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Assessment of turning performance and muscle coordination in individuals post-stroke

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ABSTRACT

Turning is an important activity of daily living and often compromised post-stroke. The fall rate for individuals post-stroke while turning is nearly four times as high compared to healthy adults, with most falls resulting in injury. Thus, there is a need for evidence-based rehabilitation targets to improve turning performance for individuals post-stroke. To produce well-coordinated movements, muscles can be organized into muscle modules (i.e., groups of co-excited muscles). Post-stroke these modules can be merged, leading to impaired muscle coordination and walking performance. However, the relationship between impaired coordination and turning performance is not well understood. Thus, the purpose of this study was to analyze the influence of impaired muscle coordination (i.e., merged modules) on turning performance (i.e., time and number of steps required to complete a turn, and smoothness and balance control during the turn). Individuals post-stroke and healthy controls performed three tasks including overground straight-line walking, a 90-degree turn, and a 180-degree turn. The number of muscle modules during straight-line walking were determined using non-negative matrix factorization. During 180-degree turning, those with two modules took longer to turn, used more steps and had less smooth movement. Those with reduced module complexity exhibited diminished balance control during both 90-degree and 180-degree turning. These results suggest obtaining independent modules should be an important aim in locomotor therapies aimed at improving turning performance. In addition, the time it takes to complete a 180-degree turn may provide useful clinical insight into impaired muscle coordination post-stroke.

1. Introduction

An important goal of post-stroke locomotor therapy is to help individuals return to activities of daily living. Straight-line walking has been the primary focus of gait and rehabilitation studies, however turning steps represent 35–45% of all steps taken during activities of daily living (Glaister et al., 2007). In addition, the fall rate for individuals post-stroke while turning is nearly four times as high compared to healthy adults (Simpson et al., 2011). Falling while turning can be extremely dangerous, with 70% of those falls leading to injury (Talbot et al., 2005). Currently, no specific exercise protocols exist for improving turning performance (ACSM, 2013). Thus, there is a need for evidence-based rehabilitation targets to improve turning performance for individuals post-stroke.

Successful performance of a given motor task is dependent upon proper coordination of muscle activity. To produce well-coordinated straight-line walking, muscles can be organized into muscle modules (i.e., groups of co-excited muscles) (e.g., Cappellini et al., 2006; Ivanenko et al., 2005; Neptune et al., 2009). Experimental studies using surface EMG data have identified four primary lower-limb modules present in healthy subjects during straight-line walking (e.g., Clark et al., 2010; Gizzi et al., 2011; Routson et al., 2014): Module 1 (gluteus medius, vasti, and rectus femoris), Module 2 (soleus and gastrocnemius), Module 3 (rectus femoris and tibialis anterior), and Module 4 (biarticular hamstrings). Post-stroke, these modules often merge or become co-activated, leading to impaired walking performance (Clark et al., 2010). Fewer modules are associated with altered propulsion patterns, increased braking, step asymmetries and impaired balance control (e.g., Allen et al., 2013; Brough et al., 2019). However,

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the influence of impaired muscle coordination on turning performance is less understood.

Turning requires changes in muscle coordination from straight-line walking (Ventura et al., 2015), which could affect turning performance for individuals post-stroke with impaired muscle coordination. Previous studies have shown that individuals post-stroke demonstrate poor turning performance in 90-degree and 180-degree turning compared to healthy controls, using an increased number of steps and a longer amount of time to complete the turn (Faria et al., 2016; Lam and Luttmann, 2009). In addition, they often display staggering during a turn (Faria et al., 2016), which is indicative of less smooth movement. Such a reduction in smoothness in performing a movement task is often characteristic of motor impairment post-stroke (Rohrer et al., 2002). These metrics could be used to assess turning performance and inform locomotor therapies.

In addition, impaired muscle coordination could affect balance control strategies used during turning by individuals post-stroke. Regulating whole body angular momentum (H), the sum of the angular momentum of the body segments about the center of mass (COM), is important for maintaining balance and preventing falls (Pijnappels et al., 2004). During straight-line walking, those with impaired balance control exhibit a higher range of H (H_R) in the frontal-plane during paretic single-leg stance (Nott et al., 2014), which is consistent with others showing a reduction of muscle modules is associated with increased frontal-plane H_R (Brough et al., 2019). H is regulated through foot placement and generation of appropriate ground reaction forces. Unlike walking, turning requires additional changes in ground reaction force impulses and joint kinetics to direct the body COM along a curved trajectory (Orendurff et al., 2006), which may be a challenge for individuals post-stroke who already exhibit altered balance and motor control strategies in straight-line walking.

The aim of this study was to analyze turning performance (90-degree and 180-degree turns) in individuals post-stroke and healthy controls. We expected that impaired muscle coordination in individuals post-stroke (i.e., fewer modules) influences turning performance, which was assessed by analyzing the time and number of steps required to complete a turn, and smoothness and balance control during the turn. Specifically, we hypothesized that those individuals post-stroke with merged modules would exhibit increased turn times, a higher number of steps required to complete the turn, less smooth turning (both linear and angular) and diminished balance control (higher frontal-plane H_R during the turn).

