INDIVIDUAL LIMB MECHANICAL ANALYSIS OF GAIT FOLLOWING STROKE

Caitlin E. Mahon

A thesis submitted to the faculty of the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Master of Science in the Department of Biomedical Engineering

Chapel Hill
2013

Approved by:

Michael D. Lewek, PT, PhD

Gregory S. Sawicki, PhD

Paul S. Weinhold, PhD

ABSTRACT

CAITLIN E. MAHON: Individual Limb Mechanical Analysis of Gait Following Stroke
(Under the direction of: Michael D. Lewek, PT, PhD)

Following stroke, hemiparesis can lead to gait impairment, characterized by limb mechanical asymmetry and metabolic inefficiency. Due to the importance of ambulation at home and in the community, post-stroke rehabilitation often focuses on recovery of gait, including increasing self-selected walking speeds. An in depth comparison between individual limb mechanics and their role in gait inefficiencies would allow for improvements to rehabilitation programs by guiding more specific therapies. Recently, the step-to-step transition of a stride has been studied more closely in unimpaired individuals through the pendulum model of walking, and has been shown to be a period of high mechanical and metabolic power output. The purpose of this study was to perform an individual limb analysis of post-stroke hemiparetic walking during different phases of a stride, including the step-to-step transition, in order to assess limb mechanical asymmetries and how they relate to severity of gait impairment.

ACKNOWLEDGEMENTS

For the individual limb method calculations performed as part of this analysis, I would like to acknowledge the use of MATLAB code written by Dominic Farris, PhD.

Additionally, I would like to acknowledge the Rehabilitation Engineering Center at the University of North Carolina at Chapel Hill and North Carolina State University for providing a seed grant, which was applied to the work conducted for this analysis.

TABLE OF CONTENTS

LIST OF TABLES

Table

LIST OF FIGURES

Figure

LIST OF ABBREVIATIONS AND SYMBOLS

α	Alpha level
AFO	Ankle-foot orthosis
COM	Center of mass
DSL	Leading double-support phase
DST	Trailing double-support phase
GRF	Ground reaction force
Hz	Hertz
ILM	Individual Limbs Method
m	Meter
m/s	Meter per second
$+\eta_{work}$	Metabolic Efficiency of positive work
$+P_{avg}$	Mean positive average mechanical power
$-P_{avg}$	Mean negative average mechanical power
P_{avg}	Mean average mechanical power
$+P_{cavg}$	Combined mean positive average mechanical power
P_{inst}	Instantaneous mechanical power
P_{met}	Net metabolic power
p-value	Probability value
SS	Single-support phase
TS	Treadmill Speed
W/kg	Watt per kilogram
%	Percent
°	Degree
*	Statistical significance

CHAPTER 1

INTRODUCTION

Hemiparesis is a common result of stroke [1], often leading to gait inefficiencies due to weakness in the paretic lower-limb [2]. Inactivity may result from gait impairment, and in turn, cause further negative consequences to overall health [3]. Increasing self-selected walking speed, while maintaining metabolic economy, is often the primary goal of post-stroke rehabilitation programs, due to the role of ambulation in a healthy life-style and within the community. It is important that we improve upon rehabilitation methods used to treat gait impairment, since stroke is prevalent in our society and the leading cause of long-term disability [4]. Analyzing the fundamentals of post-stroke gait will provide more successful rehabilitation targets, to ultimately decrease disability and improve quality of life.

Previously, post-stroke gait has been described mechanically as asymmetric and metabolically as inefficient. Mechanical asymmetry is most often a result of unilateral (i.e., paretic) limb weakness and compensatory movements provided by the contralateral (i.e., non-paretic) limb [5]. Metabolically, post-stroke gait has been shown to exhibit approximately twice the metabolic cost of unimpaired gait [6,7]. Metabolic inefficiency has been attributed to the increase in mechanical work performed during post-stroke gait [6,7]. Therefore, one approach to lowering metabolic cost, and potentially enabling individuals to walk greater distances and with greater self-selected speed, may be to lower mechanical work requirements.

In order to develop rehabilitation programs focused on lowering mechanical work requirements, a better understanding of these requirements and the influence of asymmetry at an individual limb level must be gained. Recent studies indicate that separate phases of a stride contribute uniquely to total mechanical and metabolic work during gait [8]. This has been illustrated by the pendulum model, which describes the motion of an individual's center of mass (COM) velocity during the single-support of one limb as an inverted pendulum [8]. During step-to-step transitions between pendulum arcs of each limb, metabolic requirements account for sixty percent (%) to seventy % of the total metabolic cost of a stride [9]. Deviation from normal step-to-step coordination may increase mechanical work requirements, and therefore possibly metabolic requirements [10]. Research examining the individual limbs of post-stroke gait during each phase of a stride, particularly during step-to-step transitions, is important in determining these asymmetries and may provide insight into gait inefficiencies.

Acquiring knowledge to improve rehabilitation strategies is imperative to increasing the potential for improved mechanical and metabolic walking efficiency of individuals post-stroke, which may lead to greater self-selected walking speed. Self-selected walking speed is used as a determinant of rehabilitation success, and due to the influence it has on the daily lives of individuals at home and within the community, is also related to functional classification [4]. This emphasizes the importance research to improve rehabilitation, in order to increase social advantages and general well-being.

CHAPTER 2

INDIVIDUAL LIMB MECHANICAL ANALYSIS

The objective of this analysis was to compare mechanical asymmetry between individual limbs across three functional levels during post-stroke hemiparetic treadmill walking at self-selected speed. Using the individual limb method (ILM) [8], mechanical power produced on the COM was calculated during the trailing double-support (DST), leading double-support (DSL), and single-support (SS) phases of a stride, as well as over a complete stride. Across all functional levels, the non-paretic limb produced significantly more positive net mechanical power than the paretic limb during all three phases and over a complete stride, indicating mechanical asymmetries. A variation in average net mechanical power production during SS between functional levels existed, however a variation in mechanical asymmetry between functional levels did not exist during any phase. These results suggest that mechanical limb asymmetries are consistent between functional levels. Therefore, with lower functional ability, greater limb mechanical asymmetry is not the cause of reduced self-selected walking speeds.

INTRODUCTION

In unimpaired gait, the motion of the COM velocity can be described as an inverted pendulum during the SS phase of a stride [8]. Due to this pattern of motion, minimal mechanical work is required during SS when compared to the total mechanical work requirements of a stride [9]. During step-to-step transitions, mechanical work is required to redirect the COM velocity between the pendulum arcs of each limb [8,10]. Redirection comes from the combination of: (1) positive work produced during the trailing limb's DST phase and (2) negative work produced during the leading limb's DSL phase [8,10]. Minimizing total mechanical work is desirable to minimize metabolic cost [9,11], and can occur when the timing and magnitude of the leading limb's negative work is equal to the trailing limb's positive work [9,10]. However, even when this occurs, the step-to-step transition requires a significant amount of metabolic energy [9,11]. Typical mechanical work production during the DSL, SS and DST phases of unimpaired gait are shown in Figure 1. During the SS phase, both positive and negative work is produced. During the DSL and DST phases, net negative and net positive work is produced, respectively.

