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Neuromuscular Adaptations Following Slope Walking in Individuals Post-Stroke

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NEUROMUSCULAR ADAPTATIONS FOLLOWING SLOPE WALKING
IN INDIVIDUALS POST-STROKE

By

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A doctoral project submitted in partial fulfillment
of the requirements for the

Doctor of Physical Therapy

Department of Physical Therapy
School of Allied Health Sciences
Division of Health Sciences
The Graduate College

University of Nevada, Las Vegas
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ABSTRACT

Background: The excitability of the H-reflex pathway in the non-impaired nervous system can be augmented by altering the different parameters of a walking task, specifically slope. We sought to examine the adaptations in soleus H-reflex excitability and foot force control following an acute bout of upslope or downslope treadmill walking in people post-stroke compared to those who are non-impaired.

Methods: We recruited 12 individuals with chronic post-stroke hemiparesis and 9 age-similar non-neurologically impaired individuals. Each subject was tested over 2 sessions separated by at least 7 days. For each session, subjects walked at a self-selected walking speed on an instrumented treadmill for 20 minutes under a level and then an upslope condition, or a level and then a downslope condition, with at least an hour rest between the conditions. The vertical component of ground reaction force was used to determine the stance and swing phase of the gait cycle. Peak propulsion and braking forces were analyzed offline for the first (T1) and last minute (T20) of each walking condition to examine adaptations in foot force control. Soleus H-reflexes (H_{max}) were tested before and after each walking condition in the paretic legs of the post-stroke group and the right legs of the control group. To ensure consistency, a control M wave (M_{max}) preceding the H_{max} was kept constant across all conditions for each subject. Peak to peak amplitudes of the maximal H-reflexes and maximal M waves were measured offline and expressed as an H_{max}/M_{max} ratio.

Results: The paretic legs generated higher propulsion force during upslope (11.75 ± 1.04 %BW), but comparable propulsion forces during downslope, when compared to level walking (6.14 ± 0.67 %BW). However, we did observe statistical significance in main effect for slope in paretic ($F(2,22)=33.178$, $p < 0.001$), non-paretic ($F(1.144, 12.585)=23.246$, $p < 0.001$) and non-impaired legs ($F(1.137, 10.998)=22.766$, $p < 0.001$). Pairwise comparisons between slope types indicated that on average, peak braking forces were higher when walking downslope and lower when walking upslope, when compared

to level walking. We observed an overall change in H_{\max}/M_{\max} ratio following 20 minutes of walking, and the change was different for post-stroke compared to control group, as suggested by the significant interaction between time and group ($F(1,19)=16.84$, $p=0.001$).

Conclusion: Our observations suggest that when the biomechanics of the walking task is altered, through adjusting the slope of the walking surface, paretic legs exhibit increased propulsion forces during upslope walking. Paretic propulsive forces were greatest in the upslope condition and lowest in the downslope condition. Regardless of group, individuals had greatest braking forces during the downslope condition and lowest during the upslope condition. We believe, based on current studies, that increased paretic propulsion forces in the upslope condition may be due to the increased difficulty of the environmental condition. In the level condition, spinal circuits in the stroke-impaired nervous system are trending towards adaptations similar to the non-impaired nervous system, such that the H_{\max}/M_{\max} ratios were depressed. However, in the more challenging upslope condition, adaptations of the paretic soleus H-reflexes were impaired such that the H_{\max}/M_{\max} ratios were trending towards elevated. Future studies will examine the optimal walking duration and degree of slope to induce neural adaptations, as well as determine any long-term retention of plasticity.

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INTRODUCTION

According to the American Heart Association 795,000 people experience a new or recurrent stroke every year with approximately 610,000 of these being first events and 185,000 are recurrent stroke events (Mozaffarian et al., 2015). One of the primary risk factors for stroke recurrence is physical inactivity, and specifically, a lack of walking, as walking is an essential component of community dwelling, including socialization and safety (Benjamin et al., 2017). Following a stroke, people fear dependency on other individuals more than any other factor, and independent walking has been found to be the greatest protector against dependency (Solomon, Glick, Russo, Lee, & Schulman, 1994).

