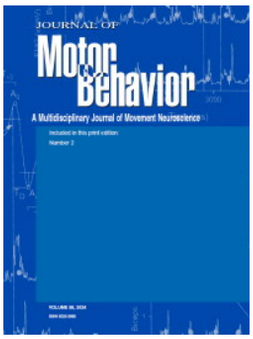

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RESEARCH ARTICLE

Interlimb Coordination during Double Support Phase of Gait in People with and without Stroke

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ABSTRACT. This study aims to identify differences between participants with and without stroke regarding the ipsilesional and contralesional lower limbs kinematics, kinetics, muscle activity and their variability during double support phase of gait. Eleven post-stroke and thirteen healthy participants performed 10 gait trials at a self-selected speed while being monitored by an optoelectronic motion capture system, two force plates and an electromyographic system. The following outcomes were evaluated during the double support: the time and the joint position; the external mechanical work on the centre of mass; and the relative electromyographic activity. Both, contralesional/ipsilesional and dominant/non-dominant of participants with and without stroke, respectively, were evaluated during double support phase of gait in trailing or leading positions. The average value of each parameter and the coefficient of variation of the 10 trials were analysed. Post-stroke participants present bilateral decreased mechanical work on the centre of mass and increased variability, decreased contralesional knee and ankle flexion in trailing position, increased ipsilesional knee flexion in leading position and increased variability. Increased relative muscle activity was observed in post-stroke participants with decreased variability. Mechanical work on the centre of mass seems to be the most relevant parameter to identify interlimb coordination impairments in post-stroke subjects.

Keywords: Contralesional, Ipsilesional, Lower Limbs, Biomechanics, Quality Movement, Variability

Introduction

Stroke, in 2019, remained the second-leading cause of death and the third-leading cause of death and disability combined in the world (Feigin et al., 2021). The sensorimotor repercussions of a stroke can cause limitations in daily activities and participation restriction, both in the professional and social context (Rajsic et al., 2019). Gait is one of the most impaired functional task with a great impact on the autonomy (Newman et al., 2006). Therefore, gait recovery is one of the main objectives for post-stroke patients and their rehabilitation (Belda-Lois et al., 2011; Bohannon et al., 1991; Langhorne et al., 2009; Mudge & Stott, 2007; Shumway-Cook & Woollacott, 2017; Sousa et al., 2013b).

Stroke commonly occurs in the middle cerebral artery territory (Crafton et al., 2003; Schiemanck et al., 2006), compromising the transcallosal interhemispheric interaction (Cleland & Madhavan, 2021, 2022; Grefkes & Fink, 2011; Shimizu et al., 2002; Ward & Cohen, 2004), the predominantly contralesional (CONTRA) cortical motor areas (Kandel et al., 2014; Shumway-Cook & Woollacott, 2017), but also the predominantly ipsilesional (IPSI) subcortical structures involved in neural control of gait (Kandel et al., 2014; Rothwell, 2009, 2012). These above-mentioned aspects may justify the influence of both lower limbs on post-stroke gait impairments (Matsuyama et al., 2004). Some authors demonstrated that post-stroke sequelae involves impairment of function and performance in both CONTRA and IPSI limbs (Cramer et al., 1997; Genthon et al., 2008; Silva et al., 2012, 2015; Sousa et al., 2013b; 2015), with a less severe expression in IPSI side compared to the CONTRA one (Alagona et al., 2001; Schaefer et al., 2009). In fact, the main changes in post-stroke gait are mainly expressed in an asymmetric behaviour of both lower limbs (Arya & Pandian, 2014; Bajwa et al., 1992; Crafton et al., 2003;

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This article has been corrected with minor changes. These changes do not impact the academic content of the article.

Olney et al., 1991, 1994; Olney & Richards, 1996; Schiemanck et al., 2006; Sousa et al., 2013b; Sousa & Tavares, 2015). From a biomechanical point of view, impairments in lower limbs (Beyaert et al., 2015; Kwakkel et al., 2017) have been extensively explored and described at a kinematic level (joint position and double support phase) (Balaban & Tok, 2014; Chen et al., 2005; Kim & Eng, 2004; Nadeau et al., 2013; Oken & Yavuzer, 2008; Woolley, 2001), kinetic (ground reaction forces) (Balaban & Tok, 2014; Bensoussan et al., 2006; Kim & Eng, 2004; Moseley et al., 1993; Nadeau et al., 2013; Woolley, 2001) and muscle activation (distal and proximal muscles) (Balaban & Tok, 2014; Daly et al., 2011; Den Otter et al., 2007; Lamontagne et al., 2000, 2002; Woolley, 2001). However, according to our knowledge in a previous scoping review, only some variables were considered in the study of the impairments of both limbs on post-stroke interlimb coordination, as joint kinematics, double-support time, external mechanical work and electromyographic activity and the combination of the different outcomes has been suggested (Couto et al., 2022).

The importance of studying the interlimb coordination in gait is based on the metabolic cost impact (Donelan et al., 2002a, 2002b; 2004; Krasovsky et al., 2012; Kuo, 2007; Kuo et al., 2005; Sousa & Tavares, 2015) and can be described at a biomechanical level (Donelan et al., 2002a, 2002b; Kuo, 2007; Kuo et al., 2005), at spinal cord level (Bajwa et al., 1992; Carson et al., 2014; Dietz, 1992), and at a supraspinal level (Drew et al., 2004; Matsuyama et al., 2004). Gait efficiency is, particularly, determinant in the double support phase of gait (Kuo, 2002, 2007; Kuo et al., 2005), in which the two lower limbs, in contact with the ground, cooperate to maintain the requirement of postural control and centre of mass progression (Kandel et al., 2014; Shumway-Cook & Woollacott, 2017). These gait requirements arise from the positive work of the posterior lower limb during propulsion (toe-off event) for centre of mass progression, that compensates for the energy expended on negative work by the anterior lower limb in initial contact (heel-strike event) and accommodation of body mass (Donelan et al., 2002a; Kuo, 2007). Therefore, the importance of exploring the interlimb coordination in people with stroke sequelae, in the double support phase of gait, is related, not only to the impact of the CONTRA limb on the IPSI limb, but also, to the impact of IPSI on the performance of the CONTRA, according to the requirements that each position of the lower limb (anterior or posterior) demands in this phase of gait.

