

PROGRAMA DOUTORAL EM SEGURANÇA E SAÚDE OCUPACIONAIS

Coordenação entre Membros na Marcha - Avaliação, Aplicações e Implicações Clínicas em Pessoas com Sequelas de Acidente Vascular Cerebral

Ana Sofia Garcez Barbosa do Couto

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Coordenação entre Membros na Marcha

Avaliação, Aplicações e Implicações Clínicas

em Pessoas com Sequelas de Acidente Vascular Cerebral

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Tese elaborada sob a orientação de:

Professora Doutora Andreia S. P. Sousa (Orientador), Área Técnico-Científica da
Fisioterapia, Escola Superior de Saúde do Instituto Politécnico do Porto

Professor Doutor Mário A. P. Vaz (Coorientador), Departamento de Engenharia
Mecânica, Faculdade de Engenharia da Universidade do Porto

Ana Sofia Garcez Barbosa do Couto

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PALAVRAS-CHAVE: ACIDENTE VASCULAR CEREBRAL; CONTRALESIONAL; IPSILESIONAL; MEMBROS INFERIORES; COORDENAÇÃO ENTRE MEMBROS; FASE DE DUPLO APOIO; BIOMECÂNICA; VARIABILIDADE

"O que é mais importante: a viagem ou o destino?"

"A companhia."

James Norbury

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LISTA DE ESTUDOS ORIGINAIS

A presente Tese é baseada nos seguintes estudos originais, referenciados em numeração romana:

Artigo I.

Couto, A.G.B., Vaz, M.A.P., Pinho, L., Félix, J., Silva, S., Silva, A., & Sousa, A.S.P. (2022). Methodological Considerations in Assessing Interlimb Coordination on Poststroke Gait: A Scoping Review of Biomechanical Approaches and Outcomes. *Sensors*, 22, 2010. <https://doi.org/10.3390/s22052010>

Artigo II.

Couto, A.G.B., Vaz, M.A.P., Pinho, L., Félix, J., Moreira, J., Pinho, F., Mesquita, I. A., Mesquita Montes, A., Crasto, C., & Sousa, A.S.P. (2023). Repeatability and Temporal Consistency of Lower Limbs Biomechanical Variables Expressing Interlimb Coordination during Double-Support Phase in People with and without Stroke Sequelae. *Sensors*, 23, 2526. <https://doi.org/10.3390/s23052526>

Artigo III.

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RESUMO

Introdução: O comprometimento motor dos membros inferiores após a ocorrência de um Acidente Vascular Cerebral tem um impacto significativo na capacidade para a marcha. A evidência do comprometimento bilateral, contralesional e ipsilesional, nesta condição patológica, tem vindo a crescer, emergindo a necessidade de desenvolver estudos com recurso a medidas objetivas. O reflexo deste impacto bilateral poderá ser observado na coordenação entre os membros, expressa durante a fase de duplo apoio.

Objetivo: Caracterizar a coordenação entre os membros inferiores na marcha, especificamente na fase de duplo apoio, em pessoas com e sem sequelas de Acidente Vascular Cerebral, através da análise biomecânica multi-instrumental (cinemática, cinética e eletromiográfica).

Métodos: Para dar resposta ao objetivo apresentado foram realizados 3 estudos originais. No estudo I foi realizada uma *Scoping Review*, cumprindo os pressupostos da metodologia *PRISMA-ScR*, e a estratégia População, Conceito e Contexto, com o objetivo de identificar e sumariar a metodologia de avaliação biomecânica da coordenação entre membros, na marcha, após Acidente Vascular Cerebral. O estudo II caracterizou-se por um Teste-Retestes com o objetivo de determinar o número mínimo de repetições necessário para obter repetibilidade e consistência temporal dos parâmetros biomecânicos dos membros inferiores na fase de duplo apoio em pessoas com e sem sequelas de Acidente Vascular Cerebral, e o estudo III foi do tipo Analítico Transversal com o objetivo de identificar diferenças nas variáveis cinemáticas, cinética e de atividade muscular dos membros inferiores, ipsilesional e contralesional, durante a fase de duplo apoio da marcha em pessoas com sequelas de Acidente Vascular Cerebral. Nos estudos II e III, onze participantes com sequelas de Acidente Vascular Cerebral e treze saudáveis realizaram 10 ensaios de marcha a uma velocidade auto-selecionada, em dois momentos distintos, com intervalo de 72h a 7 dias. Para a recolha das variáveis, recorreu-se a um sistema optoeletrónico de captura de movimento, a duas plataformas de forças e um sistema de eletromiografia de superfície. Os resultados obtidos foram avaliados durante

a fase de duplo apoio da marcha tendo sido utilizado para análise o tempo de duplo apoio e a posição articular, o trabalho mecânico externo sobre o centro de massa e a percentagem relativa da atividade eletromiográfica. O membro dominante/ não dominante nos participantes saudáveis, e o membro contralesional/ ipsilesional nos participantes pós-Acidente Vascular Cerebral foram avaliados nas posições *leading* e *trailing*.

Resultados: No estudo I, sete artigos sumariaram as principais considerações metodológicas na avaliação da coordenação entre os membros inferiores, durante a marcha, em pessoas com sequelas de Acidente Vascular Cerebral. No estudo II, a consistência temporal demonstrada pelo número mínimo de repetições necessário para atingir uma boa estabilidade de desempenho ($ICC > 0,75$), na maioria das variáveis cinemáticas e cinéticas estudadas em cada sessão, foram necessárias 2 a 3 tentativas para ambos os grupos, membros e posições, enquanto as variáveis eletromiográficas apresentaram maior número de repetições. As diferenças entre grupos para as variáveis biomecânicas analisadas foram observadas em ambas as posições. No estudo III, nas pessoas pós-Acidente Vascular Cerebral observou-se redução da flexão do joelho e flexão plantar do tornozelo contralesional na posição *trailing* e aumento bilateral da flexão do joelho em posição *leading* e aumento da variabilidade; um menor trabalho mecânico externo no centro de massa com uma maior variabilidade bilateral nas duas posições e uma maior percentagem relativa de atividade muscular bilateral com uma menor variabilidade distal, bilateralmente.

Conclusão: O estudo I permitiu sumariar as condições da marcha, os instrumentos e as variáveis para a avaliação da coordenação entre membros, em pessoas com sequelas de Acidente Vascular Cerebral. O estudo II, permitiu concluir que a estabilidade de desempenho das variáveis cinemáticas e cinéticas mostraram uma boa repetibilidade e consistência temporal com um baixo número de repetições, enquanto as variáveis eletromiográficas necessitaram de um maior número de repetições para atingir essa estabilidade. No último estudo, foi identificado que as diferenças cinemáticas, cinéticas e eletromiográficas não se encontram limitadas ao membro contralesional, observando-se também diferenças no ipsilesional. Neste sentido, sugere-se precaução no uso do membro ipsilesional como controlo do contralesional. As diferenças na variabilidade no

movimento foram observadas em ambos os membros, apontando esta variável como um potencial fator relevante na comparação entre pessoas com e sem sequelas de Acidente Vascular Cerebral.

PALAVRAS-CHAVE: ACIDENTE VASCULAR CEREBRAL; CONTRALESIONAL; IPSILESIONAL; MEMBROS INFERIORES; COORDENAÇÃO ENTRE MEMBROS; FASE DE DUPLO APOIO; BIOMECÂNICA; VARIABILIDADE

ABSTRACT

Introduction: After stroke, lower limb motor impairment has a significant impact on walking ability. In this pathological condition, evidence on bilateral involvement, both contralesional and ipsilesional, has been growing, resulting in the needing to develop studies using objective measurements. This bilateral impact can be observed in the coordination between lower limbs, expressed during the double support phase.

Aim: To characterize the coordination between lower limbs in gait, specifically in the double support phase, in people with and without stroke sequelae, through multi-instrumental biomechanical analysis (kinematics, kinetics and electromyography).

Methods: To answer this study purposal, 3 original studies were developed. In study I, a Scoping Review was carried out, fulfilling the assumptions of the PRISMA-ScR methodology, and the Population, Concept and Context strategy, with the aim to identify and summarize the methodology for biomechanical assessment of interlimb coordination, in gait, after a stroke. Study II was characterized by a Test-Retest in order to determine the minimum number of repetitions necessary to obtain repeatability and temporal consistency on biomechanical parameters of lower limbs, in the double support phase, in people with and without sequelae of stroke. Study III was a Cross-sectional Analytical type one, with the aim to identify differences in the kinematic, kinetics and muscle activity variables in ipsilesional and contralesional lower limbs, during double support phase, in gait, in people with stroke sequelae. In studies II and III, eleven participants with stroke sequelae and thirteen healthy performed 10 gait trials at a self-selected speed, at two different times, with an interval of 72 hours to 7 days. To collect the variables, an optoelectronic motion capture system, two force plats and a surface electromyography system were used. The results obtained were evaluated during the double support phase of gait, using the time of double support and the joint position, the external mechanical work on the center of mass and the relative percentage of electromyographic activity analysis. The dominant/ non-dominant limb in healthy

participants, and the contralesional/ ipsilesional limb in post-stroke were evaluated in leading and trailing positions.

Results: In study I, seven literature units summarized the main methodological considerations in the evaluation of interlimb coordination in lower limbs, during gait, in people with stroke sequelae. In study II, most of kinematic and kinetic variables studied in each session, showed temporal consistency, regarding the minimum number of repetitions necessary, considering 2 to 3 attempts to achieve good performance stability (ICC >0.75), for both groups, limbs, and positions. The electromyographic variables demonstrated a higher number of repetitions. Differences between groups, for the analyzed biomechanical variables, were observed in both positions. In study III, in people after stroke, a reduction in knee flexion and plantar flexion of the contralesional ankle in the trailing position and a bilateral increase in knee flexion in the leading position and increased variability were observed, less external mechanical work on the center of mass with greater bilateral variability in both positions and a greater bilateral relative percentage of muscle activity with less distal variability bilaterally.

Conclusions: Study I allowed to summarize the gait conditions, instruments, and variables to access interlimb coordination, in people with stroke sequelae. Study II concluded that the performance stability of the kinematic and kinetic variables showed good repeatability and temporal consistency with a low number of repetitions, while the electromyographic variables required a greater number of repetitions. In the last study, it was identified that the kinematic, kinetic and electromyographic differences are not limited to the contralesional limb. Differences were also observed in the ipsilesional limb. In this sense, caution is suggested in the use of the ipsilesional limb as a control of the contralesional one. Differences in variability in movement were observed in both limbs, pointing to this variable as a potentially relevant factor in the comparison between people with and without stroke sequelae.

KEYWORDS: STROKE; CONTRALESIONAL; IPSILESIONAL; LOWER LIMBS; INTERLIMB COORDINATION; DOUBLE SUPPORT PHASE; BIOMECHANICS; VARIABILITY

LISTA DE ABREVIATURAS E SÍMBOLOS

AVC	Acidente Vascular Cerebral
BF	Biceps Femoris
CONTRA	Contralesional
DOM	Dominante
EMG	Eletromiografia
GasM	Gastrocnemius Medialis
GMax	Gluteus Maximus
ICC	Coeficiente de Correlação Intraclassa
IPSI	Ipsilesional
LEAD	Leading limb
NDOM	Não Dominante
p (P)	p-value
RF	Rectus Femoris
RMS	Root mean square
SOL	Soleus
TA	Tibialis Anterior
TRAIL	Trailing limb
WCOM	Trabalho mecânico externo sobre o centro de massa
VM	Vastus Medialis

CAPÍTULO I – INTRODUÇÃO GERAL

O Acidente Vascular Cerebral (AVC) é uma condição de incapacidade com incidência significativa na população adulta (Langhorne et al., 2011; Sturm et al., 2002). A recuperação da função motora do membro inferior contralesional (CONTRA) e da capacidade de marcha são considerados objetivos primários da reabilitação (Lord et al., 2004; Vestling et al., 2003). A reabilitação das sequelas da marcha ocorre em cerca de 95% das pessoas nas primeiras 11 semanas pós-AVC (Jorgensen et al., 1995). No entanto, o padrão da marcha desvia-se frequentemente do padrão típico das pessoas saudáveis e cerca de um terço dos sobreviventes não saem de casa sem supervisão (Lord et al., 2004).

A recuperação das sequelas após um AVC é um processo heterogêneo e complexo que ocorre pela combinação de processos espontâneos e dependentes da aprendizagem (Kwakkel et al., 2004). Atualmente, supõe-se que as alterações na capacidade motora podem ocorrer pelo processo de restituição (o que reflete o processo de “verdadeira recuperação”) ou de compensação (Bernhardt et al., 2017). A recuperação da função motora do membro inferior (domínio das estruturas e funções do corpo), até à fase crónica é conseguida, principalmente, através do processo de restituição, enquanto a recuperação da capacidade da marcha (domínio das atividades) reflete o efeito combinado dos processos de restituição e compensação (Frenkel-Toledo et al., 2021). Embora as estratégias compensatórias possam, a curto prazo, ajudar na realização das tarefas para alcançar a autonomia nas atividades básicas diárias o mais rapidamente possível, e sejam a opção face às limitações impostas pelos custos de tratamento (Krakauer & Carmichael, 2017), a longo prazo podem conduzir a problemas como a redução da amplitude de movimento articular e a dor (Levin, 1996). Assim, o foco da reabilitação das sequelas pós-AVC deverá centrar-se na melhoria da capacidade funcional, mas de acordo com a restituição dos padrões de movimento pré-lesão.

A distinção da recuperação entre os padrões de movimento pré-lesão e os padrões de movimento alternativos, adotados ou ensinados à pessoa para compensar os défices, pode ser feita a partir da seleção de medidas de avaliação qualitativa e quantitativa do desempenho sensório-motor (Langhorne et al., 2011; Levin et al., 2009; Veerbeek et al., 2014). Nesta perspetiva, a análise quantitativa da marcha, em contexto laboratorial, permite uma avaliação precisa e repetível da função motora do membro inferior durante a marcha, possibilitando a medição de parâmetros objetivos para a deteção dos défices

de movimento (Kwakkel et al., 2017; Mohan et al., 2021), assim como a monitorização da evolução sensório-motora face ao programa de reabilitação (Mohan et al., 2021). A análise biomecânica, através de variáveis cinemáticas, cinéticas e eletromiográficas, é considerada uma das melhores ferramentas para diferenciar o processo de restituição e de compensação, e para melhorar a compreensão sobre estes mecanismos que orientam a recuperação sensório-motora (Beyaert et al., 2015; Kwakkel et al., 2017). A identificação dos métodos e das medidas, específicos para esta análise, poderá contribuir para a compilação de parâmetros de avaliação, auxiliar no processo de raciocínio clínico para orientar as estratégias de intervenção e demonstrar a eficácia da intervenção do programa de neuro-reabilitação (Kwakkel et al., 2017). A avaliação do desempenho motor da marcha, na fase crónica pós-AVC, pode refletir, não só o impacto dos danos na neuroanatomia funcional do controlo motor, mas também nos fatores neurofisiológicos que podem condicionar os processos de restituição e a capacidade de adotar estratégias compensatórias do comportamento motor (Frenkel-Toledo et al., 2021).

Atualmente, na reabilitação das sequelas pós-AVC na fase crónica, ambas as estratégias de restituição e de compensação são utilizadas de forma cooperativa (Frenkel-Toledo et al., 2021; Krakauer & Carmichael, 2017). No entanto, será importante refletir sobre o foco da reabilitação se centrar na melhoria da capacidade funcional, visando a recuperação dos padrões de movimento aproximada aos padrões típicos saudáveis prévios à lesão.

A coordenação entre membros na marcha resulta numa interação de parâmetros biomecânicos, sinergicamente controlados para o desempenho motor ótimo (Dietz & Schrafl-Altermatt, 2016), adaptável ao contexto e à tarefa (Krasovsky & Levin, 2010). A avaliação e monitorização quantitativa de variáveis biomecânicas que expressam a coordenação entre os membros inferiores na fase de duplo apoio e que refletem a eficiência da marcha, para além de permitir a melhor compreensão da evolução clínica (Kwakkel et al., 2017), poderá auxiliar na identificação criteriosa dos défices a considerar na intervenção, permitindo direcionar para a (re)aprendizagem motora, de acordo as características biomecânicas típicas do movimento, nesta fase específica da marcha (Beyaert et al., 2015).

Considerando que a marcha humana é controlada por comandos motores neurais, através de uma orquestração complexa de forças musculares e movimentos articulares, diversas variáveis biomecânicas têm sido utilizadas para a avaliação da coordenação entre membros, incluindo a cinemática articular, as forças de reação ao solo, e a atividade eletromiográfica. A fase de duplo apoio da marcha corresponde à fase de transição entre passos, refletindo o principal determinante do trabalho mecânico na marcha e na sua eficiência (Kuo, 2007). Apesar de existir um vasto número de medidas clínicas e laboratoriais que pretendem medir a coordenação entre membros, até onde sabemos, não existe consenso quanto aos métodos e à sua adequação para avaliar esta coordenação, especificamente durante a fase de duplo apoio, em pessoas com sequelas pós-AVC.

Os instrumentos e as medidas da análise biomecânica devem ser selecionados de forma a permitir a identificação dos défices de movimento que expressam a coordenação entre membros. Para além de quantificar os défices e as diferenças entre os membros, a quantificação da variabilidade dessas medidas pode também traduzir o impacto do AVC (Frykberg & Vasa, 2015) no desempenho da marcha, na medida em que, uma maior variabilidade parece estar relacionada com um menor desempenho nas medições (Balasubramanian et al., 2009). Por outro lado, a variabilidade superior parece também refletir uma menor capacidade de adaptação da marcha ao contexto e à tarefa (Cavanaugh & Stergiou, 2020; Hausdorff, 2005).

Reconhecendo que a análise da variabilidade pode permitir identificar possíveis estratégias adaptativas na recuperação pós-AVC, o seu estudo surge como um novo ponto de análise para uma melhor interpretação da expressão neuromotora desta condição patológica.

A compreensão da neuro-anatomia e neuro-fisiologia do controlo motor saudável e do comprometimento do controlo neural supraspinal da marcha após um AVC, acrescentam conhecimento às, já comumente descritas, alterações biomecânicas identificadas no membro CONTRA. Sabe-se que o membro ipsilesional (IPSI), pelo processo de compensação, desenvolve estratégias adaptativas face às alterações da lesão do lado CONTRA (Clark et al., 2010; Hendrickson et al., 2014; Kim & Eng, 2004; Lewek et al., 2014;

Olney et al., 1994; Olney et al., 1991; Parvataneni et al., 2007; Peterson et al., 2010). Uma das principais alterações refere-se, por exemplo, ao aumento do trabalho mecânico do lado IPSI para compensar a diminuição do trabalho do lado CONTRA na propulsão anterior do corpo e o maior dispêndio energético na colisão do pé no contacto inicial, de forma a influenciar a eficiência da marcha (Beyaert et al., 2015; Nadeau et al., 2013; Peterson et al., 2010; Sheffler & Chae, 2015).

De acordo com a compreensão da influência neural bilateral, as alterações que o membro IPSI apresenta poderão ser, também, uma expressão da própria lesão neural (sob a forma de disfunção) (Silva et al., 2017). O AVC pode conduzir a alterações neurais e biomecânicas, com expressão bilateral, CONTRA e IPSI, que justificam o comportamento atípico de ambos os membros inferiores, com maior impacto na coordenação entre membros na fase de duplo apoio (Arya & Pandian, 2014; Bajwa et al., 1992; Crafton et al., 2003; Matsuyama et al., 2004; Schiemanck et al., 2006; Sousa et al., 2013; Sousa & Tavares, 2015).