One of the causes for this dependency due to walking following stroke is impaired gait performance, with diminished strength and inappropriately graded muscle activity as the primary contributors (Olney & Richards, 1996). Diminished strength has been associated with post-stroke hemiparesis, decreased walking speed, and longer stance phase on the paretic limb (Teixeira-Salmela, Nadeau, McBride, & Olney, 2001). Stiff joints on the paretic limbs of individuals post-stroke causes altered magnitude of movement as well as simplified muscle contractions (Wang, Li, Yue, Yin, & Wei, 2017). However, currently available measures such as electromyography (EMG) give an incomplete assessment of the coordinated output of the paretic leg while walking (Bowden, Balasubramanian, Neptune, & Kautz, 2006). For this reason, the anterior-posterior (A-P) ground reaction forces (GRF) may represent a more appropriate method of measuring the contribution of the paretic leg to the coordinated task of forward propulsion during walking (Bowden et al., 2006). Exaggerated dorsiflexor muscle activity of the paretic limb may counteract the effects of the plantarflexors by offloading the leg and interfering with the paretic limb's ability to generate appropriate A-P ground reaction forces (Turns, Neptune, & Kautz, 2007). In individuals post-stroke, the paretic leg generates less propulsion and greater braking forces when compared to the non-paretic leg and non-impaired legs (Bowden et al., 2006).

In a non-stroke impaired population during uphill slope walking, propulsive forces increase, and braking forces decrease in comparison to level walking (A.N. Lay, Hass, & Gregor, 2006). During downslope walking, propulsive forces decrease and braking forces increase compared to level walking (A.N. Lay et al., 2006). Although there is literature showing that force modulation is possible in people post-stroke, the short-term adaptive paradigm has not yet been examined. It is known that the impaired nervous system is able to adapt to a short bout of exercise and correctly activate lower extremity muscles by being exposed to environmental demands (Alibiglou & Brown, 2011). Short-term exposures to these somatosensory inputs are able to temporarily generate a more normal motor output post-stroke (Alibiglou & Brown, 2011). The intent of our study is to assess the extent of appropriate neuro-modulation in an acute setting through both ground reaction forces and H-reflex adaptation. Due to literature supporting that treadmill walking and over-ground walking both demonstrate similar muscle activity of the soleus (SOL), rectus femoris, vastus medialis and tibialis anterior muscles (Paul, Bhattacharyya, Chatterjee, & Majumdar, 2016), we will be assessing acute adaptations via treadmill walking to assure consistency and reliability.

In addition to diminished strength, inappropriately graded muscle activity contributes to impaired gait in individuals post-stroke (Mochizuki et al., 2009). This is demonstrated through an exaggerated reflex and muscle spasticity, which is evidenced by an impaired Hoffmann Reflex (H-reflex). The H-reflex is typically measured through the H-reflex/M-wave (H/M) ratio due to the high variability of the H-reflex response between individuals, both impaired and non-impaired. The H-reflex is the estimate of the number of motor neurons an individual is capable of activating in any given state, whereas the M-reflex represents activation of the entire motor neuron pool and, therefore, its maximum muscle activation (Palmieri, Ingersoll, & Hoffman, 2004). This concept, comparable to the clinical presentation of spasticity, is why H-reflexes have been shown to be larger in individuals post-stroke when compared to non-impaired individuals both at rest (Alwan, 2014) and during locomotion

(Schindler-Ivens et al., 2008). Changing the parameters of walking by altering the slope has been shown to affect both the H-reflex excitability and A-P ground reaction forces (Carda, Invernizzi, Baricich, Cognolato, & Cisari, 2013; M. J. Sabatier et al., 2015). It has been shown that both downslope and level walking can lead to a decrease in reflex excitability as measured by the H/M ratio in non-impaired individuals, with the ratio decrease being more significant for downslope walking (M.J. Sabatier et al., 2015). This is likely because downslope walking requires more cortical activation due to eccentric muscle contractions (Fang, Siemionow, Sahgal, Xiong, & Yue, 2001). On the other hand, upslope walking has been shown to cause both an increase in the H/M ratio (M.J. Sabatier et al., 2015) and similar excitability to that of downslope (Garnier, Lepers, Stapley, Papaxanthis, & Paizis, 2017). Downslope walking is also less metabolically demanding with a smaller cardiovascular response than level or upslope walking (Navalta, Sedlock, & Park, 2004), so if the H/M ratio is reduced in subjects post-stroke following downslope walking, it may be a potential option for future exercise prescription.