Human gait is characterised by natural variations in motor performance, known as optimal movement variability in healthy adults (Hausdorff, 2005; Ó'Reilly & Federolf, 2021; Stergiou et al., 2006). Variability emerges as an aspect of motor control which can be quantified through linear measures, such as coefficient of variation (Harbourne & Stergiou, 2009), which may translate the variability measure and the impact of stroke on gait performance (Balasubramanian et al., 2009). Among healthy adults, both high- and low- variability can be associated with gait stability (Beauchet et al., 2009; Ó'Reilly & Federolf, 2021), even though traditionally, a higher variability in gait may correspond to decreased mechanical stability, whereas a lower variability corresponds to a more stable and cooperative system (Stergiou et al., 2006). Lower gait variability allows for better adaptability of the gait pattern to environmental changes (Cavanaugh & Stergiou, 2020; Hausdorff, 2005). Variability above or below the optimal level can be indicative of pathology (Hausdorff, 2005; Manor et al., 2010). In pathological conditions variability tends to be higher and studies in people with stroke sequelae have demonstrated high variability in spatiotemporal parameters (Allen et al., 2011; Balasubramanian et al., 2009; Chisholm et al., 2014).

Recognising the quantification of variability post-stroke as a possible result of a lower capacity for motor control or an adaptive strategy, its study emerges as a new point of analysis for a better understanding of the expression of the post-stroke lesion, adding knowledge to the neuromotor changes in this condition.

The neuro-anatomical and neuro-physiological perspective of motor control raises the hypothesis that the IPSI side may present an expression of the neuronal lesion itself (dysfunction). These findings add knowledge to the, already commonly described, biomechanical changes that identify adaptive strategies developed by the neuronal compensation mechanisms of the IPSI side in the face of CONTRA side injury (Kim & Eng, 2004; Olney et al., 1991, 1994; Parvataneni et al., 2007). Thus, it can be considered that the neural control of post-stroke interlimb coordination, may often be compromised (Dietz & Berger, 1984; Roerdink et al., 2007; Wu et al., 2009), in both lower limbs, due to the primary encephalic lesion and/or be the result of adaptive changes (Crone et al., 2003; Lamy et al., 2009; Murase et al., 2004).

Considering the above-mentioned information, it is suggested that post-stroke gait, as a coordinated rhythmic task of the lower limbs, involves bilateral neural and biomechanical alterations, which justify the asymmetric behaviour of both lower limbs. This has a greater impact on the step-to-step transition and related variability on the double support phase.

This study aims to compare biomechanical variables in the double support phase of gait expressing interlimb coordination between people with stroke sequelae and healthy people. Specifically, kinematic (double support time and joint position of the hip, knee, and ankle), kinetic (external work on the centre of mass) and electromyographic (mean muscle activity and muscle activation ratios) measures and their variability quantification in

lower limbs during double support phase of gait will be considered.

Methods

Study Design

A cross-sectional observational study was conducted to compare kinematic, kinetic and electromyographic variables between people "with stroke sequelae", named in the present study as the "stroke" group, and people "without history of a stroke and without self-reported disabilities" named as the "healthy" group. This group was used as a reference for the typical movement performance.

Participants

A group of eleven subjects (3 females and 8 males) with history of a single unilateral ischaemic stroke affecting the right ($n = 3$) and left ($n = 8$) hemispheres, which resulted in a motor control dysfunction of the contralesional lower limb (CONTRA), and a group of thirteen healthy subjects (4 females and 9 males) participated in the present study (Table 1). The target population was comprised of people from Porto and Braga regions. The final sample composed by volunteers (convenience sampling) was divided into the stroke group and the healthy group. The healthy group included sedentary adults without self-reported disabilities, recruited by direct invitation. Participants were excluded from the healthy group if they had one or more of the following criteria: altered mental state with interference in communication and cooperation (Lamontagne et al., 2000, 2002; Sousa et al., 2013b); history or sign of neurological dysfunction (Sousa et al., 2012); presence of pain that interfered with the performance of walking (Sousa et al., 2013a); history of anatomical deformities, osteoarticular, musculotendinous injury or lower limb surgery in the last 6 months (Sousa et al., 2012, 2013a); exposition to medication with interference in the motor performance of the lower limbs (Lamontagne et al., 2000, 2002); and practice of moderate (i.e., at least 30 min 5 days a week) or vigorous (i.e., at least 20 min 3 days a week) levels of physical activity (Thompson & Thompson, 2014).

To be included in the stroke group the participants need to fulfil the following criteria: diagnosis of ischaemic first-ever stroke, in chronic phase (Lamontagne et al., 2000, 2002; Sousa et al., 2013b), involving of middle cerebral artery territory, at the subcortical level, confirmed by Axial Computed Tomography (Lamontagne et al., 2002; Sousa et al., 2013b); lower limb sensorimotor impairment was screened by Fugl-Meyer Assessment of Sensorimotor Recovery After Stroke (Sousa et al., 2013b); and ability to walk at least 10 metres, with close supervision, if necessary, but without physical assistance (Lamontagne et al., 2000; Sousa et al., 2013b). In turn, people who had any of the exclusion criteria mentioned for the healthy group, or the presence of a lesion involving the brainstem or cerebellum were excluded (Lamontagne et al., 2000, 2002; Sousa et al., 2013b).

Ethical approval was obtained by the Institutional Ethics Committee of School of Health Polytechnic of Porto (CE 1372). All participants provided their written informed consent, before the data collection began, according to Declaration of Helsinki.

Instruments

Sample Selection and Characterization

A questionnaire was used to verify the participants inclusion and exclusion criteria for the participants characterisation regarding age, sex, dominance, and time of evolution in the stroke group. The body mass (kg) and height (m) were assessed through a seca® 760 scale (seca—Medical Scales and Measuring Systems®, United Kingdom), with a scale of 0,1Kg; and a seca® 222 stadiometer (seca—Medical Scales and Measuring Systems®, United Kingdom), with a 1 mm scale.

The physical activity level was assessed trough the Brief Physical Activity Assessment Tool (Cruz et al., 2021). It consists of a simple and quick (<5 min) questionnaire, to classify the patients as sufficiently/insufficiently active (Marshall et al., 2005). Its classification

TABLE 1. Participant characteristics (sociodemographic and clinical). Data presented from the mean (standard deviation (sd)). *P*-value reflects a comparison between the stroke and healthy groups.

