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Endpoint accuracy of goal-directed ankle movements correlates to over-ground walking in stroke

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HIGHLIGHTS

- Stroke individuals show decreased endpoint accuracy during fast, goal-directed ankle movements.
- Impaired endpoint accuracy is associated with increased coactivation of agonist-antagonist muscles.
- Endpoint accuracy of goal-directed ankle movements correlates to over-ground walking in stroke.

ABSTRACT

Objectives: Goal-directed movements are essential for voluntary motor control. The inability to execute precise goal-directed movements after stroke can impair the ability to perform voluntary functions, learn new skills, and hinder rehabilitation. However, little is known about how the accuracy of single-joint, goal-directed ankle movements relates to multi-joint, lower limb function in stroke. Here, we determined the impact of stroke on the accuracy of goal-directed ankle movements and its relation to over-ground walking.
Methods: Stroke (N = 28) and control (N = 28) participants performed (1) goal-directed ankle dorsiflexion movements to accurately match 9 degrees in 180 ms and (2) over-ground walking. During goal-directed ankle movements, we measured the endpoint error, position error, time error and the activation of the agonist and antagonist muscles. During over-ground walking, we measured the walking speed, paretic stride length, and cadence.
Results: The stroke group demonstrated increased endpoint error than the controls. Increased endpoint error was associated with increased co-activation between agonist-antagonist muscles. Endpoint error was a significant predictor of walking speed and paretic stride length in stroke.
Conclusions: Impaired accuracy of goal-directed, ankle movements is correlated to over-ground walking in stroke.
Significance: Quantifying accuracy of goal-directed ankle movements may provide insights into walking function post-stroke.

1. Introduction

Goal-directed movements are necessary for voluntary motor control during activities of daily living. The inability to execute precise goal-directed movements can interfere with the ability to perform voluntary functions, learn new motor skills, and potentially hinder motor rehabilitation after a neurological injury such as stroke (Mastos et al., 2007; Chen et al., 2014). Reports suggest that accuracy of goal-directed upper-limb movements is reduced (increased error) following stroke (Winstein and Pohl, 1995; Leonard et al., 2006; Schaefer et al., 2012). However, evidence regarding the control of single-joint, goal-directed movements of the lower-limb is lacking. Further, functional motor tasks, such as reaching and walking, require accurate coordination of multiple joints. To date, it remains unknown whether the control of single-joint, goal-directed ankle movements is impaired and whether such impairment is associated with multi-joint, lower limb deficits post-stroke.

Over-ground walking involves goal-directed movements of the lower-limb for accurate and timely foot placements to proceed forward, change directions, cross obstacles, and terminate gait (Ble

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et al., 2005; Den Otter et al., 2005; Dennis et al., 2009). Impairments in executing goal-directed ankle movements could be associated with lower-limb function during walking and increased risk of loss in balance and falls in individuals with stroke. A previous study in individuals with ataxia demonstrated that the impaired accuracy of fast, goal-directed movements was associated with individual's capacity to maintain posture and walk (Casamento-Moran et al., 2015). Here we compare stroke and age-matched control groups on a goal-directed, ankle movement and an over-ground walking task.

Fast, goal-directed movements provide an excellent model for studying the execution of pre-planned ballistic movements (Christou et al., 2007; Chen et al., 2014; Casamento-Moran et al., 2017a). Because goal-directed movements are rapid, they do not allow for online corrections (Cordo et al., 1994; Elliott et al., 2010). Thus, performance on goal-directed movements relies on the execution of a preconceived motor plan that remains unmodified by online movement feedback. Fast and accurate goal-directed movements require appropriate activation and coordination of agonist and antagonist muscles (Gribble et al., 2003; Poston et al., 2008b). Dysfunctions in performing goal-directed movements can highlight the impact of aging or disease on underlying neuromuscular mechanisms. For example, older adults showed impaired performance during goal-directed contractions as demonstrated by decreased spatial (position error) and temporal accuracy (time error) (Christou et al., 2007; Chen et al., 2014). Decreased accuracy was associated with an altered activation of agonist muscle and an altered co-activation of the agonist-antagonist muscles (Christou et al., 2007; Chen et al., 2014). While neural mechanisms underlying coupled, multi-joint movements such as walking have been examined (Arene and Hidler, 2009), it remains unknown whether stroke impairs the neuromuscular activation underlying fast, single-joint, goal-directed ankle movements.

Therefore, the purpose of this study was to determine the impact of stroke on the accuracy of single-joint, goal-directed ankle movements and examine its association with over-ground walking function. We determine the neural mechanisms underlying the performance of goal-directed ankle movements by examining the activation and co-activation of agonist and antagonist muscles. Specifically, we examined the endpoint error of goal-directed ankle dorsiflexion movements and measured the amplitude, duration, and temporal overlap of agonist and antagonist muscles. To investigate whether single-joint, goal-directed movements are associated with over-ground walking in individuals with stroke, we conducted multiple linear regression to examine if the endpoint error predicts the walking speed, paretic stride length and cadence. We hypothesized that individuals with stroke will demonstrate impaired accuracy of single-joint, goal-directed ankle movements that will relate with the deficits in over-ground walking function.

2. Methods

2.1. Participants

Twenty-eight individuals with chronic stroke (63.38 ± 14.67 years) and 28 healthy control participants (68.08 ± 11.12 years) volunteered to participate in the study. Table 1 shows the clinical characteristics of the participants. Control participants were healthy adults without any neurological impairments. Inclusion criteria for stroke participants were: (1) diagnosed with a single unilateral cerebrovascular accident at least 9 months prior to testing; (2) ability to perform plantar and dorsiflexion of the paretic ankle on command; and (3) with a minimum of 10 degrees of active movement (combined dorsiflexion-plantarflexion); (4) ability to walk 3 minutes without assistance. Exclusion criteria included self-reported presence of any other neurological or musculoskeletal disorder, uncorrected visual and hearing impairments, visual neglect, and pain in lower extremity or elsewhere that could interfere with walking and ankle movements. The University of Florida's Institutional Review Board approved all experimental procedures. All individuals read and provided a written informed consent prior to participation.

Table 1
Demographics of the stroke and control groups.

Participant characteristics	Control (N = 28)	Stroke (N = 28)
Age (years)	68.08 ± 11.12	63.38 ± 14.67
Sex (Male/Female), N	18/10	11/17
Hemiparetic side (left/right), N	n/a	10/18
Time since stroke (years)	n/a	5.35 ± 4.30
Stroke location	n/a	14 cortical, 2 sub-cortical, 12 unavailable
Fugl-Meyer Assessment-lower extremity	n/a	26.39 ± 6.33
Ankle dorsiflexion range of motion (degree)	18.75 ± 6.76	15.21 ± 6.69
Ankle dorsiflexion MVC (newton)	139.75 ± 61.90	129.21 ± 64.85

The maximum score on Fugl-Meyer Assessment for lower extremity motor recovery is 34. The values denote Mean ± SD.

