Paretic propulsion and trailing limb angle are key determinants of long-distance walking function after stroke

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Abstract

Background—Elucidation of the relative importance of commonly targeted biomechanical variables to poststroke long-distance walking function would facilitate optimal intervention design.

Objectives—To (1) determine the relative contribution of variables from three biomechanical constructs to poststroke long-distance walking and (2) identify the biomechanical changes underlying posttraining improvements in long-distance walking function.

Methods—Forty-four individuals > 6 months after stroke participated in this study. A subset of these subjects (n = 31) underwent 12 weeks of high-intensity locomotor training. Cross-sectional (pretraining) and longitudinal (posttraining change) regression quantified the relationships between poststroke long-distance walking function, as measured via the 6-Minute Walk Test (6MWT), and walking biomechanics. Biomechanical variables were organized into stance phase (paretic propulsion and trailing limb angle), swing phase (paretic ankle dorsiflexion and knee flexion), and symmetry (step length and swing time) constructs.

Results—Pretraining, all variables correlated with 6MWT distance ($r's$ = 0.39 to 0.75, $p's$ < 0.05); however, only propulsion (Prop) and trailing limb angle (TLA) independently predicted 6MWT distance (R^2 = 0.655, $F(6,36)$ = 11.38, p < .001). Interestingly, only ΔProp predicted Δ6MWT; however, pretraining Prop, pretraining TLA, and ΔTLA moderated this relationship (moderation model R^2s = 0.383, 0.468, 0.289, respectively).

Conclusions—The paretic limb's ability to generate propulsion during walking is a critical determinant of long-distance walking function after stroke. This finding supports the development of poststroke interventions that impact deficits in propulsion and trailing limb angle.

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Keywords

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INTRODUCTION

More than 5.5 million Americans are currently living with stroke – the leading cause of disability in the USA[1]. For the majority of stroke survivors, the restoration of walking is the ultimate goal of rehabilitation[2]. As such, a major focus of rehabilitation research has been on the development and testing of poststroke gait rehabilitation programs. However, activity and participation are often limited even after rehabilitation[3–8]. Indeed, persons poststroke walk less than 3500 steps per day; in contrast, even the most sedentary healthy adults walk more than 5000 steps per day[9,10]. Given the relationship between physical inactivity and diseases such as heart disease and diabetes[1], a critical need exists for the development of interventions capable of increasing the physical activity of persons who have sustained a stroke.

The development of interventions directed toward improving poststroke walking function is confounded by the fact that improvements in walking function are achievable through a variety of recovery mechanisms – from improved neuromotor control to better compensation for lost neuromotor function[11–14]. Because compensatory strategies such as stiff-legged and circumduction gait are associated with a higher energy cost of walking, reduced endurance, and slower speeds[5,15–17], recovery strategies that rely on gait compensations may limit the gains in long-distance walking function that are achievable through intervention. This is important as persons poststroke indicate that a major contributor to their lack of engagement in the community is a deficit in their ability to walk farther distances[18]. As such, training someone to walk faster may not be sufficient to improve their ability to walk farther if the strategy they use to walk faster is not economical and sustainable. Given the relationships between long-distance walking function[19], community walking participation[20], and the energy cost of walking, a better understanding of the biomechanical determinants of poststroke long-distance walking function is needed.

Improvements in poststroke long-distance walking function may be achieved through any number of biomechanical mechanisms. Indeed, previous investigations have shown relationships between various biomechanical variables and walking function after stroke[21–32]. For example, Sibley et al demonstrated that changes in spatiotemporal asymmetry were associated with less distance walked during the final two minutes of the 6-Minute Walk Test (6MWT) in those with the worst long-distance walking function[21]. Others have identified stance phase variables related to the propulsive force generating ability of the paretic limb as major contributors to poststroke walking function[22–29]. Still, others have shown that deficits in variables related to swing phase ground clearance correlate with poststroke walking function[30–32]. The primary purpose of this study was to determine the relative importance of variables from each of these biomechanical constructs – spatiotemporal symmetry, stance phase, or swing phase – to the long-distance walking function of persons in the chronic phase of stroke recovery. A secondary aim of this study

was to identify the biomechanical changes underlying posttraining improvements in long-distance walking function. We hypothesized that the stance phase construct would be the best predictor of long-distance walking function and that improvement in stance phase mechanics would account for improvements in long-distance walking function. Moreover, we hypothesized that baseline stance phase function would moderate the relationship between stance phase improvements and improvements in long-distance walking function; specifically, that improvements in stance phase mechanics would be more meaningful in those most impaired at baseline.

