Relationships between post-stroke mediolateral stepping characteristics, walking speed, balance performance, and balance confidence

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Word count: 4457

Number of figures and tables: 6

Abstract

Background. Many people with chronic stroke exhibit walking balance deficits, which are linked with increased fall risk and decreased balance confidence. One potential contributor to these deficits is a decreased ability to adjust where the paretic foot is placed based on pelvis motion, which is an important factor in mediolateral walking balance. Objective. The objective of this study was to investigate altered stepping behavior for steps taken with the paretic leg, and to explore the relationship between this behavior and clinical measures of balance and balance confidence. Methods. 93 individuals with chronic stroke walked on a treadmill and completed a series of common clinical tests. We used partial correlations to quantify the step-by-step relationship between mediolateral pelvis displacement and both step width (ρ_{SW}) and mediolateral foot placement relative to the pelvis (ρ_{FP}), a metric that focuses on swing leg repositioning. Results. ρ_{SW} was weaker for steps taken with the paretic leg, whereas ρ_{FP} did not differ between non-paretic and paretic steps. Participants who walked faster had larger ρ_{SW} and ρ_{FP} values, but instructing participants to walk faster did not consistently increase ρ_{SW} or ρ_{FP} for paretic steps. Significant, albeit weak, relationships were observed between paretic ρ_{SW} and Functional Gait Assessment score, as well as between paretic ρ_{FP} and Activities-Specific Balance Confidence score, fear of falling, and fall history. Conclusions. A reduced ability to adjust where the paretic foot lands when taking a step may discourage fast walking speeds after a stroke and contribute to clinical observations of decreased walking balance.

Key Words: gait disorders, neurologic; locomotion; postural balance; stroke; walking

Introduction

Walking balance deficits limit function in many people with chronic stroke (PwCS). The annual fall incidence rate among community-dwelling PwCS has been estimated as 1.4-5.0 falls/person-year, substantially higher than the 0.65 falls/person-year among older adults without neurological injuries[1]. Walking is the most common activity at the time of a fall for PwCS[1], suggesting a link between fall rate and walking imbalance. Additionally, many PwCS report a fear of falling[2] and reduced walking balance self-efficacy[3], each linked to decreased community participation and quality of life[3–9]. Likely as a result of this functional importance, balance improvements are a primary rehabilitation goal[10].

Numerous methods have been developed to assess walking balance. Commonly used clinical assessments include the Functional Gait Assessment (FGA) to evaluate balance during walking tasks with a range of difficulty[11], the Activities-specific Balance Confidence scale (ABC) to quantify balance self-efficacy during community tasks[12], and directly asking if a patient is afraid of falling[2]. While each of these measures has been linked to functional mobility[2,9,13–16] and may be used to aid in clinical decision making, they do not provide direct insight into the mechanical underpinnings of post-stroke walking imbalance. An alternative approach is to quantify aspects of post-stroke walking mechanics that may be indicative of balance deficits. For example, relative to age-matched controls, PwCS tend to walk with slower speeds, shorter and wider steps, and more variable step lengths, times, widths, and margins of stability[17–20].

The present study focuses on mediolateral walking balance, which model simulations[21] and human experiments[22] suggest is more challenging to control than

anteroposterior balance. Many post-stroke falls (~31-38%) occur sideways in both the chronic[4] and subacute[23] phases, supporting the clinical importance of mediolateral balance. Further, post-stroke falls can be prospectively predicted based on mediolateral motion of the pelvis during walking (e.g., lower mediolateral dynamic stability[24]; decreased mediolateral displacement[25]).

One strategy that plays an important role in mediolateral walking balance involves step width modulation. In neurologically intact controls, step width varies with pelvis dynamics; steps tend to be wider when the pelvis has a larger mediolateral displacement or velocity away from the stance foot[26–28]. This relationship is likely due in part to inverted pendulum dynamics of the stance leg and trunk, as larger displacements of the pelvis toward the swing leg would directly contribute to wider steps. In addition, active control of swing leg position contributes to this step width modulation, as larger mediolateral pelvis displacements or velocities are linked with greater activation of the swing leg hip abductors and more lateral foot placement[29]. This strategy of actively modulating step width and swing leg positioning based on pelvis motion is also observed when perturbations are applied during walking[29–31], suggesting its generalized use in preventing losses of balance.

Among PwCS, deficits in modulating step width could contribute to the many falls that are caused by intrinsic movement errors[6] rather than external perturbations such as slips or trips. In a small-scale study (n=20), we found that during unperturbed walking, the relationship between mediolateral pelvis displacement and step width was weakened for steps taken with the paretic leg, in comparison to steps taken with the non-paretic leg[32]. Increased walking speed did not influence this deficit, suggesting that

it was not simply an artifact of walking slowly. However, we did not investigate whether the observed deficit was due to altered swing leg control[33–35]. Additionally, this altered step width modulation has not been compared to clinical measures of walking balance or balance confidence.

The purpose of this cross-sectional study was to investigate mediolateral walking balance in PwCS, focusing on altered stepping behavior with the paretic leg and potential relationships with common clinical measures. Based on our prior results[32], our primary hypothesis was that the relationship between pelvis displacement and step width would be weaker for paretic steps than non-paretic steps, with this deficit unaffected by walking speed. Secondarily, we hypothesized that the relationship between pelvis displacement and mediolateral foot placement relative to the pelvis would also be weaker for paretic steps, reflecting a deficit in controlling the position of the paretic swing leg[33–35] rather than altered control of the stance leg. Finally, we performed exploratory analyses of whether biomechanical measures of mediolateral stepping behavior are associated with clinical measures of walking balance function and confidence.

