Merged plantarflexor muscle activity is predictive of poor walking performance in post-stroke hemiparetic subjects

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ABSTRACT

Stroke is the leading cause of long-term disability and individuals post-stroke often experience impaired walking ability. The plantarflexor (PF) muscles are critical to walking through their contributions to the ground reaction forces and body segment energetics. Previous studies have shown muscle activity during walking can be grouped into co-excited muscle sets, or modules. Improper co-activation, or merging of modules, is a common impairment in individuals post-stroke. The purpose of this study was to determine the influence of merged PF modules on walking performance in individuals post stroke by examining balance control, body support and propulsion, and walking symmetry. Muscle modules were identified using non-negative matrix factorization to classify subjects as having an independent or merged PF module. The merged group had decreased balance control with a significantly higher frontal plane whole-body angular momentum than both the independent and control groups, while the independent and control groups were not significantly different. The merged group also had higher paretic braking and nonparetic propulsion than both the independent and control groups. These results remained when comparisons were limited to subjects who had the same number of modules, indicating this was not a general effect due to subjects with merged PF having fewer modules. It is likely that a merged PF module is indicative of general PF dysfunction even when some activation occurs at the appropriate time. These results suggest an independent PF module is critical to walking performance, and thus obtaining an independent PF module should be a crucial aim of stroke rehabilitation.

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1. Introduction

Stroke is the leading cause of long-term disability in the United States with approximately 795,000 people experiencing a stroke each year (Benjamin et al., 2017). Individuals post-stroke often experience reduced mobility and impaired balance control, with over 70% falling at least once within six months of leaving the hospital (Forster and Young, 1995). Falls can lead to long-term injuries, limited community involvement and reduce quality of life. Thus, there is a need for more targeted and efficient rehabilitation to maximize walking performance while reducing time spent in therapy.

The ankle plantarflexors (PFs), including the soleus (SOL) and gastrocnemius (GAS), are critical to walking performance as they are primary contributors to important biomechanical functions such as body support, forward propulsion and leg swing initiation (Anderson and Pandy, 2003; McGowan et al., 2008; Neptune et al., 2001). However, PF output is often impaired post-stroke. For example, simulation analyses have shown PF activation and function is impaired in individuals post-stroke (Higginson et al., 2006; Knarr et al., 2013), while others have shown those with limited community ambulation can have reduced paretic PF contributions to propulsion (Peterson et al., 2010) and paretic propulsion has been shown to correlate with hemiparetic severity (Bowden et al., 2006).

The PFs also contribute to balance control through mediolateral acceleration of the body's center of mass (Pandy et al., 2010) and regulation of frontal plane angular momentum (H) (Neptune and McGowan, 2016). The range of frontal plane angular momentum (H_R) has been shown to correlate with common clinical balance measures and can reveal the underlying mechanisms affecting balance in hemiparetic walking, with post-stroke subjects exhibiting a higher H_R due to poor regulation of H in early stance (Nott et al., 2014).

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Appropriate muscle activity is critical to the execution of these biomechanical functions and previous studies have shown that muscle activity during walking can be grouped into sets of co-excited muscles or modules (Cappellini et al., 2006; Clark et al., 2010; Ivanenko et al., 2004). These modules may originate from neuroplasticity caused by repeated activities and may reduce the computational expense involved in choosing muscle coordination strategies (Ting et al., 2015). Simulation studies have shown that well-coordinated walking in healthy subjects can be produced by five co-activation modules: (1) hip and knee extensors in early stance, (2) ankle plantarflexors (PFs) in late stance, (3) tibialis anterior and rectus femoris in swing, (4) hamstrings in late swing and early stance, and (5) hip flexors in pre- and early swing (Allen and Neptune, 2012; Neptune et al., 2009). Most experimental studies do not include hip flexor activity measurements (e.g., iliopsoas), and thus only identify modules 1–4 (e.g., Clark et al., 2010).