2.2. Experimental protocol

Each experimental testing session lasted ~90 min. At the beginning of each task, experimental procedures were explained to the participants. During the experimental session, we conducted: (1) clinical assessments that included Fugl-Meyer Assessment (FMA), range of motion (ROM), (2) maximum voluntary contraction (MVC); (3) goal-directed movements with ankle dorsiflexion; and (4) over-ground walking task. All tasks (except over-ground walking) were performed with the paretic limb in stroke and the non-dominant limb in control participants.

2.3. Experimental procedures

2.3.1. Clinical assessments

Fugl-Meyer Motor Assessment (FMA): We assessed the leg and foot motor impairments of the stroke participants using the lower extremity subsection of the FMA. The FMA ranged between 0 and 34 such that the lower scores indicate more severe motor impairments. *Range of Motion (ROM):* We measured the ankle range of motion using a goniometer to ensure that participants had adequate ROM to perform the goal-directed task.

2.3.2. Maximum voluntary contraction (MVC)

Apparatus and protocol: The MVC was measured for ankle dorsiflexor muscle (Tibialis Anterior; primary agonist muscle during the goal-directed movement). All participants sat comfortably with the hip joint aligned at ~90° flexion, the knee at ~90° flexion, and the ankle in a neutral position. Participants were instructed to quickly exert and maintain maximum dorsiflexion force at the ankle for 3 s. They were required to exert force at the ankle while maintaining a stable posture and avoiding any movement at the hip, knee, or trunk. Participants exerted 3–5 MVCs or until two MVCs were within 5% of each other. To minimize fatigue, we provided one-minute of rest between consecutive trials.

Data acquisition and analysis: The force produced during MVC task were measured with a force transducer (model 41BN, Honeywell, Morristown, NJ, USA) embedded in a custom foot device that

allowed measurement of isometric forces by applying a lock to prevent any ankle movement (Fig. 1A). The ankle force signals were sampled at 1000 Hz using an NI-DAQ card (model USB6210, National Instruments), high-pass filtered at 0.03 Hz, and amplified 50 times (Bridge-8, World Precision Instruments). The force data were stored on a personal computer for data analysis.

Strength: We quantified the ankle dorsiflexion strength with the maximum force produced during the MVC tasks. For each trial, peak force was determined. The strength was the highest force generated during the two MVC trials that were within 5% of each other. We recorded the peak EMG during MVC (average of 360 ms around the peak EMG) and used it to normalize the EMG activity during the goal-directed movements.

2.3.3. Goal-directed movements

Apparatus: The participants sat in an upright position facing a 32-inch monitor (Sync Master 320MP-2, Samsung Electronics America, Resolution: 1920 × 1080, Refresh Rate: 60p Hz). The visual feedback of the ankle movement was displayed on the monitor located 1.25 m away at eye level. Before beginning the task, all participants affirmed that they could see the display clearly. The hip joint was flexed to ~90° with 10° abduction, the knee was