Methods

Participants

93 PwCS participated in this study, which used data collected in a screening session for a separate clinical trial (NCT02960439). Inclusion criteria were: experience of a stroke ≥ 6 months prior; age ≥ 21 years; ability to walk independently on a treadmill. Exclusion criteria were: resting heart rate above 110 beats/min; resting blood pressure

above 200/110 mm Hg; history of congestive heart failure, unstable cardiac arrhythmias, hypertrophic cardiomyopathy, severe aortic stenosis, angina or dyspnea; preexisting neurological disorders or dementia; history of major head trauma; severe visual impairment; history of deep vein thrombosis or pulmonary embolism within 6 months; uncontrolled diabetes with recent weight loss, diabetic coma, or frequent insulin reactions; orthopedic injuries or conditions (e.g., hip replacements) with the potential to alter the ability to adjust foot placement while walking.

Summary demographic and functional characteristics are presented in Table 1. The sample size for this study was based on the recruitment needs of the clinical trial. However, based on our goal of exploring associations between biomechanical metrics and clinical measures of balance, we surpassed the 84 participants needed to detect a moderate correlation (r=0.3) with 80% power and an alpha value of 0.05. All participants provided written informed consent, signing a form approved by the Medical University of South Carolina Institutional Review Board (Pro00059390).

Experimental Procedures – Clinical Testing

We calculated self-selected overground walking speed from the time required to traverse the middle 6-meters of a 10-meter path. We quantified lower extremity Fugl-Meyer motor and sensory scores as global measures of sensorimotor function[36], FGA scores to assess walking balance, ABC scores to quantify balance self-efficacy, and asked participants “Do you have a fear of falling?” to assess fear of falling[37]. We assessed self-reported fall history by asking participants “Have you had any fall in which you lost your balance and landed on a lower level in the past year?”[38]. All clinical testing

procedures were performed by a licensed and trained physical therapist or physical therapist assistant.

Experimental Procedures – Treadmill Walking

Biomechanical metrics were calculated from data collected during treadmill walking. Participants did not hold on to a handrail, and wore a harness attached to an overhead rail that did not support body weight but prevented participants from falling to the ground in case of a loss of balance.

We first identified each participant's self-selected speed, defined as the speed used to walk around the house or a store. The treadmill started at 0.1 m/s and was increased in increments of 0.05 m/s until participants reported that the treadmill speed was faster than their self-selected speed, at which point the speed was reduced by 0.05 m/s to serve as the self-selected speed. Participants walked for 3-minutes at the self-selected speed. We then identified each participant's fastest comfortable speed, defined as the fastest speed at which the participant felt safe. Starting at the self-selected speed, the treadmill speed was increased in increments of 0.05 m/s until participants reported that they no longer felt safe or could not maintain their position on the treadmill. Participants walked for 3-minutes at the fastest comfortable speed. A maximum speed of 1.2 m/s was allowed for this condition, a speed often considered sufficient for safe pedestrian travel[39]. During both 3-minute walking trials, participants were permitted to take breaks if they reported being unable to continue or were unable to maintain their position on the treadmill. A maximum of three trials was required to complete the 3-minute walking periods.

Data Collection and Step Identification

Active LED markers (PhaseSpace; San Leandro, CA) were placed on participants, with present analyses focused on markers placed on the sacrum and bilateral heels. Marker position was measured at 120 Hz. Generally, steps for a given leg were defined to start when the foot's velocity changed from posterior to anterior and to end when the foot's velocity changed from anterior to posterior. However, initial analyses revealed unusual foot velocity patterns in a subset of participants, motivating the use of a more complex method to avoid the detection of false gait events. This method is described in the Supplementary Material. Mediolateral pelvis dynamics were calculated based on the average of four markers spaced around the boundaries of a rigid cluster placed over the sacrum.

Calculation of Biomechanical Metrics

All metrics were calculated separately for paretic and non-paretic steps. Following prior methods[32,40,41], throughout each step we calculated the mediolateral displacement of the pelvis (relative to the stance heel at the start of the step) and the mediolateral velocity of the pelvis, with the positive direction defined as toward the swing leg. Pelvis displacement and velocity trajectories within each step were resampled to create 101-sample vectors, allowing us to perform analyses across these time-normalized steps. For each step, step width was calculated as the mediolateral displacement between the stance heel at the start of a step and the swing heel at the end of the step. The effect of swing leg repositioning on step width was quantified as the mediolateral displacement between the pelvis and swing heel at the end of the step (termed *foot placement*). The effect of stance leg repositioning was quantified as the

mediolateral displacement between the pelvis at the end of the step and the stance heel at the start of the step (termed *final pelvis displacement*). These metrics summed to equal step width.

To quantify the relationship between pelvis displacement and step width, we calculated the partial correlation between mediolateral pelvis displacement at each normalized time point during a step and the step width at the end of a step, accounting for mediolateral pelvis velocity (ρ_{SW}). Our focus on pelvis displacement is justified by our prior finding of no paretic side deficits in the contribution of pelvis velocity to step width[32]. Our use of partial correlations instead of R^2 magnitudes is due to our previous observation of negative partial correlation values in some participants, which artificially inflate R^2 values[32]. To assess potential paretic side deficits in the swing leg contributions to step width, we quantified the partial correlation between mediolateral pelvis displacement and foot placement (ρ_{FP}) following the same methods as above. Finally, to assess potential deficits in stance leg contributions to step width, we quantified the partial correlation between mediolateral pelvis displacement throughout the step and the final pelvis displacement (ρ_{PD}).

