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DEPARTMENT OF KINESIOLOGY

CHANGES IN KINEMATICS, ELECTROMYOGRAPHY, AND METABOLIC COST WITH ASYMMETRICAL WALKING IN HEALTHY YOUNG ADULTS

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ABSTRACT

The purpose of this study was to determine if, following an adaptive period, gait kinematics, muscle activity and metabolic cost of asynchronous walking were greater than that of synchronous walking. I hypothesize that compared to synchronous walking, asynchronous walking will result in increased metabolic cost; increased fast-leg and decreased slow-leg muscle activity of the TA, BF, RF, and LG; increased stance time, double support, and step width; and decreased stride time, swing time, and step length. Thirteen healthy college students, 7 men and 6 women, completed the protocol that consisted of level walking on a split-belt treadmill in a series of synchronous and asynchronous walking phases. With respect to the slow leg, the adaptive phase resulted in decreased stride times, decreased swing time, decreased stance time, decreased periods of double support, increased step length, and increased step width (Table 3.1). These changes with asynchronous gait were often coupled with changes on the fast leg, which included increased swing time, decreased stance time, increased periods of double support, increased step length, and increased step width (Table 3.1). Asynchronous walking resulted in greater muscle activity during periods of the gait cycle atypical for the respective muscle. Specifically, asynchronous walking resulted in significantly greater right TA activity in swing (Figure 3.7), greater right VL activity in mid to terminal stance, greater left VL activity during terminal swing (Figure 3.8), greater left LG activity during late stance through initial swing, greater right LG activity during terminal swing (Figure 3.9), greater left BF activity during terminal stance, and greater right BF activity from mid through terminal swing (Figure 3.10). Additionally, metabolic cost during the adaptive phase was 26.5% greater than the control, although this value decreased significantly through the adaptive phase, and was 22.6% less than the control by the end of the post-adaptive phase. The data confirm previous studies that suggest that adult human neural systems are capable of adjusting to asynchronous interlimb patterns and that initially exaggerated asymmetries trend toward control values after a period of adaptation.

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Chapter 1

Introduction

Walking is the most basic form of human locomotion. From an evolutionary standpoint, it has allowed humans to navigate diverse terrain to obtain food, escape predators, and interact since the advent of our species. From a biomechanical standpoint, it is the cyclic, symmetrical process by which all humans travel from one place to another, identifiable in general pattern, but highly individual in execution.

A variety of pathologies, injuries, etc. can lead to an atypical gait. Cerebral stroke survivors (Marks & Hirschberg, 1958), unilateral transfemoral amputees (Boonstra et al., 1994), and patients with neurological disorders, ranging from cerebral palsy (Galli et al., 2010) to Parkinson's Disease (Roiz et al., 2010) to traumatic brain injuries (Chow et al., 2010), each endure a range of physical abnormalities such as an asymmetrical, or asynchronous gait. Both asymmetrical and asynchronous gait are synonymous with gait that is characteristically uneven and may demonstrate altered kinematic profiles, spatiotemporal relationships, and muscle activation, as well as increased metabolic cost. Similarly, gait asymmetry is highly correlated with the degree of motor recovery, and therefore can be used as a measurement of functional improvement following a gait-altering incident (Brandstater et al., 1983). Stroke survivors, in particular, demonstrate significant and well-documented changes in gait. Such changes are characterized by a variety of altered kinematic factors, including overall decreased preferred walking speed, greater periods of bilateral double limb support and stride time, as well as greater swing time on their paretic side and greater stance time and duration of single limb support on their non-paretic side (Vonschroeder et al., 1995). The asymmetrical gait typical of stroke survivors is also accompanied by electromyographic changes, such as an overall decrease in

EMG amplitude and increase in activation on the paretic side (Marks & Hirschberg, 1958), as well as increased metabolic cost (Zamparo et al., 1995).

In an experimental setting, asynchronous gait can be simulated by the use of a split-belt treadmill. Similarly, adult human neural muscular systems are capable of adapting to asynchronous interlimb patterns after a brief period of training by extending the stance and shortening the swing phase of the slower limb and the opposite on the faster limb (Dietz et al., 1994; Reisman et al., 2007; Reisman et al., 2009).

As previous studies suggest, asynchronous walking patterns can be remediated poststroke using the split-belt treadmill as a long-term rehabilitation strategy (Reisman et al., 2007; Tyrell et al., 2011). However, such intervention programs range in scope and effectiveness standard. Most programs aim to increase preferred walking velocity, as increased walking velocity been shown to be key in both motor and sensory recovery post stroke (Hsu et al., 2003; Schmid et al., 2007). Residual gait asymmetries could also have functional importance (Schmid et al., 2007), ranging from increased metabolic cost during activities of daily living (Perry et al., 1995) to a greater risk of falls (Lamb et al., 2003). Ultimately, understanding the metabolic cost of normal, adaptive, and post-adaptive asynchronous gait would provide rehabilitation specialists a more comprehensive understanding of the impact of their current therapy methods.

The purpose of this study is to compare the gait kinematics, muscle activity, and metabolic cost of synchronous versus asynchronous walking. The study will assess the amount of time necessary to adapt to an asynchronous gait, as well as the changes in gait kinematics, muscle activity, and metabolic cost experienced throughout each testing interval.

I. Synchronous Level Walking

Locomotion is the cyclic movement pattern by which animals transport themselves from one location to another. This encompasses starting and stopping, alterations in speed and direction, as well as adjustments for terrain, grade, obstacles, etc. In walking, this is based off of a basic pattern in continual forward movement. Walking is a cyclical activity composed of a series of inherently unstable positions, requiring constant adjustments and sound neural control to maintain a balanced, vertical position. Though the particulars of walking differ on an individual basis, steady-state human walking optimizes muscle activation to minimize the metabolic energy required to traverse a given distance.

Gait Cycle

In human bipedialism, walking is marked by the cyclical motion of each foot from one position of support to the next in the direction of progression. During human walking, the center of mass passes over the supporting leg, during which time the alternate leg swings forward to prepare for the subsequent support phase. At least one foot is always in contact with the ground, generating continuing ground reaction forces to support the body, and weight transfer occurs during the period when both feet are in contact, mutually supporting the center of mass (Rose & Gamble, 2006).

During self-paced, rhythmic walking, the sequence of reoccurring gait events is consistent across individuals and allows for a general analysis based on one cycle of those consistently observed events. Using one leg as a reference, these events include (1) foot strike, (2) opposite foot-off, (3) opposite foot strike, and (4) foot-off (Sutherland & Cooper, 1981). A complete gait cycle is the period of time during which one complete sequence of these events occur, most commonly referenced in terms of percentage with initial foot strike at 0% and a second foot strike of the same leg at 100% (Rose & Gamble, 2006).

The basic gait cycle is divided into two phases: stance and swing. The stance phase consists of the initial 62% of the gait cycle, beginning at foot strike (0%) and ending at ipsilateral foot-off (62%), and is the period during which the reference limb is in a weight bearing position and is in contact with the ground. This phase can be divided further into periods of initial double limb support (foot strike to contralateral foot-off), single limb support (contralateral foot-off to contralateral foot strike), and second double limb support (contralateral foot strike to ipsilateral foot-off). During initial double limb support, the ipsilateral limb experiences rapid loading beginning at foot strike (0%) and acts in shock absorption and deceleration of the body's center of mass as weight is transferred onto that forward limb to prepare for contralateral foot-off (12%) and the subsequent period of single limb support (12-50%). During single limb support, the body travels over the grounded foot until contralateral foot-strike (50%), marking the end of single limb stance. This begins the period of second double limb support (50-62%), or preswing. As before, the forward limb experiences rapid loading as the ipsilateral limb extends at the hip and prepares for foot-off (62%) (Rose & Gamble, 2006).

