MECHANICAL ENERGETICS OF POST-STROKE HEMIPARETIC GAIT

by

Benjamin David Roewer

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Approved: ____

Jill Higginson, Ph.D.
Professor in charge of thesis on behalf of the Advisory Committee

Approved: ____

Anette Karlsson, Ph.D.
Chair of the Department of Mechanical Engineering

Approved: ____

Michael J. Chajes, Ph.D.
Dean of the College of Engineering

Approved: ____

Charles G. Riordan, Ph.D.
Vice Provost for Graduate and Professional Education

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ABSTRACT

Stroke is the third leading cause of death in the United States. Survivors are often left with residual muscle weakness and spasticity that lead to slower self-selected walking speeds and increased metabolic energy expenditure when compared to their healthy peers. They also exhibit asymmetric movement patterns and compensatory strategies which may adversely affect mechanical work production. *Objective:* This study examined the effect of speed modulation on mechanical work production and mechanical recovery in post-stroke gait. It also investigated the ability of post-stroke kinematics and kinetics to predict which stroke survivors have the capacity to increase their mechanical recovery. Ten chronic stroke survivors and 6 healthy young adults were recruited for this study. Kinematic data were collected while all subjects walked on an instrumented split-belt treadmill at speeds slower and faster than their self-selected speed. Internal work, external work, mechanical recovery, circumduction, swing asymmetry, paretic ankle and hip work, peak knee flexion during swing, peak hip extension, and paretic leg kinetic energy at toe-off were measured or calculated. Spearman rank correlation analysis was used to detect significant relationships among all gait variables of interest. Wilcoxon signed rank tests were used to detect individual changes in gait variables across the ranges of increasing and decreasing mechanical recovery. Mechanical recovery exhibited a parabolic relationship with walking speed in healthy adults with peak recovery occurring near self-selected walking speed. Mechanical recovery improved in most

stroke subjects at walking speeds up to 1 m/s. Increased mechanical recovery was also accompanied by increased peak knee flexion, paretic ankle work, and paretic leg kinetic energy. Only paretic leg kinetic energy was able to predict mechanical recovery post-stroke. Internal work production in the frontal plane increased linearly with walking speed in healthy adults, but not in stroke survivors. Speed modulation up to 1 m/s is critical for achieving optimal mechanical recovery in stroke survivors. Stroke survivors with the slowest self-selected walking speeds benefit most from speed modulation regardless of the strategies they employ to clear the paretic limb during swing. The use of compensation strategies may actually be beneficial because they allow faster walking speeds to be attained without significantly increasing mechanical work production.

Chapter 1

INTRODUCTION

1.1 Background

Nearly 800,000 people suffer from a stroke each year in the United States (AHA 2008). It is the leading cause of disability in the United States and the third leading cause of death (Association 2008). The effects of stroke can vary and depend on the severity of the stroke insult and its location in the brain (Stein, Harvey et al. 2009). The effects of stroke can include blurry vision, slurred speech, changes in behavior, and altered movement patterns (Stein, Harvey et al. 2009). This thesis addresses the individual and cumulative effects of changes in movement patterns and mechanical work production in post-stroke hemiparetic gait.

The metabolic energetic cost of walking has been shown to increase post-stroke (Zamparo, Francescato et al. 1995; Detrembleur, Dierick et al. 2003; Reisman, Rudolph et al. 2009). Increased metabolic energy cost has been attributed to hip-hiking compensation strategies and poor mechanical energy exchange between potential and kinetic energy of the whole-body COM, but has only been studied at self-selected walking speeds (Olney 1996).

Mechanical energy represents the ability of a body to perform work. From classical mechanics, the total energy of a body is the sum of its potential energy, linear kinetic energy, and rotational kinetic energy. Mechanical energy analysis has been employed in previous studies to evaluate work production during normal walking

(Cavagna, Thys et al. 1976; Cavagna and Kaneko 1977; Burdett, Skrinar et al. 1983; Willems, Cavagna et al. 1995; Donelan, Kram et al. 2002; Doke, Donelan et al. 2005; Mian, Thom et al. 2006; Ortega and Farley 2007), relate mechanical work and metabolic energy consumption during normal and pathological walking (Burdett, Skrinar et al. 1983; Detrembleur, Dierick et al. 2003; Mian, Thom et al. 2006; Ortega and Farley 2007), and estimate energy transfers between body segments (Aleshinsky 1986; Willems, Cavagna et al. 1995; Doke, Donelan et al. 2005). Mechanical analysis based on kinematics is limited because it cannot measure simultaneous positive and negative work production, musculo-tendon work, or isometric work against gravity (Winter 2005). Despite some of its inherent limitations, this approach is simple to employ and has shown promise in evaluating increased mechanical work production in pathological populations (Olney, Monga et al. 1986; McGibbon, Krebs et al. 2001; McGibbon, Puniello et al. 2001; Detrembleur, Dierick et al. 2003; Bennett, Abel et al. 2005; Chen and Patten 2008). To perform these calculations total mechanical work must be separated into components internal and external to the body.

Internal energy of the body is comprised of the kinetic energy of the limbs with respect to the whole body center of mass (COM). Positive internal work is calculated by summing the positive changes in internal energy; however, assumptions must be made about the types of energy transfers that occur. Willems et al. concluded that during healthy walking energy transfers primarily occur between the segments of the ipsilateral limb (Willems, Cavagna et al. 1995). For example, during swing the thigh may transfer some of its kinetic energy to the shank. This can be seen as the thigh decreasing its linear or rotational velocities and the shank increasing its linear or rotational velocities. Ignoring these energy transfers will result in an overestimation of internal work (Willems, Cavagna et al. 1995).

It has been shown that healthy adults produce more internal work in the sagittal plane when they increase their walking speed (Willems, Cavagna et al. 1995; Mian, Thom et al. 2006; Ortega and Farley 2007). Mian et al. showed that internal work in the frontal plane is a negligible contributor to total work production in healthy adults (Mian, Thom et al. 2006). Increased frontal plane limb movement is a characteristic of post-stroke gait, yet only one attempt has been made to evaluate internal work in stroke populations (Detrembleur, Dierick et al. 2003). Detrembleur concluded that stroke survivors with faster self-selected walking speeds produce less total internal work in the frontal plane than those with slower self-selected speeds (Detrembleur, Dierick et al. 2003).

External work evaluates the work required to support and propel the COM against external forces. During normal walking, ground reaction forces are transmitted through the legs to support and propel the COM as gravity acts to pull it down. Energy transfers can also occur among the components of COM energy. In early stance, forward kinetic energy of the COM is used to propel itself over the stance leg and increase its potential energy. The COM falls forward over the stance leg as potential energy is converted back into forward kinetic energy, and the process continues. Taking these properties into account, energy of the COM (E_{com}) is calculated as the sum of whole body potential and kinetic energy.

Not all E_{com} is conserved from step to step. During heel strike, negative work occurs as the leading leg acts to brake the COM. To overcome this braking force the trailing limb produces positive work. Energy lost during the braking process and other inefficient movements can act to reduce the recovery of mechanical energy. This percent of mechanical energy recovery can be quantified by separating the work done on the COM into its orthogonal components and then subtracting the total external work applied to the system after accounting for energy transfers.

In healthy walkers, mechanical recovery has been assessed across a range of walking speeds. It has been found that peak mechanical recovery (65-70%) occurs near self-selected walking speed (SSWS) (Cavagna and Kaneko 1977; Mian, Thom et al. 2006). In stroke populations, mechanical recovery has only been assessed at each subject's SSWS (Detrembleur, Dierick et al. 2003). It has been shown that stroke survivors with faster SSWS have greater mechanical recovery. Post-stroke treatment paradigms aim to increase walking speed, yet previous research has not evaluated whether changes in walking speed affect mechanical work production and mechanical recovery. This led to the development of Aim 1.

AIM 1: Assess how mechanical work production and mechanical recovery change when healthy and stroke subjects modulate their walking speed.

Hypothesis 1.1: Healthy mechanical recovery will peak at self-selected walking speed. Stroke mechanical recovery will continually increase with increased walking speed.

Hypothesis 1.2: Stroke subjects will produce greater frontal plane internal work than healthy subjects.

Slower walking speed and asymmetric movement patterns are common characteristics of post-stroke gait (Olney 1996; Chen, Patten et al. 2005). One primary consequence of these altered movement patterns is decreased recovery of mechanical energy (Detrembleur, Dierick et al. 2003). Stride-to-stride mechanical recovery is an essential energy-preserving mechanism in pendulum walking models, and in healthy adults it is maximized at self-selected walking speed (Cavagna and Kaneko 1977; Mian, Thom et al. 2006). In stroke populations a relationship has not yet been firmly established between walking speed and mechanical recovery, however, it appears that faster self-selected speed is associated with greater mechanical recovery (Detrembleur, Dierick et al. 2003). Some of the primary goals of post-stroke physical therapy are to increase walking speed and improve symmetry of movement. Previous studies have established some relationships between the two, yet the parameters responsible for improved mechanical energy expenditure are unclear.