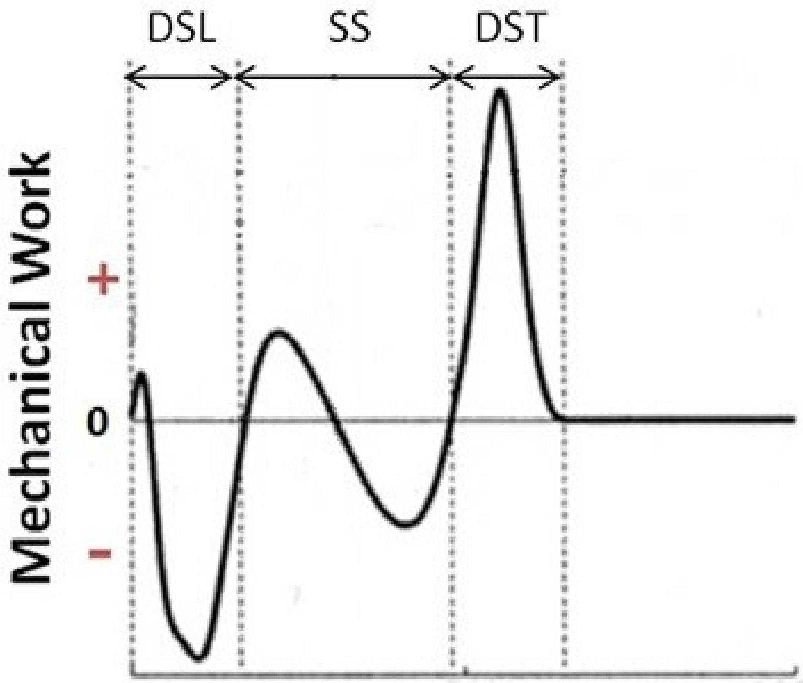

Figure 1. Typical unimpaired gait mechanical work production over a stride

Divergence from metabolic optimization has been shown to arise from inter-limb mechanical work asymmetries during step-to-step transitions. Specifically, the affected limb of individuals following transtibial amputation and total ankle arthroplasty exhibited less positive work production during DST and the unaffected limb exhibited greater negative work production during DSL [12,13]. These asymmetries are consistent with those exhibited by unimpaired individuals when mechanical restrictions are placed on the trailing limb [10]. It was concluded that greater negative work production from the leading limb was produced to redirect the COM [10,12,13] while greater positive work production during SS provided compensation for the resulting change in net work during the step-to-step transitions [12,13].

In individuals following stroke, similar impairments in muscle function have been identified, however inter-limb mechanical asymmetries for the separate phases of DST, DSL and SS have yet to be examined. Studies conducted at the joint level indicate that post-stroke gait produces less positive work during DST due to paretic plantar-flexor weakness [14,15]. An individual limb analysis examining the SS and DST phases, in combination, determined that greater positive mechanical work was produced by the non-paretic limb in order to raise the COM, and greater mechanical work production was positively correlated with metabolic cost [6]. However, this analysis did not examine individual limb contributions during the separate phases of step-to-step transitions, from which asymmetry appears to be an important factor in gait efficiency. We propose that increased mechanical requirements following stroke are, in part, the result of mechanical asymmetries between limbs during DST, DSL and SS, due to deficits and compensations exhibited uniquely for each phase.

The purpose of this study was to quantify mechanical asymmetry from an individual limb perspective in individuals with post-stroke hemiparesis, in order to aid future insight into the increased metabolic cost of post-stroke gait. We chose to examine a range of walking ability, due to the positive correlation between paretic ankle plantar-flexor weakness, which may affect mechanical asymmetries throughout a stride, and hemiparetic severity [2]. We hypothesize that: (1) mechanical asymmetries between limbs will increase in individuals with reduced gait ability. Based on previous analyses examining individual limb mechanics in similar patient populations [12,13], we additionally hypothesize that: (2) individuals post-stroke will exhibit less positive power production from the paretic limb during DST, greater negative power production from the non-paretic limb during DSL, and greater positive power production from the non-paretic limb during SS (each compared to the contralateral limb).

METHODOLOGY

We recruited individuals who presented with chronic (greater than six months post-stroke) hemiparesis following unilateral, non-cerebellar brain lesion due to stroke. We intentionally sought individuals with a range of walking ability, however all individuals had to be capable of walking at least ten meters (m) overground and two minutes on a treadmill without therapist assistance. Exclusion criteria consisted of Botox injection to the lower extremities in the three months preceding testing, or any musculoskeletal, cardiorespiratory/metabolic, or additional neurological disorder (e.g., Parkinson's disease) that could affect gait. Individuals were stratified into "functional" level groups based on self-selected overground speed: high gait function (>0.8 meters per second (m/s)), moderate gait function (0.5 m/s-0.8 m/s), and low gait function (<0.5 m/s). The range of speed defining each group was based on previous classification [4] with consideration of clinical evaluation (e.g. lower extremity Fugl-Meyer testing). Self-selected overground gait speed was determined from three passes across a 4.27 m-GAITRite mat (CIR Systems, Sparta, New Jersey) [16]. Individuals used assistive devices and bracing below the knee (e.g., ankle-foot orthosis (AFO)) if required for ankle stability. Prior to participation all individuals signed a University of North Carolina at Chapel Hill Institutional Review Board-approved informed consent form.

All experimental trials took place on a dual-belt treadmill (Bertec Corp, Columbus, Ohio), which was instrumented with two six-component force platforms that measured ground reaction force (GRF) data to be sampled at 1080 hertz (Hz) by a Vicon MX system (Vicon/Peak, Los Angeles, California). We chose an individualized treadmill speed that we believed could be maintained for a typical gait training session for each individual. If bracing below the knee was used for overground walking, it was retained for treadmill walking. If needed, individuals held one or both treadmill handrails, each instrumented with a load cell (MLP-150; Transducer Techniques, Temecula, California) capable of recording vertical force. All individuals wore a safety harness (Protecta PRO, Capital Safety, Red Wing, Minnesota) while walking which did not restrict lower extremity movements or provide unweighting during testing. Individuals walked on the treadmill for at least two minutes, with the second minute used for analysis. Steps were removed from a trial if an individual's feet did not fall on separate force platforms or if an individual experienced a stumble. For five subjects we were unable to obtain a minimum of ten consecutive steps from the second minute of walking and instead analyzed a later minute.