Research has found that in post-stroke hemiparetic walking, modules in the paretic leg can merge, or become co-activated, resulting in reduced modular complexity. Merged modules prevent the independent activation of specific muscles, which causes suboptimal execution of biomechanical functions (Clark et al., 2010). Improper co-activation of muscles is a common impairment in individuals post-stroke, with the PFs often co-activating prematurely with the hip and knee extensors in early stance (Den Otter et al., 2007; Higginson et al., 2006; Knutsson and Richards, 1979). One mechanism identified for increased co-activation of the PFs and knee extensors is an increase in intersegmental facilitative pathways between the paretic leg knee extensors and soleus, suggesting that the co-activation is related to changes in neural pathways post-stroke (Dyer et al., 2014). However, it is unclear how a PF module merging with any other module would influence walking performance. Based on the critical functional roles of the PFs in unimpaired walking, we hypothesize that a merged PF module would result in (1) increased paretic leg braking, (2) reduced paretic leg body support, (3) increased stepping asymmetry, and (4) higher frontal plane H_R compared to both healthy controls and individuals post-stroke with an independent PF module.

2. Methods

2.1. Experimental data

Kinematic, kinetic and electromyography (EMG) data were collected from 56 hemiparetic post stroke individuals (22 left hemiparesis, 29 female; age: 57 ± 13 years) and 17 healthy controls (7 female, age 55 ± 8 years). Subjects gave informed consent to participate in this IRB-approved study; see Tables A1–4 in the Appendix for demographics. The subjects walked on a split-belt instrumented treadmill (Bertec, Columbus, Ohio) at their self-selected speed. Prior to data collection, the subjects practiced treadmill walking to get comfortable with the experimental setup. Subjects walked for at least 10 s to reach a steady-state walking pattern before each 30-second trial. Kinematic data were collected at 120 Hz using a twelve-camera motion capture system (PhaseSpace, Inc.) and a modified Helen Hayes marker set. Kinematic and kinetic data were processed with a low-pass fourth-order Butterworth filter at 9 Hz and 20 Hz, respectively. EMG data were collected (Motion Labs) at 1000 Hz from bilateral electrodes placed on the GAS, SOL, tibialis anterior (TA), rectus femoris (RF), gluteus medius (GM), vastus medialis (VM), lateral hamstrings (LH), and medial hamstrings (MH). EMG data were high-pass filtered with a zero-lag fourth-order Butterworth filter at 40 Hz, demeaned, rectified and low-pass filtered with a zero-lag fourth-order Butterworth filter at 4 Hz. For each muscle, the filtered signal was normalized to its peak value during each trial. Each step was normalized to 100 percent of the gait cycle and then averaged across steps.

2.2. Data analysis

The processed EMG signals were analyzed using nonnegative matrix factorization (NNMF) as previously described (Clark et al., 2010). NNMF determined the minimum number of muscle modules required to account for >90% of the EMG variability and the weighted contribution of each muscle to the module. Each subject's module compositions were compared to the four average control modules using Pearson's correlations. If a subject had a PF module with a Pearson's correlation

coefficient of 0.8 or greater compared to the average control PF module, they were classified as having an independent PF module. If that criterion was not met, the subject was classified as having a PF module that was merged with another muscle group.

Walking performance was assessed by examining balance control, body support and propulsion, and walking symmetry. Balance control was assessed using the range of frontal plane whole-body angular momentum (H_R). Whole-body angular momentum (H) was determined using a 13-segment inverse dynamics model created in Visual3D (C-Motion, Germantown, MD) and summing the angular momentum of each body segment about the whole-body center of mass in the frontal plane. Whole-body angular momentum was normalized by subject mass, walking speed and leg length. H_R was defined as the difference between the highest positive and lowest negative peaks of H and averaged over all strides. Contributions from each leg to body support were calculated from the time integral of the vertical GRF, averaged across all steps and normalized by body weight. Contributions from each leg to braking and propulsion were calculated from the time integral of the negative and positive regions of the anterior and posterior GRF, respectively. For each subject, braking and propulsion were averaged across all steps and normalized by body weight and walking speed. Walking symmetry between the paretic and nonparetic legs was assessed using both stance and stepping measures. Stance symmetry measures were defined as the paretic propulsion ratio (PP), paretic braking ratio (PB) and paretic body support ratio, which were defined as the paretic value divided by the sum of the paretic and non-paretic values, with 0.5 being perfectly symmetric. Stepping symmetry measures were defined as the percentage of total stance on the paretic and nonparetic legs and paretic step ratio, or paretic step length divided by the sum of paretic and nonparetic step lengths (Balasubramanian et al., 2009).