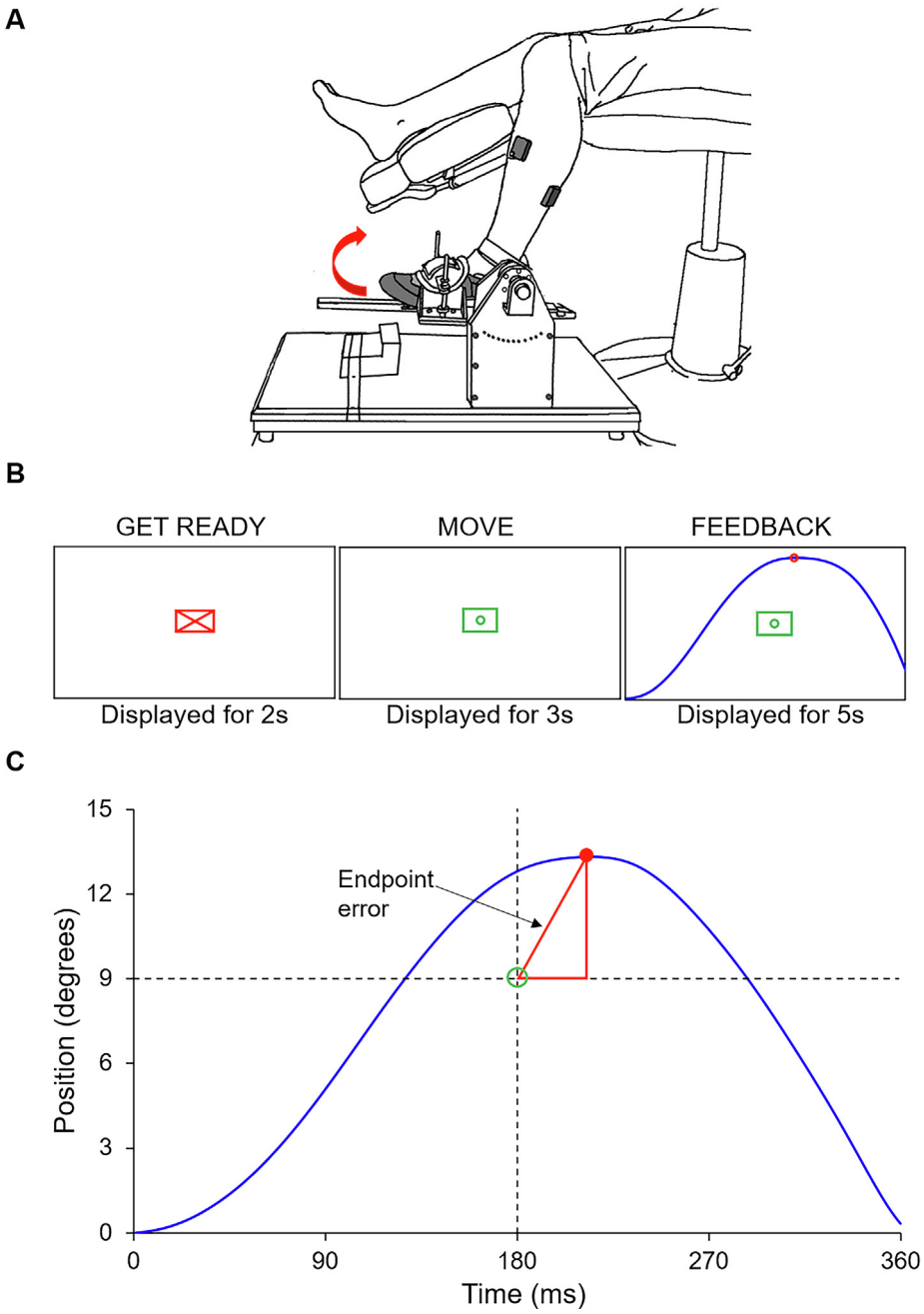

Fig. 1. (A) Experimental set-up for goal-directed ankle dorsiflexion movements: Participants were seated in a chair with hip and knees in ~90° flexion. The ankle was stabilized in the ankle device. The potentiometer connected in with line with ankle joint measured the ankle position during goal-directed movements. (B) Participants completed the goal-directed task in three phases – GET READY, MOVE and FEEDBACK. During the GET READY phase they observed the red target on the screen for 2 s. During the MOVE phase, they initiated the dorsiflexion movement as soon as the red target turned green. The green target appeared for 3 s. Finally, in the FEEDBACK phase participant's movement trajectory in relation to the target position and time was displayed on the screen for 5 s. (C). The endpoint error was calculated as the hypotenuse of the position and time values. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

flexed to ~90°, and the ankle was plantarflexed to 10°. The foot rested on a custom device that allowed only dorsiflexion and plantarflexion of the ankle. To ensure simultaneous movement between the device and the foot, we secured the foot by strapping the metatarsals (Fig. 1A). The protocol for fast goal-directed movements was adapted from previously published work (Christou et al., 2007; Casamento-Moran et al., 2015).

Protocol: Participants performed unloaded goal-directed movements to accurately match a target by performing a single, rapid ballistic ankle dorsiflexion. Following practice of 3–5 goal-directed movements at a target different from the actual target, each participant performed 10 goal-directed movement trials with 10 s of rest period between trials. The goal was to accurately match peak displacement of the ankle movement to the displayed target. The target consisted of a peak foot displacement of 9° in 180 ms. The movement had to be performed in 180 ms, not as fast as possible. No other instructions were given about how to perform the goal-directed movement to allow participants the freedom to plan their movement to best achieve the target.

The goal-directed task involved three phases: (1) GET READY; (2) MOVE; and (3) FEEDBACK. In the GET READY phase a red target was presented on the monitor for 2 s. The red target prompted the participants to get ready for the next phase. The MOVE phase began as the red target changed to green. The green target was presented on the monitor for 3 s. The green target prompted the participants to execute the goal-directed movement. Participants were instructed to initiate the movement at their convenience (not a reaction time task; but within 3 s). The task was recorded as soon as the participant initiated the movement. We did not provide online feedback of the movement to eliminate adjustments while performing the task. The FEEDBACK phase started at the end of MOVE phase and lasted for 5 s. The participants received visual feedback of their performance (movement trajectory) relative to the target position-time (9° and 180 ms) (Fig. 1B). This feedback allowed the participants to understand the task and plan the subsequent trials. The visual gain was kept constant at 1° for all trials (Vaillancourt et al., 2006). This protocol was adapted from previous studies of goal-directed movement control in healthy and diseased populations (Chen et al., 2014; Kwon et al., 2014).

Data acquisition and analysis: Data were analyzed offline using custom-written programs in Matlab® (Math Works™ Inc., Natick, Massachusetts, USA).

Limb displacement: The ankle position was measured using a low-friction potentiometer (SP22G-5K, Mouser Electronics, Mansfield, TX, USA) placed in line with the fibular malleolus. The ankle position data were sampled at 1000 Hz using an NI-DAQ card (model USB6210, National Instruments) and stored on a personal computer for data analysis.

Position, time and endpoint error: To calculate the endpoint error, we first quantified the position and time errors (Fig. 1C). Position error was computed as the absolute vertical deviation from the targeted position to the peak displacement. Time error was quantified as the absolute horizontal deviation from the targeted time to peak displacement. We normalized the position and time errors to have similar units (%). The position error was normalized to the targeted peak (9°) displacement (Eq. (1)), and the time error was normalized to the targeted time to peak (180 ms) displacement (Eq. (2)). The endpoint error was quantified as the hypotenuse between the position error and time error distance (Fig. 1C, Eq. (3)). To measure the motor output variability, we calculated coefficient of variation (CV) of the endpoint error (Eq. (4)).

$$Position\ error(\%) = \frac{peak\ displacement\ (degrees)}{targeted\ peak\ displacement\ (degrees)} \times 100 \quad (1)$$

$$Time\ error(\%) = \frac{time\ to\ peak\ displacement\ (ms)}{targeted\ time\ to\ peak\ displacement\ (ms)} \times 100 \quad (2)$$

$$Endpoint\ error(\%) = \sqrt{(time\ error)^2 + (position\ error)^2} \quad (3)$$

$$CV\ of\ endpoint\ error\ (\%) = \frac{Standard\ deviation\ of\ endpoint\ error}{Mean\ endpoint\ error} \times 100 \quad (4)$$

Neuromuscular Activation: The muscle activity of tibialis anterior (TA, agonist) and medial gastrocnemius (MG, antagonist) were measured using the Trigno wireless EMG system (Delsys Inc., Boston, MA, USA). The electrodes (Trigno™ wireless, Delsys Inc., Boston, MA) were placed parallel to the muscle fibers and secured to the skin. The location for the electrode was determined according to the European Recommendations for Surface Electromyography (Hermens et al., 2000). The EMG signals were band pass filtered at 20–450 Hz, amplified with a gain of 1000 (EMGworks® 4.0.5, Trigno™ wireless, Delsys Inc., Boston, MA), sampled at 1000 Hz with a NI-DAQ card (Model USB6210, National Instruments, Austin Tx, USA) and stored on a personal computer.

The interference EMG for each trial was rectified and smoothed with a fourth-order Butterworth filter with a cutoff frequency of 6 Hz (Poston et al., 2008a). This filtered data was used to identify the amplitude, onset, and offset of the EMG burst of the primary agonist (TA) and the primary antagonist (MG) ankle muscles.