Using Visual3D software (C-Motion, Germantown, Maryland), GRF data were first filtered with a twenty-five Hz low pass filter. The ILM described by Donelan et al. [8], was used to calculate external mechanical work and power performed on the COM by the paretic and non-paretic limbs through custom written MATLAB (MathWorks, Natick, Massachusetts) programs. Vertical force data from handrail support, when produced, were included in net force data prior to calculation of COM acceleration. Since the ILM assumes symmetric gait, and spatiotemporal asymmetries are often exhibited following stroke [17,18], the calculations in Donelan et al. [8] were adjusted by: (1) assuming symmetry over strides, instead of steps, and (2) subtracting average COM acceleration over a trial from instantaneous COM acceleration prior to integration. COM acceleration integration constants for the X, Y and Z directions were calculated by minimizing the average COM velocity over a stride. Additionally, in the Y direction, an integration constant equaling treadmill speed was added. Instantaneous mechanical power generated by each limb was calculated as the dot product of that limb's GRF vector and the COM velocity, as per the ILM. For each stride, instantaneous mechanical power was then normalized to 101 points and averaged for each subject to produce mean instantaneous mechanical power (P_{inst}).

To obtain positive and negative average mechanical work done on the COM, instantaneous mechanical power generated by each limb was cumulatively integrated over the following phases: DST (from heel-strike of the contralateral limb to toe-off of the reference limb), DSL (from heel-strike of the reference limb to toe-off of the contralateral limb), and SS (from toe-off of the contralateral limb until heel-strike of the contralateral limb), and over a complete stride, restricting integration to positive or negative areas of the integrand, respectively. The positive and negative average mechanical work values for each limb were then multiplied by phase frequency over a trial (for the measures of average mechanical work produced over DST, DSL and SS) or stride frequency over a trial (for the measures of average mechanical work produced over a stride) to yield average positive mechanical powers ($+P_{avg}$) and average negative mechanical powers ($-P_{avg}$). $+P_{avg}$ and $-P_{avg}$ were summed to obtain total average mechanical power (P_{avg}) for each phase and over a stride. P_{inst}, $+P_{avg}$, $-P_{avg}$ and P_{avg} were normalized to body mass. The main outcome variables were: paretic and non-paretic limb peak P_{inst} during DSL and DST, P_{avg} during DSL, DST, SS and over a stride, $+P_{avg}$ over a stride and $-P_{avg}$ over a stride.

Statistical analyses were performed with SPSS (ver 21, Chicago, Illinois). For all subjects a paired sample t-test ($\alpha=0.05$) was used to evaluate differences between self-selected overground gait speed and the treadmill speed used for testing. For the low, moderate, and high groups, descriptive statistics (i.e., mean and standard deviation) were calculated for each variable. Eight separate two-way (limb x functional group) within-subject ANCOVAs ($\alpha=0.05$) were performed to examine differences in peak P_{inst} during DSL and DST, P_{avg} during DSL, DST, SS and over a stride, $+P_{avg}$ over a stride and $-P_{avg}$ over a stride. Given the known effect of gait speed on limb mechanical power output [8] we expect differences between functional groups, therefore treadmill speed was assigned as a covariate.

RESULTS

Twenty-six individuals with chronic stroke were recruited for this study: thirteen high gait function, six moderate gait function, and seven low gait function. A description of the individuals representing each functional group is listed in Table 1. The mean treadmill speed of all individuals (.70 ± .28 m/s) was slower than the mean self-selected overground gait speed (.78 ± .32 m/s) (p=.004).

Mean P_{inst} produced over a complete stride is shown in Figure 2 a-c for each limb and functional group. The two-way ANCOVAs analyzing peak P_{inst} during DST and DSL showed the paretic limb (compared to the non-paretic limb) produced a significantly less positive P_{inst} peak during DST ($p<.0005$), and no asymmetry between the non-paretic limb and paretic limbs for peak P_{inst} during DSL (p=.053); no difference between functional group during DST (p=.213) or DSL (p=.378); and no interaction effect between limb and functional group during DST (p=.136) or DSL (p=.978).

P_{avg} produced during the DST, DSL and SS phases for each limb and functional group are shown in Figure 2 d-f. The two-way ANCOVA's analyzing P_{avg} produced during DST, DSL and SS, showed the paretic limb (compared to the non-paretic limb) produced significantly less positive P_{avg} during DST ($p<.0005$), and the non-paretic limb (compared to the paretic limb) produced significantly less negative P_{avg} during DSL ($p<.0005$) and significantly greater positive P_{avg} during SS ($p<.0005$); a difference between functional group during SS ($p=.049$), but none during DST ($p=.133$) or DSL ($p=.472$); and no interaction effect between limb and functional group during DST ($p=.433$), DSL ($p=.460$) or SS ($p=.502$).

$+P_{avg}$ and $-P_{avg}$ produced during the DST, DSL and SS phases for each limb and functional group are shown in Figure 2 g-i.

$+P_{avg}$, $-P_{avg}$ and P_{avg} produced over a stride for each limb and functional group are shown in Figure 3 a-c. The two-way ANCOVA's analyzing $+P_{avg}$, $-P_{avg}$ and P_{avg} produced over a stride showed the paretic limb (compared to the non-paretic limb) produced significantly less positive $+P_{avg}$ ($p<.0005$), significantly more negative $-P_{avg}$ ($p<.0005$) and significantly less positive P_{avg} ($p<.0005$); no difference between functional group ($p=.747$, $p=.186$, $p=.278$, respectively); and no interaction effect between limb and functional group ($p=.250$, $p=.938$, $p=.495$, respectively).

Peak P_{inst} values for DST and DSL, P_{avg} values for DST, DSL, SS and over a stride, $+P_{avg}$ over a stride, and $-P_{avg}$ over a stride are listed in Table 2 for each functional group, along with corresponding ANCOVA p-values.

Table 1. Functional group description

	High (n=13)	Moderate (n=6)	Low (n=7)
Self-Selected Overground Speed (m/s)			
Range (minimum/maximum)	.83/1.3	.52/.78	.19/.49
Mean	1.0 ± .16	.69 ± .10	.37 ± .13
Treadmill Speed (m/s)			
Range (minimum/maximum)	.49/1.3	.60/.70	.15/.60
Mean	.90 ± .20	.65 ± .045	.37 ± .15
Gender (Male/Female)	7/6	4/2	3/4
Age (years)	56 ± 8.4	51 ± 12	56 ± 13
Time Post Stroke (months)	103 ± 92	26 ± 17	33 ± 17
Height (centimeter)	175 ± 8.4	176 ± 11	170 ± 7.5
Weight (kilogram)	91 ± 18	87 ± 14	99 ± 11
Lower Extremity Fugl-Meyer	28 ± 2.1	26 ± 3.1	20 ± 2.4
Paretic Limb (Right/Left)	7/6	4/2	3/4
Step Length (centimeter)			
Non-paretic	59 ± 5.9	47 ± 4.0	26 ± 9.9
Paretic	60 ± 9.3	49 ± 6.0	39 ± 9.0
Step Length Ratio	1.1 ± .11	1.1 ± .081	1.7 ± .71
Swing Time (second)			
Non-paretic	.39 ± .030	.36 ± .044	.32± .064
Paretic	.43 ± .049	.59 ± .13	.57 ± .14
Stance Time (second)			
Non-paretic	.78 ± .090	1.0 ± .16	1.5 ± .45
Paretic	.73 ± .080	.80 ± .045	1.3 ± .42

