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Neurodynamics under Different Walking Speeds in Individuals with Chronic Post-Stroke Hemiparesis

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NEURODYNAMICS UNDER DIFFERENT WALKING SPEEDS IN INDIVIDUALS WITH CHRONIC POST-STROKE HEMIPARESIS

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ABSTRACT

Background and Purpose: Stroke is the leading cause of long-term disability in adults worldwide. The ability to return to walking is often a main goal of rehabilitation in individuals with chronic post-stroke hemiparesis. To increase walking speed, non-neurologically impaired individuals produce greater ankle propulsion force at push-off with greater ankle dorsiflexion angles in swing phase with no change in ankle muscle co-contraction index in the swing phase. It remains unclear if individuals post-stroke would adopt similar neuromuscular strategies. Therefore, our aim was to examine the effect of altered walking speeds on propulsion force at push-off, ankle dorsiflexion angle during swing, and co-contraction of the lower leg musculature in individuals with chronic post-stroke hemiparesis.

Subjects: We recruited 7 participants with chronic post-stroke hemiparesis and 7 age-similar, non-neurologically impaired controls. Inclusion criteria were 1) > 6 months post stroke with hemiparesis, 2) able to walk without an assistive device for 2 minutes, and 3) able to follow cues and adhere to instructions. Exclusion criteria were 1) had a history of cerebellar stroke(s) and/or 2) unable to walk without an assistive device for more than 2 minutes.

Methods: All subjects were tested under three different walking speed conditions: self-selected walking speed (SSWS), fast walking speed (FWS), and slow walking speed (SWS). We examined the propulsion force at push-off, ankle dorsiflexion angle during swing phase, and co-contraction index of the tibialis anterior and gastrocnemius muscles during stance and swing phases. A 2-factor mixed factorial ANOVA was used to assess each variable between leg and the speed condition (FWS, SSWS, SWS). The legs examined were the paretic limb of participants post-stroke, the non-paretic limb of participants post-stroke, and the non-impaired limb of non-neurologically impaired controls.

Results: The ANOVA and post-hoc analyses revealed that there were significant increases in ankle dorsiflexion angle during swing phase and propulsion force at push-off in the FWS

(4.6±4.3° and -1.1±0.6 N/kg respectively) condition when compared to the SSWS (5.6±4.8° and -0.9±0.5 N/kg respectively) and SWS (5.3±4.6° and -0.7±0.4 N/kg respectively) conditions across the 3 limbs examined. Additionally, the speed and limb had no main effect ($p=0.233$ and $p=0.554$ respectively) on co-contraction index between the tibialis anterior and gastrocnemius at peak dorsiflexion during swing and had a trending main effect ($p=0.082$) on co-contraction index between the tibialis anterior and gastrocnemius at push off.

Discussion: Faster walking speeds may help people post-stroke to improve their propulsion force and ankle kinematics during gait. Future studies should investigate individuals with different types of strokes as well as the percentage of speed increase that evokes consistent improvements in gait mechanics in people post-stroke for physical therapy interventions.

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INTRODUCTION

Stroke is the leading cause of long-term disability.[1] Post-stroke hemiparesis results in significant impairments in locomotor function[2], which negatively impacts functional and mobility independence and safety, resulting in increased energy expenditure, slower walking speeds, and increased risks for falls.[2] The ability to modulate walking speed is essential to address changes in environmental demands.[2]

In non-neurologically impaired individuals, during walking under constant speeds, the body center of mass is accelerated by propulsion force and decelerated by the braking ground reaction forces (GRFs)[3], and the activities of leg muscles are coordinated such that these GRFs are symmetric bilaterally.[4] Furthermore, non-neurologically impaired individuals can increase their speed by increasing step length, increasing cadence, reducing the duration of their stance phase, and/or increasing the duration of their swing phase.[5] In individuals with post-stroke hemiparesis, however, insufficient propulsion and excessive braking GRFs have been observed in the paretic limbs.[2, 6] During the swing phase of gait, individuals post-stroke also exhibit reduced dorsiflexion joint angles on the paretic side in comparison to non-neurologically impaired individuals.[7-9] These deficits together contribute to inefficient gait patterns and slower walking speeds, which have been found to increase risk for falls in individuals with chronic post-stroke hemiparesis.[10] These deficits may be attributed to co-contractions of the gastrocnemius and tibialis anterior muscles, which affect the gastrocnemius' ability to generate propulsion force and tibialis anterior's function for foot clearance during swing phase.[9-11]

In the gait cycle, ankle plantarflexion is an important contributor to propulsion force and dorsiflexion allows foot clearance for smooth transition through the swing phase.[12, 13] As mentioned previously, propulsion and braking forces are the forces leading to the acceleration and deceleration components of walking respectively, where an increase in force results in further acceleration or deceleration.[3] From a previous study looking at a non-neurologically impaired population, a change in speed was shown to positively correlate with a change in propulsion and braking forces.[14] However, it was found that while there is a significant correlation between co-contraction ratio of the shank with the metabolic cost of walking, there is no significant change in the co-contraction ratio of the gastrocnemius and the tibialis anterior muscles with changes in walking speed during treadmill walking in a healthy, older population.[15, 16] The co-contraction ratio in individuals living post-stroke is highly variable thus, more research is needed to investigate changes in the co-contraction ratio of the gastrocnemius and tibialis anterior muscles associated with changes in walking speed for the post-stroke population.[17]

Electrical stimulation has been utilized on both the plantarflexors and the dorsiflexors to improve foot clearance during gait in individuals post-stroke.[18] When using electrical stimulation, correctly timed activation of the plantarflexors and the dorsiflexors corrected gait deficits in propulsion, plantarflexion, and knee flexion during the overall gait cycle.[18] These strategies were shown to improve gait parameters in persons post-stroke after 12 weeks of training, demonstrating that people with chronic post-stroke hemiparesis were able to learn and develop improved gait patterns.[19-21]

