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Characteristics of horizontal force generation for individuals post-stroke walking against progressive resistive forces

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Abstract

Background—Walking, while experiencing horizontal resistive forces, can allow researchers to assess characteristics of force generation in a task specific manner for individuals post-stroke.

Methods—Ten neurologically nonimpaired individuals (mean age 52 years) and fourteen chronic stroke survivors (mean age 54 years) with hemiparesis walked in the treadmill-based KineAssist Walking and Balance System, while experiencing twelve progressive horizontal resistive forces at their comfortable walking speed. Slope coefficients of the observed force–velocity relationship were quantified and submitted to an iterative k-means cluster analysis to test for subgroups within the post-stroke sample. Extrapolated force values for individuals were quantified by extrapolating the line of best fit of the force–velocity relationship to the x-intercept.

Findings—Within the post-stroke group, six individuals were clustered into a high sensitivity group, i.e., large reduction in speed with resistance, and eight were clustered into a low sensitive group, i.e., small reduction in speed with resistance. The low sensitivity group was similar to non-impaired individual. The extrapolated force was significantly higher for non-impaired individuals compared to individuals post-stroke in either the high or low sensitivity group. The differences between low and high sensitivity group suggest that high sensitivity of walking speed to applied resistive force is indicative of overall weakness.

Interpretation—Individuals with high sensitivity to horizontal resistive force may be walking at or near their maximum force generating capacity when at comfortable walking speed, while low sensitivity individuals may have greater reserve force generating capacity when walking at a particular comfortable walking speed.

Keywords

Post-stroke hemiparesis; Force generation; Walking speed; Locomotion

1. Introduction

Limited locomotor ability, such as very slow walking speed, is an important factor in determining the degree of physical disability and independence for individuals post-stroke (Perry et al., 1995). Impaired force generating capacity has been suggested to be a primary cause of motor dysfunction for these individuals (Bohannon, 1991; Clark et al., 2006; Clark et al., 2010). Reduced rate of force development (Fimland et al., 2011) and decreased maximum isometric and isokinetic force output (Clark et al., 2006) has been documented and this decreased force output has been correlated to the observed walking speed of individuals post-stroke (Bohannon, 1986; Kim and Eng, 2003). Moreover, horizontal ground reaction forces, a measure of lower limb force production while walking, have been correlated with both walking speed and hemiparetic severity (Bowden et al., 2006). Collectively, these investigations provide evidence that very slow walking speed of individuals post-stroke can result from the limitations in force output capacity post-stroke.

However, individuals post-stroke are capable of expressing a greater range of walking speeds and horizontal propulsive forces, than typically observed. With respect to speed-generating capacity, in a recent investigation individuals post-stroke achieved walking speeds that were 150% greater than their maximum overground walking speed while walking on a treadmill with just a safety harness that provided no body weight support (Capo-Lugo et al., 2012). With respect to force-generating capacity, in a follow-up study, the self-selected speeds of individuals post-stroke walking against progressive horizontal resistive forces were faster than predicted if individuals were extremely limited in generating greater propulsive forces (Hurt et al., 2014). This suggests that these individuals have the capacity to generate propulsive forces necessary to walk at faster speeds.

The task of walking requires a complex interaction of muscle moments at individual and multiple joint segments (Neptune et al., 2001; Robertson and Winter, 1980) to support the center of mass and generate forward progression. Traditional strength measures that attempt to isolate individual joint moment outputs at either isometric or isokinetic speeds may not capture the entirety of force generation impairment post-stroke. While walking, force output is quantified using fore-aft ground reaction forces and/or moment generation at individual's comfortable or maximum walking speed (Bowden et al., 2006; Milot et al., 2007). More recently, investigations have considered the observed moment generation while walking relative to dynamometric measures of maximum strength (Milot et al., 2006; Milot et al., 2007). However, these measures may not fully describe the capability of the central nervous system to generate a greater range of force in a functional manner. Walking, while experiencing horizontal resistive forces, can allow researchers to assess characteristics of force generation in a task specific manner. Although investigations have utilized this methodology to assess the relationship between mechanical and metabolic work requirements during walking (Donovan and Brooks, 1977; Gottschall and Kram, 2003); no

study, to date, has utilized this methodology to characterize parameters of force generation while walking. Assessing individuals walking at their self-selected comfortable walking speed (CWS), while progressive horizontal resistance is applied, may reveal new insight into the nature of force generation impairment during walking post-stroke.