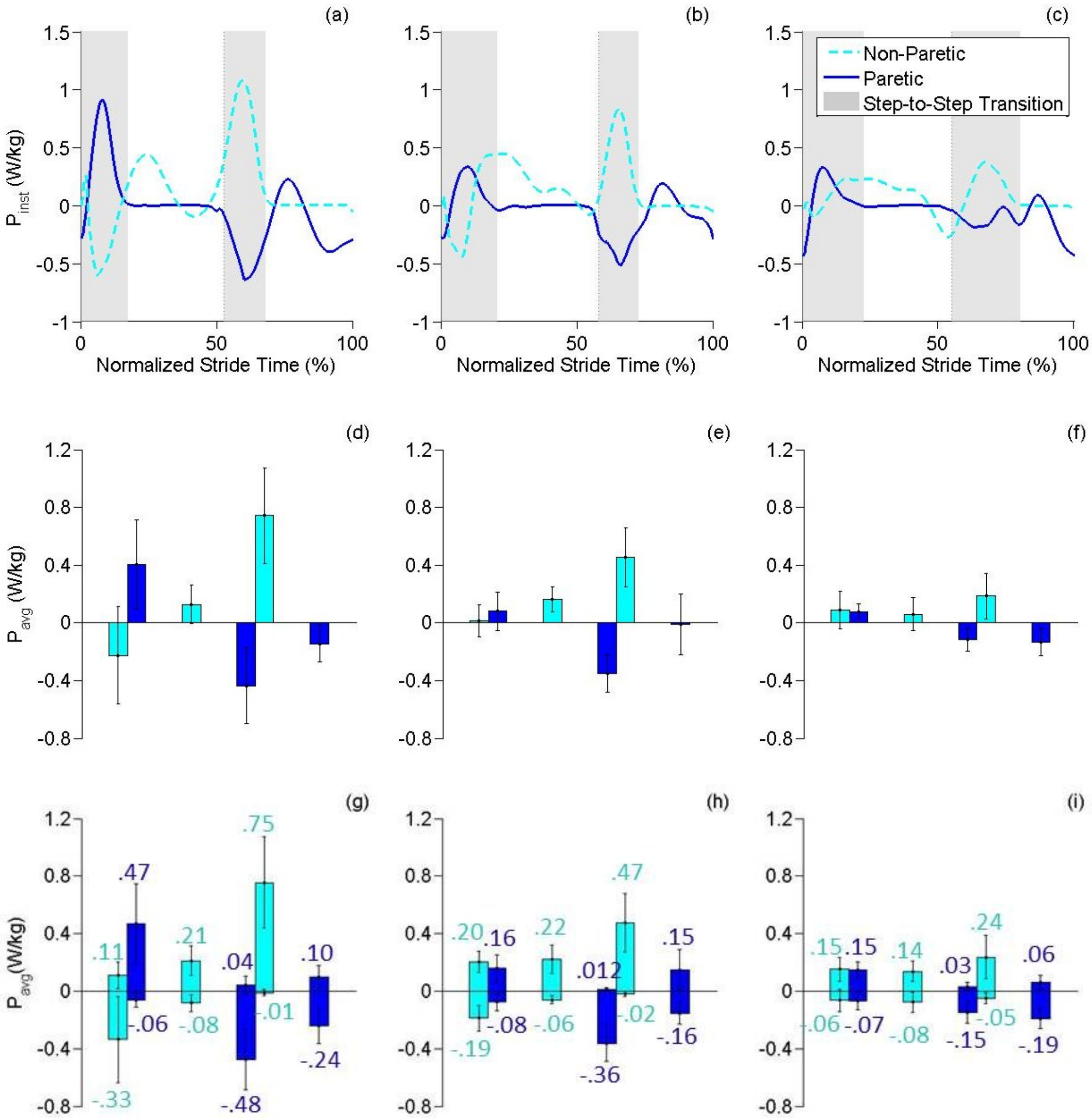

Figure 2. Mean power over each stride phase for each functional group. P_{inst} (a-c), P_{avg} (d-f) and $+P_{avg}$ and $-P_{avg}$ (g-i) results are plotted for the non-paretic and paretic limb and high (a,d,g), moderate (b,e,h) and low (c,f,i) functional groups. Error bars represent one standard deviation. W/kg = Watts per kilogram.

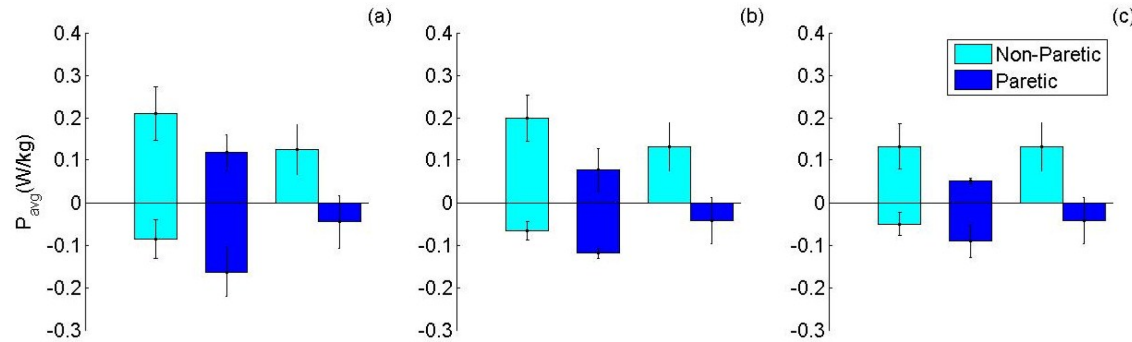

Figure 3. Mean average power over a stride for each functional group. $+P_{avg}$, $-P_{avg}$ and P_{avg} results are plotted for the non-paretic and paretic limb and high (a), moderate (b) and low (c) functional groups. Error bars represent one standard deviation.