The remaining 38% of the gait cycle is termed the swing phase, beginning at ipsilateral foot-off (62%) and ending at ipsilateral foot-strike (100%). This begins with a period of initial swing (foot-off to foot clearance), followed by mid swing (foot clearance to the point at which the tibia is perpendicular to the floor, known as tibia vertical), and ends with terminal swing (tibia vertical to ipsilateral foot-strike). In normal gait, the contralateral leg is 180 degrees out of phase, and its time in single limb support is equal to swing time in the ipsilateral leg and vice versa (Rose & Gamble, 2006).

Inverted Pendulum Mechanics

Walking dynamics minimize the total mechanical energy (TE) necessary to propel the center of mass forward. TE is the sum of kinetic energy (KE) and potential energy (PE) about the center of mass. The energy profile of walking is classically defined using an inverted pendulum model. In such a model, the leg in stance phase is assumed to be straight, and the body's center of mass (COM) generates the dominant inertia at the hip and vaults over a stiff support in a pendular arc until the COM is redirected to move upward and begin a subsequent arc at contralateral foot strike (Saibene & Minetti, 2003). In doing so, PE and KE are continually exchanged as the COM vaults over the support leg, resulting in conservation of total mechanical energy.

Kinematics

Healthy young adults adapt a mean walking velocity of 140.9 ± 2.77 cm at a cadence of 110.6 ± 2.13 steps per minute (Rose & Gamble, 2006). Additionally, the gait of young adults is characterized by an average stride length of 145.1 ± 3.06 cm and step width of 12.5 ± 1.36 cm (Rose & Gamble, 2006). Double support makes up approximately $22.0 \pm 1.70\%$ of the gait cycle of a young adult (Rose & Gamble, 2006).

As a person's stroke risk doubles every ten years after the age of 55 (Chung & Caplan, 2007), it is necessary to address age-related changes in gait. In general, elderly persons adopt a more conservative gait that is accompanied by greater spatiotemporal variability (Menz et al., 2003). Compared to young adults, older adults demonstrate 12% shorter step length (Judge et al., 1996); 17% slower preferred walking speed and 20% shorter stride length (Elble et al., 1991); greater double support time (Judge et al., 1996); and increased step width variability (Grabiner et al., 2001).

Each joint undergoes simple angular motion in the sagittal plane as well as rotational, three-dimensional motion. In the interest of simplicity, further explanation will focus entirely on planar motion about the center of rotation in the hip, knee, and ankle, respectively.

Throughout the gait cycle, the hip progresses through a pattern of flexion and extension that most closely resembles a sinusoidal curve. At initial foot strike (0%), the hip is in its most flexed position and gradually extends as it progresses through weight transfer and single support. At contralateral foot strike (50%), the ipsilateral hip is in its most flexed position. During this preswing period, weight is transferred onto the contralateral limb to prepare for foot-off (62%), at which point the ipsilateral hip begins to flex and proceed through swing phase. Flexion continues until immediately before foot-strike, when the hip extends slightly to decelerate the swinging limb and prepare for weight acceptance at foot-strike (100%) (Rose & Gamble, 2006).

The knee flexion-extension curve is the product of two flexion events throughout the course of the gait cycle. The knee is in full extension at initial foot strike (0%), after which it immediately flexes to act in shock absorption and weight acceptance. This flexion peaks at opposite foot-off (12%), and shifting anterior ground reaction forces act to passively extend the knee by mid-stance. The second flexion event occurs at foot-off (62%) and acts in toe clearance throughout the swing phase. Maximum knee flexion occurs as the ipsilateral limb passes the support limb, after which the knee gradually extends and reaches full extension immediately prior to foot-strike (100%) (Rose & Gamble, 2006).

The flexion and extension events about the ankle joint progress through four stages during a complete gait cycle. In the first stage, the ankle is in a neutral position at initial footstrike (0%), and posteriorly-oriented ground reaction forces aid in the passive extension of the ankle to a flat foot position prior to opposite foot-off (12%). As the center of mass passes over the standing leg during single support, the foot remains flat and the ankle undergoes flexion, marking the second rocker stage. The third event occurs near the end of single support phase as the supporting limb undergoes ankle extension during heel rise and preparation for contralateral foot-strike (50%). Following a brief period of second double support, the limb of reference undergoes foot-off (62%), at which point the ankle experiences maximum extension of 20-25 degrees. Finally, the ankle flexes simultaneously with the knee, as both joints act in toe clearance during the swing phase. Following toe clearance, the ankle returns to a neutral position in preparation for foot-strike (100%) (Rose & Gamble, 2006).

Muscle Activity

During normal gait, lower extremity muscle activity follows a characteristic rhythmic pattern, coupled with gait events in either the stance or swing phase. Normalized studies of such phasic contractions reveal standard electromyographic (EMG) activity during maximal isometric contraction from which conclusions about force generation, motor unit recruitment, and muscle fatigue may be drawn regarding each muscle under consideration.

Differing walking speeds, achieved by altering stride length and/or stride frequency, are accompanied by changes in EMG activity at varying speeds. In general, increases in walking speed require limb segments to move through a greater range of motion in a decreased amount of time. This demands increased muscular forces to generate requisite accelerative and decelerative forces, and subsequent increases in EMG response amplitude for involved muscles can be observed (Murray et al., 1984).

In this study, four lower extremity muscles were recorded from both legs: tibialis anterior (TA), vastus lateralis (VL), lateral gastrocnemius (LG), and biceps femoris (BF).

Tibialis Anterior (TA)

The tibialis anterior (TA) acts as the primary ankle flexor in a biphasic nature throughout the gait cycle. Following foot-strike (0%), the TA reaches its peak activity as it isometrically contracts to control foot-fall during the loading response (5%), known as the heel-strike burst (HSB) (Perry et al., 1992). The activity decreases shortly after, though it may show some activity as it acts to move the center of mass forward during mid-stance. The second burst of activity occurs immediately following foot-off, known as the toe-off burst (TOB) (62%). As the limb enters the swing phase, the TA begins a concentric contraction to gradually flex the ankle to ensure toe clearance during mid-swing. The activity of the TA continues to increase as it maintains ankle flexion through terminal swing and second foot strike (100%) (Cram & Kasman, 1998).

Across a range of increasing walking speeds, the EMG magnitude of the HSB (5%) increases almost linearly in a triangular pattern (Barrett et al., 2007). However, relative to self-selected pace, slowed walking speeds resulted in a decline of amplitude that was smaller relative to the increased amplitude reported for peaks at matched faster speeds (Byrne et al., 2007).

The TOB (62%) can be characterized as a trapezoidal shape pattern of activity at low speeds that increases in magnitude and tends toward a triangular shaped activity pattern at higher speeds (Barrett et al., 2007). This phase is marked by a greater degree of inter-individual variability. The changes in peak EMG amplitude recorded across increasing speeds decreased in a non-linear fashion and were lesser in magnitude than those recorded for the HSB (Byrne et al., 2007).

Vastus Lateralis (VL)

The vastus lateralis is a knee extensor and acts primarily in a stabilizing role. The major peak of activity occurs during weight acceptance (~10%), immediately following initial footstrike. The isometric contraction controls the amount of knee flexion and decelerates the thigh during loading before contracting concentrically to aid in knee extension during midstance. A lesser, secondary peak occurs during initial swing at which point the VL aids in halting the backward momentum of the swing limb and helps to control the knee angle during swing (Cram & Kasman, 1998). VL EMG magnitude changes linearly with changes in walking speed from 0.75 - 1.75 m/s (Hof et al., 2002).

