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Intralimb gait coordination of individuals with stroke using vector coding

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ABSTRACT

Individuals with stroke often present functional impairment and gait alteration. Among different aspects, intralimb coordination of these individuals is one of the key points that should be considered before implementing any gait intervention protocol. The purpose of this study was to investigate the effects of stroke on intralimb gait coordination of the lower limbs using a vector coding technique. Twenty-five individuals with stroke and 18 non-disabled individuals (control), between 46 and 71 years old, participated in this study. A computerized analysis system registered data from reflective markers placed on specific body landmarks to define thigh, shank, and foot of both body sides, as participants walked at self-selected comfortable speed. Coordination modes, such as in-phase, anti-phase, proximal-segment-phase, and distal-segment-phase, and variability of thigh-shank, and shank-foot were analyzed for the paretic, non-paretic and control limbs during the stance and swing periods, and the entire gait cycle using the vector coding technique. During the stance period, individuals with stroke presented higher frequency of thigh-phase and lower frequency of shank-phase for the thigh-shank coupling and higher frequency of shank-phase for the shank-foot coupling compared to non-disabled controls, indicating that the proximal segment of each pair leads the movement. During the swing period, the paretic limb presented higher frequency for in-phase than non-paretic and control limbs for the thigh-shank coupling. Adaptations in the non-paretic limb were observed in the swing period, with higher frequency than paretic and control limbs in the thigh-phase for the thigh-shank coupling, and higher frequency than the paretic limb in the foot-phase for the shank-foot coupling. No differences in coordination variability were found between paretic, non-paretic, and control limbs. The vector coding technique constitutes a useful tool for identifying gait alterations in intralimb coordination of individuals with stroke. Our coordination results demonstrate a shift from distal to more proximal control during the stance phase in both legs for the individuals with stroke and an inability to decouple segment coordination during the swing phase in the paretic limb. The results indicate that it is more suitable to consider the stance and swing periods separately instead of considering the entire gait cycle to investigate intralimb gait coordination of individuals with stroke.

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1. Introduction

Worldwide stroke is the second leading cause of death. The disability caused by a stroke episode affects individuals at the peak of their productive lives (WHO, 2018). Among the different motor disabilities, the disruption of locomotor function is one of the major issues for many individuals with stroke (Jorgensen et al., 1995a,b), which is associated with a high risk for falls in community-dwelling post-stroke individuals (Hyndman, Ashburn, & Stack, 2002). The effects of a stroke depend on which brain area is damaged and how severely it is affected (WHO, 2016, 2018), and paresis (i.e., the inability to voluntarily recruit skeletal motor units to produce movement) is one of the possible effects of the central lesions (Gracies, 2005). Weakness and spasticity are among the several characteristics related to paresis, leading to motor control disturbances (Dierick et al., 2017; Niam, Cheung, Sullivan, Kent, & Gu, 1999; Olney & Richards, 1996; Sanchez, Acosta, Lopez-Rosado, Stienen, & Dewald, 2017) and, therefore, impacting walking capabilities.

Individuals with stroke often develop compensatory motor adaptations based upon the spared functionality (Levin, Kleim, & Wolf, 2009). Compensations can occur in most of the lower limb joints during walking, such as reduced excursion of hip and knee joints of the paretic limb (Chen, Patten, Kothari, & Zajac, 2005), reduced propulsive force (Sharma, McMorland, & Stinear, 2015), and increased levels of activation and co-contraction of agonist and antagonist muscles (Ma et al., 2017). All these changes also impact the temporal organization, such as the stance and swing periods, leading to asymmetry between paretic and non-paretic sides (Chen et al., 2005).

Although the employment of different biomechanical techniques, such as discrete kinematic and EMG analysis (Chen et al., 2005; Ma et al., 2017), has provided clinicians with important insights regarding the implementation of rehabilitation protocols to improve gait in individuals with stroke, improvements in gait coordination or the relative motion between segments will be most closely aligned with regaining locomotor patterns and stability. Overall, earlier investigations concerning intralimb gait coordination of individuals with stroke have found that these individuals present asymmetry between the paretic and non-paretic limbs (Rinaldi & Monaco, 2013), and their shank and thigh are less dissociated (i.e., moving more in-phase) at the end of the gait cycle (i.e., swing period) (Barela et al., 2000).

It is important to note that those previous studies investigating gait coordination in stroke had employed the continuous relative phase (CRP) (Barela et al., 2000; Barela et al., 2002; Combs, Dugan, Ozimek, & Curtis, 2013; Rinaldi & Monaco, 2013), which allows the assessment of coordination through the gait cycle. Despite the important meaning that CRP provides to reflect the coordination dynamics based on displacement and velocity of segment or joint in the phase plane (Barela et al., 2000), CRP analysis is more appropriate on signals that are close to sinusoidal (Van Emmerik, Miller, & Hamill, 2014), which may not be the case in the gait of individuals with gait impairment. Besides that, CRP does not allow a purely spatial phasing interpretation (Van Emmerik et al., 2014).

More recently, Krasovsky and Levin (2010) presented advantages and disadvantages of different techniques used to measure locomotor coordination of individuals with stroke, and concluded that there is a lack of an appropriate technique that provides a gold standard for a complete assessment of gait coordination changes in individuals with stroke. In order to advance our understanding on motor coordination during gait in individuals after stroke, we adopted in the present study the vector coding technique, which has been employed to examine intralimb coordination based upon a segmental coupling angle (Hamill, Palmer, & Van Emmerik, 2012; Sparrow, Donovan, van Emmerik, & Barry, 1987; Van Emmerik et al., 2014), and may provide coordination measures that possibly identify alteration in movement patterns that could be improved.

The vector coding technique is a spatial measurement focusing on the angular displacement relationship of two elements. Since the calculation is based on a single variable (i.e., angular displacement), this technique allows a simple interpretation regarding the spatial relationship between elements, which in turns can be helpful to diagnose what may be improved in the task performance. In this way, the key advantage of the vector coding technique is its simplification to analyze the organization between two elements, contrary to CRP that employs displacement and velocity. The coupling angle information can provide clinicians with important insights regarding the movement state (Van Emmerik et al., 2014), which in turn can be very helpful to implement rehabilitation protocols for those with gait impairment.