2.3. Statistical tests

For each dependent measure (H_R, body support, braking and propulsion, and stance and stepping symmetries), ANOVAs and two-sample t-tests were used to test for significant differences between each group (independent PF module, merged PF module and control groups). A Bonferroni-Holm correction for multiple comparisons was used for the t-tests (uncorrected $p < 0.05$).

3. Results

3.1. Module analysis

Twenty-nine of the hemiparetic subjects had a merged PF module and sixteen hemiparetic subjects had an independent PF module (e.g., Fig. 1). Twelve subjects (one control and eleven post-stroke) were excluded from analysis due to poor EMG signals and/or missing kinematic markers. The average self-selected speeds of the merged, independent and control groups were 0.27 ± 0.12, 0.64 ± 0.12 and 0.75 ± 0.22 m/s, respectively. The merged group's self-selected speeds were significantly slower than the independent and control groups' ($p < 0.001$). The 0.12 m/s difference between the independent and control group's self-selected walking speeds did not reach significance ($p = 0.09$). See Tables A1–4 in the Appendix for individual subject demographics, Pearson's correlations and walking performance assessments.

3.2. Balance control

The merged group had a significantly higher H_R than both the independent and control groups ($p < 0.0001$) (Fig. 2). There was no difference between independent and control groups ($p = 0.12$). The increased H_R was not dependent on the differences in speed between the groups as the merged group had significantly higher H_R than the independent and control groups ($p = 0.003$ and $p < 0.001$, respectively) even without normalizing H by speed (Fig. 2).

3.3. Body support and propulsion and stance symmetry

On average, the merged group produced less body support with the paretic leg compared to the nonparetic leg (Fig. 3, Table 1). The paretic body support ratio for the merged group was significantly lower than the independent and control groups (p < 0.001 for
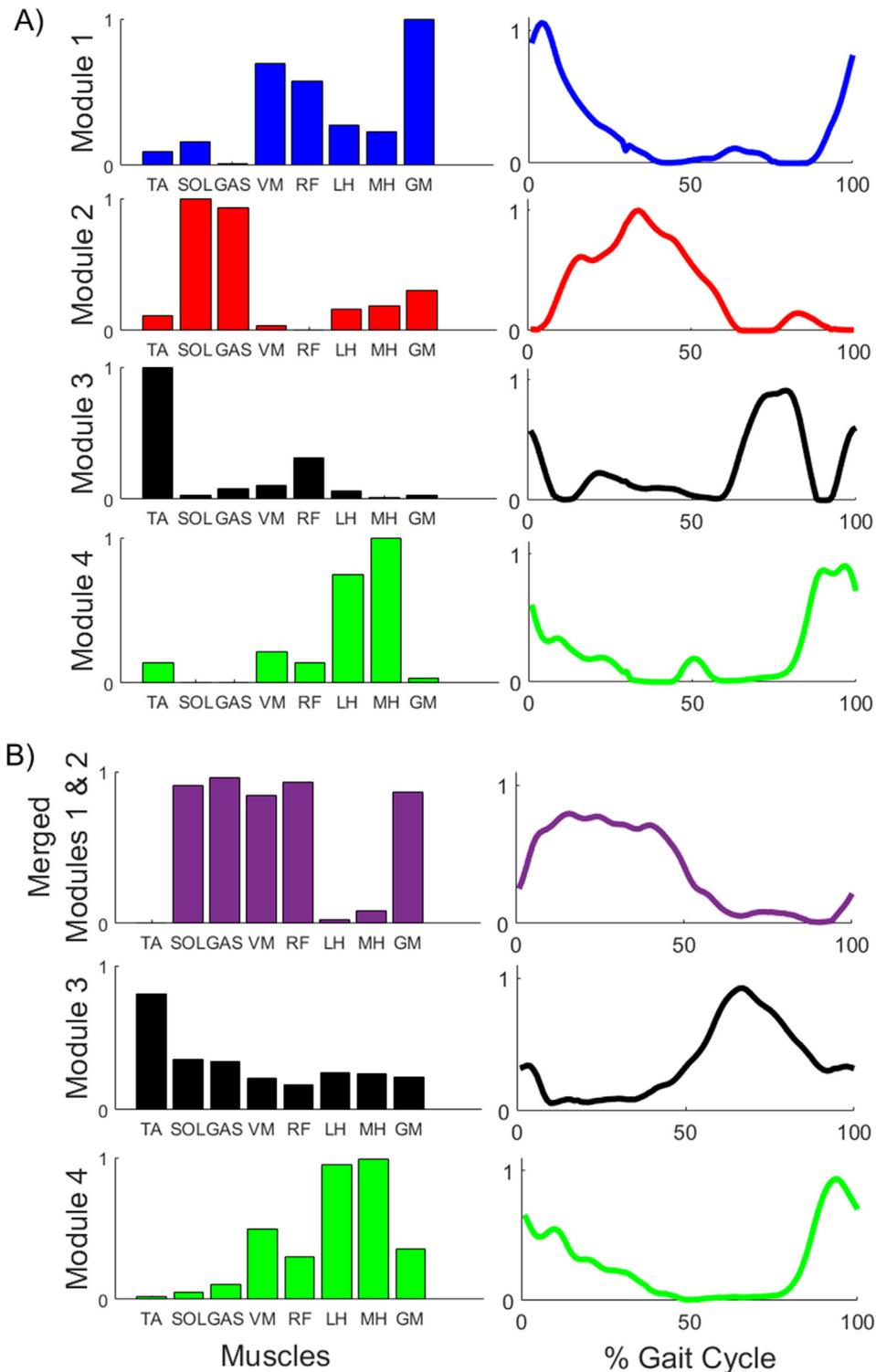
Fig. 1. Representative NNMF results, where each row represents one module and shows the weighted contribution of each muscle to that module (left column) and the activation of that module throughout a gait cycle (right column). Results are shown for: (A) control subject with four independent modules, and (B) subject with a PF module merged with module 1 (purple trace). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

