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Original Article

Title: Individual Limb Mechanical Analysis of Gait Following Stroke

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Abstract

The step-to-step transition of walking requires significant mechanical and metabolic energy to redirect the center of mass. Inter-limb mechanical asymmetries during the step-to-step transition may increase overall energy demands and require compensation during single-support. The purpose of this study was to compare individual limb mechanical gait asymmetries during the step-to-step transitions, single-support and over a complete stride between two groups of individuals following stroke stratified by gait speed (≥0.8 meters per second (m/s) or <0.8 m/s). Twenty-six individuals with chronic stroke walked on an instrumented treadmill to collect ground reaction force data. Using the individual limbs method, mechanical power produced on the center of mass was calculated during the trailing double-support, leading double-support, and single-support phases of a stride, as well as over a complete stride. Robust inter-limb asymmetries in mechanical power existed during walking after stroke; for both groups, the non-paretic limb produced significantly more positive net mechanical power than the paretic limb during all phases of a stride and over a complete stride. Interestingly, no differences in inter-limb mechanical power asymmetry were noted between groups based on walking speed, during any phase or over a complete stride. Paretic propulsion, however, was different between speed-based groups. The fact that paretic propulsion (calculated from anterior-posterior forces) is different between groups, but our measure of mechanical work (calculated from all three directions) is not, suggests that limb power output may be dominated by vertical components, which are required for upright support.

Introduction

During the single-support (SS) phase of unimpaired gait, the center of mass (COM) follows a path similar to the motion of an inverted pendulum (Donelan et al., 2002b). During the step-to-step transition, mechanical work is required to redirect the COM velocity vector between the pendulum arcs of each limb (Donelan et al., 2002b; Soo and Donelan, 2012). Redirection comes from the net combination of: (1) positive work produced during the trailing limb's double-support (DST) phase and (2) negative work produced during the leading limb's double-support (DSL) phase (Donelan et al., 2002b; Soo and Donelan, 2012). Minimizing total mechanical work is desirable to minimize metabolic cost (Donelan et al., 2002a; Kuo et al., 2005), and can occur when the timing and magnitude of the leading limb's negative work is equal to the trailing limb's positive work (Ellis et al., 2013; Kuo et al., 2005; Soo and Donelan, 2012). However, even when this occurs, both experimental and simulation studies indicate that the step-to-step transition requires a substantial amount of metabolic energy relative to the total requirements of a stride (Donelan et al., 2002a; Kuo et al., 2005; Umberger, 2010).

Divergence from metabolic optimization has been shown to arise from inter-limb mechanical asymmetries during step-to-step transitions in both healthy (Ellis et al., 2013; Soo and Donelan, 2012) and clinical (Bonnet et al., 2014; Doets et al., 2009; Feng et al., 2014; Houdijk et al., 2009) populations. For example, imposing temporal asymmetry on otherwise healthy gait leads to highly asymmetric step-to-step transition mechanics and increases metabolic cost up to 80% (Ellis et al., 2013). Similarly, the affected limb of individuals following unilateral transtibial amputation (Houdijk et al., 2009) or total ankle arthroplasty (Doets et al., 2009) exhibited less positive work production during DST and the unaffected limb exhibited greater negative work production during DSL. In these studies, impaired positive work production during DST is suggested to necessitate greater negative work production from the leading limb to redirect the COM and greater positive work production during SS; all compensations that lead to higher metabolic demand (Doets et al., 2009; Houdijk et al., 2009; Soo and Donelan, 2012).

In individuals following stroke, unilateral impairments in muscle function, commonly paretic plantar-flexors (Allen et al., 2011; Lamontagne et al., 2007b; Peterson et al., 2010; Turns et al., 2007), yield reductions in positive power during DST. An analysis using the individual limbs method (ILM; Donelan et al., 2002b), examining the SS and DST phases together, revealed greater positive mechanical work production by the non-paretic limb to raise the COM (Stoquart et al., 2012). Importantly, this greater mechanical work production was correlated with greater metabolic cost (Stoquart et al., 2012), potentially limiting gait speed and endurance. Inter-limb mechanical asymmetries for the separate phases of DST and DSL, when symmetry appears to be an important factor in gait efficiency (Ellis et al., 2013; Soo and Donelan, 2012), and SS, have yet to be comprehensively examined in individuals post-stroke. In addition, although previous studies have noted a relationship between functional recovery and gait symmetry post-stroke using spatiotemporal measures (Balasubramanian et al., 2007; Patterson et al., 2008) and anterior-posterior ground reaction forces (Bowden et al., 2006), the relationship between ILM mechanical symmetry and function remains unknown.