Table 2. Mean power values for each functional group

	Power (W/kg)					
	High		Moderate		Low	
	Non-Paretic	Paretic	Non-Paretic	Paretic	Non-Paretic	Paretic
P_{inst} DST	1.4±.52	1.0±.50	.99±.26	.36±.21	.46±.27	.36±16
P_{inst} DSL	-.70±.46	-.81±.31	-.55±.19	-.72±.19	-.17±.22	-.30±.20
P_{avg} DST	.74±.33	.40±.31	.46±.20	.080±.13	.18±.16	.079±.050
P_{avg} DSL	-.22±.34	-.44±.26	.015±.11	-.35±.13	.088±.13	-.11±.078
P_{avg} SS	.13±.13	-.14±.13	.16±.085	-.0091±.21	.060±.12	-.11±.15
P_{avg} Stride	.13±.059	-.046±.062	.13±.056	-.042±.055	.082±.056	-.040±.035
$+P_{avg}$ Stride	.21±.063	.12±.042	.20±.054	.077±.0058	.13±.053	.051±.0058
$-P_{avg}$ Stride	-.086±.046	-.164±.058	-.067±.021	-.119±.012	-.050±.028	-.091±.038

Table 3. ANCOVA p-values

	p-Value		
	Interaction Effect	Main Effect	
		Limb	Function
P_{inst} DST	.136	<.0005*	.213
P_{inst} DSL	.978	.053	.378
P_{avg} DST	.433	<.0005*	.133
P_{avg} DSL	.460	<.0005*	.472
P_{avg} SS	.502	<.0005*	.049*
P_{avg} Stride	.250	<.0005*	.747
+P_{avg} Stride	.938	<.0005*	.186
-P_{avg} Stride	.495	<.0005*	.278

*Statistical significance

DISCUSSION

These results indicate that mechanical asymmetry between the non-paretic and paretic limbs in post-stroke walking exists during all phases of a stride and over a complete stride, however contrary to our first hypothesis, asymmetry is consistent across all functional levels. Our second hypothesis was partially confirmed. We observed less positive peak P_{inst} and P_{avg} of the paretic limb (compared to the non-paretic limb) during DST, but we did not observe greater negative peak P_{inst} and P_{avg} of the non-paretic limb (compared to the paretic limb) during DSL; instead, there was greater negative P_{avg} of the paretic limb (compared to the non-paretic limb). Finally, there was greater positive P_{avg} of the non-paretic limb (compared to the paretic limb) during SS.

Although we observed significant asymmetries between limbs for all functional groups during each phase of the stride and over a complete stride we were surprised that these asymmetries did not exhibit greater differences *between* groups. These results suggest that with lower functional ability, less generation and absorption of mechanical power is exhibited by the paretic *and* non-paretic limbs during gait. Therefore, during DST, DSL and SS, non-paretic compensation does not increase in response to greater paretic weakness, and greater limb mechanical asymmetry is not the cause of reduced self-selected walking speeds. In individuals with lower functional ability, greater gait inefficiencies may result from poor intra-limb coordination through increasingly poor redistribution of muscle or joint powers. Poor coordination of muscle or joint contributions may lead to both greater positive and negative work production, undetectable at a limb level analysis due to cancelation.

Through a joint level analysis, Olney et al. [5] found that inter-limb asymmetry of positive mechanical work production over a complete stride did not relate to gait function, with the non-paretic lower-limb joints producing approximately sixty % of the total positive mechanical work across different functional levels. Although this study did not involve analysis of separate phases of a stride, it too indicates that mechanical asymmetry may not vary with functional level during post-stroke gait.

We did observe a difference in production of P_{avg} during SS between functional groups. For all functional groups, net positive P_{avg} was produced by the non-paretic limb and net negative P_{avg} was produced by the paretic limb (Table 2). The moderate group produced more net positive non-paretic P_{avg} and less net negative paretic P_{avg} than both the high and low groups. This suggests that moderate functioning individuals may compensate with greater positive power production from both limbs during SS. This may be due to a combination of less positive power generation (or more negative power absorption) during the DST and DSL phases from both limbs.

Our second hypothesis was that post-stroke walking would involve less positive mechanical power production during DST from the paretic limb, greater negative mechanical power production during DSL from the non-paretic limb, and greater positive mechanical power production during SS from the non-paretic limb (when each was compared to the opposite limb). We observed less positive peak P_{inst} and P_{avg} of the paretic limb (compared to the non-paretic limb) during DST. Although this may be due to the paretic ankle plantar-flexors producing less propulsive mechanical power than the non-paretic ankle plantar-flexors, examination of mechanical power production at the joint level is needed to confirm this.

We observed less negative peak P_{avg} of the non-paretic limb (compared to the paretic limb) during DSL, contrary to our hypothesis. We believe this is partly due to step length asymmetry between limbs, and the positive correlation between step length and negative work production during heel-strike of the same limb [11]. A greater mean paretic (compared to non-paretic) step length was exhibited across all functional groups. This may be a result of greater positive P_{avg} produced by the non-paretic limb (compared to the paretic limb) during SS, as expected in our second hypothesis, leading to greater forward propulsion during paretic (compared to non-paretic) swing [11] and therefore requiring greater power for COM redirection from the paretic limb (compared to non-paretic limb) during DSL. Kuo et al. [9], state that at any speed, the separate trailing and leading contributions of work performed on the COM nearly cancel one another out. Less positive contribution from the trailing paretic limb would be concomitant with less negative contribution from the leading non-paretic limb, which was observed.

Initiation of positive P_{inst} production by the non-paretic limb during late DSL was exhibited by all functional groups (Figure 1), which in unimpaired individuals does not begin until SS [8]. This resulted in net positive P_{avg} during DSL for the moderate and low groups. This may indicate early non-paretic limb initiation to raise the COM or compensation for less propulsive power produced by the trailing paretic limb; negative P_{inst} production by the paretic limb during late SS was exhibited by all functional groups (Figure 1), which also deviates from unimpaired individuals who produce positive P_{inst} directly prior to DSL [8]. This leads us to believe that mechanical asymmetries may be a result of weakness in the paretic limb during SS *and* DST. Across all functional groups, our results indicate that post-stroke gait exhibits: (1) less positive power production by the paretic limb (compared to the non-paretic limb) during SS and DST due to weakness and (2) greater negative power production by the paretic limb (compared to the non-paretic limb) during DSL in order to redirect the COM following non-paretic compensation, which may take place during DST, DSL and SS.

The differences we see in mechanical asymmetries of post-stroke gait compared to gait of similar patient populations [12,13] may be due to greater weakness of the affected (i.e., paretic) limb of the individuals we analyzed, a conclusion based on mean self-selected walking speeds from all studies. Greater weakness of the affected limb may lead to greater and more widespread deficits and compensations throughout the whole stride, relating to mechanical coordination.

It is important to recognize the apparent limitations of our analysis. The ILM included calculation of external mechanical power only and did not include internal mechanical power. Possible sources of internal mechanical power that could be more pronounced with lower functional ability include hip hiking, stiff-knee gait and balance control. Therefore, factors resulting in greater mechanical asymmetry, may be present, but not reflected in external mechanical power calculations.

