Title: Altered biomechanical strategies of the paretic hip and knee joints during a step-up task: a pilot study

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Abstract

Stroke often leads to chronic motor impairments in the paretic lower limb that can constrain lower extremity movement and negatively impact the ability to navigate stairs or curbs. This exploratory study investigated the differences in hip and knee biomechanical strategies during a step-up task between 5 adults with hemiparetic stroke and 5 age-matched adults without stroke. Participants were instructed to step up onto a 4-inch platform, where joint biomechanics were quantified for the hip in the frontal plane and the hip and knee in the sagittal plane. Peak joint kinematics were identified during the leading limb swing phase and peak joint moments and power were identified during the leading limb pull-up phase of stance. Mixed effects regression models estimated fixed effects of limb (3 levels: control dominant, stroke non-paretic, and stroke paretic) on biomechanical outcomes, while a random effect of participant controlled for within-participant correlations. Repeated assessments within participants (approximately 60 trials per lower limb) increased the effective sample size from 10 to between 12.0 to 19.6. Altered biomechanical strategies of the paretic lower limb included reduced flexion angles and increased pelvic obliquity angles during swing, decreased power generation in the hip frontal plane during stance, and decreased moment and power generation in the knee sagittal plane during stance. A strategy of substantial interest was the elevated hip sagittal plane moment and power generation in both stroke limbs. Overall, our findings suggest that chronic motor impairments from stroke can lead to inefficient biomechanical strategies when stepping up.

Introduction

Stroke is the leading cause of disability in the United States (Virani et al., 2020) and often leads to chronic, asymmetrical lower limb motor impairments that severely limit independent community ambulation. Biomechanical deficits include hip and knee weakness and reduced ranges of motion in the paretic (more affected) lower limb (Gracies, 2005; Sánchez et al., 2017). These impairments can constrain lower extremity movement to maladaptive strategies that are mechanically inefficient. Indeed, individuals with stroke increasingly depend on the non-paretic (less affected) limb for balance and stance loading (Bohannon and Larkin, 1985; Chen et al., 2005; Mansfield et al., 2012) and frequently hike the paretic hip to achieve proper foot clearance during level-ground walking (Li et al., 2018). However, steady-state walking only constitutes a small part of community ambulation. Maximizing daily occupational, social, and physical activities requires negotiating uneven terrain such as stairs and curbs (Alzahrani et al., 2009; Weerdesteyn et al., 2008).

Compared to level-ground walking, stepping up onto a stair or curb requires generation of (1) substantial hip and knee flexion angles during swing to achieve proper toe clearance and (2) considerable knee extension moments and power during the pull-up phase of stance to elevate the body's center of mass (Costigan et al., 2002; Nadeau et al., 2003; Riener et al., 2002). For older adults without a stroke, researchers investigating stair ascent have reported comparable hip and knee moments and power generation in the sagittal plane compared to their younger peers (Novak et al., 2011; Novak and Brouwer, 2011). However, older adults exhibited increased peak hip abduction moments during pull-up (Novak and Brouwer, 2011). This increase in the hip frontal plane

is thought to be a strategy for older adults to maximize stability and mitigate risk of falling (Novak and Brouwer, 2011). Weakness in the paretic limb hip abductors following stroke may challenge use of this strategy (Sánchez et al., 2017). Novak et al. reported similar kinematic profiles but decreases in peak hip abduction and knee extension moments in the paretic limb compared to non-paretic and control limbs during pull-up phase of stance (Novak and Brouwer, 2013). Nevertheless, this analysis did not consider the substantial hip and knee angles required to swing the limb onto the step or the power generation required to pull the body up onto the step. Characterization of these biomechanical strategies during the first step is of high clinical relevance. Hip abduction moments are higher during the first step compared to the second step (Wang and Gillette, 2018), making the first step likely more difficult post-stroke.

The purpose of this study was to quantify the kinematics and kinetics of the leading hip and knee during a step-up task in individuals with and without stroke. We focused on frontal plane metrics in the hip and sagittal plane metrics in the hip and knee to characterize the joints with the greatest contributions to swinging and pulling up onto a step (McFadyen and Winter, 1988; Novak and Brouwer, 2011; Riener et al., 2002). Our main kinematic hypothesis was that the paretic limb would have decreased peak hip abduction, hip flexion, and knee flexion angles but increased peak pelvic obliquity angles while swinging onto a step compared to the non-paretic and control limbs. Our main kinetics hypotheses were: 1) the paretic limb would have decreased peak hip abduction, hip extension, and knee extension moments while pulling up onto a step compared to the non-paretic and control limbs, and 2) the paretic limb would have decreased peak hip

frontal plane, hip sagittal plane, and knee sagittal plane power generation in the same stance phase compared to the non-paretic and control limbs.