both). The independent and control groups did not have significantly different paretic body support ratios ($p = 0.070$).

The control group had a significantly lower paretic (left) braking ratio and significantly higher paretic (left) propulsion ratio than the merged and independent groups ($p < 0.001$ for both groups and ratios, Table 1). The merged group had significantly higher

paretic braking and nonparetic propulsion compared to both the independent ($p = 0.002$, $p = 0.002$) and control groups ($p < 0.001$, $p < 0.001$, Fig. 4). For paretic propulsion and nonparetic braking, differences were not significant between any groups.

The merged group spent less time in stance on their paretic leg than on their nonparetic leg as a percentage of the total gait cycle
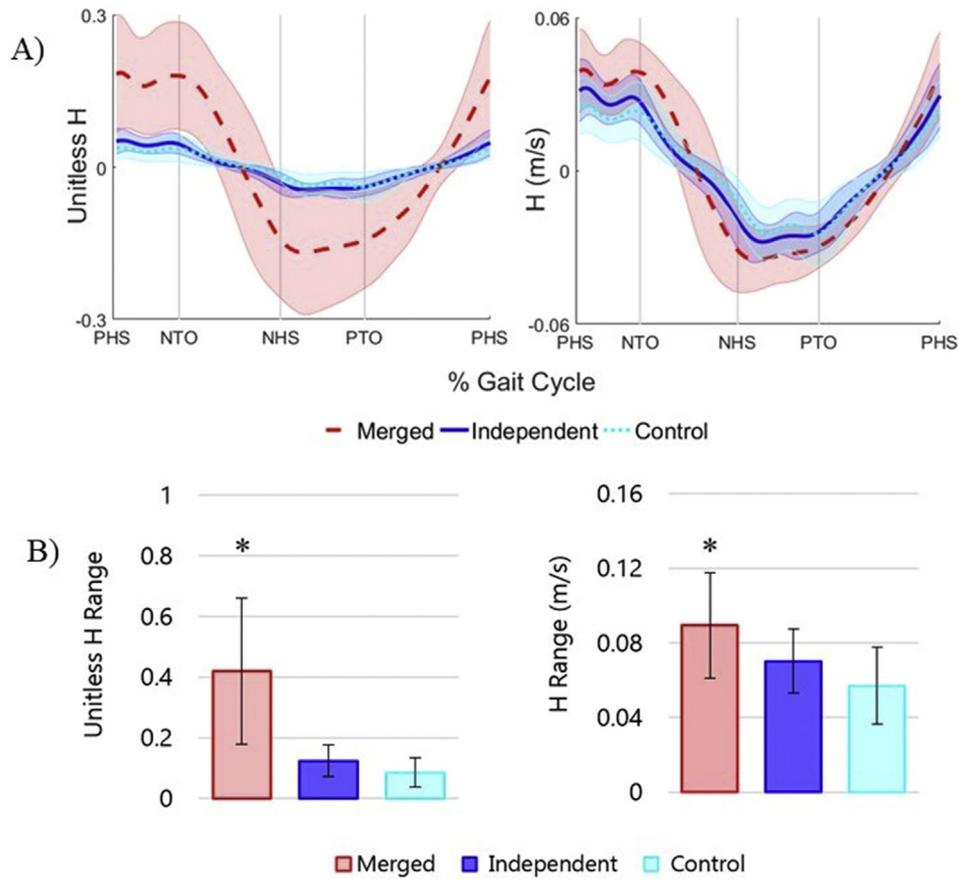
Fig. 2. Group averaged frontal plane (A) H and (B) $H_R \pm$ one standard deviation for the merged, independent and control groups. Results are shown for unitless H (normalized by body mass, leg length, and walking speed) and H not normalized by walking speed (units = m/s, normalized by body mass and leg length). Vertical lines show average paretic and nonparetic heel strikes (PHS, NHS) and toe-offs (PTO, NTO) as indicated on the horizontal axis. '*' indicates a significant difference between the H_R of the merged group and the independent and control groups.