The purpose of this study was to examine gait asymmetry in individuals with post-stroke hemiparesis by quantifying asymmetry from a mechanical power perspective. Based on previous analyses examining individual limb mechanics in patient populations with unilateral impairments (Doets et al., 2009; Houdijk et al., 2009), we hypothesized that: (1) individuals post-stroke would 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), and (2) mechanical asymmetries between limbs would be greater in the group of individuals with reduced gait speed.

Materials and Methods

Experimental Protocol

A retrospective analysis was conducted at the University of North Carolina at Chapel Hill, using data formerly collected through two research protocols examining gait characteristics in individuals following stroke. Data from 47 individuals who presented with chronic hemiparesis were analyzed; 26 individuals met inclusion/exclusion criteria. Inclusion criteria included: unilateral, non-cerebellar brain lesion due to stroke; > 6 months since stroke; ability to walk ≥ ten meters overground; ability to walk ≥ two minutes on a treadmill without therapist assistance, or harness unweighting. Exclusion criteria included: Botox injection to the lower extremities in the three months preceding testing; musculoskeletal, cardiorespiratory, metabolic, or additional neurological disorder that could affect gait.

Individuals presented with a range of walking abilities, and were stratified into two groups based on self-selected overground gait speed (Perry et al., 1995): 13 individuals walking at a speed classifying them as ‘community’ walkers (≥0.8 meters per second (m/s)) were considered high gait function and 13 individuals walking at a slower speed (<0.8 m/s) were considered low gait function. Overground gait speed was determined from three passes across a 4.3 m GAITRite mat (CIR Systems, Sparta, New Jersey) (Lewek and Randall, 2011). Individuals used assistive devices and bracing below the knee (e.g., ankle-foot orthosis; AFO) if necessary. Prior to participation, all individuals signed a University of North Carolina at Chapel Hill Institutional Review Board approved informed consent form.

Data Collection

Data collection took place on a dual-belt treadmill (Bertec Corporation, Columbus, Ohio), which was instrumented with two six-component force platforms that sampled ground reaction force (GRF) data at 1080 Hz by a Vicon MX system (Vicon, Los Angeles, California). Some individuals had not been on a treadmill since their stroke and thus did not feel comfortable walking at their self-selected overground gait speed. We therefore chose the fastest treadmill speed that we believed could be maintained for each individual (Rhea et al., 2012). If bracing was used for overground walking, it was retained for treadmill walking. All subjects in the slow-speed group and four subjects in the fast-speed group held onto one or both side-mounted 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 a stumble occurred. For five individuals, we were unable to obtain a minimum of ten consecutive steps of usable data from the second minute of walking (due to stumbles or cross-over while walking) and instead analyzed a later minute.

Data Management and Processing

GRF data were low-pass filtered at a cut-off of 25 Hz in Visual3D software (C-Motion, Germantown, Maryland). Instantaneous contributions to external mechanical power from each limb were calculated according to the ILM described by Donelan et al. (Donelan et al., 2002b) using custom written MATLAB (MathWorks, Natick, Massachusetts) programs. Briefly, this method computes the COM velocity from external forces (we included vertical handrail reaction force, as necessary) and body mass. Net forces were divided by mass and then integrated to calculate COM velocity. The dot product of COM velocity and each limb's GRF gives the instantaneous external mechanical power provided by each limb. An assumption of the ILM is that gait is periodic and integration is performed over each successive periodic cycle. This cycle is normally a step (Donelan et al., 2002b) but because of the step asymmetries that exist in walking post-stroke (Lewek and Randall, 2011; Patterson et al., 2010), we modified the procedure by performing integration over successive strides. For each stride, instantaneous external mechanical power was normalized to 101 points/stride and averaged for each individual to produce mean instantaneous external mechanical power (P_{inst}).