Therefore, the purpose of this study was to investigate characteristics of horizontal force generation of individuals post-stroke walking at self-selected comfortable walking speeds under progressive horizontal resistive forces in order to characterize maximum horizontal force generating capacity, compared to nonimpaired individuals. Due to the previously reported reduced force generation capacity, we hypothesized that individuals post-stroke, compared to non-impaired individuals, would demonstrate lower estimated maximum force output capability. In addition, we hypothesized that the rate of speed reduction relative to the progressively increasing horizontal force (i.e., sensitivity) will be steeper in individuals post-stroke, especially in individuals with slower CWS, due to their lack of ability to increase propulsive force proportional to the requirements of the task. This study represents the first description of a new measurement tool that can assess progressive capacity to generate propulsive force during walking post-stroke.

2. Methods

Fourteen chronic stroke survivors (>6 months post ictus; age range: 42–82; 54 SD 12 years old) with hemiplegia and 10 non-impaired individuals (age range: 41–61; 52 SD 7 years old) volunteered to participate in this institutionally reviewed and approved study. All data were collected at the Department of Physical Therapy and Human Movement Sciences at Northwestern University. Northwestern University IRB approved the study for human subject participation. The inclusion criteria of post-stroke were as follows: unilateral stroke, able to walk 10 m independently without walking aids other than ankle foot orthoses, medically stable (controlled blood pressure, no arrhythmia, and stable cardiovascular status), and can provide written informed consent. Recruitment criteria for the nonimpaired individuals were: over 40 years of age, no musculoskeletal, cardiovascular or neurological disorders that affected their gait performance, and able to walk 10 m without assistance.

Participants from the above groups were excluded when they had the following conditions: (1) severe cardiac disease (New York Heart Association classification II–IV), (2) systolic blood pressure reduction >20 mm Hg of quiet standing, resting systolic pressure >140 mm Hg and diastolic blood pressure >90 mm Hg, (3) poorly controlled or brittle diabetes mellitus history, (4) lower limb amputation history, (5) presence pre-morbid gait disorder from any cause, (6) a history of lower limb non-healing ulcer, (7) simple commands following failure, and (8) weight >113 kg (due to weight restriction of robotic device). The Berg Balance Scale and Fugl–Meyer assessments were performed on individuals post-stroke (Blum and Korner-Bitensky, 2008; Fugl-Meyer et al., 1975). Participant's characteristics are shown in Table 1. All participants provided written informed consent prior to participation in this investigation.

2.1. Experimental task

The KineAssist Gait and Balance Training System™ (HDT Global, Solon OH, USA, Fig. 1) was employed for the study (Capo-Lugo et al., 2012; Patton et al., 2007). The robotic device was stationary and connected with a Biodex-RMS treadmill (Biodex, Shirley, NY, USA) for this investigation. Individuals interacted with the robot through a pelvic harness. Embedded in the pelvic harness are bi-lateral, force transducers, which, based on a linear relationship between the measured horizontal force and treadmill belt velocity, allow the treadmill to be self-driven at a large range of speeds (0–3.0 m/s). The progressive horizontal resistive forces were generated by modifying the relationship between forces required to move the treadmill belt at a given speed. All participants were tested as follows: 1) three trials each of 10-meter walk test (10-MWT) overground at self-selected comfortable walking speed (CWS) and self-selected maximum walking speed (MWS), 2) a horizontal resistive test while walking in a robotic device, and 3) twelve progressive horizontal resistive forces walking trials that were determined from the horizontal resistive test. A tether that was anchored to the KineAssist was attached to a vest and was used to limit the forward trunk flexion to 10°. As described above, only measured force signals from transducers in the pelvic interface are used to dictate the treadmill belt speed. The tether was utilized to limit the extent that individuals could use their body weight to drive the treadmill belt. This was particularly important for high resistive force magnitudes. Participants were allowed to undergo a short familiarization process with the device.