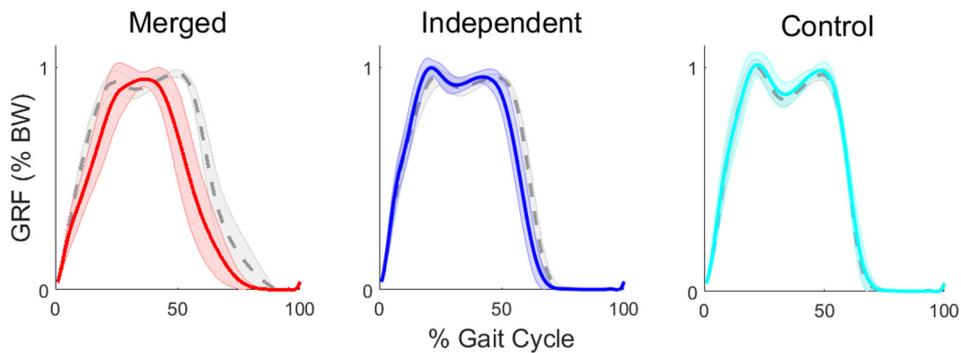

Fig. 3. Group averaged vertical ground reaction forces ± one standard deviation for the paretic leg (solid line) and non-paretic leg (dashed line) for the merged and independent groups (left and right sides of the control group, respectively).

Table 1
Paretic body support, braking and propulsion ratios ± one standard deviation for the merged, independent and control groups. A bolded value with '*' indicates the result was significantly different from both the other groups.

Group	Paretic body support (%)	Paretic propulsion (%)	Paretic braking (%)
Merged	**43 ± 4**[*]	33 ± 23	63 ± 18
Independent	49 ± 2	39 ± 10	58 ± 9
Control	50 ± 0	**55 ± 5**[*]	**46 ± 6**[*]

($p < 0.001$). There were no significant differences between paretic and nonparetic stance percentage in the independent and control groups.

3.4. Stepping symmetry

The merged group had highly variable step length asymmetries with a higher paretic step length ratio on average (0.58 ± 0.9) com-
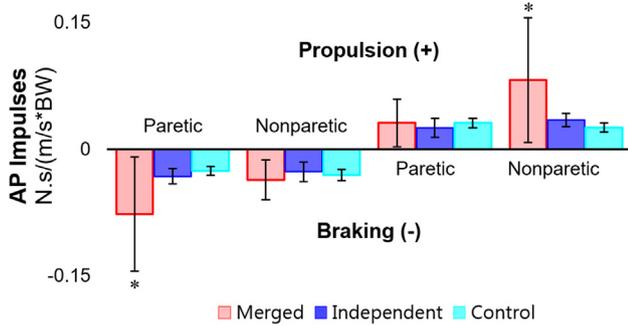
Fig. 4. Group averaged braking and propulsion ± one standard deviation (normalized by subject weight and walking speed) for the paretic and non-paretic legs of the merged and independent groups (left and right legs for the control group, respectively). '*' indicates significance between the merged group and both other groups.
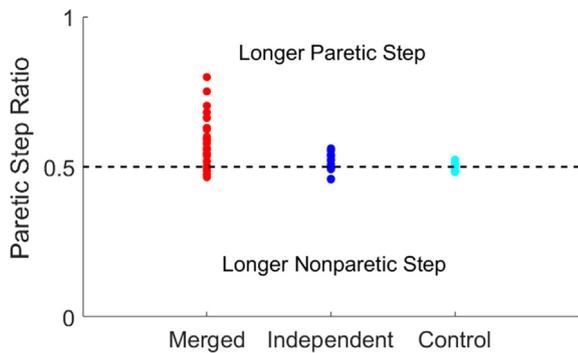