	Stroke ($n = 11$) Mean (sd)	Healthy ($n = 13$) Mean (sd)	Between-groups comparison p Value
Age (years)	51.82 (12.92)	49.92 (14.91)	0.745
Weight (kg)	70.73 (11.15)	77.69 (15.87)	0.235
Height (m)	1.70 (0.13)	1.71 (0.12)	0.769
BMI (kg/m^2)	24.58 (3.38)	26.29 (3.79)	0.261
Post-stroke time (months)	52.18 (29.89)	—	—

categories showed good construct validity ($0.40 \leq \kappa \leq 0.64$; sensitivity = 0.75 95%CI: 0.70– 0.79, specificity = 0.74 95%CI: 0.71–0.77 (Marshall et al., 2005)) in patients with various health conditions, when compared to accelerometery and to other physical activity questionnaires.

The Mini-Mental State Examination scale was used to assess mental status. It is an 11-question measure that tests 5 areas of cognitive function: memory, attention, calculation, language and praxis, with a maximum score of 30 points (Folstein et al., 1975). It is considered that a person has cognitive deficit when the score is: ≤15 for illiterates, ≤22 for people with 1 to 11 years of schooling, or ≤27 for people with more than 11 years of schooling (M. S. Guerreiro, A. P. Botelho, M.; Leitão, O.; Castro-Caldas, A.; Garcia, C., 1994). This instrument has been adapted and validated for the Portuguese population (M. Guerreiro, 1998; M. S. Guerreiro, A. P. Botelho, M.; Leitão, O.; Castro-Caldas, A.; Garcia, C., 1994), with a sensitivity between 63% and 73.4% and a specificity between 90 and 96.8% (M. Guerreiro, 1998).

The Fugl-Meyer Assessment of Sensorimotor Recovery After Stroke was used to assess post-stroke sensorimotor impairment in the adult population in 5 domains: upper limb, lower limb, balance, sensitivity, passive movement and pain (Fugl-Meyer et al., 1975). It is considered that a person has sensorimotor impairment of the lower limb if a score lower than 34 in obtained in the respective subsection of the FMA (Fugl-Meyer et al., 1975; Sanford et al., 1993). The Portuguese version of the scale was adapted and validated for the Portuguese population (Costa, 2003; Santos, 2005) with excellent internal consistency (α Cronbach = 0.99) (Costa, 2003).

Kinematic Data

The joint position of the hip, knee, and ankle in the sagittal plane, on the dominant (DOM) and non-dominant (NDOM) sides in the healthy group and of ipsilesional (IPSI) and contralesional (CONTRA) sides in the stroke group and time of the double support phase were assessed using an optoelectronic system, the Qualisys Motion Capture System (Qualisys AB, Sweden). The spatial position of reflector markers, placed on the participant, was collected using twelve infra-red cameras, eight Oqus 500 and four Miqus M3, connected to the Qualisys USB Analog Acquisition interface, at a sampling frequency of 100 Hz.

Kinetic Data

Ground reaction forces (GRF) and respective torques were used to assess the external mechanical work on the centre of mass (WCOM). The kinetic data was collected using two force platforms (FP4060-08 and FP4060-10 models from Bertec Corporation, United States of America), placed in series near the midpoint of the walkway, and connected to a Bertec amplifier AM 6300, at a sampling frequency of 1000 Hz. The capture hardware was connected to the Qualisys Motion Capture System analogue board.

Electromyographic Data

Surface electromyography (sEMG) was monitored using the wireless Trigno TM acquisition system (Delsys Inc., United States of America) to assess bilaterally the muscle activity of the Tibialis Anterior (TA), Soleus (SOL), Gastrocnemius Medialis (GasM), Rectus Femoris (RF), Vastus Medialis (VM), Biceps Femoris (BF) and Glutaeus Maximus (GMax). Pre-amplified bipolar differential electrodes (Trigno Avanti Sensor model) with a rectangular configuration of two Ag bars in parallel (inter-electrode distance of 10 millimetres) and a gain of 1000 were used to collect the surface electromyography (sEMG) signal, with an acquisition frequency of 1000 Hz. The sEMG signal was integrated into the Qualisys Motion Capture System through an analogue board (National Instruments, United States of America). EMGworks software (Delsys Inc., United States of America) was used to analyse the sEMG signal quality. An Electrode Impedance Checker® (Noraxon, United States of America) was also used to measure the level of skin impedance.

Qualisys Track Manager software (Qualisys AB, Sweden) was used to display and acquire kinematic, kinetic and electromyographic data, which were analysed using Visual 3D software, v6x64 (C-Motion, United States of America). The above-mentioned outcomes were considered to describe the double support phase of gait, of both lower limbs, in the leading limb (LEAD) position (initial contact and loading response) the trailing limb (TRAIL) position (pre-swing) (Kuo et al., 2005).

Procedures

Data collection took place at a biomechanical laboratory, the Rehabilitation Research Centre of Health School of the Polytechnic Institute of Porto, in a controlled environment. To avoid inter-rater error, each researcher was responsible for only one of the following tasks. Prior to data collection, anthropometric measures, body mass and height, were recorded for each participant. Then, the Body Mass Index (BMI), expressed in kg/m2, was calculated, according to the theoretical formula (1):

$$\left(BMI = \frac{\text{weigth}}{heigth^2}\right) \quad (1).$$

For the collection of kinematic data, 46 reflective markers were placed bilaterally in anatomical references (identified by manual palpation): vertex of the head, ear lobules, jugular notch, xiphoid process, spinous apophysis of the 7th cervical vertebra, lateral surface of the acromion, anterior-superior iliac spines, posterior-

superior iliac spines, greater trochanters, lateral and medial epicondyles of the femur, tibial tuberosities, head of the fibula, lateral and medial malleolus, calcaneal tendons (Achilles), 1st, 2nd and 5th metatarsophalangeal joints, lateral and medial epicondyle of the humerus, styloid process of the ulna and radius, 2nd and 5th metacarpophalangeal joints (C-Motion, United States of America) (Cappozzo et al., 1995). To place the markers on the head of the 1st, 2nd, and 5th metatarsals, we carefully explained and demonstrated to each participant the location where he should feel the pressure; then, the markers were placed through palpation of the points on top of their shoes. The1st and 5th metatarsals were easily located on top of the shoes, the 2nd was placed considering the anatomical and biomechanical proximity with the 1st metatarsal. These markers allowed building of a biomechanical full-body model in the Visual 3D software.

To collect the sEMG signal, the electrodes were placed over the muscles belly following the recommendation of the Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000) and the study by *Sousa et al. (2013a)* (Table 2). Electrode placement was confirmed by palpation. Before placing the electrodes, the skin was shaved, exfoliated to remove dead cells from the skin surface, and cleaned with isopropyl alcohol (70%) to remove oil and remaining dead cells. The electrode impedance checker was used to ensure that impedance levels were lower than 5 kΩ (Hermens et al., 2000).