The primary output of vector coding technique is a series of coupling angles that allow the identification of two elements, such as segments or joints, which may act: (1) in the same direction (in-phase); (2) in opposite direction (anti-phase); (3) the proximal element leading the movement during performance (proximal-element-phase); and (4) the distal element leading the movement during performance (distal-element-phase). These four coordination modes might be interpreted according to the direction of the coupling angles that are calculated using the vector coding technique (Arnold, Caravaggi, Fraysse, Thewlis, & Leardini, 2017; Chang, Van Emmerik, & Hamill, 2008; Needham, Naemi, & Chockalingam, 2014; Van Emmerik et al., 2014).

Another important aspect that the vector coding technique provides is the coordination variability between multiple cycles or repetitions. The coordination variability has been calculated from multiple gait cycles of running (Hafer & Boyer, 2017; Takabayashi et al., 2018), walking (Needham et al., 2014; Tepavac & Field-Fote, 2001; Van Emmerik et al., 2014; Van Emmerik, Hamill, & McDermott, 2005), and during the triple jump (Wilson, Simpson, Van Emmerik, & Hamill, 2008). The coordination variability might be interpreted as the individual's ability to explore the degrees of freedom inherent to the task, which in turn provides a way for understanding how flexible or consistent the nervous system is to reproduce an action (Hamill et al., 2012; Van Emmerik et al., 2005; Van Emmerik & Van Wegen, 2000).

Based upon the unique and important information that the vector coding technique can provide, the main purpose of this study was to investigate the effects of stroke on gait intralimb coordination using a vector coding technique. More specifically, we aimed to identify the coordination between thigh-shank and shank-foot segments of the paretic, non-paretic, and control lower limbs on the

Table 1
General characteristics (mean ± SD) for non-disabled controls and individuals with stroke.

Characteristics	Control	Stroke	*p* value
Male/Female	3/15	12/13	–
Age (years)	60.8 ± 8.2	60.0 ± 6.21	0.718
Mass (kg)	63.33 ± 13.61	66.76 ± 11.90	0.917
Height (m)	1.56 ± 0.06	1.61 ± 0.09	0.150
Time of lesion (months)	–	58.40 ± 50.63	–
Hemiparesis (Left/Right)	–	13/12	–
Ischemic/Hemorrhagic	–	19/6	–

basis of the frequency of four identified modes of coordination, namely in-phase, anti-phase, proximal-segment-phase, and distal-segment-phase during the entire gait cycle as well as during stance and swing periods of gait separately. In addition, we examined the coordination variability between gait cycles of these segmental couplings. Because previous studies have demonstrated that lower limb segments are less dissociated for both paretic and non-paretic limbs (Barela et al., 2000; Barela et al., 2002; Combs et al., 2013; Rinaldi & Monaco, 2013), we hypothesized that the intralimb coordination of both paretic and non-paretic limbs would be predominantly in-phase, especially during the swing period. We also hypothesized that, due to the lack of movement control of the paretic limb during walking, the coordination variability of paretic limb would be higher than the non-paretic and control limbs.

2. Materials and methods

2.1. Participants

Twenty-five individuals with hemiparesis due to stroke and 18 non-disabled controls, aged between 46 and 71 years old, conveniently sampled, participated in this cross-sectional study. The inclusion criteria for the individuals with stroke were chronic hemiparetic gait after an ischemic or hemorrhagic stroke, present at least 6 months post stroke episode; absence of cardiac (or medical clearance for participation), orthopedic, pulmonary disease or neurologic impairment other than stroke; ability to follow verbal commands; and capacity to walk 10 m with or without assistance. The inclusion criteria for the non-disabled controls were no known musculoskeletal injuries or neurological disorders that could impair their ability to walk. Table 1 presents general characteristics of participants from both groups.

The Institutional Review Ethics Board of the Cruzeiro do Sul University approved the procedures of this study. All participants provided informed consent before the experimental session.

2.2. Procedures

Walking was performed on a 7-m walkway equipped with 2 embedded force platforms (Kistler, model 9286BA). A computerized gait analysis system (VICON, Oxford Metrics, Ltd.) with 7 infrared cameras was used to acquire data from reflective markers that were placed on the main body landmarks according to the Vicon Plug-In Gait model (Vicon, 2010). After a calibration trial, participants were asked to walk barefoot at a comfortable self-selected speed with no assistive device.

Before data acquisition, all participants practiced for a few trials until they felt comfortable with the laboratory environment and with the walking task. After this familiarization, at least five trials were recorded for each participant at a frequency of 100 Hz.

2.3. Data analysis

One intermediate stride per trial for each limb (paretic and non-paretic for individuals with stroke, and right and left for non-disabled controls) by each participant was analyzed, for a total of 5 trials. The trial selection was based on the best visualization of the markers and on participants walking with no interruption throughout the walkway. Based on the ground reaction force data from the force platforms and through visual inspection, a gait cycle (stride) was defined by two consecutive initial contacts of the same limb with the ground along the progression line. In addition, foot contacts and toe-offs during each stride were identified based on the foot vertical velocity (O'Connor, Thorpe, O'Malley, & Vaughan, 2007) for subsequent calculation of the spatio-temporal organization of walking.

When necessary, interpolation of missing signals from the markers (maximum 5 consecutive frames) was performed with the Nexus software (VICON, Oxford Metrics, Ltda.). The calculation of thigh, shank and foot angles was performed with TheMotionMonitor software (Innovative Sports Training, Inc.). Subsequent analyses, including spatio-temporal parameters, coordination modes and their frequency, and coordination variability, were performed using custom routines written in Matlab (MathWorks, Inc.).

2.4. Spatio-temporal gait parameters

The main gait parameters included mean walking speed, stride length, stride speed, and stance duration. The mean walking speed