With increased walking speed, non-neurologically impaired individuals demonstrated increased ankle plantarflexion power and increased ankle dorsiflexion angle during the swing phase.[22, 23] It remains unclear if increasing walking speed would have the same effects on ankle kinetics, kinematics, and muscle activation in individuals post-stroke. To date, no one has manipulated walking speed to investigate the effects on propulsion forces at push-off, dorsiflexion angle during swing phase, or muscle co-contraction of gastrocnemius and tibialis anterior during gait in persons post-stroke. In addition, there has been inconclusive research on propulsion force, where walking at a faster speed did not evoke a uniform increase in ankle propulsion force at toe off between paretic and non-paretic limbs. Previous research has not contributed to demonstrating that there is a more symmetric propulsion force between paretic and non-paretic limbs at a faster walking speed for individuals post-stroke.[7] There is also little research on ankle dorsiflexion angle in unassisted gait of individuals post-stroke.[8, 24]

It should also be noted that none of the previously mentioned studies have examined these qualities in individuals with hemiparesis due to stroke, while walking with full weight bearing on a treadmill, without the use of handrails or ankle-foot orthosis. Thus, the purpose of this study is to analyze the changes in propulsion and braking forces, ankle joint kinematics, focusing on ankle dorsiflexion in swing phase, and tibialis anterior and gastrocnemius muscle activity at three different walking speeds, self-selected walking speed (SSWS), fast walking speed (FWS), and slow walking speed (SWS), in individuals with chronic post-stroke hemiparesis. This will provide evidence for activity-dependent spinal plasticity in the stroke-impaired nervous system, and could lead to development of a novel targeted intervention to enhance post-stroke locomotor control and interaction with challenging environmental conditions during daily life such as speed walking.

METHODS

Subjects

We recruited seven participants with chronic post-stroke hemiparesis (58.0±12.1 years old, 3 males; 30.6±14.6 months post-stroke) and 7 age-similar, non-neurologically impaired controls (60.1±9.8 years, 3 males) were recruited. The pool of post-stroke candidates for the study were recruited from local clinics, community centers, and support groups using IRB-approved advertisements and word of mouth. Inclusion and exclusion criteria are detailed in Table 1. A sample size calculation suggested seven individuals were needed in each group to detect a group difference in dorsiflexion angle using a two-sided paired t-test with 95% power and α value of 0.05.

INCLUSION CRITERIA	EXCLUSION CRITERIA
• 6+ months post-single, unilateral, cortical or subcortical stroke with residual hemiparesis • Able to walk without an assistive device or ankle foot orthosis or brace for at least 2 minutes as tested by the 2 Minute Walk Test • Able to follow and adhere to verbal instructions	• Cerebellar strokes • Unable to walk without an assistive device or ankle foot orthosis or brace for at least 2 minutes.

Table 1: Subject inclusion and exclusion criteria

Study Aims

The first aim was to compare changes in propulsion force between limbs under three different walking speeds (SSWS, FWS, and SWS) in individuals with chronic post-stroke hemiparesis and non-neurologically impaired controls.

Hypothesis 1: We hypothesize that as walking speed increases individuals post-stroke will have an increased paretic limb propulsion force percentage, but the magnitude of that change will be smaller than the change in non-neurologically impaired controls.

The second aim was to compare the changes in dorsiflexion angle between legs during the swing phase under three different walking speeds (SSWS, FWS, and SWS) in individuals with chronic post-stroke hemiparesis and non-neurologically impaired controls.

Hypothesis 2: We hypothesize that as walking speed increases individuals with chronic post-stroke hemiparesis will have increased peak ankle dorsiflexion angle during swing phase, but the magnitude of that change will be smaller than the change in non-neurologically impaired controls.

The third aim was to compare the changes in the muscle co-contraction ratio of the tibialis anterior (TA) and gastrocnemius (GA) between legs under three different walking speeds (SSWS, FWS and SWS) in individuals with chronic post-stroke hemiparesis and non-neurologically impaired controls.

Hypothesis 3: We hypothesize that as walking speed increases individuals with chronic post-stroke hemiparesis will have a decreased muscle co-contraction ratio at the times of peak ankle dorsiflexion angle during swing phase and peak propulsion force during stance phase, but the magnitude of that change will be smaller than the change in non-neurologically impaired controls.

Instrumentation

A 12-camera motion analysis system (Vicon, Oxford Metrics Ltd., Oxford, UK) was used to capture kinematic data of the lower extremity at 200 Hz. GRFs were collected at a sampling rate of 2000 Hz using instrumented dual-belt treadmill (Fully Instrumented Treadmill, Bertec Corp., Columbus, OH). Electromyography (EMG) of the gastrocnemius and tibialis anterior were collected using wireless surface EMG electrodes (Delsys Inc., Natick, MA, USA) at 2000 Hz.

Procedures

Participants were tested in a single session at the UNLV biomechanics lab. Upon subject's arrival, their medical history, vitals (blood pressure, resting heart rate and oxygenation), Lower Extremity Fugl-Meyer Assessment, and 2 Minute Walk Test were collected first to assess their function and ability to participate in the study safely. Next, subjects were tested under three different walking conditions (SSWS, FWS, and SWS). Each subject started with their SSWS condition, followed by the FWS or SWS conditions, the order in which alternated between subjects. The SSWS condition always preceded the SWS and the FWS so that the experimental speeds could be calculated from the SSWS (further explained in the Data Collection Procedure section). The alternating order eliminated walking speed bias, in which the first speed may influence the subsequent speed's data.