Although short-term adaptations of the A-P ground reaction forces and H/M ratios following sloped walking have been evaluated and studied in non-impaired individuals, it has not yet been evaluated in individuals post-stroke. Assessment of downslope, level, and upslope walking and measurement of the H/M ratio, as well as the A-P ground reaction forces, additionally has yet to be assessed while comparing non-impaired to stroke-impaired populations. Therefore, the purpose of the study is to investigate the short-term neurological adaptations in individuals post-stroke after a single bout of treadmill walking. 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 slope walking.

To address this purpose, we developed the following two aims:

METHODS

Subjects

Table 1. Subject Demographics

	Post-Stroke		Controls	
Total number (n)	12		9	
Female	4		5	
Male	8		4	
Age (mean)	70.5		59.4	
Average walking speed (m/s)	0.63		1.04	
Years post stroke (mean)	7.4		-	
Average Motor Function FM Score (Out of 34)	28.4		-	
Hemiparetic/Stimulator limb	10 (L)	2 (R)	2 (L)	7 (R)

Subject demographics are shown in Table 1 above. Subjects in the post-stroke group were included in the study if they had a single, unilateral, cortical or subcortical stroke, more than 6 months post-ictus. Subjects were excluded if they had other neurological conditions, used anti-spastic medication, had cognitive or affective disorders, expressive or receptive aphasia, severe concurrent medical problems such as severe cardiac disease, history of poorly controlled diabetes, active cancer, etc., orthopedic conditions affecting the legs, history of hip or knee replacement, or peripheral nerve injury in the lower limb. Prior to participation, each subject received written and verbal information about the experimental procedures before signing an approved consent form. The protocol was approved by the Institutional Review Board at the University of Nevada Las Vegas.

Instrumentation

Foot forces were measured using a treadmill with Vicon force plates (Fully Instrumented Treadmill, Bertec Corporation, Columbus, OH) (Vicon, Culver City, CA). EMG signals of the SOL muscle

were captured from the test leg using a wireless surface EMG system (Trigno Wireless System, Delsys Inc., Natick, MA). SOL H-reflexes and M-waves were elicited using a constant current stimulator and isolation unit (DS7A, Digitimer Ltd, Welwyn Garden City, UK) with a square pulse stimulus of 1ms duration, a current range of 50 μ A~200 μ A and total output capacity of 400V. GRFs and surface EMG activity during walking were sampled at 2000Hz and H-reflex data was sampled at 3200Hz.

Procedure

Each subject was tested over two sessions separated by at least seven days. Three subjects in the control group and two subjects in the post-stroke group were separated by more than seven days. One subject in the control group and four subjects in the post-stroke group were separated by exactly seven days. Each session included a level walking condition followed by either upslope or downslope walking. Additionally, the first session included a screening, which included consent, years since stroke, medical history including: medications and previous LE surgeries, vitals: blood pressure and heart rate, weight, Fugl-Meyer lower extremity assessment (for the post-stroke group), baseline H/M ratio, and assessment of the subject's ability to walk on a treadmill at a self-selected walking speed without any assistive devices. Following the screening, the subjects' skin was prepped for EMG application. Surface EMG electrodes were placed bilaterally over the Tibialis anterior muscle (TA) and medial and lateral muscle bellies of the Gastrocnemius muscles to record muscle activity during walking. A surface EMG was applied to the subjects' SOL muscle allowing capture of the H-reflex recruitment curves prior to and following each of the walking conditions. To ensure consistent background EMG activity of the target SOL muscle and inhibition from antagonistic TA muscle, EMG activity was monitored 100ms prior to the electrical stimulus across trials. For all H-reflex recordings, subjects were tested in an upright standing position with arms relaxed by sides. A stimulating electrode was placed over the Tibial nerve at the Popliteal fossa on the test leg. To record the H-reflex recruitment curve, the intensity of the electrical

stimulus was gradually increased in small increments, until a maximal M-wave was attained. Pre-assessment H-reflex and vitals were obtained prior to each 20-minute bout of walking. During the 20-minute bout of walking, GRFs and EMG activity were collected in 1-minute trials every other minute to yield 10 trials per condition. Post-test H-reflex and vitals were then recorded. This was followed by a 1-hour break. Using pseudo randomization to determine the order of the second condition, subjects then completed a pre-test assessment and a second 20-minute session of either upslope (US) or downslope (DS) walking. Following pseudo randomization of each group, the order was alternated for each proceeding subject to ensure an equal amount of upslope and downslope walking was performed throughout the study. The US walking was measured at 5 degrees incline, and the DS walking was measured at 5 degrees decline for both the post-stroke and control groups. Immediately following any condition of walking, H-reflex and vitals were recorded.