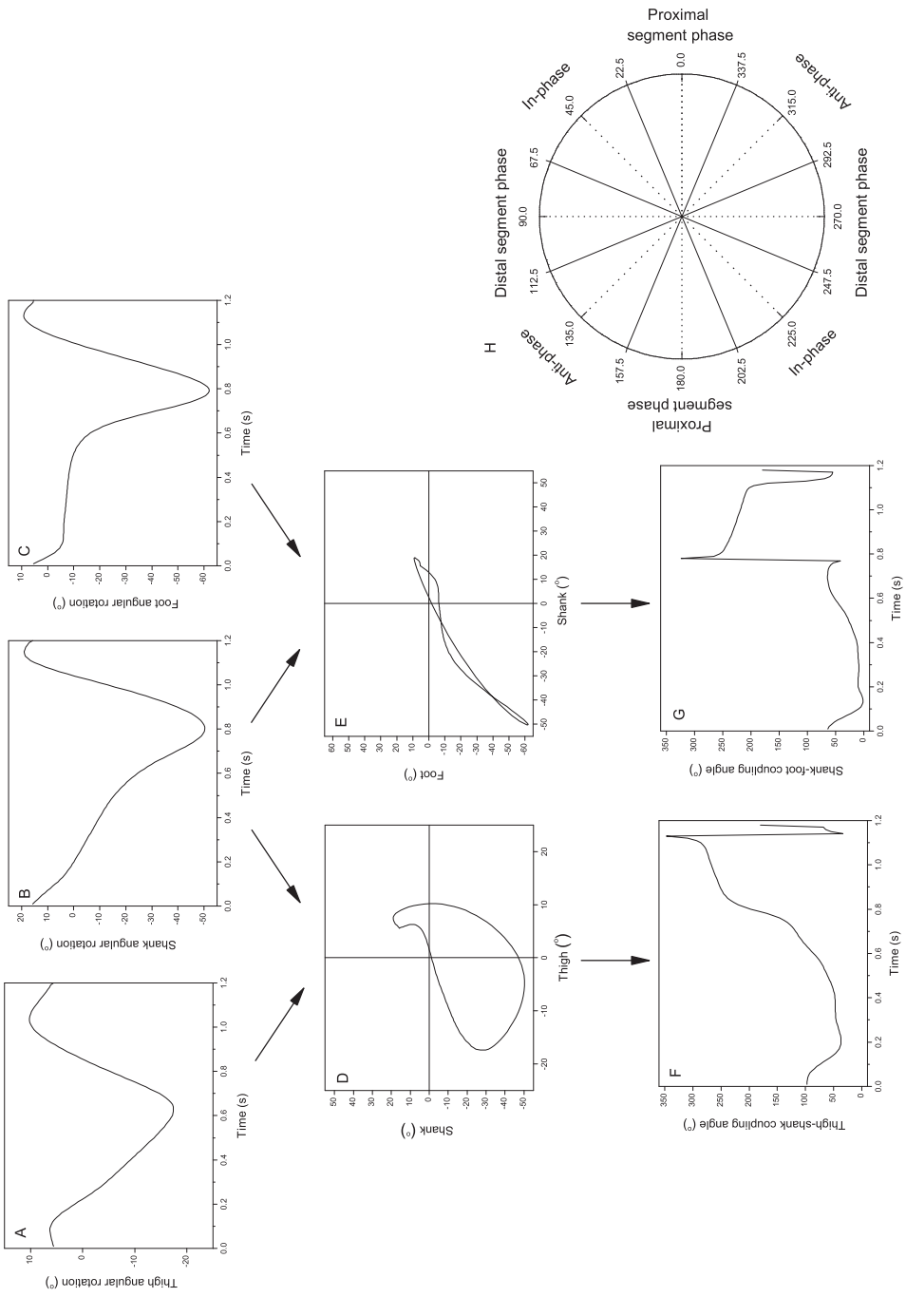

Fig. 1. Angular displacement of thigh (A) shank (B) and foot (C), and their respective angle-angle plots of thigh-shank (D) and shank-foot (E) displacement. Phase angles for the entire gait cycle were obtained through vector coding for thigh-shank (F) and shank-foot (G) couplings. Representation of the four different coordination modes identified from the phase angle data (H). Note: the coupling angle definitions are $0^o \leq \gamma < 22.5^o$, $157.5^o \leq \gamma < 202.5^o$, $337.5^o \leq \gamma < 360^o$ for proximal-segment-phase; $22.5^o \leq \gamma < 67.5^o$, $202.5^o \leq \gamma < 247.5^o$ for in-phase; $67.5^o \leq \gamma < 112.5^o$, $247.5^o \leq \gamma < 292.5^o$ for distal-segment-phase; and $112.5^o \leq \gamma < 157.5^o$, $292.5^o \leq \gamma < 337.5^o$ for anti-phase.

was calculated as the ratio between the distance covered by two strides and its duration, determined by the position of the sacrum marker; stride length was calculated from the distance between two successive initial contacts of each foot to the ground, determined by the position of the heel marker; stride speed was calculated by the ratio between stride length and its duration; and stance duration was calculated by the time (in % of the gait cycle) of foot contact to the ground during the gait cycle (Perry, 1992).

2.5. Intralimb coordination

The coordination patterns between thigh-shank and shank-foot were analyzed for the paretic, non-paretic and control limbs using the vector coding technique (Boyer, Freedman Silvernail, & Hamill, 2014; Freedman Silvernail, Van Emmerik, Boyer, Busa, & Hamill, 2018; Hafer & Boyer, 2017; Van Emmerik et al., 2014). Fig. 1 shows the angular displacement in the sagittal plane of thigh (A), shank (B), and foot (C), angle-angle plot of thigh-shank (D) and shank-foot (E), the coupling angles of thigh-shank (F) and shank-foot (G), and the representation of the coordination modes (H). The coupling angle was calculated, using the following equation:

$$\gamma_{j,i} = \tan^1 (y_{j,i+i} - y_{j,i} / x_{j,i+i} - x_{j,i}) \quad (1)$$

where, γ corresponds to coupling angle, i corresponds the consecutive data points in a cycle, and j corresponds to the gait cycle considered for the proximal (x) and the distal (y) segment angular displacement.

From the coupling angle data (Fig. 1F and G), the coordination modes can be identified (Fig. 1H). More specifically, for both coupling angles (i.e., thigh-shank and shank-foot), the four identified coordination modes represent the following:

(1) *in-phase mode*; both segments rotate in the same direction, either clockwise (backward) or counter-clockwise (forward);
(2) *anti-phase mode;* both segments rotate in opposite direction, with the proximal segment (thigh/shank) rotating in clockwise direction and the distal segment (shank/ft) in counter-clockwise direction, or vice-versa;
(3) *proximal segment-phase mode*; both limbs rotate in the same direction (either clockwise or counter-clockwise), but proximal segment is ahead of distal segment;
(4) *distal segment-phase mode*; both limbs rotate in the same direction (either clockwise or counter-clockwise), but distal segment is ahead of proximal segment.

Finally, the frequency of each coordination mode was calculated as the percentage of occurrence during stance period, swing period, and the entire gait cycle.