Statistical Analyses

We first investigated the relationship between pelvis displacement and step width when walking at a self-selected speed. We used one-dimensional statistical parametric mapping[42] to assess potential differences between paretic and non-paretic step ρ_{SW} throughout the step, a method that identifies the presence and timing of significant differences in these trajectories. We subsequently focused our analyses on ρ_{SW} values

at two time points previously found to differ between paretic and non-paretic steps[32]: 1) start of the step, providing insight into adjustments in body configuration occurring within a step; 2) end of the step, providing insight into the ultimate effect of any within-step adjustments. For both step start ρ_{SW} and step end ρ_{SW}, we performed ANCOVA to assess potential differences between paretic and non-paretic steps, and interparticipant effects of self-selected walking speed on these metrics.

While the ANCOVA provided insight into interparticipant effects of walking speed, we additionally investigated within-participant effects of speed on four metrics: non-paretic step start ρ_{SW}; non-paretic step end ρ_{SW}; paretic step start ρ_{SW}; paretic step end ρ_{SW}. For each of these metrics, we used one-way repeated measures ANOVA to test for significant differences between three conditions: 1) experimental value at self-selected speed; 2) experimental value at fastest-comfortable speed; 3) predicted value at the fastest-comfortable speed. This predicted value was calculated based on each participant's increase in speed for the fastest-comfortable condition, and the best-fit slope between walking speed and metric value from the previously performed ANCOVA. Essentially, we calculated the metric value that would be observed if each participant followed the interparticipant relationship when increasing speed. In the event of a significant main effect, we performed Tukey-Kramer post-hoc tests to identify differences between conditions. These analyses did not include one participant who reported being too fatigued to perform the fastest-comfortable trial or eight participants who reached the maximum allowed speed of 1.2 m/s, as they may not have been truly walking at their fastest-comfortable speed. These comparisons thus included 84 participants.

Moving beyond the focus on step width, we subsequently investigated the effects of stepping leg and walking speed on step start ρ_{FP}, step end ρ_{FP}, and step start ρ_{PD}. We did not include step end ρ_{PD}, as this metric trivially converges to one at the end of the step. For these analyses, we performed ANCOVA and repeated measures ANOVA following the previously described structure.

Finally, we explored potential relationships between our biomechanical metrics when walking at self-selected speed and several measures of clinical interest – FGA score, ABC score, fear of falling, and fall history. For FGA and ABC scores, we performed Spearman correlations with each of our biomechanical metrics. If the clinical measure of interest was significantly ($p<0.05$) correlated with self-selected walking speed, we corrected for this relationship by subtracting the best-fit linear relationship and re-performing the Spearman correlation for any biomechanical metrics that exhibited a significant positive relationship with the clinical measure. The goal of this step was to determine whether our biomechanical metrics can provide insight beyond walking speed. For fear of falling and fall history, we performed logistic regressions with each of our biomechanical metrics. In the case of a significant effect ($p<0.05$) of self-selected speed, we re-performed these regressions including speed as an independent variable for any biomechanical metrics that made a significant positive contribution to predicting the clinical measure.

For all comparisons, an alpha value of 0.05 was considered significant. Given the exploratory nature of many of our analyses, we did not correct for multiple comparisons.

Results

Relationship between Pelvis Displacement and Step Width (ρ_{SW})

The partial correlation trajectory between mediolateral pelvis displacement and step width was higher for steps taken with the non-paretic leg than with the paretic leg (Fig. 1A). Statistical parametric mapping analysis revealed a significant set level difference between legs ($p=0.0002$), with significantly different clusters from 0-42% ($p=0.028$) and 53-100% of the step ($p=0.025$) (Fig. 1B).

Effects of Speed on the Relationship between Pelvis Displacement and Step Width

Across participants, step start ρ_{SW} and step end ρ_{SW} were greater for non-paretic steps than paretic steps, and were greater for participants who walked faster. Specifically, both stepping leg ($p=0.0006$) and speed ($p<0.0001$) had a significant main effect on step start ρ_{SW} (Fig. 2A). While the differences between non-paretic and paretic values appeared smaller for participants who walked faster, no significant interaction was detected ($p=0.15$). Stepping leg ($p=0.008$) and speed ($p<0.0001$) also had a significant main effect on step end ρ_{SW} (Fig. 2B). However, these main effects were moderated by a significant interaction ($p=0.042$), as no evidence for paretic step deficits was observed for the small number of participants who walked faster than ~0.6 m/s.

The within-participant effects of increasing walking speed from self-selected to fastest-comfortable (by 0.23±0.14 m/s; mean±s.d.) were less clear. Non-paretic step start ρ_{SW} varied significantly ($p<0.0001$) across conditions, increasing for the fastest-comfortable condition to approximately the value predicted from the interparticipant relationship (Fig. 2C). Non-paretic step end ρ_{SW} also varied significantly ($p<0.0001$) across conditions, increasing to a higher level than predicted from the interparticipant

relationship when participants walked faster (Fig 2D). In contrast, paretic step start ρ_{SW} did not increase significantly when participants walked faster, with a significant effect across conditions ($p<0.0001$) due to these values remaining lower than predicted from the interparticipant relationship (Fig. 2E). Finally, paretic step end ρ_{SW} varied significantly across conditions ($p<0.0001$), increasing when participants walked faster, although not up to the level predicted from the interparticipant relationship (Fig. 2F).