Wall reported no relationship between preferred walking speed and single-support stance time asymmetry (Wall and Turnbull 1986). However, Chen reported a significant improvement in swing-time symmetry and an increase in paretic leg kinetic energy at toe-off when walking speed was increased by 30% over self-selected (Chen, Patten et al. 2005). Reduced capacity of the paretic plantar flexors to support and propel the body may be the root cause of this asymmetry (Higginson, Zajac et al. 2006). It is also a primary inhibitor to increasing walking speed (Bowden, Balasubramanian et al. 2006; Jonkers, Delp et al. 2007), but it can be augmented by increasing hip flexion to propel the paretic limb during swing (Jonkers, Delp et al. 2007; Lewis and Ferris 2008). Therefore, stroke patients who have the capacity to increase their walking speed by increasing plantar flexor output, hip flexion output, or a combination of the two during pre-swing, may also improve mechanical recovery.

Reduced plantar flexion strength during pre-swing has also been related to reduced knee flexion during swing (Anderson, Goldberg et al. 2004). The resulting stiff-knee gait necessitates the employment of hip hiking and circumduction strategies to clear the paretic leg during swing (Wall and Turnbull 1986; Olney 1996; Kerrigan, Frates et al. 2000; Stein, Harvey et al. 2009). To evaluate these strategies, Cruz used a stepwise regression model to predict gait speed and pelvic obliquity velocity – a possible facilitator of circumduction - from post-stroke changes in gait (Cruz and Dhaher 2009). Similar analyses have been used to evaluate changes in gait due to other disabilities. McGibbon was able to use joint power analysis to identify hip and low-back strategies in older adults with and without orthopedic or neurologic injury (McGibbon, Krebs et al. 2001; McGibbon, Puniello et al. 2001).

The extent to which the aforementioned movements present themselves can vary from patient to patient (Wall and Turnbull 1986; Kerrigan, Frates et al. 2000). Consequently, physical therapy protocols to increase walking speed and reduce asymmetrical movement patterns may not be best suited for all stroke patients. While increasing walking speed and improving symmetry are considered successful outcomes of post-stroke physical therapy, what are the implications of those outcomes on mechanical recovery? If they come with a cost of reducing mechanical recovery then endurance could be compromised. Additionally, increasing walking speed or reducing asymmetries may not address the individual needs of each patient (Wall and Turnbull 1986). By understanding the relationships among changes in mechanical recovery and changes in movement patterns, physical therapists might be better able to prioritize which post-stroke gait deviations to address. This led to the development of Aim 2.

AIM 2: Identify the relationships among stroke gait parameters and determine which of those parameters are the best predictors of mechanical recovery.

Hypothesis 2.1: Plantar flexor work of the paretic ankle during pre-swing and reduced circumduction will best predict mechanical recovery.

Hypothesis 2.2: Increased walking speed will correlate positively with symmetry, paretic limb work, and paretic leg kinetic energy.

Hypothesis 2.3: Paretic ankle work and paretic hip work will correlate positively with paretic leg kinetic energy and peak knee flexion during swing.

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Chapter 2

EFFECT OF SPEED MODULATION ON MECHANICAL WORK PRODUCTION IN POST-STROKE HEMIPARETIC GAIT

2.1 Introduction

Chronic symptoms of stroke often include decreased self-selected walking speeds and asymmetric movement patterns (Olney 1996; Chen, Patten et al. 2005). Improvements in self-selected walking speeds relate to stroke survivors being better able to complete activities of daily living and lead independent lives (Perry, Garrett et al. 1995). Asymmetric movement patterns in stroke survivors and their use of compensation strategies can lead to greater energy expenditure and less endurance than their healthy peers (Pohl, Duncan et al. 2002; Detrembleur, Dierick et al. 2003). Training stroke survivors to walk at faster speeds can improve gait symmetry and lead to more energy-efficient movement patterns (Lamontagne and Fung 2004; Chen, Patten et al. 2005; Reisman, Rudolph et al. 2009).

Metabolic energy analyses have shown that after a stroke the muscles may still have the same work production capability as healthy adults (Detrembleur, Dierick et al. 2003), yet they consume more metabolic energy when walking at matched speeds (Zamparo, Francescato et al. 1995; Waters and Mulroy 1999). This suggests that increased energy consumption may be a result of inefficient movement patterns. Many studies have noted inefficient and asymmetric movement patterns by examining kinematic and spatiotemporal data (Olney 1996; Chen, Patten et al. 2005; Chen, Patten et al. 2005), but few studies have examined the mechanical work production in persons after stroke (Olney, Monga et al. 1986; Detrembleur, Dierick et al. 2003). Those studies were performed only at self-selected speeds and did not evaluate the effects of speed modulation. Mechanical analyses can reveal the effectiveness of the pendulum-like movement of the whole-body center of mass and the individual sources of energy expenditure.

Mechanical recovery is one measure that identifies how much mechanical energy is preserved from step to step through the normal pendulum-like motion of the whole-body center of mass (COM). Studies performed on healthy subjects have revealed that mechanical recovery is maximal (~65%) near self-selected walking speed (SSWS) indicating optimal exchange of potential and kinetic energy (Cavagna, Thys et al. 1976; Mian, Thom et al. 2006). At speeds slower and faster than SSWS, mechanical recovery decreases (Cavagna and Kaneko 1977; Burdett, Skrinar et al. 1983; Willems, Cavagna et al. 1995; Mian, Thom et al. 2006). Very few studies have examined mechanical work production in stroke survivors. A study by Olney et al. (1986) reported that 10 adults with post-stroke hemiparesis exhibited lower amounts of mechanical recovery (22-62%) than healthy walkers when walking at their preferred walking speed. These reduced mechanical recovery values were explained by the in-phase forward and vertical components of mechanical work in the stroke survivors, and by walking speeds which were too slow to create enough forward kinetic energy to create a useful mechanical energy exchange (Olney, Monga et al. 1986). A study by Detrembleur et al. (2003) found similar mechanical recovery values (25-70%), with increased mechanical recovery exhibited by stroke survivors with faster self-selected speeds (Detrembleur, Dierick et al. 2003). Again, it was hypothesized that the slower walkers did not walk fast enough to create a useful mechanical energy exchange; however, neither of these studies explored the effect of increasing speed on recovery within individuals. The use of compensation strategies may have also disrupted the natural energy exchange as well as increased the amount of mechanical work performed, but these differences have not been quantified.

Internal work measures the amount of work performed by the limbs with respect to the whole-body center of mass. By evaluating internal work, the contributions of the legs to total mechanical work production can be evaluated. In healthy subjects, internal work of the legs increases linearly with increased walking speed above SSWS (Cavagna and Kaneko 1977; Mian, Thom et al. 2006). Internal work of the legs is primarily done in the sagittal plane, with less than 5% of all mechanical work comprised of frontal-plane internal work of the legs (Mian, Thom et al. 2006). Detrembleur et al. (2003) reported that frontal plane internal work and total internal work was reduced in stroke subjects when they had faster self-selected walking speeds (Detrembleur, Dierick et al. 2003).

One primary limitation of the previous studies is that most mechanical energy analyses were performed in the sagittal plane (Burdett, Skrinar et al. 1983; Olney, Monga et al. 1986; Olney 1996). The only study that included mechanical work in the frontal plane did so at self-selected walking speeds (Detrembleur, Dierick et al. 2003). Furthermore, a cross-sectional sample of stroke survivors has been used to illustrate how speed is related to mechanical measures instead of studying how post-stroke individuals respond to increases in walking speed. In the current study, we investigated the effect of modulating walking speed on mechanical energy expenditure in healthy young adults and individuals with post-stroke hemiparesis. Our aim was to assess how mechanical energy expenditure and mechanical recovery would change when healthy and stroke subjects walked at speeds faster and slower than their self-selected walking speed. Our hypothesis was that stroke survivors walk at a self-selected speed which is slower than the speed that corresponds to their optimal mechanical recovery. Based on this hypothesis, we expected to see an increase in mechanical recovery when stroke survivors increased their walking speed. We also hypothesized that stroke survivors would produce greater amounts of internal work in the frontal plane than the healthy subjects because of the use of compensatory strategies to clear the foot during swing phase. We expected that frontal plane internal work would decrease with increased walking speeds above self-selected speed. Our long-term objective is to understand which characteristics of post-stroke gait contribute to poor recovery of mechanical energy and increased mechanical work.

2.2 Materials and Methods

2.2.1 Subject Population

Chronic stroke survivors were recruited from local physical therapy clinics in the greater Philadelphia area. Subjects were included if they were between 30 and 80 years of age, suffered from their first and only cortical or sub-cortical stroke at least 6 months prior, were ambulatory with some gait deficit, and were able to walk for at least five minutes at their SSWS. Subjects were excluded if they exhibited signs of a cerebellar stroke, had uncontrolled hypertension (> 190/110 mmHg), peripheral artery disease with claudication, active cancer, pulmonary or renal failure, unstable angina, severe aphasia, or dementia (Mini-Mental State exam score < 22).

Healthy subjects were recruited from the community at the University of Delaware. Subjects were included if they were between 18 and 40 years of age with no history of musculoskeletal or neurological injury. Subjects were excluded if they were pregnant at the time of collection, had muscle or nervous system disorders, or had Body Mass Index (BMI) of 30 or greater.

All subjects gave informed consent for this study which was approved by the University of Delaware Human Subjects Review Board.