To obtain average net external mechanical work, instantaneous external mechanical power generated by each limb was 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), SS (from toe-off of the contralateral limb until heel-strike of the contralateral limb), and over a complete stride. The average net external mechanical work values for each limb were then multiplied by phase frequency over a trial (for the measures of average net external mechanical work produced over DST, DSL and SS) or stride frequency over a trial to obtain total average net external mechanical power (P_{avgNET}) for each phase and over a stride. The main outcome variables were therefore: paretic and non-paretic limb peak P_{inst} during DSL and DST, and P_{avgNET} during DSL, DST, SS, and over a stride.

Secondary measures included spatiotemporal measures, paretic propulsion, and peak vertical handrail forces obtained during treadmill walking. The step length of the paretic and non-paretic limbs was used to calculate step length asymmetry as the maximum of the non-paretic and paretic step lengths divided by the sum of the non-paretic and paretic step lengths (Awad et al., 2014). Propulsive impulse was calculated as the integral of positive anterior-posterior GRF over a complete stride for the paretic and non-paretic limbs. Paretic propulsion (Pp) was then calculated as the propulsive impulse of the paretic limb divided by the sum of the propulsive impulse of the paretic and non-paretic limbs (Bowden et al., 2006). Vertical handrail forces were normalized to body mass, and the peak vertical handrail force was selected for each stride. The mean of these peak forces was then calculated over all strides for each subject.

Statistical Analyses

Statistical analyses were performed with SPSS (version 21, IBM, Chicago, Illinois). For the high and low speed-based groups, descriptive statistics (i.e., mean and standard deviation) were calculated for each variable. For all individuals a paired samples t-test (α=0.05) was performed to evaluate differences between self-selected overground gait speed and the treadmill speed used for testing. To examine a relationship previously identified between step length and work production during collision of the same limb (Donelan et al., 2002a), we performed a partial correlation (α=0.05) to relate step length to both peak P_{inst} and P_{avgNET} during DSL for each limb. Six separate two-way (limb x speed-based group) ANCOVAs (α=0.05) were performed to examine differences in peak P_{inst} during DSL and DST, P_{avgNET} during DSL, DST, SS and over a stride. Separate one-way (speed-based group) ANCOVAs (α=0.05) were performed to examine the difference in group for step length asymmetry and Pp over a stride. Given the known effect of gait speed on limb mechanical power output (Donelan et al., 2002b) we controlled for treadmill speed when performing the partial correlation and all ANCOVAs.

Results

The mean treadmill speed of all individuals was slower than the mean self-selected overground gait speed (p=0.004) (Table 1). Step length asymmetry was not different between the high and low groups (p=0.648; see Table 1); within these groups respectively, 7 (of 13), and 9 (of 13) had longer paretic (compared to non-paretic) step lengths. There was a significant correlation between the paretic limb's step length and peak negative P_{inst} during DSL (r=-0.446, p=0.026), but no relationship for the non-paretic limb (r=0.047, p=0.822). There was no relationship between step length and P_{avgNET} during DSL for the paretic (r=-0.367, p=0.071) or non-paretic limbs (r=0.331, p=0.107). Pp was significantly greater (p=0.050) in the high compared to the low group. Peak vertical handrail forces from the non-paretic upper extremity were significantly lower (p<0.001) in the high group compared to the low group.

For all measures of power (P_{inst} during DSL and DST, P_{avgNET} during DSL, DST, SS and over a stride), there was a significant difference between paretic and non-paretic limb, no difference between speed-based groups, and no interaction effect between limb and speed-based groups (Table 2, Figures 1-2). The paretic limb produced significantly less positive peak P_{inst} and P_{avgNET} during DST, the non-paretic limb produced significantly less negative peak P_{inst} and P_{avgNET} during DSL, and the non-paretic limb produced significantly greater positive P_{avgNET}

during SS (each compared to the contralateral limb). The paretic limb produced significantly less positive P_{avgNET} over a stride compared to the non-paretic limb.