O conhecimento de que a alteração bilateral dos membros após um AVC tem impacto na coordenação entre membros, tem vindo a ser reconhecido na investigação e na neuro-reabilitação desta condição patológica (Arya & Pandian, 2014; Silva et al., 2017), mas continua a ser pouco aprofundado. O membro inferior IPSI (muitas vezes mencionado como “não afetado”) ainda é usado como a medida de referência e controlo para o membro CONTRA. No entanto, se o membro IPSI também apresentar alteração do comportamento motor, a sua influência sobre a coordenação entre membros também deverá ser revista e esta comparação direta entre ambos os membros das pessoas com sequelas de AVC deverá ser questionável, uma vez que poderá levar a que a recuperação do desempenho do CONTRA possa estar a ser subestimado (Raja et al., 2012), limitando a identificação do potencial máximo de recuperação. Assim, surge a necessidade de comparar e compreender o comportamento de variáveis biomecânicas que expressam a coordenação entre membros (parâmetros da cinemática, cinética e eletromiografia), em ambos os lados, CONTRA e IPSI, bem como o impacto sobre a variabilidade das mesmas, face à referência comparativa dos membros inferiores de pessoas saudáveis, na fase de duplo apoio da marcha, à mesma velocidade.

De modo a dar resposta às questões de investigação e aos respetivos objetivos, esta Tese encontra-se organizada em 7 capítulos. O **Capítulo I** compreende uma visão global da pertinência desta investigação. A revisão da literatura sobre tópicos de relevância para a compreensão do tema de investigação desta Tese, nomeadamente o impacto do AVC na coordenação entre membros na fase de duplo apoio da marcha, e as principais alterações descritas na literatura, a pertinência da avaliação biomecânica do desempenho motor da marcha para a interpretação dos mecanismos de recuperação sensório-motora nas pessoas com sequelas de AVC encontra-se no **Capítulo II**. O **Capítulo III** explora as três questões de investigação e os respetivos objetivos. O **Capítulo IV** é composto por três estudos originais desenvolvidos no âmbito desta investigação. A discussão integrada dos resultados, as considerações metodológicas e as implicações para a prática clínica e a investigação futura encontram-se no **Capítulo V**. O **Capítulo VI** descreve as principais conclusões desta Tese, seguindo-se as referências bibliográficas da presente Tese. Contudo, as referências bibliográficas dos estudos originais estão apresentadas no final de cada um, para manter a estrutura original destes.

CAPÍTULO II – REVISÃO DA LITERATURA

O impacto do Acidente Vascular Cerebral

O Acidente Vascular Cerebral (AVC) é caracterizado por um déficit neurológico, de causa vascular, resultante da interrupção focal súbita da irrigação sanguínea ao Sistema Nervoso Central. Em 2019, o AVC permaneceu como a terceira principal causa de morte e incapacidade no mundo (Feigin et al., 2021). A evolução na qualidade dos cuidados de saúde, na fase aguda, conduziu a uma maior proporção de sobrevivência das pessoas com sequelas de AVC e, por consequência, a maior proporção de pessoas com limitações nas atividades de vida diária e restrições à participação no contexto profissional e social (Rajsic et al., 2019).

Os défices neuromotores consequentes do AVC dependem da gravidade, da localização, da extensão da lesão, do tempo pós-lesão, da reabilitação e das diferenças individuais (Olney & Richards, 1996). No entanto, o comprometimento mais comum pós-AVC é o comprometimento motor, considerado como uma perda ou limitação da função muscular e do movimento (Langhorne et al., 2009). O comprometimento motor pós-AVC pode influenciar no controlo do movimento da face, do membro superior, do tronco e do membro inferior, afetando cerca de 80% das pessoas (Langhorne et al., 2009). As consequências físicas, a longo prazo, representam um dos maiores impactos para as pessoas e familiares, assim como para a sua economia (Langhorne et al., 2011). A recuperação da independência para o retorno à participação individual na comunidade e nas atividades ocupacionais são dos principais objetivos para as pessoas com sequelas de AVC, cuidadores e profissionais de saúde (Belda-Lois et al., 2011; Bohannon et al., 1991; Langhorne et al., 2009; Mudge & Stott, 2007; Shumway-Cook, 2017; Sousa et al., 2013).

O foco da reabilitação pós-AVC, e em particular da intervenção da Fisioterapia, centra-se na recuperação do movimento e das funções associadas (Langhorne et al., 2009). Entre as diversas repercussões motoras decorrentes do AVC, a diminuição da capacidade de produzir força, ou paresia, constitui o componente primário associada às alterações funcionais, nomeadamente da marcha (Chen & Patten, 2008; Knarr et al., 2013; Lamontagne et al., 2002; Perry & Burnfield, 2010). As alterações na marcha destacam-se como as que mais interferem com o nível de participação da pessoa no seu contexto quotidiano (Combs et al., 2013; Lord et al., 2004). A prevalência da limitação de

mobilidade na marcha é elevada, por exemplo, após 11 semanas, apenas 50% das pessoas com sequelas de AVC recuperam a capacidade de andar independente (Woolley, 2001) e cerca de 30% são incapazes de andar ou necessitam da supervisão de outra pessoa ou do uso de auxiliar de marcha (Lord et al., 2004; Woolley, 2001). Neste sentido, a recuperação da marcha independente é um dos objetivos primordiais para as pessoas com sequelas de AVC, bem como para a intervenção da Fisioterapia na reabilitação neuromotora (Balaban et al., 2011; Bohannon et al., 1991; Harris & Eng, 2004).

A marcha humana

A marcha humana é uma tarefa funcional extraordinariamente complexa, que envolve todo o corpo e, portanto, requer a coordenação de vários músculos e articulações. Pode ser caracterizada por três requisitos essenciais: a progressão, o controlo postural e a adaptação (Massion et al., 2004; Shumway-Cook, 2017; Winter, 2004). A progressão é assegurada por um padrão locomotor básico que coordena padrões rítmicos de ativação muscular para o movimento do corpo na direção desejada (Massion et al., 2004; Shumway-Cook, 2017; Winter, 2004). O controlo postural envolve a organização de múltiplos sistemas no corpo para alcançar tanto a orientação como a estabilidade postural (Massion et al., 2004; Shumway-Cook, 2017; Winter, 2004). Durante a marcha a progressão e o controlo postural atuam em sinergia para assegurar um movimento anterior do corpo estável. A adaptação refere-se à capacidade de modificar o padrão de marcha, incluindo estratégias para alcançar a progressão e o controlo postural face à mudança das tarefas e das exigências ambientais (Patla et al., 1991; Reisman et al., 2010; Shumway-Cook, 2017). Estes requisitos são dependentes de uma interação multifatorial que resulta da organização biomecânica (dinâmica músculo-esquelética) e neurofisiológica (circuito medular do gerador de padrão central, do controlo neural supraspinal e da informação sensorial periférica), adaptável às exigências individuais e ambientais (Lacquaniti et al., 2012; MacKay-Lyons, 2002; Rossignol et al., 2006).

A biomecânica da marcha humana

A marcha, num ciclo, pode ser dividida em duas principais fases: a fase de apoio, que associa a forças horizontal para a progressão do corpo e vertical para o suporte da massa corporal contra gravidade (controlo postural); e a fase de balanço, que permite a progressão da marcha e a preparação para o posicionamento do membro inferior para o contacto com o solo e acomodação da massa corporal (Shumway-Cook, 2017).

A transição entre as fases envolve o controlo dinâmico do deslocamento do centro de massa na base de suporte. Num ciclo de marcha, a trajetória do centro de massa no plano sagital é uma curva sinusoidal, deslocando-se duas vezes na vertical e lateralmente no plano horizontal, sendo estas duas curvas similares (Gard et al., 2004; Levingie & Norkin, 2011). A teoria mecânica do pêndulo invertido (Kuo, 2007) propõe que a marcha é energeticamente menos dispendiosa se o membro inferior se comportar como um pêndulo, dependendo do trabalho muscular para acelerar e desacelerar este pêndulo (Cavagna & Margaria, 1966).

A fase de duplo apoio da marcha

A fase de duplo apoio representa 20% da fase de apoio, num ciclo de marcha, e corresponde a dois eventos em que ambos os pés estão em contacto com o solo: o *heel-strike* (10% inicial) e o *toe-off* (10% final) (Neumann, 2017; Perry & Burnfield, 2010). Esta fase de duplo apoio corresponde à fase de transição entre passos, na qual os membros inferiores formam uma cadeia cinemática fechada, e a conservação da energia da fase de apoio unipodálico é interrompida pela colisão do membro oscilante com o solo (Kuo et al., 2005). Neste momento de colisão ocorre a libertação de energia de forma inelástica ao longo do membro e de outras partes do corpo, aquando do contacto com o solo (Donelan et al., 2002b; Yeom & Park, 2011). De facto, e de acordo com Donelan et al. (2002a), o trabalho realizado na transição entre passos pode explicar 60-70% do dispêndio energético da marcha. Para além da perda energética, o maior efeito da colisão é a alteração da velocidade dos membros e do centro de massa. A alteração da velocidade ocorre devido à troca dos membros em apoio, em que o início do contacto ao solo de

um, desaceleração, é seguido do término do apoio do outro, aceleração (Donelan et al., 2002b). A colisão inelástica com dissipação de energia, traduzida no trabalho negativo sobre o centro de massa, pode refletir-se na ativação muscular e na deformação de tecidos moles (Kuo et al., 2005). Este trabalho negativo, e a consequente redução da magnitude da velocidade, permite que a fase de apoio unipodálica seguinte inicie com menor velocidade do que a precedente. A perda de energia pode ser reduzida em 75% através do impulso de propulsão do membro de apoio imediatamente antes da colisão, expressa no trabalho positivo (Kuo, 2002).

De facto, na fase de duplo apoio, o trabalho positivo sobre o centro de massa, realizado pelos músculos do membro inferior, na fase de pré-balanço, denominado de *trailing limb* (TRAIL), deriva da energia metabólica e mecânica armazenada nos componentes elásticos, durante a fase precedente ao trabalho negativo realizado pelo membro inferior na fase de contacto inicial, denominado de *leading limb* (LEAD) (Donelan et al., 2002b). Idealmente, esta transição deve ser curta em duração e distância, e deve ser de igual magnitude (Donelan et al., 2002a, 2002b; Kuo, 2007; Kuo et al., 2005). O modelo teórico proposto por Kuo (2007), evidencia que a transição entre passos, retratada num pêndulo invertido, atinge um nível ótimo quando a fase final do apoio e o contato inicial têm a mesma magnitude e a mesma duração, refletindo assim o principal determinante do trabalho mecânico na marcha e na eficiência desta.

Relativamente à cinemática, no contacto inicial com o solo, a coxo-femoral encontra-se em aproximadamente 20° de flexão, progredindo para extensão, até cerca de 20° de hiperextensão à medida que transita na fase de apoio (Neumann, 2017; Sheffler & Chae, 2015). De seguida, na fase de pré-oscilação, inicia-se a redução da hiperextensão para dar início à flexão para a fase de oscilação (Neumann, 2017; Sheffler & Chae, 2015). O joelho contacta o solo com 5° de flexão para a absorção da força na colisão com o solo. Gradualmente, o joelho entra na extensão até o suporte da massa corporal, voltando à flexão até cerca de 40° na fase de pré-balanço, para facilitar a progressão anterior do corpo até à saída do pé do solo (Neumann, 2017; Sheffler & Chae, 2015). No tornozelo, a flexão plantar é crítica a partir do momento de contacto com o solo e na fase de pré-balanço a antecipar a saída do pé. A dorsiflexão gradual do tornozelo até à fase média de

apoio é necessária para a progressão anterior da tíbia (Neumann, 2017; Sheffler & Chae, 2015).

No que diz respeito à ativação muscular na marcha, esta visa alcançar os dois requisitos relacionados com o suporte do corpo contra gravidade, na subfase de contacto inicial e acomodação da carga (controlo postural), e na impulsão do corpo para anterior na subfase de pré-balanço (progressão). Na subfase de contacto inicial com o solo, a maior desaceleração do membro inferior é conseguida pelos músculos extensores uni e bi-articulares da coxo-femoral (*Gluteus Maximus*, Isquiotibiais) que, pela ação excêntrica, promovem o controlo da extensão da coxo-femoral, a redução da velocidade para o contacto com o solo e, por conseguinte, a acomodação da carga (Neptune et al., 2004; Perry & Burnfield, 2010). Nesta subfase, o joelho mantém uma pequena flexão controlada pela coativação dos Isquiotibiais (excentricamente) e do *Quadríceps* (concentricamente), favorecendo a absorção do impacto resultante da colisão do pé com o solo e consequente acomodação da massa corporal. O tornozelo contacta com o solo em posição neutra ou ligeira flexão plantar, pela ação excêntrica do *Tibialis Anterior*, para a desaceleração do movimento do pé até ao solo. Na subfase seguinte, acomodação da carga, a atividade excêntrica dos flexores plantares permitem o controlo do movimento anterior da tíbia e perónio sobre o pé. Na subfase de pré-balanço, a flexão concêntrica da coxo-femoral é dependente da ação dos flexores uni e bi-articulares (Iliopsoas, *Rectus Femoris*), que influenciam excentricamente e de forma passiva a flexão do joelho (*Rectus Femoris*) (Neptune et al., 2004). A atividade dos flexores plantares (*Soleus* e *Gastrocnemius*) também promove alguma flexão ao joelho, mas atinge o seu pico de atividade na flexão plantar concêntrica do tornozelo, subfase em que são responsáveis pela aceleração vertical e anterior do membro inferior, ou seja, ao gerar a força propulsiva para manter a progressão do corpo (Chen & Patten, 2008; Kepple et al., 1997; Neptune et al., 2001b; Perry & Burnfield, 2010; Sadeghi, 2000).

Durante a marcha, o efeito da massa corporal no apoio é quantificado pela força de reação do solo, que tem uma magnitude igual, mas oposta, à força produzida no solo pelo membro inferior no contacto inicial.

A coordenação entre membros na marcha humana

Krasovsky and Levin (2010) definem a coordenação entre membros na marcha como a capacidade de manter uma relação cíclica entre os diferentes segmentos ou articulações do corpo, no domínio espacial e temporal, dependente do contexto e da fase. A coordenação entre membros é alcançada pelo acoplamento dos movimentos destes (Dietz & Michel, 2009; Zehr & Duysens, 2004), numa interação entre a cinemática, a cinética e atividade muscular em movimentos que requerem ritmo, sequência e cooperação de ambos os lados do corpo, como é o exemplo da marcha (Arya & Pandian, 2014). Este acoplamento pode ser definido, numa perspetiva neural, como uma ligação neuroanatômica e neurofisiológica específica, mas flexível e adaptável ao contexto (Arya & Pandian, 2014), descrita pelo controlo sinérgico de ambos os membros para o desempenho ideal (Dietz & Schrafl-Altermatt, 2016).

O estudo da coordenação entre membros baseia-se no impacto sobre o custo metabólico (Donelan et al., 2002a, 2002b; Kuo, 2007; Kuo et al., 2005), e assenta em evidência biomecânica (Donelan et al., 2002a, 2002b; Kuo, 2007; Kuo et al., 2005) e neurofisiológica a nível espinal (Bajwa et al., 1992; Carson et al., 2014; Dietz, 1992) e supraspinal (Drew et al., 2004; Matsuyama et al., 2004).

Na transição entre passos, a coordenação entre os membros inferiores assume um papel crucial para responder às exigências do controlo postural e do movimento. A fase de duplo apoio responde a esta exigência associada à economia da marcha (Donelan et al., 2002a; Donelan et al., 2004; Krasovsky et al., 2012; Kuo, 2007; Sousa & Tavares, 2015), e tem sido apontada como um fator determinante na eficiência da marcha (Kuo, 2002, 2007; Kuo et al., 2005). O trabalho mecânico externo sobre o centro de massa realizado por cada membro inferior pode refletir a coordenação entre os membros. Os dois membros inferiores desempenham um papel determinante na progressão anterior e vertical do centro de massa decorrente do trabalho positivo do membro inferior posterior durante a propulsão – TRAIL que compensa a energia despendida pelo membro inferior anterior na fase de contacto inicial – LEAD (Donelan et al., 2002a; Kuo, 2007).

A neurofisiologia da marcha humana

A marcha humana é um movimento rítmico e automático dos dois membros inferiores (Dietz, 2002a). Este movimento é baseado num controlo tripartido, nomeadamente: o circuito medular, as estruturais neurais supraspinais e a informação aferente (MacKay-Lyons, 2002).

O circuito localizado na Medula Espinal, determinado geneticamente, é capaz de gerar padrões de atividade rítmica, na ausência de *feedback* externo ou do controlo supraspinal, denominado de gerador de padrão central (Grillner, 2006; MacKay-Lyons, 2002; Rossignol et al., 2006). Os geradores de padrão central promovem o movimento rítmico, coordenado e alternado dos membros inferiores na marcha (Dimitrijevic et al., 1998). São redes de neurónios e interneurónios espinais, localizados nos segmentos lombo-sagrados, que modulam com precisão a atividade dos neurónios motores inferiores para produzir a flexão e extensão alternada dos membros inferiores (Danner et al., 2015; MacKay-Lyons, 2002). Cada membro inferior possui um gerador de padrão central, dividido em dois “meios-centros”, um flexor e outro extensor, controlados separadamente (Grillner et al., 2008; Yang et al., 2005). Ambos os geradores são coordenados por sinais na comissura anterior da Medula Espinal (Butt et al., 2002), de modo que quando ocorre flexão de um membro inferior, no outro ocorre extensão. Estes geradores de padrão central regulam o nível de atividade dos neurónios motores inferiores para a formação do padrão e a duração das fases do ciclo de marcha para a geração do ritmo (Frigon, 2012; Lacquaniti et al., 2012; McCrea & Rybak, 2008). Para além de produzir o padrão repetitivo e estereotipado de atividade, os geradores de padrão central recebem múltiplas aferências propriocetivas (com origem nos Fusos Neuromusculares, Órgãos Tendinosos de Golgi e recetores articulares) para o controlo minucioso da posição, do movimento e da força muscular durante o ciclo da marcha (Edgerton et al., 2004).

A flexão e extensão alternada dependente do circuito básico dos geradores de padrão central, não é o único mecanismo responsável pela marcha, pois o controlo supraspinal postural automático e cortical voluntário têm um papel relevante na modulação deste

circuito (Capaday et al., 1999; Duysens et al., 2013; Pearson & Gordon, 2018; Zelik et al., 2014).