2. Methods

2.1. Experimental data collection

Electromyography (EMG) and full body 3D kinematic data were collected from 34 individuals with chronic post-stroke hemiparesis (12 left hemiparesis, 11 female, age: 61 ± 11 years) and 19 healthy age-matched controls (11 female, age: 59 ± 10 years). All participants provided informed consent prior to study participation in accordance with the Institutional Review Board of the Medical University of South Carolina. Patient demographics are provided in the Appendix (Tables A1 and A2). Subjects performed three tasks including overground straight-line walking, a 90-degree turn, and a 180-degree turn. For straight-line walking trials, subjects were instructed to walk at their comfortable walking speed. For turning trials, two lines were placed on the floor approximately four meters apart. Individuals were instructed to begin at the first line, walk in a straight line at a comfortable speed, and turn as quickly as possible once they heard a clicker. Research personnel pressed the clicker once individuals reached the second line. For consistency in task difficulty, individuals post-stroke were instructed to turn toward their paretic side. Healthy individuals were instructed to turn toward a randomized side, with randomization performed using a custom Matlab (Mathworks, Natick, MA) script (Suresh, 2011).

Kinematic data were collected at 120 Hz with a 16-camera motion capture system (PhaseSpace, Inc., San Leandro, CA) using a modified Helen Hayes marker set. Marker coordinates were interpolated over gaps smaller than 20 samples and resampled at 100 Hz. EMG data were collected (Motion Labs Systems, Inc., Baton Rouge, LA) at 2000 Hz using bilateral electrodes placed on the medial gastrocnemius, soleus, tibialis anterior, rectus femoris, vastus medialis, biceps femoris and semitendinosus. For the straight-line walking trials, EMG signals were high-pass filtered at 40 Hz with a zero-lag 4th order Butterworth filter, demeaned, rectified and low-pass filtered at 4 Hz with a zero-lag 4th order Butterworth filter. For each gait cycle, each EMG signal was normalized by its peak value, resampled at 100 Hz and averaged across steps (Clark et al., 2010).

2.2. Data analysis

While each subject completed several trials, only one trial of each task was analyzed. Trials with the cleanest marker data were chosen for analysis. Muscle coordination was assessed using the number of muscle modules during the straight-line walking task determined using non-negative matrix factorization (Clark et al., 2010). For each leg, the number of modules was increased until the variability accounted for reached ≥90% for each of the 8 muscles and 6 regions of the gait cycle (i.e., first double support, first half of ipsilateral single-leg stance, second half of ipsilateral single-leg stance, second double support, first half of ipsilateral swing and second half of ipsilateral swing). To determine the timing of each gait region, heel strike and toe off events were identified by analyzing the heel and toe marker kinematics, with the timing determined as the point when the respective marker reached a minimum height in the gait cycle.

Turning performance was assessed using five measures: time to complete the turn, number of steps taken, turning smoothness (both linear and angular) and balance control. Time to complete the turn was defined as the time it takes the body COM velocity

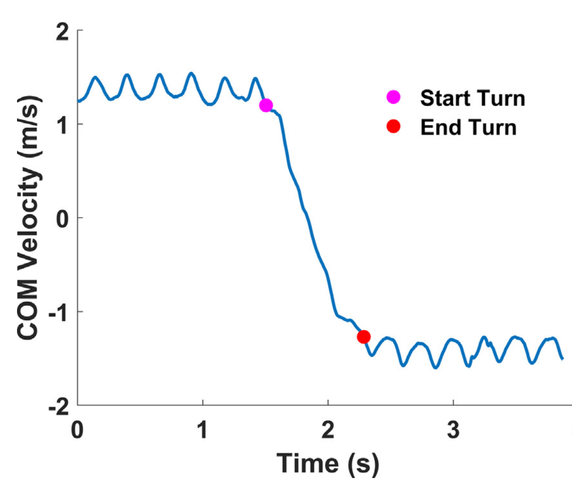

Fig. 1. Center of mass (COM) velocity during a 180-degree turn for a representative healthy subject. The direction of positive velocity is defined as the direction of initial forward motion in the fixed laboratory reference frame. Start and end times were defined by the duration the body COM velocity was outside the range of steady-state straight line walking values.

to return to the range of steady-state straight-line walking values (Fig. 1). Body COM velocity was determined as the time derivative of the body COM position, which was identified using a 13-segment model with custom scripts in LabVIEW (National Instruments, Austin, TX). COM velocity data were filtered using a 4th order Savitsky-Golay filter with a 30 point moving average window. One step was defined as one heel strike to the next. Because every start and end turn time did not coincide with a heel strike, partial steps were included. A partial step was defined as the portion of the step length that occurred during the turn.

Linear turning smoothness was assessed using a velocity peaks metric, which was defined as the number of body COM forward velocity peaks during a turn normalized by the number of heel strikes during the turn (Rohrer et al., 2002). The body COM velocity was first transformed from the fixed lab coordinate frame (x_s, y_s, z_s) into the body trajectory coordinate frame (x_T, y_T, z_T) using θ_T:

$$\theta_T = \tan^{-1}\frac{\dot{y}_s}{\dot{x}_s} \quad (1)$$

where $\dot{y}_s$ and $\dot{x}_s$ are the body COM velocity vectors defined by the fixed laboratory reference frame. A transformation matrix was used to rotate the global body COM velocity vectors into the body trajectory reference frame as previously described (Imai et al., 2001). Body COM forward velocity was low-pass filtered at 6 Hz with a zero-lag 4th order Butterworth filter. The number of peaks was determined using the *findpeaks* function in Matlab (Mathworks, Natick, MA) with a minimum peak prominence of 0.01. A smoother movement corresponds to fewer periods of acceleration and deceleration, thus leading to fewer peaks in COM velocity.