Muscle activity: We analyzed the EMG bursts to quantify the following variables:

- *EMG burst duration* was measured as the time (ms) elapsed between muscle activity onset (>15% of the peak EMG) and offset (<15% of the peak EMG).
- *EMG burst amplitude (%)* was measured as the peak EMG activity within the burst duration normalized to the peak EMG activity during the MVC.
- *Agonist-antagonist muscle temporal overlap (%)* was defined as the period of simultaneous activation of TA and MG muscles relative to the total duration of combined activation of TA and MG (Eq. (5)).

$$TA-MG\ temporal\ overlap\ (\%) = absolute\left[\frac{TA\ offset - MG\ onset}{MG\ offset - TA\ onset} * 100\right] \quad (5)$$

2.3.4. Over-ground walking

Apparatus and protocol: Each participant wore six inertial movement sensors on their body; two sensors were placed on the wrist, two sensors on the ankle, one on chest, one on lumbar spine (Mobility Lab system, APDM, Inc, Oregon, USA). Participants were instructed to walk a short distance (7 m) over-ground at their comfortable natural pace. Three trials were performed with one minute rest period between trials to minimize fatigue.

Data acquisition and analysis: The gait parameters were measured using wireless inertial movement monitors (Mobility Lab, APDM, Inc, Oregon, USA). Each trial was validated, analyzed and, saved to a personal computer. A trial was considered validated when the inertial sensors were detected throughout the trial and primary outcomes were successfully calculated. We computed the average of three trials to quantify mean gait speed (m/s), stride length (m) of the paretic leg for stroke and non-dominant leg for control participants, and cadence (steps/min).

2.4. Statistical analysis

We assessed the normality of endpoint error for goal-directed movement and EMG (EMG burst amplitude, duration and TA-MG temporal overlap) using the Shapiro-Wilk test. The endpoint error, position error, time error, and EMG variables did not meet the test of normality. Therefore, we compared the stroke and control groups using the Man Whitney-U test. We conducted a secondary analyses to examine whether the endpoint error of goal-directed movements in the stroke group was related to the strength or motor output variability. Here, we performed Pearson's bivariate correlation analysis between endpoint error, MVC of dorsiflexion and CV of endpoint error.

To examine whether single-joint goal-directed movements predicted over-ground walking in stroke group, we conducted a linear regression. We log transformed the endpoint error, position error and time error for regression analysis. We ran three separate linear regressions with walking speed, paretic stride length and cadence as the criterion variables and endpoint error as the predictor variable. In addition, to examine the contribution of position and time error to over-ground walking in stroke, we conducted separate backward, multiple linear regression analyses with walking speed, paretic stride length, and cadence as criterion variables and position error and time error as predictor variables. All statistical tests were conducted with an alpha level set at 0.05. All statistical analyses were performed using SPSS 25.0 (IBM, Armonk, NY, USA).

3. Results

3.1. Clinical variables

The mean FMA lower extremity score in stroke group was 26.39 ± 6.33 (Table 1). The dorsiflexion range of motion was not statistically different ($|t_{54}| = -1.96$, $p > 0.05$) between the stroke and the control group. The ankle dorsiflexion strength was not significantly different between the two groups ($|t_{54}| = -0.62$, $p > 0.05$).

3.2. Goal-directed movements

The accuracy of goal-directed movements was quantified with endpoint error. The goal-directed movements across ten consecutive trials of representative stroke and control participants are shown in Fig. 2. The stroke group exhibited significantly increased endpoint error compared with the control group (Z = – 2.17, $p = 0.02$; Fig. 3). Compared with the controls, the stroke group showed a trend towards increased position error and time error; however, the difference did not reach statistical significance ($p > 0.05$; Table 2).

There was no relationship between endpoint error of goal-directed movements and ankle dorsiflexion MVC ($r = -0.23$, $p = 0.23$) and CV of endpoint error ($r = -0.13$, $p = 0.50$) in the stroke group. Thus, the accuracy of goal-directed ankle movements was not related to dorsiflexion strength or motor output variability. To ensure that motor learning was not a confounding factor, a paired sample t-test was conducted on endpoint error. The results confirmed that there was no difference in endpoint error between the first and tenth trials for each group (Stroke: $t_{27} = -0.83$, $p > 0.05$; Control: $t_{27} = 0.84$, $p > 0.05$).

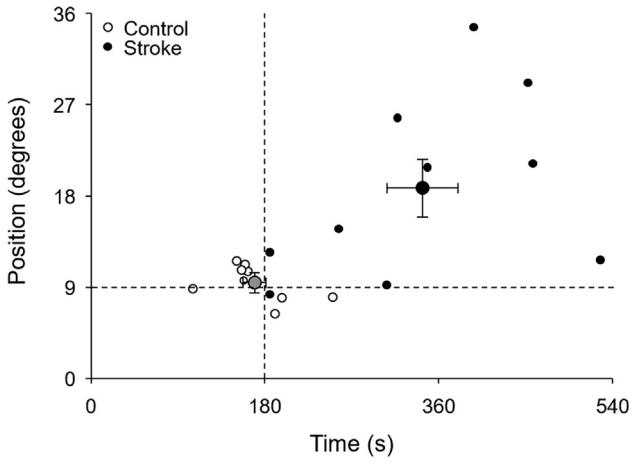

Fig. 2. For a representative stroke (small black circles) and control (small open circles) participant, the performance for ten consecutive trials. The broken lines represent the target. The average endpoint values are shown in large black circle (stroke) and grey circle (control) with error bars.

3.3. Neuromuscular activation

The TA muscle activation data for 2 individuals (1 stroke and 1 control) and MG data for 5 individuals (1 stroke and 4 controls) were not available due to corrupt files and were removed from the EMG analysis. The stroke group demonstrated increased amplitude of TA activation as compared with the control group (Z = – 2.39, $p = 0.01$; Fig. 4A). The two groups did not differ on the amplitude of MG activation (Z = –1.85, $p = 0.06$; Fig. 4C). The stroke group demonstrated prolonged duration of TA activation than the control group (Z = –5.42, $p < 0.001$; Fig. 4B). However, the two groups did not differ on the duration of MG activation (Z = –0.72, $p = 0.47$; Fig. 4D). We quantified the co-activation between agonist and antagonist muscle with temporal overlap between TA and MG activation. The stroke group showed increased co-activation between TA and MG muscles as compared with the control group (Z = –2.88, $p < 0.01$; Fig. 4E and F). We found a significant positive

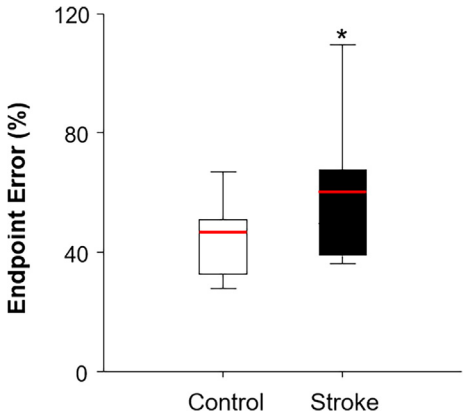

Fig. 3. The group data on endpoint error in goal-directed movements. The red line represents group means. The stroke group had significantly increased endpoint error as compared with the control group ($^{*}p < 0.05$). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 2
Endpoint error, position error, and time error for the two groups.