Swing Leg Contributions to Step Width Modulation (ρ_{FP})

Unlike with ρ_{SW}, no significant main effect of stepping leg ($p=0.064$) on step start ρ_{FP} was detected (Fig. 3A). Step start ρ_{FP} was significantly greater ($p<0.0001$) for participants who walked faster, with no significant interaction between stepping leg and speed ($p=0.94$). Similarly, step end ρ_{FP} was not significantly influenced by stepping leg ($p=0.065$), was greater for faster walkers ($p<0.0001$), and was not influenced by an interaction between stepping leg and speed ($p=0.54$) (Fig. 3B).

The effects of increasing within-participant walking speed on ρ_{FP} were similar to those observed with ρ_{SW}. Both non-paretic step start ρ_{FP} ($p<0.0001$; Fig. 3C) and non-paretic step end ρ_{FP} ($p<0.0001$; Fig. 3D) increased with speed, and approximately matched interparticipant predictions. Both paretic step start ρ_{FP} ($p<0.0001$; Fig. 3E) and paretic step end ρ_{FP} ($p<0.0001$; Fig. 3F) increased with speed but did not reach the level predicted from the interparticipant relationship.

Stance Leg Contributions to Step Width Modulation (ρ_{PD})

Stepping leg had a significant main effect ($p<0.0001$) on step start ρ_{PD}, while walking speed did not ($p=0.27$). The main effect of stepping leg was moderated by a significant interaction ($p=0.018$), as the differences between legs were only apparent for participants who walked slowly (Fig. 4A).

Within-subject effects of walking speed on ρ_{PD} were minimal. No significant effect ($p=0.22$) was observed across conditions for non-paretic step start ρ_{PD} (Fig. 4B). A significant effect ($p=0.002$) was detected across conditions for paretic step start ρ_{PD}, as this metric did not increase to the level expected from interparticipant results (Fig. 4C).

Relationships between Biomechanical Metrics and Clinical Measures

Many biomechanical metrics of mediolateral stepping behavior were related to clinical measures of interest, albeit only moderately or weakly (Table 2). Most biomechanical metrics (including all paretic step metrics) were positively correlated with FGA score, although only paretic step start ρ_{SW} was still significantly correlated with FGA score after correcting for walking speed. Only paretic step start ρ_{FP} was significantly correlated with ABC score. While several gait metrics made a significant contribution to predicting fear of falling, only paretic step start ρ_{FP} survived the correction for walking speed. Finally, only paretic step start ρ_{FP} made a significant contribution to predicting a participant's self-reported fall history.

Discussion

The relationships between stepping leg, walking speed, and the modulation of mediolateral stepping were more complex than expected, only partially supporting our hypotheses. The relationship between pelvis displacement and step width was weaker

for steps taken with the paretic leg, as expected. However, walking speed influenced this relationship in ways that differed for interparticipant and within-participant analyses, contradicting our hypothesis. Additionally, we did not observe the expected weaker relationship between pelvis displacement and mediolateral foot placement for steps taken with the paretic leg, but instead observed asymmetries in the stance leg contributions to step width modulation. Our exploratory analyses of relationships between our biomechanical metrics and clinical measures revealed preliminary evidence for a role of modulating paretic step width and foot placement in walking balance performance and confidence.

The partial correlation between pelvis displacement and step width was weaker for paretic steps than non-paretic steps. These paretic-side deficits were present both early and late in a step, supporting our pre-planned investigation of step start ρ_{SW} and step end ρ_{SW}. The magnitudes of the observed deficits (0.09 and 0.07, respectively) were quite similar to those reported in our earlier small-scale study (0.09 for both metrics)[32] despite the slight differences in step definition described in the Supplementary Material, further indicating that the dynamics-dependent control of step width differs for paretic and non-paretic steps.

However, the expected paretic-side deficits in the swing leg contribution to step width modulation were not observed. Instead, ρ_{FP} values were generally (albeit not significantly) more positive for paretic steps than non-paretic steps, contradicting our expectation that a reduced ability to accurately control paretic swing leg position[33,35] would preferentially weaken paretic ρ_{FP}. Perhaps this unexpected result was due to our inclusion of PwCS with a broad range of function, as clear alterations in the active

control of paretic swing leg motion were previously only observed for PwCS with clinically-identified walking balance deficits[34]. Alternatively, perhaps the control of foot placement was disrupted for both paretic and non-paretic steps. Bilaterally, the partial correlation between pelvis displacement and foot placement was negative for many participants who walked slowly, indicating that the swing foot was placed more medially for steps in which the pelvis was displaced farther from the stance foot. This behavior contrasts with the expected positive relationship between pelvis displacement and foot placement and suggests a bilateral alteration in the underlying balance control strategy.

In contrast to the equivocal foot placement results, paretic and non-paretic steps differed in the stance leg contribution to step width modulation. Step start ρ_{PD} was lower for steps taken with the paretic leg, indicating asymmetry in the control of stance leg motion. The correlation between mediolateral pelvis displacement early and late in a step is likely due in part to passive inverted pendulum dynamics, as with anteroposterior motion in model simulations[43]. Speculatively, perhaps PwCS are less willing to allow the stance leg to act as an inverted pendulum during paretic steps, instead actively braking pelvic motion toward the paretic side to reduce the subsequent mechanical demands upon paretic foot contact[44]. Future work could investigate this possibility by using electromyography to quantify the activity of muscles with the potential to control mediolateral motion of the stance leg, an approach previously used to investigate the control of swing leg motion[29,34].