Angular turning smoothness was assessed using an H peaks metric, defined as the number of H peaks during a turn normalized by the number of heel strikes during the turn. H was assessed using a 13 segment model with custom scripts in LabVIEW as:

$$\vec{H} = \sum_{i=1}^{n}\left[\left(\vec{r}_i^{\,COM} - \vec{r}_{body}^{\,COM}\right) \times m_i\left(\vec{v}_i^{\,COM} - \vec{v}_{body}^{\,COM}\right) + I_i\vec{\omega}_i\right] \quad (2)$$

where $\vec{r}_i^{\,COM}$, $\vec{v}_i^{\,COM}$ are the position and velocity vectors of the i^{th} segment's CoM, respectively. $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ are the position and velocity vectors of the whole-body CoM, m_i, I_i and $\vec{\omega}_i$ are the mass, moment of inertia and angular velocity vector of the i^{th} segment, respectively, and n is the number of body segments. H was transformed into each anatomical plane using equation (1) and low-pass filtered at 4 Hz with a zero-lag 4th order Butterworth filter. The number of peaks was determined using the *findpeaks* function in Matlab (Mathworks, Natick, MA) with a minimum peak prominence of 0.01.

Balance control was defined as the peak-to-peak range of H (H_R) in each anatomical plane normalized by the respective subject's mass, height and speed before the turn. Speed before the turn was defined as the average COM forward velocity of the stride (i.e., heel strike to heel strike) before the start of the turn. H was assessed using the same 13 segment model and again transformed into each anatomical plane using equation (1).

2.3. Statistical tests

To test the hypothesis that impaired muscle coordination affects turning performance, we compared turning measures between the following groups: controls and individuals post-stroke with 2, 3, 4 and 5 modules. One-way ANOVAs were used to test for significant differences between groups in each of the five turning performance measures. Bonferroni post-hoc corrections were used to identify differences between groups. Significance was defined as $p \leq 0.05$. Furthermore, we performed diagnostics of the residuals (Quantile-Quantile plots) to confirm normality and other assumptions. All assumptions were adequately met.

3. Results

On the paretic leg, one individual post-stroke had 5 modules during the straight-line walking trials, 12 had 4 modules, 14 had 3 modules and 7 had 2 modules. On the non-paretic leg, one individual post-stroke had 5 modules, 18 had 4 modules, 10 had 3 modules and 5 had 2 modules (Table 1). The control subjects used an average of 3.79 ± 0.42 modules (15 controls had 4 modules and 4 controls had 3 modules) on their outside leg and 3.74 ± 0.45 modules (14 controls had 4 modules and 5 controls had 3 modules) on their inside leg.

One subject (4 modules on their paretic and nonparetic legs) was excluded in the 180-degree turning analysis due to missing kinematic markers during those trials. In addition, we excluded the 5 module group (n = 1) from further analysis because the sample size was not large enough to statistically analyze.

3.1. Time to turn

Individuals post-stroke with two modules on their inside (paretic) leg took longer to complete the 180-degree turn compared to individuals post-stroke with four modules and controls ($p \leq 0.032$) (Fig. 2). Those with two modules on their outside (non-paretic) leg took a longer time to complete the 180-degree turn compared to controls ($p = 0.019$). No differences were observed between groups during the 90-degree turns ($p \geq 0.466$). The remaining p-values for all comparisons can be found in the Appendix (Tables A3 and A4).

3.2. Total number of steps

Individuals post-stroke with two modules on their inside (paretic) leg took more steps (5.6 ± 1.0 steps) during a 180-degree turn compared to controls (4.6 ± 0.9 steps) ($p = 0.036$; Fig. 3). No differences were observed between groups with different modules on their outside (non-paretic) leg for 180-degree turning trials ($p = 0.059$). For 90-degree turning trials, no differences were seen between groups ($p \geq 0.462$).

3.3. Turning smoothness

Those with two modules or four modules on their outside (non-paretic) leg had a higher velocity peaks metric compared to controls ($p \leq 0.033$) (Fig. 4). Those with two modules on their inside (paretic) leg were less smooth compared to controls ($p = 0.036$). There were no differences in smoothness between groups during the 90-degree turning task ($p \geq 0.496$).

Individuals post-stroke with two modules on their outside (non-paretic) or inside (paretic) leg had a higher H peaks metric (i.e., less smooth balance control) in the transverse plane for the 180-degree turn compared to controls ($p \leq 0.035$) (Fig. 5). In addition, individuals post-stroke with 4 modules on their outside leg ($p = 0.021$) and 3 modules on their inside leg ($p = 0.027$) had a higher H peaks metric in the transverse plane for the 90-degree turn compared to controls. No differences were observed between groups in the sagittal ($p \geq 0.478$) or frontal ($p \geq 0.142$) planes for

Table 1
Number of modules for the paretic and non-paretic legs.

Group	Total Number of Subjects	
	Paretic Leg	*Non-Paretic Leg*
Controls	19	19
Post-Stroke- 5 Modules	1	1
Post-Stroke- 4 Modules	12	18
Post-Stroke- 3 Modules	14	10
Post-Stroke- 2 Modules	7	5

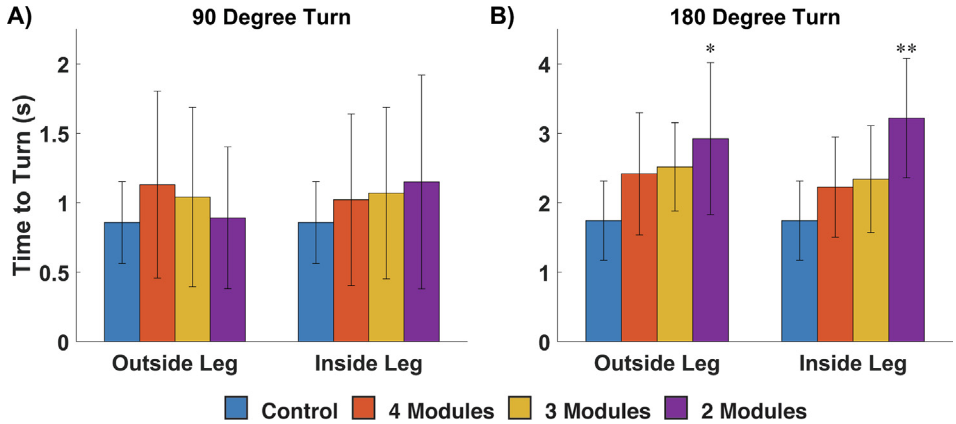

Fig. 2. Time to turn for (A) 90-degree trials, and (B) 180-degree trials, grouped by controls and the number of modules used during straight line walking for post-stroke individuals. "*" indicates a significant difference from controls. "**" indicates a significant difference from controls and 4 modules. Error bars represent standard deviation.

