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Altered post-stroke propulsion is related to paretic swing phase kinematics

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Abstract

Background—Gait propulsion is often altered following a stroke, with clear effects on anterior progression. Changes in the pattern of propulsion could potentially also influence swing phase mechanics. The purpose of the present study was to investigate whether post-stroke variability in paretic propulsion magnitude or timing influence paretic swing phase kinematics.

Methods—29 chronic stroke survivors participated in this study, walking on an instrumented treadmill at their self-selected and fastest-comfortable speeds. For each participant, we calculated several propulsion-related metrics derived from anteroposterior ground reaction force or from center of mass power, as well as knee flexion angle and circumduction displacement during the swing phase. We performed a series of linear mixed model analyses to determine whether the propulsion metrics for the paretic leg were related to paretic swing phase mechanics.

Findings—A subset of the stroke survivors exhibited unusual braking forces late in the paretic stance phase, when strong propulsion typically occurs among uninjured controls. Beyond the effects of walking speed or walking condition, these braking forces were significantly linked with altered paretic swing phase mechanics. Specifically, large braking impulses were associated with reduced paretic knee flexion ($p=0.039$) and increased paretic circumduction ($p=0.023$).

Interpretation—The present results suggest that braking forces late in stance are particularly indicative of deficits in the production of typical swing phase kinematics. This relationship

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suggests that therapies designed to address altered swing kinematics should also consider altered force generation in late stance, as these behaviors appear to be coupled.

Keywords

braking; circumduction; propulsion; stiff-knee gait; stroke; walking

1. Introduction

Forward propulsion is a primary mechanical sub-task of human walking, driving the anterior progression of the body's center of mass (CoM) (Neptune et al. 2001; Gottschall and Kram 2003). An inability to effectively generate propulsion may limit mobility, motivating the development of metrics to quantify propulsion magnitude. One such metric, "propulsive impulse", is calculated by integrating the anterior component of the ground reaction force (GRF) under each leg (Bowden et al. 2006), and is proportional to anterior CoM acceleration. A slightly more complex metric applies the "individual limbs method" (ILM) to calculate the mechanical CoM power generated by the forces under each leg (Donelan et al. 2002).

Propulsion magnitude is often altered following a stroke, diverging from the typical pattern of bilaterally-symmetric push-off from the trailing leg late in the stance phase. Compared to uninjured controls walking at matched speeds, the average propulsive impulse during push-off is often reduced for the paretic leg, but near-normal for the non-paretic leg (Combs et al. 2012; Raja et al. 2012). In contrast, recent work found that average paretic push-off ILM power is not significantly reduced relative to controls, while average non-paretic push-off power is increased (Farris et al. 2015). This discrepancy between propulsion metrics may be due to the large vertical component of ILM power (Mahon et al. 2015), as vertical CoM motion can be exaggerated post-stroke (Stoquart et al. 2012). While interesting, these group-average propulsive asymmetries should be interpreted cautiously, as the propulsive pattern can vary substantially across individual stroke survivors (Bowden et al. 2006; Peterson et al. 2010; Allen et al. 2011).

Propulsion timing also appears to change after a stroke. Healthy controls modulate their leg joint torques to produce maximum push-off at a constant time in the gait cycle (Toney and Chang 2016), likely to ensure coordination with forces produced upon contralateral ground contact (Kuo 2002; Kuo and Donelan 2010). Although not yet investigated systematically, the onset of anteriorly-directed GRF among stroke survivors often seems to shift in comparison to controls walking at matched speeds. Specifically, anterior forces tend to be produced later in a step for the paretic leg, and earlier in a step for the non-paretic leg (Bowden et al. 2006; Raja et al. 2012). Similarly, paretic propulsive ILM power tends to begin later in a step than for speed-matched controls (Farris et al. 2015). As with propulsion magnitude, propulsion timing can vary substantially across stroke survivors (Allen et al. 2011).