METHODS

Subjects

Forty-five subjects with poststroke hemiparesis participated in this study. A subset of these subjects (n = 31) underwent 12 weeks of locomotor training as described below. Subjects were recruited over two years from Delaware, New Jersey, and Pennsylvania health care facilities, advertisements, and patient support groups. Subjects were at least 6 months post a single stroke, able to walk at a self-selected pace for six minutes without orthotic support but with observable gait deficits, were able to passively dorsiflex the ankle to a neutral position with the knee extended (tested in the prone position), and were able to passively extend the hip at least ten degrees (tested in a side lying position). Subjects were excluded if they had evidence of moderate to severe chronic white matter disease or cerebellar stroke on MRI, a history of lower extremity joint replacement due to arthritis, an inability to communicate with the investigators, neglect (tested via the star cancellation test[33]) or hemianopia, a score of >1 on question 1b and >0 on question 1c on the NIH Stroke Scale, or unexplained dizziness in the last 6 months. All subjects signed written informed consent forms approved by the Human Subjects Review Board of the University of Delaware, received written medical clearance from their physician, and completed a submaximal stress test to determine exercise safety prior to participation in the intervention protocol described below. Subjects completed clinical and biomechanical evaluations prior to (pretraining) and immediately following 12 weeks of training (posttraining).

Clinical Testing

Clinical evaluations were conducted by licensed physical therapists and included the 6-meter walk test[34] and the 6-minute walk test (6MWT)[35]. Derived from the 6-meter walk test was each subject's self-selected and maximum walking speeds (m/s), which are reported in Table 1 as an indication of baseline walking disability[7]. The distance walked during the 6MWT served as our a priori measure of long-distance walking function. The 6MWT is thought to be reflective of a person's ability to maintain a moderate amount of exertion over a period of time similar to the activities of daily living, has been identified as an excellent measure of poststroke walking capacity and community ambulation[36,37], and as indicative of community reintegration following stroke[5,38]. Subjects were allowed the use of their regular assistive device (e.g. cane) during testing, if necessary. Subjects who used an assistive device at their pretraining evaluation also used one during their posttraining evaluation.

Motion Analysis

Previous work has described in detail the methods used during this investigation[19,39,40]. Briefly, kinetic and kinematic data were collected using an 8-camera motion analysis system (Motion Analysis 3D Eagle, Santa Rosa, CA, USA) as subjects walked for thirty seconds at the maximum walking speed they could maintain for four minutes. For baseline motion analysis testing, this maximum walking speed was determined during an acclimatization session conducted prior to the start of training. The speed used for posttraining motion analysis was determined during the final week of training. During motion analysis testing, subjects walked on a dual-belt treadmill instrumented with two independent 6-degree of freedom force platforms. Ground reaction force (GRF) data were collected at 2000Hz (Bertec Corporation, Worthington, OH). Kinematic data were sampled at a rate of 100 Hz and based on the motion of retro-reflective markers placed over the pelvis, and bilaterally over the thigh, shank, and foot segments, and on the medial and lateral malleoli, at the medial and lateral femoral condyles, the greater trochanters, and the iliac crests. All kinematic and kinetic variables were computed for each stride and averaged across the first 15 strides recorded during motion analysis testing using a custom-written LabVIEW program (National Instruments, Austin, TX, USA).

Six biomechanical variables, divided into three biomechanical constructs – 1) stance phase, 2) swing phase, and 3) spatiotemporal symmetry – quantified biomechanical function during walking. These constructs were selected due to their prevalent study[13,21–32,41–44] and common consideration by clinicians during poststroke rehabilitation. Peak paretic propulsion and peak paretic trailing limb angle (measured during the paretic double support phase) comprised the stance phase construct, peak knee flexion and peak ankle dorsiflexion angles comprised the swing phase construct, and step length symmetry and swing time symmetry comprised the spatiotemporal symmetry construct. Peak propulsion was defined as the maximum anterior GRF recorded during the paretic double support phase, normalized to body weight. Peak trailing limb angle was defined as the maximum sagittal plane angle between the vertical axis of the lab and a vector joining the paretic limb's lateral malleolus and greater trochanter. Peak ankle dorsiflexion was defined as the maximum ankle dorsiflexion angle during the paretic swing phase. Peak knee flexion was defined as the maximum knee flexion angle during the paretic swing phase.