Lateral Gastrocnemius (LG)

Activation of the lateral gastrocnemius results in ankle extension, rendering it the primary mover in walking in a biphasic nature. A secondary burst begins just prior to initial foot-strike and peaks as the LG aids in knee flexion during weight acceptance (~10%). After this point, the LG undergoes an eccentric contraction, the magnitude of which determines the amount of forward rotation about the ankle and compensatory knee flexion through mid-stance. Its primary burst of activity occurs through a concentric contraction to extend the ankle and propel the body forward during terminal stance (~40-45%). Following this power movement, some activity may continue to exist through swing phase as the LG contributes to knee flexion (Cram & Kasman, 1998).

At speeds ranging from 0.75 - 1.75 m/s, normalized LG EMG profiles showed a linear relationship between increasing walking speed and EMG magnitude and demonstrated similar timing patterns of EMG activity (Hof et al., 2002). However, at slow speeds, (0.5 m/sec) the

onset of the primary burst of activity is delayed by 3-6% and extended by 8-11% of the gait cycle (Clancy et al., 2004).

Biceps Femoris (BF)

The biceps femoris plays a primary role in knee flexion as well as hip extension throughout the gait cycle. Activity increases through terminal swing as the BF undergoes an isometric contraction to aid in the deceleration of the swinging limb (Rose & Gamble, 2006). Peak activity occurs immediately after initial foot-strike (~4%), at which point the BF is cocontracted with the other knee flexors and extensors to stabilize the knee position. It also plays a role in hip extension during weight acceptance (Cram & Kasman, 1998).

The BF also showed a prominent activity peak early in swing phase (~60%) at higher speeds. The amplitude of the peak appears to increase in a non-linear (quadratic) fashion with increasing speed, though it is entirely absent at very low speeds (Hof et al., 2002).

Metabolic Cost

The human body uses energy in the form of ATP to support continual physiological processes, including those that control and support human walking. Skeletal muscles hydrolyze ATP to release chemical energy and generate mechanical and chemical work. The basal metabolic rate, or resting metabolic rate, represents the requisite energy necessary to support only the most basic physiological functions, measured while lying down, ~3-4 hours following food consumption (Rose & Gamble, 2006). According to the Mayo Foundation Normal Standards, the basal metabolic rate is 40 kcal/m²/hr and 37 kcal/m²/hr for males and females, respectively. The maintenance of a standing position requires approximately 25% more energy than lying supine.

Similarly, the energy requirement of walking is approximately 50% greater than that at rest, which reflects the increased energy need of the skeletal muscles responsible for accelerating and decelerating the limbs against the force of gravity (Rose & Gamble, 2006).

General reviews of the existing literature suggest that the metabolic cost of walking does not change significantly with increasing age (Lamontagne et al., 2007). Alternatively, total body weight is strongly associated with increased metabolic cost, though distal loads have a greater impact on increasing metabolic cost because of the energetic requirement to accelerate and decelerate the limb (Martin, 1985). It is important to consider the units in which oxygen uptake is expressed. When normalized to body weight, most studies have found no significant differences in metabolic cost of walking in adults across age, weight, or genders (Lamontagne et al., 2007).

Preferred walking speed has been found to be nearly synonymous with optimal walking speed, or the speed at which walking is the least metabolically expensive (Ralston, 1958). Therefore, walking speed is determined by a step frequency and step length that is energeticallyoptimal. The metabolic cost of walking as a function of speed corresponds with a "J-shaped" curve. At preferred walking speed, metabolic cost is minimized. At gradually slower speeds, metabolic cost increases. Similarly, with increased speed, the energy cost of walking gradually increases until it is more metabolically effective to transition to a run (Ralston, 1976).

II. Asynchronous Level Walking

Asynchronous walking characterizes gait that differs significantly from the temporal and spatial parameters of regularly observed locomotion. Asynchronous walking is characteristic of a variety of pathologies, ranging from those of neurodegenerative nature such as Parkinson's Disease, to multiple sclerosis, to victims of traumatic brain injuries, to amputees, to cerebral stroke victims, etc. Stroke victims demonstrate a characteristic asynchronous gait and will serve as the asynchronous walking basis for comparison. Nearly 2/3 of stroke victims demonstrate impaired walking post-stroke (Jorgensen et al., 1995), though a variety of studies have reported improvement of walking function with time and/or following intervention, most of which were assessed by an increase in preferred walking speed or successful completion of a task-oriented training (Hsu et al., 2003). Furthermore, asymmetry is itself characteristic of post-stroke gait. In a study by Patterson *et al.*, 55.5% of prior test participants demonstrate temporal asymmetry and 33.3% demonstrate spatial asymmetry. Asynchronous walkers experience challenges to balance control, an increased potential for injury, metabolic inefficiency, and subsequent decreased functional ability to complete the requisite activities of daily living (Patterson et al., 2008).

Kinematics

A variety of altered kinematic factors characterize typical post-stroke gait. Most noticeably, stroke patients demonstrate decreased preferred gait speed compared to age-matched, healthy adults (Goldie et al., 1996), reported to range between 0.48 – 1.22 m/s (Da Cunha-Filho et al., 2003; Platts et al., 2006). Such decreased gait speed is also associated with decreased stride length and an increased duration of the entire gait cycle (Nakamura et al., 1988). Other common kinematic changes post-stroke include decreased step frequency, decreased step length (Nakamura et al., 1988), increased step width (Chen et al., 2005a), and increased double support time (Goldie et al., 2001). However, in a later study, Chen *et al.* (2005b) determined that step length is highly variable between individuals, as some stroke patients demonstrated greater step length on the paretic side while some showed greater step length on the non-paretic side (Chen et al., 2005b). As compared to age and gender-matched healthy adults, stroke patients spent a greater proportion of the gait cycle in stance on both the paretic and non-paretic sides (Chen et al., 2005b). Similarly, patients spent a greater proportion of the gait cycle in stance on the non-

paretic side and a greater proportion in double support on the paretic side (Lamontagne et al., 2000).

Slowed gait is associated with decreased peak joint displacements about the hip, knee, and ankle joints in the sagittal plane in both healthy individuals (Murray et al., 1966) and in individuals post-stroke (Chen et al., 2003). In stroke victims, the paretic limb often displays an increased degree of ankle flexion at initial foot strike (Kim & Eng, 2004) as well as decreased knee flexion and a lack of ankle flexion in swing (Kuan et al., 1999). At higher walking speeds, many of these observed kinematic changes tend to improve, suggesting a role for rehabilitation at higher gait speeds (Lamontagne & Fung, 2004). However, some kinematic deviations appear to be independent of speed. In speed-matched trials with comparing the gait of stroke patients and healthy controls, stroke patients demonstrate less hip extension at foot-off bilaterally, decreased knee and ankle flexion on the paretic side, and increased knee flexion during swing on the nonparetic side (Chen et al., 2005b).

Muscle Activity

Interindividual differences in muscle activity and patterning in the gait of post-stroke patients are common. These deviations from the averaged normal values are much more apparent in paretic leg, though some atypical muscle activity and patterning is also clear in the non-paretic leg (Marks & Hirschberg, 1958). Stroke patients demonstrate an overall decrease in EMG amplitude on the paretic side and an overall increase in activation on the paretic side. Furthermore, the temporal EMG abnormalities of asymmetrical gait are seen bilaterally and reflect the timing of the paretic limb (Marks & Hirschberg, 1958).

Tibialis Anterior (TA)

As compared to matched healthy controls, stroke victims demonstrate increased duration of paretic-side tibialis anterior activity during swing (73% versus 60% in controls) and decreased total duration of activity during single support phase (28% vs 48%) (Den Otter et al., 2007). Additionally, the mean activity of the TA is positively correlated with increases in speed (Hesse et al., 2001).