Task

Participants were asked to walk for 10 metres, without technical aids, with their usual footwear, at a self-selected speed, and without explicit instructions ("you can walk this route whenever you want"). Prior to data collection, sufficient time was given till the participants became familiar with the experimental setting (Sousa et al., 2013a; 2013b).

The trials were considered valid if one lower limb had a full contact with the first platform, arranged longitudinally along the route (Sousa et al., 2013a). The participants had to perform 20 valid trials: 10 trials with the CONTRA and NDOM limb of the stroke and healthy group, respectively, in the LEAD position and with the IPSI and DOM limb in the TRAIL position (Figure 1(A)), and 10 trials with the opposite combination (Figure 1(B)) (Fotiadou et al., 2018), to assess the two lower limbs in the two positions, TRAIL and LEAD. A resting time of two-minutes between trials was established to prevent fatigue (Sousa et al., 2013a; 2013b).

Data Processing

Marker trajectories were processed through the Qualisys Track Manager software. Trajectory deviations or interruptions were interpolated using the linear, polynomial, and relational calculations built into the software. Subsequently, the resulting data was exported to the Visual 3D software, in which a full-body biomechanical model was built (according to the appropriate C-motion recommendations). Prior to exporting the data, a 6 Hz low-pass Butterworth filter was used for tracing the markers.

The hip, knee and ankle joint position values in the sagittal plane were recorded during the heel strike of the LEAD limb and toe-off of the TRAIL limb. The heel strike was determined through the maximum horizontal distance between the marker of the ipsilateral calcaneus and the marker of the contralateral lateral malleolus (Banks et al., 2015; French et al., 2020). The toe-off was determined through the minimum horizontal distance between the

TABLE 2. Anatomical references for the electrode placement (Hermens et al., 2000; Sousa et al., 2013a).

Muscle	Anatomical references
Tibialis anterior	Proximal third proximal of the line between the tip of the fibula and the tip of the medial malleolus
Soleus	2 centimetres distal to the lower border of the gastrocnemius medialis muscle belly and 2 centimetres medial to the posterior midline of the leg
Gastrocnemius medialis	Most prominent portion of muscle belly
Rectus femoris	50% on the line between the anterior superior iliac spine and the upper border of the patella
Vastus medialis	4 centimetres above the superior chord of the patella and 3 cm measured medially and oriented 55° from a reference line between the anterior superior iliac spine and the centre of the patella
Biceps femoris	50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia
Glutaeus maximus	50% on the line between the sacrum and the greater trochanter

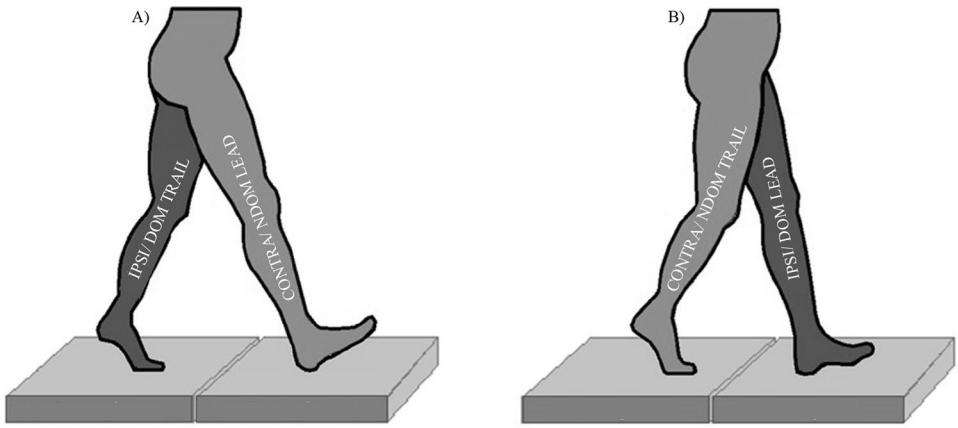

FIGURE 1. A) CONTRA and NDOM limb of the stroke and healthy group, respectively, in the LEAD position; B) IPSI and DOM limb of the stroke and healthy group, respectively, in the LEAD position.

calcaneal marker and the sacral marker (French et al., 2020; Zeni et al., 2008). Exploratory analyses were carried out, in some participants, that validated the similar identification of the events in the double support phase between the adopted kinematic methods and the ground reaction forces data. A positive variation in the range of movement refers to flexion for hip and knee joints and to plantar flexion for ankle joint. The time of the double support phase was calculated through the difference between the time of the heel strike and toe-off events.

The ground reaction forces signal was used to calculate the WCOM, of each lower limb during the double support phase assuming that the external mechanical power by a limb is equal to the dot product of the external force ($\vec{F}$) acting on the limb and the velocity of the centre of mass ($\vec{v}_{com}$) (Donelan et al., 2002b). Accordingly, to calculate the external work done on the centre of mass, the velocity of the centre of mass (in the F_y component) was first calculated through the first derivative of centre of mass displacement. Then, the mechanical energy of the centre of mass for each limb (LEAD and TRAIL) was calculated by multiplying the respective ground reaction force (normalised by the mass of the participant) by the velocity of the centre of mass for F_y direction. Finally, the external work done on the centre of mass for each limb was computed by calculating the mechanical energy integral of each limb in the double support phase, according to the equations (2) and (3).

$$P_{trail} = \vec{F}_{trail} \cdot \vec{v}_{com} = F_{y,trail} v_{y,com} \quad (2)$$

$$P_{lead} = \vec{F}_{lead} \cdot \vec{v}_{com} = F_{y,lead} v_{y,com} \quad (3)$$

The electromyographic data processing was carried out using Matlab software, version 3.9.0. (MathWorks, United States of America). A second-order digital band pass Butterworth with a cut of frequency between 20 Hz and 450 Hz and the root mean square (RMS) value was calculated using a moving average window of 100 samples (Lamontagne et al., 2000). The mean of RMS during double support was normalised by the maximum RMS value obtained during the gait cycle (Sousa et al., 2013a; Sousa & Tavares, 2012). The coactivation ratio was calculated according to the following equation (4) (Kellis et al., 2003):

$$Coactivation\ ratio = \frac{\text{agonist activity}}{\text{agonist activity} + \text{antagonist activity}} \quad (4)$$

This coactivation ratio reflects a relative measure of agonist activity in a specific limb and in a specific position (LEAD or TRAIL) of double support phase (Kellis et al., 2003). The RF, VM and TA muscle were considered the agonists for the LEAD position while the GMax, BF, GasM and SOL were considered the agonists in the TRAIL position. The coactivation ratio can vary between 0 and 1, with 0 indicating no agonist activity and 1 indicating no antagonist activity. In turn, a coactivation ratio of 0.5 indicates an agonist and antagonist coactivation in which their percentage of activation intensity were equal.