No tronco encefálico está identificada uma área, denominada de “Região Locomotora Mesencefálica”, com um papel importante no controlo da iniciação da marcha. Os neurónios desta região do Mesencéfalo ativam a Formação Reticular Ponto-Medular, especificamente o núcleo *Gigantocellularis* que tem uma ação excitatória sobre os geradores de padrão central (Brocard & Dubuc, 2003; Drew et al., 2004; Schepens et al., 2008). Estes circuitos medulares são ainda modulados por influências do Cerebelo, que recebe *feedback* aferente dos recetores sensoriais relacionados com a marcha (MacKay-Lyons, 2002) (vias espinocerebelares: dorsal – transmite informação aferente dos recetores musculares; e ventral – transmite informação dos neurónios espinais relativos ao *output* do gerador de padrão central) (Grimaldi & Manto, 2012; Massaquoi, 2012; Morton & Bastian, 2004). O Cerebelo envia sinais moduladores para núcleos do Tronco Encefálico (Morton & Bastian, 2004), destacando os núcleos vestibular, rubro e reticular, que têm influência direta ou indireta nos neurónios motores inferiores, pelas vias motoras descendentes (vestibulospinal, rubrospinal e reticulospinais), que refinam o movimento, momento a momento, e de acordo com a exigência da tarefa (Cullen, 2012; Morton & Bastian, 2004; Takakusaki, 2017). Tem ainda um papel preponderante no *timing* de ativação muscular para a modulação do ciclo de passos face à informação dos sistemas somatossensoriais, visuais e vestibulares, não só para a correção, mas também para o ajuste do padrão do passo pela influência sobre o Córtex Cerebral (Massaquoi, 2012; Morton & Bastian, 2004; Takakusaki et al., 2008). Os Núcleos da Base, através do Tronco Encefálico e do Córtex Cerebral, exercem influência nas vias descendentes da Medula Espinal contribuindo para o controlo automático e voluntário do movimento da marcha e para a manutenção do tónus postural adequado, sendo muito importantes na iniciação e finalização do programa motor da marcha (MacKay-Lyons, 2002; Takakusaki, 2013, 2017; Takakusaki et al., 2004). O Córtex, apesar de não ter grande relevância na marcha automática, está envolvido no controlo voluntário pela modificação dos geradores de padrão central face às mudanças das características do contexto, como superfícies instáveis ou irregulares e na orientação visual da marcha (Lajoie et al., 2010; Takakusaki, 2008). A Área Suplementar Motora está implícita no controlo postural

(mecanismo antecipatório) com uma estreita conexão neuronal com a Formação Reticular, através da via cortico-reticular, importante na preparação postural para a iniciação da marcha (Takakusaki, 2017; Takakusaki et al., 2008). A Área Suplementar Motora associada ao Córtex Pré-Motor contribuem para o planeamento e programação do movimento preciso da marcha (Miyai et al., 2002; Takakusaki, 2017; Takakusaki et al., 2008).

A modulação neural da marcha, pela ação das estruturas supraspinais, atinge os neurónios e interneurónios do circuito medular (Dietz, 2002b), pelos sistemas motores descendentes dorsolaterais e ventromedianos (Kandel et al., 2014; Takakusaki, 2017). Estes sistemas motores têm características específicas para o controlo da marcha, nomeadamente, os sistemas motores dorsolaterais, predominantemente contralaterais (rubro-espinal e cortico-espinal), envolvidos no controlo do movimento distal e principalmente relacionados com a flexão, enquanto os sistemas ventromedianos (retículo-espinal e vestibulo-espinal), predominantemente ipsilaterais, são responsáveis pelo controlo postural e atividade antigravítica, e têm muitas projeções que terminam bilateralmente estando envolvidos no controlo bilateral dos músculos axiais (pescoço e tronco) e proximais (Kandel et al., 2014; Rothwell, 2012).

Para além da modulação supraspinal, as entradas sensoriais também são fundamentais para a modulação da atividade da marcha na adaptação às mudanças das condições ambientais (MacKay-Lyons, 2002; Rossignol et al., 2006), nomeadamente a informação somatossensorial, a vestibular e a visual (Shumway-Cook, 2017). A informação aferente assume um papel importante: na regulação detalhada do padrão temporal da sequência de ativação muscular (Ivanenko et al., 2006; Pearson & Collins, 1993); no reforço da atividade muscular, particularmente dos músculos que suportam a massa corporal, principalmente os extensores dos membros inferiores na fase de apoio da marcha (Cattaert, 2004; Pearson & Collins, 1993; Sinkjaer et al., 2000; Stephens & Yang, 1996) e no controlo da transição entre as fases da marcha (Cattaert, 2004; Lacquaniti et al., 1999; Pearson & Collins, 1993).

A variabilidade da marcha humana

A variabilidade é uma característica fundamental, inerente ao controlo do movimento humano, ou seja, reflete diferentes recursos neuromotores disponíveis para produzir estratégias de movimento, necessárias para alcançar o objetivo motor (Stergiou & Decker, 2011; Stergiou et al., 2006). O conceito de variabilidade de movimento é definido como a expressão de variações típicas no desempenho motor quando uma tarefa é executada repetidamente (Stergiou et al., 2006), um aspeto que tem vindo a ter relevo nos estudos de investigação clínica. A variabilidade ótima do movimento é crucial para um controlo motor saudável (Stergiou & Decker, 2011) e um alto nível de automatismo e variabilidade relativamente constante parece ser expectável quando uma tarefa bem conhecida é executada repetidamente (Carson et al., 2014).

A marcha humana representa uma forma de movimento altamente qualificada, que se caracteriza por flutuações naturais, em adultos saudáveis (Stergiou et al., 2006). Tradicionalmente, o aumento da variabilidade na marcha pode corresponder a uma diminuição da estabilidade mecânica, enquanto a diminuição da variabilidade corresponde a um sistema mais estável e cooperativo (Stergiou et al., 2006). A menor variabilidade da marcha permite uma melhor adaptabilidade do padrão de marcha às mudanças ambientais (Cavanaugh & Stergiou, 2020; Hausdorff, 2005).

Estudos cinemáticos que descreveram os movimentos dos segmentos do corpo sugerem que as estratégias de movimento são semelhantes entre pessoas saudáveis, ou seja, a marcha parece ter um padrão cinemático consistente. Em contraste, estudos que descreveram os músculos e as forças associadas à marcha sugerem que existe uma grande variabilidade na forma como estes movimentos são alcançados. Assim, parece haver uma grande variedade de padrões de ativação muscular utilizados por pessoas sem condição patológica, para cumprir os requisitos da tarefa de marcha (Shumway-Cook, 2017).

A marcha humana após um Acidente Vascular Cerebral

A marcha humana reflete a integridade dos circuitos medulares sob a modulação supraspinal adequada que conferem o padrão automático e coordenado bilateralmente entre os membros inferiores, (Arya & Pandian, 2014; Dietz, 2002a). A lesão do Sistema Nervoso Central, consequente de um Acidente Vascular Cerebral (AVC), resulta numa alteração do controlo neural supraspinal da marcha (Corbetta et al., 2015). As alterações da marcha pós-AVC apresentam-se sob a forma de um padrão assimétrico, realçando a necessidade do conhecimento e compreensão sobre como o AVC pode interferir com a modulação da marcha (Hendrickson et al., 2014; Kim & Eng, 2003; Lewek et al., 2014; Morita et al., 1995).

O AVC compromete a função cortico-subcortical, mas preserva a função da Medula Espinal. Ocorre, comumente, no território da Artéria Cerebral Média (Crafton et al., 2003; Schiemanck et al., 2006), afetando áreas motoras corticais, mas também estruturas subcorticais envolvidas no controlo neural da marcha.

Numa perspetiva cortical, a interação transcalosal inter-hemisféricas (Grefkes & Fink, 2011; Shimizu et al., 2002; Ward & Cohen, 2004) pode ser comprometida pós-AVC, dado que a lesão de um hemisfério cerebral pode interferir com o processamento entre ambos os hemisférios (Cleland & Madhavan, 2021, 2022). Por outro lado, o hemisfério cerebral é responsável pelo controlo do movimento do corpo contralateral ou contralesional pós-AVC (CONTRA), pela ação do feixe descendente com origem no Córtex (corticospinal), que se projeta para a região dorsolateral da Medula Espinal (Kandel et al., 2014). No entanto, sabe-se que 10% a 15% dessas fibras corticospinais não decussam para o lado contralateral (Shumway-Cook, 2017), permanecem ipsilateralmente, sendo responsáveis pelo controlo do movimento distal ipsilateral ou ipsilesional pós-AVC (IPSI) (Soteropoulos et al., 2011). Nesta perspetiva, existem autores que referem que as sequelas pós-AVC podem levar, também, ao comprometimento motor do lado ipsilateral do corpo, tendo este uma expressão menos severa do que o lado CONTRA (Alagona et al., 2001; Schaefer et al., 2009). Numa perspetiva subcortical, o AVC compromete feixes que conectam o Córtex com estruturas do Tronco Encefálico, como a Formação Reticular, através do feixe cortico-reticular (Silva et al., 2017), que cursa paralelamente ao corticospinal, mas que se

projeta bilateralmente para os núcleos reticulares (Cleland & Madhavan, 2021). Destes, emergem sistemas ventromedianos que atingem a Medula Espinal, os feixes reticulospinais, responsáveis pelo controlo ipsilateral e contralateral do movimento (Rothwell, 2009, 2012), originando alterações bilaterais. Os feixes vestibulares têm também ação bilateral sobre os músculos axiais e proximais (Gjelsvik & Syre, 2016). Portanto, um AVC pode afetar as vias motoras de um lado do Cérebro, resultando numa alteração bilateral, com expressão distinta em ambos os lados do corpo (Silva et al., 2017). Dependendo da localização da lesão (Matsuyama et al., 2004), o desempenho do membro CONTRA pode ser comprometido pela lesão das vias neuronais responsáveis pelo controlo do movimento, enquanto o desempenho do membro IPSI pode ser afetado pelas vias neuronais responsáveis pelo controlo postural (Matsuyama et al., 2004; Rothwell, 2009, 2012).

Estes aspetos neurofisiológicos suprajacentes poderão fundamentar o comprometimento bilateral dos membros inferiores na marcha, pós-AVC (Silva et al., 2017), mas resultando numa afeção do desempenho motor com expressão distinta em ambos os membros inferiores (Arya & Pandian, 2014; Bajwa et al., 1992; Crafton et al., 2003; Matsuyama et al., 2004; Schiemanck et al., 2006; Sousa et al., 2013; Sousa & Tavares, 2015).

Esta hipótese de que o lado IPSI pós-AVC poderá apresentar uma expressão da própria lesão neuronal (disfunção), pode acrescentar conhecimento às já comumente descritas alterações adaptativas biomecânicas em ambos membros inferiores (Kim & Eng, 2004; Olney et al., 1994; Olney et al., 1991; Parvataneni et al., 2007), ou seja, associadas aos mecanismos neuronais de compensação do lado IPSI, que acontecem pelos ajustes face à lesão do lado CONTRA (Silva et al., 2017).

Assim, o comprometimento da coordenação entre membros, na marcha, pode ocorrer após um AVC (Dietz & Berger, 1984; Roerdink et al., 2007; Wu et al., 2009) devido à lesão primária cerebral e/ ou ser resultante de alterações adaptativas em estruturas neurais (Beyaert et al., 2015; Crone et al., 2003; Gale & Pearson, 2012; Jang et al., 2013; Lamy et al., 2009; Murase et al., 2004; Schmid et al., 2013).

A importância de explorar a coordenação entre membros em pessoas com sequelas de AVC depende, não são só do impacto da lesão do membro CONTRA sobre o membro IPSI, mas também da disfunção do IPSI na performance do CONTRA.

As alterações na marcha das pessoas com sequelas de um Acidente Vascular Cerebral

As sequelas pós-AVC expressam-se em alterações sensorio-motoras, incluindo fraqueza muscular (paresia), défices no controlo postural e no controlo motor seletivo, nomeadamente o aumento da oscilação postural, alteração da intensidade e tempo de ativação muscular, desorganização de sinergias musculares, co-contracção excessiva e assimetria na distribuição da massa corporal para o lado IPSI (Balaban et al., 2011; den Otter et al., 2004; Krasovsky & Levin, 2010; Lamontagne et al., 2000; Nadeau et al., 2013; Neckel et al., 2006; Olney & Richards, 1996; Perry et al., 1995; Rosa et al., 2014; Roth et al., 1997; Sheffler & Chae, 2015; Wang et al., 2020). Estas alterações conduzem ao desenvolvimento de padrões de movimento compensatórios, resultando num padrão de marcha com características atípicas e assimétricas em ambos os membros inferiores (Krasovsky & Levin, 2010; Oken & Yavuzer, 2008). As alterações do membro CONTRA têm sido vastamente exploradas (Kim & Eng, 2003; Lamontagne et al., 2000; Olney et al., 1994; Olney et al., 1991; Olney & Richards, 1996; Sheffler & Chae, 2015; Titianova et al., 2003) comparativamente às do membro IPSI (Lamontagne et al., 2000; Silva et al., 2012; Silva et al., 2015; Sousa et al., 2015; Sousa et al., 2013).

Considerando o comprometimento neuromotor da marcha pós-AVC, esta é frequentemente caracterizada por diferenças nos parâmetros espaço-temporais, alterações nos padrões cinemáticos, cinéticos e na atividade muscular, exprimindo um maior gasto de energia nesta condição patológica (Arya & Pandian, 2014; Bowden et al., 2006; Chen & Patten, 2008; Hendrickson et al., 2014; Krasovsky & Levin, 2010; Lamontagne et al., 2000; Lamontagne et al., 2007; Mahon et al., 2015; Mohan et al., 2021; Perry & Burnfield, 2010; Sheffler & Chae, 2015; Yavuzer et al., 2001).

Parâmetros espaço-temporais

As principais assimetrias nos parâmetros espaço-temporais são referentes, maioritariamente, a alterações temporais das fases da marcha, como na fase de apoio e de balanço, ou espaciais como a diminuição do comprimento do passo e do ciclo (Balasubramanian et al., 2007; Kim & Eng, 2003; Lewek et al., 2014; Patterson et al., 2010; Titianova et al., 2003; Woolley, 2001). A assimetria temporal descrita refere um maior tempo da fase de balanço do membro CONTRA, e um maior tempo da fase de apoio do membro IPSI (Kim & Eng, 2003; Mohan et al., 2021; Titianova et al., 2003). A diminuição da velocidade da marcha e da cadência também são alterações comuns associadas às alterações das fases, como por exemplo, um maior tempo da fase de duplo apoio representa uma menor velocidade da marcha (Mohan et al., 2021; von Schroeder et al., 1995). As assimetrias espaciais são menos consistentes (Balaban & Tok, 2014), sendo reportado um maior comprimento do passo, normalmente do lado CONTRA, mas o mesmo também pode ser observado do lado IPSI (Balasubramanian et al., 2007; Woolley, 2001).

As diferenças mencionadas dependem, em larga escala, da gravidade, da localização e do tipo do AVC, mas também das sequelas e complicações motoras associadas, determinando o nível de disfunção (Balaban & Tok, 2014).

Cinemática

As principais alterações cinemáticas reportadas incluem a redução da amplitude articular dos membros inferiores (Nadeau et al., 2013; Mohan et al., 2021).

No membro inferior CONTRA destaca-se, na fase de pré-balanço, a diminuição da extensão da coxo-femoral (Moseley et al., 1993; Nadeau et al., 2013; Sheffler & Chae, 2015), a diminuição da flexão do joelho (*recurvatum*) (de Quervain et al., 1996; Nadeau et al., 2013) e a diminuição da flexão plantar do tornozelo (Moseley et al., 1993; Nadeau et al., 2013). A hiperextensão do joelho compromete a progressão anterior da tíbia e do

perónio na fase média de apoio, resultando numa inadequada flexão plantar na fase de apoio final.

Na fase de contacto inicial com o solo, é característico encontrar uma menor flexão da coxo-femoral (Nadeau et al., 2013; Sheffler & Chae, 2015), um aumento da flexão do joelho (Moore et al., 1993; Nadeau et al., 2013), associado à instabilidade postural para a colisão e absorção da carga de impacto, e um aumento da flexão plantar para o contacto do calcanhar com o solo (Moore et al., 1993; Sheffler & Chae, 2015).

Cinética

As principais alterações cinéticas documentadas referem-se à redução da força produzida no solo que determina os momentos e a potência dos músculos (Mohan et al., 2021; Nadeau et al., 2013; Olney et al., 1991; Parvataneni et al., 2007; Sheffler & Chae, 2015). A força de reação do solo do membro CONTRA na posição TRAIL é menor enquanto a da posição LEAD é maior, demonstrando uma assimetria entre estas duas forças na fase de duplo apoio (Beyaert et al., 2015; Bowden et al., 2006; Kim & Eng, 2003; Morita et al., 1995). Nestas duas posições, destacam-se os flexores plantares na propulsão anterior do corpo, os extensores da coxo-femoral (Morita et al., 1995; Neptune et al., 2001a; Olney et al., 1991; Turns et al., 2007) na fase de apoio inicial com o solo (LEAD), e os flexores da coxo-femoral na fase de pré-balanço (TRAIL), estando associados à menor velocidade da marcha, característica das pessoas com sequelas de AVC (Bowden et al., 2006; Jonkers et al., 2009).

Estas alterações resultam num aumento do gasto energético da marcha pós-AVC (Detrembleur et al., 2003). Este excesso de energia consumido parece dever-se ao maior trabalho positivo necessário do membro IPSI, para propulsionar o corpo na direção vertical e anterior (Beyaert et al., 2015; Chen et al., 2005; Mahon et al., 2015), no sentido de compensar a diminuição da energia gerada pelos flexores plantares no membro CONTRA na fase de pré-balanço (Beyaert et al., 2015; Mahon et al., 2015; Peterson et al., 2011).

Eletromiografia

As alterações da atividade eletromiográfica identificadas nesta condição patológica são relativas ao tempo e/ ou magnitude de ativação muscular (Jonkers et al., 2009; Mohan et al., 2021). No entanto, a eletromiografia tem vindo a ser menos explorada comparativamente com a cinemática e com a cinética.

Na fase de pré-balanço, a diminuição da extensão da coxo-femoral pode ser justificada pela diminuição da atividade dos Isquiotibiais e do *Gluteus Maximus* (Mulroy et al., 2003; Sousa et al., 2013). A diminuição da flexão do joelho (*recurvatum*) pode ser explicada pela hiperatividade (espasticidade ou contratura) dos flexores plantares, hiperatividade do *Quadríceps* ou diminuição da atividade dos Isquiotibiais (Mulroy et al., 2003; Sheffler & Chae, 2015). A alteração da posição do joelho pode influenciar a atividade dos flexores plantares na progressão para a fase média de apoio e, por consequência, reduzir a força necessária destes músculos para a fase de pré-balanço (Cooper et al., 2012; Kim & Eng, 2004; Krasovsky et al., 2013; Mulroy et al., 2003; Olney et al., 1998; Sheffler & Chae, 2015; Wong et al., 2004). O GasM e o SOL são fundamentais na estabilidade do tornozelo durante o aumento da dorsiflexão desde a fase média de apoio até à fase de pré-balanço, sendo a diminuição da sua contribuição nesta fase uma das principais alterações que caracterizam a marcha pós-AVC (Mulroy et al., 2003; Peterson et al., 2010).