Additionally, handrail support was measured in the vertical direction only and did not include force produced in the fore-aft and medial-lateral directions. However, we do not believe handrail support in either of these directions were large enough to have a significant impact on our final results and conclusions. However, vertical handrail support used during treadmill walking may affect asymmetry within a stride. Handrail support was predominantly used on the non-paretic side by the moderate and low groups and was greatest during paretic SS. However, handrail support was used differently and sometimes periodically by individuals across all groups, and it would be difficult to make generalized conclusions regarding its effect on our results. Our main aim for this analysis was to examine inter-limb contributions during post-stroke hemiparetic walking, which is often best represented by including upper-limb support as replication for assistive walking devices (e.g., cane).

CHAPTER 3

CONCLUSION

This study examined individual limb mechanics of post-stroke hemiparetic walking during different phases of a stride, in order to assess mechanical limb asymmetries and how they relate to severity of walking impairment. Long-term disability from stroke is a significant problem in our society today, and a mechanical analysis will allow insight into the cause of gait inefficiencies.

We found that robust differences in mechanical power produced between limbs exist in post-stroke gait. Across all functional groups, the paretic limb produced less positive peak P_{inst} and P_{avg} during DST and the non-paretic limb produced less negative P_{avg} during DSL and greater positive P_{avg} during SS (each compared to the contralateral limb). A variation in P_{avg} production during SS between functional groups existed, but we did not observe variation in mechanical asymmetry between functional groups during any phase.

Understanding individual limb mechanics and how they relate to gait impairments will aid development of improved rehabilitation programs. With improved rehabilitation, selection of increased walking speeds at home and in the community will be possible, leading to higher functional ability of individuals post-stroke.

APPENDIX

Mechanics and Metabolics of Gait Following Stroke

The purpose of this data collection was to examine individual limb mechanical powers, metabolic power and positive work efficiency over a complete stride for post-stroke gait at varying speeds and surface gradients.

Two males and one female who presented with chronic (greater than six months post-stroke) hemiparesis following unilateral, non-cerebellar brain lesion due to stroke were recruited for this study. All individuals had to be capable of walking at least ten m overground and four minutes on a 4.8 degree (°) (8.3%) inclined treadmill, without therapist assistance. Exclusion criteria consisted of Botox injection to the lower extremities in the three months preceding testing, or any musculoskeletal, cardiorespiratory/metabolic, or additional neurological disorder (e.g., Parkinson's disease) that could affect gait. Self-selected overground gait speed was determined from two ten-m overground walking trials. One individual used an AFO, which was required for ankle stability; other assistive devices were not used. Further descriptions of each individual are listed in Table A-1. Prior to participation all individuals signed a University of North Carolina at Chapel Hill Institutional Review Board-approved informed consent form.

Table A-1. Individual description

	Individual 1	Individual 2	Individual 3
Self-Selected Overground Speed (m/s)	.86	.79	.92
Treadmill Speed (m/s)	.65	.65	.92
Gender	Male	Male	Female
Age (years)	69	64	57
Time Post Stroke (months)	205	32	366
Height (centimeter)	183	183	170
Weight (kilogram)	93	85	91
Paretic Limb (Right/Left)	Left	Right	Left

All experimental trials took place on a dual-belt treadmill (Bertec Corp, Columbus, Ohio), which was instrumented with two six-component force platforms that measured GRF data to be sampled at 960 Hz by a Vicon MX system (Vicon/Peak, Los Angeles, Calirofnia). We chose a treadmill speed (TS) for each individual that we believed could be maintained for a four minute, 4.8° inclined gait training session. If bracing below the knee was used for overground walking, it was retained for treadmill walking. When needed, individuals held one or both treadmill handrails, each instrumented with a load cell (MLP-150; Transducer Techniques, Temecula, Calirfonia) capable of recording vertical force. All individuals wore a safety harness (ZeroG, Aretech, Ashburn, Virgina) while walking which did not restrict lower extremity movements or provide unweighting during testing. Individuals walked on the treadmill for a total of five four-minute trials, consisting of one trial at TS and zero grade, two trials that varied speed (by ± ten % TS) and two trials that varied grade (by ± 4.8°). Table 2 includes a description of each trial. The grade of the inclined and declined trials was selected based on the 2010 Americans with Disabilities Act Standards for Accessible Design (Section 405.2). The second minute of each trial was used for analysis. Steps were removed from a trial if an individual's feet did not fall on separate force platforms or if an individual experienced a stumble. For one trial we were unable to obtain a minimum of ten consecutive steps from the second minute of walking, and instead analyzed the third minute.

Rates of oxygen and carbon dioxide production were recorded in thirty second intervals using a portable metabolic system (Oxycon Mobile, Viasys Healthcare, Yorba Linda, California) during each walking trial and during one four-minute standing trial taken prior to the walking trials. Visual inspection of the rates of oxygen consumption during the last one minute of each trial confirmed that individuals were at steady-state during this period.

The ILM described by Donelan et al. [8], was used to calculate external mechanical power performed on the COM by the paretic and non-paretic limbs through custom written MATLAB (MathWorks, Natick, Massachusetts) programs. GRF data were first filtered with a twenty-five Hz low pass filter. Vertical force data from handrail support, when produced, were included in net force data prior to calculation of COM acceleration. Since the ILM assumes symmetric gait, and spatiotemporal asymmetries are often exhibited following stroke [17,18], the calculations in Donelan et al. [8] were adjusted by: (1) assuming symmetry over strides, instead of steps, and (2) subtracting average COM acceleration over a trial from instantaneous COM acceleration prior to integration. COM acceleration integration constants for the X, Y and Z directions were calculated by minimizing the average COM velocity over a stride. Additionally, in the Y direction, an integration constant equaling treadmill speed was added. For all trials, force vectors were lateral (X direction), parallel (Y direction) and perpendicular (Z direction), respectively, to the treadmill surface. Body weight was adjusted accordingly for net force calculations of inclined and declined trials.

Instantaneous mechanical power generated by each limb was calculated as the dot product of that limb's GRF vector and the COM velocity, as per the ILM [8]. For each stride, instantaneous mechanical power was then normalized to 101 points and averaged for each individual to produce mean P_{inst}.

To obtain positive and negative average mechanical work done on the COM, instantaneous mechanical power generated by each limb was cumulatively integrated over the following phases: DST (from heel-strike of the contralateral limb to toe-off of the reference limb), DSL (from heel-strike of the reference limb to toe-off of the contralateral limb), and SS (from toe-off of the contralateral limb until heel-strike of the contralateral limb), and over a complete stride, restricting integration to positive or negative areas of the integrand, respectively. The positive and negative average mechanical work values for each limb were then multiplied by phase frequency over a trial (for the measures of average mechanical work produced over DST, DSL and SS) or stride frequency over a trial (for the measures of average mechanical work produced over a stride) to yield $+P_{avg}$ and $-P_{avg}$. $+P_{avg}$ and $-P_{avg}$ were summed to obtain P_{avg} for each phase and over a stride. Combined positive average mechanical power performed over one stride ($+P_{cavg}$) was obtained by summing $+P_{avg}$ for the non-paretic and paretic limbs. P_{inst}, $+P_{avg}$, $-P_{avg}$, P_{avg} and $+P_{cavg}$ were normalized to body mass.