Step length, stride duration, and swing time were calculated bilaterally per stride to allow calculation of the symmetry measures of interest. Step length was defined as the distance between heel markers at the leading limb's initial contact. Stride duration was defined as the time from one initial contact to the subsequent ipsilateral initial contact. Swing time was defined as the time between toe off and initial contact. Swing time was normalized to stride duration. As per a previous study[45], to calculate step length symmetry, the following equation was used: [larger step length / (larger step length + smaller step length)]. To calculate swing time symmetry, the following equation was used: [longer swing time / (longer swing time + shorter swing time)]. A value of 0.50 reflects perfect symmetry. For step length asymmetry, a value of 1.00 reflects a step-to gait pattern and values greater than 1.00 reflect a walking pattern where one limb does not pass the other.

Training Protocol

A subgroup of subjects (n = 31) completed 12 weeks of high intensity locomotor training. Subjects walked at their maximum walking speeds with (n = 15) or without (n = 16) the application of functional electrical stimulation to the paretic ankle dorsiflexors during swing phase and plantarflexors during late stance phase. The training protocol used has been previously described[39,40]. Regardless of whether subjects trained with or without FES, the training provided task-specific practice of thousands of steps per treatment session. Training occurred at a frequency of 3 sessions per week for 12 weeks. Approximately 36 minutes of total walking were completed during each session. Because the present manuscript is only concerned with a mechanistic investigation of the biomechanical changes underlying changes in long-distance walking function, change-score data from these treatment groups have been combined. A subsequent manuscript will test treatment efficacy by investigating group-specific effects as they relate to a control group.

Statistical Analyses

All statistical tests were performed in SPSS version 21. Sequential and moderated regression analyses[46,47] of cross-sectional (ie, pretraining) and longitudinal (ie, posttraining change) data were performed. Centered variables were used in the analysis. Standardized regression coefficients (β) are reported, allowing us to infer the strongest predictor of long-distance walking function based on magnitude. Residuals for each of the regression models were screened for the presence of outliers. Alpha level of 0.05 was set as the threshold for statistical significance. One-tail tests were used for effects with an *a priori* directional hypothesis.

Sequential linear regression was used to test our hypothesis that the stance phase construct would be the strongest predictor of long-distance walking function. The order by which the swing phase and spatiotemporal symmetry constructs were added to the model was based on the magnitudes of the bivariate correlations, with strongest added first. With 44 subjects, and alpha set at 0.05, this study had power = 0.80 to detect an R^2 increase between 0.20 (1-tail) and 0.24 (2-tail) when adding the swing phase and spatiotemporal symmetry constructs (ie, four variables) to the model containing the stance phase construct (ie, two variables).

Bivariate correlations of the longitudinal data were used to test our hypothesis that improvements in stance phase function would relate to improvements in long-distance walking function. Moderated regression was used to test our hypothesis that baseline stance phase function would moderate the relationship between changes in stance phase mechanics versus changes in long-distance walking function. Because only changes in paretic propulsion correlated to changes in long-distance walking function (see Results), only interactions with change in paretic propulsion were tested. The available sample size precluded us from examining all interactions in one model, so to avoid model over fit and to maintain adequate power, independent moderation models were generated to examine each interaction.

Specifically, the first model tested the interaction between pretraining propulsion and change in propulsion and the second tested the interaction between pretraining trailing limb

angle and change in propulsion. Based on our finding of moderation by pretraining trailing limb angle (see Results), the third moderation model tested moderation by changes in trailing limb angle with an *a priori* hypothesis that changes in propulsion would have a stronger relationship to changes in long-distance walking function in those with concomitant changes in trailing limb angle. With 30 subjects, at an alpha level of 0.05, each moderated regression model was 80% powered to detect an R^2 increase (1-tail) of 0.22.

RESULTS

Clinical data were available for all subjects (see Table 1 for subject characteristics); however, due to technical issues during data collection, pretraining biomechanical data were not available for 1 of the 45 subjects studied. Moreover, a single subject was found to be a statistical outlier and was removed prior to the analyses presented. These 2 subjects were also among the cohort (n = 31) who underwent training. As such, the cross-sectional analyses presented reflect the data collected for 43 subjects and the longitudinal analyses reflect the data collected for 29 subjects. Table 2 presents means, standard deviations, minimums, and maximums for the pretraining and change-score variables included in the regression analyses conducted.