Beyond the primary effects of propulsion on CoM progression, the combined changes in post-stroke propulsion magnitude and timing may also contribute to altered swing phase kinematics. Trailing leg propulsion has been suggested to provide mechanical energy to the

swing leg, thus contributing to swing phase knee flexion (Neptune et al. 2001; Lipfert et al. 2014). Based on this idea, deficits in paretic propulsion magnitude have been indirectly related to reduced paretic knee flexion during swing (Campanini et al. 2013), which may in turn require increased circumduction to allow safe ground clearance (Chen et al. 2005). Additionally, manipulating propulsion timing in healthy controls can influence the activity of muscles crossing the knee joint near the beginning of the swing phase (Malcolm et al. 2015), a finding attributed to earlier push-off causing greater knee flexion. An altered pattern of propulsion may thus influence swing phase mechanics.

The primary purpose of this exploratory study was to investigate whether post-stroke changes in paretic propulsion magnitude or timing influence the swing phase kinematics of the paretic leg. To address this question, we quantified the magnitude and timing of two sets of candidate metrics related to gait propulsion (calculated from anteroposterior GRF and ILM power). We determined whether these metrics of paretic leg propulsion were related to subsequent paretic swing leg motion across individual stroke survivors. We hypothesized that reduced propulsion magnitude and delayed propulsion timing would be linked with reductions in paretic swing phase knee flexion and increases in paretic circumduction.

2. Methods

2.1. Experimental population

The primary analyses for the present study involved 29 chronic stroke survivors (6 months post-stroke), with general characteristics presented in Table 1. The inclusion criteria were: hemiparesis secondary to a unilateral stroke; absence of significant lower-limb joint pain, limb contractures, or sensory deficits; daily home walking; self-selected treadmill walking speed of 0.3 m/s. Exclusion criteria were: orthopedic or neurologic conditions beyond stroke that would be expected to influence gait; cardiovascular impairments contraindicated to walking. All participants provided informed consent using a form approved by the University of Florida Institutional Review Board and consistent with the Declaration of Helsinki.

2.2. Experimental protocol

Participants walked on a dual-belt instrumented treadmill (Tecmachine; Andezvieux-Boutheon, France) at their self-selected and fastest-comfortable speeds. All trials lasted 30 seconds, separated by at least a one-minute break. Participants were not permitted to hold onto handrails, but wore a harness attached to an overhead rail that would prevent a fall in case of a loss of balance. Our choice to investigate treadmill walking rather than overground walking was based on our reliance on GRF measurements for the propulsion measures of interest. Using an instrumented treadmill allows GRF to be quantified for many consecutive steps and avoids the risk of participants “aiming” their steps toward the force plates during overground walking, factors that have long been cited to justify the use of instrumented treadmills in clinical gait analysis (Dierick et al. 2004). However, extending the results from treadmill walking experiments to overground walking requires caution, as further described in the Discussion.

2.3. Data collection and analysis

Reflective markers (Vicon; Denver, Colorado; USA) in rigid clusters were placed on 13 body segments (head, trunk, pelvis, and bilateral thighs, shanks, feet, upper arms and lower arms). Marker locations were sampled at 100 Hz using a 12-camera Vicon system, and a custom model consisting of 12 segments was created using Visual3D software (Germantown, MD; USA). These segments correspond to the rigid cluster locations listed above, with the trunk and pelvis combined into a single segment. This model was used to determine body segment kinematics, with segment CoM locations calculated using established anthropometric and inertial properties (de Leva, 1996). GRF data were sampled at a rate of 1000 Hz. Both kinematic marker data and GRF data were low-pass filtered at 10 Hz using a zero-lag 4th order Butterworth filter.

Initial contact and foot off events were identified based on the vertical GRF crossing a threshold of 1% body weight. Strides were defined from initial contact to subsequent initial contact with the same leg. Cross-over steps were excluded from analyses.