Methods

I. Participant summary

A total of 10 participants were included in this study: 5 adults with hemiparetic stroke (4 with right hemiparesis, 2 females, age = 59.0 ± 7.6) and 5 age-matched adults without stroke (3 females, age = 60.4 ± 5.0). Adults aged 21-65 years who could independently walk up to 1000 ft and step up and down a 4-inch platform were eligible for this study. Furthermore, participants with stroke were included if they had a clinical presentation of hemiparesis from a diagnosed unilateral stroke at least one year prior to study participation. Exclusion criteria were 1) orthopedic surgeries or fractures in the trunk, pelvis, or lower limb joints, 2) current use of botulinum toxin injections or other muscle relaxants in the lower extremity, 3) inability to stand or take a step independently, and 4) health risks from comorbidities. This study was approved by Northwestern University's Institutional Review Board, and all participants provided informed consent.

II. Experimental Set-Up

Participants performed a series of stair-stepping trials in a gait laboratory. Two custom-machined 4-inch wooden steps were designed to cover the entire surface area of two individual force plates. A 4-inch step height was selected as it proved sufficiently challenging and achievable for all participants with stroke.

A 12-camera motion capture system (Qualisys, Gothenburg, Sweden) and four in-ground force plates (AMTI, Glenview, IL) were used to record the 3D positions of retro-reflective markers and ground reaction forces, respectively. Kinematic data was recorded

at 60 Hz and ground reaction force data was recorded at 1200 Hz. Markers were placed bilaterally on anatomical landmarks of the pelvis, thigh, shank, and foot, including: anterior superior iliac spines, posterior superior iliac spines, greater trochanters, lateral femoral epicondyles, lateral malleoli, calcanei, and the 2nd and 5th metatarsals. A single marker was placed on the sternum and 4-marker tracking clusters were affixed bilaterally to the thigh and shank.

III. Experimental Procedure

Prior to the stair-stepping trials, a licensed physical therapist administered clinical assessments to evaluate walking and balance function. Clinical measures included the Lower Extremity (LE) Fugl-Meyer (Crow and Harmeling-Van Der Wel, 2008), the Activities-specific Balance Confidence (ABC) Scale (Botner et al., 2005), and the Step Test (Hong et al., 2012). Lower limb dominance was self-reported for the control group and established as the non-paretic limb for the stroke group. Vital signs were taken at prescribed intervals (pre-, during, post-experiment) to ensure participants had an appropriate physiological response to physical activity.

To minimize the risk of falling, participants were secured with a harness (Aretech, Ashburn, VA) attached to a passive overhead trolley system that allowed for unrestricted anteroposterior movement. Participants with stroke were asked not to use assistive devices such as canes or walkers during the experiment but used their ankle foot orthoses (AFOs) if typically worn for community ambulation. Participants started each trial with their feet on two individual force plates posterior to the force plates with the wooden steps. Prior to data collection, participants first practiced stepping up, alternating their leading leg. After becoming comfortable with the task, participants were instructed to take a full

step up as they normally would while avoiding the use of their hands for assistance. Each participant completed 12 trials per leading lower limb, for approximately 60 trials per leading lower limb (control dominant, control non-dominant, stroke non-paretic, stroke paretic).

IV. Data Processing & Analysis

Data were visually inspected in Qualisys Track Manager to confirm that markers were labeled appropriately and then processed in Visual 3D (C-Motion, Germantown, MD) to calculate joint kinetics and kinematics using inverse dynamics. To account for the step height during inverse dynamic calculations, two virtual force platforms were created in the software. Marker and force data were filtered with a 4th order Butterworth filter with a cut-off of 6Hz to remove high-frequency fluctuations in the data. Stepping events for lift-off and initial contact were determined by a 5 N threshold on each force plate, and these events were then visually inspected to confirm accuracy.