Before treadmill walking in the second session, subjects first had their vitals and H-reflex recruitment curves assessed. Each subject completed LS walking for 20 minutes to achieve a second baseline. Immediately following LS walking, vitals and H-reflex recruitment curves were assessed again. Subjects were then given a 1-hour break, then performed a second 20-minute session of either US or DS walking depending on whichever was not assigned during the first session. Similarly, vitals and H-reflex recruitment curves were assessed prior and immediately after each of the walking conditions. M_{max} was assessed as a constant across both days in order to demonstrate that electrodes were in the same position. Utilizing a level walking condition on both days served to establish this as a control as well.

	SCREENING	20-min LS Walk	1-hour break	20-min US/DS Walk
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INITIAL VISIT	<ul style="list-style-type: none"> • Consent • History • Vitals • Fugl Meyer (post-stroke only) • H/M ratio • Walking demonstration 	Pre/Post-Test: <ul style="list-style-type: none"> • H/M ratio • Vitals • GRF and EMG taken for 1-min increments every other min to yield 10 trials per condition 		<ul style="list-style-type: none"> • Pseudo-randomization Pre/Post-Test: <ul style="list-style-type: none"> • H/M ratio • Vitals • GRF and EMG taken for 1-min increments every other min to yield 10 trials per condition
SECOND VISIT		Pre/Post-Test: <ul style="list-style-type: none"> • H/M ratio • Vitals • GRF and EMG taken for 1-min increments every other min to yield 10 trials per condition 		<ul style="list-style-type: none"> • Dependent on initial visit Pre/Post-Test: <ul style="list-style-type: none"> • H/M ratio • Vitals • GRF and EMG taken for 1-min increments every other min to yield 10 trials per condition

Table 2. Procedures

Data analysis

All forces, EMG and H-reflex data were processed in Vicon and analyzed offline using custom MATLAB programs (MATLAB, Proton Station ON NOC 1LO, Canada). For each trial, peak-to-peak amplitudes of all H-reflexes and the maximal M waves were calculated using a custom MATLAB analysis program. For each condition, the 5 maximal H-reflexes and maximal M wave amplitudes were averaged. Ground reaction forces (GRF) were filtered with a 20 Hz low-pass, 2nd order, zero-lag Butterworth filter. The vertical component of GRF was used to determine the stance phase of gait cycle. The time point where the vertical GRF exceeded zero newton’s (N) and remained for a continuous period of at least 50 ms was denoted as heel contact. Subsequently, the time point where the vertical GRF reached zero N and remained for a continuous period of at least 50 ms, was denoted as toe off. The period from heel

contact to toe off was considered the stance phase. Subsequently, horizontal forces in the anterior-posterior direction at the corresponding time points were used to determine the propulsive and braking forces for each gait cycle. For each subject, we analyzed the data from the first and final 60 seconds of walking for each condition. For each 60-second trial, these forces of the first and last five stance phases were removed, and the mean propulsive and braking force were averaged over the rest of stance phases and normalized to the individual's body weight. The same procedure was repeated for GRF data from the other force platform.

Statistical analysis

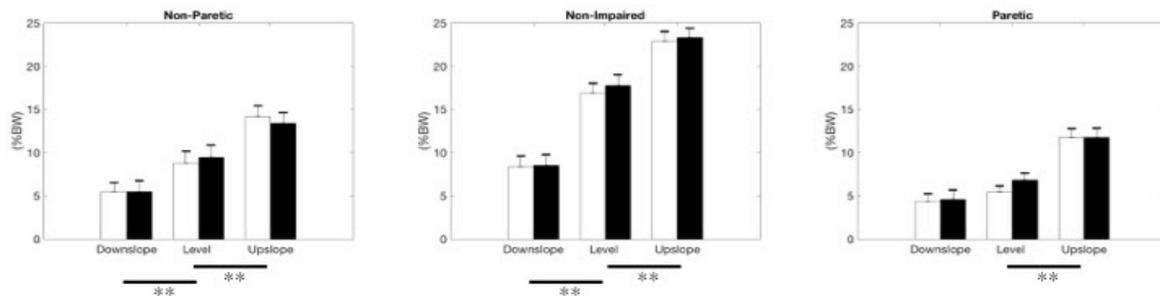
To test the hypothesis that the change in H_{max}/M_{max} ratio and foot forces were different in the post-stroke versus control group, a three-way mixed ANOVA with repeated measures was used. For comparison of foot forces, paretic and non-paretic foot forces of the post-stroke group were contrasted to compared with the right foot forces of the control group as a between-subjects factor (group). The independent variables include time (2 levels: pre and post treadmill walking), and slope (3 levels: level, downslope, and upslope) as within-subjects factors. Where the Mauchley's test violated sphericity, we reported the more conservative Greenhouse-Geisser corrected values. Significant main effects were reported if there were no significant interactions. Where there was a significant interaction effect of group, a further two-way ANOVA analysis was conducted for each group. If a significant main effect of slope was reported, a further pairwise comparisons analysis was conducted for level, downslope and upslope walking. A p value less than or equal to 0.05 was considered statistically significant.