Discussion

The external mechanical power results, computed using the ILM, provide strong evidence of interlimb mechanical asymmetry during gait in individuals following stroke, during all phases of a stride and over a complete stride, however this mechanical asymmetry was not more severe for our group of slower walkers. This finding that external mechanical power asymmetries were largely unchanged with speed-based group may impact how we think about walking speed as a measure of function.

Individual Limb Mechanical Power

Over a complete stride, we observed that P_{avgNET} was positive for the non-paretic limb and negative for the paretic limb for both speed-based groups. Within the gait cycle, our data revealed less positive external mechanical power production during paretic DST, less negative external mechanical power production during non-paretic DSL, and more positive external mechanical power production during non-paretic SS (each compared to the contralateral limb). Evaluation of these sub-phases of gait provides enhanced understanding of how limb kinetic compensations are made during gait following stroke.

For example, the DST phase corresponds with push-off at the end of stance; a frequently studied period of the gait cycle following stroke (Allen et al., 2011; Peterson et al., 2010; Stoquart et al., 2012), likely due to the presence of profound plantar-flexor weakness (Allen et al., 2011; Lamontagne et al., 2007a; Peterson et al., 2010; Turns et al., 2007). The plantar-flexors have a primary responsibility to provide limb propulsion (McGowan et al., 2008; Neptune et al., 2001; Peterson et al., 2010), which is often asymmetric following stroke (Balasubramanian et al., 2007; Bowden et al., 2006). Our subjects demonstrated Pp less than 0.50 (0.50= symmetric) and there was less positive peak P_{inst} and P_{avgNET} from the paretic limb (compared to the non-paretic limb) during DST across both speed-based groups. Although this is likely due to plantar-flexor weakness on the paretic side (Peterson et al., 2010), examination of mechanical power production at the joint level is needed to confirm this.

During DSL, we observed less negative P_{inst} and P_{avgNET} from the non-paretic limb (compared to the paretic limb), which was contrary to our hypothesis based on analyses of other patient populations (Doets et al., 2009; Houdijk et al., 2009). Less negative P_{avgNET} from the non-paretic limb could be attributed to the presence of positive mechanical power production during late non-paretic DSL (as seen in Figure 1), which in unimpaired individuals does not typically begin until SS (Donelan et al., 2002b). The functional consequences of this phase advancement in non-paretic positive power production is unclear, but may indicate earlier or greater non-paretic limb initiation to compensate for less paretic limb propulsive power during DST (Raja et al., 2012). In addition, a majority of individuals within this study exhibited longer steps in the paretic versus non-paretic limb that caused a step length asymmetry (Donelan et al., 2002a), which is common following stroke (Patterson et al., 2010). Step length is positively correlated with negative mechanical work production during heel-strike (Donelan et al., 2002a), which corresponds with the DSL phase of our analysis. For our individuals, a correlation was observed between step length and peak negative P_{inst} during DSL for the paretic limb, but not for the non-paretic limb. It appears, therefore, that in addition to phase advancement of positive power production from the non-paretic limb, deficits in step length symmetry may be responsible, in part, for the mechanical asymmetry during DSL.

During SS, we observed greater P_{avgNET} from the non-paretic limb (compared to the paretic limb) across both speed-based groups. This is likely due, in part, to the continuation of positive mechanical power produced during late non-paretic DSL into SS. Forward dynamic models have previously suggested that this early phase of SS is a critical period for raising the body's COM (Neptune et al., 2004). In addition, we observed negative mechanical power production by the paretic limb during late SS, which persisted into paretic DST and may have contributed to the reduction in P_{avgNET} during paretic DST. The combination of these results yields a profound interlimb mechanical asymmetry during SS that produces acceleration/deceleration and rise/fall of the COM with each non-paretic/paretic stance, respectively. Rather than maintaining a smooth trajectory of COM motion as observed in unimpaired individuals (Donelan et al., 2002b), the result appears to be an inefficient method of maintaining forward progress during walking (Stoquart et al., 2012).