2.6. Intralimb coordination variability

To obtain the coordination variability during stance period, swing period, and entire gait cycle, distinct steps were considered since arithmetic mean calculus to a series of polar angles can result in errors. Therefore, mean coupling angles were calculated using circular statistics (Batschelet, 1981), applied to all trials for each group using the following equations:

$$\overline{x}_i = 1/n \sum_{j=1}^{n} (\cos\ \gamma(j,i)) \quad (2)$$

$$\overline{y}_i = \frac{1}{n} \sum_{j=1}^{n} (\sin\ \gamma(j,i)) \quad (3)$$

where, n refers to number of gait cycles ($n = 5$), and $\overline{x}$ and $\overline{y}$ refer to mean horizontal and vertical components, respectively, across the multiple gait cycles (j) for each data point (i) (% stride cycle). Eqs. (2) and (3) are then applied to all data points (i) in the gait cycle from 0 to 100%. Finally, in eqs. 4–5 the mean coupling angle is calculated and normalized in the range from 0° to 360°:

$$\overline{\gamma} = \arctan (\overline{y}_i / \overline{x}_i), \text{ if } \overline{x}_i > 0 \quad (4)$$

$$\overline{\gamma}_i = 180 + \arctan (\overline{y}_i / \overline{x}_i), \text{ if } \overline{x}_i < 0 \quad (5)$$

The coordination variability is then derived from the vector length ($\overline{r}$) of the horizontal ($\overline{x}$) and vertical ($\overline{y}$) components of this mean coupling angle (Batchelet, 1981):

$$\overline{r}_i = \sqrt{\overline{x}^2 + \overline{y}^2} \quad (6)$$

As shorter vector length, relative to the unit circle represents more variability in the coupling angles, the angular variance in coupling angles is expressed as:

$$2 * (1 - \overline{r}_i) \quad (7)$$

The coordination variability measure (CV) for each % of the gait cycle (i) is then expressed as the angular standard deviation and normalized to the range 0–180°:

$$CVi = \sqrt{2 * (1 - \overline{r}_i)} * 180/\pi \quad (8)$$

Finally, the CV was averaged across the stance phase, swing phase and the entire gait cycle.

3. Statistical analyses

One-way analysis of variance (ANOVA) was employed to compare the mean walking speed between both groups (stroke and non-disabled controls). For the following analyses we assumed that gait of non-disabled controls is symmetric and data from right and left strides were averaged together before making comparisons with individuals with stroke. Data from non-disabled controls were considered the "control limb". For spatial-temporal gait parameters, one-way multivariate analysis of variance (MANOVA) was employed, using limb (control, paretic, non-paretic) as a factor. The dependent variables were stride length, stride speed, and stance duration. Since there was a group difference for mean walking speed and stride speed, for the remaining analyses, we ran multivariate analyses of covariance (MANCOVAs) with the same factor (limb) but having the stride speed as covariate. Six MANCOVAs were employed. The dependent variables were the frequency for in-phase, anti-phase, thigh-phase, and shank-phase during the stance period (1), swing period (2), and the entire gait cycle (3) for the thigh-shank coupling; and frequency for in-phase, anti-phase, shank-phase, and foot-phase during the stance period (4), swing period (5), and the entire gait cycle (6) for the shank-foot coupling. Finally, six ANCOVAs were employed for the coordination variability of the thigh-shank and shank-foot for the stance period, swing period, and the entire gait cycle.

When necessary, univariate analysis and post hoc pairwise comparisons with Bonferroni adjustments were employed. An alpha level of 0.05 was set for all statistical tests, which were performed using SPSS software.

4. Results

4.1. Spatial-temporal gait parameters

ANOVA involving the mean walking speed revealed that individuals with stroke walked slower (0.52 ± 0.25 m/s) than their peers (1.09 ± 0.09 m/s) ($F_{1,41} = 58.78$, $p < .001$). Table 2 depicts the mean (± SD) values for stride length, stride speed, and stance duration for the control, paretic and non-paretic limbs. MANOVA revealed a main effect of limb (Wilks' Lambda = 0.47, $F_{4,128} = 14.41$, $p < .001$). Univariate analyses revealed differences for stride length ($F_{2,65} = 26.18$, $p < .001$) and speed ($F_{2,65} = 35.29$, $p < .001$), and for the stance period ($F_{2,65} = 36.45$, $p < .001$). Post hoc tests revealed shorter and slower stride for the paretic and non-paretic limbs compared to the control limb, longer stance duration for the paretic and non-paretic limbs than the control limb, and longer stance duration for the non-paretic than the paretic limb.

4.2. Intralimb coordination

Fig. 2 shows sample time series for the coupling angles of control, paretic, and non-paretic limbs. Overall, the pattern differs at the beginning (0–20%) and end (90–100%) of the gait cycle, and at the end of stance period for both thigh-shank (Fig. 2A) and shank-foot couple (Fig. 2B). The vertical lines indicate the temporal lag in the instant of toe-off when comparing the paretic and non-paretic limbs to the control limb.

4.2.1. Thigh-shank coupling

Fig. 3 shows mean (± SD) values of frequency in the four coordination patterns for the thigh-shank coupling during stance and swing periods and the entire gait cycle. MANCOVA for the stance period (Fig. 3A) revealed a main effect of limb (Wilks' Lambda = 0.71, $F_{8,124} = 2.78$, $p = .007$). Univariate analysis revealed differences for thigh-phase ($F_{2,64} = 11.75$, $p < .001$) and shank-phase ($F_{2,64} = 18.47$, $p < .001$) coordination patterns. Post hoc tests indicated that paretic and non-paretic limbs presented higher frequency in the thigh-phase ($p < .001$) and lower frequency in shank-phase ($p < .001$) coordination patterns than the control limb. No differences between paretic and non-paretic limbs were observed for the stance period.

MANCOVA for the swing period (Fig. 3B) revealed a main effect of limb (Wilks' Lambda = 0.38, $F_{8,122} = 9.48$, $p < .001$). Univariate analysis revealed differences for in-phase ($F_{2,64} = 33.16$, $p < .001$) and thigh-phase ($F_{2,64} = 20.93$, $p < .001$) coordination patterns. Post hoc tests indicated that the paretic limb presented higher frequency in the in-phase coordination pattern than the control ($p < .001$) and non-paretic limbs ($p < .001$). The non-paretic limb presented higher frequency in the thigh-phase coordination pattern than the control ($p < .001$) and paretic limb ($p < .001$).

MANCOVA for the entire gait cycle (Fig. 3C) revealed a main effect of limb (Wilks' Lambda = 0.41, $F_{8,122} = 8.68$, $p < .001$).