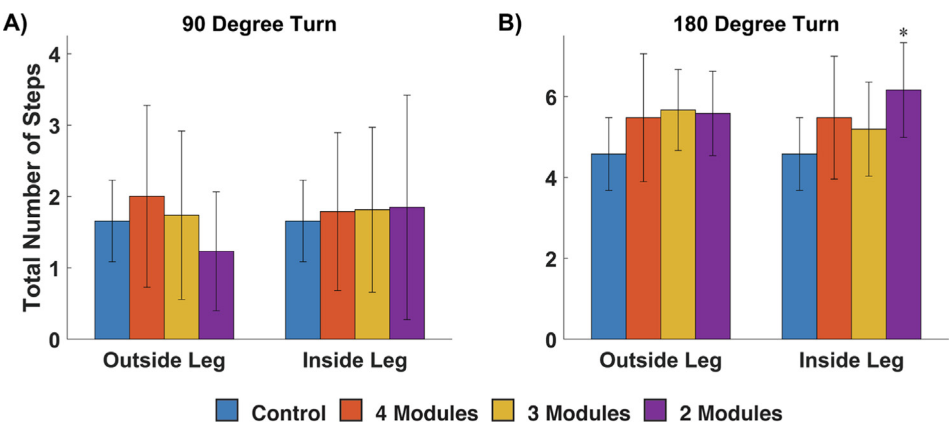

Fig. 3. Total number of steps taken for (A) 90-degree turning trials, and (B) 180-degree turning trials, grouped by controls and the number of modules used in straight line walking for post-stroke individuals. "*" indicates a significant difference from controls. Error bars represent standard deviation.

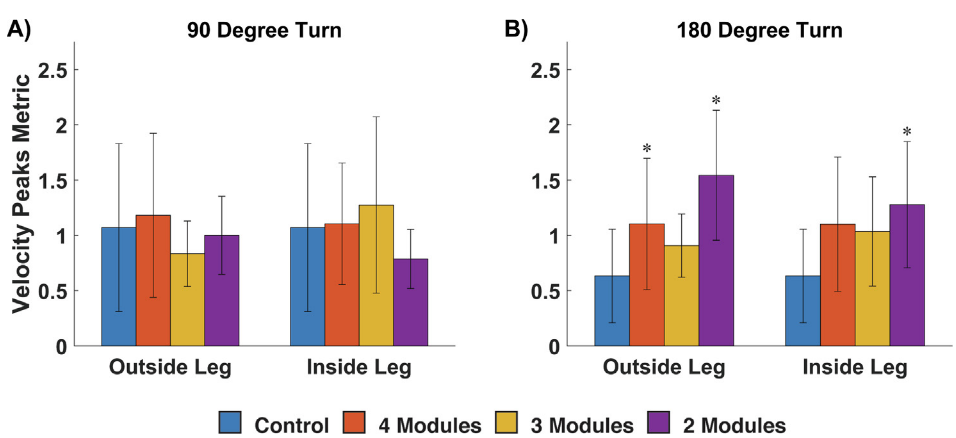

Fig. 4. Velocity peaks metric (i.e., number of body COM forward velocity peaks during a turn normalized by the number of steps taken) for (A) 90-degree turning trials, and (B) 180-degree turning trials, grouped by controls and the number of modules used in straight line walking for post-stroke individuals. "*" indicates a significant difference from controls. Error bars represent standard deviation.

the 180-degree turn. No differences in the H peaks metric were observed between groups during the 90-degree turning task in any plane ($p \geq 0.140$).

3.4. Balance control

For both turning tasks, there were differences in peak-to-peak H in the frontal ($p \leq 0.001$), sagittal ($p \leq 0.001$) and transverse ($p \leq 0.048$) planes for 180-degree and 90-degree turning trials (Fig. 6). On average, subjects with reduced module complexity on the paretic (inside) limb had higher H_R compared to controls in all planes.

4. Discussion

The aim of this study was to analyze the influence of impaired muscle coordination on turning performance in individuals