For each stride, we calculated several propulsion-related metrics based on anteroposterior GRF (see Fig. 1A). We calculated propulsive impulse as the area under the positive portion of the GRF-time trajectory, surrounding the point of maximum anterior GRF. We calculated braking impulses as the areas under the negative portions of this trajectory that preceded (early braking) and followed (late braking) the propulsive impulse. Each impulse value was normalized by body weight. Finally, we identified propulsion onset as the time when the propulsive impulse began, quantified relative to contralateral initial contact as a percentage of stride time.

Whole-body CoM position was calculated from the weighted sum of all individual body segments defined with our 12-segment model. CoM velocity was calculated as the time derivative of CoM position, with anteroposterior velocity corrected to account for treadmill speed. This method of calculating CoM velocity differs from the previous approach (e.g. Farris et al. 2015) of integrating CoM acceleration values calculated from GRF. We chose to use a marker-based approach because post-stroke participants often exhibited anteroposterior drift on the treadmill, violating the assumption of periodic strides necessary for the prior method. We calculated the ILM power contributed by the forces under each leg as the dot product of the three-dimensional CoM velocities and the three-dimensional GRF (Donelan et al. 2002).

For each stride, we calculated several metrics related to propulsion based on ILM power. We first identified the start and end of the step-to-step transitions surrounding double support, as the time points when vertical CoM velocity reached its extreme values (Adamczyk and Kuo 2009; Farris et al. 2015). These time points were restricted to fall within 250 ms of the double support phase (Farris et al. 2015). Based on these time points, the stance phase was divided into three periods: leading leg step-to-step transition; pendulum phase; and trailing leg step-to-step transition (see Fig. 1B). Within each period, we integrated the areas under the power-time trajectories to calculate positive and negative ILM work values, normalized by body mass. Finally, we calculated the time when ILM power first became positive during

the trailing leg step-to-step transition. This timing metric was quantified relative to contralateral initial contact as a percentage of stride time.

In addition to propulsion magnitude and timing, we quantified several swing phase metrics that may be affected by paretic propulsion deficits. Specifically, we identified the peak paretic knee flexion angle using marker kinematic data, and quantified paretic circumduction as the maximum lateral displacement of the paretic heel marker relative to its location at the start of swing (Tyrell et al. 2011; Zissimopoulos et al. 2014).

Across all strides in each trial, we calculated the average value of our metrics of propulsion magnitude, propulsion timing, and swing phase mechanics.

2.4. Statistics

We used a series of linear mixed models to determine whether paretic knee flexion and circumduction during the swing phase were related to our measures of paretic propulsion. We focused on metrics quantifying propulsive behavior near the end of the stance phase, when the majority of propulsion typically occurs. The first set of mixed models were based on anteroposterior GRF metrics, and included the independent variables of trial condition (self-selected or fastest-comfortable speed), walking speed, propulsive impulse, late braking impulse, and propulsive force onset time. We determined whether these five independent variables were significantly related to either swing phase knee flexion or circumduction. The second set of mixed models were based on ILM power metrics, and included the independent variables of trial condition, walking speed, trailing leg transition positive work, trailing leg transition negative work, and trailing leg positive power onset time. We again determined whether these five independent variables were related to either swing phase knee flexion or circumduction. A p-value less than 0.05 was interpreted as a significant effect.

Each mixed model analysis thus included five independent variables and a total of 58 observations (29 participants; 2 walking conditions). This sample size met our *a priori* goal of having at least 10 observations per independent variable, as established effect sizes between propulsion metrics and swing phase kinematics were not available for this exploratory study. To address the potential risk of multicollinearity between our continuous independent variables, we calculated the variance inflation factor (VIF) for our two sets of mixed models. In all cases, the VIF was less than 3.7, below the value of 10 typically interpreted as indicating problematic multicollinearity (Belsey et al. 1980).