2.2.2 Data Collection – Stroke Subjects

SSWS was calculated using a handheld stopwatch while subjects walked unassisted down a six-meter walkway at their preferred speed. Subjects were allowed to wear prescribed ankle-foot orthoses. A total of 41 reflective markers were placed on the bony landmarks of the foot, shank, thigh, pelvis, and trunk in order to record the motions of each body segment during the walking trials. Subjects walked on an instrumented split-belt treadmill (AMTI, Watertown, MA), with speeds tied. Subjects were allowed to use a handrail and wore an overhead harness (no body weight support) for safety. Two 20-second walking trials were performed at multiple walking speeds. Walking speeds were determined by the overground SSWS and the fastest speed that each subject could safely walk on the treadmill (FAST). Each stroke subject performed walking trials at the following speeds: 80% of SSWS, SSWS, FAST, and two or three intermediate speeds that were equally partitioned between the SSWS and FAST walking speeds. The order in which each walking speed was performed was chosen at random to reduce any potential fatigue effects. The 3D positions of each marker were collected and recorded using an 8 camera Vicon motion capture system (Vicon MX, Lake Forest, CA) sampling at 120 Hz.

2.2.3 Data Collection – Healthy Subjects

SSWS was calculated using a handheld stopwatch while subjects walked unassisted down a six-meter walkway at their preferred speed. A total of 21 reflective markers were placed on the bony landmarks of the foot, shank, thigh, pelvis, and trunk in order to record the motions of each body segment during the walking trials. Subjects walked on an instrumented split-belt treadmill (Bertec Corp., Columbus, OH), with speeds tied. One 20-second walking trial was performed at six different walking speeds. Walking speeds were determined by the overground SSWS. Each subject performed walking trials at the following speeds: 25% of SSWS, 50% of SSWS, 75% of SSWS, SSWS, 125% of SSWS, and 200% of SSWS. The 3D positions of each marker were collected using 6 Motion Analysis cameras sampling at 60Hz.

2.2.4 Data Processing

8-segment subject-specific models were created in Visual 3D v4.99 (Rockville, MD). Models were comprised of two feet, two shanks, two thighs, a pelvis, and a Head-Arm-Trunk (HAT) segment. The mass and inertial properties of each body segment were determined using anthropometric data (Winter 2005). The COM of each body segment was calculated from the marker position data in Visual 3D. The COM of each segment was filtered using a low-pass Butterworth filter with a cutoff frequency of 6Hz in order to reduce the noise that can result from differentiating the COM displacements. Segment velocities and accelerations were calculated by 1st and 2nd differentiation of the limb COM positions relative to the whole-body COM.

All energy and work calculations were made during walking trials between the first and last heel strikes of the right leg. Initial contact was determined using kinematic marker data from the foot and pelvis (Zeni, Richards et al. 2008). All work calculations were normalized to body mass (kg) and distance walked (m). Distance walked for each trial was determined by multiplying treadmill speed with amount of time between the first and last heel strikes.

Internal work was divided into anterior-posterior (W_{int_ap}) and medial-lateral (W_{int_ml}) components (i.e. sagittal plane and frontal plane components, respectively). First, we calculated the anterior-posterior, vertical and rotational kinetic energies of each leg segment relative to the whole-body COM. Energy transfers were assumed to occur between segments of the ipsilateral limb in the sagittal plane (Cavagna and Kaneko 1977). Total energy for each leg was calculated as follows:

$$KE_{leg\_ap} = (\tfrac{1}{2}mv^2{}_{ap} + \tfrac{1}{2}mv^2{}_{v} + \tfrac{1}{2}Iw^2)_{foot} + (\tfrac{1}{2}mv^2{}_{ap} + \tfrac{1}{2}mv^2{}_{v} + \tfrac{1}{2}Iw^2)_{shank} + (\tfrac{1}{2}mv^2{}_{ap} + \tfrac{1}{2}mv^2{}_{v} + \tfrac{1}{2}Iw^2)_{thigh} \qquad 1$$

where *m* is the mass of the body segment, *v* is the velocity of the segment COM in the sagittal plane, *I* is the momentum of inertia of the segment, and *w* is the angular velocity of the segment. Similarly, energy of the leg in the frontal plane (KE_{leg_ml}) was calculated as follows:

$$KE_{leg\_ml} = (\tfrac{1}{2}mv^2{}_{ml})_{foot} + (\tfrac{1}{2}mv^2{}_{ml})_{shank} + (\tfrac{1}{2}mv^2{}_{ml})_{thigh} \qquad 2$$

Any positive changes in KE_{leg_ap} and KE_{leg_ml} were assumed to be caused by positive mechanical work. Therefore, W_{int_ap} and W_{int_ml} was calculated in Matlab (v R2006b) by summing all positive increments in KE_{leg_ap} and KE_{leg_ml} respectively. Energy exchanges were not assumed to occur between segments because we believe that the vertical displacements of the segments were primarily used to accelerate the limbs in the forward direction.

W_{ext} describes the work done on the whole-body COM by external forces acting through the legs. W_{ext} was calculated using a kinematic method because it is easy to implement and provides accurate results (Gard, Miff et al. 2004). W_{ext} was determined by first calculating the kinetic and potential energy of the whole-body COM (E_{com}). Energy exchanges were assumed to occur in the vertical, anterior-posterior, and medial-lateral directions. E_{com} was calculated as follows:

$$E_{com} = mgh + \frac{1}{2}mv_v^2 + \frac{1}{2}mv_{ap}^2 + \frac{1}{2}mv_{ml}^2 \qquad 3$$

where *m* is the mass of the whole body, *g* is the gravitational constant, *h* is the height of the whole-body COM, and v_{ap}, v_v, and v_{ml} are the linear velocities of the whole body COM in the anterior-posterior, vertical, and medial-lateral directions, respectively. External work (W_{ext}) was calculated by summing the positive increments of E_{com} from the first to last right heel strikes of each gait trial. The vertical, anterior-posterior, and medial-lateral components of E_{com} were also calculated in order to compute the recovery index. Those components were calculated as follows:

$$E_v = mgh + \frac{1}{2}mv_v^2 \qquad 4$$

$$E_{ap} = \frac{1}{2}mv_{ap}^2 \qquad 5$$

$$E_{ml} = \frac{1}{2}mv_{ml}^2 \qquad 6$$

W_v, W_{ap}, and W_{ml} were calculated by summing the positive changes in E_v, E_{ap}, and E_{ml} respectively.

The mechanical recovery index is a measure that describes the completeness of the energy exchange between the vertical, anterior-posterior, and medial-lateral directions of the whole-body COM (Detrembleur, Dierick et al. 2003). In this study, mechanical recovery was calculated as follows:

$$R = \frac{W_v + W_{ap} + W_{ml} - W_{ext}}{W_v + W_{ap} + W_{ml}} * 100\% \qquad 7$$

where W_v, W_{ap} and W_{ml} are the vertical, anterior-posterior, and medial-lateral components of W_{ext} respectively.

Based on previous findings (Cavagna and Kaneko 1977; Willems, Cavagna et al. 1995; Detrembleur, Dierick et al. 2003; Mian, Thom et al. 2006), linear regression analysis was used to assess the effect of walking speed on W_{int_ap} and W_{int_ml} in healthy adults and stroke survivors. A 2nd order polynomial regression was used to analyze W_{ext} and mechanical recovery in healthy adults (Cavagna and Kaneko 1977; Willems, Cavagna et al. 1995; Detrembleur, Dierick et al. 2003; Mian, Thom et al. 2006). Linear and 2nd order polynomial regressions were used to analyze W_{ext} and mechanical recovery in stroke survivors in order to identify the best fit (Detrembleur, Dierick et al. 2003). Linear regression was included in this case because of our hypothesis that stroke survivors may show improvements in mechanical recovery, thus decreasing their W_{ext} when walking at speeds faster than SSWS. A priori significance was set at .05.

2.3 Results

Six healthy adults (age: 25 ± 7 years, SSWS: 1.15 ± 0.13 m/s) and 10 stroke survivors were tested (Table 2.1). Overall, the stroke survivors had slower SSWS (0.72 ± 0.14 m/s) than the healthy subjects. Healthy subjects exhibited work and recovery patterns similar to previous studies (Cavagna and Kaneko 1977; Mian, Thom et al. 2006). For stroke survivors, increasing walking speed did not have the same effect on internal work, external work and mechanical recovery.

Table 2.1 Stroke subject clinical data. Abbreviations: LE - lower extremity, SSWS - self-selected walking speed, FAST = fastest walking speed.

Stroke Subject Clinical Data							
Subj #	Age (yr)	Sex	Time since stroke	LE Fugl Meyer Score	AFO	SSWS (m/s)	FAST (m/s)
1	47	M	17	23	No	0.80	1.39
2	52	M	40	24	Yes	0.81	1.43
3	47	M	18	16	Yes	0.63	1.70
4	61	M	17	28	No	0.81	1.16
5	66	F	7	22	No	0.40	0.58
6	72	M	20	28	No	0.80	1.20
7	71	M	59	19	Yes	0.70	0.90
8	77	M	28	19	No	0.70	1.10
9	75	F	22	19	No	0.90	1.10
10	45	M	30	27	No	0.70	1.30

2.3.1 Internal Work

Wint_ap production of the healthy subjects increased linearly with walking speed (Figure 2.1 A; r2 = .925, p<.05). Across the range of walking speeds, Wint_ap increased from 0.08 to 0.55 J/kgm.

In the group of stroke survivors, Wint_ap production also increased linearly with walking speed (Figure 2.1 B; r2 = .497, p<.05). Across the range of walking speeds, Wint_ap increased from 0.09 to 0.33 J/kgm.