Biomechanical Marker Definition

To assure appropriate biomechanical marker placements, the participant was fitted with a harness and connected to an overhead support system prior to testing in order to ensure landmark visibility. Once in the harness, biomechanical markers were placed on the individual for the Vicon motion capture system. The reflective biomechanical markers were placed on the following anatomical landmarks: the most distal aspect of the individuals' shoes, 1st and 5th metatarsal heads, medial and lateral malleoli, medial and lateral femoral epicondyles, the joint space between L5–S1 and bilaterally over the greater trochanters, iliac crests, and anterior superior iliac spines (ASIS).[25, 26] In addition, rigid clusters of reflective tracking markers were placed on the lateral surfaces of each subject's thigh, lower leg, and heel of the shoe.[26] After obtaining a static calibration trial, all anatomical markers (with the exception of those attached to the pelvis and the rigid clusters of reflective tracking markers) were removed as is standard with this camera system.[25, 26]

EMG Placement

In addition to the biomechanical markers, wireless surface EMG sensors were applied to the tibialis anterior and medial gastrocnemius muscles to measure muscle activation. The sensors were applied to subjects following the manufacture manual (Delsys Inc., Natick, MA, USA) guidelines as well as related EMG protocols.[11, 27, 28] Initially, the skin overlying muscle bellies was shaved, lightly exfoliated with gauze, and cleaned with alcohol wipes.[28] Initial positioning of the electrodes was determined by palpation of each muscle during a manually resisted dorsiflexion contraction (for tibialis anterior) and a heel raise (for gastrocnemius muscles). Sensors were placed over the muscle bellies perpendicular to the muscle fibers, and secured with tape.[11, 27, 28]

Data Collection Procedure

Prior to participant arrival, researchers ensured the dual-belt treadmill set up/calibration was complete. This set up consisted of preparing the over-head harness above the dual-belt treadmill that the participants were placed in for data collection. The Vicon camera system was masked and calibrated to reduce the chance of erroneous motion capture. The camera system was then calibrated using a calibration wand to direct the camera system to collect data at the treadmill where the subjects were walking. Next, three reflective markers were used to set the orientation of the plane of the treadmill. The force plates within the treadmill were zeroed to ensure no incline or residual GRFs. The force plates in the treadmill were tested by having a researcher provide force through the force plate by stepping on the plates to ensure proper functioning.

Once the participant arrived, anthropometric data was collected, they were placed in the harness and biomechanical marker and EMG placements were completed, as described in prior sections. The participant was then assessed in the overhead harness over the dual belt treadmill to ensure that there was harness slack so that subjects were fully weight bearing. Assessment began with three 30-second static readings of all Vicon biomechanical marker placements. When the static reading was complete, certain anatomical markers were removed for the dynamic trials according to the Vicon Motion System protocol.[25, 26] The participant then began with a practice walk for two minutes. The participant was instructed to walk with their right foot on the right belt and their left foot on the left belt of the dual belt treadmill. Any crossover steps during trials were removed from data after collection. Participants were then asked to verbally request increases or decreases in treadmill speed to achieve a comfortable SSWS. Once the SSWS was maintained for at least 30 seconds, data collection began. Data from their gait cycle were collected for three 30-second trials. The three 30-second trials were continuous without breaks between trials, totaling one minute and 30 seconds of data collection per walking condition. Data collection was electronically stopped for the first trial and instantly started again, without the subject stopping or changing walking speed for the second and third trials. This method was used to maximize the amount of viable data collected and to allow the participants with stroke to achieve an adequate number of gait cycles during data collection without having to stop between trials. Multiple gait cycles were needed due to the high variability of gait in this population.

Once data from the SSWS was collected, either the FWS or SWS was tested next. The order of FWS and SWS alternated between subjects. FWS was determined with an increase from the SSWS by 20% in increments of 0.01 m/s. Participants then cued the researchers as to whether or not they need to go slower in order to prevent falls. The researchers provided verbal encouragement to the participants to achieve their FWS. Then, three 30-second trials were collected under the FWS condition as previously described. The participant was asked to stop if at any point both limbs were simultaneously off of the treadmill, suggesting that participants were running instead of walking. For the SWS condition, the SSWS was decreased by 20%. Researchers ensured that at all times the participants were able to continue walking on the treadmill or the speed was adjusted. Participants verbally requested changes in treadmill speed to achieve a walkable SWS. Three 30-second trials were collected at the SWS and then the session was completed. Participants were given a break if needed between speeds to reduce the chance of fatigue during data collection.

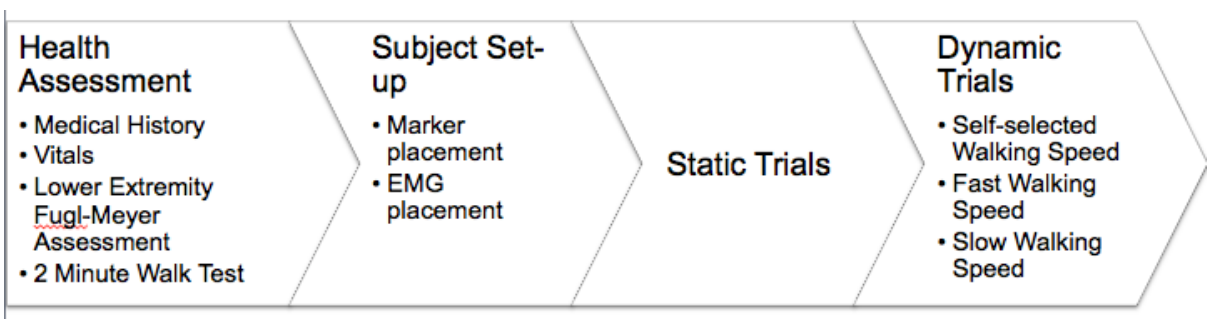

Figure 1: Subject screening and data collection.

Data Analysis

Propulsion force was analyzed using GRFs collected at a sampling rate of 2000 Hz using the instrumented force plates in the dual-belt treadmill. Peak propulsion force and peak braking force was analyzed in stance phase of gait throughout the three 30-second trials. Stance phase of gait was identified by the researchers as the time that the foot was in contact with the ground and presented with a positive GRF as recorded by the force plate.

Peak dorsiflexion angle during swing phase was calculated through use of the Vicon Motion Capture System. Reflective markers were labeled and digitized using Vicon Nexus software. Visual 3D software (C-Motion, Rockville, MD) was used to quantify sagittal plane joint motions of the ankle joint. MATLAB (MathWorks Inc., Natick, MA, USA) software was used to calculate and interpret the peak dorsiflexion angle during swing phase. The swing phase of gait was identified by the researchers as the time that the foot had no contact with the ground and there was no GRF being recorded by the force plate.