The relationships between walking speed and stepping behavior differed between interparticipant and within-participant comparisons. From our interparticipant comparisons, participants who walked faster tended to have larger values of step start

and step end ρ_{SW} (paretic and non-paretic steps), step start and step end ρ_{FP} (paretic and non-paretic steps), and step start ρ_{PD} (only paretic steps). From our within-participant comparisons, increases in walking speed were accompanied by changes in the non-paretic step metrics that either matched or exceeded the change expected from the interparticipant relationships. In contrast, the paretic step metrics increased less than expected from the interparticipant relationships. This type of distinction between interparticipant and within-participant effects has previously been seen for step length asymmetry among PwCS[45], indicating caution in over-interpreting interparticipant results. Here, we suggest that individuals able to achieve stronger correlations between pelvis displacement and step width / foot placement / final pelvis displacement are more willing to self-select faster speeds (interparticipant results). However, asking some individuals to walk faster may challenge their ability to increase the modulation of paretic step width contributors that is warranted at the faster speed, producing the smaller than expected metric increases (within-participant results).

The modulation of paretic step width and mediolateral foot placement was related to clinical measures of walking balance function and confidence. Most notably, a stronger relationship between step start pelvis displacement and paretic step width was significantly correlated with better FGA scores, beyond the effect of walking speed. A stronger positive relationship between step start pelvis displacement and paretic mediolateral foot placement was significantly linked with better ABC scores, a lower probability of self-reported falls, and a lower probability of fear of falling beyond the effect of walking speed. No metrics of non-paretic stepping behavior or the stance leg contribution to step width modulation exhibited similar relationships with our clinical

measures. In combination, these results suggest that control of where the swing foot lands is related to walking performance and self-efficacy in PwCS, both of which can influence real-world behavior[14]. However, the results of these exploratory analyses should be interpreted with caution, as many comparisons were performed, and the revealed relationships were only weak to moderate. Replication of these findings is needed to increase confidence, although it is perhaps not surprising that mechanical measures of mediolateral stepping behavior were only weakly linked to clinical performance. The ability to adjust where the foot lands is only one component of the tasks assessed in the FGA, and many factors other than foot placement surely contribute to balance confidence and the likelihood of real-world falls.

A major limitation of this work was our focus on a single biomechanical strategy for ensuring mediolateral walking balance – modulating the body's configuration at the end of a step based on pelvis dynamics. While this strategy allows large adjustments in the base of support, it is unable to influence body dynamics *within* a step. A faster alternative is a "center of pressure (CoP) shift strategy", in which ankle musculature activation can shift the CoP location medially or laterally within the boundaries of the stance foot[46–49]. A "push-off strategy" in which the ankle musculature modulates trail leg propulsion[48], may also help ensure controlled mediolateral weight shift toward the new stance leg. These balance strategies are interdependent[50], with their relative importance depending on context and population. For example, a foot placement strategy appears to be of reduced importance when young adults walk at slow speeds (0.4 m/s)[51], whereas older adults walking at relatively fast speeds (1.1 m/s) exhibit an increased

reliance on a foot placement strategy and less reliance on the CoP shift strategy[52]. Future work should investigate the use of these strategies in PwCS.

Other limitations should be considered when interpreting the present results. Our focus was on unperturbed walking, which may not predict the ability to recover from perturbations in populations at a risk for falls[53]. Also, our results were derived from treadmill walking rather than overground walking. While this approach was necessary to feasibly measure many steps without risking participant fatigue, walking patterns can differ somewhat between these two contexts[54], most notably in the form of substantially lower speeds. Finally, falls were self-reported and retrospective, a method that can yield unreliable results[55]. Despite these limitations, the information gained from this type of experiment has value. Gait characteristics measured during steady-state treadmill walking have recently been found to be more promising predictors of prospective falls in PwCS than either clinical assessments or perturbation responses[24,56]. The lack of clear relationships between the response to mediolateral perturbations and clinical assessments of function[57] may be partially attributable to post-stroke balance deficits being largely due to internal movement errors[6], and not simply altered reactions to external forces.

Conclusions

The relationship between mediolateral pelvis displacement and step width is weaker for paretic steps than non-paretic steps. This paretic side deficit is observed in the stance leg contribution to modulations in step width (pelvis displacement), but not the swing leg contribution (foot placement). Despite this lack of a group-wide deficit in the swing leg contribution, the importance of this biomechanical behavior is indicated by

individuals' reduced ability to strengthen paretic foot placement modulation when asked to walk faster and the observed correlations with clinical measures of balance performance and confidence. The control of both stance leg and swing leg contributions to mediolateral gait motion may provide unique insight into walking balance deficits after a stroke.

Acknowledgements

The views expressed by the authors are their own and do not necessarily reflect the official policy of the VA or the U.S. Government.

Funding

The authors disclosed receipt of the following financial support for the research, authorship, and/or publication of this article: This work was supported by the Department of Veterans Affairs [grant number I01 RX002256) and the National Institutes of Health [grant number P20 GM109040].

Declaration of Conflicting Interests

The Authors declare that there is no conflict of interest.

Tables

Characteristic	Value
Gender	42 F / 51 M
Paretic side	39 L / 54 R
Age (years)	60 [23 85]
Height (cm)	170 [150 196]
Mass (kg)	81 [50 143]
Time since stroke (months)	31 [6 242]
Fugl Meyer motor score	24 [14 34]
Fugl Meyer sensory score	11 [2 12]
Self-selected overground walking speed (m/s)	0.79 [0.11 1.39]
Self-selected treadmill walking speed (m/s)	0.30 [0.10 1.20]

Table 1. Summary participant information. For numerical measures, we present the group median and the group range in brackets.