A horizontal resistive force test preceded the progressive horizontal resistive test. Up to three trials, lasting 90 s each, were collected for each participant. Resistive force was increased every 30 s. With each trial the increasing steps of resistive force were smaller as the force level was reached in which no noticeable movement of the treadmill belt was detected. Participants were encouraged “to try your best to keep walking no matter how hard it gets” as the horizontal resistance was increased. The maximum value for each participant was then used to determine the range of resistive forces that individuals experienced for the subsequent walking trials.

Participants were then asked to walk at “the speed that feels the most comfortable” against twelve progressively increasing horizontal resistance levels. If the maximum force from the horizontal resistive force test was 120 N, twelve intervals of 10 N increments were randomly presented. In each walking trials, approximately 20 continuous steps were collected at each force interval. At least 30 s of rest was permitted between each trial.

2.2. Data processing and statistical analysis

In order to assure that we selected data from a steady-state walking condition, we used custom software (MATLAB, MathWorks, Natick, MA, USA) to determine the lowest coefficient of variation, i.e., the standard deviation of speed/the mean speed, of a moving ten second window of the treadmill belt speed over the duration of the trials to ensure that individuals were walking at a constant speed. This window of data was then used for the subsequent analysis. We utilized a simple regression to quantify the relationship between horizontal resistive force and walking speed. This presumed a negative linear relationship represented the sensitivity of individuals' comfortable walking speed to progressive

horizontal resistive force. Only individuals with a significant linear fit were used for subsequent analyses. The slope coefficients were then submitted to an iterative k-means cluster analysis. This analysis was used to explore whether subgroups were present in the sensitivity of individuals post-stroke walking speed to progressive resistive force. We were particularly interested whether individuals in the post-stroke group were similar to the control group and thus performed this analysis with data from both groups. Using the negative linear relationship we also extrapolated the regression line to the x-intercept (i.e., the magnitude of force that would result a walking speed of 0 m/s) to estimate the maximal horizontal propulsive force of the individual, referred to heretofore as the extrapolated force value. We also quantified the difference between the maximum overground walking speed and the comfortable overground walking speed, which is referred to as the reserve speed. We were particularly interested in whether this value would relate to the characteristics of force generation of individuals walking while experiencing progressive horizontal resistance. The extrapolated force value was normalized by the body mass of all participants. The statistical analysis was performed with SPSS 18 (IBM Armonk NY, USA). Kolmogorov–Smirnov testing indicated normal distribution of data related to speed and force for individuals post-stroke and the age-matched control group.

The difference in the sensitivity and extrapolated force values between groups was examined by one-way ANOVA. Multiple comparisons were tested using a Bonferroni correction. The association analyses were analyzed by Pearson's r correlations. Effect size is reported as Cohen's d . Statistical significance was set at $P < 0.05$.

3. Results

We observed a significant reduction in walking speed between the overground CWS for post-stroke compared with non-impaired individuals (0.88 SD 0.20 m/s vs. 1.36 SD 0.16 m/s respectively $P < 0.001$), as well as the maximum overground walking speed (1.22 SD 0.26 m/s vs. 2.24 SD 0.59 m/s respectively $P < 0.001$). Significant linear fits were found for all participants with the exception of one healthy non-impaired participant (S6, Table 2). Thus, fourteen post-stroke and 9 nonimpaired controls were submitted for further analysis.

Overall, the sensitivity of the relationship between CWS and horizontal resistive force was similar to non-impaired individuals for a subset of individuals post-stroke. At the group level, despite a 28% difference in the sensitivity of this relationship, we did not detect a statistically significant difference between individuals post-stroke and non-impaired control participants (-0.31 SD 0.14 vs. -0.23 SD 0.08 m·kg/N·sec respectively, $P = 0.07$). However, using a cluster analysis technique, individuals post-stroke appeared to be of two groups: those similar in sensitivity to non-impaired individuals and those who had a higher sensitivity. The two identified clusters had significantly different sensitivity values when compared with each other ($P < 0.001$) with centers of the clusters at slope values of -0.21 m·kg/N·sec ($n = 7$) and -0.44 m·kg/N·sec ($n = 16$) for the two clusters. Six individuals post-stroke were clustered together and represented a high sensitivity to horizontal resistive force (-0.41 SD 0.1 m·kg/N·sec), whereas eight of the post-stroke individuals were clustered into a group that included a less sensitive relationship (-0.18 SD 0.04 m·kg/N·sec, Table 2), similar to non-impaired individuals. A significant difference in sensitivity was detected in