All data were exported into MATLAB (MathWorks, Inc., Natick, MA) for further processing. Kinematic data were extracted when the leading lower limb was swinging onto the step, between leading foot lift-off and initial contact. This window was normalized to 100% of the swing phase and peak angles in hip abduction (HABDa), hip flexion (HFLEX), knee flexion (KFLEX), and pelvic obliquity were considered for statistical analysis (Goyal et al., 2021). Kinetic data were normalized to body weight and extracted when the leading lower limb was in stance pulling the body up onto the step, between leading foot initial contact and trailing foot initial contact. Kinematic data were also extracted for this time period to calculate angular velocity and to generate power profiles. This window was normalized to 100% of the stance phase. Peak moments in hip

abduction (HABD), hip extension (HEXT), and knee extension (KEXT) and peak power in the hip frontal plane (H1), hip sagittal plane (H2), and knee sagittal plane (K2) were considered for statistical analysis (Goyal et al., 2021).

V. Statistical Analysis

All statistical analyses were carried out in Stata IC 14.1 (StataCorp LLC, College Station, TX, USA). Independent two-sample t-tests were used to compare height, weight, and ABC scores between the two groups. A paired t-test was used for the Step Test to compare differences in step count between the two lower limbs within each group. For subsequent analyses, only the dominant limb of individuals without stroke was considered because previous research has shown no significant between-limb differences in this group (Novak and Brouwer, 2011). Mixed effects regression models estimated fixed effects of limb (3 levels: control dominant, stroke non-paretic, and stroke paretic) on biomechanical outcomes, while a random effect of participant controlled for within-participant correlations. The residuals of each regression were visually inspected to confirm normal distribution. Each regression model included approximately 60 trials per lower limb, for a total of 180 trials. The *a priori* threshold for statistical significance was set to 0.05.

Results

I. Participant Metrics

All participant metrics are displayed in Table 1. There were no significant differences in weight, height, or ABC scores between the groups. The p-value for ABC scores was close to the significance threshold ($p = 0.063$) and standard deviation for the stroke group was much larger than the control group (Table 1). For the Step Test, the

non-paretic lower limb had a significantly greater step count than the paretic lower limb ($p = 0.002$), but there were no significant differences between the lower limbs for the control group. The average LE Fugl-Meyer score for the stroke group was just under 21 points (Table 1), indicating a moderate-to-high level of functional mobility (Kwong and Ng, 2019). Given the observed within-participant correlations of 0.460 to 0.826 in the regression models, the effective sample size ranged from 12.0 to 19.6, indicating increased statistical power to detect significant effects.

Table 1. Mean (SD) participant-specific metrics and clinical assessment outcomes.

Outcome Measure	Group	
	Stroke (n = 5)	Control (n = 5)
Weight (kg)	68.6 (12.2)	66.9 (12.9)
Height (in)	66.7 (3.5)	64.7 (2.6)
ABC Score	84.3 (13.9)	98.0 (3.3)
LE Fugl-Meyer	20.4 (5.0)	--
Step Test		
Non-Paretic/Dominant	11.5 (4.7)	18.3 (3.7)
Paretic/Non-Dominant	8.1 (4.1)	18.0 (3.7)

II. Joint Biomechanics

All kinematic and kinetic data are summarized in Table 2. Joint kinematic curves during the leading limb swing phase are displayed in Figure 1. The main effect of limb was not significant for HABDa angles. There was a significant main effect of limb on HFLEX, KFLEX, and Pelvic Obliquity angles. The non-paretic limb exhibited higher peak hip flexion angles compared to the paretic limb ($p = 0.012$). For KFLEX and Pelvic

Obliquity, both the dominant and non-paretic limbs exhibited higher peak angles compared to the paretic limb ($p < 0.001$ for all comparisons).

Table 2. Mean (SD) peak joint (1) angles in hip abduction, hip flexion, knee flexion, and pelvic obliquity (2) moments in hip abduction, hip extension, and knee extension and (3) powers in the hip frontal plane, hip sagittal plane, and knee sagittal plane, all for each leading lower limb when swinging and pulling up onto a step.