3. Results

3.1. Descriptive characteristics of propulsion-related metrics

Individual examples of the quantified propulsion metrics are illustrated in Figure 1. However, the pattern of propulsion varied substantially across participants, so this individual behavior should not be considered typical of all post-stroke propulsive behavior. For descriptive purposes, we here present the range of propulsion-related metrics derived from anteroposterior GRF (Fig. 2) and from ILM-method CoM power (Fig. 3). While the primary focus of this study is on the effects of propulsion variability across individual stroke survivors, more detailed comparisons between stroke survivors and age-matched control

participants walking at matched speeds are presented in Appendix A. Of particular note, a subset of the stroke survivors exhibited large paretic late braking impulses (20/29 for the self-selected walking speed, and 19/29 for the fastest-comfortable walking speed), as defined by an impulse magnitude more than two standard deviations greater than that observed in control participants at a matched speed (see Appendix A).

3.2. Relationships between paretic propulsion and swing phase motion

A subset of the propulsion-related metrics derived from anteroposterior GRFs were related to paretic swing phase motion (Table 2). Specifically, paretic swing phase knee flexion was significantly related to walking speed and the paretic late braking impulse. Knee flexion was reduced with slower walking speeds and larger late braking impulses. Paretic swing phase circumduction was significantly related to walking condition (self-selected vs. fastest-comfortable speed), paretic propulsive impulse, and paretic late braking impulse. Circumduction increased when participants walked at their fastest-comfortable speeds, as well as with smaller propulsive impulses and larger late braking impulses.

In contrast to the propulsion-related metrics derived from anteroposterior GRFs, metrics based on ILM power were not uniquely related to paretic swing phase motion (Table 3). Swing phase knee flexion was only significantly related to walking speed, with reduced knee flexion observed with slower speeds. For these models, circumduction was significantly related to walking conditions and walking speed. Circumduction increased when participants were asked to walk at their fastest-comfortable speed, but also when participants walked more slowly for a given speed condition.

4. Discussion

The pattern of propulsion varied substantially across individual stroke survivors, as quantified by metrics derived from anteroposterior GRF and ILM power. Partially supporting our hypothesis, paretic swing phase knee flexion and circumduction were significantly linked to several propulsion-related metrics, most notably the magnitude of the late braking impulse under the paretic leg. We found no direct evidence that propulsion timing or propulsion metrics derived from ILM power were linked to swing phase mechanics.

The observed patterns of gait propulsion were highly variable across individual stroke survivors. As a general principle, the weaker paretic leg is often thought to contribute less to propulsion and the non-paretic leg to contribute more, scaling with the legs' available mechanical capacity (Milot et al. 2006). While this is indeed the case for many stroke survivors (Bowden et al. 2006; Mahon et al. 2015; Hsiao et al. 2016), a subset of this population actually exhibits greater propulsive impulses under the paretic leg (Balasubramanian et al. 2007). This variability in propulsion symmetry is partially due to differences in trailing leg angle at push-off (Peterson et al. 2010; Hsiao et al. 2015), and is investigated in more detail in Appendix A. For the primary goal of this exploratory study, the inter-subject variability in propulsion metrics allowed us to use statistical methods to investigate potential links between propulsion and swing phase mechanics.

Propulsion metrics based on anteroposterior GRF were significantly linked to subsequent swing phase mechanics. Most notably, large braking impulses late in the stance phase were associated with reduced knee flexion and increased circumduction during the swing phase. Small propulsive impulses were similarly linked with increased circumduction, while the moderate sample size used in this exploratory study may have been insufficient to detect a significant link between propulsive impulse and knee flexion ($p=0.092$). In general, these results are consistent with the idea that less effective push-off provides less mechanical energy to the swing leg to help drive knee flexion (Neptune et al. 2001; Lipfert et al. 2014), and subsequently requires circumduction to achieve ground clearance (Chen et al. 2005). However, we found no evidence that delays in the production of propulsive forces uniquely contribute to reduced knee flexion and increased circumduction.