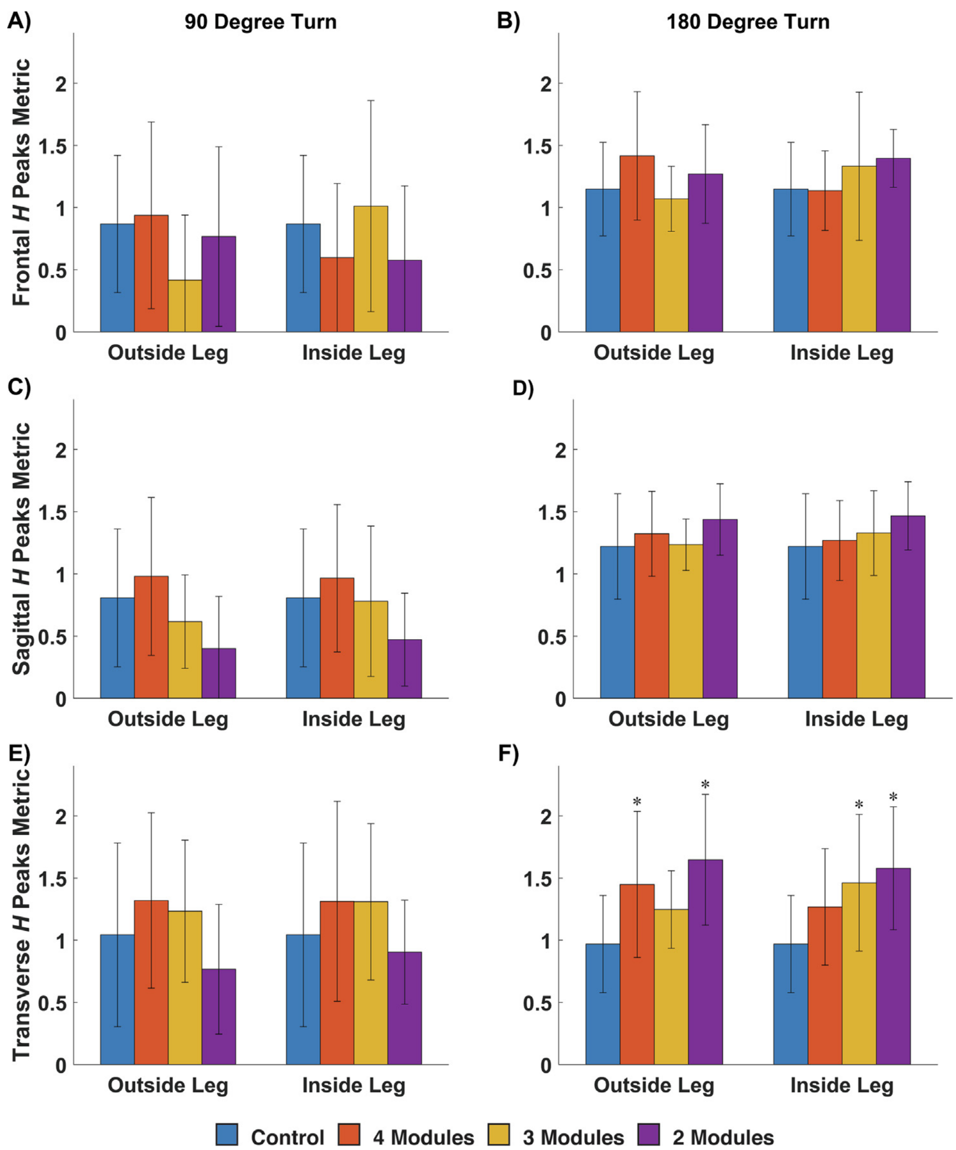

Fig. 5. Whole body angular momentum (H) peaks metric for 90-degree turning trials (A, C, E) and 180-degree turning trials (B, D, F) for the frontal (A, B), sagittal (C, D) and transverse (E, F) planes, grouped by controls and the number of modules used in straight line walking for post-stroke individuals. "*" indicates a significant difference from controls. Error bars represent standard deviation.

post-stroke. We hypothesized that those with merged modules would have impaired 90-degree and 180-degree turning performance, which was defined by the time to complete a turn, number of steps taken, turning smoothness and balance control. We found that individuals post-stroke with 2 modules had the most impaired 180-degree turning performance.

Contrary to our hypothesis, differences in time to turn, turning smoothness, and number of steps were only observed during the 180-degree turn, but not the 90-degree turn. A 180-degree turn is a more complex task requiring more steps and a greater range of body rotation, further exposing an individual's degree of motor impairment. Individuals post-stroke with 2 modules took a longer time to complete the 180-degree turn, suggesting that the time it takes to complete a 180-degree turn may give clinical insight into post-stroke muscle coordination impairment. Currently, a limited number of tests assess turning performance for individuals post-stroke, such as the Emory Functional Ambulation Profile (Wolf et al., 1999), the performance oriented mobility assessment (Tinetti, 1986) and the Berg Balance Scale (Berg et al., 1992). These tests assess a timed up-and-go (TUG) activity and a 360-degree turn test where individuals walk in a complete circle. Further, a Component TUG test separately assesses the 180-degree component of the TUG test (Clemens et al., 2018). This present study provides further support for the use of 180-degree assessments to evaluate motor impairment in individuals post-stroke.

As expected, those with 2 modules took a greater number of steps during a 180-degree turn compared to controls. Studies show elderly adults, who report difficulty in turning, use a multiple step strategy (5 or more steps) to execute a 180-degree turn (Thigpen et al., 2000) instead of a pivot strategy (i.e., pivoting on their back leg to change direction). In the present study, post-stroke individuals with two modules took an average of 5.6 ± 1.0 steps. Thus, it is likely those with 2 modules may be using a multiple step strategy to increase their perception of stability.