	Stroke	Control	*p*
Endpoint error	60.41 ± 5.79	46.77 ± 4.18	0.02*
Position error	40.08 ± 3.99	30.86 ± 3.67	0.08
Time error	37.17 ± 5.31	27.70 ± 3.67	0.08

Scores indicate mean ± SE.
* Indicate p < 0.05.

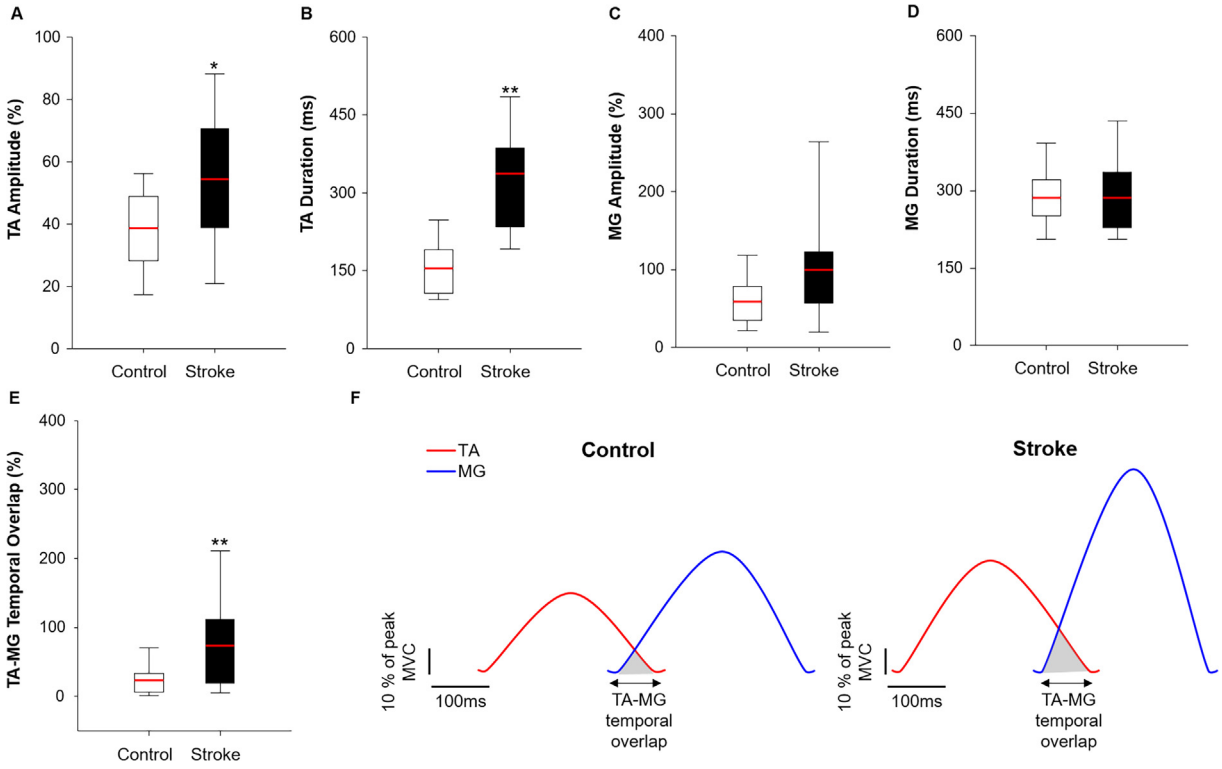

Fig. 4. This figure shows the differences in agonist (tibialis anterior, TA) and antagonist (medial gastrocnemius, MG) EMG variables between the groups. (A) TA amplitude, (B) TA duration, (C) MG amplitude, (D) MG duration and, (E) the temporal overlap between TA and MG muscles. There red line represents group means. The stroke group had significant increased TA amplitude and duration, as well as increased TA-MG temporal overlap ($^{*}p < 0.05$; $^{**}p < 0.01$). (F) Temporal overlap of TA and MG muscle activity in stroke and control groups. The increased TA-MG temporal overlap in stroke was contributed by both increased duration and amplitude of TA activation.

relationship between TA-MG co-activation and the endpoint error in the stroke group ($r = 0.42$, $p = 0.02$).

3.4. Relation between goal-directed movements and over-ground walking

We found a significant negative relationship between over-ground walking speed and the endpoint error ($r = -0.52$, $p < 0.01$) in the stroke group. Thus, decreased accuracy of the goal-directed ankle movements was associated with reduced walking speed. Additionally, we found a significant negative relationship between paretic stride length and the endpoint error ($r = -0.60$, $p < 0.01$) in the stroke group. The decreased accuracy on the goal-directed ankle movements was associated with shorter stride length of the paretic limb. We found no significant relationship between the endpoint error of goal-directed movements and cadence ($r = -0.33$, $p > 0.05$).

The linear regression model revealed that endpoint error was a significant predictor of the over-ground walking speed ($R^2 = 0.27$, adjusted $R^2 = 0.25$; $p < 0.01$, Fig. 5A), and stride length of the paretic limb ($R^2 = 0.36$, adjusted $R^2 = 0.35$; $p < 0.01$, Fig. 5B) in the stroke group. The endpoint error did not predict cadence ($R^2 = 0.10$, adjusted $R^2 = 0.07$; $p > 0.05$) in the stroke group. In addition, we performed backward multiple regression to examine whether

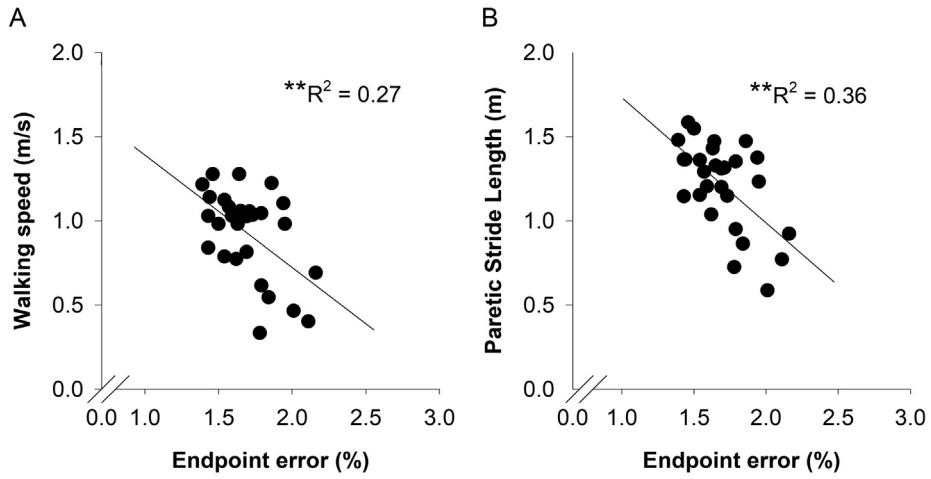

Fig. 5. Prediction of (A) over-ground walking speed and (B) stride length of the paretic limb from endpoint error of the goal-directed ankle movements ($^{*}p < 0.05$; $^{**}p < 0.01$).