The tibialis anterior (TA) and gastrocnemius muscle (GA) co-contraction index was derived from the wireless EMG surface electrodes. Overlap of EMG activity between the medial gastrocnemius and tibialis anterior was used to determine the co-contraction index (GA/TA at peak dorsiflexion, TA/GA at peak toe off). Taking the root mean square of the EMG activation values for each muscle (medial gastrocnemius and tibialis anterior), the co-contraction calculation was made using a previously developed formula.[17]

Measurement Reliability of Current Research

To establish intra- and inter-rater reliability of the Vicon marker placement for the static and dynamic lower extremity measurements, researchers performed repeated measurements of the static and dynamic lower extremity alignment of five non-neurologically impaired subjects that were not included in the data collection on two different days, separated at least a week apart. The same method of marker placement was performed for both the reliability study as well as the actual data collection: each evaluator was responsible for placing the same anatomical markers both between and within subjects throughout the study, with one evaluator placing markers on the pelvis, thigh and knee and the other evaluator placing the markers on the lower leg and foot. Subjects were asked to walk at a SSWS for three 30-second trials in the first session and then the same speed and duration were used during the follow up session. Intraclass correlation coefficients Model 3 ($ICC_{3,3}$) were calculated to assess the test-retest reliability of the two examiners. ICCs were calculated on measurements of peak dorsiflexion angle (DF) during swing phase of gait and peak plantar flexion (PF) angle during stance phase of gait.

	Peak L DF Angle: Swing	Peak R DF Angle: Swing	Peak L PF Angle: Toe Off	Peak R PF Angle: Toe Off
Lower Extremity Alignment $ICC_{3,3}$	0.80	0.81	0.96	0.93

Table 2: Intra-rater reliability for Vicon marker placement

Statistical Analysis

The primary variables examined were 1) peak propulsion force during stance; 2) peak ankle dorsiflexion angle during swing phase; 3) co-contraction index (GA/TA) at peak dorsiflexion during swing phase; 4) co-contraction index (TA/GA) at toe off. We also explored other secondary variables, including 1) peak braking force during stance; 2) braking impulse during stance; 3) propulsion impulse during stance phase; 4) ankle angle at heel strike; 5) ankle angle at toe off; 6) peak plantarflexion angle during swing phase. A 2-factor mixed factorial ANOVA was used to compare each outcome variable between 3 legs and between the 3 speed conditions (FWS, SSWS, SWS). The 3 legs examined were 1) paretic limb of participant post-stroke 2) non-paretic limb of participant post-stroke 3) non-impaired limb of non-neurologically impaired control. When there was a significant interaction effect, we further examined simple main effects using a repeated measures ANOVA with a Bonferroni correction for each limb. Significant main effects and the results of post-hoc t-tests were reported if there were no significant interactions. *A priori* significance was set at $p<0.05$.

RESULTS

Walking speeds

There was not a statistically significant difference for any of the 3 walking conditions between the stroke and control groups using independent t tests ($p > 0.05$). The speeds for the

SSWS, FWS, and SWS conditions in the stroke group were 0.56±0.11 m/s, 0.67±0.17 m/s, and 0.45±0.07 m/s, respectively. The speeds for the SSWS, FWS, and SWS conditions in the control group were 0.72±0.09 m/s, 0.87±0.13 m/s, and 0.58±0.06 m/s, respectively.

Propulsion and Braking Forces

Peak Propulsion Force during Stance

There was a statistically significant main effect of speed for the peak propulsion during the stance phase of gait (p<0.001). Post-hoc analysis showed that the peak propulsion force was significantly greater during the FWS condition (-1.1±0.6 N/Kg) when compared to the SSWS (-0.9±0.5 N/Kg, p<0.001) and SWS (-0.7±0.3 N/Kg, p<0.001) conditions. There was a significant difference between SSWS and SWS (p<0.001). We did not observe a significant main effect of leg (p=0.293) or an interaction between speed and leg (p=0.347) (Table 3 and Figure 2A).

Propulsion Impulse during Stance

There was a statistically significant main effect of speed for the propulsion impulse during the stance of gait (p<0.001). Post-hoc analysis showed that the propulsion impulse was significantly greater during the FWS condition (-18.1±9.3 N/Kg*%) when compared to SSWS (-14.2±8.0 N/Kg*%, p<0.001) and SWS (-13.0±7.1 N/Kg*%, p<0.001) conditions. There was not a statistically significant difference between SSWS and SWS (p=0.835). Additionally, there was not a statistically significant main effect for leg (p=0.236) or an interaction between speed and leg (p=0.620) (Table 3 and Figure 2B).

Peak Braking Force during Stance

We observed a statistically significant main effect of speed for the peak braking force during the stance phase of gait (p<0.001). Post-hoc analysis showed that the peak braking force was significantly greater during the FWS condition (1.2±0.6 N/Kg) when compared to SSWS (0.9±0.5 N/kg, p<0.001) and SWS (0.8±0.4 N/Kg, p<0.001) conditions. Additionally, there was a statistically significant difference between SSWS and SWS ($p<0.001$). There was no statistically significant main effect observed for leg ($p=0.890$) or an interaction between speed and leg ($p=0.893$) (Table 3 and Figure 2C).

Braking Impulse during Stance

There was a main effect of speed ($p<0.001$) for the braking impulse during stance. Post-hoc analysis showed that the braking impulse was significantly greater during FWS (17.7±8.8 N/Kg*%) when compared to SSWS (14.7±7.4 N/Kg*%, $p<0.001$) and SWS (13.3±6.6 N/Kg*%, $p<0.001$). There was a significant difference between SSWS and SWS ($p<0.001$). There was not a main effect found for leg ($p=0.395$) or an interaction between speed and leg ($p=0.316$) (Table 3 and Figure 2D).