No contacto inicial com o solo, a atividade reduzida do *Quadríceps* (*Vastus Medialis*) parece estar relacionada com o aumento da flexão do joelho (Kim & Eng, 2004; Krasovsky et al., 2013; Sheffler & Chae, 2015). O aumento da flexão plantar no contacto do calcanhar com o solo, pode estar relacionada com uma atividade excessiva ou prolongada do GasM e do SOL (Cappellini et al., 2010; Chen et al., 2007; Clark et al., 2010; Sheffler & Chae, 2015; Wong et al., 2004) ou com uma alteração da atividade excêntrica do TA, no controlo da descida do pé ao solo (Clark et al., 2010; Kim & Eng, 2004; Sousa et al., 2013)

Variabilidade

Em pessoas pós AVC, o comprometimento neuromotor parece estar relacionado com uma maior variabilidade da marcha e um menor desempenho nas medições clínicas de equilíbrio e função (Balasubramanian et al., 2009). Guzik et al. (2018) e Guzik et al. (2019) confirmaram a maior variabilidade nos parâmetros espaço-temporais da marcha em pessoas com sequelas de AVC comparativamente às pessoas sem condição neurológica. Patel et al. (2022) referiu que a variabilidade motora elevada é uma alteração proeminente pós-AVC, mostrando que uma maior variabilidade dos parâmetros temporais tem um impacto negativo na coordenação entre os membros na marcha, apesar de reconhecer que o aumento da variabilidade da marcha pós-AVC ainda não é bem compreendido. Algumas evidências sugerem que pessoas com sequelas de AVC apresentam alteração da coordenação dos membros inferiores durante o movimento voluntário do tornozelo na marcha em passadeira (Roerdink et al., 2007; Tseng & Morton, 2010). Dado que a marcha humana é um movimento altamente repetível, uma maior variabilidade nos seus parâmetros pode refletir um erro na seleção ou execução de programas motores, interferindo com o seu desempenho (Patel et al., 2022).

Tradicionalmente, uma variabilidade superior na marcha pode corresponder a uma diminuição da estabilidade mecânica, enquanto uma variabilidade mais baixa corresponde a um sistema mais estável e cooperativo (Stergiou et al., 2006). Assim, a menor variabilidade na marcha permite uma melhor adaptabilidade do seu padrão às mudanças ambientais (Cavanaugh & Stergiou, 2020; Hausdorff, 2005). Em condições patológicas, a variabilidade tende a ser superior (Kao et al., 2014; Rosano et al., 2007) e, especificamente em estudos em pessoas com sequelas de AVC, foi demonstrada uma elevada variabilidade nos parâmetros espaço-temporais (Allen et al., 2011; Balasubramanian et al., 2009; Chisholm et al., 2014; Patel et al., 2022).

Avaliação da marcha pós-Acidente Vascular Cerebral e os mecanismos de recuperação sensório-motora

Os instrumentos e as medidas de avaliação da qualidade do desempenho motor devem ser selecionados de forma a permitir distinguir a recuperação de padrões de movimento pré-mórbidos dos padrões de movimento alternativos, adotados ou ensinados à pessoa para compensar a perda destes padrões de movimento (Levin et al., 2009). Muitos estudos utilizam medidas clínicas para avaliar os défices e a alteração da função após o AVC. As escalas que avaliam os défices, como a Escala de Avaliação *Fugl-Meyer* (FMA) (Kwakkel et al., 2017; Pohl et al., 2020) e o teste de 10 metros de marcha (Pohl et al., 2020), medem aspetos motores específicos que podem limitar, mas que não estão relacionados com a realização de atividades (por exemplo, a força e o movimento articular isolado), enquanto que as escalas funcionais, tais como o *Time Up & Go* e a Escala de Equilíbrio de *Berg* (Pohl et al., 2020), medem o nível de sucesso da atividade (Levin et al., 2009; Pohl et al., 2020). As pontuações das escalas funcionais podem resultar em melhorias face à reabilitação, independentemente de se restituir o padrão motor original ou de se desenvolverem padrões motores compensatórios crescentes, e a distinção entre elas não é clara (Levin et al., 2009). A análise qualitativa convencional da marcha, tipicamente utilizada na prática clínica, baseia-se principalmente na observação da marcha e é, portanto, também subjetiva e largamente influenciada pela experiência do observador (Mohan et al., 2021). Apesar das escalas que avaliam os défices oferecerem ao clínico uma apreciação dos défices motores específicos (Levin et al., 2009), elas são fortemente influenciadas pela experiência do observador (Patterson et al., 2011), o que as torna subjetivas.

Para além dos défices comumente quantificados na marcha, sabe-se que esta atividade é caracterizada por variações naturais no desempenho motor, identificadas como a variabilidade ótima de movimento em pessoas saudáveis (Stergiou et al., 2006). A variabilidade do movimento poder ser avaliada em termos da quantidade de variabilidade, através de ferramentas estatísticas lineares (como, por exemplo, o valor do ICC pelo número de repetições mínimo para a estabilidade de desempenho; o desvio padrão ou o coeficiente de variação) (Harbourne & Stergiou, 2009) e pode traduzir o impacto do AVC no desempenho da marcha (Balasubramanian et al., 2009).

Reconhecendo a quantificação da variabilidade como uma possível estratégia adaptativa pós-AVC, o seu estudo surge como um novo ponto de análise para uma melhor compreensão da expressão da lesão pós-AVC, acrescentando conhecimentos às alterações neuromotoras nesta condição.

A avaliação quantitativa da marcha, em contexto laboratorial, permite uma medição criteriosa e repetível da função motora do membro inferior (Nadeau et al., 2013). Assim, a quantificação de parâmetros objetivos para detetar os défices de movimento (Kwakkel et al., 2017; Mohan et al., 2021), e também para monitorizar os programa de intervenção, surge com muita relevância para a área da reabilitação (Mohan et al., 2021). Atualmente, a análise biomecânica, através de variáveis cinemáticas, cinéticas e eletromiográficas, é considerada como uma das melhores formas para diferenciar os mecanismos de recuperação neuromotora, como a restituição e a compensação, permitindo melhorar a compreensão sobre a evolução da recuperação sensoriomotora (Kwakkel et al., 2017).

A recuperação pós-AVC é um processo heterogéneo e complexo que ocorre pela combinação de processos espontâneos e dependentes da aprendizagem (Kwakkel et al., 2004). A recuperação espontânea refere-se a melhorias na recuperação do comportamento, na ausência de um tratamento específico e direcionado, ocorre numa janela temporal que se inicia logo após o AVC e diminui lentamente (Bernhardt et al., 2017). A (re)aprendizagem motora depende da reaquisição de padrões motores elementares (como sinergias musculares ou de movimento, reaprendendo a aplicá-los em diferentes combinações para realizar tarefas motoras desejadas) ou, na ausência de reaquisição, adaptação de remanescentes ou integração de elementos motores alternativos (Levin et al., 2009).

A reabilitação das sequelas do AVC visa facilitar e melhorar a recuperação da função sensoriomotora através da combinação de estratégias de intervenção orientadas para a restituição e/ ou para a compensação. A recuperação da função motora do membro inferior (domínio das estruturas e funções do corpo) até à fase crónica, é conseguida principalmente através do processo de restituição, enquanto a recuperação da capacidade de marcha (domínio das atividades) reflete o efeito combinado dos processos de restituição e compensação (Frenkel-Toledo et al., 2021). No entanto, embora as

estratégias compensatórias possam, a curto prazo, ajudar na realização das tarefas para alcançar a independência nas atividades básicas da vida diária o mais rapidamente possível, e sejam a opção face às limitações impostas pelos custos de tratamento (Krakauer & Carmichael, 2017), a longo prazo, podem conduzir a problemas como a restrição da amplitude de movimento articular e a dor (Levin, 1996). Assim, o foco da reabilitação pós-AVC deve centrar-se na melhoria da capacidade funcional através da recuperação dos padrões de movimento prévios à lesão.

As abordagens orientadas para a restituição consideram o processo de recuperação natural direcionado para a neuroplasticidade cerebral (capacidade de reorganização adaptativa das relações da estrutura-função no Sistema Nervoso Central danificado), e conduzem à reaquisição, tanto quanto possível, das características do movimento voluntário pré-mórbido (Krakauer & Carmichael, 2017; Levin et al., 2009), refletindo o processo de “verdadeira recuperação” (Bernhardt et al., 2017). Na estratégia orientada para a restituição, o padrão motor que se deseja restituir é analisado no domínio das “estruturas e funções do corpo” segundo a Classificação Internacional de Funcionalidade. As abordagens orientadas para a compensação consideram o processo de recuperação direcionado para a aquisição de novas funções do tecido neural, que podem ser diferentes das prévias à lesão, e conduzem à realização das tarefas motoras de uma nova forma. Esta estratégia orientada para a compensação refere-se ao “domínio das atividades” da Classificação Internacional de Funcionalidade, associada ao objetivo de atingir um melhor desempenho nas atividades básicas e instrumentais da vida diária, com a máxima autonomia e segurança (Krakauer & Carmichael, 2017; Levin et al., 2009). Por outro lado, a compensação visa apenas a conclusão bem-sucedida da tarefa, com a utilização dos membros da forma mais eficaz. Isto pode requerer o uso de diferentes recursos ou padrões de movimento motores (não necessariamente da forma mais próxima do padrão pré-mórbido) como, por exemplo, a recuperação da capacidade de marcha sem ou com recurso de ajudas técnicas, como auxiliares de marcha (Krakauer & Carmichael, 2017; Levin et al., 2009), ou o aumento da oscilação do membro superior e inferior IPSI para melhorar a velocidade da marcha (Kwakkel & Wagenaar, 2002). Quanto maior a gravidade do comprometimento sensório-motor, maior a tendência para o

desenvolvimento de padrões de movimentos compensatórios para melhorar a capacidade funcional (Levin et al., 2009).

Atualmente, na reabilitação pós-AVC, ambas as estratégias são utilizadas de forma cooperativa (Krakauer & Carmichael, 2017). No entanto, será importante refletir sobre o foco da reabilitação pós-AVC que deverá centrar-se na melhoria da capacidade funcional através da recuperação dos padrões de movimento prévios, o mais próximo do padrão típico saudável.

A compreensão deste conceito de recuperação sensório-motora é importante para a neuro-reabilitação, em particular para a reabilitação pós-AVC. A identificação de medidas e métodos específicos poderá permitir contribuir para a compilação de dados, a interpretação dos mecanismos subjacentes à recuperação motora, bem como auxiliar no processo de raciocínio clínico, para orientar as estratégias de intervenção e demonstrar a eficácia desta intervenção na recuperação pós-AVC (Kwakkel et al., 2017).

Muitos investigadores exploraram a marcha pós-AVC, incluindo instrumentos e técnicas de avaliação, parâmetros-chave da marcha e importância na recuperação funcional (Mohan et al., 2021). Contudo, até onde sabemos, não existe consenso quanto ao método, e à sua adequação, para avaliar a coordenação entre membros, especificamente durante a fase de duplo apoio da marcha, em pessoas com sequelas pós-AVC.

CAPÍTULO III – OBJETIVOS DE INVESTIGAÇÃO

Considerando a contextualização científica, e destacando i) a marcha como a atividade motora que proporciona a maior autonomia, referida como um dos objetivos primordiais das pessoas com sequelas de Acidente Vascular Cerebral (AVC); ii) a coordenação entre membros, um aspecto do controlo motor da marcha comprometida na lesão pós-AVC, expressa durante a fase de duplo apoio; iii) a crescente, mas ainda pouco reconhecida, evidência do comprometimento neuromotor bilateral (lesão/ disfunção) traduzido nas alterações dos parâmetros biomecânicos, mas também na variabilidade motora pós-AVC; iv) a necessidade da comparação entre pessoas com e sem sequelas de AVC; v) a exigência da análise cinemática, cinética e eletromiográfica do desempenho motor de pessoas pós-AVC de acordo com as recomendações mais atuais, o objetivo principal desta tese foi caracterizar a coordenação entre os membros inferiores na marcha, especificamente na fase de duplo apoio, em pessoas com e sem sequelas de AVC, através da análise cinemática, cinética e eletromiográfica.

Apesar da evidência apontar para a relevância da coordenação entre membros na marcha como o fator determinante na sua eficiência, e existirem alguns estudos que avaliaram a coordenação de diferentes formas, ainda não existe uma orientação clara para identificar quais os instrumentos e as variáveis mais adequadas e relevantes para avaliar a coordenação na fase de duplo apoio. A necessidade de definir a metodologia mais adequada e de acordo com as orientações científicas suportou o objetivo de investigação 1.

Objetivo de investigação 1

Rever e sumariar os métodos e estratégias para avaliar a coordenação entre membros, nomeadamente as condições da marcha, os principais instrumentos e as respetivas variáveis biomecânicas, em pessoas com sequelas de um AVC.

Este objetivo de investigação foi abordado no artigo I: Couto, A.G.B., Vaz, M.A.P., Pinho, L., Félix, J., Silva, S., Silva, A., & Sousa, A.S.P. (2022). Methodological Considerations in Assessing Interlimb Coordination on Poststroke Gait: A Scoping Review of Biomechanical Approaches and Outcomes. *Sensors*, 22, 2010. <https://doi.org/10.3390/s22052010>.

Permanece ainda em falta recomendações para a análise da consistência temporal através do número mínimo de repetições necessário para atingir uma estabilidade de desempenho ($ICC > 0,75$), numa sessão e entre sessões de avaliação biomecânica. Esta estabilidade de desempenho pode traduzir o impacto do AVC na variabilidade das variáveis biomecânicas que expressam na coordenação entre membros, face ao comprometimento bilateral, contralesional (CONTRA) e ipsilesional (IPSI), na fase de duplo apoio.

Objetivo de investigação 2

Analisar a repetibilidade e a consistência temporal de variáveis biomecânicas ao longo de múltiplas tentativas, intra-sessão e inter-sessões, através do número mínimo de repetições de marcha necessário para obter uma boa estabilidade de desempenho ($ICC > 0,75$) de parâmetros cinemáticos (tempo de duplo de apoio e posição articular da anca, do joelho e do tornozelo), cinéticos (trabalho externo sobre o centro de massa) e eletromiográficos (média de atividade muscular e rácios de ativação muscular) dos membros inferiores durante a fase do duplo apoio da marcha em pessoas com e sem sequelas de AVC.

Este objetivo de investigação foi abordado no artigo II: Couto, A.G.B., Vaz, M.A.P., Pinho, L., Félix, J., Moreira, J., Pinho, F., Mesquita, I. A., Mesquita Montes, A., Crasto, C., & Sousa, A.S.P. (2023). Repeatability and Temporal Consistency of Lower Limbs Biomechanical Variables Expressing Interlimb Coordination during Double-Support Phase in People with and without Stroke Sequelae. *Sensors*, 23, 2526. <https://doi.org/10.3390/s23052526>.

O comprometimento bilateral dos membros inferiores nesta condição patológica, ainda carece de estudo, nomeadamente no que se refere à identificação de possíveis alterações, principalmente no membro IPSI, comparativamente a pessoas sem sequelas de AVC, como sendo a referência típica e o potencial máximo de reabilitação.

Objetivo de investigação 3

Identificar variáveis biomecânicas, nomeadamente parâmetros cinemáticos (tempo de duplo de apoio e posição articular da anca, do joelho e do tornozelo), cinéticos (trabalho externo sobre o centro de massa) e eletromiográficos (média de atividade muscular e rácios de ativação muscular) dos membros inferiores durante a fase do duplo apoio da marcha que permitam diferenciar pessoas com e sem sequelas de AVC.

Este objetivo de investigação foi abordado no artigo III: Couto, A.G.B., Vaz, M.A.P., Pinho, L., Félix, J., Moreira, J., Pinho, F., Mesquita, I. A., Mesquita Montes, A., Crasto, C., & Sousa, A.S.P. (2023). Interlimb coordination during double support phase of gait in people with and without stroke sequelae. Submetido para publicação.

CAPÍTULO IV – ESTUDOS ORIGINAIS

ARTIGO I

Methodological Considerations in Assessing Interlimb Coordination on Poststroke Gait: A Scoping Review of Biomechanical Approaches and Outcomes

Ana G. B. Couto ^{1,2,3}, Mário A. P. Vaz ⁴, Liliana Pinho ^{2,5,6}, José Félix ⁷, Sandra Silva ⁵, Augusta Silva ⁸ and Andreia S. P. Sousa ⁸

1 Escola Superior de Saúde de Santa Maria, Travessa Antero de Quental, 173, 4049-024 Porto, Portugal; ana.couto@santamariasauade.pt

2 Centro de Investigação em Reabilitação, Centro de Estudos de Movimento e Atividade Humana, Rua Dr. António Bernardino de Almeida, 400, 4200-072 Porto, Portugal; liliana.pinho@ipsn.cespu.pt

3 Faculdade de Engenharia, Universidade do Porto, Rua Dr. Roberto Frias, s/n, 4200-465 Porto, Portugal

4 Departamento de Engenharia de Mecânica, Faculdade de Engenharia, Universidade do Porto (INEGI/ Labiomep), Rua Dr. Roberto Frias, s/n, 4200-465 Porto, Portugal; gmavaz@fe.up.pt

5 Escola Superior de Saúde do Vale do Ave, Cooperativa de Ensino Superior Politécnico e Universitário, Rua José António Vidal, 81, 4760-409 Vila Nova de Famalicão, Portugal; sandra.silva@ipsn.cespu.pt

6 Faculdade de Desporto, Universidade do Porto, Rua Dr. Plácido da Costa, 91, 4200-450 Porto, Portugal

7 Departamento de Física, Centro de Investigação em Reabilitação, Centro de Estudos de Movimento e Atividade Humana, Escola Superior de Saúde do Porto, Instituto Politécnico do Porto, Rua Dr. António Bernardino de Almeida, 400, 4200-072 Porto, Portugal; josefelixfelix15@gmail.com

8 Área Científica de Fisioterapia, Centro de Investigação em Reabilitação, Centro de Estudos de Movimento e Atividade Humana, Escola Superior de Saúde do Porto, Instituto Politécnico do Porto, Rua Dr. António Bernardino de Almeida, 400, 4200-072 Porto, Portugal; smaugusta@gmail.com

Couto, A.G.B., Vaz, M.A.P., Pinho, L., Félix, J., Silva, S., Silva, A., & Sousa, A.S.P. (2022). Methodological Considerations in Assessing Interlimb Coordination on Poststroke Gait: A Scoping Review of Biomechanical Approaches and Outcomes. *Sensors*, 22, 2010. <https://doi.org/10.3390/s22052010>

Abstract

Objective: To identify and summarize biomechanical assessment approaches in inter-limb coordination on poststroke gait. **Introduction:** Interlimb coordination involves complex neurophysiological mechanisms that can be expressed through the biomechanical output. The deepening of this concept would have a significant contribution in gait rehabilitation in patients with an asymmetric neurological impairment as poststroke adults. **Inclusion criteria:** Poststroke adults (>19 years old), with assessment of interlimb coordination during gait, in an open context, according to the Population, Concept, Context framework. **Methods:** A literature search was performed in PubMed, Web of Science™, Scopus, and gray literature in Google Scholar™, according to the PRISMA-ScR recommendations. Studies written in Portuguese or English language and published between database inception and 14 November 2021 were included. Qualitative studies, conference proceedings, letters, and editorials were excluded. The main conceptual categories were “author/ year”, “study design”, “participant’s characteristics”, “walking conditions”, “instruments” and “outcomes”. **Results:** The search identified 827 potentially relevant studies, with a remaining seven fulfilling the established criteria. Interlimb coordination was assessed during walking in treadmill (n = 3), overground (n = 3) and both (n = 1). The instruments used monitored electromyography (n = 2), kinetics (n = 2), and kinematics (n = 4) to assess spatiotemporal parameters (n = 4), joint kinematics (n = 2), anteroposterior ground reaction forces (n = 2), and electromyography root mean square (n = 2) outcomes. These outcomes were mostly used to analyze symmetry indices or ratios, to calculate propulsive impulse and external mechanical power produced on the centre of mass, as well as antagonist coactivation. **Conclusions:** Assessment of interlimb co-ordination during gait is important for consideration of natural auto-selected overground walking, using kinematic, kinetic, and EMG instruments. These allow for the collection of the main biomechanical outcomes that could contribute to improve better knowledge of interlimb coordination assessment in poststroke patients.