In addition, as expected, those with reduced module complexity had higher H_R in the frontal plane for both turns. However, contrary to our hypothesis, those with reduced module complexity also had higher H_R in the sagittal and transverse planes compared to controls. This finding is partially in agreement with previous analyses of straight-line walking that found individuals with

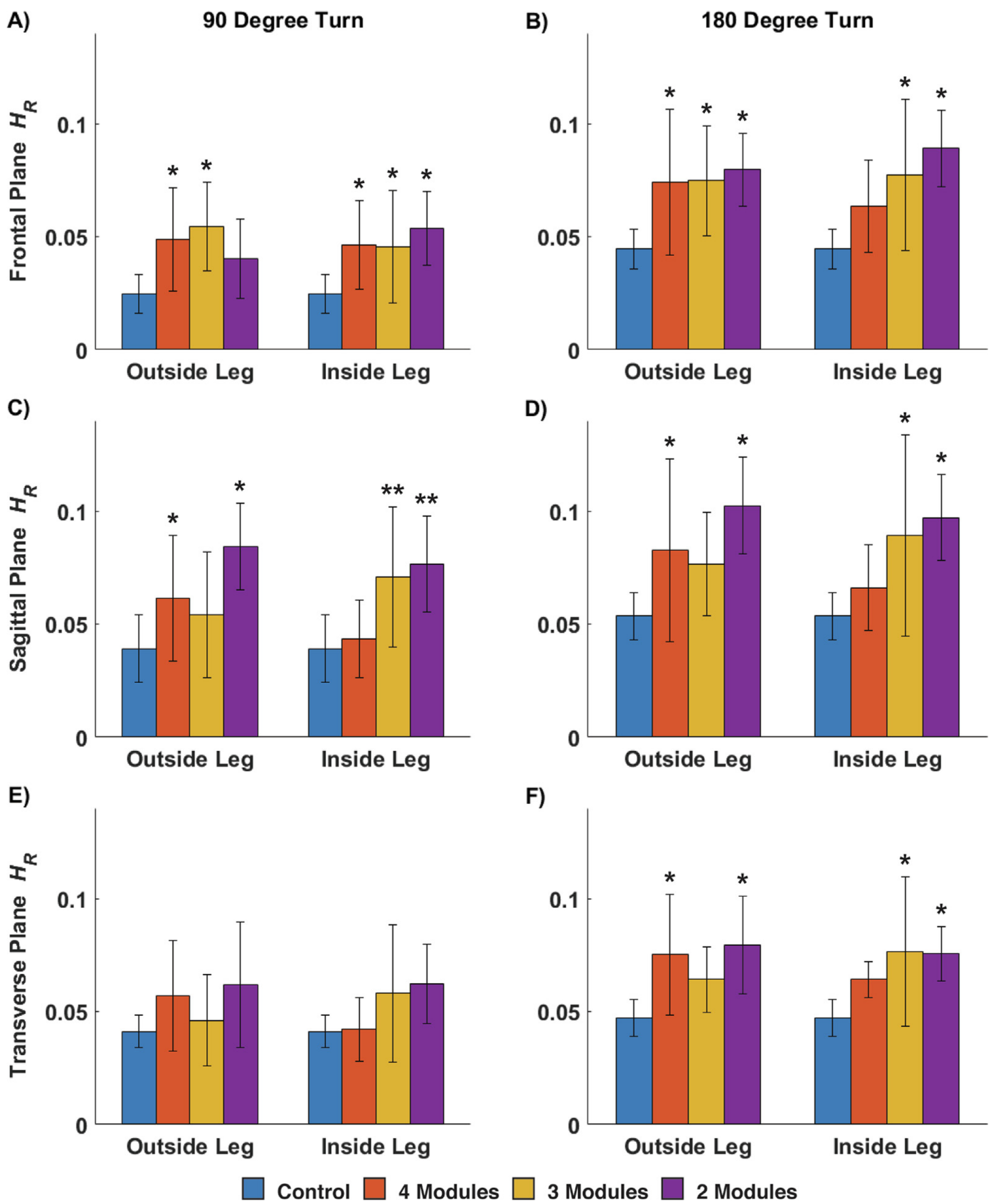

Fig. 6. Peak-to-peak whole body angular momentum normalized by mass, height, and speed before the turn (H_R) for 90-degree turning trials (A, C, E) and 180-degree turning trials (B, D, F) for the frontal (A, B), sagittal (C, D) and transverse (E, F) planes, grouped by controls and the number of modules used in straight line walking for post-stroke individuals. "*" indicates a significant difference from controls. "**" indicates a significant difference from controls and 4 modules. Error bars represent standard deviation.

lower-limb amputations have higher H_R in the sagittal and frontal planes, with no reported differences in the transverse plane (Silverman and Neptune, 2011). Because turning involves rotation of the body around the transverse plane, more transverse plane balance control is required, which would lead to higher H_R during 180-degree and 90-degree turns for those with reduced module complexity.

We found differences in smoothness for all module groups, with the largest difference observed in the 2 module group. Those with 2 modules had less smooth movement (both linear and angular metrics) during 180-degree turns. This is consistent with research showing an association between upper limb motor impairment and a larger number of peaks in hand velocity during a planar reaching task (Rohrer et al., 2002). Similarly, others have shown that slower circular turning speeds result in a non-uniform body COM trajectory, with sharp peaks in the trajectory during single-leg stance of the outer limb (Orendurff et al., 2006). Our results show individuals post-stroke with impaired muscle coordination, who take a longer time to complete a 180-degree turn, have a less smooth velocity profile. These differences were only observed in the transverse plane where the majority of turning movement occurs.

4.1. Limitations and future work

One potential limitation of this study is that we did not instruct all subjects to use the same turning strategy. Individuals used a variety of 90-degree turning strategies, including step turns (i.e., a change in direction with the outside limb) and spin turns (i.e., a pivot on the inside limb). Similarly, individuals can complete a 180-degree turn with one pivot turn, where they turn their body over the foot in one controlled movement, or with multiple steps and weight shifts (Thigpen et al., 2000). This results in different turning velocities and COM trajectory changes with the type of strategy used (Yamaguchi et al., 2017). As a result, we could not directly compare each turn. This could also affect turning performance metrics. Future research should focus on comparing turning metrics between different turning strategies in order to further evaluate the effect of muscle coordination impairment on turning performance.