Table 2
Mean (± SD) for stride and speed length, and stance period of control, paretic (P) and non-paretic (NP) limbs.

Measures	Limb			*p* value		
	Control	*P*	*NP*	*ControlxP*	*ControlxNP*	*PxNP*
Stride length (m)	1.15 ± 0.13	0.73 ± 0.23	0.73 ± 0.24	< 0.001	< 0.001	1.000
Stride speed (m/s)	1.07 ± 0.18	0.53 ± 0.25	0.53 ± 0.26	< 0.001	< 0.001	1.000
Stance period (%)	60.32 ± 1.45	63.24 ± 5.84	73.79 ± 7.00	< 0.001	< 0.001	0.047

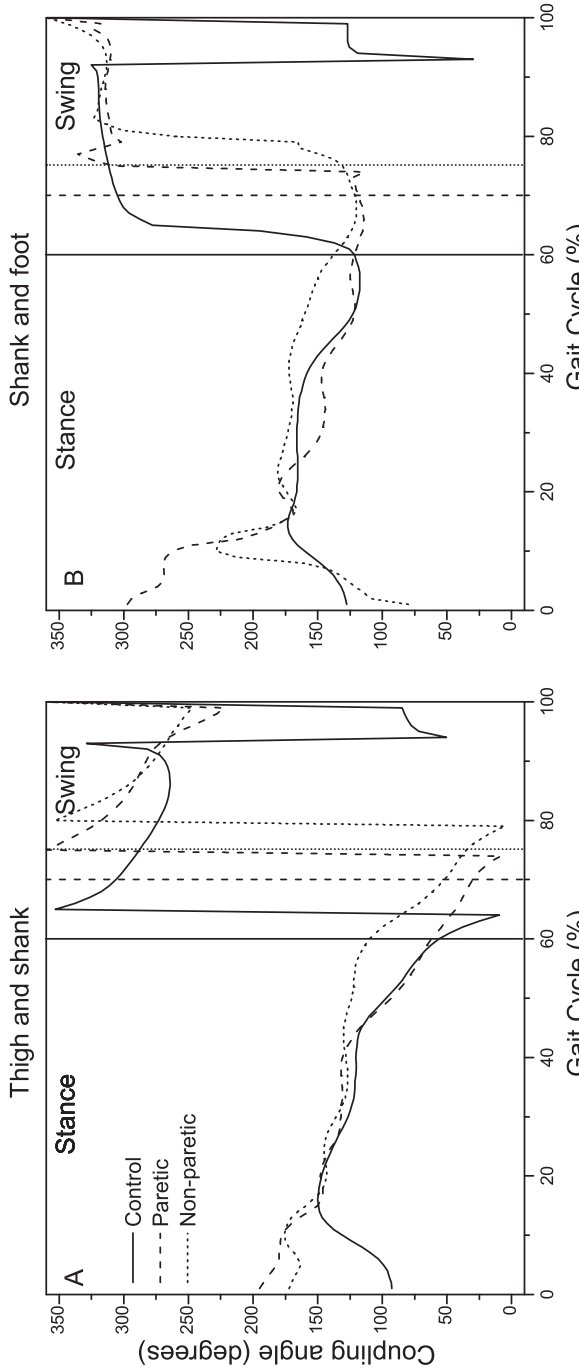

Fig. 2. Exemplar time series of thigh-shank (A) and shank-foot (B) coupling angles during the gait cycle of control (continuous), paretic (dash), and non-paretic (short dash) limbs. Note: vertical lines indicate the transition between stance and swing periods.

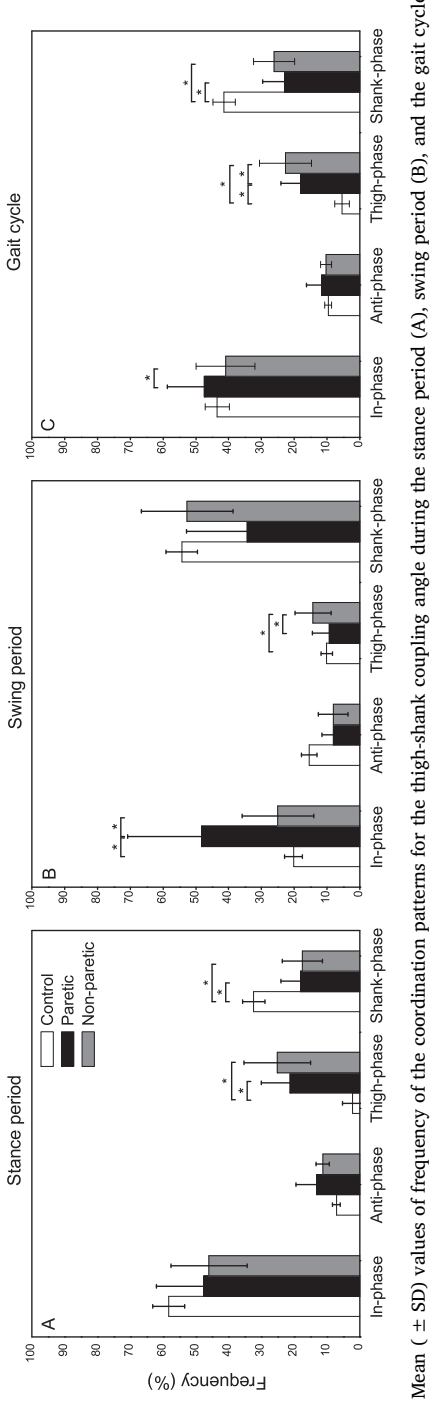

Fig. 3. Mean (± SD) values of frequency of the coordination patterns for the thigh-shank coupling angle during the stance period (A), swing period (B), and the gait cycle (C).