Wint_ml production of the healthy subjects increased linearly with walking speed. The relationship with walking speed was not as strong as in the sagittal plane, but still significant (Figure 2.1 C; r2 = .632, p<.05). Wint_ml ranged from 0.007 to 0.060 J/kgm over the range of walking speeds.

Wint_ml production was not significantly correlated with walking speed in the stroke population when viewed as a whole (Figure 2.1 D; r2 = .004, p = .66). Wint_ml ranged from 0.005 to 0.037 J/kgm across speeds.

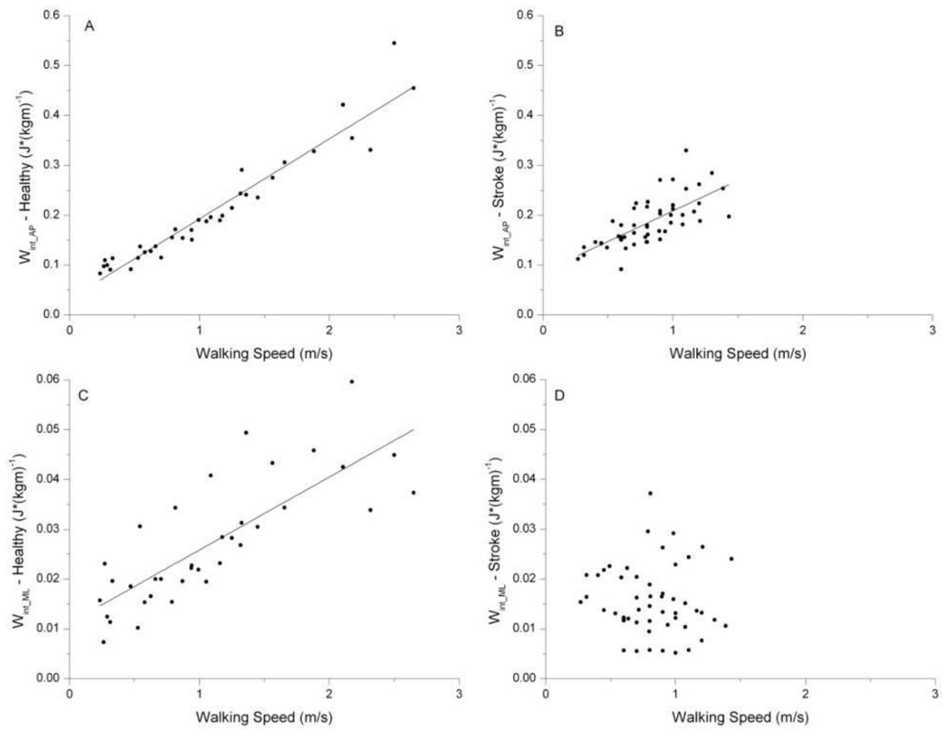

Figure 2.1 Internal work production for healthy (left) and stroke (right) subjects. Figures A and B represent lower-limb internal work in the anterior-posterior direction (W_{int_ap}). Figures C and D represent lower-limb internal work in the medial-lateral direction (W_{int_ml}). Linear regression lines are shown in solid black.

2.3.2 External Work

A 2nd order polynomial regression found a significant relationship between walking speed and W_{ext} production in the healthy subjects (Figure 2.2 A; $r^2 = .368$, $p<.05$). W_{ext} ranged from 0.19 to 0.50 J/kgm over the range of walking speeds.

W_{ext} production was not significantly explained by changes in walking speed in the stroke survivors (Figure 2.2 B) using either a linear regression ($r^2 = .032$, p=.21) or a 2nd order polynomial regression (r^2 = .082, p=.132). W_{ext} ranged from 0.26 to 0.62 J/kgm across speeds in that group.

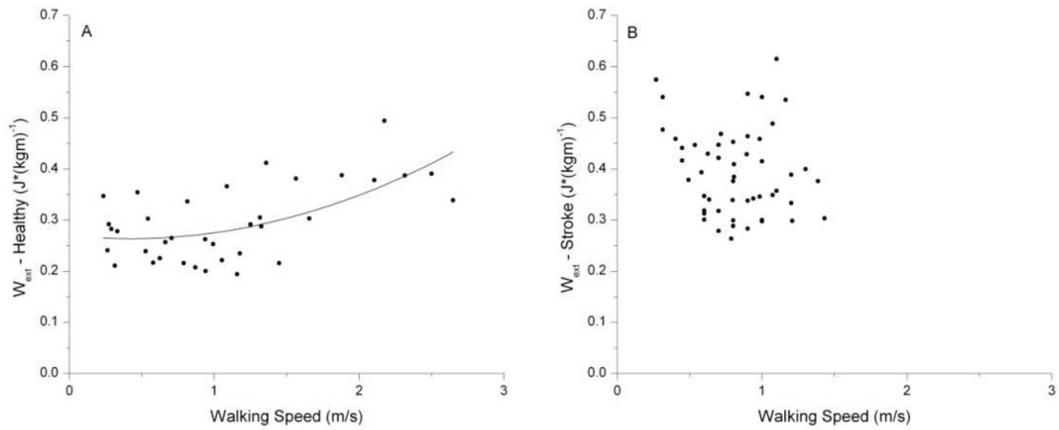

Figure 2.2 External work production for (A) healthy and (B) stroke subjects. A 2nd order polynomial regression found a significant relationship (r2 = .368, p<.05) between external work and walking speed in healthy adults (regression line shown in solid black). No significant relationship was found in stroke survivors.

2.3.3 Mechanical Recovery

For the healthy subjects, a significant relationship was found between walking speed and mechanical recovery. The data were best fit by a 2nd order polynomial regression (Figure 2.3 A; r^2 = .636, p<.05) which agrees with previous findings (Cavagna and Kaneko 1977; Mian, Thom et al. 2006). Mechanical recovery peaked (64-75%) around SSWS (1.05-1.66 m/s). Mechanical recovery decreased when walking faster or slower than SSWS.

Stroke mechanical recovery ranged from 33-62% across speeds (Figure 2.3 B). It peaked (44-62%) between 0.54 and 0.94 m/s. Mechanical recovery was analyzed using both a linear (r^2 = .020, p = .319) and a 2nd order polynomial regression (r^2 = .021, p = .597). Both types of regression were analyzed because we hypothesized that the stroke subjects might not walk fast enough to exhibit the same parabolic relationship between walking speed and mechanical recovery as the healthy subjects. In other words, the stroke mechanical recovery value might only reflect the linearly increasing portion of the parabolic curve. However, neither the linear or 2nd order polynomial regressions revealed a significant relationship between mechanical recovery and walking speed.

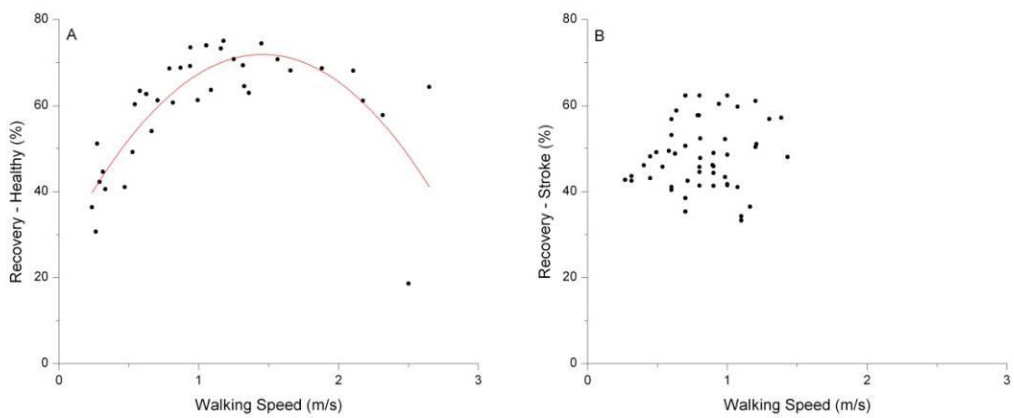

Figure 2.3 Mechanical recovery for (A) healthy and (B) stroke subjects. A 2nd order polynomial regression found a significant relationship (r2 = .636, p<.05) between mechanical recovery and walking speed in healthy adults (regression shown with solid line). No significant relationship was found in stroke survivors.

We examined the effect of walking speed on mechanical recovery for each stroke survivor (Figure 2.4). Three of the stroke survivors were not able to reach 1 m/s; they were classified as slow walkers (Figure 2.4 A). They reached optimal mechanical recovery (46-50%) when walking at speeds between 0.54 and 0.9 m/s. These speeds were all faster than self-selected. The other seven stroke survivors were able to walk on the treadmill at speeds faster than 1 m/s; they were classified as fast walkers (Figure 2.4 B). Only one of those seven fast walkers improved their mechanical recovery (from 49% to 50%) when walking faster than 1 m/s. The other six fast walkers had diminished recovery above 1 m/s. Among those seven fast walkers, optimal recovery (44-62%) occurred between 0.60 - 0.94 m/s. Again, these represent speeds that were faster than self-selected.