While post-stroke deficits in propulsive force have previously been investigated in detail (e.g. Bowden et al. 2006; Peterson et al. 2010; Allen et al. 2011; Hsiao et al. 2015; Hsiao et al. 2016), the unusual late braking forces observed in the present study are relatively unexamined. A single study (Turns et al. 2007) has reported that net impulse in late stance (the sum of propulsive and late braking impulses) is correlated with unusual leg flexor activity during this period, suggesting a deficit in coordination. In addition, deficits in controlling the direction of the force produced by the paretic leg on the environment have been reported during pedaling tasks (Rogers et al. 2004; Liang and Brown 2014), results that have been theoretically linked to commonly observed changes in post-stroke gait (Boehm and Gruben 2016). From the present results, it seems that posteriorly-directed forces just before the foot leaves the ground are particularly indicative of an inability to achieve typical swing phase mechanics. This apparent importance of paretic late braking impulse motivated a *post hoc*, small-scale investigation of factors that may contribute to this unusual gait behavior. Specifically, we quantified the trailing leg angle (sagittal plane angle from pelvis CoM to foot CoM) at the end of stance, as push-off could feasibly cause posteriorly directed forces if the ground contact point is anterior relative to the trunk. We also quantified the stance leg unloading rate (defined as the rate of change in vertical GRF over the final 200 ms of stance), as we observed an apparent inability to rapidly shift their weight off the paretic leg in several participants. We found that larger paretic trailing leg angles were significantly ($p<0.001$; Pearson $r=-0.57$) associated with smaller late braking impulses. Similarly, faster unloading rates of the paretic stance leg were also significantly ($p<0.001$; Pearson $r=-0.61$) associated with smaller late braking impulses. From this limited analysis, it appears that braking forces late in the stance phase are related to the body's configuration at push-off, as well as an inability or unwillingness to rapidly unload the paretic leg. Further investigation of these forces seems justified by the present exploratory results.

Unlike propulsion metrics derived from anteroposterior GRF, those derived from ILM power were not significantly linked with swing phase mechanics. This discrepancy is likely due to the different mechanical constructs that are quantified using these two methods. Most obviously, ILM power includes contributions in the anteroposterior, mediolateral, and vertical directions (Donelan et al. 2002), with individual differences perhaps dominated by the relatively large vertical component (Mahon et al. 2015). The present results suggest that dysfunction in propulsion specific to the anteroposterior direction is most strongly related to swing phase behavior.

Beyond the potential effects of the quantified propulsion metrics, walking speed and walking condition were linked with swing phase knee flexion and circumduction. Specifically, faster walking speeds were associated with increased knee flexion, consistent with previous results in the chronic stroke population (Tyrell et al. 2011; Stanhope et al. 2014). The effects of walking speed and condition on circumduction were more complex, likely due in part to the structure of our mixed model analyses. In these analyses, potential effects of walking condition (self-selected vs. fastest-comfortable) derive from within-subject comparisons, while potential effects of walking speed are from between-subject comparisons. Based on both mixed model analyses, the within-subject comparisons found that participants walked with greater circumduction when walking at their fastest-comfortable speed than when walking at their self-selected speed. However, the ILM-based analyses also found that for the between-subject comparisons, individuals who walked faster for a given condition exhibited less circumduction. This complexity may explain the lack of consistency between previous studies investigating the relationship between gait speed and circumduction (Tyrell et al. 2011; Stanhope et al. 2014). Speculatively, it may be the case that challenging stroke survivors to walk faster (the fastest-comfortable condition) accentuates frontal plane gait deviations, while individuals who naturally walk at fairly fast speeds are less likely to have this underlying limitation.