		Slow Walking Speed (SWS)	Self-selected Walking Speed (SSWS)	Fast Walking Speed (FWS)
Peak Propulsion Force (N/Kg)	Paretic	-0.5±0.3	-0.7±0.5	-0.8±0.6
	Non-paretic	-0.7±0.4	-0.9±0.5	-1.1±0.5
	Non-impaired	-0.8±0.4	-1.1±0.5	-1.3±0.6
	Overall	**-0.7±0.4**	**-0.9±0.5‡**	**-1.1±0.6†‡**
Propulsion Impulse (N/Kg* %)	Paretic	-9.3±6.0	-10.8±7.1	-13.1±9.5
	Non-paretic	-15.1±8.6	-14.4±9.3	-20.5±9.2
	Non-impaired	-14.6±5.7	-17.2 ± 7.2	-20.7±8.1
	Overall	**-13.0±7.1**	**-14.2 ± 8.0**	**-18.1±9.3†‡**
Peak Braking Force (N/Kg)	Paretic	0.8±0.4	0.9±0.4	1.2±0.6
	Non-paretic	0.7±0.4	0.8±0.6	1.1±0.7
	Non-impaired	0.8±0.4	1.0±0.5	1.3±0.6
	Overall	**0.8±0.4**	**0.9±0.5‡**	**1.2±0.6†‡**
Braking Impulse (N/Kg* %)	Paretic	14.8±5.6	15.0±6.7	18.4±6.9
	Non-paretic	10.4±6.9	11.7±8.2	14.2±10.2
	Non-impaired	14.6±7.1	17.4±7.1	20.5±9.0
	Overall	**13.3±6.6**	**14.7±7.4‡**	**17.7±8.8†‡**

Table 3: The comparisons of peak propulsion force, propulsion impulse, peak braking force, and braking impulse during the stance phase of gait between the SWS, SSWS, and FWS conditions across 3 legs (paretic limb, non-paretic limb, and non-impaired limb).

Positive values indicate braking and negative values indicate propulsion forces.
† indicates a significant difference from the SSWS condition ($p<0.001$) and ‡ indicates a significant difference from the SWS condition ($p<0.001$) using a 2-factor ANOVA with repeated-measures and a post-hoc t-test.

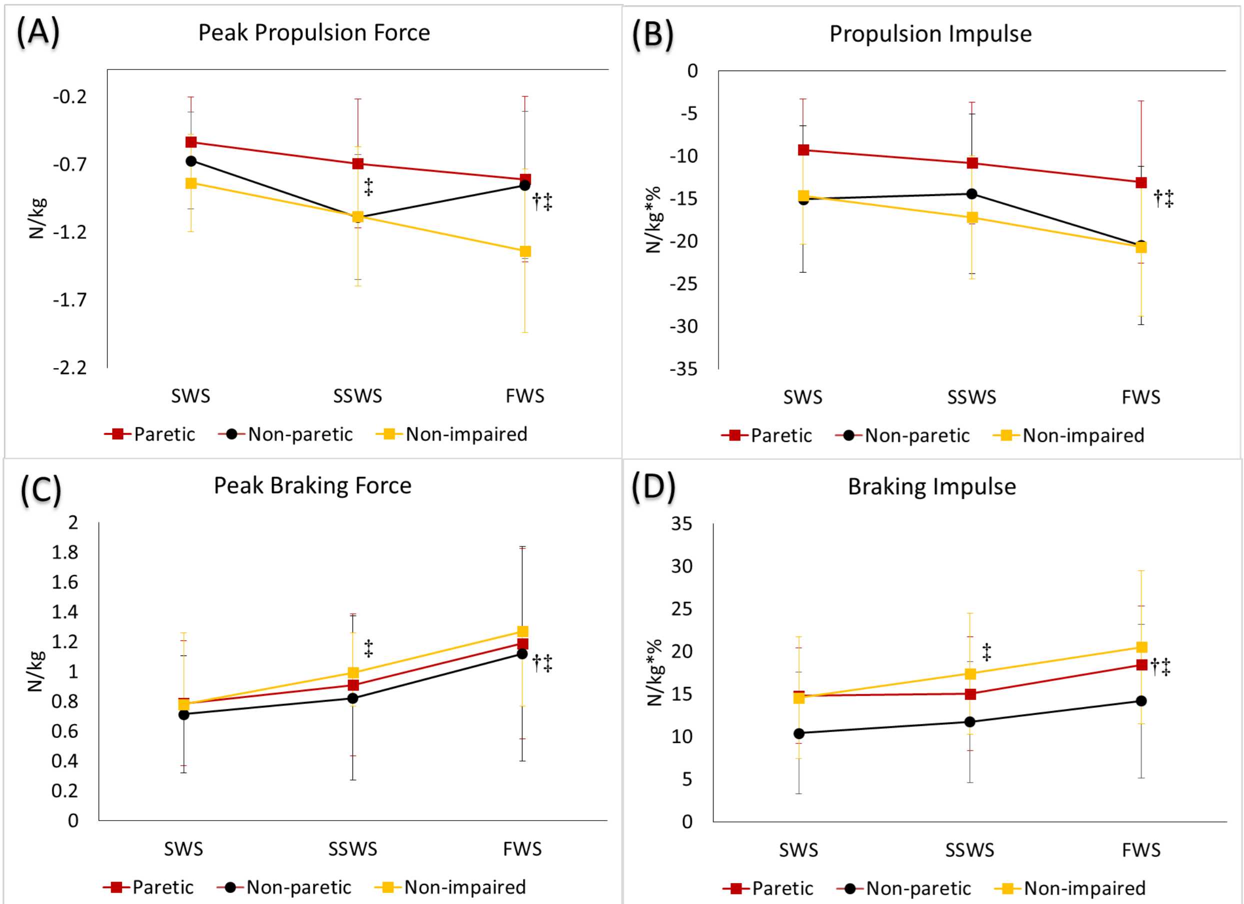

Figure 2: (A) peak propulsion force, (B) propulsion impulse, (C) peak braking force, and (D) braking impulse during the stance phase of gait in SWS, SSWS and FWS conditions across 3 legs (paretic limb, non-paretic limb, and non-impaired limb). The error bars indicate the standard deviations.

† indicates a significant difference from the SSWS condition (p<0.001) and ‡ indicates a significant difference from the SWS condition (p<0.001) using a 2-factor ANOVA with repeated-measures and a post-hoc t-test.

Kinematics

Ankle Angle at Heel Strike

There was a speed main effect (p=0.016) for the ankle angle at heel strike. Post-hoc analyses showed that ankle angle was significantly greater during FWS (0.4±5.2°) compared to the SSWS (1.2±5.6°, p<0.001) and SWS (1.2±5.6°, p<0.001). There was not a significant difference between SSWS and SWS for the ankle angle at heel strike (p=0.519). Neither a main effect for leg ($p=0.163$) nor an interaction effect between the leg and speed ($p=0.569$) was found for the ankle angle at heel strike (Table 4 and Figure 3A).