Biomechanical metric	FGA	Speed-corrected FGA	ABC	Fear of falling (FOF)	Speed-corrected FOF	Self-reported falls
Self-selected speed	**0.67 (<0.0001)**	-	0.20 (0.058)	**0.50 (0.019)**	-	0.25 (0.25)
Non-paretic step start ρ_{SW}	0.11 (0.30)	-	0.10 (0.35)	0.14 (0.64)	-	0.20 (0.48)
Non-paretic step end ρ_{SW}	0.03 (0.77)	-	-0.01 (0.95)	-0.11 (0.72)	-	0.11 (0.71)
Non-paretic step start ρ_{FP}	**0.40 (0.0001)**	0.13 (0.22)	0.13 (0.20)	**0.55 (0.006)**	0.42 (0.075)	0.33 (0.10)
Non-paretic step end ρ_{FP}	**0.35 (0.0005)**	0.10 (0.33)	0.05 (0.64)	**0.48 (0.038)**	0.29 (0.25)	0.29 (0.21)
Non-paretic step start ρ_{PD}	-0.30 (0.004)	-	0.01 (0.90)	-0.98 (0.094)	-	0.01 (0.99)
Paretic step start ρ_{SW}	**0.43 (<0.0001)**	**0.23 (0.028)**	0.08 (0.45)	**0.57 (0.033)**	0.35 (0.25)	0.25 (0.35)
Paretic step end ρ_{SW}	**0.43 (<0.0001)**	0.16 (0.13)	0.02 (0.83)	0.37 (0.12)	-	0.26 (0.28)
Paretic step start ρ_{FP}	**0.38 (0.0002)**	0.06 (0.58)	**0.24 (0.020)**	**0.70 (0.0007)**	**0.63 (0.013)**	**0.45 (0.036)**
Paretic step end ρ_{FP}	**0.31 (0.003)**	-0.07 (0.53)	0.12 (0.25)	**0.62 (0.005)**	0.48 (0.064)	0.42 (0.060)
Paretic step start ρ_{PD}	**0.22 (0.034)**	0.14 (0.17)	-0.13 (0.22)	-0.27 (0.45)	-	-0.53 (0.14)

Table 2. Linear correlation and logistic regression results. For FGA, speed-corrected FGA, and ABC (shaded columns), Spearman correlation coefficients are presented, with corresponding p-values in parentheses. Positive correlations with p-values less than 0.05 are bolded. Positive correlation values indicate that a larger biomechanical metric was linked with better performance on clinical tests. For fear of falling, self-reported fear of falling, and self-reported falls, regression coefficients are presented, with corresponding p-values in parentheses. Positive regression coefficients with p-values less than 0.05 are bolded. Positive coefficients indicate that a larger biomechanical metric was linked with a lower probability of a fear-of-falling or self-reported fall.

Figures

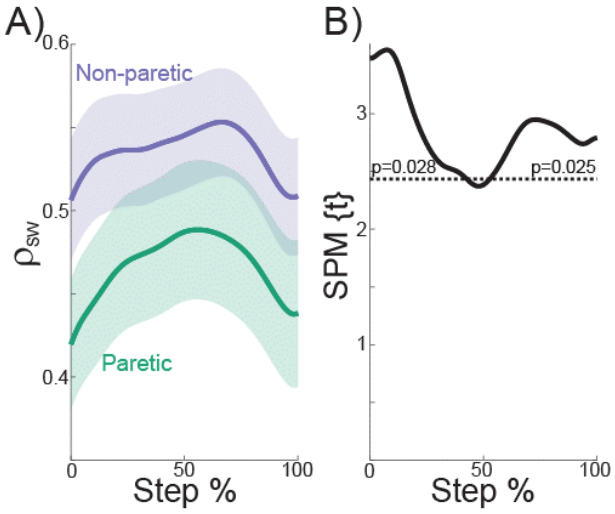

Figure 1. Partial correlation trajectory between step width and mediolateral pelvis displacement over the course of a step. A) The magnitude of ρ_{SW} was higher for non-paretic steps than paretic steps. Lines indicate mean values, and shaded areas indicate 95% confidence intervals. B) The difference between non-paretic and paretic steps reached significance both early and late in the step, as indicated by values greater than the critical t-value shown as the dashed line.