Vastus Lateralis (VL)

Data on specific changes with asymmetry in vastus lateralis activity were limited. However, overall quadriceps activity was found to occur at an earlier onset in stroke patients than in speed-matched, healthy controls. Additionally, the mean activity of the quadriceps is positively correlated with increases in speed (Hesse et al., 2001).

Lateral Gastrocnemius (LG)

Changes in gastrocnemius activity were subtle and varied in relation to the severity of disability. The average relative duration of medial gastrocnemius activity in paretic leg was (51%) was significantly longer than in the control group (36%) during first double support phase (Den Otter et al., 2007).

Biceps Femoris (BF)

Stroke patients demonstrated a significantly longer duration of biceps femoris EMG activity during swing phase on the paretic side (70%) as well as on the non-paretic side (71%) as

compared to matched controls (45%) which was coupled with an extended duration of coactivation with the rectus femoris bilaterally (Den Otter et al., 2007). Additionally, mean muscle activity was positively correlated with increasing walking speed (Hesse et al., 2001).

Metabolic Cost

Due to the nature of asymmetrical gait, the pattern of mechanical energy exchange between potential and kinetic energy is disrupted, rendering asymmetrical gait more metabolically costly (Olney et al., 1986). Specifically, the average mechanical energy cost per stride is higher because of the increased energetic demand during preswing and swing of the paretic limb (Chen et al., 2005b), causing stroke victims to expend a greater amount of energy per unit distance at a given speed than healthy matched controls (Zamparo et al., 1995). The metabolic cost of walking in stroke survivors is 1.5-2 times greater than healthy, age-matched controls, which has practical consequences such as limited daily activities (Macko et al., 1997). The relative energy cost per unit distance decreased when stroke patients were instructed to walk more quickly, suggesting that greater walking speeds facilitate a less metabolically-expensive gait pattern (Hesse et al., 2001).

III. Purpose Statement and Hypotheses

The purpose of this study was to determine if, following an adaptive period, gait kinematics, muscle activity, and metabolic cost of asynchronous walking were greater than that of synchronous walking. Previous studies have determined that asynchronous gait is marked by greater metabolic cost and muscle activity. However, no study has yet addressed the energetic cost of *adapting* to such a gait, in terms of gait kinematics, muscle activity, and metabolic cost

both during and following the adaptive phase. First, I hypothesize that metabolic cost in asymmetrical walking will be greater than during synchronous walking. Second, I hypothesize that, as compared to muscle activity during regular, synchronous walking, the muscle activity of the TA, VL, LG, and BF will be greater during asynchronous walking for the fast leg and less for the slow leg. Furthermore and with respect to the slow leg, I hypothesize that step width, double support time, and stance time will be greater, and step length, swing time, and stride time will be less during asynchronous walking.

Chapter 2

Methods

I. Participants

Thirteen healthy college students, 7 men and 6 women (age = 21.846 (*1.908*) yr, height = 1.73 (*0.1266*) m, mass = 68.57 (*13.62*) kg, mean (*standard deviation*) completed the protocol. All of the participants gave written informed consent that followed the guidelines of The Pennsylvania State University Human Research Committee.

II. Protocol

Each participant completed a standing trial and a series of synchronous and asynchronous walking conditions at slow, moderate, and fast speeds. Participants were asked to walk on a custom-built treadmill consisting of two independently motorized belts, which allowed for the control of individual belt speed, and subsequently leg speed.

Throughout the protocol, participants would walk on the treadmill with both belts moving at the same speed ('tied' configuration) or at different speeds ('untied' configuration). During the tied condition, the belt speeds were either 'slow' (0.5 m/s), 'moderate' (1.0 m/s) or 'fast' (1.5 m/s). During all untied conditions, the participant's right leg was matched to the fast speed belt ('fast leg,' 1.5 m/s), and the left leg was matched to the slow speed belt ('slow leg,' 0.5 m/s). The left-leg moderate condition served as the basis of comparison to which all measured parameters were normalized and will henceforth be referred to as the 'control.'

Each testing session consisted of three phases. In the control phase, the belts were tied, moving first at a moderate pace for 10 minutes to allow the participant to acclimatize to the treadmill apparatus, then at a slow speed, and finally at a fast speed for 5 minutes each. During the adaptation phase, participants walked with the belts untied for 16 minutes. During the post-adaptation phase, the belts were again tied at a slow pace for 10 minutes.

CONTROL	RANDO	DMIZED		
Moderate 1.0 m/s 10 min	Slow 0.5 m/s 5 min	Fast 1.5 m/s 5 min	Adaptation (Untied) 1.5 m/s and 0.5 m/s 16 min	Post-Adaptation (Tied) 0.5 m/s 10 min

Figure 2.1: Experimental Design. The 'moderate' speed condition is referred to as the control. The order of the slow and fast conditions were randomized for each participant. 'Moderate,' 'slow,' 'fast,' and 'post-adaptation' denote tied conditions at the designated speeds. Similarly, 'adaptation' indicates the untied, or asynchronous walking phase during which the left limb was moving at 0.5 m/s and the right limb at 1.5 m/s.

The time between each speed shift and phase was brief (<1min). During this time,

participants were instructed to use the side handrails to lift themselves off of the treadmill belt surface until the belt speeds were reset, minimizing the neutralizing effects of steady standing between each condition.

Prior to the testing period, participants were not given any practice walking on the treadmill apparatus, in either tied or untied conditions, though they were informed that the belts would move at two different speeds at some point during the testing. For the purpose of safety, participants were required to hold lightly onto the side handrails throughout the duration of the trials.

III. Kinematics

I collected kinematic data with a six-camera, passive marker, 3D photogrametry system (Motion Analysis Corporation, Santa Rosa, CA). The calibration residual was less than 0.5 mm in a capture volume of approximately 2m x 2m x 2m. Prior to data collection, I placed retroflective markers on the sacral crest and lateral malleoli as well as the shoes of each participant superficial to the posterior calcaneus and superior hallux. Additional markers were placed bilaterally at the fibular head and anterior superior iliac spine, as well as at the C7 vertebrate, right shank, left thigh, and right hip. I collected the marker data at 100 Hz and postprocessed the data with EVaRT software (Version 3.21, Motion Analysis Corporation, Santa Rosa, CA). A purpose-written Matlab program (Version R2006b, The Mathworks, Natick, MA) was written for subsequent data processing that included a lowpass filter for the marker trajectories at 7 Hz (fourth-order, dual-pass, Butterworth). Minutes 1-2 of each walking phase were excluded from later data analysis due to high variability. Average values of the remaining time in the control, slow, and fast phases were reported as the basis of comparison for the respective phase. Similarly, average values of the first and fifth minutes of the adaptive and postadaptive phases served as representative samples and will henceforth be referred to as Adapt 1, Adapt 5, Post 1, and Post 5, respectively.

IV. Electromyography

I measured electromyography (EMG) signals using a wired amplifier system (Bortec Octopus AMT-8, Calgary, AB, Canada) with a bandpass filter setting of 5~500 Hz. I collected the data at 2000 Hz using EVaRT software, which also synchronized the marker and muscle data. Prior to electrode placement, I prepared the skin with fine sandpaper and rubbing alcohol. I

placed 1 cm x 1.5 cm² bipolar, silver-silver chloride, surface electrodes (Vermed, A10041, Bellows Falls, VT) over the tibalis anterior (TA), lateral gastrocnemius (LG), vastus lateralis (VL), and biceps femoris (BF) muscles of both legs according to the recommendations by Cram and Kasman (1998). The interelectrode distance was 2 cm. I verified that the position of the electrodes was functionally correct and that cross talk between the muscles was negligible with a series of flexion and extension exercises suggested by Winter et al. (1994), and Cram and Kasman (1998).