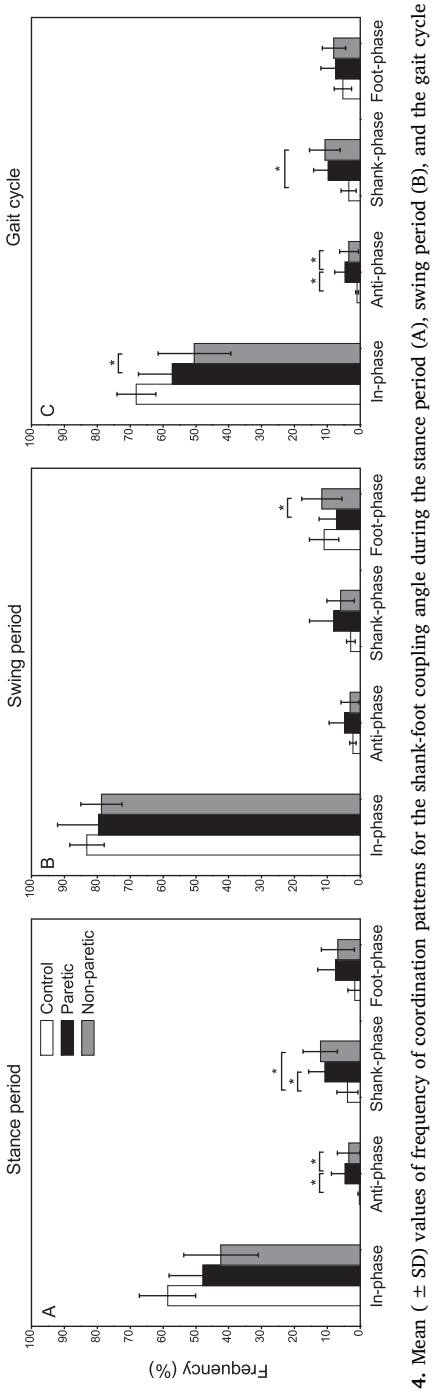

Fig. 4. Mean (± SD) values of frequency of coordination patterns for the shank-foot coupling angle during the stance period (A), swing period (B), and the gait cycle (C).

Univariate analysis revealed differences for in-phase ($F_{2,64}$ = 3.37, p = .041), thigh-phase ($F_{2,64}$ = 13.96, p < .001), and shank-phase ($F_{2,64}$ = 19,68, p < .001) coordination patterns. Post hoc tests indicated that the paretic (p = .010) and non-paretic limb (p < .001) presented higher frequency in the thigh-phase and lower frequency in the shank-phase (p < .001) coordination pattern than the control limb. Finally, the paretic limb presented higher frequency in the in-phase (p = .049) and lower frequency in the thigh-phase (p = .011) coordination patterns than the non-paretic limb.

4.2.2. Shank-foot coupling

Fig. 4 shows mean (± SD) values of frequency in the four coordination patterns for the shank-foot coupling during stance period, swing period, and the entire gait cycle. MANCOVA for the stance period (Fig. 4A) revealed a main effect of limb (Wilks' Lambda = 0.71, $F_{8,122}$ = 2.78, p = .007). Univariate analysis revealed differences for anti-phase ($F_{2,64}$ = 4.73, p = .012) and shank-phase ($F_{2,64}$ = 6.23, p = .003) coordination patterns. Post hoc tests indicated that the paretic limb presented higher frequency in the anti-phase (p < .05) and shank-phase (p < .001) coordination patterns than the control limb. The paretic limb also presented higher frequency in the anti-phase (p < .05) coordination pattern compared to the non-paretic limb. Finally, the non-paretic limb presented higher frequency in the shank-phase (p < .005) coordination pattern than the control limb.

MANCOVA for the swing period (Fig. 4B) revealed a main effect of limb (Wilks' Lambda = 0.69, $F_{8,122}$ = 3.08, p = .003). Univariate analysis revealed differences for foot-phase ($F_{2,64}$ = 6.30, p = .003) coordination patterns. Post hoc tests indicated that the paretic limb presented lower frequency in the foot-phase (p < .005) coordination pattern than the non-paretic limb.

MANCOVA for the entire gait cycle (Fig. 4C) revealed a main effect of limb (Wilks' Lambda = 0.64, $F_{8,122}$ = 3.84, p < .001). Univariate analysis revealed differences for in-phase ($F_{2,64=}$5.86, p = .005), anti-phase ($F_{2,64}$ = 5.18, p = .008), and shank-phase ($F_{2,64}$ = 3.76, p = .028) coordination patterns. Post hoc tests indicated that the paretic limb presented higher frequency in the anti-phase (p < .05) coordination pattern and the non-paretic limb presented higher frequency in shank-phase (p < .05) coordination patterns than the control limb. Finally, the paretic limb presented higher frequency in the in-phase (p < .01) and anti-phase (p < .05) coordination patterns than the non-paretic limb.

4.3. Intralimb coordination variability

Fig. 5 shows mean values of sample time series for coordination variability during the gait cycle of control, paretic, and non-paretic limbs. Overall, it appears that the paretic and non-paretic limbs presented more variability than the control limb, with the paretic limb presenting variability more similar to non-paretic limb. These differences visually seem to occur more in the relationship between shank-foot. The vertical lines indicate the temporal lag in the instant of toe-off when compared the paretic and non-paretic limb to control limb.

Fig. 6 shows mean (± SD) values of coordination variability of the thigh-shank and shank-foot coupling angles. ANCOVA for the thigh-shank coupling revealed no effect of limb during the stance period ($F_{2,64}$ = 1.90, p = .157), swing period ($F_{2,64}$ = 0.65, p = .524) and for the entire gait cycle ($F_{2,64}$ = 1.13, p = .328). Similarly, ANCOVA for the shank-foot coupling revealed no effect of limb during the stance period ($F_{2,64}$ = 0.80, p = .454), swing period ($F_{2,64}$ = 0.71, p = .495) and for the entire gait cycle ($F_{2,64}$ = 0.25, p = .781).

5. Discussion

This study investigated the effects of stroke on gait intralimb coordination using the vector coding technique. We hypothesized that paretic and non-paretic limbs would be moved predominantly in-phase, especially during swing period. We also hypothesized that the coordination variability of paretic limb would be higher than the non-paretic and control limbs. Based upon our results, the first hypothesis was confirmed, indicating that stroke affects the coordination and rotation of thigh and shank segments. The second hypothesis was refuted as no difference was observed in the coordination variability among limbs. Our results showed that during the stance period individuals with stroke presented higher frequency of thigh-phase and lower frequency of shank-phase for the thigh-shank coupling and higher frequency of shank-phase for the shank-foot coupling compared to the non-disabled control leg. These results suggest that as consequence of stroke, the proximal segments lead the movement during stance period for both, paretic and non-paretic limbs. During the swing period, the paretic limb showed higher frequency in the in-phase than non-paretic and control limbs, and the non-paretic showed higher frequency in the thigh-phase than paretic and control limbs for the thigh-shank coupling. The non-paretic limb presented higher frequency in the foot-phase than paretic limb for the shank-foot coupling. Moreover, our results showed that intralimb coordination using the vector coding in individuals with stroke should be conducted considering stance and swing periods separately rather than the entire gait cycle.