Keywords

Double support; interlimb; lower limbs; poststroke adults; walking

Introduction

Interaction between posture and movement demands a higher challenge during walking (Horak, 2006; Shumway-Cook, 2017). This requires a close coordination of muscle activity between the two lower limbs (Dietz, 2002) in a complex control of a moving center of mass (CoM), that is not within the base of foot support (Horak, 2006; Shumway-Cook, 2017).

The importance of this interlimb coordination during gait is even more highlighted by its strong association with gait economy (Donelan et al., 2002a, 2002b; Kuo, 2007; Kuo et al., 2005). This association has been supported by biomechanical factors (Donelan et al., 2002a, 2002b; Kuo, 2007; Winter & Eng, 1995), that seem to reflect both spinal (Bajwa et al., 1992; Corna et al., 1996; Dietz, 1992; Rossignol et al., 2006; Zehr & Duysens, 2004) and supraspinal mechanisms (Drew et al., 2004; Lacquaniti et al., 2012a, 2012b; Mac-Kay-Lyons, 2002; Matsuyama et al., 2004; Rossignol et al., 2006; Zehr & Duysens, 2004) involved in interlimb coordination. Based on this, and on its definition, the timing of lower limbs motor cycles in relation to one another (Swinnen & Carson, 2002), it can be stated that interlimb coordination would reflect the principles of central networks organization to generate muscle activity patterns, that determine body kinematics, kinetics and efficiency (Sousa & Tavares, 2015).

From the first step, Central Nervous System (CNS) seeks to achieve the perfect coordination of body segments to maintain postural stability during forward progression (Winter & Eng, 1995). The relation between movement and postural control, depending, respectively, of the dorsolateral and ventromedial systems activation, is expressed through the functional unit composed by both lower limbs during walking to keep the body CoM over the feet (Dietz, 2002). Specifically, this neural coupling unit depends on cortico-ponto-reticular-spinal interneuronal system and vestibular system together with the spinal circuits (Jiang et al., 2015). In fact, the maintenance of reciprocal control of interlimb coordination is ensured by two systems of spinal interneurons (Takakusaki, 2013). This is particularly critical for stable human bipedal walking during step-to-step transition, considering its impact on the mechanical work on gait (Donelan et al., 2002b; Kuo, 2007).

During the double support of walking, lower limbs forces are needed to redirect the CoM velocity from a downward and forward direction to an upward and forward direction (Donelan et al., 2002a; Kuo, 2007). Particularly, the interruption of the energy-conserving motion during single support by an inelastic collision of the swing leg with the ground is related to negative work on the CoM that leads to changes in velocities of the lower limbs and the CoM (Geyer et al., 2006; Kuo, 2007). Consequently, a substantial amount of positive mechanical work, predominantly related to the propulsion impulse in the trailing limb immediately before collision of the leading limb, is needed to redirect the CoM between steps (Donelan et al., 2002a, 2002b; Kuo, 2002, 2007; Kuo et al., 2005; Sousa & Tavares, 2015). Transition between steps reaches an optimum level when the terminal stance propulsion and the initial contact-loading response have the same magnitude and a short duration (Kuo, 2007).

Considering that human walking is controlled by neural motor commands, through a complex orchestration of muscle forces and joint motions, several biomechanical variables, including electromyographic (EMG) activity, ground reaction forces (GRF) and resulting limb motions, have been used to assess interlimb coordination during walking. However, to the best of our knowledge there is no consensus regarding the method and its suitability for assessing gait interlimb coordination.

Thus, the purpose of this study was to review and summarize the methods/ strategies used to assess the interlimb coordination, for a better understanding of its underpinning mechanisms and its role in gait performance and rehabilitation.

Review Questions

The two main questions have been addressed in this review:

- Which motor task has been considered to assess the interlimb coordination in poststroke adults?
- What are the main instruments and outcomes that have been used to assess the interlimb coordination in poststroke adults?

Inclusion criteria

Eligibility criteria were established a priori using the acronym PCC (Population, Concept, Context), defining the search strategy (Peters et al., 2020) as:

- Population: Poststroke adults (> 19 years old) (Sawyer et al., 2018).
- Concept: Interlimb coordination assessment during gait.
- Context: Open.
- Types of source evidence: It was included studies in Portuguese or English language, on poststroke adult population assessed in terms of interlimb coordination during gait. Qualitative studies, conference proceedings, letters and editorials were excluded.

Methods

The scoping review was conducted in accordance with the Preferred Reporting Items for Systematic Reviews and Meta-Analysis extension for Scoping Reviews (PRISMA-ScR) (Peters et al., 2020).

Search Strategy

The review was performed, aiming to identify relevant evidence on the theme, by combining specific terms with the Boolean logic strategies in the following expression: (interlimb AND (walk* OR gait OR locomotion) AND ("stance phase" OR "double support" OR "support phase" OR "step-to-step")) AND (stroke OR "Cerebral Vascular Accident" OR CVA), in the electronic databases PubMed, Web of ScienceTM, Scopus[®] and grey literature in Google ScholarTM, between database inception and 14 November 2021.

Source of Evidence Screening and Selection

The process of source of evidence screening and selection involved several stages. Elimination of duplicated records was automatically performed by the EndNote software. For the eligibility process the search strategy was limited to titles and abstracts and was performed by two independent reviewers (AGBC & LOP). This process was synthesized using worksheets of Excel 2021 (Microsoft®). The reference lists of all studies were also scanned to identify other potential eligible articles. When discrepancies occurred between reviewers on whether the study should be included in the review, the reasons for disagreement were analyzed, the trial report was consulted, and a consensus was achieved. This process consisted of a selection of 25 randomized titles/ abstracts to consider; the independent reviewers screened the selected sample using the eligibility criteria and definitions/ elaboration for this scope; then they met to discuss discrepancies and made modifications to the eligibility criteria and definitions/ elaboration to be analyzed. The reviewers only started screening when 75% (or greater) agreement was achieved. The results of the search were presented in a PRISMA-ScR flow diagram as shown in Figure 1.

Data Extraction

Data from publications were extracted based on the following main conceptual categories: “author/ year”, “study design”, “participant’s characteristics”, “motor task”, “instruments” and “outcomes”.

Analysis and Presentation of Results

The screening and selection process were presented in a diagrammatic form. A tabular form and a descriptive summary were used to describe the main conceptual categories, answering to the review objectives and questions.

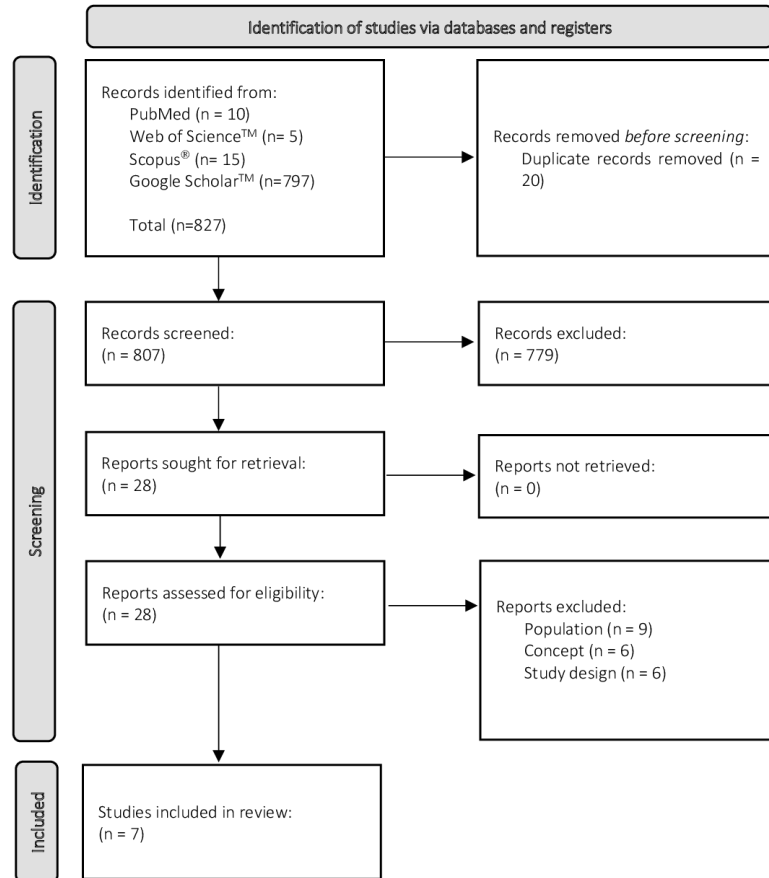


Figure 1. Flow diagram for the scoping review process adapted from the PRISMA statement (Page et al., 2021).

Results

The search identified 827 potentially relevant studies, 10 from PubMed, 5 from Web of Science™, 15 from Scopus® and 797 from Google Scholar™. After removing 20 duplicated records, 807 remained, of which 779 were excluded after screening the title and abstract. Over the 28 lasting, 21 studies were excluded after full-text reading, by not addressing the proposed questions for this review, remaining 7 studies fulfilling the established criteria.

Study Design

Four studies included in this review were Observational, Transversal, Analytical (Mahon et al., 2015; Puh & Baer, 2009; Silva et al., 2015; Sousa et al., 2013), 2 were Quasi-Experimental (Krasovsky et al., 2013; Reisman et al., 2009) and 1 was Randomized Controlled Trial (Družbicki et al., 2016). The publication year ranged from 2009 to 2016. It should be noted that the older studies were published 12 years ago (Puh & Baer, 2009; Reisman et al., 2009) (see Table 1).

Table 1. Descriptive poststroke participant's characteristics and walking conditions descriptions.

Author, Year	Study Design	Poststroke Participant's Characteristics	Walking Conditions
Reisman et al., 2009	Quasi-experimental	• $n = 11$ (2 F, 9 M) • Age: 55 (35–70) years old Stroke characteristics • Time poststroke onset: (8–128) months • Stroke side: left (6), right (5) • Stroke type: ischemic (7), hemorrhagic (4) • Stroke localization: parietal (2), hemisphere (6), basal ganglia (1), caudate head, anterior limb internal capsule (1), posterior temporoparietal (1) Clinical measures • Lower Extremity Fugl–Meyer Assessment	Overground walking • Self-selected speed
Puh & Bear, 2009	Observational, transversal, analytical	• $n = 10$ (5 F, 5 M) • Age: 37.3+/-10.6 (19–54) years old • Height: 171.5+/-7.2 (163–185) cm • Body weight: 71.5+/-14.2 (53.5–100) kg Stroke characteristics • Time poststroke onset: 26+/-19.6 (9–95) months • Hemiparetic side: left (2), right (8)	Overground walking and treadmill walking • Self-selected speed
Sousa et al., 2013	Observational, transversal, analytical	• $n = 16$ (8 F, 8 M) • Age: 53.87+/-7.17 years old • Height: 1.65+/-0.10 m • Body weight: 75.29+/-7.03 kg Stroke characteristics • Time poststroke onset: 26+/-9 months • Stroke type: ischemic (16) • Stroke side: left (11), right (5) • Stroke localization: middle cerebral artery Clinical measures • Lower Extremity Fugl–Meyer Assessment • Mini-Mental State Examination	Overground walking • Self-selected speed

Krasovsky et al., 2013	Quasi-experimental	• $n = 10$ (0 F, 10 M) • Age: 60.3+/-11.3 years old • Height: 176.3+/-8.4 cm • Body weight: 87.1+/-13.5 kg Stroke characteristics • Time poststroke onset: 28.9+/-23.4 months • Stroke type: ischemic (8), hemorrhagic (2) • Stroke side: left (4), right (6) • Stroke localization: middle cerebral artery-internal capsule (1), middle cerebral artery (2), bulbar (1), caudate nucleus (1), thalamus-internal capsule (1), subcortical (2), anterior choroidal artery (1), putamen-internal capsule (1) Clinical measures • Functional Gait Assessment • Activities-Specific Balance Confidence Questionnaire • Lower Extremity Fugl-Meyer Assessment	Treadmill walking • Conditioned at different speeds	
Mahon et al., 2015	Observational, transversal, analytical	High-speed group • $n = 13$ (6 F, 7 M) • Age: 56+/-8.4 years old • Height: 175+/-8.4 cm • Body weight: 91+/-18 kg Stroke characteristics • Time poststroke onset: 103+/-92 months • Hemiparetic side: left (7), right (6) Clinical measures • Lower Extremity Fugl-Meyer Assessment	Low-speed group • $n = 13$ (6 F, 7 M) • Age: 54+/-12 years old • Height: 173+/-9.3 cm • Body weight: 93+/-13 kg Stroke characteristics • Time poststroke onset: 30+/-17 months • Hemiparetic side: left (7), right (6) Clinical measures • Lower Extremity Fugl-Meyer Assessment	Treadmill walking • Conditioned at different speeds
Silva et al., 2015	Observational, transversal, analytical	• $n = 16$ (8 F, 8 M) • Age: 53.87+/-7.17 years old • Height: 1.65+/-0.10 m • Body weight: 75.29+/-7.03 kg Stroke characteristics • Time poststroke onset: 26+/-9 months	Overground walking • Self-selected speed	

		- Stroke type: ischemic (16) - Stroke side: left (11), right (5) - Stroke localization: middle cerebral artery Clinical measures - Lower Extremity Fugl–Meyer Assessment - Mini-Mental State Examination	
Družbicki et al., 2016	Randomized Controlled Trial	Intervention group $n = 15$ (6 F, 9 M) - Age: 61.9+/-11.4 years old Stroke characteristics - Time poststroke onset: 36 (8–120) months - Hemiparetic side: left (6), right (9) Clinical measures - Mini-Mental State Examination - Brunnström - Ashworth Modified Scale	Control group $n = 10$ (5 F, 5 M) - Age: 59.8+/-11.7 years old Stroke characteristics - Time poststroke onset: 38.2 (8–110) months - Hemiparetic side: left (6), right (4) Clinical measures - Mini-Mental State Examination - Brunnström - Ashworth Modified Scale Treadmill walking - Self-selected speed

Poststroke Participant's Characteristics

In the 7 studies included, 3 were performed exclusively with poststroke participants (Druzbicki et al., 2016; Mahon et al., 2015; Puh & Baer, 2009), and 4 included also healthy participants (Krasovsky et al., 2013; Reisman et al., 2009; Silva et al., 2015; Sousa et al., 2013). The sample sizes of poststroke participants varied from 10 (Puh & Baer, 2009) to 26 participants (Mahon et al., 2015).

A total of 114 poststroke participants were included in the considered studies, totalizing 68 males and 46 females, with mean age of 54.67 years old. Only 2 studies did not characterize anthropometric data of the sample, namely weight and height (Druzbicki et al., 2016; Reisman et al., 2009).

All included participants were in a chronic stage, ranging from 6 months poststroke (Krasovsky et al., 2013) to 16.25 years (Mahon et al., 2015), presenting a single and unilateral episode of stroke.

4 studies presented information about stroke type (Krasovsky et al., 2013; Reisman et al., 2009; Silva et al., 2015; Sousa et al., 2013): 2 studied ischemic and hemorrhagic stroke (Krasovsky et al., 2013; Reisman et al., 2009) and 2 analyzed only ischemic stroke (Silva et al., 2015; Sousa et al., 2013). According to lesion location, 2 studies defined the middle cerebral artery as the lesion area (Silva et al., 2015; Sousa et al., 2013) and 2 described several specific stroke location (Krasovsky et al., 2013; Reisman et al., 2009). 3 studies did not present information about stroke type or lesion location (Druzbicki et al., 2016; Mahon et al., 2015; Puh & Baer, 2009).

All studies were performed on left and right stroke participants, totalizing 65 left side strokes and 49 right side strokes.

To accomplish the understanding of following instructions in the requested tasks, 3 studies used Mini-Mental State Examination to ensure it (Druzbicki et al., 2016; Silva et al., 2015; Sousa et al., 2013). To assess sensorio-motor lower limb impairment, Fugl-Meyer Assessment was applied in 5 studies (Krasovsky et al., 2013; Mahon et al., 2015; Reisman et al., 2009; Silva et al., 2015; Sousa et al., 2013). Other clinical measures were

included in the 2 of the studies, namely Functional Gait Assessment and Activities-Specific Balance Confidence Questionnaire (Krasovsky et al., 2013), Brunnström and Ashworth Modified Scale (Druzbicki et al., 2016). Only 1 study did not identify any clinical measure to accomplish eligibility criteria or sample characterization (Puh & Baer, 2009) (see Table 1).

Walking Conditions

3 studies assessed interlimb coordination during walking on the treadmill (Druzbicki et al., 2016; Krasovsky et al., 2013; Mahon et al., 2015), using different speeds (Krasovsky et al., 2013; Mahon et al., 2015), or self-selected gait speed (Druzbicki et al., 2016). All 3 studies that assessed interlimb coordination during overground walking, used a self-selected gait speed (Reisman et al., 2009; Silva et al., 2015; Sousa et al., 2013). Only 1 study included both treadmill and overground walking at a self-selected gait speed (Puh & Baer, 2009) (see Table 1).

Instruments

To explore interlimb coordination, several instruments were used to assess EMG, kinetics and kinematics data. 4 studies included kinematic analysis (Druzbicki et al., 2016; Krasovsky et al., 2013; Puh & Baer, 2009; Reisman et al., 2009), 1 only kinetic analysis (Mahon et al., 2015), 1 kinetic and EMG analysis (Sousa et al., 2013) and 1 only EMG analysis (Silva et al., 2015) (see Table 2).

To assess kinematic data, three different motion capture systems were used: OPTOTRACK (Reisman et al., 2009), ViconTM (Krasovsky et al., 2013; Puh & Baer, 2009) and Bioengineering Tracker Analyser Soft (BTS) (Druzbicki et al., 2016), with a number of cameras ranging from five (Puh & Baer, 2009) to twelve (Krasovsky et al., 2013). One study did not describe the number of the cameras (Reisman et al., 2009). Reflective markers were located on lower limbs, pelvis and sacrum in 2 studies (Druzbicki et al., 2016; Puh & Baer, 2009), 1 study considered lower limbs, pelvis and shoulder (Reisman

et al., 2009) and 1 study only on feet and hands (Krasovsky et al., 2013). Within each identified segment, different anatomical references were considered, according to each author (see Table 2).

Kinetic data was evaluated through two Bertec force plates (Mahon et al., 2015; Sousa et al., 2013).

To assess EMG activity, a bioPLUX system was applied bilaterally to record data from several lower limb muscles: Soleus, Gastrocnemius medialis, Tibialis Anterior in 2 studies (Silva et al., 2015; Sousa et al., 2013), and Rectus Femoris, Biceps Femoris and Vastus Medialis in 1 study (Sousa et al., 2013) (see Table 2).

Table 2. Instruments, main outcomes, and significant findings to assess interlimb coordination, during gait.