In addition, research has shown that the number of modules change depending on the type of task performed (Barroso et al., 2014; Allen et al., 2019). Because different turning strategies were used (e.g., a step turn versus a spin turn), we chose not to directly compare the number of modules used during turning. Instead, we used the number of modules during straight-line walking to classify muscle coordination impairment. Since turning and straight-line walking execute similar biomechanical functions (e.g., body support, body angular control, foot clearance, leg propulsion), we believe the number of modules used during straight-line walking is a useful measure of muscle coordination impairments during turning. To test this assumption, we ran a post-hoc analysis to determine the average VAF when turning modules were forced to use the same muscle weightings as calculated during walking. The average VAF for all turns across all muscles was 90.63%. Thus, we believe our walking module groupings are generalizable to turning.

Another potential limitation of this study was the absence of kinetic data. Because we allowed participants to choose any turning strategy and did not constrain their steps to force plates, we were not able to record kinetic data. Such data could help identify additional differences in turning performance (e.g., contributions to body support and propulsion, and joint mechanical work). Individual modules may be organized to perform specific biomechanical functions during straight-line walking (Neptune et al., 2009), and merging of modules may impair the ability to effectively accomplish these tasks. Thus, future studies with kinetic data during turning could provide additional insight into the relationships between impaired muscle coordination and turning performance.

Finally, we instructed participants to turn as quickly as possible when completing the 180-degree and 90-degree turns. Because we used time to turn as a performance metric, we wanted to obtain their best performance. This might not represent how they turn in real life and could affect other turning performance metrics. Future studies could seek to assess the influence of the instructions given on turning performance metrics.

5. Conclusions

In summary, when modules are merged post-stroke, 180-degree turning performance is impaired. Individuals post-stroke with 2 modules took longer to turn, used more steps and had less smooth movement. In addition, those with reduced module complexity exhibited diminished balance control during 90-degree and 180-degree turning. Recent studies have shown that individuals post-stroke can gain independent control of modules through rehabilitation (Routson et al., 2013). Thus, obtaining independent modules should be an important aim in turning rehabilitation. In addition, post-stroke clinical assessments of motor impairment severity are necessary for prescribing effective therapy. The time it takes to complete a 180-degree turn may provide useful clinical insight into impaired muscle coordination post-stroke.

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Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Appendix A

Tables A1–A4.

Table A1
Control group subject demographics and results.

Subject	Age (years)	Sex	Mass (kg)	Height (m)	# Modules		Time to Turn	
					Left Leg	Right Leg	90° (s)	180° (s)
1	56	Male	113	1.85	3	4	0.78	1.99
3	48	Male	88	1.70	4	4	0.71	1.56
4	62	Male	112	1.82	3	4	0.83	1.19
5	47	Female	90	1.65	3	3	0.84	1.92
6	60	Female	78	1.65	4	4	1.55	1.71
7	76	Female	70	1.64	4	4	0.91	1.34
8	63	Male	90	1.80	4	4	0.9	1.06
9	44	Male	79	1.74	4	4	0.81	1.04
10	52	Female	79	1.71	4	4	0.51	1.31
11	68	Female	73	1.60	4	4	0.95	1.6
12	57	Female	99	1.70	4	3	1.26	2.62
13	79	Male	80	1.75	4	4	0.47	1.42
14	62	Male	118	1.74	4	3	0.46	2.61
15	55	Female	114	1.72	4	4	0.48	1.62
16	61	Female	63	1.58	4	4	1.02	1.62
17	69	Female	87	1.75	3	4	0.98	3.2
18	51	Female	74	1.65	4	4	1.34	2.05
20	62	Female	82	1.62	4	3	0.81	1.93
28	40	Male	76	1.88	4	3	0.69	1.33
Mean ± SD	**59 ± 10**	F=11,M=8	**88 ± 16**	**1.71 ± 0.08**	**3.8 ± 0.4**	**3.7 ± 0.5**	**0.86 ± 0.29**	**1.74 ± 0.57**

Table A2
Post-stroke subject demographics and results. * indicates the type of stroke unknown.

Subject	Age (years)	Gender	Mass (kg)	Height (m)	Time Since Stroke		Paretic Side	Type of Stroke
					years	months		
19	72	Male	59	1.75	3	3	R	Ischemic
21	64	Male	74	1.70	1	4	L	Ischemic
22	66	Female	87	1.64	7	7	R	Ischemic
23	72	Female	69	1.52	2	4	L	Ischemic
24	55	Male	100	1.82	1	5	L	Hemorrhage
25	48	Female	88	1.65	3	6	R	Ischemic
26	58	Male	114	1.75	5	11	L	Ischemic
27	57	Female	111	1.68	4	5	L	Ischemic
29	68	Male	93	1.67	7	0	R	Ischemic
30	52	Male	99	1.93	3	8	R	Ischemic
31	52	Male	93	1.83	0	6	R	Ischemic
32	76	Male	84	1.78	2	4	L	Hemorrhage
33	74	Male	102	1.52	11	2	R	Ischemic
34	46	Female	60	1.57	2	7	R	Ischemic
35	62	Female	103	1.54	0	8	R	Ischemic
36	65	Male	76	1.61	5	3	R	Hemorrhage
37	70	Female	77	1.66	2	10	R	Ischemic
38	73	Male	87	1.83	5	2	R	*
39	50	Male	122	1.78	2	8	R	Ischemic
40	54	Male	149	1.82	4	11	L	Ischemic
41	55	Female	63	1.52	20	3	R	Hemorrhage
42	57	Male	86	1.75	3	2	R	Ischemic
43	69	Male	83	1.74	0	7	R	Ischemic
44	64	Male	77	1.69	11	2	R	Hemorrhage
46	65	Male	74	1.67	1	2	R	Hemorrhage
47	54	Male	112	1.85	1	5	R	Ischemic
49	62	Female	90	1.68	2	2	L	Ischemic
51	68	Female	83	1.65	1	7	L	Ischemic
52	57	Male	115	1.69	6	4	R	*
53	68	Male	108	1.76	2	7	L	Ischemic
54	55	Male	139	1.80	10	1	R	Ischemic
55	34	Female	64	1.50	2	2	R	Ischemic
56	37	Male	70	1.62	15	3	L	Hemorrhage
57	78	Male	100	1.74	5	4	L	Ischemic
Mean	**61 ± 11**	F=11,M=23	**92 ± 21**	**1.70 ± 0.11**	**4.4 ± 4.5**	**4.9 ± 2.9**		