The vector coding technique clearly showed that, although intralimb coordination of individuals with stroke can be examined throughout the entire gait cycle as previous studies have examined employing CRP (Barela et al., 2000; Barela et al., 2002), it seems more suitable to consider the stance and swing periods separately. The main reason for that is because each period demands specific requirements, leading to different relationships between lower limb segments (Inman, Ralston, Todd, Childress, & Gard, 2006; Perry, 1992; Winter, 1991). Considering the entire gait cycle could mask information regarding important changes in the intralimb relationship of a pair of segments during walking.

As mentioned previously, the vector coding technique has been often employed to investigate coordination during the performance of different motor tasks (Boyer et al., 2014) (Wilson et al., 2008) (Chang et al., 2008; Hafer & Boyer, 2017), but its use to examine pathological gait is still lacking. Our results showed that the vector coding analysis can reveal changes in intralimb

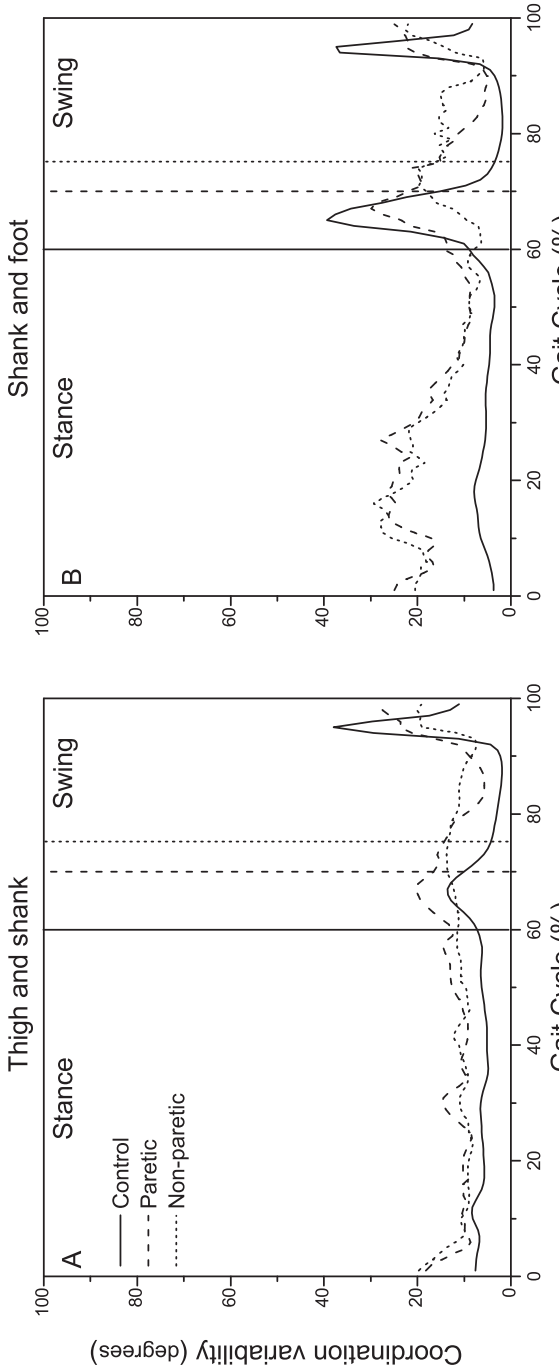

Fig. 5. Mean time series of coordination variability for thigh-shank (A) and shank-foot (B) coupling angles during the gait cycle of control (continuous), paretic (dash), and non-paretic (short dash) limbs. Note: vertical lines indicate the transition between stance and swing periods.

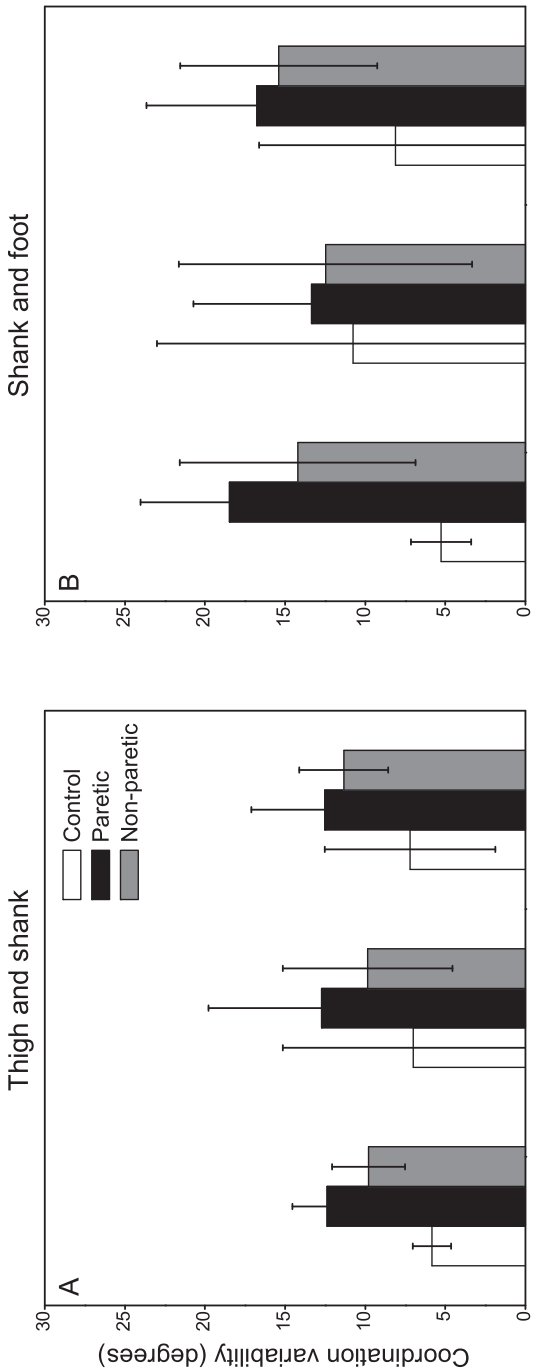

Fig. 6. Mean (± SD) values of coordination variability during the stance and swing period, and the gait cycle of the control, paretic, and non-paretic limbs for the thigh-shank (A) and shank-foot (B) coupling angles.

coordination in both the paretic and non-paretic limbs after stroke, which will be discussed below.