RESULTS

Propulsion force

No changes in peak propulsion forces were observed between 1st and 19th minute of walking overall in paretic, non-paretic and non-impaired legs, as suggested by the lack of statistical significance in main effects of time ($p>0.05$). Additionally, when we compared the 3 slope conditions, the change in peak propulsion forces generated in response to each slope condition were different between the paretic, non-paretic and non-impaired legs, as suggested by the statistically significant interaction between slope and group ($F(2.861, 44.351)=10.122, p<0.001$). We observed a statistically significant main effect for slope in paretic ($F(1.088, 13.424)=21.611, p<0.001$), non-paretic ($F(2, 22)=55.630, p<0.001$) and non-impaired ($F(1.222, 10.998)=155.283, p<0.001$) legs. (See Graphs 1-3 below).

Graphs 1-3. Peak Propulsive Forces

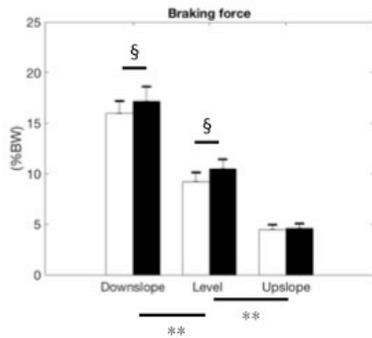


Pairwise comparisons indicated that paretic legs generated higher propulsion force during upslope ($11.75\pm 1.04\%BW$), but comparable propulsion forces during downslope, when compared to level walking ($6.14\pm 0.67\%BW$). In contrast, both non-paretic and non-impaired legs generated higher peak propulsion during upslope (non-paretic= $13.81\pm 1.22\%BW$; non-impaired= $23.10\pm 1.07\%BW$) and lower peak propulsion during downslope (non-paretic= $5.47\pm 1.19\%BW$; non-impaired= $8.43\pm 1.25\%BW$), compared to level walking (non-paretic= $8.66\pm 1.46\%BW$; non-impaired= $17.20\pm 1.18\%BW$).

Braking force

The changes in peak braking forces between first and last minute of walking generated by paretic, non-paretic, and non-impaired legs were different for the three slope conditions, as suggested by the statistically significant interaction between slope and time ($F(2,62)=3.290$, $p=0.044$). Specifically, peak braking forces increased with 20 minutes of level and downslope walking but did not change with upslope walking. Additionally, when we compared the 3 slope conditions, the change in peak braking forces generated in response to each slope condition were not different between the paretic, non-paretic and non-impaired legs, as suggested by the lack of statistical significance in slope and group interaction ($F(2.294, 35.550)=1.710$, $p=0.192$). However, we did observe statistical significance in main effect for slope in paretic ($F(2,22)=33.178$, $p<0.001$), non-paretic ($F(1.144, 12.585)=23.246$, $p<0.001$) and non-impaired legs ($F(1.137, 10.998)=22.766$, $p<0.001$). (See Graph 4 below). Pairwise comparisons between slope types indicated that on average, peak braking forces were higher when walking downslope and lower when walking upslope, when compared to level walking.