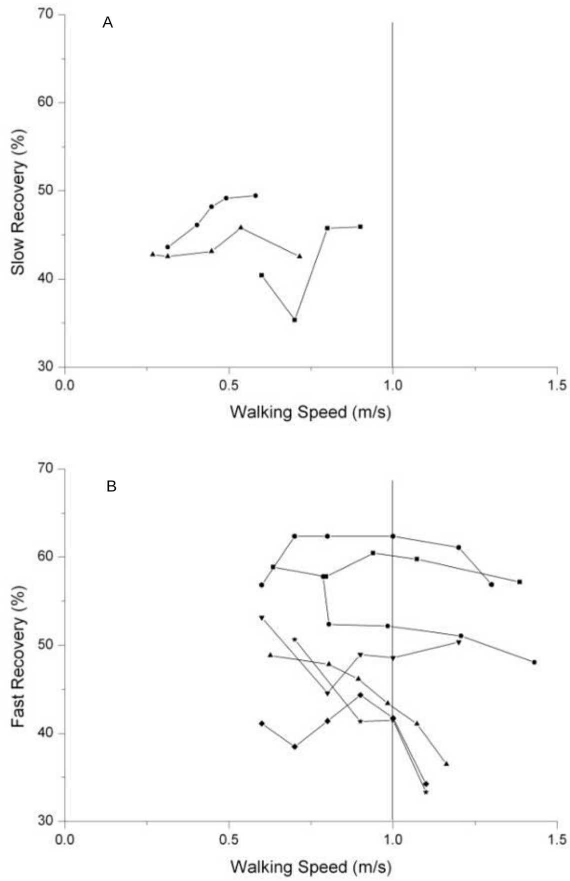

Figure 2.4 Mechanical recovery grouped by stroke subjects who (A) did not achieve 1m/s walking speed (slow walkers), and those who (B) did achieve 1 m/s (fast walkers). All slow walkers increased their mechanical recovery when walking faster than SSWS. For fast walkers, mechanical recovery varied when walking faster than SSWS. Mechanical recovery diminished at speeds >1m/s for all but one fast walker.

2.4 Discussion

The aim of this study was to understand how mechanical work production and mechanical recovery change when healthy adults and stroke survivors walk at speeds faster and slower than their SSWS. In healthy adults, we found that internal work of the lower limbs increased linearly with walking speed, and mechanical recovery was optimal around SSWS. While stroke survivors exhibited more frontal plane limb moment than healthy adults (Olney 1996), it comprised a negligible component of overall mechanical work. We also discovered that for stroke survivors, mechanical recovery generally improved when walking at speeds faster than SSWS, but consistently declined at speeds above 1 m/s.

Internal work in the sagittal plane increased steadily with increased walking speed between the healthy and stroke groups. Previous studies have also noted the same relationship in healthy populations (Cavagna and Kaneko 1977; Willems, Cavagna et al. 1995; Mian, Thom et al. 2006). In the frontal plane, more apparent differences emerged between groups. Healthy subjects increased their frontal plane internal work steadily with increased walking speed, but this was not observed in the stroke survivors. It was expected that stroke survivors might have elevated internal work in the frontal plane due to limb movement brought on by the use of frontal plane swing phase gait compensations such as hip hiking and circumduction. With increased walking speed, frontal plane internal work of the stroke survivors was equal to or less than the healthy adults. This is consistent with recent findings suggesting that hip hiking and circumduction do not increase with increased walking speed (Malecka C 2008). It should be noted that all stroke survivors held a handrail and exhibited minimal arm swing during their walking trials which could have lead to limited frontal plane movement and improved gait kinematics (Chen, Patten et al. 2005). It is also unknown how AFO usage among stroke survivors affected gait kinematics. Regardless of these patterns and limitations, W_{int_ml} was small, comprising less than 10% of total internal work production and less than 5% of total work production (Mian, Thom et al. 2006).

A significant relationship was identified between walking speed and external work production in the healthy adults. As healthy adults increased their walking speed above self-selected speed, the exchange of potential and kinetic energy became less efficient; external work in the vertical and forward directions moved out of phase as walkers approach running speeds (Cavagna and Margaria 1966; Cavagna and Kaneko 1977). No such speed-external work relationship was found in the group of stroke survivors. It is possible that this relationship was not observed because of the limited increase in speed that could be accomplished by this group.

Healthy adults produced 0.19 to 0.50 J/kgm of external work across walking speeds, which agrees with previous findings (Willems, Cavagna et al. 1995; Mian, Thom et al. 2006). The stroke survivors exhibited slightly elevated external work compared to the healthy adults. External work values for the stroke group ranged from 0.28 and 0.55 J/kgm at SWSS, which is similar to those found previously (Detrembleur, Dierick et al. 2003). Greater external work indicates greater amount of work required to support and accelerate the COM during walking at a given walking speed. Increased external work production among the stroke survivors may reflect a poor energy exchange between the vertical, forward, and lateral components of external work regardless of speed. Although the post-stroke subjects were significantly older than the healthy adults, Mian et al. (2006) reported that total mechanical work was similar in young and older healthy men suggesting that age is not responsible for our findings (Mian, Thom et al. 2006).

For healthy adults, the relationship between mechanical recovery and walking speed was similar to results reported previously (Cavagna, Thys et al. 1976; Mian, Thom et al. 2006). Mechanical recovery was optimal (60-75%) between 1.09 and 1.66 m/s which corresponded with speeds between 100-125% of self-selected for that group. Mechanical recovery decreased at faster and slower speeds. Previous studies only evaluated mechanical recovery with energy exchanges in the sagittal plane (Cavagna, Thys et al. 1976; Willems, Cavagna et al. 1995; Mian, Thom et al. 2006). We included frontal plane energy exchanges because we hypothesized that excessive frontal plane movement in the stroke survivors might explain some of the presumed decreases in mechanical recovery. We expected to see an increase in mechanical recovery with increased walking speed because we hypothesized that stroke subjects choose a slower, sub-optimal SSWS; however, changes in stroke mechanical recovery did not appear to be significantly explained by changes in walking speed. Differences in ambulation ability may have altered mechanical recovery trends among stroke survivors. To evaluate this we explored individual traces in mechanical recovery. We noted that improvements in recovery were limited at faster walking speeds (Figure 2.4B). All of the slow walkers improved their mechanical recovery when walking at faster speeds. Results varied among the fast walkers, but at about 1 m/s mechanical recovery generally diminished. These results suggest that speed modulation may be most beneficial for stroke survivors with slower walking speeds. These results are consistent with changes in energy efficiency observed at faster walking speeds post-stroke (Reisman, Rudolph et al. 2009).

2.4.1 Clinical Implications

This study demonstrated that stroke survivors have the capacity to increase mechanical work production and improve mechanical recovery by increasing walking speed. Instead of analyzing a cross-sectional sample of stroke survivors walking at self-selected speeds, we quantified the change in measures of mechanical work that can occur due to walking at faster speeds. Others have determined that walking economy, gait symmetry, and gait kinematics can improve by walking faster, especially in lower functioning stroke survivors (Lamontagne and Fung 2004; Reisman, Rudolph et al. 2009). Together these results suggest that stroke survivors who have the slowest SSWS might benefit the most from speed modulation. Adequate speed modulation should be augmented with proper measures to ensure that stroke survivors can maintain proper balance and safety, improve confidence, improve cardiovascular fitness, and perform sufficiently in other tasks that might make them more effective community ambulators (Lord and Rochester 2005).

2.4.2 Conclusion

Intersegmental coordination and adequate walking speed are crucial for maintaining fluidity of movement patterns and the cyclical transfer of potential and kinetic energy within the body. By increasing walking speed, most stroke subjects can improve mechanical recovery without penalties in internal work production. In particular, frontal plane internal work does not appear to increase as a result of speed-related compensation strategies, and is not a significant contributor to total mechanical work. Future studies will aim to identify the kinematic and spatiotemporal variables that are the best predictors of mechanical work and mechanical recovery.

2.4.3 Acknowledgements

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Chapter 3

REGRESSION ANALYSIS OF POST-STROKE MOVEMENT PATTERNS AND MECHANICAL RECOVERY

3.1 Introduction

Post-stroke changes in gait are both mechanically and metabolically costly. Many stroke survivors have reduced walking speeds and poor endurance. During normal walking, mechanical energy is preserved from stride to stride through the exchange of potential and kinetic energy. In healthy walkers, optimum recovery of mechanical energy (~65%) occurs around self-selected walking speed (Cavagna and Kaneko 1977; Mian, Thom et al. 2006). This mechanism deteriorates post-stroke and peak mechanical recovery can range from 25% and 70% (Olney, Monga et al. 1986; Detrembleur, Dierick et al. 2003). The cause of this reduction has been attributed to the vertical and forward movement of the whole body center of mass becoming more in-phase (Olney, Monga et al. 1986). Mechanical recovery may also be impacted by asymmetric movement patterns and compensatory strategies to clear the paretic leg during swing.