Author, Year	Instruments	Main Outcomes	Significant Findings to Assess Interlimb Coordination
Reisman et al., 2009	Kinematic · OPTOTRACK (Northern Digital, Waterloo, Ontario, Canada) · Infrared emitting diodes: · foot (fifth metatarsal head), ankle (lateral malleolus), knee (lateral joint space), hip (greater trochanter), pelvis (iliac crest), and shoulder (acromion process), bilaterally	Spatiotemporal parameters: · Step length (m) · Double support time (% gait cycle)	Interlimb symmetry of the spatiotemporal parameters
Puh & Bear, 2009	Kinematic · Vicon™ 370 · 5 cameras · Reflective markers: · second sacral vertebra, on the anterior superior iliac spine, the femoral epicondyle, lower lateral surface of the thigh and calf, lateral malleolus, calcaneus, and the second metatarsal head, bilaterally	Spatiotemporal parameters: · Step length (m) · Step time (s) · Stance time (absolute (s) and relative (% gait cycle)) · Single support time (absolute (s) and relative (% gait cycle)) · Double support time (absolute (s) and relative (% gait cycle)) Joint kinematics: · Ankle, knee and hip (o)	Interlimb symmetry of the spatiotemporal parameters using the formula: Interlimb symmetry ratio = $2NHP/(NHP + HP)$ Interlimb symmetry of the mean joint kinematics amplitude
Sousa et al., 2013	Kinetic · 2 force plates (Bertec Corp, 6171 Huntley Rd, Ste J, Columbus, OH 43229; models FP4060–10 and FP4060–08) · Electromyography · bioPLUX (PLUX Wireless Biosignals, Arruda dos Vinhos, Portugal) · Bilateral lower limbs EMG activity: · SOL, GASm, TA, BF, RF and VM	Ground reaction force: · Integral Fy Electromyographic activity: · Magnitude by root mean square (mV)	Percentage of the propulsive impulse = $\int Fy_{trail} / (\int Fy_{trail} + \int Fy_{lead}) \times 100$ Statistic analisys through a correlation coefficient: Relation between contralesional limb electromyography and ipsilesional limb propulsive/braking impulse

Krasovsky et al., 2013	Kinematic · Vicon™ 512 · 12 cameras · Reflective markers: · back of the heel, on the forefoot (head of the 2nd metatarsal), and on the hand (head of 2nd metacarpal), bilaterally	Temporal parameters: · Double support time (s)	Interlimb symmetry of the double support duration
Mahon et al., 2015	Kinetic · 2 force plates (Bertec Corporation, Columbus, Ohio)	Ground reaction force: · Integral Fy Spatial parameters: · Step length	Individual limb mechanical power on the CoM Propulsive impulse = Integral of positive Fy over a complete stride (HP and NHP) Paretic propulsion = $PI_{HP} / (PI_{HP} + PI_{NHP})$ Step length asymmetry = $\frac{Max_{NHP} + Max_{HP}}{(NHP + HP)}$
Silva et al., 2015	Electromyography · bioPLUX (PLUX Wireless Biosignals, Arruda dos Vinhos, Portugal) · Bilateral activity of the ankle agonist muscles: · TA in the lead limb and GASm, SOL in the trail limb	Electromyographic activity: · Magnitude by root mean square (mV)	Ankle antagonist coactivation (%) = $\frac{(\text{antagonist activity} / (\text{agonist} + \text{antagonist activity})) \times 100}{}$
Družbicki et al., 2016	Kinematic · Bioengineering Tracker Analyzer Software · 6 infrared cameras · Reflective markers (internal protocol of the system Helen Hayes (Davis) Marker Placement): · sacrum, pelvis (anterior posterior iliac spine), femur (lateral epicondyle, greater trochanter and in lower 1/3 of the shank), fibula (lateral malleolus, lateral condyle and in lower 1/3 of the shank), and foot (metatarsal head and heel), bilaterally	Spatiotemporal parameters: · Step length (m) · Stance time (s) · Swing time (s) · Joint kinematics: · Knee and hip (o)	Interlimb symmetry index for spatiotemporal parameters and mean joint kinematics amplitude: = $\frac{[(\text{paretic} - \text{non-paretic}) / 0.5(\text{paretic} + \text{non-paretic})] \times 100\%}{}$ Interlimb symmetry ratio: = paretic/non-paretic of the spatiotemporal parameters and mean joint kinematics amplitude

Main Outcomes and Significant Findings to Assess Interlimb Coordination, during Gait

Regarding kinematics, 4 studies explored symmetry of spatiotemporal parameters namely step length (Družbicki et al., 2016; Puh & Baer, 2009; Reisman et al., 2009), double support time (Krasovsky et al., 2013; Puh & Baer, 2009; Reisman et al., 2009), stance time (Družbicki et al., 2016; Puh & Baer, 2009), step time and single support time (Puh & Baer, 2009), and swing time (Družbicki et al., 2016). Only one of these considered interlimb symmetry ratio through the above spatiotemporal parameters (Puh & Baer, 2009). Joint kinematics were studied by 2 authors considering the range of motion of hip and knee joints (Družbicki et al., 2016; Puh & Baer, 2009) and ankle joint (Puh & Baer, 2009) (see Table 2).

2 studies defined as kinetic outcome the integral of the anteroposterior component GRF, using different strategies to calculate propulsive impulse (Mahon et al., 2015; Sousa et al., 2013). This impulse allowed to calculate the paretic propulsion in 1 study (Mahon et al., 2015). Still in kinetic context, 1 study referred the external mechanical power and step length asymmetry (Mahon et al., 2015) (see Table 2).

Concerning EMG data, 2 studies contemplated the magnitude of the muscle activity through root mean square analysis (Silva et al., 2015; Sousa et al., 2013). This measure was used to be correlated with the propulsive impulse kinetic variable in 1 study (Sousa et al., 2013), and to calculate antagonist coactivation ratio in another (Silva et al., 2015) (see Table 2).

Discussion

This scoping review gathered the literature that analyzed methods/ strategies to assess interlimb coordination during gait on poststroke adults. Specifically, the motor task, the main instruments and the related outcomes have been analyzed.

This information will allow a better understanding of the mechanisms underpinning interlimb coordination and its role in gait performance and rehabilitation, while its assessment, promoting a reflection on key information to consider in future research.

Walking Conditions

According to the present review, interlimb coordination during gait was assessed in both treadmill (Drużbicki et al., 2016; Krasovsky et al., 2013; Mahon et al., 2015; Puh & Baer, 2009) and overground walking (Puh & Baer, 2009; Reisman et al., 2009; Silva et al., 2015; Sousa et al., 2013). Because there are different task demands and characteristics between treadmill and overground walking, these should be considered in the analysis of interlimb coordination.

Despite providing valuable information about interlimb coordination as a result of better monitoring conditions, walking on the treadmill may have some differences when comparing to individuals self-selected speed, in their natural walking (Carpinella et al., 2010). While on a treadmill the surface is moving without changes in environment, static surface overground requires constant adaptation to it (Reisman et al., 2009; Riley et al., 2007).

Kinematic and kinetic parameters (Lee & Hidler, 2008; Riley et al., 2007) and muscle activity (Lee & Hidler, 2008) were described with small differences when analyzed between treadmill and overground gait on healthy participants, suggesting qualitatively and quantitatively similarity. These small differences are generally within the normal variability of gait parameters (Riley et al., 2007), showing a quite similar behavior (Lee & Hidler, 2008).

Puh and Baer (2009) suggested with their study, on poststroke participants, that the treadmill training may be a useful tool to some areas of rehabilitation, but it cannot replace gait training overground. It may, however, be useful to use treadmill to focus on improving specific deficits, if combined with overground walking to ensure transferability of an improved gait pattern (Puh & Baer, 2009), like an improvement on step length asymmetry, being more energetically advantageous (Awad et al., 2015). From a therapeutic perspective, this seems to suggest that training individuals with neurological injuries on a treadmill appears to be justified (Lee & Hidler, 2008).

Despite similarities overlap differences between treadmill and overground walking, interlimb coordination assessment, regarding poststroke participants, seems to be more suitable on overground context, considering individual previous accommodation and possible behavior changes on gait parameters when using treadmill. Overground walking may interfere less with natural characteristics of disabled participants (Reisman et al., 2009).

Instruments

To evaluate kinematic data, three optoelectronic systems were used: OPTOTRACK (Reisman et al., 2009), ViconTM (Krasovsky et al., 2013; Puh & Baer, 2009) and BTS (Druzbicki et al., 2016). These measurement systems seem to be the most accurate ones and are described in literature as the gold standard in motion capture (Corazza et al., 2010).

Accuracy of the systems depends on the cameras locations relative to each other, the distance between the cameras and the markers, the position, number, and type of the markers, and the motion of the markers within the capture volume (Maletsky et al., 2007), being this one dependent on the maximum number of cameras and the field of view of each camera (van der Kruk & Reijne, 2018).

Concerning the number of cameras, there were identified five (Puh & Baer, 2009) to twelve (Krasovsky et al., 2013). Knowing that less cameras can interfere markers gaps, using a smaller number might be seen as a disadvantage (Panjkota et al., 2009).

The optoelectronic systems found included active and passive marker systems. ViconTM (Krasovsky et al., 2013; Puh & Baer, 2009) and BTS (Druzbicki et al., 2016) passive motion capture systems use markers that reflect light back to the sensor. Active marker optical system, like OPTOTRACK (Reisman et al., 2009), utilize markers that contain the source of light for the sensors (often infrared) (Richards, 1999). The benefit of active markers over passive ones could be the robustness of the measurements. However, active markers

require additional cables and batteries, so the freedom of movement is more limited (Stancic et al., 2013).

Different models for motion capture were found, with differences in marker sets and protocols that details different number and position of markers. This fact can interfere in a large effect on the kinematic angles generated (Mantovani & Lamontagne, 2017). One of the biggest challenges in marker-based motion capture is maintaining accuracy and precision. Appropriate models for one's specific research purpose and a single examiner precisely placing the markers, may be important factors to maintain these assumptions (Della Croce et al., 2005).

Overall, to ensure the accuracy of the instruments, it seems important a description of all parameters worrying their well use. Also, to increase the robustness of the data and to keep up with technological developments, researchers could consider a prior test to experiment protocol and improve the most precise setups.

To evaluate kinetic data, two Bertec force plates were identified (Mahon et al., 2015; Sousa et al., 2013). A force plate measures the GRF over time during static or dynamic motions such as gait. They provide information on the three-dimensional components of force (X, Y and Z), which can then be used to look at more in-depth joint loadings (Garza-Ulloa, 2018). They are considered the gold standard model, with high accuracy, to determine spatiotemporal gait parameters related to GRF characteristics (Garza-Ulloa, 2018).

In this review, two authors resorted surface EMG to assess muscle activity during a dynamic task (Silva et al., 2015; Sousa et al., 2013). Surface EMG provides a non-invasive, global measurement of muscle activity, which may be more suitable for applications in movement analysis that require frequent assessments or information on the patterns of activation of multiple muscles (Agostini et al., 2020; McManus et al., 2020; Papagiannis et al., 2019). Careful skin preparation and choice of electrode type, electrode placement and recording configuration, including filter settings and amplifier gain, are essential to record high quality EMG signals with low noise (Agostini et al., 2020; McManus et al., 2020; Papagiannis et al., 2019).

To record this muscle activity, a bioPlux device was used (Silva et al., 2015; Sousa et al., 2013). The bioPlux is a device that collects and digitizes the signals from the electromyographic sensors transmitting them to a computer, where they are viewed in real time (PLUX wireless biosignals S.A., biosignalsplux wearable body sensing platform, 2013). Despite its usefulness, the connection between the sensors and the equipment is ensured by cables that might interfere in movement freedom during the dynamic assessment. Nowadays, different type of solutions could be considered, using wireless technology, providing better data quality.

Several muscles were identified to evaluate interlimb coordination. The lower limb distal muscles were the most assessed ones in the included studies (Silva et al., 2015; Sousa et al., 2013). The number and selection of the assessed muscles might be justified by the number of channels available on the equipment used, as well as the specific research purpose of each study.

It is suggested that surface EMG combined with kinematic and kinetic data is a useful tool for decision making of the appropriate method needed.

Main Outcomes and Significant Findings to Assess Interlimb Coordination, during Gait

To assess the interlimb coordination, several outcomes have been considered including kinematics, kinetics, and EMG data.

A higher variety of kinematic outcomes was found, and the most pointed were related to spatiotemporal parameters as step length (Druzbicki et al., 2016; Puh & Baer, 2009; Reisman et al., 2009) and double support time (Krasovsky et al., 2013; Puh & Baer, 2009; Reisman et al., 2009). Literature reports that asymmetry of step length following stroke is related to decreased propulsive force of the contralesional limb, decreased work, and power of the contralesional plantarflexors and decreased walking speed (Balasubramanian et al., 2007). Moreover, asymmetric double support time is related to decreased speed (Olney et al., 1994). In fact, according to the theoretical model in biomechanical study proposed by Kuo (2007), the double support corresponds to step-

to-step transition, and this transition from one stance limb inverted pendulum to the next appears to be the major determinant of the mechanical work of walking. Considering the aforementioned, the transition between steps reaches an optimum level when the propulsion of the trailing and the initial contact loading response of the leading have the same magnitude and short duration (Kuo, 2007). Thus, step length and double support time symmetry between both lower limbs seem to allow a better understanding on interlimb coordination behavior.

Considering these assumptions, some authors also propose to combine the study of joint kinematics symmetry with spatiotemporal parameters, to improve this knowledge (Druzbicki et al., 2016; Puh & Baer, 2009).

Regarding kinetic analysis, different outcomes were identified in this review: anteroposterior component GRF, to study the propulsive impulse (Mahon et al., 2015; Sousa et al., 2013), and the external mechanical power (Mahon et al., 2015). Despite some symmetry is required between lower limbs, the studies that evaluated the CoM mechanical work of leading and trailing showed that the trailing needs to generate more work than the leading one (A. S. P. Sousa, Silva, & Tavares, 2013), probably through feedforward mechanism expressed through a lag between the peak of anteroposterior GRF trailing and leading limbs (Sato & Yamada, 2017). According to the theoretical model, about 75% of the energy loss during leading limb heel strike is compensated by the trailing limb during the propulsion in the double support phase (Kuo, 2002, 2007).

Concerning EMG outcomes, muscle activity magnitude was expressed by root mean square (RMS) (Silva et al., 2015; Sousa et al., 2013), allowing to study the correlation between lower limbs muscle activity and propulsive impulse (Sousa et al., 2013). This is supported by a previous study, on healthy subjects, showing a correlation between the EMG activity of the trailing limb and the magnitude of the leading anteroposterior GRF, during step-to-step transition, finding a higher correlation between them, which corroborates its major importance in forward propulsion (Sousa et al., 2012).

Another outcome identified in this study results was the antagonist coactivation ratio (Silva et al., 2015), showing that the dynamic relationship between limbs during walking

may also be analyzed through the levels of the antagonist coactivation ratio considering the functional position of each limb (trailing vs. leading) and the subsequent role of each.

Considering the exposed, it is suggested to assess interlimb coordination resorting to more than one outcome, through the combination of the different EMG, kinetic and kinematic data.

Despite the importance to summarize the main parameters to consider in the interlimb coordination, it is not only enough the identification of the better outcomes to consider. It is also important to understand their behavior and to know their psychometric characteristics to find the better ones to study interlimb coordination and allowing to monitor clinical improvements.

Conclusions

The interlimb coordination assumes a crucial role to meet the demands of postural and movement control. The knowledge gathered from this review was useful to understand which motor task main instruments and outcomes has been considered.

Since gait is a functional task from the perspective of rehabilitation, natural auto-selected walking should be pondered. Kinematic, kinetic and EMG activity were instruments found, allowing to study: spatiotemporal parameters and joint kinematics; anteroposterior GRF and propulsive impulse; RMS and antagonist coactivation ratio; respectively, improving a better understanding of the mechanisms supporting interlimb coordination assessment and its role in gait performance and rehabilitation.

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ARTIGO II

Repeatability and Temporal Consistency of Lower Limbs Biomechanical Variables Expressing Interlimb Coordination during Double-Support Phase in People with and without Stroke Sequelae

Ana G. B. Couto ^{1,2,3,4}, Mário A. P. Vaz ^{5,6}, Liliana Pinho ^{2, 7,8}, José Félix ^{2,9,10}, Juliana Moreira ^{2,11}, Francisco Pinho ^{7,12}, Inês Albuquerque Mesquita ^{2,4,13}, António Mesquita Montes ^{1,4,11}, Carlos Crasto ^{1,4,11} and Andreia S. P. Sousa ¹¹

1 Department of Physiotherapy, Santa Maria Health School, 4049-024 Porto, Portugal

2 Centre for Rehabilitation Research (CIR), School of Health of Polytechnic Institute of Porto, 4200-072 Porto, Portugal

3 Faculty of Engineering, University of Porto, 4200-465 Porto, Portugal

4 Research Centre and Projects (NIP), Santa Maria Health School, 4049-024 Porto, Portugal

5 Institute of Mechanical Engineering and Industrial Management, Faculty of Engineering, University of Porto, 4200-465 Porto, Portugal

6 Porto Biomechanics Laboratory (LABIOMEP), University of Porto, 4200-450 Porto, Portugal

7 College of Health Sciences—Escola Superior de Saúde do Vale do Ave, Cooperative for Higher, Polytechnic and University Education, 4760-409 Vila Nova de Famalicão, Portugal

8 Faculty of Sport, University of Porto, 4200-450 Porto, Portugal

9 Department of Physics, School of Health of Polytechnic Institute of Porto, 4200-072 Porto, Portugal

10 Department of Medical Sciences, University of Aveiro, 3810-193 Aveiro, Portugal

11 Department of Physiotherapy, School of Health of Polytechnic Institute of Porto, 4200-072 Porto, Portugal

12 Human Movement Unit (H2M), Cooperative for Higher, Polytechnic and University Education, 4760-409 Vila Nova de Famalicão, Portugal

13 Department of Functional Sciences, School of Health of Polytechnic Institute of Porto, 4200-072 Porto, Portugal

Couto, A.G.B., Vaz, M.A.P., Pinho, L., Félix, J., Moreira, J., Pinho, F., Mesquita, I. A., Mesquita Montes, A., Crasto, C., & Sousa, A.S.P. (2023). Repeatability and Temporal Consistency of Lower Limbs Biomechanical Variables Expressing Interlimb Coordination during Double-Support Phase in People with and without Stroke Sequelae. *Sensors*, 23, 2526. <https://doi.org/10.3390/s23052526>

Abstract

Reliable biomechanical methods to assess interlimb coordination during the double-support phase in post-stroke subjects are needed for assessing movement dysfunction and related variability. The data obtained could provide a significant contribution for designing rehabilitation programs and for their monitorisation. The present study aimed to determine the minimum number of gait cycles needed to obtain adequate values of repeatability and temporal consistency of lower limb kinematic, kinetic, and electromyographic parameters during the double support of walking in people with and without stroke sequelae. Eleven post-stroke and thirteen healthy participants performed 20 gait trials at self-selected speed in two separate moments with an interval between 72 h and 7 days. The joint position, the external mechanical work on the centre of mass, and the surface electromyographic activity of the tibialis anterior, soleus, gastrocnemius medialis, rectus femoris, vastus medialis, biceps femoris, and gluteus maximus muscles were extracted for analysis. Both the contralesional and ipsilesional and dominant and non-dominant limbs of participants with and without stroke sequelae, respectively, were evaluated either in trailing or leading positions. The intraclass correlation coefficient was used for assessing intra-session and inter-session consistency analysis. For most of the kinematic and the kinetic variables studied in each session, two to three trials were required for both groups, limbs, and positions. The electromyographic variables presented higher variability, requiring, therefore, a number of trials ranging from 2 to >10. Globally, the number of trials required inter-session ranged from 1 to >10 for kinematic, from 1 to 9 for kinetic, and 1 to >10 for electromyographic variables. Thus, for the double support analysis, three gait trials were required in order to assess the kinematic and kinetic variables in cross-sectional studies, while for longitudinal studies, a higher number of trials (>10) were required for kinematic, kinetic, and electromyographic variables.