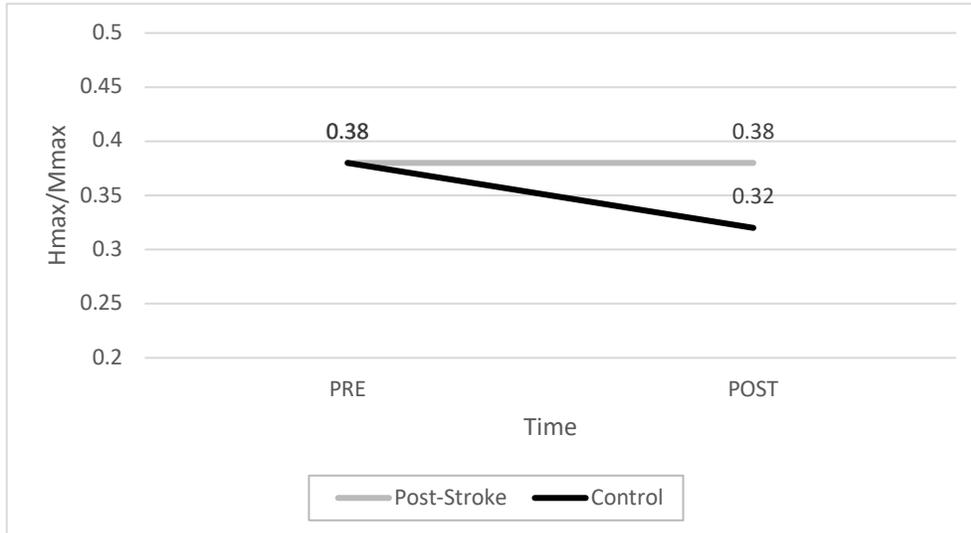
Graph 4. Peak Braking Force



H_{max}/M_{max}

We observed an overall change in H_{max}/M_{max} ratio following 20 minutes of walking, and the change was different for post-stroke compared to control group, as suggested by the significant interaction between time and group (F(1,19)=16.84, p=0.001). Specifically, in individuals post-stroke, on average across all slope conditions, we observed no change in post walking H_{max}/M_{max} ratio (0.38±0.06) when compared to pre-walking H_{max}/M_{max} ratio (0.38±0.06) (p=0.766). In the control group, however, on average across all slope conditions, H_{max}/M_{max} ratio was lower immediately after treadmill walking (0.32±0.06), compared with pre-walking H_{max}/M_{max} ratio (0.38±0.06) (p=0.00). (See Graph 5 below). Furthermore, the effects of slope conditions were not stronger for one than for the other, as suggested by the lack of interaction between time and slope (F(2,38)=2.79, p=0.07).

Graph 5. Change in H_{max}/M_{max} Over Time



DISCUSSION

This was the first study to examine the effects of sloped walking on excitability of the spinal reflex pathway and propulsive and braking forces in individuals post-stroke. The self-selected walking speeds for all three slope conditions were lower for individuals post-stroke than controls. Choosing to allow individuals to self-select treadmill speed allows the results to be interpreted for day to day activity. Forcing individuals to walk at an uncomfortable treadmill speed can cause a change in the H-reflex adaptation secondary to increased difficulty (Arnold et al., 2017).

Immediately following 20 minutes of walking, regardless of the slope condition, the H_{max}/M_{max} ratio was decreased in non-impaired individuals. These results are consistent with the post-activation depression phenomenon associated with H-reflex inhibition following activity (Masakado et al., 2005; Tahayori, Tahayori, & Koceja, 2015; Voigt & Sinkjaer, 1998). The post-stroke individuals had no significant change in the H_{max}/M_{max} . Other studies have shown downslope walking and level walking H_{max}/M_{max} ratios were depressed in non-impaired individuals, with the downslope being greater than the level (M. J. Sabatier et al., 2015; Thompson, Pomerantz, & Wolpaw, 2013). It has been hypothesized that

the H_{\max}/M_{\max} ratio is reduced to a greater extent in downslope walking because eccentric control requires more motor complexity (M. J. Sabatier et al., 2015). Downslope walking requires predominantly eccentric control and with the stretch of the muscle during these contractions, there is an altered stretch related afferent feedback and increase in cortical mediated spinal inhibition (Arnold et al., 2017). These manifest as a trending reduction in spinal excitability in the non-impaired nervous system; however, no such trend was observed in the stroke-impaired nervous system suggesting impairment in this pathway. There has been evidence of a dose response for treadmill walking in non-impaired individuals, with greater H_{\max}/M_{\max} soleus depression following increased walking time and increased percentage of decline (Arnold et al., 2017). No such dose response has been observed in stroke-impaired individuals, which may indicate that to achieve H_{\max}/M_{\max} depression in stroke-impaired individuals walking time and percentage decline may need to be altered.