Table 3
Summary of multiple regression analysis to predict over-ground walking from position and time errors.

	Gait speed		Paretic stride length		Cadence	
	SB	*p*	*SB*	*p*	*SB*	*p*
Position error	−0.41	0.01	−0.43	0.00	−0.25	0.17
Time error	−0.41	0.01	−0.45	0.00	−0.32	0.08
	R^2	*p*	R^2	*p*	R^2	*p*
Model combining position and time errors	0.34	0.00	0.40	0.00	0.17	0.08

SB: Standardized beta. We performed backward multiple regression analysis to predict gait speed, paretic stride length, and cadence from position error and time error. We found that both position and time error predicted gait speed and paretic stride length.

position and time error predicted over-ground walking performance. These results are presented in Table 3. We found that both position and time error are significant predictors for gait speed and paretic stride length in the stroke group. Overall, reduced accuracy of goal-directed, ankle movements was associated with decline in over-ground walking function in individuals with stroke.

4. Discussion

In this study, we determined the impact of stroke on the accuracy of single-joint, goal-directed ankle movements and examined its relation to over-ground walking function. Our results demonstrate that individuals with stroke exhibit decreased endpoint accuracy and altered muscle activation compared with the healthy controls. The decreased endpoint accuracy following stroke was related to increased co-activation of agonist-antagonist muscles. Interestingly, the accuracy of goal-directed ankle movements in stroke predicted walking speed and stride length of the paretic limb during over-ground walking. These novel findings, therefore, suggest that individuals with stroke show altered co-activation of agonist-antagonistic muscles and decreased endpoint accuracy of single-joint ankle movements that correlates to over-ground walking function.

4.1. Accuracy of fast goal-directed movements in stroke

Individuals with stroke demonstrate increased endpoint error during fast, goal-directed ankle dorsiflexion movements. Our goal-directed task required a fast voluntary dorsiflexion to achieve a targeted position of 9° within a targeted time of 180 ms. We did not provide any real-time feedback of the movement to encourage pre-planning of each movement trial and to prevent online error corrections. Therefore, decreased endpoint accuracy suggests impaired ability to execute fast and precise goal-directed movements in individuals with stroke. Further, the stroke group showed a trend towards increased position error and time error indicating that the decline in endpoint accuracy had partial contribution from both spatial and temporal inaccuracy. These findings are consistent with the upper extremity studies demonstrating that stroke impairs the overall accuracy to reach a pre-determined target during rapid voluntary movements (Cirstea and Levin, 2000; Velicki et al., 2000; McCombe Waller et al., 2016). For example, Velicki et al., (2000) reported reduced directional accuracy during goal-directed elbow flexion with the non-paretic arm (Velicki et al., 2000). Further, individuals with stroke exhibited impaired goal-directed reaching with paretic arm as shown by the longer time to initiate a movement in the absence of a visual cue (Dean et al., 2012).

The current study extends previous work by providing evidence that impaired endpoint accuracy is related to altered neural activation of the agonist and antagonist muscles during goal-directed ankle movements. Our results show that stroke group demonstrated increased amplitude and duration of activation of tibialis anterior (i.e. agonist muscle). In addition, stroke participants demonstrated increased temporal overlap of agonist-antagonist muscles that was related to increased endpoint error. The amplitude and timing of activation of agonist-antagonist muscle are essential to perform accurate goal-directed movements (Christou et al., 2007; Casamento-Moran et al., 2017b). Previous studies reported increased amplitude, altered timing and co-activation of agonist-antagonist muscles during walking and voluntary balance tasks among stroke survivors (Cheng et al., 2004; Arene and Hidler, 2009; Garland et al., 2009; Peters et al., 2016). Specifically, Peters et al. (2016) demonstrated that stroke survivors with larger amplitude of muscle activation and higher co-activation of rectus femoris and biceps femoris prior to heel strike had poor balance on the community balance and mobility scale (Peters et al., 2016). Further, individuals with stroke demonstrate reduced co-activation of ankle muscles during stance phase which may contribute to reduced postural stability while walking (Lamontagne et al., 2000). Our results are in line with the previous work and highlight that stroke affects the ability to select and time appropriate muscle activation patterns to execute an accurate goal-directed movement.

4.2. Goal-directed movements and over-ground walking

The most important finding of the current study is the relation between simple, isolated, goal-directed movement of a single-joint and multi-joint lower-limb function. Our results provide novel evidence that impaired accuracy of goal-directed ankle movements is associated with reduced walking ability following stroke. In our study, we found that endpoint error is a significant predictor of over-ground walking speed in individuals with stroke. Walking speed is an important determinant of community ambulation in stroke survivors (van de Port et al., 2008; Fulk et al., 2017). Consider the situation when a pedestrian approaches a crosswalk to cross a road and looks at the crossing signal. If the light indicates STOP signal, the individual prepares to slow down or if the light indicates WALK signal, then the individual walks faster to cross the road timely. These motor actions that seem implicit involve integration of relevant environmental information and execution of precise spatio-temporal limb dynamics to prevent any errors in modulating the speed and position of lower-limbs. While walking speed is frequently used as an indicator of functional status and rehabilitation outcomes, stride length and cadence are also considered important measures of walking function (Allen et al., 2011; Pohl et al., 2011; Nascimento et al., 2015). Our results suggest that decreased endpoint accuracy is a significant predictor of walking speed and paretic stride length in individuals with stroke. Thus, impaired accuracy of fast, goal-directed movement following stroke may reflect deficits in planning and execution of movements to terminate, maintain or modulate speed and stride length during walking.