Statistical Analysis

IBM Statistical Package for the Social Science® software version 28.0 (IBM Corporation, Armonk NY, United States of America) was used for descriptive and inferential data analysis, with significance set at $p < 0.05$.

To ensure that there were no significant differences between groups (stroke vs. healthy) regarding age, mass, height, and BMI, the t-test for 2 independent samples was used. The assumption of normality was guaranteed

using the Shapiro-Wilk test. For the comparison between groups regarding gender, Fisher test was used.

Mean and standard deviation were used as descriptive statistics for quantitative variables. In the variables in which the assumption of normality is not verified, the median and the 25th and 75th percentiles presented a behaviour similar to the mean and standard deviation, therefore the former were not presented. For the qualitative variables of sample characterisation (gender and injured/dominant side), the absolute and relative frequency was used (Marôco, 2018).

To compare the groups (stroke vs. healthy) in the quantitative variables the t-test for 2 independent samples was used, or the Mann-Whitney test, upon the response of the Shapiro-Wilk normality test. Regarding the kinematic, kinetic and electromyographic variables, the mean and coefficient of variation of the 10 repetitions performed by each participant were considered for the analysis.

Results

The characteristics of the participants are summarised on Table 1. No significative values were found, thereby both groups were comparable. Specifically, the double support phase time, in LEAD position, also did not present significative differences between groups, ensuring the same comparison (Table 4)

Joint Position

Through the analysis of Table 3 differences between groups were observed in lower limb joints position in both TRAIL and LEAD positions. The stroke group presented lower knee flexion ($p < 0{,}001$) and ankle plantar flexion ($p < 0{,}001$) in the CONTRA limb in TRAIL position and higher knee flexion of both, IPSI ($p < 0{,}001$) and CONTRA ($p < 0{,}001$) limbs in LEAD position. When analysing the coefficient of variation of the joint positions, a surrogate for variability, higher variability was found in the knee and ankle of both limbs, CONTRA ($p = 0{,}011$; $p = 0{,}001$ respectively) and IPSI ($p = 0{,}007$; $p = 0{,}007$ respectively), in TRAIL position. In LEAD position, the coefficient of variation was higher in the stroke group in the CONTRA and IPSI hip ($p = 0{,}019$; $p = 0{,}041$ respectively) and ankle joint positions ($p = 0{,}009$; $p < 0{,}001$ respectively), but a lower coefficient of variation was observed in knee joint position in the CONTRA limb ($p = 0{,}035$). In the remaining joint positions, no significant differences were observed between the groups ($p > 0.05$).

Double Support Time and External Mechanical Work on the Centre of Mass

Table 4 presents the values of WCOM and double support time by group, limb, and position. The results demonstrate that stroke group presented a lower WCOM when CONTRA limb was both in LEAD ($p = 0{,}010$) and TRAIL positions ($p < 0{,}001$), and when IPSI limb was in LEAD position ($p = 0{,}023$). Higher coefficient of variation was found on stroke group in double support time (CONTRA $p = 0{,}001$; IPSI $p = 0{,}012$). The same result was found in WCOM in TRAIL position on both limbs (CONTRA $p = 0{,}013$; IPSI $p = 0{,}012$) and in the WCOM when IPSI limb is in LEAD position ($p = 0{,}038$).

TABLE 3. Mean and coefficient of variation values of the hip, knee, and ankle joints position.

		Stroke CONTRA	Healthy NDOM	*p* Value	Stroke IPSI	Healthy DOM	*p* Value
TRAIL							
Hip flexion (°)	Mean	3.23 (6.45)	3.02 (8.64)	0.949	7.14 (6.42)	2.80 (9.17)	0.201
	CV	0.88 (2.91)	0.46 (1.46)	0.654	0.60 (1.57)	0.23 (0.62)	0.447
Knee flexion (°)	Mean	31.57 (9.21)	46.29 (4.53)	**<0.001**	45.92 (6.55)	45.67 (5.87)	0.922
	CV	0.19 (0.16)	0.04 (0.01)	**0.011**	0.11 (0.06)	0.04 (0.02)	**0.007**
Ankle plantar flexion (°)	Mean	16.44 (9.63)	30.18 (6.04)	**<0.001**	25.47 (8.57)	29.89 (6.85)	0.174
	CV	0.45 (0.65)	0.07 (0.03)	**0.001**[a]	0.18 (0.10)	0.08 (0.04)	**0.007**
LEAD							
Hip flexion (°)	Mean	31.15 (6.88)	32.15 (8.32)	0.756	36.18 (7.85)	31.71 (8.20)	0.188
	CV	0.10 (0.07)	0.04 (0.01)	**0.019**	0.08 (0.08)	0.04 (0.02)	**0.041**[a]
Knee flexion (°)	Mean	11.05 (4.31)	0.00 (5.56)	**<0.001**	16.50 (10.13)	−0.19 (5.08)	**<0.001**
	CV	0.36 (0.39)	−2.63 (6.06)	**0.035**[a]	0.08 (0.42)	0.11 (0.55)	0.890
Ankle plantar flexion (°)	Mean	17.73 (7.73)	14.83 (2.90)	0.262	12.40 (5.64)	14.56 (3.39)	0.259
	CV	0.21 (0.24)	0.09 (0.09)	**0.009**[a]	0.51 (0.78)	0.07 (0.02)	**<0.001**[a]

Data presented by the mean (standard deviation) of the contralesional and ipsilesional lower limb in the stroke group and dominant and non-dominant lower limb in the healthy group, both in the TRAIL and LEAD positions.
[a]Nonparametric tests; Significant differences in **bold**. CV: coefficient of variation.

TABLE 4. Mean and coefficient of variation of double support time and centre of mass external work.