Ankle Angle at Toe off

Our results revealed a speed main effect ($p<0.001$) for the ankle angle at toe off. Post-hoc analysis showed that the ankle angle was significantly greater during FWS (-5.3±7.3°) compared to SSWS (-3.9±7.2°, $p<0.001$) and SWS (-2.3±5.9°, $p<0.001$). There was a significant difference between SSWS and SWS for the ankle angle at toe off ($p<0.001$). Neither a main effect for leg ($p=0.494$) or between leg and speed ($p=0.878$) was found for the ankle angle at toe off (Table 4 and Figure 3B).

Peak Dorsiflexion Angle during Swing

There was a main effect ($p=0.011$) of speed for the peak dorsiflexion angle during swing. Post-hoc analysis showed that the dorsiflexion angle was significantly greater during FWS (4.6±4.3°) compared to SSWS (5.6±4.8°, $p<0.001$) and SWS (5.3±4.6°, $p<0.001$). There was not a significant difference between SSWS and SWS for the peak dorsiflexion during swing ($p=0.295$). Neither a main effect for leg ($p=0.156$) nor an interaction between speed and leg ($p=0.835$) was found for peak dorsiflexion angle during swing (Table 4 and Figure 3C).

Peak Plantarflexion Angle during Swing

There was a main effect ($p<0.001$) of speed for the peak plantarflexion angle during swing. Post-hoc analysis showed that the plantarflexion angle was significantly greater during FWS (-6.1±7.4°) when compared to SSWS (-4.8±7.4°, $p<0.001$) and SWS (-3.5±6.3°, $p<0.001$). There was a significant difference between SSWS and SWS ($p<0.001$). There was not a main effect found for leg ($p=0.562$) or an interaction between speed and leg ($p=0.871$) (Table 4 and Figure 3D).

		Slow Walking Speed (SWS)	Self-selected Walking Speed (SSWS)	Fast Walking Speed (FWS)
Ankle Angle at Heel Strike (°)	Paretic	-1.9±5.7	-2.0±5.1	-2.3±4.7
	Non-paretic	3.7±5.5	3.9±5.9	2.5±5.5
	Non-impaired	1.8±4.8	1.8±4.9	1.1±4.8
	Overall	**1.2±5.6**	**1.2±5.6**	**0.4±5.2†‡**
Ankle Angle at Toe off (°)	Paretic	-1.7±2.6	-3.1±3.3	-4.1±4.2
	Non-paretic	-0.8±5.8	-2.1±6.3	-3.8±7.1
	Non-impaired	-4.4±8.3	-6.8±10.4	-7.9±9.8
	Overall	**-2.3±5.9**	**-3.9±7.2‡**	**-5.3±7.3†‡**
Peak Dorsiflexion Angle during Swing (°)	Paretic	2.7±3.5	3.4±2.9	2.5±3.6
	Non-paretic	7.8±3.7	8.1±4.8	6.6±3.2
	Non-impaired	5.2±5.4	5.4±5.7	4.5±5.2
	Overall	**5.3±4.6**	**5.6±4.8**	**4.6±4.3†‡**
Peak Plantarflexion Angle During Swing (°)	Paretic	-3.8±5.1	-4.8±4.7	-5.7±5.1
	Non-paretic	-1.4±5.5	-2.5±6.7	-4.2±7.1
	Non-impaired	-5.2±8.3	-6.9±10.3	-8.4±9.8
	Overall	**-3.5±6.3**	**-4.8±7.4‡**	**-6.1±7.4†‡**

Table 4: The comparisons of ankle kinematics between paretic, non-paretic, and non-impaired limbs across the 3 walking speeds.

Positive values indicate dorsiflexion and negative values indicate plantarflexion.
† indicates a significant difference from the SSWS condition ($p<0.001$) and ‡ indicates a significant difference from the SWS condition ($p<0.001$) using a 2-factor ANOVA with repeated-measures and a post-hoc t-test.