Fig. 5. Paretic step length ratio for each subject in the merged, independent and control groups. The horizontal dashed line indicates perfect symmetry.

pared to the merged and independent groups ($p = 0.002$, $p < 0.001$, Fig. 5). There was no significant step length asymmetry in the independent and control groups.

3.5. Controlling for the effect of fewer modules

The merged group had an average of 2.76 ± 0.69 total modules, the independent group had an average of 3.75 ± 0.58 total modules and the control group had an average of 3.75 ± 0.56 modules. Previous research has shown that subjects with fewer modules have slower self-selected walking speeds and greater stepping and propulsive asymmetry (Clark et al., 2010). Because the merged group had fewer modules on average than the other groups, an additional analysis was performed on the merged and independent subjects with three (n = 7 and n = 5, respectively) and four (n = 4 and n = 10, respectively) modules to isolate the effects of a merged PF module. Despite having the same number of modules, the

merged group still had significantly lower self-selected walking speeds and higher H_R, and higher paretic braking, nonparetic propulsion and propulsive asymmetries (Table 2).

4. Discussion

The goal of this study was to examine the influence of a merged PF module on walking performance post-stroke. Walking performance was assessed using the dependent measures of balance control, body support, braking and propulsion, and stance and stepping symmetries, which are directly related to the functional roles of the ankle plantarflexors in healthy walking (Higginson et al., 2006; Neptune et al., 2001; Neptune and McGowan, 2016). We found that subjects with a merged PF module had decreased performance in these measures, while participants with an independent PF module walked more similarly to the control group than the merged group.

4.1. Balance control

Previous research has shown individuals post stroke have a higher H_R than control subjects, indicating poor balance control (Nott et al., 2014). Our results support the expectation that a merged PF module would lead to poor balance control, with the merged group having a higher H_R than both the independent and control groups. Whole-body angular momentum is regulated through foot placement and GRF generation, with the plantarflexors being primary contributors to both the vertical and mediolateral GRFs (Neptune and McGowan, 2016). Thus, subjects who do not have independent control of their plantarflexor module are likely unable to modulate the timing and magnitude of their GRFs and adequately regulate frontal plane H and control their balance.

4.2. Body support and propulsion and stance symmetry

As expected, the merged PF group produced less paretic body support than the independent and control groups. While prolonged activity of modules 1 and 2 during paretic stance would seem to increase body support, quicker offloading of the paretic leg (i.e., lower percentage of paretic stance) resulted in a lower overall vertical impulse. A number of subjects in the merged group lacked the second peak in the vertical GRF associated with the PF push-off (Fig. 3), suggesting that the paretic leg functioned as a passive strut for body support rather than actively generating needed forces. While slower speeds are associated with less of a trough between the first and second peaks of the vertical GRF (Cook et al., 1997; Nilsson and Thorstensson, 1989) and the merged group did walk more slowly, the nonparetic leg did not exhibit single peak behavior to the extent of the paretic leg.

We expected that the merged group would have increased paretic leg braking due to the premature PF co-activation with the knee extensor group in early stance (Den Otter et al., 2007;

Table 2
Post-hoc comparison of merged and independent group subjects with the same total number of modules (either 3 or 4). A bolded p-value with '*' indicates a significant difference between the merged and independent groups.