the one-way ANOVA between groups ($P < 0.001$). Post-stroke individuals with higher sensitivity (HS) were significantly different than either the non-impaired individuals ($P < 0.001$, Cohen's $d = 1.98$) or the individuals post-stroke with lower sensitivity (LS, $P < 0.001$, Cohen's $d = 3.02$). We did not detect a statistical difference between the non-impaired individuals and individuals post-stroke with lower sensitivity ($P = 0.892$, Fig. 2).

The extrapolated force value that would result in zero walking speed was significantly higher for non-impaired individuals, $5.33 \text{ SD } 1.55 \text{ N/kg}$, compared to individuals post-stroke at the group level, $3.03 \text{ SD } 1.26 \text{ N/kg}$, ($P < 0.001$). Comparing groups when accounting for the subgroups showed a significant effect of group in the model ($P < 0.001$). Post hoc analysis utilizing a Bonferroni correction, revealed lower values for the HS group ($1.98 \text{ SD } 0.27 \text{ N/kg}$) when compared with both the LS ($3.80 \text{ SD } 1.12 \text{ N/kg}$, $P = 0.029$, Cohen's $d = 2.2$) and nonimpaired group ($5.33 \text{ SD } 1.55 \text{ N/kg}$, $P < 0.001$, Cohen's $d = 3.01$). Further, a lower value was calculated for the LS group when compared with the nonimpaired group ($P = 0.029$, Cohen's $d = 1.13$, Fig. 3).

We were able to detect a significant correlation between the reserve speed and sensitivity for individuals in the LS group, ($r = 0.71$, $P = 0.049$) but not the HS group ($r = 0.18$, $P = 0.73$). The reserve speed was also significantly correlated to the extrapolated force value for the LS group ($r = 0.825$, $P = 0.012$) but not for the HS group ($r = -0.463$, $P = 0.355$). For individuals post-stroke clustered in the HS and LS we were unable to detect a significant correlation between the Fugl Meyer scores and the extrapolated maximum force ($r = -0.43$, $P = 0.47$; $r = -0.63$, $P = 0.09$, respectively).

4. Discussion

The purpose of the current study was to investigate characteristics of horizontal force generation of individuals post-stroke walking at self-selected comfortable walking speeds under progressive horizontal resistive forces in order to characterize maximum horizontal force generating capacity. We hypothesized that the rate of speed reduction relative to the progressively increasing horizontal force (i.e., sensitivity) would be steeper in individuals post-stroke due to their lack of ability to increase propulsive force proportional to the requirements of the task. We were not able to detect a difference between the post-stroke group and non-impaired controls despite almost a 30% difference in sensitivity of their comfortable walking speed to horizontal resistive force. However, it was discovered that individuals post-stroke were of two subgroups: one with a high sensitivity to horizontal resistive forces and the other, which had a similar sensitivity to horizontal resistive force as the control group. Thus, our hypothesis was only partially-supported. In addition we hypothesized that individuals post-stroke, compared to non-impaired, and would demonstrate lower extrapolated force output capability. This hypothesis was supported since the extrapolated force output of individuals post-stroke was ~45% less than non-impaired controls. Even when we examined the post-stroke group respective of the individual groups, these differences persisted when compared to the nonimpaired group.

The sensitivity of walking speed to progressive horizontal resistance for the post-stroke group varied across participants. In fact a subgroup individuals post-stroke i.e., LS, was

observed to have a similar response to progressive horizontal resistance as non-impaired controls. Previously reported data has suggested that some post-stroke individuals utilized a similar relative sense of effort as non-impaired controls walking at their comfortable speeds (Milot et al., 2006). These findings suggest that some individuals possess the ability to generate greater lower limb force while walking, a requirement to walking faster under unobstructed walking conditions (Orendurff et al., 2008). From a functional standpoint, we found that the reserve speed (i.e., MWS–CWS) of individuals was strongly correlated to our sensitivity measure, however only within the LS group. This further provides some evidence that for the LS group, other factors may affect their choice of their self-selected CWS such as becoming unstable or a fear thereof.