Phase	Metric	Group		
		Stroke	Control	
Swing	Joint Angles (Degrees)	Paretic	Non-Paretic	Dominant
	Hip Abduction (HABDa)	3.65 (2.86)	8.92 (5.97)	4.04 (4.34)
	Hip Flexion (HFLEX)	45.4 (4.88) ⁺	57.5 (8.86)	53.8 (14.5)
	Knee Flexion (KFLEX)	49.5 (10.8) ⁺⁺	82.2 (8.48)	82.4 (5.31)
	Pelvic Obliquity (to ipsilateral side)	11.9 (2.08) ⁺⁺	4.13 (3.18)	4.44 (2.31)
Pull-up Stance	Joint Moment (Nm/kg)	Paretic	Non-Paretic	Dominant
	Hip Abduction (HABD)	0.734 (0.192)	0.887 (0.094)	0.852 (0.152)
	Hip Extension (HEXT)	-0.700 (0.236) [*]	-0.672 (0.155) [*]	-0.321 (0.176)
	Knee Extension (KEXT)	-0.654 (0.168) ⁺⁺	-1.49 (0.224) [*]	-0.851 (0.171)
	Joint Power (W/kg)	Paretic	Non-Paretic	Dominant
	Hip Frontal Plane (H1)	0.143 (0.077) ⁺⁺	0.285 (0.141)	0.279 (0.151)
	Hip Sagittal Plane (H2)	0.761 (0.198) [*]	0.848 (0.418) [*]	0.426 (0.250)
	Knee Sagittal Plane (K2)	0.847 (0.248) ⁺⁺	2.40 (0.603) [*]	1.37 (0.371)

^{*} $p < 0.05$ for a significantly different from dominant limb

⁺ $p < 0.05$ for a significantly different from non-paretic limb

Figure 1. Joint kinematic profiles for the leading leg swing phase. The x-axis of each subplot is time normalized to 100% of the swing phase and the y-axis is joint angles in degrees. Shaded regions represent one standard deviation. Red line = stroke paretic limb, blue line = stroke non-paretic limb, black line = control dominant limb.

Joint kinetic profiles during the leading limb pull-up stance phase are displayed in Figure 2. The main effect of limb was not significant for HADD moment. There was a significant main effect of limb on HEXT and KEXT. Peak hip extension moments were lower in the dominant limb compared to the non-paretic limb ($p < 0.001$) and the paretic limb ($p = 0.003$). For KEXT, the non-dominant limb generated higher peak knee extension moments compared to the dominant and paretic limbs ($p < 0.001$ for both comparisons), while the dominant limb also generated higher peak KEXT moments compared to the paretic limb ($p = 0.022$).

Joint power results are similar to joint moment results, with some deviations. There was a significant main effect of limb on H1, where the paretic limb generated lower peak power in the hip frontal plane compared to the non-paretic limb ($p = 0.012$) and dominant limb ($p = 0.048$). There was also a significant main effect of limb on H2 and K2. The dominant limb generated lower peak hip sagittal plane power compared to the non-paretic limb ($p = 0.038$) and paretic limb ($p = 0.004$). For K2, the non-paretic limb generated higher peak knee sagittal plane power compared to the dominant and paretic limbs ($p < 0.001$ for both comparisons), while the dominant limb also generated higher peak K2 power compared to the paretic limb ($p = 0.003$).

Figure 3. Joint moment and power profiles for the leading limb pull-up phase of stance. The x-axis of each subplot is time normalized to 100% of the stance phase and the y-axis is joint moment or power normalized to body weight. Positive joint moment indicates abduction/flexion direction. Positive joint power indicates concentric contraction, where energy is created. Shaded regions represent one standard deviation. Red line = stroke paretic limb, blue line = stroke non-paretic limb, black line = control dominant limb.

Discussion

This pilot study investigated leading hip and knee biomechanical strategies during the first step of stair ascent in adults with and without chronic stroke. The hip and knee joints are typically the largest contributors to kinematics during swing and kinetics during the pull-up phase of stair ascent, particularly in the frontal and sagittal planes (McFadyen and Winter, 1988; Novak and Brouwer, 2011; Riener et al., 2002). The first step in particular is of great interest because hip abduction moments are the highest when transitioning from standing to stepping (Wang and Gillette, 2018). The main finding was that the paretic lower limb exhibited kinematic differences during swing and joint moment and power differences during pull-up stance compared to the non-paretic and control lower limbs, especially in the knee joint.