Metabolic data from the last one minute of each trial was averaged and oxygen consumption and carbon dioxide production were converted to metabolic powers using standard equations [19]. Net metabolic power (P_{met}) during each walking trial was calculated by subtracting standing metabolic power from metabolic power during each walking trial. P_{met} was normalized to body mass.

Metabolic efficiency of positive work ($+\eta_{work}$) was determined by dividing $+P_{cavg}$ by P_{met} for each trial. Negative work was not included in the metabolic efficiency calculation. The amount of negative work that is stored by the body and returned as positive work is unknown, and therefore including negative work would introduce new inaccuracies to the calculation [20]. Additionally, negative work accounts for a minimal portion of metabolic power compared to positive work [20].

Figure A-1 displays P_{inst} for the walking trials with variation in speed and Figure A-2 displays P_{inst} for the walking trials with variation in surface gradient. Table A-2 and Table A-3 list mean P_{avg} during DSL, DST, SS and over a stride, mean $+P_{avg}$ over a stride and mean $-P_{avg}$ over a stride. Table A-4 lists $+P_{cavg}$, P_{met} and $+\eta_{work}$ for each individual.

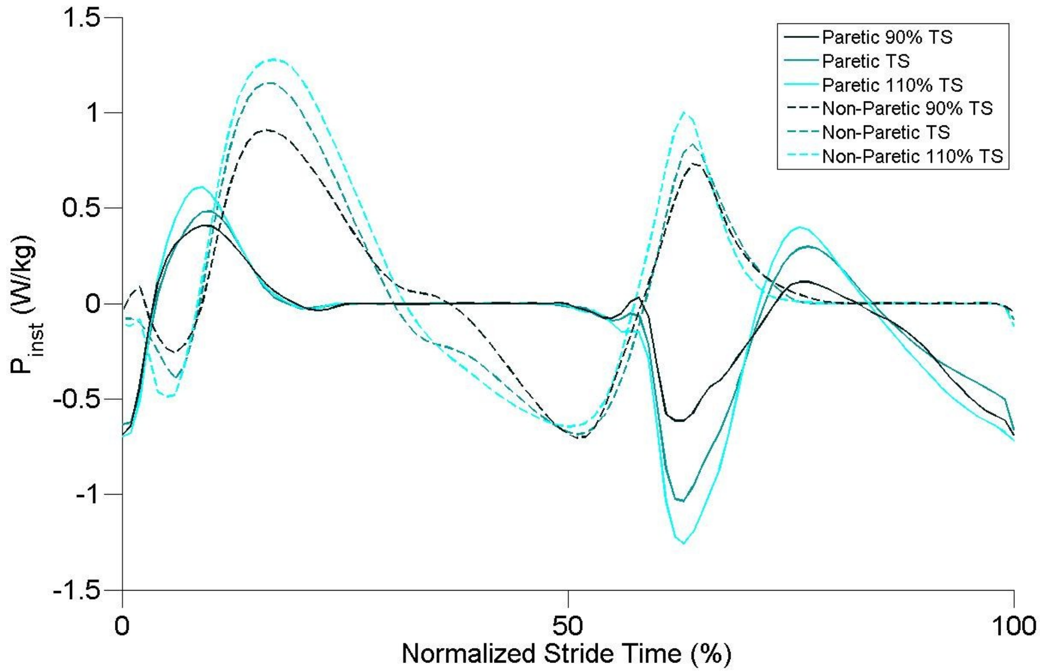

Figure A-1. Mean Pinst plotted for a complete stride for varying treadmill speeds at 0° grade

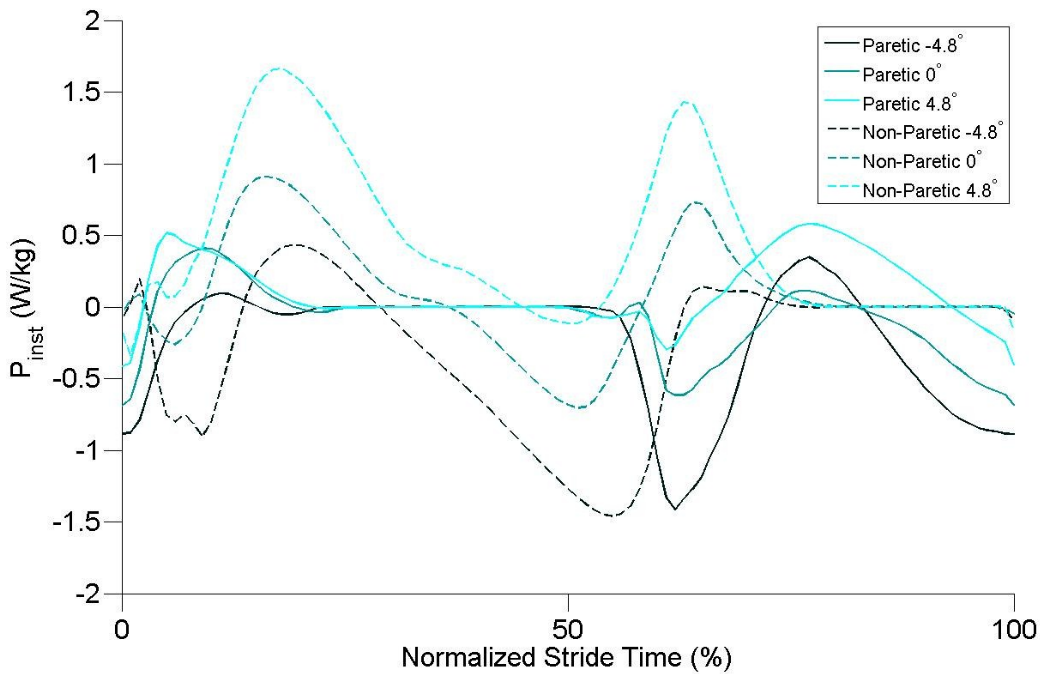

Figure A-2. Mean Pinst plotted for a complete stride for varying surface gradients at 90% TS

Table A-2. Mean average power values for varying treadmill speeds at 0° surface gradient