V. Muscle Activity Data

For the temporal analysis, a linear envelope was generated(Winter, 1991) obtained by low pass filtering (dual-pass, fourth order, Butterworth) the rectified EMG data at 30 Hz. For the amplitude analysis, I full-wave rectified the band-pass signals, calculated the mean EMG amplitude (mEMG) and averaged five strides for each condition. Each stride was further divided into eight time segments. For each trial, heel strike was calculated by marker location and stance and swing phase was determined for both legs. Swing phase was then categorized into: (1) initial swing from 0-34% of swing phase, (2) mid swing from 34-60% of swing phase, and (3) terminal swing from 60-100% of swing phase. Stance phase was categorized into: (1) initial stance 0-20% of stance phase, (2) mid stance from 20-57% of stance phase, (3) late stance from 57-81% of stance phase, and (4) terminal stance from 81-100% of stance phase. For each participant, I normalized the mEMG for each experimental condition based on the mEMG for the synchronous level walking condition.

VI. Metabolic Cost

I measured rates of oxygen consumption (VO₂) and carbon dioxide production (VCO₂) using an open circuit respiratory system (ParvoMedics, Sandy, UT, USA). Standing metabolic rate was measured prior to beginning the experimental trials. For all conditions, I allowed 3 minutes for the participants to reach steady state and then calculated the average VO₂ (mL O₂ per second) for the subsequent 4 minutes. My rationale for measuring metabolic cost was to understand the adaptive strategy taking place during asynchronous walking.

VII. Statistical Analyses

All values across all parameters were normalized to those recorded from the left leg during the warm up condition (1.0 m/s). This will be referred to as the control (100% normalized value).

Metabolic cost and bilateral muscle activity of the TA, LG, VL, and BF were analyzed across all conditions using a repeated measures design (ANOVA). When appropriate, I performed Newman-Keuls post hoc tests and paired Student's t-tests to analyze the differences between conditions and reported all values as mean \pm standard deviation. Significance was defined as $p \le 0.05$.

Chapter 3

Results

I. Kinematic Parameters

In summary, there was a significant difference between synchronous and asynchronous walking for all variables I evaluated. Specifically, compared to the control and with respect to the slow leg, the adaptive phase resulted in decreased stride times, decreased swing time, decreased stance time, decreased periods of double support, increased step length, and wider step width. These changes with asynchronous gait were often coupled with changes on the fast leg, which included increased swing time, decreased stance time, increased periods of double support, increased step length, and wider step width.

Table 3.1: Normalized Kinematic and Temporospatial Values Across Testing Conditions. All values are normalized to the bold font control (100%, left leg, 1.0 m/s walking in the Warm Up condition) for the respective parameter. Italicized values within each parameter are significantly different from the control.

		Warm Up	Slow	Fast	Adapt1	Adapt5	Post1	Post5
Stride Time	Left	100.00	148.425	82.845	99.023	97.203	138.703	137.421
	Right	99.936	148.331	82.863	82.728	97.171	138.294	137.299
Swing Time	Left	100.00	123.93	91.517	125.83	128.48	120.957	116.709
	Right	99.084	121.75	90.478	69.260	77.380	121.41	116.19
Stance Time	Left	100.00	160.35	78.693	86.062	82.045	147.294	147.407
	Right	100.36	161.19	79.221	88.750	106.85	146.80	147.60
Double Support	Left	100.00	194.01	67.084	131.24	93.273	151.426	166.932
	Right	100.33	196.46	67.637	44.971	78.916	170.01	176.49
Step	Left	100.00	75.827	123.13	123.09	116.55	85.286	74.467
Length	Right	100.06	75.819	123.05	142.94	116.55	84.612	74.328
Step Width	Left	100.00	97.407	93.972	106.08	102.99	105.43	101.27
	Right	98.253	96.299	90.684	112.96	103.02	103.75	100.20

Stride Time



Figure 3.1: Normalized Stride Time values in all conditions. Stride time was significantly greater during the slow condition and significantly less during the fast condition. Left limb stride time was significantly greater than the right limb during the first minute of the adaptive phase. This difference was insignificant by the end of the adaptive phase, and the stride time with respect to each limb approached that of the control. The stride time during all post-adaptive phases was significantly greater than the control and remained constant through the termination of the testing period.

Mean stride time was 130.99 ms in the control. Average stride time increased by nearly

48% during the slow condition (p<0.001) and decreased by 17% during the fast condition

(p<0.001). During all pre-adaptive, synchronous walking conditions, no significant differences

between the right and left limb were observed.

Stride time of the left limb during the first minute of adaptation was 16% greater than the right limb (p<0.05), though this difference was not apparent by the final minute of the adaptive phase (p= 0.3829). By the end of the adaptive phase, both limbs demonstrated a stride time that

was approximately 3% less than the control, though this difference was insignificant (p>0.15). Both post-adaptive phases resulted in significantly greater stride times as compared to the adaptive phases (p<0.001) and control (p<0.0001), which remained nearly constant throughout the post-adaptive period (p>0.10). Post-adaptive stride time did not differ significantly from the slow condition (p>0.05).



Swing Time

Figure 3.2: Normalized Swing Time values in all conditions. Swing time was significantly greater during the slow condition. Left limb swing time was significantly greater than both the right limb and the control throughout the adaptive phase. The swing time during all post-adaptive phases was significantly greater than the control and decreased slightly, though insignificantly, by the end of the post-adaptive period.

The average recorded swing time was 42.66 ms in the control. Average swing time increased by nearly 23% during the slow condition (p<0.0005) and decreased by 9% during the fast condition, though this difference was insignificant (p>0.05). During all pre-adaptive, synchronous walking conditions, no significant differences between the right and left limb were observed.

During the first minute of adaptation, left limb swing time was 57% greater than that of the right limb (p<0.0005) and 26% greater than the control (p<0.0005). By the final minute of the adaptive phase, this difference in swing time decreased slightly to 51% (p<0.0005). Mean swing time recorded during the first minute of the post-adaptive phase was approximately 21% greater than the control (p<0.0005). Swing time decreased by 5% by the end of the post-adaptive period, though this difference was insignificant (p>0.05). Furthermore, most-adaptive swing time did not differ significantly from the slow condition (p>0.10).

Stance Time



Figure 3.3: Normalized Stance Time values in all conditions. The slow and fast conditions resulted in stance times that were significantly greater than and less than the control, respectively. Stance time during the first minute of the adaptive phase was significantly less than the control, though the stance time of the right limb was significantly greater by the end of the adaptive phase. Stance time in the post-adaptive phase was significantly greater than the control and remained nearly constant through the termination of the testing period.

The average stance time recorded in the control group was 88.33 ms. Mean stance time

was 60% greater during the slow condition (p<0.0005) and 21% less during the fast condition

(p<0.01). During all pre-adaptive, synchronous walking conditions, no significant differences

between the right and left limb were observed.

The first minute of the adaptive phase resulted in a stance time that was approximately

12.5% less than the control, though that difference was insignificant (p>0.05). Furthermore, no

significant difference was observed between the right and left limb (p>0.50). By the final minute

of the adaptive phase, right limb stance time was 18% greater (p<0.05), resulting in a difference of nearly 25% between the stance time of the right and left limbs (p<0.005). Mean stance time recorded throughout the post-adaptive phase was approximately 47% greater than the control (p<0.0005), which remained nearly constant throughout the post-adaptive period (p>0.8). Postadaptive stance time was approximately 13% less than the slow condition, though this difference was not significant (p>0.05).



Double Support

Figure 3.4: Normalized Double Support values in all conditions. The duration of double support was significantly greater during the slow condition and significantly less during the fast condition. Left limb double support time was significantly greater than the right limb during the first minute of adaptation. This difference was insignificant by the end of the adaptive phase, and the double support time with respect to each limb approached that of the control. The double support time during all post-adaptive phases differed significantly between the right and left limb, and was markedly greater than all other conditions except for the slow-tied condition.