The leading paretic limb had reduced hip and knee flexion angles when swinging onto the step, confirming our first hypothesis. Significant isometric strength imbalances have been quantified between the non-paretic and paretic limbs in these joint directions (Hayes Cruz and Dhaher, 2007; Sánchez et al., 2017). In addition, knee joint stiffness in the paretic limb is often responsible for reduced knee flexion during swing. This stiffness has been attributed to decreases in amplitude and frequency of biceps femoris and gastrocnemius muscles (Wang et al., 2017) and overactive muscle reflexes of the rectus femoris muscle (Akbas et al., 2020). Similarities in hip abduction angles could be due to the fact that these joint angles are relatively small compared to sagittal plane flexion angles. We also found that hip hiking, in the form of pelvic obliquity, was exaggerated in the paretic limb during the step-up task. This compensation is common among adults with stroke during level-ground walking to achieve proper foot clearance (Li et al., 2018) and

may be especially important when stepping up given the significantly low knee flexion angles of the paretic limb.

The stroke limbs also had asymmetrical moment and power profiles during the pull-up phase of stance, indicating that they were overall less mechanically and energetically efficient in stepping up. Sánchez et al. quantified hip abduction and knee extension isometric weakness in the paretic limb compared to the non-paretic and control limbs (Sánchez et al., 2017). Though there were no significant differences in hip abduction moments during the task, the paretic limb did generate significantly less power in the hip frontal plane. These findings suggest that while hip abductor muscle force generation in the paretic limb is adequate, their concentric activity is not as efficient in producing the needed movement to stabilize the body when pulling up onto a step. We did confirm part of our joint moment hypothesis by quantifying significant decreases in knee extension moments when pulling up onto a step, similar to what has been previously reported (Novak and Brouwer, 2013). The paretic limb also exhibited decreased power generation in the knee sagittal plane, especially compared to the non-paretic limb. Analysis of energy transfer in individuals with stroke during level-ground walking revealed diminished knee power profiles in both the paretic and non-paretic limbs during stance (Novak et al., 2015). The more asymmetrical nature of our results could be due to higher demand on the knee joint during the pull-up phase of stance when stepping up.

Significant new findings were the increased peaks in hip extension moments and hip sagittal plane power generation in both stroke limbs compared to the control limb. Previous research has shown that it is typically the knee joint that contributes the highest relative moment and power when pulling-up onto a step (Novak and Brouwer, 2011;

Riener et al., 2002). However, the paretic limb displayed significantly diminished kinetic profiles in this joint. It is possible that participants with stroke increased dependence on their paretic limb hip extensors as an inefficient compensatory strategy. However, the non-paretic limb did not exhibit diminished moment or power profiles in the knee joint. High hip power generation in this limb during level-ground walking has been interpreted as compensation for lack of push-off power in the paretic limb (Novak et al., 2015). As push-off power is relatively similar between level-ground walking and stair ascent (Riener et al., 2002), the non-paretic limb may be using the same strategy to successfully complete the step-up task.

In summary, in this exploratory study, we quantified differences in leading hip and knee biomechanical strategies between the paretic and non-paretic/control limbs during a step-up task. Altered biomechanical strategies of the paretic lower limb included reduced flexion angles and increased pelvic obliquity angles during swing, decreased power generation in the hip frontal plane during stance, and decreased moment and power generation in the knee sagittal plane during stance. A strategy of substantial interest was the elevated hip sagittal plane moments and power generation in both stroke limbs. Further analysis of the trailing leg may illuminate the impact of the paretic leg in the push-off phase of stance. Study limitations include low sample size and allowed use of passive AFOs. These braces typically restrict ankle movement only (Daryabor et al., 2018; Totah et al., 2019), which was not considered in the present analysis. Future research should test these findings in a larger sample to confirm that the quantified biomechanical strategies persist in adults with stroke, especially in participants with lower functional mobility. Combining the step-up task with progressive limb loading may reveal

additional impairments such as abnormal coupling between the hip adductors and lower limb extensors, which has been quantified isometrically (Hayes Cruz and Dhaher, 2007; Sánchez et al., 2018). Overall, our findings suggest that chronic motor impairments from stroke can lead to inefficient biomechanical strategies when stepping up. Rehabilitation implications based on our study may include interventions that aim to increase hip and knee sagittal plane angles during the swing phase of step ascent, as well as substitution patterns to maximize power during step negotiation.

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Conflict of interest statement

There are no conflicts of interest.

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