Cross-sectional Analyses

Despite variables from all three constructs correlating with performance on the 6MWT (Figure 1), only paretic propulsion ($\beta = .339$) and trailing limb angle ($\beta = .564$) independently predicted 6MWT distance (Table 2; $R^2 = .655$, $F(6,36) = 11.38$, $p < 0.001$).

Longitudinal Analyses

Bivariate correlations of the longitudinal data revealed that only changes in paretic propulsion ($r = 0.435$, $p = 0.009$) correlated with changes in the 6MWT. Interestingly, changes in trailing limb angle and in the swing phase and the symmetry variables studied did not ($r's < 0.29$ and $p's > 0.05$).

Moderated regression analyses revealed three independent moderators of the relationship between changes in paretic propulsion and changes in the 6MWT (see Table 3). The first was moderation by pretraining paretic propulsion (final model testing this interaction: $F(3,25) = 5.17$, $p = 0.006$, $R^2 = 0.383$, $\Delta R^2 = 0.193$). Specifically, changes in propulsion were strongly positively related to changes in the 6MWT for those with low pretraining propulsion – that is, those with baseline propulsion lower than one standard deviation *below* the mean (ie, < 3.67% BW). For those with average propulsion (ie, 8.67% BW), changes in propulsion weakly positively related to changes in the 6MWT. For those with pretraining propulsion greater than one standard deviation *above* the mean (ie, > 13.67% BW), a weak negative relationship was observed (Figure 2, panel A).

An even stronger effect was observed when testing moderation by pretraining trailing limb angle (Table 3; $F(3,25) = 7.34$, $p = 0.001$, $R^2 = 0.468$, $\Delta R^2 = 0.279$). Similar to the effect of pretraining propulsion, the strongest relationship between changes in paretic propulsion and changes in 6MWT distance was observed in those with a pretraining trailing limb angle lower than one standard deviation *below* the mean (ie, < 7.2 degrees). A weaker positive

relationship was observed in those with the average pretraining trailing limb angle (ie, 15.35 degrees) and a weak negative relationship was observed in those with the largest pretraining trailing limb angle (ie, > 23.5 degrees) (Figure 2, panel B).

The weakest moderator of the relationship between changes in propulsion and changes in the 6MWT was changes in trailing limb angle (Table 3; $F(3,25) = 3.39$, $p = 0.034$, $R^2 = 0.289$, $\Delta R^2 = 0.094$). The relationship between changes in propulsion and changes in the 6MWT was strongest in those with the largest change in trailing limb angle (ie, > 7.41 degrees). Interestingly, for those with a change in trailing limb angle one standard deviation below the mean (ie, a decline of 1.89 degrees or greater), changes in propulsion were unrelated to changes in the 6MWT (Figure 2, panel C).

DISCUSSION

This report is the first to explore the relative importance of stance phase, swing phase, and spatiotemporal symmetry biomechanics to poststroke long-distance walking function. This investigation extends previous work that has studied the biomechanical determinants of *short-distance* walking function[12,26,28,48–50]. The present results reveal a relationship between the function of the paretic limb during stance phase – particularly the propulsive force generated during late stance – and *long-distance* walking function in persons in the chronic phase of stroke recovery. Moreover, the results of the moderated regression analyses indicate that this relationship is greatest in those persons presenting with large pretraining impairments in propulsion or trailing limb angle. Given that a majority of individuals in the chronic phase of stroke recovery identify deficits in their ability to walk farther as contributing to reduced engagement in the community[18], by identifying key biomechanical determinants of poststroke long-distance walking function, this investigation facilitates the development of targeted interventions with the potential to increase community participation after stroke.

Previous investigators have posited that an assessment of post-intervention outcomes is lacking if limited to only gross clinical measures such as walking speed[51]. The present investigation's elucidation of key biomechanical determinants of poststroke long-distance walking function therefore orients clinicians to important poststroke gait variables, ultimately informing clinical practice. Although task-specific practice forms a necessary basis for neurorehabilitation efforts[16,52–54], the present findings support structuring practice in a manner that targets the specific impairments that may be limiting performance. For example, although walking practice is commonly prescribed as a therapeutic intervention, the present results suggest that training walking at a fast speed will produce improvements in long-distance walking function associated with the recovery of paretic limb biomechanical function – especially in those most impaired. Further development and testing of hypothesis-driven targeted locomotor interventions for persons poststroke is warranted.