Reduced extrapolated force values were observed for the LS and HS post-stroke group compared to non-impaired individuals. Impaired force generating capacity has been suggested to be the primary cause of motor dysfunction for individuals post-stroke (Clark et al., 2006). Further, propulsive forces while walking have been shown to be related to the slow walking speed of individuals post-stroke, specifically of the paretic limb (Bowden et al., 2006). Indeed it should be noted that in the current investigation that both post-stroke groups were observed to have decreased extrapolated force values compared to the control group. However, a 98% increase in this value was observed between the HS and LS groups suggesting greater functional capacity of the LS group, which coincided with a significant correlation between the reserve speed and the extrapolated force value for the LS group only. A previous investigation has shown that individuals post-stroke walking under progressive horizontal resistive force generated greater propulsive forces than could be predicted if they were extremely limited in their ability to generate horizontal propulsive forces (Hurt et al., 2014). Currently we are unable to differentiate force output between the paretic and nonparetic limb. Thus, the current investigation represents a composite measure of force output by the individual. Further investigations incorporating ground reaction force measures could resolve the magnitude of force output into either the paretic or nonparetic limbs.

Walking while experiencing progressive horizontal resistive forces is an innovative way to not only assess lower limb strength and power but can also to be used to increase strength in a meaningful and task specific way. While resistance training has been shown to improve force output of individuals post-stroke (Bourbonnais et al., 2002; Kim et al., 2001; Milot et al., 2008); it has been suggested that the lack of specificity in the training task may limit transfer of gains from the resistance training regime to more functional tasks (Eng, 2004). In one investigation, despite greater than a 25% increase in the strength of the plantar flexors and hip flexors, walking speed only increased around 10% (Milot et al., 2008). One potential issue is that walking requires the coordination of force output with respect to multiple joints in a time dependent manner. Thus, the lack of task specificity of strength training may limit the extent to which these results translate functionally. The task-specific nature of the current protocol may represent a more functional test for assessing strength deficits and further be modified to be used as a strength training protocol.

We recognize several limitations of the current investigation. First, we did not assess the maximum force output of individuals post-stroke. Instead we report an extrapolated force

value in place of an actual physical measurement. It should be noted however, the protocol tested individuals at twelve different resistive force magnitudes and we were able to demonstrate a strong linear relationship for 23 of the 24 participants. Thus we feel that these values are indicative of the maximal force capabilities of the participants. Also, this study was limited by an inability to ascribe the increased force generation to either the paretic or nonparetic limb or a combination of both. Although this is important, we propose that the current investigation still provides value information related to the capability of individuals post-stroke to modulate propulsive force output while experiencing progressive horizontal resistive forces.

5. Conclusion

With the current investigation, we demonstrated that when individuals post-stroke walk while experiencing progressive horizontal resistive forces, they can be identified as two groups: one group whose walking speed was greatly reduced with high sensitivity to horizontal resistive force and one with a similar sensitivity as non-impaired individuals. This suggests that an approach to increasing CWS for individuals with higher sensitivity may require improving the range of lower-limb force generating capacity whereas individuals with lower sensitivities may benefit from learning how to remain safe and stable at faster speeds that they are capable of achieving.

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Fig. 1.

Picture of experimental setup used for this investigation. The KineAssist is interfaced with a treadmill. Force applied by the person to the pelvic harness is used to generate a command signal that dictates the speed of the treadmill belt. This linear relationship was manipulated to create a constant horizontal resistive force.

Fig. 2. Sensitivity of walking speed to progressive horizontal resistive forces displayed by group. A significant difference was observed between those clustered in the high sensitivity group (dashed line) and the low sensitivity group (solid gray line) as well as the age-match control group (solid black line, $P < 0.001$). No difference was detected between the age-matched controls and the low sensitivity group ($P = 0.892$).

Fig. 3. Extrapolated force value by group. Average extrapolated force plus standard deviation is shown. A significant difference was detected between all groups when adjustments to the *P*-value for multiple comparisons were made.

Table 1

Characteristics of post-stroke individuals compared with summary data from nonimpaired participants.