Asymmetry and Walking Speed

Although we observed significant inter-limb external mechanical power asymmetries during each phase of the stride and over a complete stride, these asymmetries were not different between speed-based groups. Olney et al. (1991) reported comparable findings through a joint level analysis, suggesting that inter-limb asymmetry of positive mechanical work production over a complete stride did not relate to gait speed. Interestingly, our subject's Pp differed between speed-based groups, similar to the results presented by Bowden et al. (2006). This suggests that measures of mechanical asymmetry based on sagittal plane kinetics (i.e. anterior-posterior ground reaction force) may be more closely related to function, as classified by walking speed, than work-based metrics that account for multiple joints, and in all three dimensions (i.e. ILM). The fact that Pp (calculated from anterior-posterior forces only) is different between groups, but our measure of mechanical work (calculated from all three directions) is not, suggests that power output of the paretic limb post-stroke may be dominated by vertical components, which are required for upright support. Furthermore, it could be that those who recover well in the anterior-posterior direction (e.g. as reflected by Pp) appear to have a better functional outcome, at least with respect to walking speed.

The finding that mechanical power asymmetries were largely unchanged with group may also impact how we think about walking speed as a measure of function. It appears that the ability to walk faster was the result of greater compensation with the non-paretic limb (Bowden et al., 2006). The use of walking speed as a primary outcome measure in many studies, while an important measure of function, may also represent the ability to compensate with the non-paretic limb. Previous analyses examining external mechanical work for individuals following stroke revealed greater positive mechanical work production by the non-paretic limb to raise the COM (Stoquart et al., 2012) which was related to metabolic energy use, another indicator of walking function. Further work will need to be done to establish the respective importance of inter-limb mechanical asymmetries in each movement direction (i.e., vertical, anterior-posterior, and medial-lateral) to functional abilities including walking speed, metabolic energy use and dynamic balance.

Limitations

Our analysis method (ILM) has some limitations. Simulation analyses performed have shown that under reasonable assumptions regarding muscle activity that external work correlates poorly with musculotendon work (Neptune et al., 2009; Neptune et al., 2004; Sasaki et al., 2009). Of concern is that external mechanical work calculations, such as those employed in ILM, exclude muscle co-contraction and thus cannot account for simultaneous negative and positive muscle work across joints (Neptune et al., 2009; Neptune et al., 2004; Sasaki et al., 2009) that may be used to stabilize the body against gravity at significant energy cost. Additionally, external mechanical work calculations cannot accurately partition contributions of muscular versus passive elastic tissue contributions to limb work (e.g. elastic energy storage and release) and do not explicitly include internal mechanical power (e.g., from the motion of the swing limb). However, Zelik and Kuo (2010) reported a qualitative correspondence between inverse dynamics and external mechanical work rates, and attributed the differences that were observed, during DSL and the beginning of SS, to energy dissipation and elastic rebound of soft tissue, respectively, which are not captured through joint-based calculations.

Individuals that required an AFO to provide ankle stability and/or prevent toe drag continued to use the AFO during data collections. In the same way, individuals that required upper limb support for stability and balance used treadmill handrail support during data collections. AFOs and handrail support may have affected power generation and absorption throughout the stride, however we felt it best to retain the use of both during testing to replicate normal every-day gait as closely as possible. The effect of AFO use is difficult to quantify in our data, however we were able to quantify handrail use in the vertical direction. Our handrail-mounted transducers indicated small vertical handrail support forces (all subjects: 7.5 ±5.6 % BW). Based on the low magnitude of observed vertical handrail forces, we expect that the unmeasured anterior-posterior handrail forces were also small. We note, however, that handrail forces do have the potential to cause an error in COM velocity calculations based on ground reaction force data alone. For example, an individual exerting large anterior-posterior handrail forces could reduce the need for the non-paretic limb to compensate during DSL. We recognize this as a limitation to our study, however, the use of upper-limb support also replicates normal every-day gait as closely as possible (i.e. the use of cane/walker).