Post-stroke rehabilitation is aimed at increasing walking speed and eliminating these undesirable post-stroke movement patterns (Wall and Turnbull 1986). Beneficial outcomes have been observed as a result of increasing walking speed (Zamparo, Francescato et al. 1995; Detrembleur, Dierick et al. 2003; Lamontagne and Fung 2004; Chen, Patten et al. 2005; Reisman, Rudolph et al. 2009). A positive relationship was established between mechanical recovery and walking speed, with greater mechanical recovery occurring in subjects with faster self-selected walking speeds (Detrembleur, Dierick et al. 2003). Increasing walking speed has also been shown to improve swing-time symmetry, muscle activation patterns, and metabolic efficiency (Zamparo, Francescato et al. 1995; Lamontagne and Fung 2004; Chen, Patten et al. 2005; Reisman, Rudolph et al. 2009). Post-stroke changes in mechanical work production at the hip and ankle have also been detected. Reduced hip and ankle strength has been shown to contribute to poor swing initiation in the paretic leg (Nadeau, Gravel et al. 1999; Chen and Patten 2008). Poor swing initiation and reduced kinetic energy of the paretic leg at toe-off can lead to reduced knee flexion during swing and limited foot clearance (Anderson, Goldberg et al. 2004; Chen and Patten 2008).

Regression models have been designed to examine gait variables in healthy and post-stroke populations, but with limited utility (Stansfield, Hillman et al. 2006; Cruz, Lewek et al. 2009). Only toe-off posture and joint strength have been used to predict leg-clearance strategies in post-stroke gait, ignoring other clinically relevant variables. Other regression studies have tried to predict six-minute walk time and step length based on clinical variables, but ignore mechanical work and recovery (Judge, Davis III et al. 1996; Patterson, Forrester et al. 2007).

Understanding the interactions among walking speed, movement asymmetries, altered joint kinetics and kinematics, and mechanical work production is vital in designing post-stroke treatment protocols to improve walking function. If walking speed, economy, and endurance are paramount for successful recovery then it is important to understand how mechanical recovery is affected by all post-stroke movement patterns. Being able to predict mechanical recovery from gait variables can aid clinicians in determining which aberrant movement patterns may be leading to diminished mechanical recovery which is otherwise undetectable by inspection. Therefore, the objective of this study was to determine if any kinematic, kinetic, or temporal-spatial variables could be used to predict mechanical recovery post-stroke. Furthermore, we wanted to understand the interrelationships among many common post-stroke movement patterns.

3.2 Methods

Stroke survivors were recruited from the greater Philadelphia area. All subjects provided informed consent which was approved by the University of Delaware Human Subjects Review Board. All subjects were required to be between 30 and 80 years of age, and have suffered from their first stroke at least 6 months prior. To be included, subjects had to have incurred either a cortical or subcortical stroke. Subjects had to be ambulatory with some gait deficit and be able to walk for 5 minutes at their self-selected walking speed. Exclusion criteria included uncontrolled hypertension (> 190/110 mmHg), active cancer, peripheral artery disease with claudication, pulmonary or renal failure, unstable angina, sever aphasia, or dementia (Mini-Mental State exam score <22).

Preferred walking speed was determined using a stopwatch while subjects walked unassisted down a 6 meter walkway. Subjects were allowed to wear their prescribed ankle-foot orthoses. Reflective markers were placed on the bony landmarks of the foot, shank, thigh, pelvis, and trunk. Subjects walked on an instrumented split-belt treadmill with speeds tied (AMTI, Watertown, MA) at their preferred walking speed, 80% of their preferred speed, and two or three speeds faster than their preferred speed up to the maximum speed at which they could safely walk. Maximum walking speed was determined by increasing treadmill walking speed until subjects indicated that they could not walk any faster or did not feel comfortable walking any faster. Subjects were attached to an overhead harness (no body weight support), and were allowed to hold on to a handrail during all walking trials. Force plate data was collected at 1080 Hz. An 8-camera Vicon motion capture system (Vicon MX, Lake Forest, CA) was used to capture kinematic data at 120 Hz. An 8-segment model was created in Visual 3D (C-Motion, Rockville, MD) to include the foot, shank, thigh, pelvis, and Head-Arm-Trunk (HAT) segment.

External work (W_{ext}) in the vertical (W_v), anterior-posterior (W_{ap}), and medial-lateral (W_{ml}) directions were calculated by summing the positive increments of kinetic and potential energy in those directions over each 20-second walking trial. Mechanical recovery was calculated as follows:

$$R = \frac{W_v + W_{ap} + W_{ml} - W_{ext}}{W_v + W_{ap} + W_{ml}} * 100\% \qquad 8$$

Details of these calculations are described in chapter 2. Heel strike and toe off events were determined using a 20N threshold force in the vertical direction. Preswing was defined as the time period between heel strike of the contralateral limb and toe-off of the ipsilateral limb. Joint work of the ankles, knees, and hips during preswing were computed in Visual 3d. Swing time was calculated in Visual 3D as the time from toe-off to heel strike of the ipsilateral limb. Swing time asymmetry was calculated as follows:

$$\frac{\left| Swing_{paretic} - Swing_{healthy} \right|}{Swing_{paretic} + Swing_{healthy}} * 100\% \qquad 9$$

Circumduction was determined by measuring the maximum distance between the heel marker during stance phase and during the subsequent swing phase of the paretic leg during each gait cycle of the walking trial. The average circumduction during each walking trial was reported. Peak hip extension was determined by measuring the peak extension angle between the femur and the pelvis during terminal stance and pre-swing of each gait cycle and averaged across the walking trial. Peak knee flexion was determined by measuring the peak knee flexion of the paretic leg during swing and averaged across the walking trial.

Data were analyzed using SPSS v17 (SPSS inc, Chicago, IL). Spearman rank correlations were performed on all data in order to detect trends among all movement variables (a priori significance level of $p < .001$ after Bonferroni correction). Any variables that were significantly correlated with mechanical recovery were entered into a hierarchical linear regression in order to test their relative abilities to predict mechanical recovery. Finally, Wilcoxon signed-rank tests, with a priori significance of .05, were performed to compare the changes in each gait variable from minimum recovery to peak recovery as walking speed was increased. The same test was performed on all gait variables from peak recovery to minimum recovery as walking speed was increased.

3.3 Results

Ten stroke subjects (8 Male, 2 Female) were recruited for the study. The average age was 61.3 years (SD ± 12.6), and the average time since stroke was 25.8

months (SD ± 14.7). Average walking speed was 0.72 m/s (SD ± 0.14). Three subjects needed to use their AFOs in order to successfully complete the treadmill walking trials. All relevant clinical information of the post-stroke cohort is presented in chapter 2 (Table 2.1). Spearman rank correlation revealed a number of relationships among gait variables (Table 3.1).

Paretic leg kinetic energy was positively correlated with speed ($\rho = .530$, $p < .001$), peak hip extension ($\rho = .473$, $p < .001$), peak knee flexion ($\rho = .782$, $p < .001$), and paretic hip work ($\rho = .692$, $p < .001$). Paretic leg kinetic energy was negatively correlated with swing time asymmetry ($\rho =-.506$, $p < .001$). Paretic hip work was positively correlated with peak knee flexion ($\rho = .672$, $p < .001$) and negatively correlated with circumduction ($\rho = -.506$, $p < .001$). Paretic ankle work was positively correlated with hip extension ($\rho = .739$, $p < .001$) and negatively correlated with swing time asymmetry ($\rho = -.833$, $p < .001$) and circumduction ($\rho = -.473$, $p < .001$). Swing time asymmetry was also positively correlated with walking speed ($\rho = -.455$, $p < .001$) and negatively correlated with peak hip extension ($\rho = -.798$, $p < .001$).

Table 3.1 Spearman's Rho values are listed in the correlation matrix. All significant correlations ($p < .001$) are shown with double asterisks. Circumduction and paretic leg kinetic energy significantly correlated with mechanical recovery ($p < .05$), shown in bold. Paretic leg kinetic energy had the largest number of significant correlations; it correlated with five of the other eight gait variables. $p < .05$ was used as a threshold for including variables in a hierarchical regression to predict mechanical recovery.

Gait Variable Correlation Matrix

Variable	Speed	Mech Recovery	Swing Time Asym	Hip Ext	Knee Flexion	Circum	Paretic Ankle Work	Paretic Leg KE
Speed								
Mechanical Recovery	.105							
Swing Time Asymmetry	-.455**	.240						
Hip Extension	.392	-.076	-.798**					
Knee Flexion	.098	.174	-.371	.238				
Circumduction	.182	**.295**	.360	-.264	-.396			
Paretic Ankle Work	.296	.116	-.833**	.739**	.290	-.473**		
Paretic Hip Work	.192	-.074	-.307	.263	.672**	-.506**	.238	
Paretic Leg KE	.530**	**.306**	-.506**	.473**	.782**	-.064	.335	.692**

Although no significant relationships were discovered between mechanical recovery and any of the gait variables after Bonferroni correction (p <

.001), there were two relationships at the $p < .05$ level. Mechanical recovery correlated positively with paretic limb kinetic energy ($\rho = .306$, $p = .031$) and circumduction ($\rho = .295$, $p = .037$). Because of these correlations, the two gait variables were entered into a hierarchical regression in order to determine their ability to predict mechanical recovery. Prior to running the hierarchical regression a Spearman rank correlation was performed to test for collinearity of the two predictor variables. The test revealed no significant relationship between circumduction and paretic leg kinetic energy ($\rho = -.064$, $p = .656$). The two variables were entered into the regression model in order of their Spearman's rho values from largest to smallest (Table 3.2). The hierarchical regression revealed that paretic limb kinetic energy was able to significantly explain about 10% of the variance in mechanical recovery (r2 = .101, p = .024). The addition of circumduction did not significantly increase the accuracy of the model (significance of change in F, p = .074).