Subject ID	Male/female	Age	Height (cm)	Weight (kg)	M.S.S.	Weak side	Brace (AFO)	BP medications	BBS	FM
S1	M	57	178	82	290	L	N	Y	46	19
S2	F	59	173	66	314	R	Y	N	53	20
S3	F	64	160	60	105	R	Y	Y	51	17
S4	M	57	180	92	105	R	N	N	52	26
S5	M	45	180	73	164	R	N	Y	51	20
S6	M	50	178	83	96	L	N	?	49	16
S7	M	46	180	96	291	R	N	N	54	21
S8	M	42	173	98	27	L	Y	Y	50	15
S9	M	43	178	64	101	L	Y	N	55	20
S10	M	57	180	95	94	R	N	Y	53	19
S11	F	60	150	62	166	L	N	Y	53	20
S12	M	53	178	93	46	R	Y	Y	46	21
S13	M	82	168	68	217	R	Y	Y	–	–
S14	M	58	170	77	38	L	Y	Y	47	18
Mean		55	173	79	147				51	19
SD		10	9	14	97				3	3
<i>Non-impaired</i>										
Mean		51	174.2	77.3						
SD		8	9.9	14.2						

M.S.S. = Months Since Stroke, AFO = Ankle Foot Orthoses, BP = Blood Pressure, BBS = Berg Balance Score, and FM = Fugl Meyer

Table 2

Overground walking speeds and regression analysis. Results of individual regression analysis as well as overground walking speed are shown. Means and standard deviations (SD) are shown. For the Post-stroke group, those individuals in the low sensitivity (LS) group are denoted in gray while the high sensitivity (HS) group is denoted by white.

Post-Stroke	Beta coefficient	R ²	p value	Overground CWS	Max CWS	Max CWS–Overground CWS
S1	−0.18 (0.04)	0.67	0.002	0.5	0.9	0.4
S2	−0.22 (0.03)	0.91	<0.001	0.7	1.3	0.6
S3	−0.31 (0.05)	0.79	<0.001	0.8	1.1	0.3
S4	−0.11 (0.02)	0.67	0.001	0.9	1.5	0.6
S5	−0.20 (0.04)	0.74	0.001	1.0	1.5	0.5
S6	−0.46 (0.04)	0.93	<0.001	0.7	0.9	0.2
S7	−0.43 (0.04)	0.93	<0.001	0.9	1.1	0.2
S8	−0.40 (0.13)	0.63	<0.019	0.8	0.9	0.1
S9	−0.42 (0.06)	0.83	<0.001	1.2	1.5	0.3
S10	−0.29 (0.03)	0.93	<0.001	0.7	0.9	0.2
S11	−0.14 (0.04)	0.65	0.003	0.8	1.3	0.5
S12	−0.21 (0.06)	0.69	0.011	1.0	1.2	0.3
S13	−0.37 (0.05)	0.89	<0.001	1.2	1.6	0.4
S14	−0.61 (0.04)	0.96	<0.002	1.1	1.4	0.3
Mean (SD)	−0.31 (0.14)			0.88 (0.20)	1.22 (0.26)	0.34 (0.15)
Non Impaired						
S1	−0.18 (0.02)	0.88	<0.001	1.5	--	0.8
S2	−0.22 (0.05)	0.71	0.001	1.4	2.2	0.8
S3	−0.41 (0.06)	0.88	<0.001	1.2	2	0.5
S4	−0.25 (0.03)	0.95	<0.001	1.1	1.6	0.7
S5	0.05 (0.42)	0.00	0.911	1.4	2.1	1.77
S6	−0.26 (0.02)	0.96	<0.001	1.4	3.17	1.6
S7	−0.16 (0.05)	0.49	0.001	1.5	3.1	0.6
S8	−0.18 (0.04)	0.62	0.002	1.3	1.9	0.6
S9	−0.19 (0.06)	0.50	0.014	1.2	1.8	0.5

Post-Stroke	Beta coefficient	R ²	p value	Overground CWS	Max CWS	Max CWS–Overground CWS
S10	−0.17 (0.05)	0.54	0.01	1.6	2.1	
Mean (SD)	−0.23 (0.08)			1.36 (0.16)	2.22 (0.55)	0.87 (0.47)

Note, for one non-impaired (dark gray highlight) individual a non-significant linear relationship was detected and thus this individual was not included in the group analyses.