Interestingly, despite not relating to changes in long-distance walking function, changes in trailing limb angle moderated the relationship between changes in propulsion and changes in long-distance walking function. Specifically, only in those with a large improvement in the

paretic trailing limb angle did gains in propulsion relate to gains in long-distance walking (see Figure 2, panel C). One explanation for these apparently contradictory findings is that improvements in trailing limb angle are not meaningful – in terms of improving long-distance walking function – if they do not result in improvements in propulsion. Indeed, although increasing the paretic trailing limb angle yields a more effective biomechanical position for the generation of propulsive forces by the ankle musculature[27], it is important to note that persons poststroke often use the hip flexors to advance the paretic limb during the stance to swing transition[24,55,56] – which is a strategy known to negatively correlate with the propulsive forces generated[15,27,57,58]. That is, merely providing better resources (ie, a larger paretic trailing limb angle) may be insufficient to alter the strategy used to walk faster. The specific training of use of the ankle musculature may be necessary.

Multiple factors may influence performance on the 6MWT – our measure of long-distance walking ability. We have previously shown that changes in maximum walking speed account for greater than 50% of the variance in changes in 6MWT performance[59]. Other factors certainly contribute. One possible factor is changes in the energy cost of walking. Although the present report does not directly investigate the role that changes in walking energetics may play in modifying long-distance walking function, recent work from our laboratory demonstrates a meaningful relationship between posttraining changes in walking kinematics, specifically step length asymmetry, and changes in the energy cost of walking[45]. Surprisingly, the present investigation revealed that changes in walking kinematics were unrelated to changes in 6MWT performance, suggesting that deficits in walking kinematics were not limiting long-distance walking function as measured via the 6MWT. In contrast, it has been shown that compensatory kinematic strategies are energetically costly[15–17,60] and previous work from our laboratory has shown that those with lower walking energy costs travel farther distances during the 6MWT[19]. Moreover, although not directly related to changes in the 6MWT in the present investigation, changes in the paretic trailing limb angle moderated the influence that changes in paretic propulsion had on changes in 6MWT performance. As such, further investigation of the interplay between walking kinematics, the energy cost of walking, and long-distance walking function is warranted.

Based on the findings of this investigation, a reasonable hypothesis would be that an intervention targeting deficits in paretic propulsion through specific effects on the paretic trailing limb angle could produce improvements in the functional status of persons in the chronic phase after stroke. Indeed, our laboratory recently published a preliminary study that supports this hypothesis by demonstrating improvements across the domains of the World Health Organization's International Classification of Functioning, Health, and Disability following training targeting deficits in paretic propulsion through specific effects on the function of the paretic limb during late stance[40]. The findings from this preliminary study validate the present study's emphasis on stance phase mechanics. However, future work is necessary to determine the efficacy of interventions targeting paretic propulsion across subgroups of patients stratified according to baseline biomechanical function.

An important point is that although this investigation considered the biomechanical constructs studied independently, for an individual, these variables are likely interrelated. That is, events during stance phase may have a direct impact on swing phase function, and

changes in both stance phase and swing phase underlie changes in spatiotemporal symmetry. For example, increased propulsive force during late stance is one mechanism posited to increase knee flexion during swing[61–63]. Moreover, improvements in step length symmetry may result from a larger trailing limb angle – which would effectively increase the contralateral step length – or better propulsion – which may increase ipsilateral step length through its swing phase effects. Even so, by examining the relative importance of each of these constructs – particularly how changes in each relate to the changes in long-distance walking function observed after gait rehabilitation – this report reveals important information regarding the mechanisms that may be driving the recovery of poststroke walking function. Future work that examines how changes in other biomechanical measures, such as mechanical work or power, account for the variance in changes in long-distance walking function after poststroke locomotor intervention would further extend this work.

Limitations

A potential limitation of this study is that biomechanical testing occurred on a treadmill. Conceptually, relating changes in overground gait mechanics to changes in overground long-distance walking function would have been preferable; however, it should be noted that treadmill biomechanical assessment has several advantages over overground testing. These include the averaging of consecutive strides, the ability to control speed – a major determinant of gait mechanics, increased patient safety, and a marked reduction in efforts by both patient and researcher to generate data for a large number of strides. Previous work has also shown that treadmill biomechanical data provides relevant information for understanding overground walking[64,65].