Following upslope walking, we observed trending lower H_{\max}/M_{\max} ratios in non-impaired individuals, but a trending increase in the H_{\max}/M_{\max} ratios in the paretic legs of individuals post-stroke. Upslope walking is unique in that it is associated with more motor unit recruitment than level or downslope (Franz & Kram, 2012; A. N. Lay, Hass, Richard Nichols, & Gregor, 2007) and requires greater force-related feedback compared to level or downslope walking (A.N. Lay et al., 2006). This may suggest that the more challenging upslope condition, with more motor unit recruitment required, does not allow the paretic soles H-reflex to adapt short-term.

Similar to results seen in previous research, after an acute bout of level walking, non-impaired individuals had a trending decrease in H_{\max}/M_{\max} ratios (M. J. Sabatier et al., 2015; Thompson et al., 2013). Additionally, stroke-impaired individuals had a trending decrease from pre to post H_{\max}/M_{\max} ratios. Based off of our relatively high lower extremity Fugl-Meyer scores, the post-stroke individuals in our study were higher functioning. With the relatively simple level walking, higher functioning post-

stroke individuals were able to exhibit a similar trend to non-impaired individuals, suggesting appropriate soleus H-reflex adaptation in this condition.

When comparing the paretic peak propulsion forces, we observed that the propulsion forces were lower during level walking compared to the upslope, and level walking was greater compared to the down-slope condition. This pattern corresponds to the pattern seen in non-impaired individuals (A.N. Lay et al., 2006) and demonstrates that a short-term adaptive paradigm of force modulation is possible in people post-stroke with altered walking conditions. In non-impaired individuals, it has been shown that propulsive forces increase with upslope walking compared to level walking largely because it takes greater effort to propel the COM forward (Damavandi, Dixon, & Pearsall, 2012). Propulsive forces are reduced in individuals with hemiparesis, and propulsive forces from the paretic leg have been shown to be affected by the severity of paresis, such that those with more severe paresis have less propulsion than those with less severe paresis (Bowden et al., 2006). Peak force has been found to increase in the first metatarsal and hallux regions relating to increase power production stemming from gastrocnemius to propel the body forward in an upslope condition gradient of +8.5% (Grampp, Willson, & Kernozek, 2000). Because of the effect of paresis on propulsive forces and the biomechanical differences in sloped walking, it can be concluded that paretic legs have overall decreased propulsive forces, and the pattern of propulsion is similar to that seen in non-impaired individuals for level, upslope, and downslope walking tasks; thus, demonstrating that individuals with post-stroke hemiparesis still maintain the appropriate propulsive force modulation pattern for sloped walking.

For the paretic peak braking forces, despite slower self-selected walking speeds, we observed inappropriately higher peak braking forces during all three sloped conditions. Individuals post-stroke did not exhibit the same motor control pattern with braking forces for sloped walking tasks as non-impaired individuals, and as seen in previous studies (A.N. Lay et al., 2006). The force pattern may be altered in individuals post-stroke due to altered muscle activation during ambulation. An increase in lower

extremity muscle activation of the non-affected lower extremity has been associated with reduced motor ability of the plantarflexor moment and dynamic ankle strength of the affected limb (Rosa, Marques, Demain, & Metcalf, 2014). In the acute stage post-stroke, foot drop is common secondary to dorsiflexor weakness, which further limits the ability of the dorsiflexors to contribute to walking stability required in community environments, such as on slopes and ramps. The dorsiflexor strength increases with recovery and muscle co-contraction of the Tibialis anterior and Gastrocnemius complex increases ankle stability in order to allow safe community ambulation (Rosa et al., 2014). This co-contraction in individuals post-stroke may represent the cause of the increased braking forces throughout all slopes. Although individuals post-stroke were able to demonstrate appropriate propulsive force modulation with varying slope tasks, this motor control was not seen with braking forces.