Individuals post-stroke, exhibit a number of movement patterns, such as hip hiking, stiff-knee gait, and drop foot (De Quervain et al., 1996; Mulroy et al., 2003), which may be more pronounced in individuals in the lower speed-based group. These factors could result in greater mechanical asymmetry but may not be reflected in external mechanical power calculations. An alternative approach to studying mechanical energetics post-stroke is to use forward dynamic modeling which can be performed at the individual muscle-level, producing values that should include co-contraction (Peterson et al., 2011). Musculotendon modeling results in healthy gait have corresponded to joint-based results during DST and DSL (Neptune et al., 2009), but have been shown to exhibit the greatest positive and net mechanical work over a gait cycle during the beginning of SS (Neptune et al., 2009; Neptune et al., 2004; Sasaki et al., 2009). This is contrary to the pendulum model and inverse dynamic calculations (where the greatest positive and net mechanical work over a gait cycle occurs during DST), and suggests that a significant amount of work that occurs during the beginning of SS is due to muscle co-contraction, believed to control hip and knee flexion and provide lower-limb stability (Neptune et al., 2009). In short, more studies including simultaneous measurements of symmetry using multiple metrics based on varied analysis techniques (e.g. temporal and spatial kinematics, paretic propulsion (Pp), ILM, inverse dynamics, forward dynamics computer simulations, ultrasound imaging) are needed to elucidate the impact of symmetry on mechanical and metabolic energy expenditure post-stroke.

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Conflict of Interest: None.

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Figure Captions

Figure 1. Mean P_{inst} over separate phases of a stride (1: non-paretic DSL/paretic DST; 2: non-paretic SS; 3: paretic DSL/non-paretic DST; 4: paretic SS) for the (A) high, and (B) low speed-based groups. Light grey lines represent non-paretic limb and black lines represent paretic limb. Average non-paretic limb heel strike occurs at 0 normalized stride time and dark grey shading indicates phases of step-to-step transitions.

Abbreviations: P_{inst}, mean instantaneous external mechanical power; DSL, leading double-support; DST, trailing double-support; SS, single-support; W/kg, Watts per kilogram

Figure 2. Mean P_{avgNET} over separate phases of a stride (1: non-paretic DSL/paretic DST; 2: non-paretic SS; 3: paretic DSL/non-paretic DST; 4: paretic SS) for the (A) high, and (B) low speed-based groups. Light grey bars represent non-paretic limb and black bars represent paretic limb. Dark grey shading indicates phases of step-to-step transitions. Error bars represent one standard deviation.

Abbreviations: P_{avgNET}, total average net external mechanical power; DSL, leading double-support; DST, trailing double-support; SS, single-support; W/kg, Watts per kilogram

Table 1. Speed-based group description

	High (n=13)	Low (n=13)
Self-Selected Overground Speed (m/s)		
Range (min/max)	.83/1.3	.19/.78
Mean	1.0±.16	.52±.20
Treadmill Speed (m/s)		
Range (min/max)	.49/1.3	.15/.70
Mean	.90±.20	.50±.18
Gender (male/female)	7/6	7/6
Age (years)	56±8.4	54±12
Time Post Stroke (months)	103±92	30±17
Height (cm)	175±8.4	173±9.3
Weight (kg)	91±18	93±13
Lower Extremity Fugl-Meyer	28±2.1	22±4.0
Paretic Limb (right/left)	7/6	7/6
Swing Time (s)		
Non-paretic	.38±.04	.37±.07
Paretic	.42±.06	.56±.10
Stance Time (s)		
Non-paretic	.81±.13	1.3±.30
Paretic	0.77±.11	1.1±.27
Step Length (cm)		

Non-paretic	50±7.7	37±10
Paretic	50±8.4	40±13
Step Length Asymmetry	0.52±0.02	0.55±0.04
P_p Stride	0.41±0.07	0.29±0.13
Peak Vertical Handrail Force (%BW)	3.0±4.0	11.0±4.0

Abbreviations: m/s, meter per second; cm, centimeter; kg, kilogram; s, second; Pp, paretic propulsion; %BW, percent body weight