While the present work quantified multiple aspects of post-stroke propulsion, several limitations should be noted. We quantified propulsion-related metrics based on two common approaches, but did not investigate joint mechanics. Therefore, we are unable to determine whether an individual's propulsion deficits are due to weakness or delayed production of specific joint actions (e.g. paretic plantarflexion). At an even more detailed level, we are unable to differentiate between possible distinct effects of soleus and gastrocnemius (Neptune et al. 2001), or of more proximal muscle activity (Riley and Kerrigan 1998) on leg swing kinematics. Additionally, all presented data were collected during treadmill walking, which has been reported to allow stroke survivors to walk with a more typical, symmetric gait pattern (Harris-Love et al. 2001). However, this prior study allowed participants to hold onto handrails for support, likely reducing the level of challenge that may drive stroke survivors to walk with altered propulsion and swing phase mechanics. Therefore, participants in the present study were not permitted to hold onto external supports during the walking trials. More recent work found that the pattern of anteroposterior ground reaction forces (the strongest correlate of swing phase mechanics in the present study) was quite similar between overground and treadmill walking in healthy controls (Goldberg et al. 2008). While stroke survivors tend to slow down when walking on a treadmill, it appears that their motor control deficits (as quantified by step length asymmetries) remain consistent or are even amplified compared to overground walking (Kautz et al. 2011). Therefore, we believe that the increased challenge of treadmill walking is an appropriate environment to identify the factors that may prevent typical swing phase motion.

The present work demonstrates that altered anteroposterior propulsion is linked to post-stroke deficits in swing phase kinematics, but the causality of this relationship is not entirely clear. It is feasible that an inability to generate strong propulsive forces (and no braking) late in the stance phase prevents mechanical energy from being effectively transferred to the upcoming swing phase and driving knee flexion, consistent with previous suggestions

(Neptune et al. 2001; Lipfert et al. 2014). However, the lack of a clear relationship between push-off work and swing phase kinematics may indirectly argue against this interpretation. Alternatively, the relationship between braking force and swing phase kinematics may be due to an inability to transition from the stance phase to the swing phase in a coordinated fashion. For example, stroke survivors often exhibit excessive activation of the paretic knee extensors following movements into hip extension that typically occur near the end of the stance phase (Lewek et al. 2007), which may prevent the swing foot from being lifted off the ground. Additionally, activation of the paretic hip flexor musculature in severely affected stroke survivors prior to fully shifting their weight off of this leg (Turns et al. 2007) could cause a resistive braking force late in stance, due to friction under the foot preventing anterior swing. This lack of paretic leg extensor and flexor coordination could thus contribute to the observed deficits in both propulsion and swing phase mechanics. Future work should investigate whether patient characteristics such as joint range of motion, strength, or spasticity can provide insight into the subset of individuals who walk with unusual late braking forces and altered swing phase kinematics. Identifying the mechanistic cause of the observed relationship may allow the development of interventions to eliminate the posteriorly-directed forces late in the stance phase, and normalize swing phase mechanics. Ultimately, our goal is to use the mechanistic insight resulting from this type of study to promote improved activity and participation among the chronic stroke population through increased gait function.

5. Conclusions

Stroke survivors with a wide range of walking abilities exhibited substantial variation in their patterns of propulsion. Propulsion metrics derived from anteroposterior ground reaction force were significantly linked with swing phase mechanics. Of particular interest, large late braking impulses were associated with reduced swing phase knee flexion and increased circumduction, suggesting the importance of eliminating these late posterior forces as a part of interventions targeted to restoring a typical gait pattern.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Highlights

- Gait propulsion and swing phase kinematics are often altered following a stroke
- Substantial variability is present in post-stroke gait propulsion patterns
- Some stroke survivors exhibit unusual braking forces late in paretic stance
- Late braking paretic impulses are linked to altered swing phase mechanics

Figure 1.

Patterns of propulsion in individual participants. Panel A illustrates anteroposterior ground reaction forces (positive = anterior), with shaded areas indicating paretic leg braking and propulsive impulses. The dashed vertical line indicates the average time of non-paretic leg initial contact. Panel B illustrates CoM power calculated using the individual limbs method, with the shaded areas indicating negative and positive work values. Vertical dashed lines are used to demarcate the leading leg step-to-step transition (LL-SST), the pendulum phase (PP), and the trailing leg step-to-step transition (TL-SST). Both panels illustrate an average stride, from paretic leg initial contact to the subsequent paretic leg initial contact.

Figure 2.