Keywords

Post-stroke; gait; human movement variability; kinematic parameters; kinetic parameters; electromyographic parameters; test–retest reliability

Introduction

In 2019, stroke remained the third leading cause of death and disability in the world (Feigin et al., 2021). The sensorial and motor repercussions of a stroke can cause limitations in activities of daily living and participation restrictions, both in the professional and social context (Rajsic et al., 2019). Gait is one of the most frequent altered functional tasks in this pathological condition, with a great impact on independence; 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; A. S. P. Sousa et al., 2013).

The main changes in post-stroke gait are mainly expressed in the asymmetric behaviour of both lower limbs (Arya & Pandian, 2014; Olney et al., 1994; Olney et al., 1991; Olney & Richards, 1996; A. S. P. Sousa et al., 2013; Sousa & Tavares, 2015). From a biomechanical point of view, these changes have been extensively explored and described (Balaban & Tok, 2014; Chen et al., 2005; Kim & Eng, 2004; Lamontagne et al., 2002; Lamontagne et al., 2000; Lamontagne et al., 2007; Nadeau et al., 2013; Verma et al., 2012; Woolley, 2001) at a kinematic level (joint position and double support phase) (Oken & Yavuzer, 2008), kinetic (ground reaction forces) (Bensoussan et al., 2006; Moseley et al., 1993) and muscle activation (distal and proximal muscles) (Daly et al., 2011; Den Otter et al., 2007), on both lower limbs.

To improve the understanding of the underlying mechanisms of post-stroke motor recovery and to differentiate compensation and restitution, the Stroke Recovery and Rehabilitation Roundtable (SRRR) (Kwakkel et al., 2017) recommended, in 2017, the inclusion of kinematic and kinetic analysis in studies on post-stroke motor recovery. However, the recommendation of specific measures and methods for this analysis is challenging, and the same authors, in 2019, published a consensus of recommendations for specific upper limb tasks. In this consensus, a recommendation of 15 repetitions was purposed to achieve good repeatability (Kwakkel et al., 2019). By contrast, Frykberg et al. (2021) observed that it took 2 to 3 repetitions to reach good inter-repetition (intra-session) performance stability, and at least 3 inter-session repetitions for upper limb kinematic measurements in people with and without stroke sequelae.

Although there are no recommendations of the SRRR for the lower limb, this entity reported that the motor performance and movement quality could be analysed by coordination measures to establish the (as)symmetry between the contralesional (CONTRA) and ipsilesional (IPSI) sides, such as the spatiotemporal parameters, the ground reaction forces and their torques, and the centre of mass displacement (Kwakkel et al., 2017). Specifically in the double support phase, Couto et al. (2022) identified some methodological considerations for the analysis, namely the time of double support phase, the external mechanical work performed by the lower limbs in the step-to-step transition, the muscle activity developed by the lower limbs and the coactivation ratio, considering the functional position and the role of each lower limb in this gait phase. Although variables can be used to calculate indices and ratios that show (as)symmetry, the combination of individual analyses also seems to be recommended (Couto et al., 2022).

Beyond the detailed and objective information on the motor performance and movement quality provided by biomechanical gait analysis, another aspect that is important to be considered is the movement variability. This characteristic is inherent to the human movement control, and different neuromotor processes are available to produce automatic movement strategies, which allow to achieve the same motor purpose (Stergiou & Decker, 2011; Stergiou et al., 2006). Optimal movement variability is crucial for a healthy motor control (Stergiou & Decker, 2011) and a high level of automatism, with a relatively constant variability, that can be expected when a well-known task is performed repeatedly (Carson et al., 2014). As mentioned above, one of the aspects that shows the movement variability, and still not well investigated is the performance stability of biomechanical variables, observed by the number of trials necessary to achieve this stability. As far as the minimum number of trials is concerned, to the best of our knowledge, there are no studies in this population for gait analysis so it is relevant to analyse the intra-session and inter-session variability of the kinematic variables (double-support time and position of the hip, knee and ankle joints), kinetics variables (external mechanical work) and electromyographic variables (muscle activity mean values and muscle activation ratios) in the double support phase, in people with and without stroke sequelae.

Methods

Study design

A quantitative methodology, with a test-retest study design was conducted to analyse the performance stability of the kinematic, kinetic and electromyographic variables during the double support phase of gait, in people “with stroke sequelae”, named in the present study as the “stroke” group, and in people “with no history of a stroke and without self-reported disabilities” named as the “healthy” group. The performance stability over a set of multiple trials intra-session (between-repetition analysis) and inter-session was analysed. For this purpose, all participants performed 20 gait trials at a self-selected speed, in two different moments.

Participants

A group of eleven subjects (3 females and 8 males) with history of a single unilateral ischemic 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 group of thirteen healthy subjects (4 females and 9 males) (Table 1). Participants were excluded from the healthy group if they had an altered mental state with interference in communication and cooperation (Lamontagne et al., 2002; Lamontagne et al., 2000; A. S. P. Sousa et al., 2013), history or sign of neurological dysfunction (Sousa et al., 2012), presence of pain that interfered with the performance of walking (A. S. Sousa et al., 2013), history of osteoarticular or musculotendinous injury, history of lower limb surgery in the last 6 months, anatomical deformities of the lower limb (A. S. Sousa et al., 2013; Sousa et al., 2012), taking medication that interferes with the motor performance of the lower limb (Lamontagne et al., 2002; Lamontagne et al., 2000), moderate-intensity (i.e., at least 30 minutes 5 days a week) or vigorous-intensity aerobic physical activity (i.e., at least 20 minutes 3 days a week) (Thompson, 2014).

To be included in the stroke group the participants need to fulfil the following criteria: diagnosis of ischemic first-ever stroke with at least 6 months of evolution (Lamontagne

et al., 2002; Lamontagne et al., 2000; A. S. P. Sousa et al., 2013), involving middle cerebral artery territory, at the subcortical level, confirmed by Axial Computed Tomography (Lamontagne et al., 2002; A. S. P. Sousa et al., 2013); lower limb sensorimotor impairment (A. S. P. Sousa et al., 2013); ability to walk at least 10 meters, with close supervision if necessary, but without physical assistance (Lamontagne et al., 2000; A. S. P. Sousa et al., 2013). In turn, people who had any of the exclusion criteria mentioned for the healthy group, or the presence of a lesion involving the brainstem and/ or cerebellum were excluded (Lamontagne et al., 2002; Lamontagne et al., 2000; A. S. P. Sousa et al., 2013).

Table 1. Participants characteristics (sociodemographic and clinical). Data presented from the mean (standard deviation). p value reflects a comparison between the stroke and healthy groups.

	Stroke (n=11)	Healthy (n=13)	Between-groups comparison
	Mean (sd)	Mean (sd)	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
Double support time (s)			
Ipsilesional/ Dominant	0,29 (0,31)	0,20 (0,03)	0,313
Contralesional/ Non-Dominant	0,24 (0,10)	0,19 (0,03)	0,162
Post-stroke time (months)	52,18 (29,89)	---	---

Instruments

Sample selection and characterization

A questionnaire was used to verify the inclusion and exclusion criteria defined for the selection of participants, and to characterize the sample, namely age, sex, dominance, and time of evolution in the stroke group. This questionnaire included the Brief Physical Activity Assessment Tool (Cruz et al., 2021) to verify the people's level of physical activity.

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 the person has cognitive deficit depending on the level of schooling: score ≥ 15 for illiterates, ≥ 22 for people with 1 to 11 years of schooling, or ≥ 27 for people with more than 11 years of schooling (Guerreiro, 1994). This instrument has been adapted and validated for the Portuguese population by (Guerreiro, 1998; Guerreiro, 1994), with a sensitivity between 63% and 73.4% and a specificity between 90 and 96.8% (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 the person has sensorimotor impairment of the lower limb if a score < 34 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 by Costa (2003) and Santos (2005) with excellent internal consistency (α Cronbach=0.99) (Costa, 2003).

To characterize the sample, anthropometric measurements were performed, such as body mass (kg) and height (m), using 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 1mm scale.

Kinematic data

The joint position of the hip, knee, and ankle in the sagittal plane, on the dominant (DOM) side in the healthy group/ ipsilesional (IPSI) side in the stroke group, and on the non-dominant (NDOM)/ contralesional (CONTRA) side, respectively, and time of the double support phase (space-time variable) 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 infrared cameras, eight

Oqus 500 and four Miquis M3, connected to the Qualisys USB Analog Acquisition interface, at a sampling frequency of 100Hz.

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 1000Hz. The capture hardware was connected to the Qualisys Motion Capture System analogue board.

Electromyographic data

Surface electromyography (EMG) 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 Gluteus Maximus (GMax), on the DOM side in the healthy group/ IPSI in the stroke group, and on the NDOM/ CONTRA, respectively. 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 EMG signal, at an acquisition frequency of 1000 Hz. The EMG signal was integrated into the Qualisys Motion Capture System, connecting its analogue board to the BNC-2090a board (National Instruments, United States of America). EMGworks software was used to analyse the EMG 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 task. 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 $BMI = \frac{weight}{height^2}$.

For the collection of kinematic data, 46 reflective markers were placed bilaterally in anatomical references (identified by manual palpation) of the participant body: apex of the head, earlobes, jugular notch, xiphoid apophysis, spinous apophysis of the 7th cervical vertebral, lateral part of the acromion, anterior iliac spines superior, posterior superior iliac spines, greater trochanters, lateral and medial epicondyles of femur, anterior tuberosities of tibia; head of the fibula, lateral and medial malleoli, posterior surface of the calcaneus, head of the 1st, 2nd and 5th metatarsals, lateral and medial epicondyle of the humerus, styloid apophysis of the ulna and radius, 2nd and 5th metacarpals (C-Motion, United States of America) (Cappozzo et al., 1995). These markers would allow the built of a full-body biomechanical model in the Visual 3D software.

To collect the EMG signal, the electrodes were placed over the muscles under analysis followed the recommendation of the Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000) and the study by A. S. P. Sousa et al. (2013) (Table 2). Electrode placements were confirmed by palpation of the muscle belly and muscle contraction with the subject in the test

position. Before placing the electrodes, the participant's hair was shaved, an abrasive cream was used to remove dead cells from the skin's surface, and the skin was cleaned with isopropyl alcohol (70%) to remove oil and remaining dead cells. The electrode impedance checker was used to ensure that impedance levels were less than 5 K Ω , which was considered to signify the satisfactory acquisition of the EMG signal (Hermens et al., 2000). Finally, the electrodes were placed parallel to the orientation of the underlying muscle fibers, 5 minutes after skin preparation.

Table 2. Anatomical references for the electrode placement (Hermens et al., 2000; A. S. Sousa et al., 2013).

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 degrees 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
Gluteus maximus	50% on the line between the sacrum and the greater trochanter

The collections were carried out in two moments, with an interval of 72 hours to 7 days between them. This interval was defined according to the literature for studies with similar samples (Cho et al., 2015; Fotiadou et al., 2018; Geiger et al., 2019).

Task

Participants were asked to walk for 10 meters, 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 (A. S. Sousa et al., 2013; A. S. P. Sousa et al., 2013).

The participants performed 10 valid trials with the DOM/ IPSI lower limb in the TRAIL position and 10 valid trials with the NDOM/ CONT lower limb in the TRAIL position (Fotiadou et al., 2018), with a resting time of two-minutes between trials to prevent fatigue (A. S. Sousa et al., 2013; A. S. P. Sousa et al., 2013).

For the kinetic variable, the trials were considered valid if the 1st lower limb made full contact on the 1st platform – TRAIL, and the 2nd lower limb also made full contact on the 2nd platform – LEAD (A. S. Sousa et al., 2013).

Data processing

Kinematic data processing was carried out using the Qualisys Track Manager software, with the aim of recognizing the trajectory of the markers and verifying whether they were correctly identified during data collection. Path deviations/ interruptions were interpolated using the linear, polynomial, and relational calculations built into the software. Subsequently, the resulting data were 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 and toe-off events of the NDOM/ CONTRA and DOM/ IPSI limb. The double support phase was identified as the interval between heel strike of the LEAD limb and toe-off of the TRAIL. To identify these events, the methods proposed by Banks et al. (2015) for the heel strike and by Zeni et al. (2008) for the toe-off were considered. The Banks et al. (2015) method refers to the horizontal distance between the calcaneus and the ankle, and the heel strike is determined through the maximum displacement between the marker on the ipsilateral calcaneus and the marker on the contralateral lateral malleolus (Banks et al., 2015; French et al., 2020). The Zeni et al. (2008) method refers to the horizontal distance between the sacrum and the calcaneus, and the toe-off is determined through the minimum displacement between the calcaneal marker and the sacral marker (French et al., 2020; Zeni et al., 2008). Regarding the position of the hip and

knee joints, in the sagittal plane during each of the events, the positive variation of the values refers to flexion and the negative variation to extension. For the position of the ankle joint, the positive variation of the values refers to plantar flexion and the negative variation to dorsal flexion. The identification of the heel strike and toe-off events also allowed extracting the spatiotemporal parameter relative to the time of the double support phase of the gait (time difference between the two events).

Kinetic data processing was carried out by calculating the WCOM of each lower limb during the double support phase. The power generated on the body's centre of mass (external mechanical power) by a limb is equal to the dot product of the external force acting on the limb and the velocity of the centre of mass (Donelan et al., 2002). 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 (normalized 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 member was computed by calculating the mechanical energy integral of each member in the double limb support phase.

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

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

The electromyographic data processing was carried out using Matlab software, version 3.9.0. (MathWorks, United States of America). A second-order digital filter Infinite Impulse Response (Butterworth) – one at 20 Hz (high pass) and the other at 450 Hz (low pass) was applied to the EMG to remove electrical noise and sensor movement. Subsequently, the root mean square (RMS) value was calculated using a moving average window set to 100 samples (Lamontagne et al., 2000). The percentage of activation intensity of each muscle under analysis was obtained by normalizing the EMG signal by calculating the quotient between the mean RMS of each muscle during the double

support phase and the maximum RMS of the same muscle during the gait cycle, multiplied by 100 (A. S. Sousa et al., 2013; Sousa & Tavares, 2012). In turn, the coactivation ratio was calculated according to the following equation (Kellis et al., 2003):

$$\text{Coactivation ratio} = \frac{\text{agonist activity}}{\text{agonist activity} + \text{antagonist activity}}$$

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 role of each agonist muscle in the double support phase was considered for each of the events. For the ankle coactivation ratio, in the LEAD position, TA was considered agonist, and in the TRAIL position, SOL and GasM were considered. In the knee ratio, RF and VM were considered agonists, in both positions, in LEAD and TRAIL. For the hip coactivation ratio, in the LEAD position, RF was considered agonist, and in the TRAIL position, GMax and BF were considered. 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's 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 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's test was used.

Mean and standard deviation were used as descriptive statistics for quantitative variables, and absolute and relative frequency for qualitative variables of sample characterization (gender and injured/ dominant side).

Intra-session and inter-session performance stability was verified by analysing the intraclass correlation coefficient (ICC). ICC values greater than 0.75 were considered good to excellent (Koo & Li, 2016). To determine the number of trials necessary to reach a good level of reliability ($ICC > 0.75$), the ICC values were calculated for each variable based on n consecutive trials ($n=2$ to $n=10$ for inter-trials comparison and $n=1$ to $n=10$ for inter-session comparison). When in a variable the ICC did not reach values greater than 0.75, in any of the comparisons, then the number of trials necessary to reach a good level of reliability was classified as “ >10 ”. For ICC processing, the absolute agreement was calculated for an average of measurements for all comparisons, except for the ICC referring to the comparison between-sessions of only one repetition. In this case, an ICC of absolute agreement was used for a measurement (Marôco, 2018).

Results

The number of trials required to achieve intra-session and inter-session ICC values higher than 0.75 for kinetic and kinematic variables is presented in Table 3. It can be observed that, the number of trials needed in each session was generally low, ranging between 2 and 3 trials for most of the analysed variables, while between sessions the values ranged from 1 to >10 trials. In some variables more trials are required, as observed in TRAIL, namely hip joint position in both limbs in both groups, knee joint position in both limbs in healthy group, and in LEAD namely hip joint position in both limbs in stroke group, in CONTRA knee joint position in stroke group and in ankle joint position in both limbs in the healthy group. These variables with low temporal consistency do not seem to show a clear pattern regarding differences between healthy and stroke groups nor between IPSI/ DOM and CONTRA/ NDOM limbs. However, in the LEAD position it can be observed by the analysis of Table 3 that a higher number of trials for the hip position and a lower number for the ankle position are needed in the stroke group, contrasting with an opposite behaviour in the healthy group.

The double support phase time and WCOM were consistent intra-session and inter-session in both, healthy group and in the stroke group, with a number of trials required ranging from 1 to 3. In exception to this, the WCOM of the CONTRA in TRAIL position intra-session required 7 trials, and the WCOM of the DOM in LEAD position inter-session required 9 trials.

Table 3. Number of trials needed to reach good performance stability intra-session and inter-session ($ICC > 0.75$) of the kinematic variables (hip, knee, and ankle joint position) and kinetics (work on the centre of mass) with the dominant lower limb (healthy)/ ipsilesional (stroke) in the TRAIL and LEAD positions.

	Intra-session				Inter-session			
	Healthy		Stroke		Healthy		Stroke	
	DOM	NDOM	IPSI	CONTRA	DOM	NDOM	IPSI	CONTRA
Joint position								
TRAIL								
Hip	2	2	2	3	>10	>10	6	>10
Knee	2	2	2	2	>10	5	2	3
Ankle	2	2	3	2	1	2	4	2
LEAD								
Hip	2	2	2	2	2	2	>10	>10
Knee	2	2	2	3	2	1	2	>10
Ankle	2	4	2	2	5	>10	2	2
Time								
Double support	2	2	2	2	1	1	1	1
WCOM								
TRAIL	2	2	2	7	2	2	3	1
LEAD	3	2	2	2	9	2	2	2

Number of trials to reach $ICC > 75\%$; >10 (without reaching the goal)

The number of trials required to intra-session and inter-session ICC values higher than 0.75 for electromyographic variables is presented in Table 4. As in the kinematic variables, the number of trials necessary to reach an intra-session $ICC > 0.75$ is, in general, lower compared to inter-session values. However, in both, the variability in most electromyographic variables seems to be higher compared to the kinematics and kinetics, with many variables in which, even with 10 trials, an $ICC > 0.75$ is not reached. Intra-session the trials required ranged from 2 to >10, while inter-session ranged from 1 to >10.

Despite there is not a clear pattern in the number of trials required between groups, nor between sides it can be observed by the analysis of Table 4 that when the CONTRA and the IPSI limbs are in the LEAD and TRAIL positions, respectively, a smaller number of trials is needed intra-session compared to the NDOM and DOM sides in the LEAD and TRAIL position, respectively, in the healthy group. When the limbs assume opposite positions, the same does not happen, being verified, in most variables, an increase in the number of trials in both groups.

Intra-session, the muscles that seem to need a smaller number of trials (2) are in the stroke group, namely the SOL and the GasM, and the ankle ratio, the VM and the hip and knee ratio of the CONTRA limb in LEAD. In TRAIL, can also be highlighted the GasM in both limbs, SOL, and the ankle ratio of the IPSI, the RF and BF of the IPSI, the GMax in the CONTRA, and the hip ratio of the CONTRA and the knee of the IPSI.