Table 3.2 Results of hierarchical regression to predict mechanical recovery. The table shows the two data models, the respective r2 values, the change in F value from the previous model, and the significance of the change in F from the previous model (i.e. p value). A significant F change of .05 was used as the cutoff for including variables in the hierarchical regression model.

Model	r^2	F change	Sig F change
Paretic limb kinetic energy	.101	5.417	.024
Paretic limb kinetic energy + Circumduction	.125	3.331	.074

Wilcoxin signed rank tests were performed in order to detect changes in each gait variable when mechanical recovery improved from minimum to maximum and again when recovery declined from maximum to minimum. The results are shown in Figure 3.1. Seven of the ten stroke subjects were able to increase their mechanical recovery when they increased their walking speed. Those subjects were able to significantly increase their walking speed ($Z=-2.366$, $p<.05$), peak knee flexion ($Z=-2.366$, $p<.05$), paretic hip work ($Z=-2.366$, $p<.05$), and paretic leg kinetic energy ($Z=-2.366$, $p<.05$), when their mechanical recovery increased from minimum to maximum ($Z=-2.366$, $p<.05$). Mean walking speed increased from .60 m/s to .84 m/s as mechanical recovery increased from its mean minimum value of 45.6% to its mean maximum value of 51.3%. Over that same range, peak knee flexion increased from 43.7 deg to 47.1 deg, paretic hip work increased from 0.045 J/kg to 0.063 J/kg, and paretic leg kinetic energy increased from 0.022 J/kg to 0.037 J/kg.

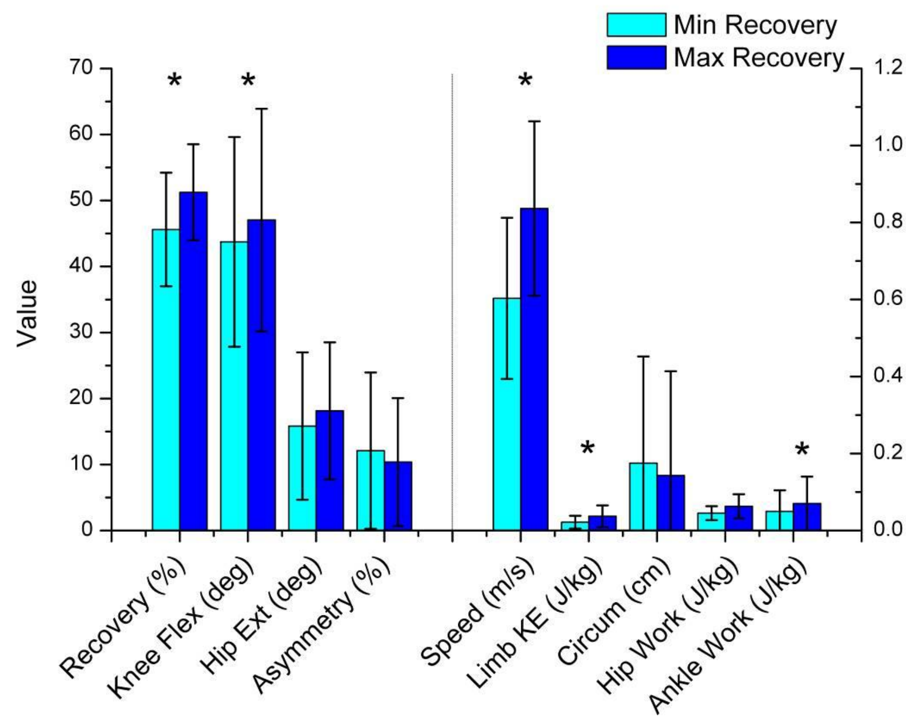

Figure 3.1 Results of the Wilcoxon signed ranks for each gait variable between minimum mechanical recovery at slow walking speeds and maximal mechanical recovery (N=10). Shown are average values (± 1 standard deviation) for each gait variable. Results are displayed in two groups (large values and small values) separated by the dashed vertical line for ease of display. Speed, paretic limb kinetic energy, paretic hip work, and peak knee flexion were significantly increased when mechanical recovery increased from minimum to maximum (*) (p<.05).

Seven of the ten stroke subjects were able to walk fast enough to exceed their optimal walking speed and experience reductions in mechanical recovery. Although these subjects were able to significantly increase their walking speed (Z=-2.366, p<.05) and decrease their mechanical recovery (Z=-2.366, p<.05), no other significant changes were detected during decreasing mechanical recovery. Mean walking speed increased from 0.76 m/s to 1.17 m/s as mechanical recovery decreased from its mean maximum value of 52.9% to its mean minimum value of 44.1% (Figure 3.2).

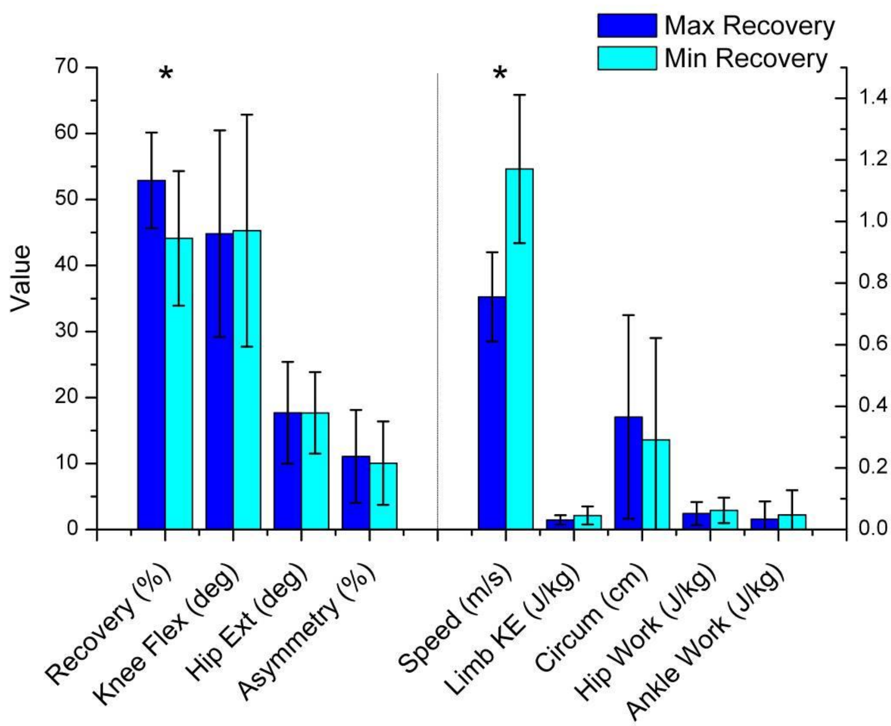

Figure 3.2 **Results of the Wilcoxon signed ranks for each gait variable between maximal mechanical recovery and minimum mechanical recovery at fastest walking speeds (N=10). Shown are average values (± 1 standard deviation) for each gait variable. Results are displayed in two groups (large values and small values) separated by the dashed vertical line for ease of display. Walking speed significantly increased when mechanical recovery decreased from maximum to minimum (*) (p<.05).**

3.4 Discussion

The main objective of this study was to examine whether speed-related changes in post-stroke gait are related to changes in mechanical recovery. Mechanical recovery can only be partially explained by paretic limb kinetic energy at toe off, since changes in mechanical recovery are accompanied by a range of kinematic and kinetic changes. This information could be used to determine which gait characteristics limit mechanical recovery post-stroke.

We found no significant relationships between mechanical recovery and any of the gait variables after Bonferroni correction ($p < .001$) but we did detect correlations with paretic limb kinetic energy and circumduction at the $p < .05$ level. Only paretic limb kinetic energy was a significant component of the hierarchical regression model, yet it was only able to explain about 10% of the variance in mechanical recovery.

Circumduction is generally considered an undesirable compensation strategy and most post-stroke interventions aim to eliminate it in order to regain symmetry of movement. The addition of circumduction to the regression model did not significantly increase its predictive ability. Despite the correlation, the results of this study suggest that circumduction may not adversely affect mechanical recovery. Although circumduction may be a mechanically effective strategy to achieve optimal mechanical recovery it may still be metabolically costly. Determining the extent to which increased metabolic expenditure undermines increased mechanical efficiency remains a complex issue (Burdett, Skrinar et al. 1983; Mian, Thom et al. 2006). Increased circumduction was also correlated with decreased paretic ankle work and decreased hip work suggesting that it is an adopted strategy to overcome hip and ankle weakness. In fact, paretic hip work positively correlated with paretic limb kinetic energy and peak knee flexion during swing (Nadeau, Gravel et al. 1999). Sufficient limb kinetic energy and knee flexion are important conditions for successful foot clearance during swing. Since paretic ankle work negatively correlated with circumduction but did not correlate with limb kinetic energy and peak knee flexion then it is possible that increased hip flexion work was used to compensate for plantar flexion weakness (Olney 1996; Jonkers, Delp et al. 2007; Chen and Patten 2008; Lewis and Ferris 2008). Hip work and power have been identified as limiting factors in the ability of healthy adults and stroke survivors to increase their walking speed (Judge, Davis III et al. 1996; Nadeau, Gravel et al. 1999; Hsu, Tang et al. 2003). In the current study, paretic hip work was not significantly correlated with walking speed. However, the results of the Wilcoxon signed rank test showed that individuals increasedparetic hip work and walking speed when mechanical recovery increased from minimum to maximum (Figure 3.1). This tells us is that even though hip work is not a reliable predictor of absolute walking speed or mechanical recovery it appears to be a mechanism for speed modulation post-stroke (Nadeau, Gravel et al. 1999; Hsu, Tang et al. 2003).