Individual values for metrics derived from anteroposterior GRF. The distribution of early braking impulse magnitudes (A), propulsive impulse magnitudes (B), late braking impulse magnitudes (C), and propulsion onset times (D) are illustrated across all participants. NP indicates the non-paretic leg and P indicates the paretic leg, while SS indicates the self-selected condition and FC indicates the fastest-comfortable condition. Each violin plot (created in Matlab using the `rst_data_plot.m` function) illustrates the estimated probability density function for the indicated metric, with individual data points shown as circles (o) or crosses (+) for identified outliers. The thick horizontal line indicates the median of the distribution, and the vertically-oriented rectangle indicates the 95% High Density Interval.

Figure 3.

Individual values for metrics derived from ILM power. Violin plots illustrate the distribution of positive work magnitudes during the leading leg transition (A), pendulum phase (B), and trailing leg transition (C), the onset of trailing leg positive power (D), and negative work magnitudes during the leading leg transition (E), pendulum phase (F), and trailing leg transition (G). Panels follow the same structure as Figure 2.

Table 1.

Basic demographic, clinical, and gait characteristics.

Characteristic	Value
Gender (female / male)	11 / 18
Paretic side (right / left)	14 / 15
Age (years) [mean, SD]	59, 12
Time post-stroke (months) [median; range]	47; 7–411
Fugl-Meyer Lower Extremity motor score [*] [mean, SD]	26, 5
Berg Balance Scale score [*] [mean, SD]	51, 5
Self-selected speed (m/s) [mean, SD]	0.49, 0.19
Fastest-comfortable speed (m/s) [mean, SD]	0.70, 0.26

^{*} Fugl-Meyer and Berg Balance Scale scores are unavailable for 2 participants.

Table 2.

Results of linear mixed model analyses based on metrics derived from anteroposterior GRF. The two rows present the results of analyses focused on the indicated gait metric. For each independent variable, the parameter estimate is presented, followed by the 95% confidence interval for the parameter estimate in brackets, and the p-value for this independent variable. For the categorical Condition variable, the parameter estimate quantifies the average difference in each gait metric between the SS condition (comparison group) and the FC condition (reference group). For the remaining continuous variables, the parameter estimates quantify the linear regression coefficients. Significant contributing factors are indicated in bolded italics.

	Condition		Walking speed (m/s)	Propulsive impulse (Ns/BW)	Late braking impulse (Ns/BW)	Propulsion onset time (%GC)	Constant term
	SS	FC					
Swing leg knee flexion (degrees)	0.3 [-2.6 3.3] <i>p=0.81</i>	0	<i>15 [1 28]</i> <i>p=0.032</i>	4.2 [-0.7 9.2] p=0.092	<i>-14.9 [-29.0 - 0.8]</i> <i>p=0.039</i>	-0.02 [-0.33 0.28] p=0.87	36 [26 47]
Swing leg circumduction (mm)	<i>-7.6 [-12.8 -2.1]</i> <i>p=0.009</i>	0	-14 [-37 10] p=0.25	<i>-9.9 [-18.7 -1.1]</i> <i>p=0.029</i>	<i>25.9 [3.9 47.9]</i> <i>p=0.023</i>	-0.35 [-0.92 0.22] p=0.21	39 [22 57]

Table 3.

Results of linear mixed model analyses based on metrics derived from ILM-method power, following the same structure as Table 2 above.

	Condition		Walking speed (m/s)	Trailing leg positive work (J/kg)	Trailing leg negative work (J/kg)	Positive power onset time (%GC)	Constant term
	SS	FC					
Swing leg knee flexion (degrees)	0.5 [-2.5 3.5] $p=0.74$	0	16 [3 29] $p=0.016$	37 [-45 119] $p=0.36$	39 [-65 143] $p=0.45$	0.0 [-0.8 0.8] $p=0.99$	33 [21 45]
Swing leg circumduction (mm)	-10.5 [-16.1 -4.8] $p=0.001$	0	-29 [-52 -6] $p=0.015$	-123 [-286 41] $p=0.14$	-37 [-244 171] $p=0.72$	1.2 [-0.5 2.8] $p=0.16$	52 [32 72]