Condition	Power (W/kg)					
	90% TS, 0°		TS, 0°		110% TS, 0°	
	Non-Paretic	Paretic	Non-Paretic	Paretic	Non-Paretic	Paretic
P_{avg} DST	.40±.12	.11±.051	.45±.19	.11±.10	.59±.23	.16±.066
P_{avg} DSL	.29±.16	-.31±.32	.35±.14	-.61±.53	.31±.10	-.79±.67
P_{avg} SS	-.10±.11	-.18±.10	-.12±.14	-.094±.11	-.10±.18	-.14±.084
P_{avg} Stride	.092±.064	-.082±.056	.085±.091	-.089±.078	.12±.11	-.10±.085
+P_{avg} Stride	.24±.054	.068±.021	.26±.043	.081±.024	.30±.040	.10±.027
-P_{avg} Stride	-.15±.039	-.15±.073	-.17±.051	-.17±.090	-.18±.067	-.20±.11

Table A-3. Mean average power values for varying surface gradients at 90% treadmill speed

Condition	Power (W/kg)					
	90% TS, -4.8°		90% TS, 0°		90% TS, +4.8°	
	Non-Paretic	Paretic	Non-Paretic	Paretic	Non-Paretic	Paretic
P_{avg} DST	-.18±.15	-.15±.20	.40±.12	.11±.051	.76±.18	.22±.021
P_{avg} DSL	-.27±.27	-.97±.50	.29±.16	-.31±.32	.66±.097	.042±.15
P_{avg} SS	-.52±.099	-.29±.17	-.10±.11	-.18±.10	.41±.17	.20±.028
P_{avg} Stride	-.28±.068	-.23±.12	.092±.064	-.082±.056	.42±.12	.11±.029
$+P_{avg}$ Stride	.10±.044	.056±.12	.24±.054	.068±.021	.46±.094	.17±.017
$-P_{avg}$ Stride	-.38±.075	-.29±.0071	-.15±.039	-.15±.073	-.040±.030	-.063±.042

Table A-4. Individual $+P_{tavg}$, P_{met} and $+\eta_{work}$ values

Condition		Individual 1			Individual 2			Individual 3		
Speed (m/s)	Grade (°)	$+P_{cavg}$	P_{met}	$+\eta_{work}$	$+P_{cavg}$	P_{met}	$+\eta_{work}$	$+P_{cavg}$	P_{met}	$+\eta_{work}$
TS	0	.34	--	--	.29	3.3	.089	.39	3.9	.10
110% TS	0	.42	--	--	.33	3.4	.096	.46	4.4	.10
90% TS	0	.39	--	--	.24	2.7	.088	.30	3.6	.082
90% TS	4.8	.61	--	--	.56	4.5	.13	.73	5.3	.14
90% TS	-4.8	.16	--	--	.12	2.8	.043	.20	2.8	.072

REFERENCES

1. Ganley, K. J., Herman, R. M., & Willis, W. T. 2008 Muscle metabolism during overground walking in persons with poststroke hemiparesis. *Top. Stroke Rehabil.* ***15*****(3)**, 218-226.

2. Bowden, M.G., Balasubramanian, C.K., Neptune, R.R., Kautz, S.A. 2006 Anterior-Posterior Ground Reaction Forces as a Measure of Paretic Leg Contributions in Hemiparetic Walking. *Stroke **37(3)**, 872-876.*

3. Dal, U., Erdogan, T., Resitoglu, B., & Beydagi, H. 2010 Determination of preferred walking speed on treadmill may lead to high oxygen cost on treadmill walking. *Gait&Posture* ***31*****(3)**, 366-369.

4. Perry, J., Garret, M., GrOlney, J.K., Mulroy, S.J. 1995 Classification of walking handicap in the stroke population. *Stroke* **26(2)**, 982-989.

5. Olney, S.J., Griffin, M.P., Monga, T.N., McBride, I.D. 1991 Work and Power in the Gait of Stroke Patients. *Arch. Phys. Med. Rehabil.* **72(5)**, 309-314.

6. Stoquart, G., Detrembleur, C., Lejeune, T.M. 2012 The reasons why stroke patients expend so much energy to walk slowly. *Gait&Posture **36(3)**, 409-413.*

7. Detrembleur, C., Dierick, F., Stoquart, G., Chantraine, F., & Lejeune, T. 2003 Energy cost, mechanical work, and efficiency of hemiparetic walking. *Gait&Posture* **18(2)**, 47-55.

8. Donelan, J.M., Kram, R.and Kuo, A.D. 2002 Simultaneous positive and negative external mechanical work in human walking. *J. Biomech* **35(1)**, 117-124.

9. Kuo, A.D., Donelan, J.M. and Ruina, A. 2005 Energetic Consequences of Walking Like an Inverted Pendulum: Step-to-Step Transitions. *Exerc. Sport Sci.* **33(2)**, 88-97.

10. Soo, C.H. and Donelan, J.M. 2012 Coordination of push-off and collision determine the mechanical work of step-to-step transitions when isolated from human walking. *Gait&Posture* **35(2)**, 292-297.

11. Donelan, J. M., Kram, R., & Kuo, A. D. 2002 Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. *J. Exp. Biol.* **205(23)**, 3717-3727.

12. Houdijk, H., Pollman, E., Groenewold, M., Wiggerts, H., Polomski, W. 2009 The energy cost for the step-to-step transition in amputee walking. *Gait&Posture* **30(1)**, 35-40.

13. Doets, H.C., Vergouw, D., Veeger, H.E.J., Houdijk, H. 2009 Metabolic cost and mechanical work for the step-to-step transition in walking after successful total ankle arthroplasty. *Hum. Movement Sci.* **28(6)**, 786-797.

14. Turns, L.J., Neptune, R.R., Kautz & S.A. 2007 Relationships between Muscle Activity and Anteroposterior Ground Reaction Forces in Hemiparetic Walking. *Arch. Phys. Med. Rehabil.* **88(9)**, 1127-1135.

15. Peterson. 2010 Pre-swing deficits in forward propulsion, swing initiation and power generation by individual muscles during hemiparetic walking. *J. Biomech* **43(12)**, 2348-2355.

16. Lewek, M.D. and E.P. Randall. 2011 Reliability of spatiotemporal asymmetry during overground walking for individuals following chronic stroke*. J. Neurol Phys Ther.* **35(3)**, 116-21.

17. Patterson, K.K., Parafianowicz, I., Danells, C.J., Closson, V., Verrier, M.C., Staines, W.R., Black, S.E., McIlroy, W.E. 2010 Gait asymmetry in community-ambulating stroke survivors. *Arch. Phys. Med. Rehabil.* **89(2)**, 304-310.

18. Lewek, M.D., Randall, E.P. 2011 Reliability of spatiotemporal asymmetry during overground walking for individuals following stroke. *J. Neurol. Phys. Ther.* **35(3)**, 116-121.

19. Brockway, J.M. 1987 Derivation of formulas used to calculate energy-expenditure in man. *Hum. Nutr. Clin. Nutr.* **41(6)**, 463-471.

20. Farris, D.J. and Sawicki, G.S. 2012 The mechanics and energetics of human walking and running: a joint level perspective. *J. R. Soc.* **9(66)**, 110-118.