The mean double support duration was 22.99 ms in the control. Average double support time during the slow condition was 95% greater than the control (p<0.0005) and 33% less than the control during the fast condition (p<0.05). During all pre-adaptive, synchronous walking conditions, no significant differences between the right and left limb were observed.

During the first minute of adaptation, the duration of double support with respect to the left limb was 86% greater than that of the right limb (p<0.0005) and 31% greater than the control (p<0.05). By the final minute of the adaptive phase, the double support time of the right and left limb differed by only 14%, which was not significant (p>0.10). The duration of double support with respect to both the right and left limbs was over 50% greater than the control throughout the post-adaptive conditions (p<0.0005), and 20% less than the slow condition, though this difference was not significant (p>0.05). During the first minute post-adaptation, double support time with respect to the right limb was approximately 37% greater than the left limb (p<0.05). This difference was only 19% by the end of the post-adaptive condition (p<0.005), though both the right and left limb continued to differ from the control by over 65% (p<0.0005).

Step Length



Figure 3.5: Normalized Step Length values in all conditions. Step length during the fast and adaptive phases was significantly greater than the control. The step length of the right limb during the first minute of adaptation was significantly greater than that of the left limb, though this difference was no longer apparent by the end of the adaptation phase. Step length during the slow and post-adaptive conditions was significantly less than the control, and the step length observed by the end of the post-adaptive condition did not differ significantly from the matched-speed slow condition.

Step length during the control condition was an average of 62.8 cm. Mean step length was 24% less than the control during the slow condition (p<0.0005) and 23% greater than the control during the fast condition (p<0.0005). Step length throughout the adaptive conditions was significantly greater than the control (p<0.005). Step length of the right limb during the first minute of adaptation was nearly 20% greater than the left limb (p<0.0005), though this difference was not apparent by the final minute of the adaptive phase (p= 0.9998), at which point both limbs demonstrated a step length that was approximately 16.5% greater than the control (p<0.005).

Both post-adaptive phases resulted in significantly shorter steps as compared to the adaptive phases (p<0.001) and control (p<0.01). Right limb step length did not differ significantly from that of the slow condition at any point during the post-adaptive period (p>0.10). Left limb post-adaptive step length was approximately 9.5% greater than that of the slow condition during the first minute post-adaptation (p<0.005), thought this difference was less than 1.5% by the final minute of testing (p>0.10).

Step Width



Figure 3.6: Normalized Step Width values in all conditions. No significant differences in step width were observed.

Average step width was 25.4cm during the control condition. Step width exhibited no significant deviations from the control during synchronous walking at slow and fast speeds.

During the Adapt 1 condition, step width was 6% and \sim 13% wider on the left and right legs respectively, though this increase was not significant (p=0.837 and p=0.153).

II. Muscle Activity

In summary, there was a significant difference in the muscle activation of the tibialis anterior (TA), vastus lateralis (VL), lateral gastrocnemius (LG), and biceps femoris (BF) between synchronous and asynchronous walking throughout the gait cycle. Asynchronous walking resulted in greater muscle activity during periods of the gait cycle different from the timing of expected activity bursts for the respective muscle. Specifically, asynchronous walking resulted in significantly greater right TA activity in swing (Figure 3.7), greater right VL activity in mid to terminal stance, greater left VL activity during terminal swing (Figure 3.8), greater left LG activity during late stance through initial swing, greater right LG activity during terminal swing (Figure 3.9), greater left BF activity during terminal stance, and greater right BF activity from mid through terminal swing (Figure 3.10).

Tibialis Anterior

There were no significant differences in TA activity during stance. Activity of the right TA in the Adapt 1 condition during initial swing is approximately 6 times greater than the TA activity of the control. Additionally, the right Adapt 1 TA activity during initial swing is also over four times greater than the slow (p<0.005), fast (p<0.005), and post-adaptive conditions (p<0.005) and 3.5 times greater than the final minute of adaption (p<0.005). These differences persist through terminal swing, though the magnitude of difference decreases substantially in mid and terminal swing (p<0.05). The right TA is 2.3 times (p<0.05) greater than the control in mid

swing during the first minute of adaptation and decreases the activity to 2 times that of the control in terminal swing (p<0.05). Similarly, the activity of the left TA in mid swing during the first minute of adaptation is 2.7 times greater than the control (p<0.005), and falls to an insignificant difference of only 31% in terminal swing (p=0.6256).



Figure 3.7: Normalized right and left tibialis anterior EMG values. TA activity is not significantly different from the control at any point during stance. Right TA activity peaks during

the first minute of adaptation in initial swing, and left TA activity peaks during the first minute of adaptation in mid swing. The magnitude of both the right and left TA activity declines substantially after the peak values in initial and mid swing, respectively.

Vastus Lateralis

No significant differences in VL activity were observed across all conditions during initial stance, initial swing, or mid swing. Activity of the right VL during the first minute of adaptation in mid stance was 9x greater than the control (p<0.005) and 7.5 times greater than the left limb. Right VL activity in mid stance decreased by 690% from the first minute of adaptation to the final minute of adaptation. Furthermore, right VL activity was only 3.5 times greater than the control (p<0.01) in late stance and 3.8 times greater than the control (p<0.001) in terminal stance. Left VL activity during the first minute of adaptation in terminal swing was 2.3 times greater than the control (p<0.005) and 1.5 times greater than the right limb (p<0.001). By the final minute of adaptation, left VL activity was only 76% greater than the control, which was insignificant (p=0.1956), and did not differ significantly from the right limb (p=0.1365).



Figure 3.8: Normalized right and left vastus lateralis EMG values. No significant differences were observed during initial stance, initial swing, or mid swing. Right VL activity was

significantly greater than the control and left limb during the first minute of adaptation in mid stance, and decreased by nearly 700% by the final minute of adaptation. Left VL activity peaked during the first minute of adaption in terminal swing, but did not differ significantly from the control or right limb by the final minute of adaptation.

Lateral Gastrocnemius

No significant differences were observed in initial or mid stance across all conditions. Activity of the left LG during the first minute of adaptation is significantly greater than the right throughout late (p<0.0005) and terminal stance (p<0.0005) as well as initial swing (p<0.0005). Furthermore, Adapt 1 left LG activity is 2.3 times greater than the control during late stance (p<0.005), over 4 times greater than the control in terminal stance (p<0.0005), and 2.3 times greater than the control in initial swing (p<0.0005). The magnitudes of each of these differences decreased significantly throughout the adaptive period and were not significantly different from the control by the final minute of adaptation (p<0.05). No significant trends were observed with respect to the right limb throughout stance as well as initial and mid swing. Right LG activity in the first minute of adaptation was 1.6 times greater than the control during terminal swing (p<0.05) and was significantly greater than the LG activity of both the right and left limbs during the final minute of adaptation (p<0.05).



Figure 3.9: Normalized right and left lateral gastrocnemius EMG values. No significant differences were observed in LG activity during initial and mid stance. Activity of the left LG

during late and terminal stance, as well as initial swing was significantly greater than both the control and the right limb. The magnitude of these differences decreased substantially and was not significant by the final minute of adaptation. Right LG activity during the first minute of adaptation in terminal swing was significantly greater than the control and the LG activity of both limbs during the final minute of adaptation.

Biceps Femoris

No significant differences in right BF activity were observed across all conditions during initial to late stance, as well as initial swing. Left BF activity was 52% greater than the control during mid to late stance, but this difference was not significant (p>0.05). Activity of the left BF in terminal stance during the first minute of adaptation is 76% greater than the control (p<0.05) and 96% greater than the right leg (p<0.005). Activity of the right BF during mid swing in the first minute of adaptation was 2 times greater than the control (p<0.05) and 51% greater than the left limb, though the latter difference was insignificant (p>0.10). BF activity in this condition was also significantly greater that the slow (p<0.05) and fast conditions (p<0.05), as well as all post-adaptive conditions (p<0.05). These differences were sustained through terminal swing (p<0.0005), though their magnitude decreased slightly.