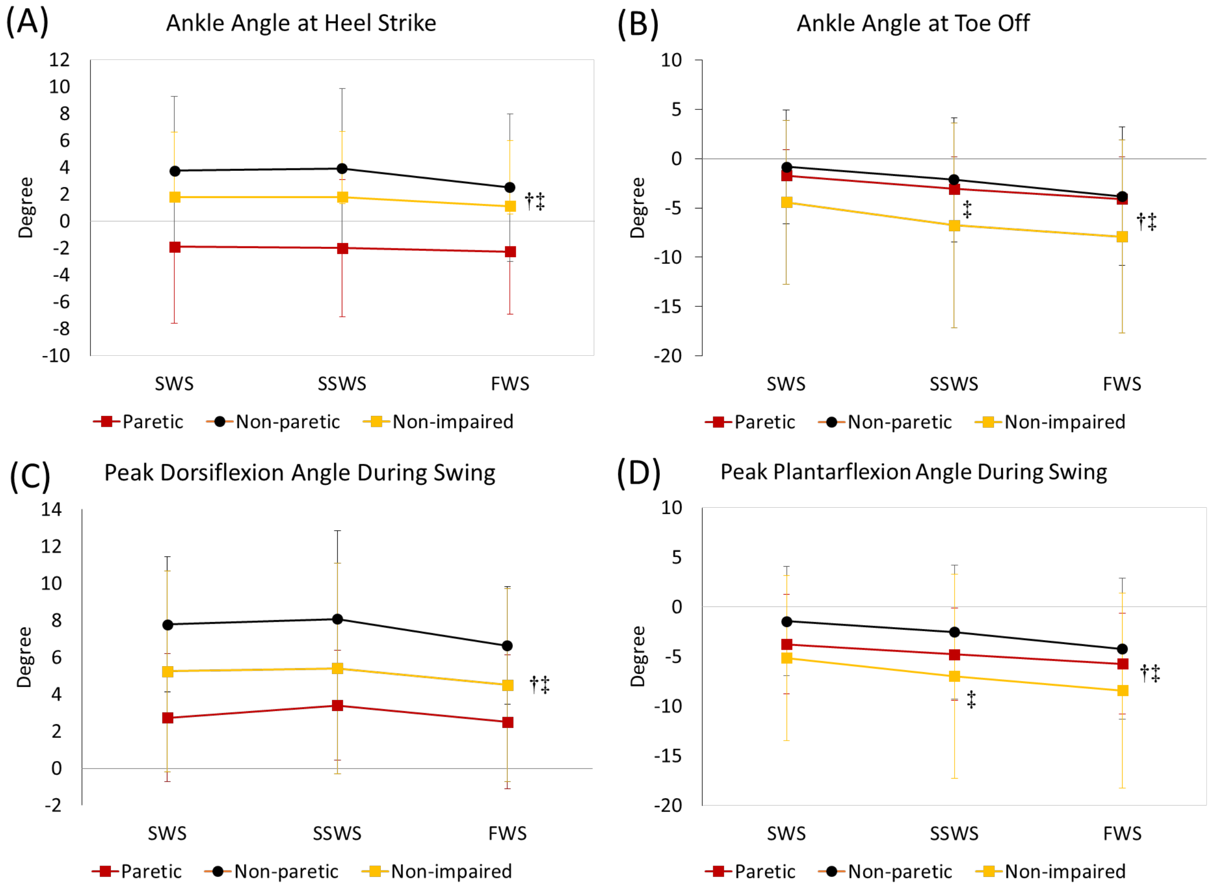

Figure 3: (A) ankle angle at heel strike, (B) ankle angle at toe off, (C) peak dorsiflexion angle during swing, and (D) plantarflexion angle during swing of gait in SWS, SSWS and FWS conditions across 3 legs (paretic limb, non-paretic limb, and non-impaired limb). The error bars indicate the standard deviations.

† indicates a significant difference from the SSWS condition ($p < 0.00$) and ‡ indicates a significant difference from the SWS condition ($p<0.001$) using a 2-factor ANOVA with repeated-measures and a post-hoc t-test.

Co-Contraction Index

Co-Contraction Index at Peak Dorsiflexion during Swing (GA/TA)

The 2-way ANOVA with repeated measures showed that there was not a speed main effect ($p=0.233$) for the co-contraction index at peak dorsiflexion during swing. There was not a main effect found for leg ($p=0.554$) or an interaction between speed and leg ($p=0.604$) (Table 5 and Figure 4A).

Co-Contraction Index at Toe Off (TA/GA)

The 2-way ANOVA with repeated measures showed that there was a trending main effect (p=0.082) for the co-contraction index at toe off. There was not a main effect found for leg (p=0.784) or an interaction between speed and leg (p=0.678) (Table 5 and Figure 4B).

		Slow Walking Speed (SWS)	Self-selected Walking Speed (SSWS)	Fast Walking Speed (FWS)
Co-Contraction Index at Peak Dorsiflexion During Swing (%)	Paretic	42.4±7.1	39.7±12.5	44.3±9.6
	Non-paretic	36.0±14.9	37.6±16.5	37.7±12.3
	Non-impaired	44.0±3.9	37.5±10.1	43.6±4.4
Co-Contraction Index at Toe Off (%)	Paretic	41.5±9.9	38.5±15.7	40.6±13.1
	Non-paretic	40.0±14.3	36.9±13.8	36.6±15.6
	Non-impaired	46.5±3.4	36.9±12.5	42.9±11.9

Table 5: The comparisons of co-contraction index between paretic, non-paretic, and non-impaired limbs across the 3 walking speeds.

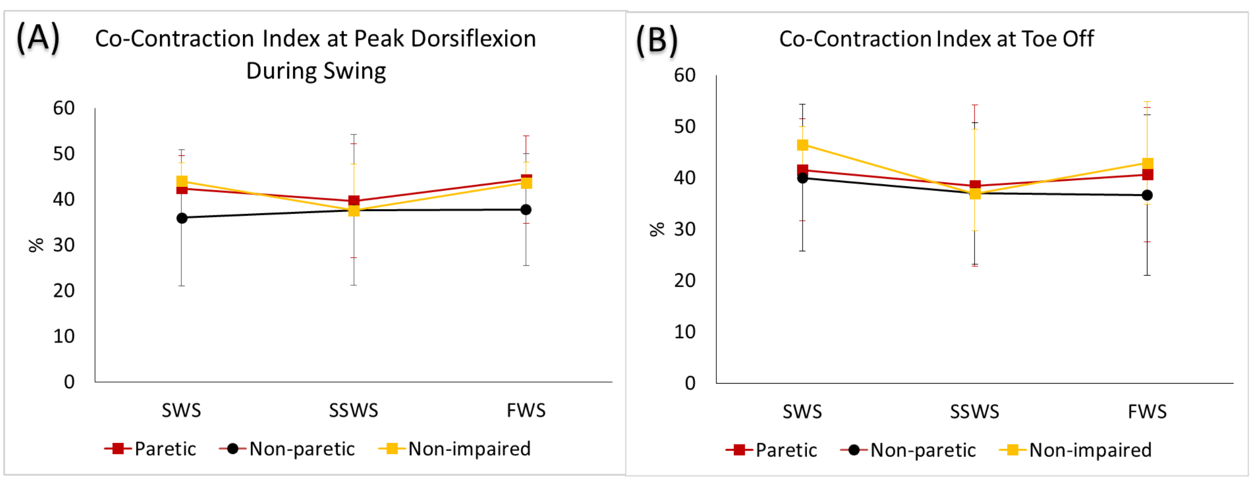

Figure 4: (A) co-contraction index at peak dorsiflexion during swing and (B) co-contraction index at peak dorsiflexion during swing in SWS, SSWS and FWS conditions across 3 legs (paretic limb, non-paretic limb, and non-impaired limb). The error bars indicate the standard deviations.