Modules	Group	Speed (m/s)	Unitless H_R	Paretic braking (N·s/BW)	Nonparetic propulsion (N·s/BW)	Paretic ratio (%)	
						Braking	Propulsion
3	Merged	0.30 ± 0.13	0.35 ± 0.17	−0.08 ± 0.06	0.09 ± 0.07	67 ± 20	27 ± 24
	Independent	0.58 ± 0.08	0.12 ± 0.04	−0.03 ± 0.01	0.04 ± 0.00	59 ± 8	35 ± 10
	p-value	**0.002**[*]	**0.015**[*]	0.119	0.118	0.442	0.485
4	Merged	0.30 ± 0.09	0.38 ± 0.14	−0.06 ± 0.03	0.07 ± 0.04	65 ± 17	29 ± 26
	Independent	0.67 ± 0.12	0.10 ± 0.03	−0.03 ± 0.01	0.03 ± 0.01	49 ± 2	51 ± 3
	p-value	**<0.001**[*]	**0.044**[*]	0.150	0.181	0.513	0.462

Higginson et al., 2006). To maintain a constant walking speed (i.e., produce net zero anterior/posterior impulse), subjects who produced higher paretic braking would then have to produce higher nonparetic propulsion. Consistent with our expectations, the merged PF group had higher mean paretic braking and nonparetic propulsion than both the independent and control groups relative to walking speed.

Hemiparetic individuals generally spend a lower percentage of the gait cycle in paretic stance than in nonparetic stance (Patterson et al., 2010; von Schroeder et al., 1995). We observed this trend in the merged PF group but it was not significant in the independent group. A lower percentage of gait spent in paretic stance versus nonparetic stance is consistent with the observed reduced paretic body support in the merged group.

4.3. Stepping symmetry

Because GAS contributes to initiating leg swing (Neptune et al., 2001), we expected that the merged PF group would have altered spatiotemporal stepping characteristics following stroke. Generally, the merged group had greater step length asymmetry than the other groups. However, there was high variability between merged subjects (Fig. 5). Previous work found that a high paretic step length ratio was related to lower paretic propulsion and poor PF function (Allen et al., 2011). However, post-hoc analysis of these individuals showed no relationship between the percentage of paretic step length and their propulsive measures.

4.4. Modules

It is unlikely that the merging of modules is the only cause of post-stroke biomechanical abnormalities during gait, as the merging identified through matrix factorization may also be associated with poorly activated PFs such that even when the combined module is activated in late stance when healthy PFs should be activated, the PFs are not producing adequate force. Thus, the merged module is likely indicative of general dysfunction of the PFs even during the period of the gait cycle when the merged activation overlaps with the usual PF activation. Impaired coordination of module 1 (hip and knee extensors) when it is merged with the PF module could also affect results (e.g., prolonged knee extensor activation may prevent the knee from flexing during the swing phase (Yelnik et al., 1999). Post-stroke individuals may be able to improve walking without improving muscle coordination (Den Otter et al., 2006). However, these results further support that PF coordination is a strong predictor of gait performance.

4.5. Limitations and future work

One potential limitation of this study is that by having subjects walk at their self-selected speed, we did not control for possible speed-dependent differences in our dependent measures (Lelas et al., 2003; Zeni and Higginson, 2009). Since the merged group was also the slowest group, reduced speed may have added to the observed differences.

This study also raised some questions about the protocol for normalizing frontal plane whole-body angular momentum. Normalizing by mass, leg length, and walking speed is a common protocol for H calculations (e.g. Herr and Popovic, 2008; Silverman and Neptune, 2011). However, due to the much slower walking speed of the merged group, we also examined H_R without normalizing by speed to better understand the behavior of absolute H_R. Frontal plane H_R was still higher in the merged group than in the independent and control groups. Future work is needed to determine whether absolute or relative H_R is more important for assessing fall risk and balance control. It is possible that frontal plane H does not

scale linearly with walking speed, and other normalization techniques should be explored to compare subjects with a high range of body weight, height and walking speed.

5. Conclusions

In summary, we found that having an independent PF module is essential to walking performance. Individuals post-stroke whose PFs were co-activated with other muscle groups had slower walking speeds, decreased balance control and decreased walking symmetry. Thus, strategies should be developed to improve PF output during walking (Clark et al., 2016; Hsu et al., 2017). Previous research has shown that individuals can gain independent modules through locomotor training, even years after a stroke occurs (Routson et al., 2013). Thus, obtaining an independent PF module should be a priority in post-stroke rehabilitation.

Conflict of interest statement

The authors have no conflict of interest to declare.

Acknowledgements

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2018.11.011.

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