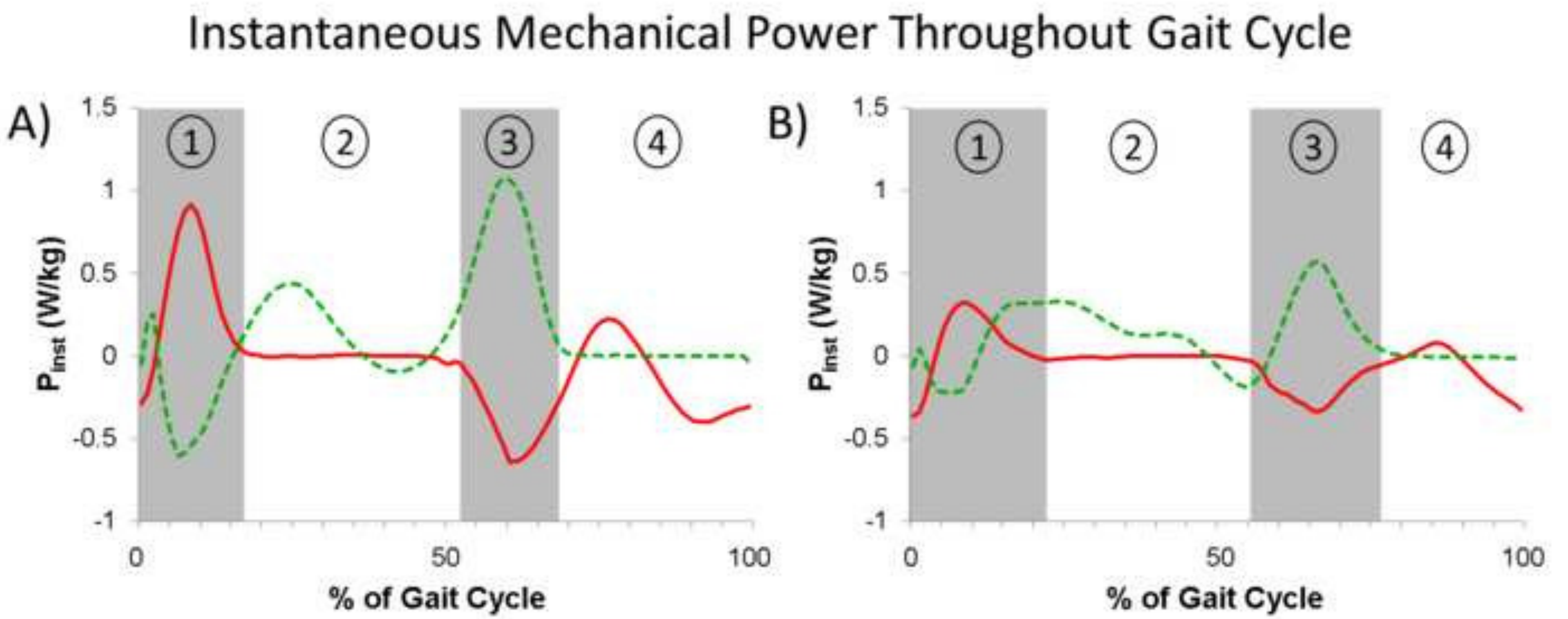
Table 2. Mean mechanical power values and corresponding ANCOVA p-values

	High (W/kg)		Low (W/kg)		p-value		
	Non-Paretic	Paretic	Non-Paretic	Paretic	Interaction Effect	Main Effect (Limb)	Main Effect (Group)
P_{inst} DST	1.30±0.63	0.88±0.53	0.71±0.38	0.36±0.18	0.336	<0.001[1]	0.920
P_{inst} DSL	-0.70±0.46	-0.81±0.31	-0.35±0.28	-0.52±0.27	0.931	0.029[1]	0.179
P_{avgNET} DST	0.74±0.33	0.40±0.31	0.31±0.22	0.08±0.09	0.617	<0.001[1]	0.900
P_{avgNET} DSL	-0.22±0.34	-0.44±0.26	0.05±0.12	-0.22±0.16	0.458	<0.001[1]	0.435
P_{avgNET} SS	0.13±0.13	-0.14±0.13	0.11±0.11	-0.08±0.16	0.774	<0.001[1]	0.334
P_{avgNET} Stride	0.12±0.06	-0.05±0.06	0.11±0.06	-0.04±0.04	0.300	<.001[1]	0.960