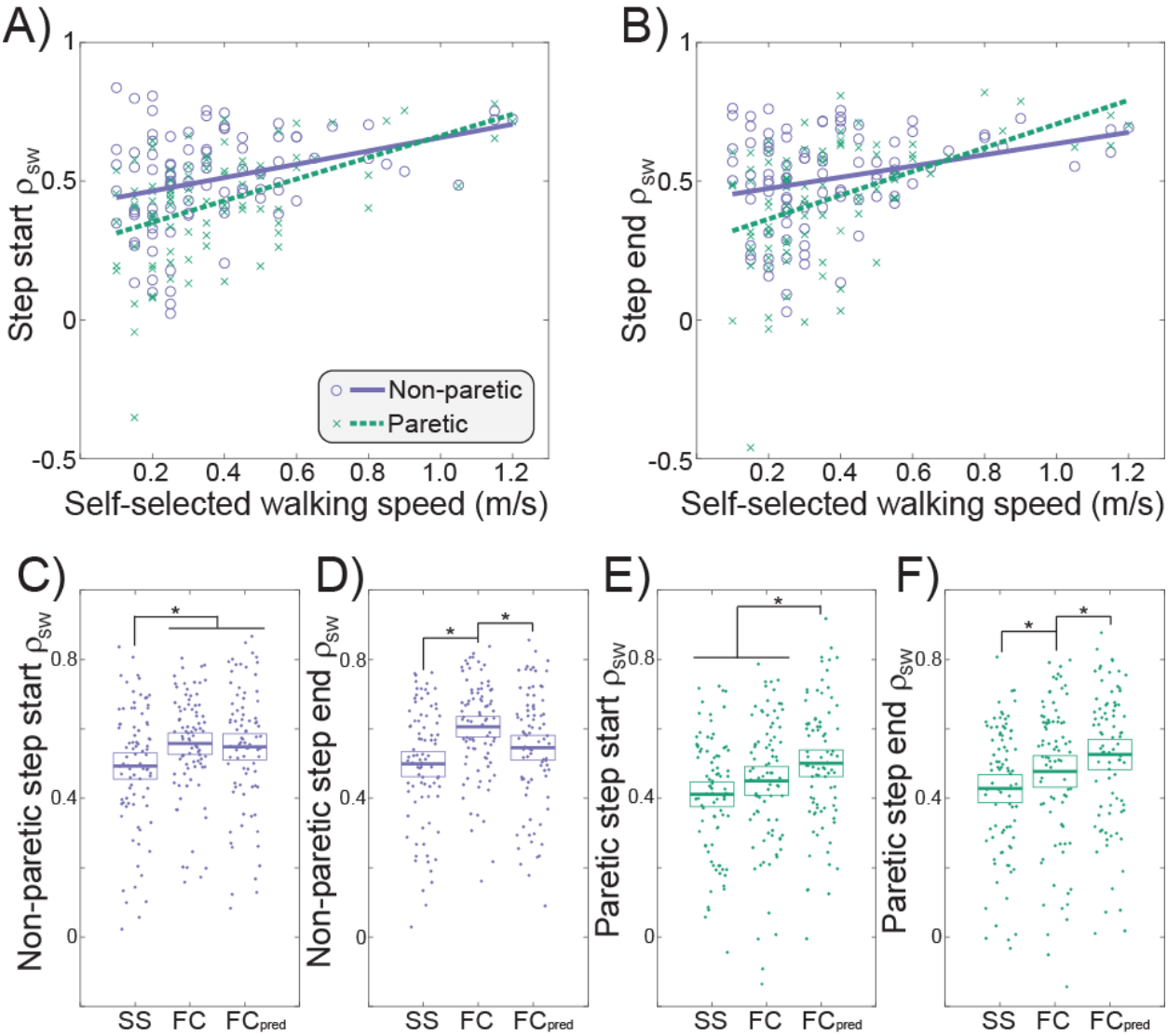

Figure 2. Effects of speed on ρsw. A) Across participants, step start ρsw values increased with walking speed, and were higher for non-paretic than paretic steps. B) Across-participants, step end ρsw values increased with walking speed. For participants who walked slowly, non-paretic step values were higher than paretic step values. For panels A and B, data points represent individual participant (n=93) values, while lines illustrate the best linear fit of these data points to walking speed. C) Non-paretic step start ρsw values were compared within participants for the self-selected speed condition (SS), the fastest-comfortable speed condition (FC), and the predicted value for the

fastest-comfortable condition based on the interparticipant relationship (FC_{pred}). Non-paretic step start ρ_{SW} increased at the faster speed, to approximately the predicted level. D) Non-paretic step end ρ_{SW} increased at the faster speed, more than predicted. E) Paretic step start ρ_{SW} did not increase significantly at the faster speed and did not reach the predicted level. F) Paretic step end ρ_{SW} increased at the faster speed, but not to the predicted level. For panels C-F, data points indicate individual participant values (n=84), thick horizontal lines indicate means, boxed areas indicate 95% confidence intervals, and asterisks indicate significant post-hoc differences.