A second potential limitation of this study is that some subjects utilized a handrail during testing. Specifically, subjects who typically used an assistive device or those who felt unsafe walking on the treadmill were allowed to use a handrail. The use of a handrail during testing may promote a forward trunk lean that could influence our measurement of trailing limb angle if the pelvis/trunk are not aligned with the vertical axis of the laboratory. However, it should be noted that subjects were only allowed to use a handrail located at the side of the treadmill. This mimicked walking with an assistive device and placed minimal constraint on the anterior/posterior displacement of the body during walking. It should also be noted that subjects were instructed to use the minimal amount of handrail support possible.

Conclusions

Because a rapid achievement of walking independence, not necessarily the reduction of impairment, is the goal of current neurorehabilitation practice[66], the high prevalence of inefficient walking strategies among persons in the chronic phase of stroke recovery is not surprising[15,16]. Maximizing posttraining outcomes for persons in the chronic phase of stroke recovery may therefore necessitate the learning of new walking strategies. The findings of this investigation support the development of poststroke locomotor interventions that include the targeting of paretic limb stance phase deficits during walking – specifically propulsion and trailing limb angle.

Acknowledgements

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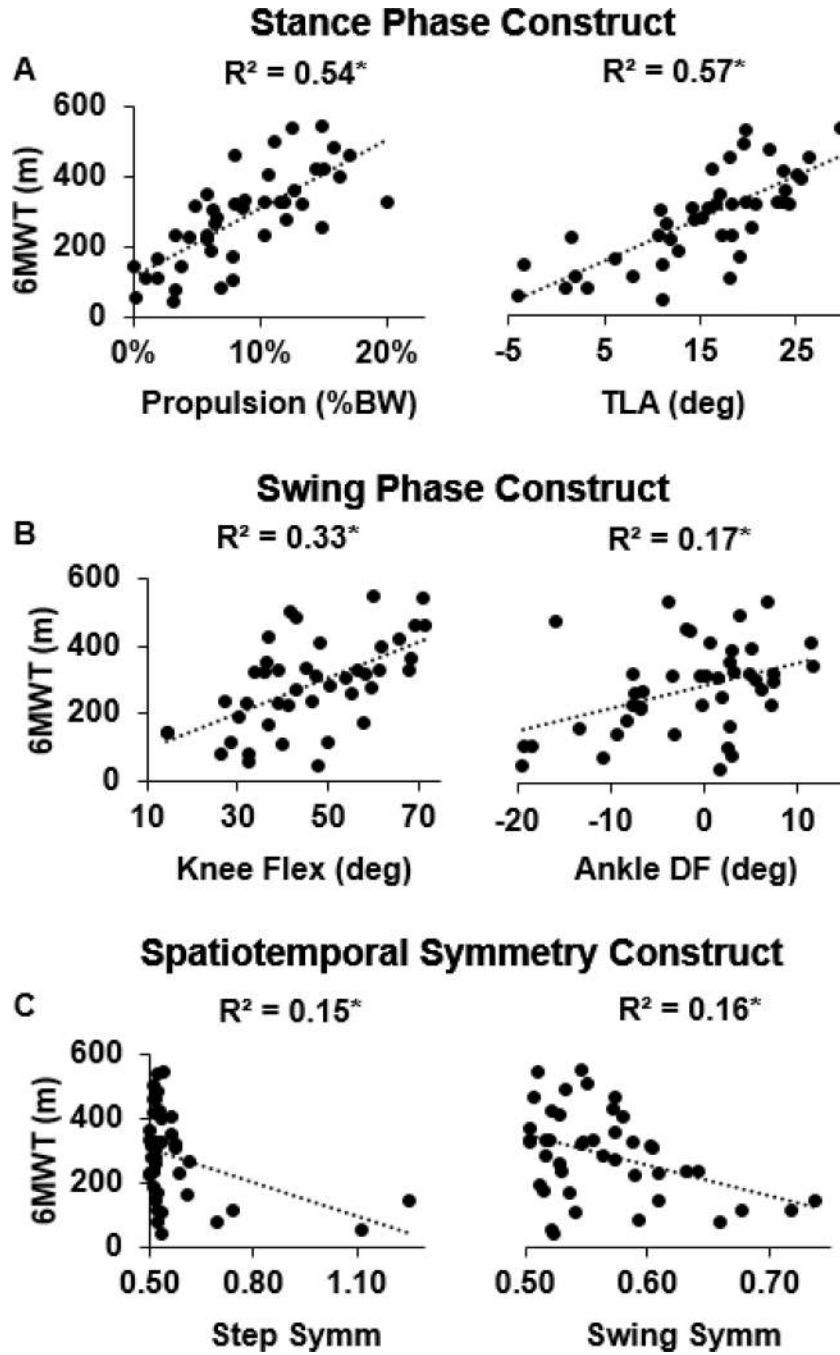
Figure 1.
Scatter plots present the relationships between stance phase (panel A), swing phase (panel B), and spatiotemporal symmetry (panel C) biomechanics versus long distance walking function. All variables considered were related to performance on the 6MWT; however, stance phase function exhibited the highest degree of correlation. * $p < .05$.
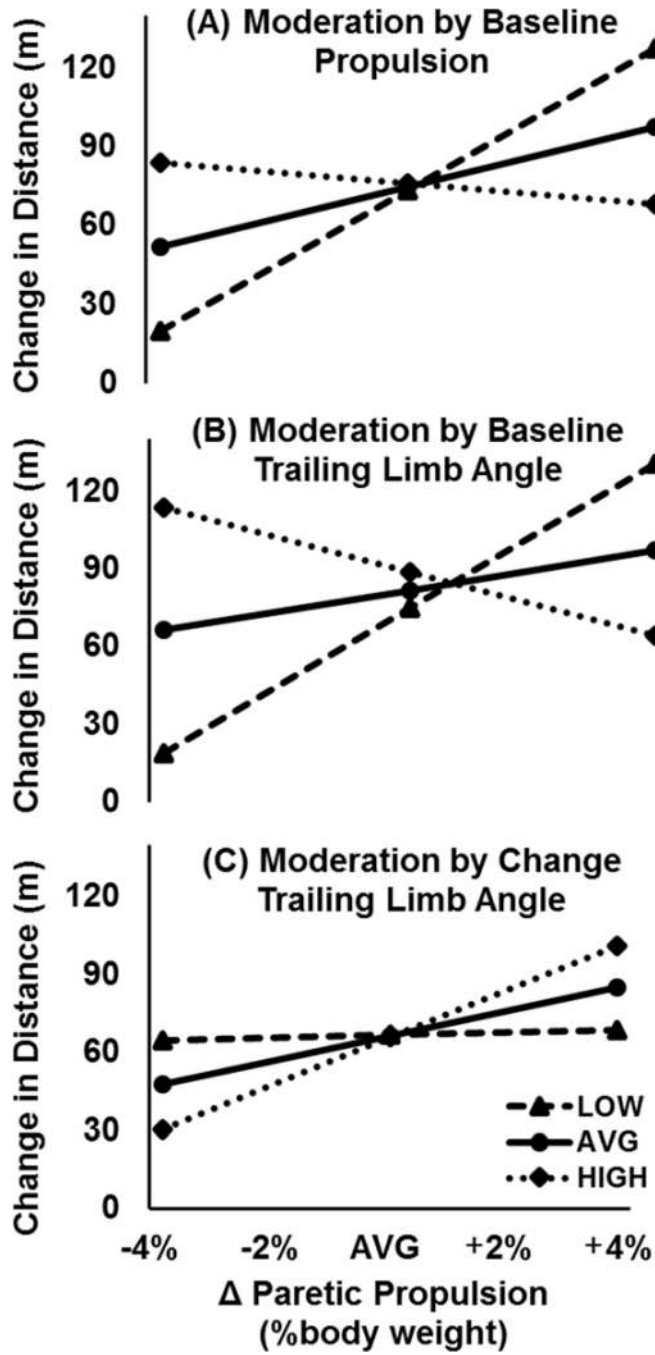
Figure 2.
Moderated regression plots present a visual representation of the relationship between changes in paretic propulsion (x-axis) versus changes in long-distance walking function (y-axis) as moderated by pretraining propulsion (panel A), pretraining trailing limb angle (panel B), and changes in trailing limb angle (panel C). This relationship was strongest in those with LOW (ie, 1 standard deviation below the mean) pretraining levels of propulsion (< 3.67% body weight, panel A) and trailing limb angle (< 7.2 degrees, panel B). Moreover, changes in paretic propulsion most strongly related to changes in long-distance walking

function in those with the largest change in trailing limb angle (ie, 1 standard deviation above the mean: > 7.41 degrees) (panel C). Please note that panels A-C present simple slopes at each level of the moderator, not a grouping of subjects.