As long as stroke survivors are capable of reaching walking speeds of at least 0.54 m/s, our data suggest they have the potential to maximize their mechanical recovery regardless of how the paretic limb is cleared. Speeds in excess of 0.94 m/s did not contribute to increased mechanical recovery. This same decline in mechanical recovery at fast walking speeds (approximately 1-1.5 m/s) has also been reported in healthy adults (Cavagna, Thys et al. 1976; Mian, Thom et al. 2006). This suggests that increasing walking speed up to healthy self-selected speeds might be the most reliable predictor of increasing mechanical recovery in healthy adults and stroke survivors

(Cavagna, Thys et al. 1976; Detrembleur, Dierick et al. 2003; Mian, Thom et al. 2006). This is not surprising since the forward component of mechanical work is not sufficiently large enough to adequately exchange with the vertical mechanical work at slow walking speeds which are common post-stroke (Olney, Monga et al. 1986). Increasing walking speed has the added benefit of improved limb symmetry, joint excursions, and muscle activation (Lamontagne and Fung 2004; Chen, Patten et al. 2005).

Kinematic and temporal-spatial symmetry are important for improving the appearance of post-stroke movement patterns. We found no indication that swing time symmetry affects mechanical recovery. However, it did negatively correlate with walking speed, hip extension, paretic ankle work, and paretic leg kinetic energy. In fact, each of those gait variables significantly correlated with each other (except for paretic leg kinetic energy and paretic ankle work). We surmise that subjects who had greater hip extension were higher functioning, therefore they could produce a larger amount of paretic ankle work (Jonkers, Delp et al. 2007). While paretic hip work did not correlate with any of these variables it could still be used in conjunction with increased paretic ankle work to increase paretic leg kinetic energy, increase knee flexion during swing, and reduce the swing time of the paretic leg (Chen, Patten et al. 2005; Jonkers, Delp et al. 2007; Chen and Patten 2008).

Some limitations of this study must be noted. First, 3 of our 10 subjects used an AFO during their walking trials. Two of those subjects were able to increase their mechanical recovery by increasing their walking speed. The use of an AFO could create an inextricable interaction between the device and ankle, hip, and knee moments (Cruz and Dhaher 2009). The AFO could restrict active ankle plantar flexion during push off or passively increase plantar flexion due to elastic energy storage in the orthotic. Second, all stroke survivors used a handrail while walking on the treadmill. Handrail hold has been shown to improve swing time symmetry compared to unassisted walking (Chen, Patten et al. 2005). Third, creating a regression model with data from 10 subjects could yield an underpowered analysis. In a multiple regression analysis it is a common convention to include at least 50 data points per independent variable used in the model. There were only 50 data points included in this regression model to determine the predictive ability of two independent variables. Similarly, the Spearman rank correlations were performed on 10 different variables, yielding a significance level of $p < .001$. This could have lead to an increase in type II errors. Six other significant relationships were identified at the $p < .05$ level but were not presented in this study.

3.4.1 Conclusion

Being able to predict mechanical work and recovery is important because they can help reveal the sources of metabolic energy expenditure which dictate walking economy and endurance. The individual effects of stroke can vary making it difficult to discern generalized relationships. Although we were unable to find any predictors of mechanical recovery at the $p < .001$ level we discovered that increased circumduction did not inhibit mechanical recovery but actually helped to improve it. Regaining symmetry of movement may be of psychological importance to many stroke survivors, but its role in mechanical recovery may not be critical. Although walking speed cannot predict mechanical recovery, speed modulation up to healthy walking speeds is critical for maximizing mechanical recovery and may have beneficial implications for therapeutic rehabilitation.

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Chapter 4

CONCLUSION

4.1 Introduction

Post-stroke hemiparetic gait involves reduced walking speeds and compensatory movement patterns, which has implications for metabolic and mechanical efficiency. Mechanical recovery is reduced post-stroke, but stroke survivors with faster self-selected walking speeds tend to recover more mechanical energy, thus behaving more like their neurologically healthy peers (Detrembleur, Dierick et al. 2003). Walking speed is considered a useful outcome measure when evaluating the effectiveness of post-stroke recovery (Goldie, Matyas et al. 1996). While previous research provides valuable information about walking speed and walking function it does so at only one or two walking speeds. These studies have not addressed the ability of stroke survivors to alter their movement patterns and mechanical recovery by modulating walking speed. Modulating walking speed has been shown to improve symmetry, joint work of the paretic leg, and walking economy (Chen, Patten et al. 2005; Reisman, Rudolph et al. 2009). This study aimed to determine if speed modulation has similar beneficial outcomes on the recovery of mechanical energy. To do this, we investigated the effect of speed modulation on mechanical work and mechanical recovery during gait in healthy adults and stroke survivors. We also evaluated the relationships among post-stroke movement patterns in order to determine which gait parameters were responsible for changes in mechanical recovery.

AIM 1: Assess how mechanical work production and mechanical recovery change when healthy and stroke subjects modulate their walking speed.

Hypothesis 1.1: Healthy mechanical recovery will peak at self-selected walking speed. Stroke mechanical recovery will continually increase with increased walking speed.

Hypothesis 1.2: Stroke subjects will produce greater frontal plane internal work than healthy subjects.

As expected, mechanical recovery peaked around self-selected walking speed, and declined at speeds which deviated from self-selected for healthy adults. Stroke recovery did not exactly behave as hypothesized. Walking speed did not correlate with mechanical recovery, but mechanical recovery did tend to increase with increased walking speed up to about 1 m/s. At speeds faster than 1 m/s only one of the subjects in this study was able to increase their mechanical recovery. Even more importantly, stroke subjects with the slowest self-selected walking speeds showed the greatest improvements in mechanical recovery.

We hypothesized that frontal plane work would be greater in the stroke group due to the use of foot clearing strategies such as circumduction. We found that frontal plane work was not responsive to changes in walking speed in the stroke group. It appeared that healthy subjects and stroke subjects produced the same amount of frontal plane leg work at matched speeds. The healthy group produced more frontal plane work as speed increased while no change was observed in the stroke group. Regardless of these relationships, the proportion of frontal plane work to total leg work was negligible.

It must be noted that stroke subjects in this study required the use of the handrail in order to successfully and safely walk on the treadmill. This introduces another set of external forces that act on the whole-body center of mass. Handrail holding has been shown to improve movement patterns, but we have no knowledge of how these forces affect mechanical work and recovery (Chen, Patten et al. 2005). In trying to establish relationships among speed modulation and walking mechanics we found inconsistent responses to increasing walking speed. Ten subjects may not be enough to discern the effects of speed modulation in the post-stroke population.

AIM 2: Identify the relationships among stroke gait parameters and determine which of those parameters are the best predictors of mechanical recovery.

Hypothesis 2.1: Plantar flexor work of the paretic ankle during pre-swing and reduced circumduction will best predict mechanical recovery.

Hypothesis 2.2: Increased walking speed will correlate positively with symmetry, paretic limb work, and paretic leg kinetic energy.

Hypothesis 2.3: Paretic ankle work and paretic hip work will correlate positively with paretic leg kinetic energy and peak knee flexion during swing.

We discovered that only paretic leg kinetic energy can partially predict mechanical recovery. This did not agree with our original hypothesis. Paretic limb kinetic energy was significantly correlated with mechanical recovery. Interestingly, we found that increased circumduction did not lead to reduced mechanical recovery; in fact it aided limb clearance and facilitated increasing mechanical recovery. The results of the Wilcoxon signed rank test also revealed that circumduction did not decrease when mechanical recovery increased. This means that eliminating circumduction is not necessary for increasing mechanical recovery regardless of how much stroke survivors may rely on it for ambulation.

Similar to previous work, we found that walking speed correlated negatively with swing time asymmetry, and positively with paretic leg kinetic energy (Chen, Patten et al. 2005). Speed did not significantly correlate with either paretic hip work or paretic ankle work. However, despite the lack of correlation for the group individual stroke subjects increased their paretic hip work and walking speed when increasing their mechanical recovery. We concluded that, as a group, hip work does not predict absolute walking speed, but it is a tool used to modulate walking speed.

Only hip work significantly correlated with paretic leg kinetic energy and peak knee flexion during swing. This did not completely agree with our original hypothesis that ankle work would also be important for increasing paretic leg kinetic energy and driving the leg into swing. Paretic leg kinetic energy was also positively correlated with peak knee flexion during swing, but we could not conclude if this was attributed to either increased paretic hip work, paretic ankle work, or a combination of the two.

This study was subject to the same limitations as the previous study in that handrail hold and AFO use can alter coordination patterns, kinetics, and kinematics and should be controlled in future studies (Chen, Patten et al. 2005; Cruz and Dhaher 2009). Ten subjects may not be enough to produce an adequately powered regression analysis. Finally, even though circumduction had no negative consequences on mechanical recovery it could still be a metabolically inefficient strategy. Overall, circumduction appears to be a necessary movement strategy to clear the paretic leg when adequate propulsion cannot be obtained by the hip flexors or ankle plantar flexors. Future work should evaluate the relationships between mechanical recovery and metabolic economy post-stroke.

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