		Stroke CONTRA	Healthy NDOM	*p* Value	Stroke IPSI	Healthy DOM	*p* Value
Double support phase time (LEAD)							
	Mean	0.29 (0.31)	0.20 (0.03)	0.313	0.24 (0.10)	0.19 (0.03)	0.162
	CV	0.19 (0.09)	0.06 (0.03)	**0.001**	0.16 (0.11)	0.05 (0.02)	**0.012**
WCOM							
TRAIL							
	Mean	0.10 (0.03)	0.21 (0.02)	**<0.001**	0.15 (0.09)	0.21 (0.02)	0.131
	CV	0.27 (0.15)	0.07 (0.03)	**0.013**	0.19 (0.08)	0.08 (0.03)	**0.012**
LEAD							
	Mean	−0.11 (0.06)	−0.17 (0.04)	**0.010**	−0.12 (0.05)	−0.16 (0.03)	**0.023**
	CV	−0.53 (0.64)	−0.11 (0.03)	0.130	−0.25 (0.15)	−0.10 (0.04)	**0.038**

Data presented by mean (standard deviation) of the contralesional and ipsilesional lower limb of the stroke group and dominant and non-dominant lower limb of the healthy group, in the TRAIL and LEAD positions.
Significant differences in **bold**. CV: coefficient of variation.

TABLE 5. Mean electromyographic activity of tibialis anterior (TA), Soleus (SOL), Gastrocnemius Medialis (GasM), Rectus Femoris (RF), Biceps Femoris (BF), Vastus Medialis (VM) and Glutaeus Maximus (GMax) muscles and the coactivation ratio of the hip, knee, and ankle joints.

	Stroke CONTRA	Healthy NDOM	*p* Value	Stroke IPSI	Healthy DOM	*p* Value
TRAIL						
Ankle						
TA	21.64 (5.52)	13.20 (4.38)	**0.001**	8.80 (4.23)	8.57 (3.27)	0.886
SOL	14.25 (4.80)	11.65 (5.19)	0.249	18.85 (11.10)	7.40 (4.27)	**0.015**
GasM	11.98 (4.89)	7.47 (4.02)	**0.028**	8.64 (6.98)	4.87 (3.58)	0.165
Ankle ratio	0.55 (0.05)	0.55 (0.14)	0.943	0.72 (0.19)	0.52 (0.17)	**0.017**
Knee and Hip						
RF	13.81 (3.74)	17.58 (5.33)	0.083	15.27 (5.54)	13.14 (4.41)	0.326
VM	10.29 (3.09)	17.33 (6.66)	**0.004**	11.04 (4.93)	11.56 (4.46)	0.796
BF	14.77 (6.65)	12.93 (5.66)	0.493	12.72 (7.19)	12.07 (4.14)	0.811
GMax	17.19 (6.22)	19.28 (6.29)	0.449	11.47 (2.55)	14.47 (4.80)	0.073
Hip ratio	0.31 (0.11)	0.36 (0.08)	0.157	0.38 (0.11)	0.32 (0.07)	0.145
Knee ratio	0.62 (0.11)	0.73 (0.08)	**0.015**	0.67 (0.13)	0.65 (0.10)	0.666
TA	24.32 (6.63)	12.36 (4.87)	**<0.001**	30.35 (5.40)	11.28 (3.32)	**<0.001**
SOL	19.45 (5.34)	14.35 (7.02)	0.081	18.23 (6.28)	8.10 (4.46)	**<0.001**
GasM	19.92 (8.07)	13.24 (11.25)	0.143	13.81 (11.17)	5.50 (4.56)	0.061
Ankle ratio	0.40 (0.07)	0.42 (0.20)	0.750	0.50 (0.11)	0.52 (0.15)	0.843
Knee and Hip						
RF	30.91 (6.50)	16.46 (4.83)	**<0.001**	30.78 (5.86)	14.84 (6.18)	**<0.001**
VM	28.95 (7.38)	18.84 (7.67)	**0.006**	31.59 (8.38)	13.44 (5.39)	**<0.001**
BF	24.54 (6.84)	13.38 (6.74)	**0.001**	23.98 (9.56)	12.10 (7.11)	**0.003**
GMax	31.22 (12.26)	17.79 (5.95)	**0.012**	35.95 (7.12)	16.59 (4.16)	**0.000**
Hip ratio	0.63 (0.07)	0.63 (0.12)	0.877	0.66 (0.06)	0.68 (0.09)	0.539
Knee ratio	0.71 (0.06)	0.73 (0.10)	0.566	0.72 (0.06)	0.71 (0.11)	0.661

Data presented by the mean (standard deviation) of the contralesional and ipsilesional lower limb of the stroke group and dominant and non-dominant lower limb of the healthy group, in the TRAIL and LEAD positions.
CV: coefficient of variation.

Electromyographic Activity

The results presented in Table 5 demonstrate that stroke group showed a higher percentage of activity against the maximum value of the gait cycle, compared to the healthy group. Specifically, in the TRAIL position, the stroke group presented higher percentage of TA ($p = 0{,}001$) and GAS ($p = 0{,}028$) muscle activity of CONTRA limb and higher percentage of SOL activity ($p = 0{,}015$) and ankle ratio ($p = 0{,}017$) of the IPSI limb, but lower percentage of VM activity ($p = 0{,}004$) and knee ratio ($p = 0{,}015$) of CONTRA limb. In the LEAD position, in both limbs, the stroke group presented higher percentage of TA (CONTRA $p < 0{,}001$; IPSI $p < 0{,}001$), RF (CONTRA $p < 0{,}001$; IPSI $p < 0{,}001$), VM (CONTRA $p = 0{,}006$; IPSI $p < 0{,}001$), BF (CONTRA $p = 0{,}001$;; IPSI $p = 0{,}003$) and GMax (CONTRA $p = 0{,}012$; IPSI $p = 0{,}000$) activity and higher percentage of SOL activity also in the IPSI limb ($p < 0{,}001$).

When analysing the coefficient of variation of the electromyographic variables expressed in Table 6, a distinct pattern was observed compared to the one observed for kinematic variables, since, globally, the coefficient of variation was lower in the stroke group than in the healthy group. In the TRAIL position, significant differences were observed in SOL (CONTRA $p = 0{,}002$; IPSI $p = 0{,}003$), GAS (CONTRA $p = 0{,}020$; IPSI $p = 0{,}005$), and ankle ratio (CONTRA $p = 0{,}014$; IPSI $p = 0{,}003$) for both limbs and in TA of CONTRA limb ($p = 0{,}015$). In the LEAD position the differences were observed in TA (CONTRA $p = 0{,}002$; IPSI $p = 0{,}001$), SOL (CONTRA $p = 0{,}030$; IPSI $p = 0{,}049$), RF (CONTRA $p = 0{,}012$; IPSI $p = 0{,}011$) and BF (CONTRA $p = 0{,}003$; IPSI $p = 0{,}008$) and hip ratio (CONTRA $p = 0{,}037$; IPSI $p = 0{,}041$) of both limbs, in GasM ($p = 0{,}002$) of the CONTRA and in ankle ($p = 0{,}027$) and knee ($p = 0{,}031$) ratio of the IPSI limb. The remaining muscles did not present significantly different values in the mean and in the coefficient of variation between the groups ($p > 0.05$).