Figure 3.10: Normalized right and left biceps femoris EMG values. No significant differences in right BF activity were observed during stance and initial swing. In terminal stance during the

first minute of adaptation, activity of the left BF was significantly greater than the control and right leg. Similarly, activity of the right BF during mid and terminal swing was significantly greater than the control and all other conditions.

III. Metabolic Cost

The average metabolic cost recorded during the control condition is 11.59 mL/kg/min. No significant differences were observed when comparing the metabolic cost of the control with the slow (p=0.422) and fast (p=0.292) conditions. As the adaptive phase begins, the cost of walking was observed to increase by 21% (p<0.000001), with respect to the control, and peak during the second minute of adaptation at 26.5% greater than the control (p<0.0001). Metabolic cost decreases gradually as the adaptive phase continues. By the final minute of the adaptive phase, metabolic cost has decreased from its peak value by nearly 14% (p<0.00005) at which point it was approximately 14% greater than the control (p<0.005) and did not differ significantly from the metabolic cost of fast walking (p>0.05). During the post-adaptive phase, metabolic cost cost of set walking (p>0.05). During the post-adaptive phase, metabolic cost of fast walking (p>0.05). During the post-adaptive phase, metabolic cost of fast walking (p>0.05). During the post-adaptive phase, metabolic cost field uring the fifth minute of the post-adaptive phase, during which the cost of walking is 22.6% lower than the control (p<0.05). As shown in the graph (Figure 3.11), a slight increase in cost occurs during the final minute of the post-adaptive phase, though this increase is insignificant (p>0.10).



Figure 3.11: Average metabolic cost throughout walking conditions. Peak metabolic cost occurred during the second minute of the adaptive phase. Metabolic cost gradually decreased through the remainder of the adaptive phase and was not significantly different from the metabolic cost of fast walking by the end of this walking phase. Metabolic cost continued to decrease gradually through the post-adaptive phase and began to out when it reached its minimum during the fifth minute post-adaptation.

Chapter 4

Discussion

In agreement with my hypothesis, asynchronous walking resulted in decreased stride time and decreased swing time with respect to the slow leg. Participants also displayed a somewhat wider step width during asynchronous walking. However, this increase was insignificant, and therefore the hypothesis regarding step width cannot be entirely confirmed. Contrary to my hypothesis, asynchronous walking resulted in decreased stance time, decreased periods of double support, and increased step length with respect to the slow leg. Though they are beyond the scope of the original hypothesis, asynchronous walking resulted in increased swing time, decreased stance time, increased periods of double support, increased step length, and wider step width with respect to the fast leg.

My hypothesis regarding greater muscle activity in the TA, VL, LG, and BF during asynchronous walking was only partially confirmed. Normalized EMG activity was greater bilaterally in all muscles, although increases in activity were isolated to specific portions of the gait cycle. Specifically, asynchronous walking resulted in significantly greater right TA activity in swing, greater right VL activity in mid to terminal stance, greater left VL activity during terminal swing, greater left LG activity during late stance through initial swing, greater right LG activity during terminal swing, greater left BF activity during terminal stance, and greater right BF activity from mid through terminal swing.

As I had hypothesized, the metabolic cost of asymmetrical walking was greater than synchronous walking. Peak metabolic cost was observed early in the adaptive phase, but metabolic cost decreased gradually as the adaptive phase continued.

I. Kinematic Parameters

Stride time was significantly increased during the slow and post-adaptive conditions and significantly decreased during the fast condition, which confirms the relationships established in previous studies between gait speed and stride time (Nakamura et al., 1988). The hypothesis that asynchronous walking would result in a shorter stride time was confirmed, though it was only significant on the right limb during the first minute of adaptation. During the adaptive phase, the left limb was the 'slow leg,' traveling at a speed equal to that of the slow and post-adaptive conditions. During the first minute of the adaptive phase, left limb stride time was significantly greater than the right limb, which follows the relationship between gait speed and stride time, with respect to both the right and left limbs. However, this difference was insignificant by the end of the adaptive phase, and stride time with respect to each limb approached that of the control. This adaption suggests a compensatory mechanism used during asynchronous gait to tend toward a normal, even gait, possibly as a means by which to minimize metabolic cost and restore the equal transfer of kinetic and potential energy throughout the gait cycle.

Stance time was significantly increased during the slow and post-adaptive conditions and significantly decreased during the fast condition. This agrees with the previous discussion regarding stride time and gait speed (Nakamura et al., 1988). With respect to both the right and left legs, stance time during the first minute of the adaptive phase was significantly less than the control. This directly contradicts both the hypothesis that stance time would increase and previous studies that observed that stroke patients spend a greater proportion of the gait cycle in stance on both sides (Chen et al., 2005a). However, the right limb demonstrated the ability to adapt to the asynchronous walking pattern resulting in a stance time that was restored to near-baseline values. Similarly, swing time increased significantly during the slow and post-adaptive conditions, which is consistent with the findings by Nakamura, *et al.* (1988). Contrary to the

original hypothesis, left limb swing time was significantly greater than both the right limb and the control throughout the adaptive phase, which may have been a compensatory mechanism to balance the increase in right stance time.

The duration of double support was significantly increased during the slow condition and significantly decreased during the fast condition, which is consistent with the findings of Nakamura, et al. (1988). Left limb double support time was significantly greater than the right limb during the first minute of adaptation, which is consistent with the findings of Lamontagne, et al. (2000) that stroke patients spend a greater proportion of the gait cycle in double support on the paretic side. This difference was insignificant by the end of the adaptive phase, and the double support time with respect to each limb approached that of the control. Overall, this disagrees with the original hypothesis and past studies that have indicated that asynchronous walking in stroke patients results in increased double support time on both limbs (Goldie et al., 2001). This may be a consequence of the predetermined gait velocity or of the inherent biomechanical gait differences between stroke patients and normal controls. All post-adaptive phases resulted in a right limb double support time that was significantly greater than the left limb, which was opposite from the relationship that existed during the first minute of adaptation. This may suggest an exaggerated movement correction by the nervous system. From a rehabilitation perspective, Reisman, et al. (2007) suggested that enhancing an asymmetry, such as an exaggerated step length on the paretic limb, signals the nervous system to correct the movement, resulting in a post-adaptive, symmetrical walking pattern. In normal patients free from an existing asymmetry prior to the adaptive period, a similar movement correction could result in the observed exaggeration.

Step length during the fast and adaptive phases was significantly greater than the control. Specifically, the step length of the right limb during the first minute of adaptation was significantly greater than that of the left limb. Contrary to my original hypothesis, both the right and left step length were greater than the control. These differences reflect with those reported by Nakamura, *et al.* (1988) which correlate step length and gait velocity. Furthermore, Chen *et al.* (2005) demonstrated that step length is highly variable between stroke patients, which agrees with the results of this study. As observed in other parameters, the interlimb difference was no longer apparent by the end of the adaptive phase, confirming previous findings that asymmetries exaggerated during the beginning of the adaptive phase can be restored to levels near to that of the control (Reisman et al., 2007).

The hypothesis that step width would increase during the adaptive phase was not confirmed. Chen *et al.* (2005) observed an increase in step width in stroke patients, however, no significant differences in step width were observed in this current study. It is likely that participants stabilized themselves by increasing their reliance on the handrails surrounding the treadmill, thus eliminating the need for a wider base of support which is typical of clinical asymmetrical walking (Chen et al., 2005a).