Until recently, much of stroke research focused on identifying the role of motor impairments such as reduced strength and increased variability in walking deficits (Olney et al., 1994; Kerrigan et al., 2001; Balasubramanian et al., 2009; Kizony et al.,

2010). Clearly, strength and variability are important to consistently generate sufficient force for forward propulsion while walking. However, our results suggest that the accuracy of single-joint, goal-directed movements was not associated with the ankle dorsiflexion strength or motor output variability. Thus, individuals with stroke demonstrated impaired accuracy that was unrelated to ankle weakness or variability. A myriad of interventions have been tested to rehabilitate walking function in stroke survivors (Morone et al., 2012; Polese et al., 2013; Dragin et al., 2014). Whether training programs focused on the identifying and addressing deficits in executing goal-directed movements could facilitate recovery of walking function remains to be determined (Hollands et al., 2015; Timmermans et al., 2016). A previous study demonstrated that motor training to improve the accuracy of ankle plantarflexion-dorsiflexion through visuomotor tracking task increased toe clearance and reduced temporal asymmetry while walking in chronic stroke survivors (Deng et al., 2012). Retraining goal-directed movements may facilitate movement planning to develop the association between movement outcomes and selected movement patterns, and consequently improve walking function after stroke.

4.3. Considerations

The location of brain lesion can influence the degree and severity of impairments in goal-directed movements following stroke. While steady state walking is considered as a rhythmic motor behavior controlled by central pattern generators and supraspinal locomotor regions (Guadagnoli et al., 2000; Rossignol et al., 2006), the role of higher cortical centers including prefrontal cortex (PFC), premotor cortex (PMC), and supplementary motor area are implicated in planning and execution of movement direction, speed and limb positioning is also well-known (Deiber et al., 1996; la Fougere et al., 2010; Zwergal et al., 2012). For example, PMC and PFC show heightened activation while participants increased the walking speed, suggesting their role in planning for acceleration of walking speed (Suzuki et al., 2004). In our study, fourteen individuals had a cortical stroke and two individuals had a sub-cortical stroke. None of the participants had a cerebellar stroke. The precise information on the location of lesion was unknown to determine the impact of lesion location on the accuracy of goal-directed movements following stroke.

The impairments in goal-directed movements and over-ground walking may be contributed by several factors including sensory and cognitive deficits, spasticity and hypertonicity (Ng and Hui-Chan, 2005; Cirstea et al., 2006; Wutzke et al., 2013). However, we did not assess these factors in the current study. Further, our results showing correlation between accuracy of goal-directed movements and over-ground walking in stroke must be interpreted as association not causation between these two factors. The current approach to studying goal-directed movements in stroke individuals has not been examined before and warrants future investigation. While our study focuses on goal-directed ankle dorsiflexion that is critical for accurate foot positioning and clearance during initial contact and swing, future studies should investigate the relationship between goal-directed plantar-and dorsiflexion to gain a more comprehensive understanding of how the control of single joint, goal-directed ankle movement influences walking function in stroke.

5. Conclusions

The current study provides novel evidence that the endpoint accuracy of single-joint, goal-directed ankle movements is impaired following stroke. The impaired endpoint accuracy is associated with increased co-activation of the agonist-antagonist muscles. Most importantly, endpoint accuracy of single-joint, goal-directed ankle movements correlates with the multi-joint over-ground walking function in stroke. The improvements in accuracy of goal-directed, single joint ankle movements may promote appropriate timing and activation of agonist-antagonist muscles to enhance motor function following stroke.

Conflict of interest

None of the authors have potential conflicts of interest to be disclosed.

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References

Allen JL, Kautz SA, Neptune RR. Step length asymmetry is representative of compensatory mechanisms used in post-stroke hemiparetic walking. Gait Posture 2011;33:538–43.
Arene N, Hidler J. Understanding motor impairment in the paretic lower limb after a stroke: a review of the literature. Top Stroke Rehabil 2009;16:346–56.
Balasubramanian CK, Neptune RR, Kautz SA. Variability in spatiotemporal step characteristics and its relationship to walking performance post-stroke. Gait Posture 2009;29:408–14.
Ble A, Volpato S, Zuliani G, Guralnik JM, Bandinelli S, Lauretani F, et al. Executive function correlates with walking speed in older persons: the InCHIANTI study. J Am Geriatr Soc 2005;53:410–5.
Casamento-Moran A, Chen YT, Kwon M, Snyder A, Subramony SH, Vaillancourt DE, et al. Force dysmetria in spinocerebellar ataxia 6 correlates with functional capacity. Front Hum Neurosci 2015;9:184.
Casamento-Moran A, Chen YT, Lodha N, Yacoubi B, Christou EA. Motor plan differs for young and older adults during similar movements. J Neurophysiol 2017a;117:1483–8.
Casamento-Moran A, Hunter SK, Chen YT, Kwon MH, Fox EJ, Yacoubi B, et al. Sex differences in spatial accuracy relate to the neural activation of antagonistic muscles in young adults. Exp Brain Res 2017b;235:2425–36.
Chen YT, Kwon M, Fox EJ, Christou EA. Altered activation of the antagonist muscle during practice compromises motor learning in older adults. J Neurophysiol 2014;112:1010–9.
Cheng PT, Chen CL, Wang CM, Hong WH. Leg muscle activation patterns of sit-to-stand movement in stroke patients. Am J Phys Med Rehabil 2004;83:10–6.
Christou EA, Poston B, Enoka JA, Enoka RM. Different neural adjustments improve endpoint accuracy with practice in young and old adults. J Neurophysiol 2007;97:3340–50.
Cirstea CM, Ptito A, Levin MF. Feedback and cognition in arm motor skill reacquisition after stroke. Stroke 2006;37:1237–42.
Cirstea MC, Levin MF. Compensatory strategies for reaching in stroke. Brain 2000;123:940–53.
Cordo P, Carlton L, Bevan L, Carlton M, Kerr GK. Proprioceptive coordination of movement sequences: role of velocity and position information. J Neurophysiol 1994;71:1848–61.
Dean PJ, Seiss E, Sterr A. Motor planning in chronic upper-limb hemiparesis: evidence from movement-related potentials. PLoS One 2012;7 e44558.
Deiber MP, Ibanez V, Sadato N, Hallett M. Cerebral structures participating in motor preparation in humans: a positron emission tomography study. J Neurophysiol 1996;75:233–47.
Den Otter AR, Geurts AC, de Haart M, Mulder T, Duysens J. Step characteristics during obstacle avoidance in hemiplegic stroke. Exp Brain Res 2005;161:180–92.
Deng H, Durfee WK, Nuckley DJ, Rheude BS, Severson AE, Skluzacek KM, et al. Complex versus simple ankle movement training in stroke using telerehabilitation: a randomized controlled trial. Phys Ther 2012;92:197–209.
Dennis A, Dawes H, Elsworth C, Collett J, Howells K, Wade DT, et al. Fast walking under cognitive-motor interference conditions in chronic stroke. Brain Res 2009;1287:104–10.
Dragin AS, Konstantinovic LM, Veg A, Schwirtlich LB. Gait training of poststroke patients assisted by the Walkaround (body postural support). Int J Rehabil Res 2014;37:22–8.
Elliott D, Hansen S, Grierson LE, Lyons J, Bennett SJ, Hayes SJ. Goal-directed aiming: two components but multiple processes. Psychol Bull 2010;136:1023–44.
Fulk GD, He Y, Boyne P, Dunning K. Predicting home and community walking activity poststroke. Stroke 2017;48:406–11.