5.1. Thigh-shank relationship

In the stance period, the thigh-shank relationship of individuals with stroke was characterized by higher frequency in the thigh-phase and lower frequency in the shank-phase coordination patterns compared to the non-disabled individuals. These thigh-shank intralimb changes might be related to the short step length observed for the paretic and non-paretic limbs, commonly observed in individuals with stroke (Olney & Richards, 1996), which leads to the forward rotation of thigh during most of the stance period. Moreover, the lower frequency of the shank-phase coordination pattern suggests that individuals with stroke experience difficulties in rotating the shank segment. These thigh-shank intralimb coordination changes may be due to the muscle weakness and spasticity (Bourbonnais & Vanden Noven, 1989; Lamontagne, Malouin, Richards, & Dumas, 2002; Neckel, Pelliccio, Nichols, & Hidler, 2006; Sanchez et al., 2017) that impact the coordination patterns.

The vector coding technique also revealed differences in the intralimb coordination between the thigh-shank during the swing period. The swing period is a potentially unstable period as it requires body support by the contralateral limb as the swing limb advances forward to execute a step (Inman et al., 2006; Winter, 1991). Individuals with stroke presented in the paretic limb higher frequency in the in-phase coordination pattern, due to reduced knee flexion movement as presented in previous studies (Moore, Schurr, Wales, Moseley, & Herbert, 1993; Raja, Neptune, & Kautz, 2012). Because of this diminished dissociation between the thigh and shank segments, the intralimb coordination pattern is kept mostly in the in-phase mode. In the non-paretic limb, individuals with stroke adopted a higher frequency in the thigh-phase coordination pattern. Such strategy might be related to the reduced step duration (Chen et al., 2005), as individuals with stroke have to move the non-paretic limb forward and perform heel-strike as soon as possible avoiding the unstable single-stance performed by the paretic limb.

5.2. Shank-foot relationship

The vector coding technique also showed differences for paretic and non-paretic limbs in the shank-foot relationship of individuals with stroke. During the stance period, a higher predominance in the anti-phase and shank-phase coordination patterns was observed for the paretic compared to the control limb, indicating that the shank and foot segments rotate in opposite directions, although in a small portion of the cycle, as the shank rotates forward in relation to the foot. These results can be related to the alterations in muscle activation of tibialis anterior of individuals with stroke (Turns, Neptune, & Kautz, 2007). The weakness of muscles responsible to perform the rotation of foot forward may be the reason that the shank leads the movement during the stance period. The differences in the shank-foot relationship were also observed during the swing period, as the non-paretic limb presented higher frequency in the foot-phase coordination patterns compared to the paretic limb. In the same way, these results may be attributed to the muscle weakness of shank and foot segments in individuals with stroke during the swing period, as previously shown by diminished dorsiflexion movement of ankle joint observed in individuals with stroke during the swing period in the paretic limb (Moore et al., 1993).

During the stance period, the non-paretic limb presented lower frequency in the anti-phase coordination pattern, and during the swing period the non-paretic limb presented higher frequency in the foot-phase coordination pattern. These results indicate that individuals with stroke present a lack of control in the paretic distal segments and compensate the movement of these segments.

5.3. Intralimb coordination variability

The examination of the coordination variability indicated no difference among the control, paretic, and non-paretic limbs when stride speed is taking into consideration. Such observation was surprising and against the raised hypothesis. Individuals with gait impairment, such as those with risk of falling (Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997), Parkinson's disease, and Huntington's disease (Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998) present high variability in terms of spatio-temporal parameters. This variability indicates the inconsistency to control the many degrees of freedom to perform cyclic skills, being dysfunctional to perform the motor task (Hamill et al., 2012). Although the sample data (Fig. 5) and the average data (Fig. 6) do seem to suggest increased coordination variability in both paretic and non-paretic limbs, the observed lack of difference in this study might be due to the relatively high variation inter-individual variation observed in all limbs, including the control limb (Fig. 6).

In addition to the high inter-individual variability, the results from the ANCOVA showing no differences in variability between the limbs, suggest that the observed walking speed differences between the stroke and control groups may be a another major factor in the lack of statistical difference between both limbs from the stroke group and the control limb. It is well known that walking speed affects coordination parameters (Haddad, van Emmerik, Wheat, Hamill, & Snapp-Childs, 2010; Krasovsky & Levin, 2010; Levin et al., 2009; Seay, Haddad, van Emmerik, & Hamill, 2006), and all participants of this study adopted comfortable speed, which was different between both groups (stroke and control). Since this aspect could influence some of the results, we used stride speed as covariate, and the results of this analysis suggests that speed indeed had an impact on coordination variability. Future studies may want to control the gait speed between groups in order to clarify this issue.

5.4. Limitations

Although this study presents promising and unique results, it also presents some limitations. First, our study focuses only in the

intralimb coordination of two coupling structures (thigh-shank and shank-foot segments). Another important issue to be considered is the number of gait cycles considered in the present study. The investigation of coordination variability from the vector coding technique in healthy individuals involved more cycles than in the present study, for example, ten strides (Hafer & Boyer, 2017). However, studies considering five strides have also been published (Chang et al., 2008; Needham et al., 2014). As the coordination variability of pathologic gait must be treated carefully (Hamill et al., 2012), future studies employing the vector coding technique to investigate individuals with stroke should consider more gait strides and a larger sample. Second, couple of other body segments and interlimb coordination should also be addressed in future studies. Even with these limitations, the use of the vector coding technique seems to be revealed a promising and useful tool to examine coordination gait in individuals with stroke that might reveal important information to implement gait intervention protocols.

6. Conclusion

Individuals with stroke present different intralimb coordination in relation to age-matched controls, and the vector coding technique was a useful tool to identify the frequency of specific coordination modes. Stance and swing periods should be examined separately because it reveals differences between paretic and non-paretic limbs that could be masked when considering the entire gait cycle. Our results revealed that differences in the intralimb coordination of individuals with stroke are mostly due to the lack of distal segment control. Qualitative information related to coordination measures is required to assist and orient gait rehabilitation protocols of individuals with stroke (Levin et al., 2009). The vector coding technique employed here may provide detailed information about intralimb coordination changes related to these gait rehabilitation protocols, and may help in identifying the 'gold parameter' to be employed in terms of coordination assessment in individuals with stroke (Reisman & Scholz, 2003).

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