II. Muscle Activity

Activity of the TA has been previously shown to increase with increases in synchronous walking speed (Barrett et al., 2007; Byrne et al., 2007). The current study observed no significant changes in TA activity correlated with changes in synchronous walking speed. This is likely due to the fact that my study was limited to a fast, control, and slow walking speed rather than a greater number of speeds separated by smaller intervals across that range. Past studies have found that stroke victims demonstrate prolonged and increased paretic-side TA activity during swing (Den Otter et al., 2007). In agreement with both past studies and the standing hypothesis, the right TA activity during the adaptive phase in this study was found to be nearly six times that of the control in initial swing and persisted, though at a lesser magnitude, through terminal swing.

Little data exists directly comparing paretic and non-paretic VL activity, though past studies indicate that general quadriceps activity is positively correlated with increases in speed and occurs at an earlier onset in asymmetrical walkers (Hesse et al., 2001). In agreement with these past findings and the hypothesis, right VL activity was nine times greater than the control and nearly 7.5 times greater than the left limb during the first minute of adaptation in mid stance. As evidenced in other parameters analyzed in this study and in agreement with the findings of Reisman, *et al.* (2007), participants were able to adapt to the split-belt training. The exaggerated asymmetry marked by tremendous activity levels of the right VL in mid stance decreased by nearly 700% throughout the course of adaptation. Similarly, the first minute of adaptation in terminal swing resulted in exaggerated left VL activity that trended to near-baseline levels by the end of the adaptive phase.

In contrast to past studies, no significant differences between the slow and fast walking conditions and the control were observed (Clancy et al., 2004; Hof et al., 2002). In agreement with Brunt, *et al.*, LG activity was significantly greater in the non-paretic, "uninvolved," or in this case "slow," limb. Peak left LG activity occurred during the first minute of the adaptive phase and remained significantly greater than the right leg and control from late stance through initial swing. These changes reflect changes in amplitude and temporal activity reported in past studies and suggest cocontraction of the LG and TA through most of stance until the second period of double support (Brunt et al., 1995). Similar to the changes noted across all muscles in this study, the magnitudes of each of these differences decreased significantly throughout the adaptive period and were not significantly different from the control by the final minute of adaptation (p<0.05), again supporting past studies suggesting the ability of human neural systems to adapt to a synchronous walking pattern (Reisman et al., 2007).

In disagreement with previous studies, no significant changes in BF activity with respect to the fast or slow gait speed conditions were observed in this study (Hesse et al., 2001; Hof et al., 2002). The observed increased activity of the left BF during the first minute of adaptation in terminal stance and mid swing as well as increased activity of the right BF during mid and terminal swing is in agreement with Den Otter, *et al.* (2007), who reported that asymmetrical walkers demonstrated extended duration of BF activity during swing, as well as extended duration of coactivation with the rectus femoris. If coactivation was not limited to the rectus femoris, this may in part explain the significant left VL activity during terminal swing.

III. Metabolic Cost

Overall, the metabolic data confirmed the hypothesis that asynchronous walking would be more metabolically-expensive than synchronous walking. This is consistent with previous studies (Chen et al., 2005b; Olney et al., 1986), which explored the relationship between mechanical energy exchange during asynchronous gait as well as mechanical energy cost during preswing and swing of the paretic limb.

The J-shaped relationship between walking speed and metabolic cost is well established and illustrates that metabolic cost is minimized at the individual's preferred gait speed (Ralston, 1958). As participants did not choose the prescribed combinations of speed and step frequency, I can assume that oxygen consumption during the synchronous conditions would be greater than the baseline conditions for a given speed. With this in mind and contrary to the findings of Hesse, *et al.* (2001), the current study found the metabolic cost of fast walking to be slightly higher than, though not significantly different from, the control, suggesting that instructing stroke patients to walk more quickly may not facilitate a less metabolically-expensive gait pattern.

This study was unique in that it addressed the energetic cost of adapting to an asynchronous gait and sought to determine the relationship between the metabolic cost of synchronous walking and asynchronous walking following that period of adaptation. The

metabolic cost increased significantly at the beginning of the adaptive phase, but a gradual decrease in metabolic cost was observed with additional time in the adaptive phase. In this study, synchronous walking resumed after sixteen minutes, at which point the metabolic cost of asynchronous walking had decreased significantly from its peak value at the beginning of adaptation, but it remained significantly greater than the control. However, the negative trend of the metabolic cost throughout the progression of the adaptation period suggests that with additional time, the metabolic cost of asynchronous walking may continue to decrease, possibly to the extent that adapted asynchronous gait is as efficient as synchronous gait. Furthermore, post-adaptive gait resulted in a metabolic cost that was significantly less than the control, suggesting a potential role for asynchronous walking in treadmill rehabilitation to yield a more metabolically efficient post-adaptive gait.

IV. Study limitations

First, this study is limited by the small number of participants. Second, participants were instructed to hold on to the side rails lightly during all trials. This aspect of the testing was not controlled, and variations between participants in the extent to which the rail was held could have impacted the collected data. Third, participants demonstrated the ability to adapt to an asynchronous walking pattern during the untied condition, resulting in kinematic, muscle activity, and metabolic cost data that approached the control by the termination of the adaptive period. Though this reinforced the results of previous research (Dietz et al., 1994; Reisman et al., 2007; Reisman et al., 2009), it did not allow for observation of continually adaptive gait. Fourth, accurate measurement of breath-by-breath oxygen consumption was limited by the collection delay inherent to the metabolic collection tube connecting the breathing apparatus to the open circuit respiratory system. Finally, the results from this study may be limited to a healthy, young

adult population free from musculoskeletal disease or injury. Though the participants' gait is not synonymous with that of a stroke patient or person with a neurological disorder, it is hoped that this limited population will serve as the normal control and basis for future comparative studies within specific patient populations.

V. Future Research

In the future, I hope to explore the kinetic aspects of asynchronous walking, particularly with respect to vertical ground reaction forces. Asymmetrical walkers, stroke patients, amputees, TBI victims, etc., often present with joint pain or injury that could be complicated by potentially high vertical ground reaction forces experienced during asynchronous locomotion. Understanding the relationship between the vertical ground reaction forces experienced during asymmetrical walking in normal controls will allow us to provide improved training and rehabilitation recommendations to patients with existing asymmetries. Furthermore, I hope to repeat the current study with the aim of preventing adaptation and the development of a new, asynchronous, steady-state walking pattern. To do so, the current adaptation phase will be replaced with a phase of constant fluctuation of the untied belts, thus preventing adaptation. Finally, I hope to assess the metabolic cost of stroke patients throughout their rehabilitation, to determine the most metabolically efficient rehabilitation strategies and the degree to which metabolic cost can be minimized with post-stroke intervention.

Chapter 5

Conclusion

Asynchronous gait resulted in decreased stride times, decreased swing time, decreased stance time, decreased periods of double support, increased step length, and increased step width on the slow, "non-paretic" limb coupled with changes on the fast leg, including increased swing time, decreased stance time, increased periods of double support, increased step length, and increased step width. Muscle activity during asynchronous walking was most noticeably increased during periods of the gait cycle beyond those typically associated with activity bursts for the respective muscle and was accompanied by a substantial degree of bilateral coactivation of opposing muscles. Finally, peak metabolic cost during asynchronous walking was 26.5% greater than the control, decreased significantly through the adaptive phase, and was 22.6% less than the control by the end of the post-adaptive phase. The data agree with previous studies that suggest that adult human neural systems are capable of adjusting to asynchronous interlimb patterns and that initially exaggerated asymmetries trend toward control values after a period of adaptation (Dietz et al., 1994; Reisman et al., 2007; Reisman et al., 2009).

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