Abbreviations: P_{inst} DST, mean instantaneous external mechanical power during trailing double-support; P_{inst} DSL, mean instantaneous external mechanical power during leading double-support; P_{avgNET} DST, total average net external mechanical power during trailing double-support; P_{avgNET} DST, total average net external mechanical power during leading double-support; P_{avgNET} SS, total average net external mechanical power during single-support; P_{avgNET} Stride, total average net external mechanical power over a stride;W/kg, Watts per kilogram

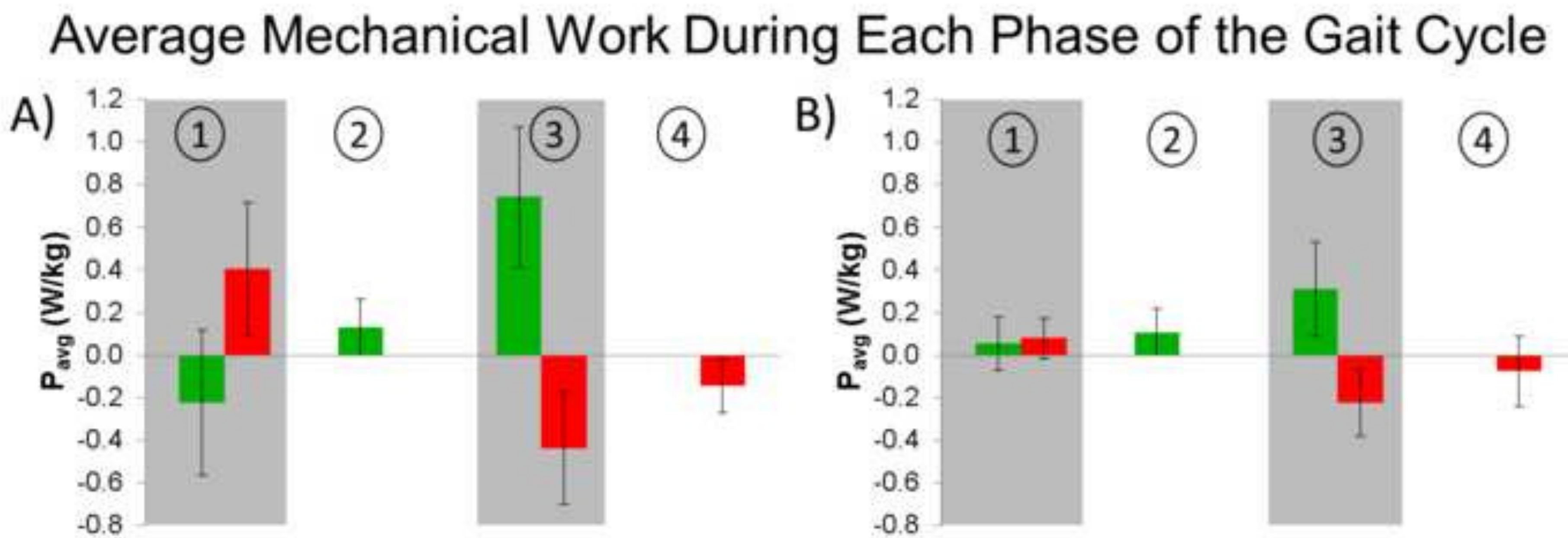
[1]Statistical significance

Figure 1



Instantaneous Mechanical Power Throughout Gait Cycle
A)
B)
1
2
3
4
P_{inst} (W/kg)
1.5
1
0.5
0
-0.5
-1
0
50
100
% of Gait Cycle

Figure 2



Average Mechanical Work During Each Phase of the Gait Cycle
A)
B)
P_{avg} (W/kg)
1.2
1.0
0.8
0.6
0.4
0.2
0.0
-0.2
-0.4
-0.6
-0.8
1
2
3
4