<u>DISCUSSION</u>

Propulsion and Braking Force

The purpose of this study was to find out how propulsion and braking forces changed in a paretic limb at faster walking speeds compared to slow and self-selected speeds. We observed that the peak braking forces and the peak propulsion forces during stance increased significantly under the FWS condition compared to the SWS and SSWS conditions.[14] This result was similarly seen in the non-paretic leg as well as the legs of the non-neurologically impaired controls. The results from our study agree with previous studies that among those post-stroke, faster walking speeds produced a greater peak propulsion force during stance and a greater peak braking force in stance, compared separately to slower and self-selected speeds.[14, 29, 30]

In addition, the propulsion and braking impulses during the FWS condition were greater compared to the SWS and SSWS conditions. This suggests that there was greater application of force throughout the time of propulsion and the time of braking during the FWS condition. Furthermore, it suggests that FWS was the primary factor that allowed for these increases in propulsion and braking impulses during the stance phase of gait. The results from our study agreed with previous research that a faster walking speed could generate a greater impulse of propulsion and greater impulse of braking during stance in the paretic limb compared to slower and normal walking speeds, which can be similarly seen in those who have not had a stroke.[29]

Kinematics

One aim of this study was to determine the role of gait speed in changing lower extremity kinematics during gait. We found a relationship between speed and ankle angles at heel strike, toe off, and during swing; people post-stroke were able to improve their walking kinematics when walking at faster speeds. The increased dorsiflexion ankle angle at heel strike and increased plantarflexion ankle angle in toe off in the FWS condition demonstrates an improved capability to absorb shock and generate larger propulsion forces, when compared to the SSWS and SWS.[14] This was similarly seen in the data of our controls. Compared to SWS and SSWS, FWS allowed for an increased peak dorsiflexion angle during swing, which increases clearance of the advancing lower extremity.[14] This improvement in the paretic limbs was also seen similarly in the control group, indicating that speed is a primary factor in changing gait kinematics across populations.

During FWS there was also a significantly different peak plantarflexion angle during swing phase of the paretic limbs compared to the angles observed during SSWS and SWS. This was also seen in the non-paretic limb and the limb of the controls. This implies that there was an increased ability to generate propulsion forces from the increased ankle plantarflexion. Previous studies did not focus on collecting data at various gait speeds, but our results do agree with previous studies that faster walking speeds do generate significantly different ankle dorsiflexion and plantarflexion angles during gait in the paretic limb.[7]

Co-contraction Index

The co-contraction index between the gastrocnemius muscle and the tibialis anterior was another important aspect of our study. We did not observe a significant difference between the co-contraction index of the limb during swing when comparing gait speed conditions, suggesting that muscle activation was not distributed to the paretic limbs in a similar manner to the non-paretic limbs and the control limbs.[11] It may also suggest that the co-contraction index of the gastrocnemius and the tibialis anterior is not a significant contributing factor to the ankle angles during the swing phase of gait. In addition, the co-contraction index at toe off was not significantly different in the FWS condition compared to the SWS and SSWS, but it was trending toward a decreased co-contraction index in the paretic limb. This does suggest that during toe off, a decreased co-contraction ratio between the two muscles may play a role in contributing to the increased forces during stance.[11]

Limitations and Future Directions

There are a few limitations to our study. We had strict inclusion and exclusion criteria including the ability to complete a 2MWT without an assistive device. The definition of assistive device included any walker, ankle-foot orthoses, or canes. This limited the sample to high-functioning individuals. Because of this, our results may not generalize to the entire stroke population. Furthermore, people post-stoke have a high degree of variability in their function. Our study excluded those with strokes involving the cerebellum. Cerebellar strokes inherently have varied, unpredictable effects on muscle movement and in decreasing the variability of our subjects, our results are further limited to individuals with movement deficits consistent with cortical strokes. The effects of stroke are incredibly varied in degree and presentation. Future research will have to explore the effects of walking speed on individuals who have experienced strokes in different areas of the brain in order for the results to benefit more people.

The application of equipment used may have also introduced limitations in the study. Although researcher reliability was tested, biomechanical marker was dependent on the researcher to determine landmarks, which introduces human error. Selection of the gait cycle for data analysis was also dependent on researchers, in which human error can be introduced. Due to the subjective input of the researchers, the exact instance of each gait cycle cannot be guaranteed and may differ between subjects; error in labeling each instance may result in incorrect calculations during data analysis. In addition, the EMG data was collected using surface electrodes which lead to increased likelihood of interference with surrounding equipment and potential to affect the data collected for the co-contraction index.[27, 28] The dynamic nature of walking as well as surrounding equipment may give off extraneous vibrations and noise that interfere with EMG electrode readings, creating motion artifacts that may have altered the accuracy of the data.[31]

Another possible limitation could be the choice in percentage increase or decrease from the self-selected walking speeds. The smaller magnitude of change in the kinematics/kinetics variables could be attributed to the smaller increase and decrease in walking speed and therefore not produce statistically significant results.

Future studies can build upon the results found in this study by focusing on people with different types of strokes and finding specific dosing requirements in gait speed to induce changes in walking kinematics. This study shows that even though individuals post-stroke show impairments in propulsion and braking forces during stance and a reduced dorsiflexion angle during swing, they are still able to increase them significantly, similar to healthy controls by walking at a faster speed. With future research designating specific dosing requirements to individuals with specific strokes, treatments and interventions can be developed to help individuals' post-stroke better their walking capabilities.

Additionally, it is important to note that after statistical analysis there was not a significant interaction found between the walking speeds (FWS, SWS, SSWS) and the groups (controls and post-stroke) for any of the dependent variables. In other words, there was no influence between the different walking speeds, and which group the limb was analyzed from (control or post-stroke) on the propulsion and braking forces, the ankle kinematics, and the co-contraction index.

CONCLUSION

Our study supports the idea that ambulating at faster speeds plays a role in generating better walking mechanics during gait. Similar to non-neurologically impaired individuals, with a faster walking speed, there was increased peak propulsion and braking forces in the paretic limb of individuals with post-stroke hemiparesis. There is also an increased dorsiflexion angle during swing phase, as well as an increased plantarflexion angle, with faster walking speeds when compared to slower and self-selected walking speeds. We did not find decreased coactivation between the medial gastrocnemius and the tibialis anterior as seen in previous studies, but there was a trend in our data toward a change in co-contraction index during toe off. Our results demonstrate that with faster walking speeds, more pronounced gait mechanics, such as angulation of the ankle, can be produced which may influence how therapy can address improving gait in those experiencing residual effects from strokes.

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