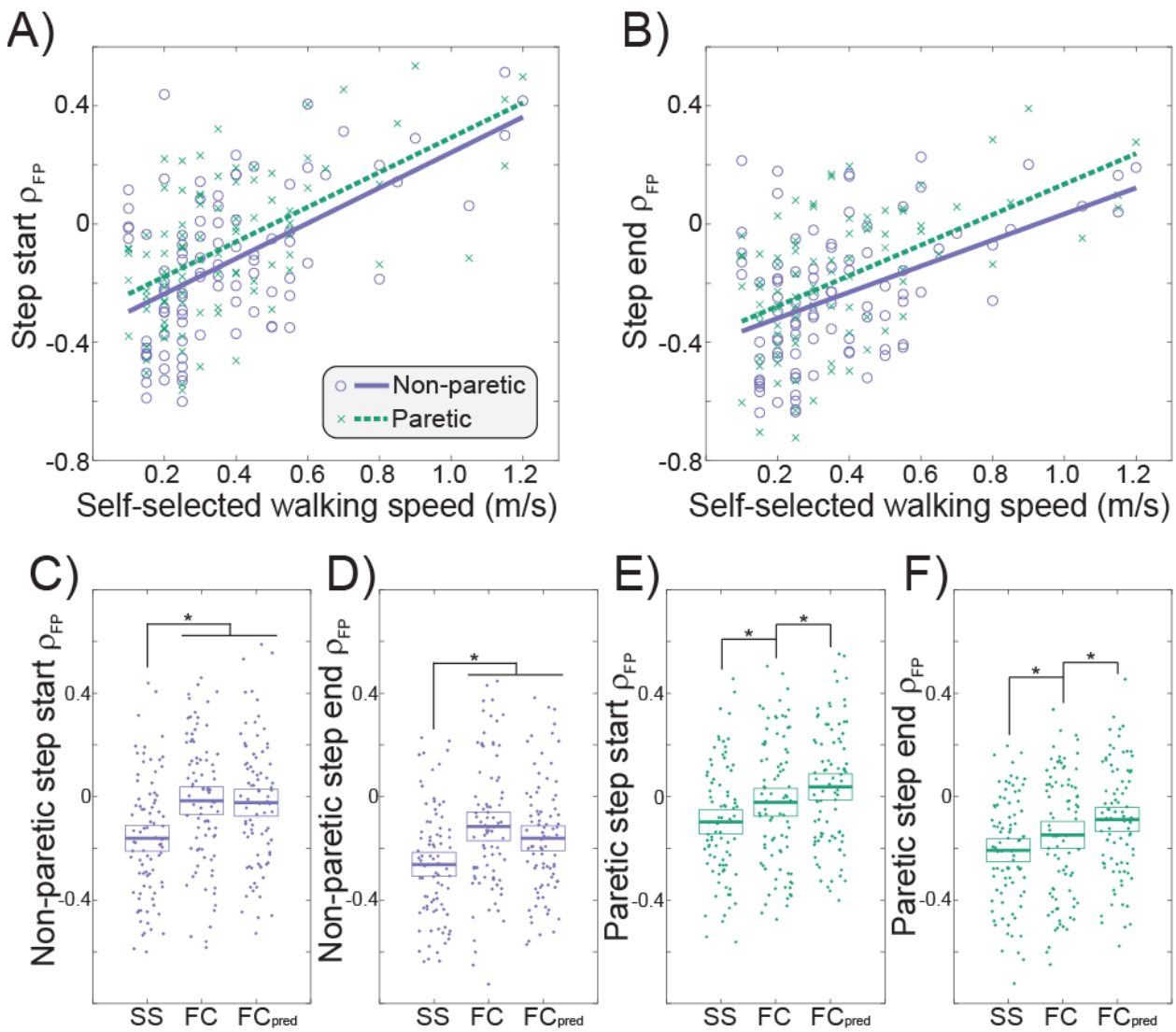

Figure 3. Effects of speed on ρ_{FP}. A) Across participants, step start ρ_{FP} values increased with walking speed, but did not differ significantly between non-paretic and paretic steps. B) Similarly, step end ρ_{FP} values increased with walking speed, but did not differ between non-paretic and paretic steps. C) Non-paretic step start ρ_{FP} increased at the faster speed, to approximately the predicted level. D) Non-paretic step end ρ_{FP} increased at the faster speed, to approximately the predicted level. E) Paretic step start ρ_{FP} increased at the faster speed but did not reach the predicted level. F) Paretic step

end ρ_{FP} increased at the faster speed but did not reach the predicted level. The figure content parallels that in Figure 2.

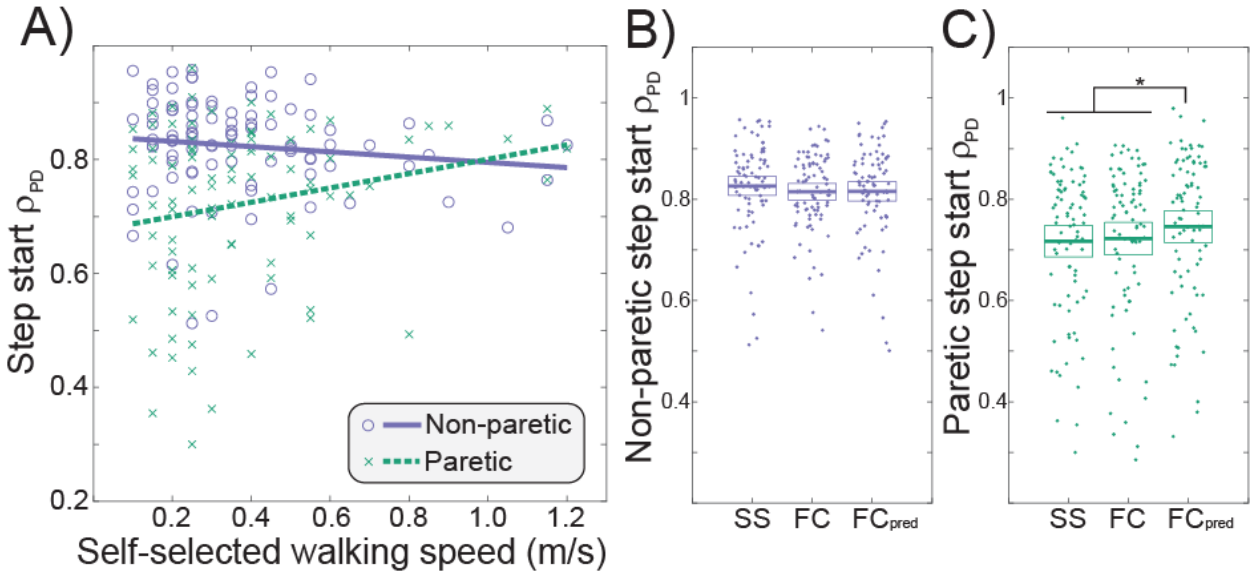

Figure 4. Effects of speed on ρ_{PD}. A) Across participants, step start ρ_{PD} values were higher for non-paretic than paretic steps. The influence of speed on this metric differed for paretic and non-paretic steps, as indicated by the illustrated line slopes. B) Non-paretic step start ρ_{PD} did not increase at the faster walking speed. C) Paretic step start ρ_{PD} also did not increase at the faster walking speed, remaining lower than the moderately increased value predicted from the interparticipant relationship. The figure content parallels that in Figure 2.

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