Discussion

This study aimed to identify possible differences between participants with and without stroke in respect to ipsilesional and contralesional lower limbs kinematics, kinetics, and muscle activity during double support phase of gait in terms of mean and coefficient of variation. In human movement the variability is a natural, often beneficial, feature that can be optimised with skill development and healthy physiologic systems (Stergiou et al., 2006). The extent to which gait variability might be considered beneficial or detrimental depends on the theoretical perspective adopted to explain the underlying control of movement (Cavanaugh & Stergiou, 2020). The amount of gait variability typically increases with pathology (Rosano et al., 2007). The study of movement variability is relevant to add a new point of analysis and observe changes in both lower limbs, during the double support phase. Movement variability is quite stable when repeatedly performing a well-known automatic task (Carson et al., 2014; Terrier et al., 2005), such as walking at a comfortable self-selected speed (Masani et al., 2002; Reisman et al., 2009; Sekiya et al., 1997; Sousa & Tavares, 2012). Results on the present study demonstrate differences between groups in the means of kinematic, kinetic and electromyographic parameters, but differences in variability between CONTRA/NDOM and IPSI/DOM limbs in both positions (TRAIL and LEAD) are even more notorious.

Regarding the mean time of double support phase, similar results were found between both stroke and healthy groups. This may raise the hypothesis that differences found in the lower limbs in CONTRA/NDOM and IPSI/DOM, in the other biomechanical parameters monitored may not be related with the differences in gait speed, commonly found between post-stroke and health participants (Lamontagne & Fung, 2004; Liu et al., 2014; Tomida et al., 2022; Tyrell et al., 2011). The similarity between groups regarding sociodemographic characteristics such as age, weight, height probably contributed to the non-existence of significant differences in double support time. However, significant differences were observed in double support time variability which is in accordance with a previous study (Woolley, 2001). This increased variability has been suggested to be the result of interlimb coordination impairments (Balasubramanian et al., 2009; Wang et al., 2020; Zukowski et al., 2019). The results of the present study reinforce that the temporal variability may be considered a robust indicator of gait, as well as a target for neuromotor intervention of individuals with stroke sequelae (Patel et al., 2022).

Kinematic results demonstrate that post-stroke group presented decreased knee flexion and ankle plantar flexion in CONTRA limb during toe-off (trailing limb) corroborating previous studies (Chen et al., 2005; Goldberg et al., 2006; Matsuda et al., 2016; Perry & Burnfield, 2010). The lower CONTRA TRAIL knee flexion and ankle plantar flexion, corresponding to the pre-swing phase, may explain the decreased WCOM (positive work on COM) observed in same limb in the present study which is also in accordance with previous studies (Beyaert et al., 2015; Bowden et al., 2006; Kim & Eng, 2003; Morita et al., 2020; Olney & Richards, 1996; Sousa et al., 2013b). Globally, high variability was observed in kinematics and WCOM in the TRAIL limb being in accordance with increased variability in double support time. Only CONTRA and IPSI hip joint did not presented variability differences. This can probably be explained by the need to seek proximal stability to compensate the distal instability produced in the limb transition, from a closed to an open kinetic chain, that occurs in the pre-swing (Fotiadou et al., 2018). The control of

TABLE 6. Electromyographic activity coefficient of variation of tibialis anterior (TA), Soleus (SOL), Gastrocnemius Medialis (GasM), Rectus Femoris (RF), Biceps Femoris (BF), Vastus Medialis (VM) and Glutaeus Maximus (GMax) muscles and the coactivation ratio of hip, knee, and ankle joints.

	Stroke CONTRA	Healthy NDOM	*p* Value	Stroke IPSI	Healthy DOM	*p* Value
TRAIL						
Ankle						
TA	0.28 (0.11)	0.55 (0.34)	**0.015**	0.49 (0.30)	0.51 (0.24)	0.888
SOL	0.39 (0.17)	0.81 (0.35)	**0.002**	0.36 (0.15)	0.65 (0.22)	**0.003**
GasM	0.49 (0.31)	0.85 (0.34)	**0.020**	0.43 (0.16)	0.95 (0.47)	**0.005**
Ankle ratio	0.20 (0.09)	0.37 (0.17)	**0.014**	0.17 (0.12)	0.38 (0.16)	**0.003**
Knee and Hip						
RF	0.45 (0.29)	0.45 (0.25)	0.999	0.41 (0.20)	0.57 (0.22)	0.102
VM	0.49 (0.22)	0.54 (0.39)	0.694	0.54 (0.21)	0.53 (0.20)	0.890
BF	0.39 (0.17)	0.58 (0.27)	0.074	0.39 (0.21)	0.57 (0.37)	0.189
GMax	0.36 (0.14)	0.42 (0.36)	0.616	0.48 (0.46)	0.38 (0.18)	0.450
Hip ratio	0.30 (0.12)	0.35 (0.19)	0.577	0.28 (0.13)	0.40 (0.17)	0.084
Knee ratio	0.18 (0.06)	0.15 (0.08)	0.391	0.19 (0.11)	0.22 (0.10)	0.604
LEAD						
Ankle						
TA	0.29 (0.12)	0.56 (0.23)	**0.002**	0.28 (0.10)	0.50 (0.16)	**0.001**
SOL	0.39 (0.16)	0.71 (0.48)	**0.030**[a]	0.39 (0.12)	0.56 (0.23)	**0.049**
GasM	0.36 (0.18)	1.08 (0.66)	**0.002**	0.50 (0.19)	0.63 (0.32)	0.283
Ankle ratio	0.29 (0.17)	0.49 (0.25)	0.054	0.23 (0.10)	0.38 (0.19)	**0.027**
Knee and Hip						
RF	0.26 (0.10)	0.47 (0.24)	**0.012**	0.28 (0.09)	0.52 (0.28)	**0.011**
VM	0.36 (0.11)	0.47 (0.28)	0.219	0.34 (0.16)	0.48 (0.16)	0.057
BF	0.29 (0.12)	0.72 (0.42)	**0.003**	0.28 (0.09)	0.59 (0.34)	**0.008**
GMax	0.33 (0.19)	0.52 (0.36)	0.169	0.26 (0.11)	0.33 (0.12)	0.223
Hip ratio	0.11 (0.07)	0.21 (0.12)	**0.037**	0.11 (0.02)	0.15 (0.05)	**0.041**
Knee ratio	0.09 (0.04)	0.19 (0.12)	0.071[a]	0.10 (0.04)	0.18 (0.10)	**0.031**

Data presented by the mean (standard deviation) of the contralesional and ipsilesional lower limb of the stroke group and dominant and non-dominant lower limb of the healthy group, in the TRAIL and LEAD positions.