Garland SJ, Gray VL, Knorr S. Muscle activation patterns and postural control following stroke. Motor Control 2009;13:387–411.
Gribble PL, Mullin LI, Cothros N, Mattar A. Role of cocontraction in arm movement accuracy. J Neurophysiol 2003;89:2396–405.
Guadagnoli MA, Etnyre B, Rodrigue ML. A test of a dual central pattern generator hypothesis for subcortical control of locomotion. J Electromyogr Kinesiol 2000;10:241–7.
Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol 2000;10:361–74.
Hollands KL, Pelton TA, Wimperis A, Whitham D, Tan W, Jowett S, et al. Feasibility and preliminary efficacy of visual cue training to improve adaptability of walking after stroke: multi-centre, single-blind randomised control pilot trial. PLoS One 2015;10:e0139261.
Kerrigan DC, Karvosky ME, Riley PO. Spastic paretic stiff-legged gait: joint kinetics. Am J Phys Med Rehabil 2001;80:244–9.
Kizony R, Levin MF, Hughey L, Perez C, Fung J. Cognitive load and dual-task performance during locomotion poststroke: a feasibility study using a functional virtual environment. Phys Ther 2010;90:252–60.
Kwon M, Chen YT, Fox EJ, Christou EA. Aging and limb alter the neuromuscular control of goal-directed movements. Exp Brain Res 2014;232:1759–71.
la Fougere C, Zwergal A, Rominger A, Forster S, Fesl G, Dieterich M, et al. Real versus imagined locomotion: a [18F]-FDG PET-fMRI comparison. Neuroimage 2010;50:1589–98.
Lamontagne A, Richards CL, Malouin F. Coactivation during gait as an adaptive behavior after stroke. J Electromyogr Kinesiol 2000;10:407–15.
Leonard CT, Gardipee KA, Koontz JR, Anderson JH, Wilkins SA. Correlation between impairment and motor performance during reaching tasks in subjects with spastic hemiparesis. J Rehabil Med 2006;38:243–9.
Mastos M, Miller K, Eliasson AC, Imms C. Goal-directed training: linking theories of treatment to clinical practice for improved functional activities in daily life. Clin Rehabil 2007;21:47–55.
McCombe Waller S, Yang CL, Magder L, Yungher D, Gray V, Rogers MW. Impaired motor preparation and execution during standing reach in people with chronic stroke. Neurosci Lett 2016;630:38–44.
Morone G, Fusco A, Di Capua P, Coiro P, Pratesi L. Walking training with foot drop stimulator controlled by a tilt sensor to improve walking outcomes: a randomized controlled pilot study in patients with stroke in subacute phase. Stroke Res Treat 2012;2012:523564.
Nascimento LR, de Oliveira CQ, Ada L, Michaelsen SM, Teixeira-Salmela LF. Walking training with cueing of cadence improves walking speed and stride length after stroke more than walking training alone: a systematic review. J Physiother 2015;61:10–5.
Ng SS, Hui-Chan CW. The timed up & go test: its reliability and association with lower-limb impairments and locomotor capacities in people with chronic stroke. Arch Phys Med Rehabil 2005;86:1641–7.
Olney SJ, Griffin MP, McBride ID. Temporal, kinematic, and kinetic variables related to gait speed in subjects with hemiplegia: a regression approach. Phys Ther 1994;74:872–85.
Peters S, Garland SJ, Miller KJ, Cochrane CK, Ivanova TD, Hunt MA. Motor planning for loading during gait in subacute stroke. Arch Phys Med Rehabil 2016;97:528–35.
Pohl PS, Kemper S, Siengsukon CF, Boyd L, Vidoni E, Herman RE. Older adults with and without stroke reduce cadence to meet the demands of talking. J Geriatr Phys Ther 2011;34:35–40.
Polese JC, Ada L, Dean CM, Nascimento LR, Teixeira-Salmela LF. Treadmill training is effective for ambulatory adults with stroke: a systematic review. J Physiother 2013;59:73–80.
Poston B, Enoka JA, Enoka RM. Endpoint accuracy for a small and a large hand muscle in young and old adults during rapid, goal-directed isometric contractions. Exp Brain Res 2008a;187:373–85.
Poston B, Enoka JA, Enoka RM. Practice and endpoint accuracy with the left and right hands of old adults: the right-hemisphere aging model. Muscle Nerve 2008b;37:376–86.
Rossignol S, Dubuc R, Gossard JP. Dynamic sensorimotor interactions in locomotion. Physiol Rev 2006;86:89–154.
Schaefer SY, Mutha PK, Haaland KY, Sainburg RL. Hemispheric specialization for movement control produces dissociable differences in online corrections after stroke. Cereb Cortex 2012;22:1407–19.
Suzuki M, Miyai I, Ono T, Oda I, Konishi I, Kochiyama T, et al. Prefrontal and premotor cortices are involved in adapting walking and running speed on the treadmill: an optical imaging study. Neuroimage 2004;23:1020–6.
Timmermans C, Roerdink M, van Ooijen MW, Meskers CG, Janssen TW, Beek PJ. Walking adaptability therapy after stroke: study protocol for a randomized controlled trial. Trials 2016;17:425.
Vaillancourt DE, Haibach PS, Newell KM. Visual angle is the critical variable mediating gain-related effects in manual control. Exp Brain Res 2006;173:742–50.
van de Port IG, Kwakkel G, Lindeman E. Community ambulation in patients with chronic stroke: how is it related to gait speed? J Rehabil Med 2008;40:23–7.
Velicki MR, Winstein CJ, Pohl PS. Impaired direction and extent specification of aimed arm movements in humans with stroke-related brain damage. Exp Brain Res 2000;130:362–74.
Winstein CJ, Pohl PS. Effects of unilateral brain damage on the control of goal-directed hand movements. Exp Brain Res 1995;105:163–74.
Wutzke CJ, Mercer VS, Lewek MD. Influence of lower extremity sensory function on locomotor adaptation following stroke: a review. Top Stroke Rehabil 2013;20:233–40.
Zwergal A, Linn J, Xiong G, Brandt T, Strupp M, Jahn K. Aging of human supraspinal locomotor and postural control in fMRI. Neurobiol Aging 2012;33:1073–84.