Table 1

Subject (n = 44) Characteristics

Variable	Median (SIQR) or Frequency (%)
Age, y	60.08 (2.49)
Time Since Stroke, y	1.72 (0.73)
Sex, Female	39%
Side of Paresis, Left	66%
Self-Selected Walking Speed, m/s	0.74 (0.12)
Maximum Walking Speed, m/s	1.03 (0.15)

Table 2

Pretraining and posttraining change-score mean (SD) values.

Pretraining (n = 43)

Variable	Mean (SD)	Min/Max
Paretic Propulsion, % body weight	8.67 (5.00)	0.00/20.12
Paretic Trailing Limb Angle, degrees	15.35 (8.15)	−3.98/29.61
Paretic Knee Flexion, degrees	46.04 (14.79)	14.43/71.39
Paretic Dorsiflexion, degrees	−1.65 (8.01)	−19.45/11.61
Step Length Symmetry	0.570 (0.147)	0.501/1.250
Swing Time Symmetry	0.565 (0.057)	0.504/0.738
Six-Minute Walk Test Distance, m	285 (134)	44/546

Change-Scores (n = 29)

Variable	Mean (SD)	Min/Max
Δ *Paretic Propulsion, % body weight*	2.26 (3.78)	−4.26/14.74
Δ *Paretic Trailing Limb Angle, degrees*	2.76 (4.65)	−4.62/16.00
Δ *Paretic Knee Flexion, degrees*	2.62 (6.82)	−11.40/20.72
Δ *Paretic Dorsiflexion, degrees*	1.12 (6.29)	−10.94/18.23
Δ *Step Length Symmetry*	0.034 (0.141)	−0.055/0.746
Δ *Swing Time Symmetry*	−0.011 (0.042)	−0.109/0.088
Δ *Six-Minute Walk Test Distance, m*	72 (61.68)	−37/207

Sequential regression models predicting *pretraining* 6MWT distance (n = 43).

	Model Statistics			Predictor Statistics			
Block	**Model R^2**	**Model F**	**Model p**	**Predictors**	**β**	**t**	**p**
1	.615	31.92	.000	Paretic Propulsion	.363	2.15	.019
				Trailing Limb Angle	.460	2.73	.005
1 + 2	.621	15.54	.000	Paretic Propulsion	.331	1.86	.035
				Trailing Limb Angle	.415	2.19	.018
				Ankle Dorsiflexion	.098	0.73	.472
				Knee Flexion	.017	0.14	.887

Sequential regression models predicting *pretraining* 6MWT distance (n = 43).

	Model Statistics			Predictor Statistics			
Block	Model R^2	Model F	Model p	Predictors	β	t	p
				Paretic Propulsion	.339	1.92	.031
				Trailing Limb Angle	.564	2.75	.005
				Ankle Dorsiflexion	.166	1.20	.237
1 + 2 + 3	.655	11.38	.000	Knee Flexion	.043	0.37	.717
				Step Length Symmetry	.156	1.19	.243
				Swing Time Symmetry	.184	1.40	.170

Table 3

Moderated regression analyses predicting *change* in 6MWT distance (n = 29).

Model	Model Statistics				Predictor Statistics			
	Model R^2	Model F	Model p	Predictors	β	t	p
Moderation of Δ Prop by Pre-tx Prop	.383	5.17	.006	Δ Prop	.340	2.11	.023
				Pre-tx Prop	.098	0.62	.539
				Δ Prop * Pre-tx Prop	−.436	−2.72	.012
Moderation of Δ Prop by Pre-tx TLA	.468	7.34	.001	Δ Prop	.227	1.43	.082
				Pre-tx TLA	.216	1.43	.082
				Δ Prop * Pre-tx TLA	−.526	−3.43	.002
Moderation of Δ Prop by Δ TLA	.289	3.39	.034	Δ Prop	.213	1.02	.160
				Δ TLA	−.049	−0.25	.403
				Δ Prop * Δ TLA	.408	1.82	.041