For propulsive and braking forces, we found a significant difference for conditions such that there were decreased propulsive and braking forces in level versus upslope and increased propulsive and braking forces in level versus downslope with upslope walking having greatest propulsive and braking forces. There was also no significant change with time, such that the first minute (T1) and the twentieth minute (T20) were not significantly different from one another in terms of propulsive forces on both the paretic and non-impaired limbs. Previous literature has studied joint mechanics while walking upslope and downslope such that hip, knee and ankle extensors are favored during upslope walking and knee extensors are favored during downslope (Franz & Kram, 2012). This can be due to the variability in the center of pressure (COP) distribution in the anterior to posterior direction when performing sloped walking (Franz & Kram, 2012). A possible explanation in our study is that individuals post-stroke may have not been able to control their COP distribution contributing to their lack of dorsiflexion control to braking forces during downslope walking. Another previous study performed on subjects with spinal cord injuries (SCI) looked into gait speed and propulsive forces (Desrosiers, Duclos, & Nadeau, 2014). Their research study found that decreased gait speed contributed to decreased power

in hip, knee and ankle joints. This is consistent with our results in that individuals post-stroke showed decreased gait speeds in all three walking conditions. However, when matched similar speeds between non-impaired and SCI group, hip and knee power was found to be equal with ankle power still decreased in the SCI group indicating that speed has no effect on power distribution when referring to the ankle joint (Desrosiers et al., 2014). This may correlate to our study due to our stroke impaired group not having the necessary eccentric strength in the plantar flexors to increase braking forces when ambulating downhill.

When comparing slope conditions in the non-impaired group there was no significant difference in braking forces found between downslope and level walking indicating that a non-impaired nervous system negotiates both conditions similarly in terms of braking forces. In both the non-paretic and paretic groups, we found a difference between downslope and level walking indicating that a change in slope or terrain during ambulation requires increased modulation and altering braking forces to perform the activity. Previous literature has shown that slope gradient plays a factor in the amount of loading in either the forefoot or hindfoot during treadmill walking in a non-impaired population (Grampp et al., 2000). Their research showed that with downhill gradients, peak forces in the heel region were found to increase at a -15% grade with decreases in the metatarsal region (Grampp et al., 2000). Our research study conducted a 5% grade in each direction, which agrees with Grampp's findings such that the gradient amount walking downhill was not enough to make an effect in force distribution throughout the foot with non-impaired individuals.

Paretic legs generated high propulsive forces during upslope walking with comparable propulsive forces when comparing downslope to level walking. Non-paretic and non-impaired legs generated higher peak propulsive forces during upslope walking and lower peak propulsion during downslope when compared to level walking. The trend of these three groups were similar in peak propulsive forces being highest in upslope walking. This can be comparable to previous literature

showing an increased COP in the metatarsal heads during upslope walking (Franz & Kram, 2012; Grampp et al., 2000). This can lead to an increased activation of the plantar flexors used to propel the subject and advance the swing limb upward during upslope walking. Despite previous literature using a 15% grade in upslope or downslope ambulation, our trend was similar in this research study while using a 5% grade.

Peak braking forces increased following 20 minutes of level and downslope walking only. This means that paretic, non-paretic and non-impaired groups were examined under the three slope conditions due to not peak brake effect. This shows that non-paretic, paretic and non-impaired limbs all trended towards utilizing peak braking forces when walking downslope and lower during upslope when compared to level walking. A decreased grade on the treadmill at 5% vs other previous research studies at 15% (M. J. Sabatier et al., 2015) could have possibly skewed these results. Having an increased grade during downslope walking may have been enough to have a significant effect between groups by having their COP's shifted (Grampp et al., 2000) and therefor affecting braking force values.

CONCLUSION

The purpose of our study was to examine the adaptation in foot force control and the soleus H-reflex excitability following an acute bout of upslope or downslope treadmill walking in individuals post-stroke compared to non-impaired individuals. Our observations suggest that when the biomechanics of the walking task is altered, through adjusting the slope of the walking surface, paretic legs exhibit increased propulsion forces during upslope walking. Paretic propulsive forces were greatest in the upslope condition and lowest in the downslope condition. Regardless of group, individuals had greatest braking forces during the downslope condition and lowest during the upslope condition. Our results have repeated what other studies have found regarding GRFs except for paretic propulsion forces. We believe, based on current studies, that increased paretic propulsion forces in the upslope condition may

be due to the increased difficulty of the environmental condition. In the level condition, spinal circuits in the stroke-impaired nervous system are trending towards adaptations similar to the non-impaired nervous system, such that the H_{\max}/M_{\max} ratios were depressed. However, in the more challenging upslope condition, adaptations of the paretic soleus H-reflexes were impaired such that the H_{\max}/M_{\max} ratios were trending towards elevated. Future studies will examine the optimal walking duration and degree of slope to induce neural adaptations, as well as determine any long-term retention of plasticity.

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