Subject	# Modules		Time to Turn	
	Paretic Leg	Nonparetic Leg	90° (s)	180° (s)
19	4	3	2.34	1.89
21	4	4	1.69	1.71
22	3	3	0.63	2.54
23	3	3	1.92	2.1
24	2	2	1.68	4.66
25	3	3	1.02	2.05
26	2	2	0.86	3.18
27	2	2	0.4	2.62
29	3	5	1.5	1.8
30	4	4	0.5	1.98
31	4	3	0.59	2.28
32	4	4	0.49	1.57
33	3	4	0.55	1.78
34	3	4	1.11	1.7
35	3	3	1.33	2.59
36	2	4	2.63	4.17
37	4	4	1.31	2.65
38	3	2	1.03	2.4
39	3	4	0.92	1.74
40	4	3	0.39	2.25
41	4	4	0.55	1.97
42	3	4	2.64	4.27
43	2	4	0.92	2.56
44	3	2	0.49	1.76
46	4	4	0.94	1.76
47	2	3	1.02	2.48
49	5	4	1.43	2.91
51	4	4	1.65	1.75
52	2	3	0.54	2.87
53	4	4	1.18	2.52
54	3	4	0.82	1.86
55	3	4	0.45	2.5
56	3	4	0.57	3.67
57	4	3	0.63	4.13
Mean ± SD	**3.2 ± 0.8**	**3.4 ± 0.8**	**1.08 ± 0.63**	**2.49 ± 0.83**

Table A3
One-way ANOVA Results. "*" denotes a p-value $\le$ 0.05.

Variable	p-values			
	180-degree turns		90-degree turns	
	Inside Leg	Outside Leg	Inside Leg	Outside Leg
Time to Turn	<0.001*	0.005*	0.566	0.466
Velocity Peaks Metric	0.014*	0.002*	0.496	0.615
H Peaks (Frontal Plane)	0.386	0.142	0.322	0.213
H Peaks (Sagittal Plane)	0.478	0.587	0.334	0.140
H Peaks (Transverse Plane)	0.009*	0.007*	0.447	0.358
Peak-to-Peak H (Frontal Plane)	<0.001*	<0.001*	<0.001*	<0.001*
Peak-to-Peak H (Sagittal Plane)	<0.001*	0.002*	<0.001*	0.001*
Peak-to-Peak H (Transverse Plane)	<0.001*	<0.001*	0.014*	0.048*
Total Steps	0.020*	0.059	0.962	0.462

Table A4
Bonferroni Post-Hoc Comparisons. "*" denotes a p-value $\le$ 0.05.

Variable	Type of Turn	Leg	*p*-value for each comparison					
			Control vs. 4 Modules	Control vs. 3 Modules	Control vs. 2 Modules	4 Modules vs. 3 Modules	4 Modules vs. 2 Modules	3 Modules vs. 2 Modules
Time to Turn	180	Inside	0.455	0.119	<0.001*	1	0.032*	0.057
Time to Turn	180	Outside	0.06	0.069	0.019*	1	1	1
Total Steps	180	Inside	0.279	0.831	0.021*	1	1	0.474
Velocity Peaks	180	Inside	0.112	0.173	0.036*	1	1	1
Velocity Peaks	180	Outside	0.033	0.914	0.003*	1	0.475	0.121
H peaks metric (Transverse Plane)	180	Inside	0.598	0.027*	0.030*	1	1	1
H peaks metric (Transverse Plane)	180	Outside	0.021*	0.808	0.035*	1	1	0.751
Peak-to-Peak H (Frontal Plane)	180	Inside	0.148	<0.001*	<0.001*	0.715	0.107	1
Peak-to-Peak H (Sagittal Plane)	180	Inside	1	0.002*	0.004*	0.211	0.121	1
Peak-to-Peak H (Transverse Plane)	180	Inside	0.130	<0.001*	0.009*	0.675	1	1
Peak-to-Peak H (Frontal Plane)	180	Outside	0.002*	0.009*	0.022*	1	1	1
Peak-to-Peak H (Sagittal Plane)	180	Outside	0.015*	0.209	0.005*	1	0.968	0.546
Peak-to-Peak H (Transverse Plane)	180	Outside	<0.001*	0.144	0.007*	0.909	1	0.862
Peak-to-Peak H (Frontal Plane)	90	Inside	0.012*	0.011*	0.004*	1	1	1
Peak-to-Peak H (Sagittal Plane)	90	Inside	1	<0.001*	0.002*	0.014*	0.014*	1
Peak-to-Peak H (Transverse Plane)	90	Inside	1	0.083	0.084	0.211	0.167	1
Peak-to-Peak H (Frontal Plane)	90	Outside	<0.001*	<0.001*	0.531	1	1	0.858
Peak-to-Peak H (Sagittal Plane)	90	Outside	0.028*	0.614	0.002*	1	0.342	0.124
Peak-to-Peak H (Transverse Plane)	90	Outside	0.101	1	0.237	0.956	1	0.866

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