[a]Nonparametric tests; Significant differences in **bold**. CV: coefficient of variation.

knee flexion and ankle plantar flexion during the pre-swing phase is mainly obtained by GasM muscle (Balasubramanian et al., 2007; Ellis et al., 2013; Richards et al., 1998; Sousa et al., 2013b) (Mazuquin et al., 2014; Olney & Richards, 1996) while the control of the vertical and anterior propulsion of the COM are dependent from the activity of both SOL and GasM muscles (Donelan et al., 2002a; Kuo, 2002; Kuo et al., 2005; Lamontagne et al., 2002). In the present study a higher percentage of activity in the distal muscles in the CONTRA (TA and GasM) and in the IPSI (SOL and ankle ratio) was observed. However, in proximal muscles, only CONTRA VM and knee ratio showed a lower percentage of activity. Only when differences in SOL and VM activities are noted, differences in ankle and knee ratio are also verified, respectively. Thus, we can hypothesise that SOL and VM activity may influence the ratio. These results contradict previous studies that demonstrate, during double support phase, different results with different normalisation procedures, namely mean dynamic method, showing a decreased muscle activity (Sousa et al., 2013b) or a decreased coactivation ratio (Silva et al., 2015); and a higher coactivation ratio, using the absolute EMG value (Lamontagne et al., 2000). The normalisation process of muscle activity, used in the present study, can explain the increased relative muscle activity observed as the mean of activity obtained in the double support phase, normalised by the maximum value obtained during gait cycle. Lower level of maximum activity will lead to higher percentage of activity in post-stroke group. Despite the feasibility of the mentioned methods, the use of peak dynamic seemed to be relevant in this study, allowing the normalisation data from neurological patients (Yang & Winter, 1984), reducing inter-subject variability, and showing itself helpful for clinical populations that are unable to attempt

maximal efforts. Some authors have already identified an atypical behaviour in distal muscle activity, in double support phase (Silva et al., 2015; Sousa et al., 2013b). Concerning TRAIL, a distal lower variability was observed in both post-stroke limbs, in accordance to a previous study in hemiplegic gait (Di Nardo et al., 2020). This reduced variability may be related with a limited muscle recruitment ability to challenges during walking task (Di Nardo et al., 2020).

Studies that evaluated the WCOM of LEAD and TRAIL showed that the TRAIL needs to generate more work than the LEAD, to compensate the energy expenditure that occurs during the collision, at initial contact phase and response to load (Kuo, 2002; Kuo et al., 2005; Sousa et al., 2013b). Our results demonstrate that when CONTRA limb is in TRAIL position, WCOM of the TRAIL is lower than the LEAD. It seems to be clear that the positive WCOM TRAIL of the CONTRA cannot compensate the energy expended by the negative WCOM LEAD of the IPSI, in agreement with Balasubramanian et al. (2007) study. There are other authors who explain this result associated with a lower capacity of the CONTRA limb to adjust to the IPSI in the LEAD position (Kim & Eng, 2004; Olney & Richards, 1996; Sousa et al., 2013b).

Kinematic results for the LEAD limb demonstrate that post-stroke group presented increased bilateral knee flexion during heel-strike which has also been reported by previous studies (Levine et al., 2012; Moore et al., 1993; Nadeau et al., 2013). This finding can explain the lower WCOM of the CONTRA and IPSI limbs, in LEAD (negative work on COM) since knee flexion has a relevant role absorption of ground reaction forces (Mazuquin et al., 2014). Like it was observed for the TRAIL position significantly higher variability was also observed in overall joint positions and WCOM. This high variability may be related to the precision of foot placement on the ground. Following this assumption, Donelan and Pearson (2004) stated that foot placement in the heel-strike implies adjustments that translate into a higher metabolic cost of walking. As mentioned about trailing position, in leading post-stroke participants presented increased percentage of muscle activity in both lower limbs in one distal (TA) and some proximal (RF, VM, BF and GMax) muscles. The IPSI LEAD SOL was the only muscle that presented differences just in IPSI. As already described, EMG data should be seen with caution and regarding the normalisation method assumed. A lower variability was observed in almost all muscles analysed, in both limbs, as mentioned in a previous study (Di Nardo et al., 2020). A reduced coefficient of variation may be, as aforementioned discuss in TRAIL, a representative of stereotypical pattern of muscular recruitment by neuromotor system in pathophysiology (Di Nardo et al., 2020; Lord et al., 2011), which translate a lower variation and adjustment of recruited muscle activity in gait motor performance.

Our results should consider some limitations, mainly related with the lack of sample characterisation according to the recommendations of the Stroke Recovery and Rehabilitation Roundtable (Kwakkel et al., 2017) although, due to the purpose of our study, we think these characteristics would not change our outcomes and results. It should also be considered that in this study, only angular positions were considered for the analysis in the events of the double support phase, since we consider that, through the joint position values in each position of the double support phase, LEAD or TRAIL, it is possible to understand the amount of angular displacements, observing the comparison between stroke and healthy limbs. For example, analysing knee and ankle joint position in CONTRA limb, in LEAD and in TRAIL, compared to the healthy limbs under the same positions, we observe that there is less movement in the stroke limbs. We can consider that the smallest difference in the position in the events may reflect a smaller joint displacement. However, future studies should consider the analysis of this variable.

To our knowledge, this is the first study to quantify variability in biomechanical variables expressing interlimb coordination in double support phase. This study findings present a relevant insight into post-stroke neuromotor impairment since they point out bilateral post-stroke chronic impairments, including motor variability, expressing interlimb coordination in double support phase. It would be interesting for future studies to explore biomechanical assessment, involving variability quantification, to better understand motor behaviour, also in acute and subacute phases.

Conclusions

The present study demonstrate that post-stroke participants presented bilateral decreased mechanical work on the centre of mass and increased variability combined with decreased contralesional knee and ankle flexion in trailing position and increased ipsilesional knee flexion in LEAD and an overall increased variability. In general, increased relative muscle activity was observed in post-stroke participants with decreased variability.

Globally, joint position, mechanical work on the centre of mass and percentage of muscular activity were relevant parameters to identify differences, as well as to quantify variability between post-stroke and healthy participants.

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