Through the analysis of Table 4, it can also be observed that less trials seem to be required in CONTRA muscle activity between sessions compared to healthy group and the IPSI limb. This was more evident in RF and GMax activity (either in LEAD or TRAIL position) compared to the IPSI side and even to the healthy group.

Table 4. Number of trials needed to reach good performance stability of intra-session and inter-session (ICC>0.75) of the electromyographic activity of muscles, Tibialis Anterior (TA), Soleus (SOL), Gastrocnemius Medialis (GasM), Rectus Femoris (RF), Biceps Femoris (BF), Vastus Medialis (VM) and Gluteus Maximus (GMax) and the hips, knee, and ankle coactivation ratio with the dominant lower limb (healthy)/ ipsilesional (STROKE) in the positions of TRAIL and LEAD.

	Intra-session				Inter-session			
	Healthy		Stroke		Healthy		Stroke	
	DOM	NDOM	IPSI	CONTRA	DOM	NDOM	IPSI	CONTRA
LEAD								
Ankle								
TA	>10	>10	10	7	>10	>10	>10	4
SOL	8	7	7	2	2	>10	>10	5
GasM	8	6	5	2	8	3	>10	>10
Ankle ratio	7	4	7	2	5	2	>10	>10
Knee and Hip								
RF	10	>10	7	7	>10	>10	>10	2
VM	5	6	7	2	3	>10	5	>10
BF	5	>10	4	8	6	>10	>10	>10
GMax	7	>10	>10	3	>10	>10	>10	2
Hip ratio	8	7	7	2	>10	>10	>10	1
Knee ratio	8	>10	9	2	4	>10	>10	>10
TRAIL								
Ankle								
TA	7	3	10	5	1	>10	4	>10
SOL	8	>10	2	7	4	>10	>10	8
GasM	>10	>10	2	2	10	>10	1	>10
Ankle ratio	5	7	2	>10	>10	7	>10	>10
Knee and Hip								
RF	9	7	2	8	>10	>10	>10	2
VM	6	10	6	>10	2	>10	>10	>10
BF	7	8	2	7	>10	>10	>10	>10
GMax	6	6	>10	2	3	4	>10	1
Hip ratio	5	7	3	2	>10	>10	>10	1
Knee ratio	7	7	2	4	>10	>10	>10	>10

Number of trials to reach ICC>75%; >10 (without reaching the goal)

Discussion

According to our knowledge, this study is the first one determining the minimum number of trials needed to reach a good performance stability, both intra- and inter-session, of biomechanical variables reflecting interlimb coordination (kinematic, kinetic and electromyographic) during the double support phase of gait, in people with and without

stroke sequelae. In post-stroke people, movement strategies seem to be more stereotyped, therefore less variable, and can be justified by the development of compensatory strategies, which result in positional fixation during post-stroke recovery (Jones, 2017; Perry et al., 1995), reducing the movement variability. Atypical muscle activation synergies and inadequate interarticular coordination can be suggested as the main causes for the development of these compensatory strategies. However, and globally the results obtained in the present study seem to point to similar intra-session ICC values between post-stroke and healthy people and, which may be related to the characteristics of the functional task assessed, gait, where a lower movement variability is expected (Carson et al., 2014).

Intra-session

The results revealed that intra-session, for most of the kinematic variables and for the kinetic variable studied, only 2 to 3 trials are enough to reach a good performance stability for an $ICC > 0.75$. The same did not happen in electromyographical variables, as a higher number of trials are required. Globally, the results of this study point to lower number of trials required comparing to the ones suggested by previous studies. Fotiadou et al. (2018) and Monaghan et al. (2007) suggested that 10 trials seem to be enough to obtain excellent performance stability for most kinematic and kinetic variables of a gait cycle for the evaluation of both lower limbs, CONTRA and IPSI in gait, in people with stroke sequelae and healthy participants, respectively. The differences between the results of the present study and the previous one could be related to the fact that in this one only the sagittal plane was considered and only the double support phase. This motion plane seems to present a better performance stability compared to the frontal and transverse planes (Fotiadou et al., 2018; Kadaba et al., 1989; Yavuzer et al., 2008). The results of joint kinematics in the sagittal plane, founded in our study, are in agreement with previous studies (Fotiadou et al., 2018; Kadaba et al., 1989; Yavuzer et al., 2008), demonstrating a low variability at a self-selected speed, which suggests that it may be acceptable to base clinical decisions according to the results of a single gait assessment.

The movement performance stability achieved with a lower number of trials for the joint position variables confirms that the movement variability is relatively stable when repeatedly performing a well-known automatic task (Carson et al., 2014), as is the example of walking at a comfortable self-selected speed. Kinematic variables can be considered spatial movement descriptors independent of the forces that generate the movement (Krasovsky & Levin, 2010). Generally, it is assumed that gait kinematics is highly stable intra-session (repeatability) in healthy people (Kadaba et al., 1989), but also in people with stroke sequelae (Yavuzer et al., 2008). The intra-session variability seems to reflect the intrinsic (physiological) variability inherent in gait movement, such as gait velocity and pattern, and soft tissue movement (Schwartz et al., 2004), which being considered an automatic rhythmic movement is characterized by low variability (Terrier et al., 2005).

The low number of intra-session trials for WCOM observed in the present study is in accordance with Caty et al. (2009) that highlighted the clinical importance of this variable. The variability of WCOM of the CONTRA limb of the stroke group in the TRAIL position seems to be in accordance with the evidence that the highest variability occurs in the limb that initiates the swing phase (TRAIL), and this could be explained by the instability produced in the transition of the lower limb from a closed kinetic chain to an open kinetic chain condition (Fotiadou et al., 2018). In the stroke group, when the CONTRA limb is in LEAD position, less variability was observed relative to healthy in both limbs and less variability of the CONTRA limb relative to IPSI in terms of lower limb muscular activity. This finding may point to a movement restriction causing a stereotyped pattern in the stroke group (Fotiadou et al., 2018) when the CONTRA limb is in LEAD position, observed by the lower number of repetitions. On the other hand, the IPSI limb may have to make corrective adjustments in the face of the CONTRA movement pattern, thus increasing the variability of the IPSI (Fotiadou et al., 2018) observed by the higher number of repetitions. The asymmetry between the two positions may be related to the specific role of each limb in each phase, where in the LEAD position the lower limb has a predominantly postural role for the initial contact of the foot with the ground and load response, while in the TRAIL position the lower limb play a propulsive role to initiate the swing phase (Neumann, 2017). A possible explanation could be related to the initial contact (LEAD),

which in the CONTRA limb, being performed with less variability, translates into a more stereotyped movement strategy with less variability in muscle activation patterns, i.e., with a smaller repertoire of muscle activation patterns, and that could somehow influence the activity of the IPSI limb in the TRAIL position. Still, the higher number of repetitions in the CONTRA limb in the TRAIL position is consistent with the higher number of repetitions found in WCOM of that limb in the same position. A possible explanation for this result may be associated with the degree of hip extension in the TRAIL position, corresponding to the toe-off and early swing event, where the hip extension, by the eccentric action of the hip flexors (Neumann, 2017), may be related to the propulsive function of this position. This hypothesis may be studied in future studies.

For electromyographic activity, the GMax in the CONTRA limb in both positions, showed a lower variability, with only 1 to 3 trials being needed for the stability in intra-session, but also in inter-session. This lower variability of GMax may raise a hypothesis related to a low level of activity of this muscle, or to the search for alternative movement strategies to overcome the low level of activity, such as in more proximal segments, i.e., in the trunk, suggesting the inclusion of the analysis in future gait assessment studies. Globally, more than 4 trials are required, and several muscles require more than 10. Overall, although electromyographic activity is considered a gold standard in the assessment of muscle activity, the variability of electromyographic variables seems to be greater than the variability of kinematics and kinetics in both groups, healthy and stroke. A critical reflection may be related to the greater variability of muscle activation patterns, the use of various muscle activation strategies – muscle synergies – may give rise to similar kinematic patterns (Grasso et al., 1998; Ivanenko et al., 2004). These findings suggest that neural circuits may somehow specify limb kinematics and appropriate muscle synergies according to mechanical constraints (Lacquaniti et al., 1999; Lacquaniti et al., 2002). Motor redundancy may support this interpretation since the same movement can be performed through different muscle activity combinations (Lacquaniti et al., 1999), which to some extent explains the lack of electromyographic studies. Although there are studies that have evaluated the repeatability of electromyographic variables in healthy people during gait (Rutherford et al., 2020; Taborri et al., 2018), to the best of our knowledge, there are no studies on this type of assessment on post-stroke people.

Inter-session

Regarding inter-session variability (temporal consistency), the number of trials to reach a good performance stability $ICC > 0.75$ was higher compared to the intra-session, demonstrating greater variability between the two moments of analysis. Inter-session variability seems to reflect extrinsic variability parameters, associated with methodological errors, related to: calibration or spatial resolution of motion capture systems, estimation of joint centres, marker application, marker movement on the skin in the case of kinematic and kinetic variables (Schwartz et al., 2004; Yavuzer et al., 2008); and the placement of the electrodes and the normalization process of the electromyographic signal. In the present study, to minimize methodological errors and their influence on the second data-gathering, identical conditions were ensured, the same calibration method, and the replacement of the reflector markers and electromyographic sensors were applied by the same researcher/ physiotherapist with experience in these assessment procedures.

The lower inter-session temporal consistency compared to the intra-session one observed in the healthy group, apart from methodological factors mentioned, may also be due to an exploratory behaviour in the search for optimal movement solutions according to the task restrictions (Davids et al., 2003). The results of this study in the healthy group demonstrated, in the LEAD position, a higher variability of the ankle joint position, compared to the other joints, that may be related to joint variations in the initial contact for positioning the foot on the force plates. In the stroke group, the opposite was observed with a less variable behaviour, which can be explained by the development of strategies to reduce degrees of freedom and increase stiffness in the ankle (Silva et al., 2017). The lower variability in the ankle joint position may reflect the higher variability in proximal segments observed in the present study.

Like observed in intra-session, the double support time and WCOM seem to be also more robust variables in both groups, healthy and stroke, inter-session. The decreased number of trials required for IPSI limb of stroke group in the LEAD position compared to the

healthy group can result from a decreased capacity of IPSI limb to adapt neuromuscular strategies to prepare the limb contact with the ground (A. S. P. Sousa et al., 2013).

The search for reliable techniques to assess gait in people with stroke sequelae is crucial to define the design and monitoring of rehabilitation programs. Considering the clinical implications of the results of this study, for people with stroke sequelae, we can highlight the sagittal gait analysis with good to excellent repeatability in most kinematic and kinetic variables and its application for gait assessment in the chronic phase of post-stroke recovery. However, in the inter-session assessment, means should be sought to minimize the interference of methodological factors, thus allowing the feasibility of monitoring the intervention and evolution. A relevant implication for clinical practice may be the possibility of applying these measurements to develop individualized rehabilitation programs, with person-centred clinical decisions (Fotiadou et al., 2018). The appropriate number of trials should consider the person under analysis and the parameter to be assessed, and therefore Fotiadou et al. (2018), it is suggested that the number of trials can be determined individually before each clinical assessment.

It would be interesting for future studies to investigate the repeatability of gait parameters in stroke patients with different gait speeds and different stages of motor recovery.

Conclusions

This is the first study to determine the number of trials needed to reach good stability performance for kinematic, kinetic, and electromyographic variables reflecting interlimb coordination during double support phase, in people with and without stroke sequelae.

The results of this study suggest that intra-session, for most of the kinematic variables in the sagittal plane (hip, knee, and ankle joint positions) and for the kinetic variable (WCOM), a low number of trials (2 to 3) seems to be sufficient to achieve a good stability of the performance of the variables ($ICC > 0.75$), while in the electromyographic variables of the lower limb muscles, a higher number of trials seems to be necessary (2 to >10).

Inter-session, while 1 trial is needed for double support time, a higher number of trials seems to be necessary for kinematic variables (1 to >10), for WCOM (1 to 9) and for electromyographic activity (1 to >10).

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ARTIGO III

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

Ana G. B. Couto ¹⁻³, Mário A. P. Vaz ⁴⁻⁵, Liliana Pinho ^{2, 6-7}, José Félix ⁸⁻⁹, Juliana Moreira ¹⁰, Francisco Pinho ¹¹, Inês Albuquerque Mesquita ¹²⁻¹³, António Mesquita Montes ^{1,10}, Carlos Crasto ^{1,10} and Andreia S. P. Sousa ¹⁰

1 Department of Physiotherapy and Research Center and Projects (NIP), Santa Maria Health School, Porto, Portugal

2 Center for Rehabilitation Research (CIR), School of Health of Polytechnic Institute of Porto, Porto, Portugal

3 Faculty of Engineering, University of Porto, Porto, Portugal

4 Institute of Mechanical Engineering and Industrial Management, Faculty of Engineering, University of Porto, Porto, Portugal

5 Porto Biomechanics Laboratory (LABIOMEP), University of Porto, Porto, Portugal

6 College of Health Sciences - Escola Superior de Saúde do Vale do Ave, Cooperative for Higher, Polytechnic and University Education, Vila Nova de Famalicão, Portugal

7 Faculty of Sport, University of Porto, Porto, Portugal

8 Department of Physics and Center for Rehabilitation Research (CIR), School of Health of Polytechnic Institute of Porto, Porto, Portugal

9 Department of Medical Sciences, University of Aveiro, Aveiro, Portugal

10 Department of Physiotherapy and Center for Rehabilitation Research (CIR), School of Health of Polytechnic Institute of Porto, Porto, Portugal

11 College of Health Sciences - Escola Superior de Saúde do Vale do Ave and Health and Human Movement Unit (H2M), Cooperative for Higher, Polytechnic and University Education, Vila Nova de Famalicão, Portugal

12 Research Center and Projects (NIP), Santa Maria Health School, Porto, Portugal

13 Department of Functional Sciences and Center for Rehabilitation Research (CIR), School of Health of Polytechnic Institute of Porto, Porto, Portugal

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Abstract

Background: Interlimb coordination impairment appears to be one of the factors leading to reduced gait efficiency in post-stroke persons. However, the knowledge regarding interlimb coordination in post-stroke is limited by the evidence focused on detecting differences in the contralesional side. A multi-instrumental biomechanical analysis of both limbs during walking remains uncommon. To identify possible differences between participants with and without stroke sequelae in respect to ipsilesional and contralesional lower limbs kinematics, kinetics, and muscle activity during double support phase of gait.

Methods: 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 with electromyography. The following outcomes were evaluated during the double support: the double support time and joint position; the external mechanical work on the centre of mass; and the relative electromyographic activity. Dominant/ non-dominant limbs in the healthy, and contralesional/ ipsilesional limbs in the stroke, were evaluated either in trailing (TRAIL) or leading (LEAD) positions. The average value of each parameter, as well as the coefficient of variation of the ten trials were used for analysis.

Findings: Differences between groups were observed in both positions. In the trailing position post-stroke participants presented: decreased contralesional knee flexion and ankle plantar flexion with bilateral increased variability; bilateral lower external mechanical work on centre of mass with increased variability; bilateral higher percentage of activity in ankle muscles; and decreased Vastus Medialis and knee ratio with related distal variability. In the leading position, post-stroke participants presented: bilateral increased knee flexion and increased hip and ankle variability; bilateral lower external mechanical work on centre of mass and increased variability; and higher bilateral percentage of activity on distal and proximal muscles and decreased variability.

Interpretation: Post-stroke participants present 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 leading position and increased variability. Globally, increased relative muscle activity was observed in post-stroke participants with decreased variability.

Keywords

Contralesional; Ipsilesional; Lower Limbs; Biomechanical Analysis; Quality Movement; Variability

Introduction

Stroke is the third leading cause of death and disability in the world (Feigin et al., 2021). The sensorial and motor repercussions of a stroke can cause limitations in daily activities and participation restriction's, both in the professional and social context (Rajsic et al., 2019). Gait is one of the most frequent altered functional tasks, with a great impact on the autonomy; therefore, the gait recovery is one of the main objectives for the post-stroke patients and for the rehabilitation (Belda-Lois et al., 2011; Bohannon et al., 1991; Langhorne et al., 2009; Mudge & Stott, 2007; Shumway-Cook, 2017; A. S. P. Sousa et al., 2013).

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, 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 involve impairment of function and performance in both CONTRA and IPSI limbs (Cramer et al., 1997; Genthon et al., 2008; Silva et al., 2012; Silva et al., 2015; Sousa et al., 2015; A. S. P. Sousa et al., 2013), 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 have been expressed in the asymmetric behaviour of both lower limbs (Arya & Pandian, 2014; Bajwa et al., 1992; Crafton et al., 2003; Olney et al., 1994; Olney et al., 1991; Olney & Richards, 1996; Schiemanck et al., 2006; A. S. P. Sousa et al., 2013; Sousa & Tavares, 2015). From a biomechanical point of view, these changes have been extensively explored and described at an individual level (Balaban & Tok, 2014; Chen et al., 2005; Kim & Eng, 2004; Lamontagne et al., 2002; Lamontagne et al., 2000; Lamontagne et al., 2007; Nadeau et al., 2013; Verma et al., 2012; Woolley, 2001), in terms of kinematic (joint position and double support stage) (Oken & Yavuzer, 2008), kinetic (ground reaction forces) (Bensoussan et al., 2006; Moseley et al., 1993) and muscle activation (distal and proximal

muscles) (Daly et al., 2011; Den Otter et al., 2007), of both lower limbs. However, according to our knowledge only some variables were considered in the study of the impairments of both limbs on post-stroke interlimb coordination (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; Donelan et al., 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 subphase (Kuo, 2002, 2007; Kuo et al., 2005), in which the two lower limbs cooperate to maintain the requirement of postural control and centre of mass progression (Kandel et al., 2014; Shumway-Cook, 2017). These gait requirements arise from the positive work of the posterior lower limb during propulsion, for centre of mass progression, that compensates for the energy expended on negative work by the anterior lower limb in initial contact 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 member 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 subphase of gait.

Human gait is characterized by natural variations in motor performance, known as optimal movement variability in healthy adults (Stergiou et al., 2006). Variability emerges as an aspect of motor control which can be quantified through linear measures (Harbourne & Stergiou, 2009), which may translate the impact of stroke on gait performance (Balasubramanian et al., 2009). 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). 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 as a possible post-stroke 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., 1994; Olney et al., 1991; 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 and without stroke sequelae. 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 of the lower limbs during double support phase of gait will be considered.

Methods

Study design

A cross sectional 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 no history of a stroke and without self-reported disabilities" named as the "healthy" group.

Participants

A group of eleven subjects (3 females and 8 males) with history of a single unilateral ischemic 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) (Table 1). Participants were excluded from the healthy group if they had an altered mental state with interference in communication and cooperation (Lamontagne et al., 2002; Lamontagne et al., 2000; A. S. P. Sousa et al., 2013), history or sign of neurological dysfunction (Sousa et al., 2012) presence of pain that interfered with the performance of walking (A. S. Sousa et al., 2013), history of osteoarticular or musculotendinous injury, history of lower limb surgery in the last 6 months, anatomical deformities of the lower limbs (A. S. Sousa et al., 2013; Sousa et al., 2012), taking medication that interferes with the motor performance of the lower limbs (Lamontagne et al., 2002; Lamontagne et al., 2000), moderate-intensity (i.e., at least 30 minutes 5 days a week) or vigorous-intensity aerobic physical activity (i.e., at least 20 minutes 3 days a week) (Thompson, 2014).

To be included in the stroke group the participants need to fulfil the following criteria: diagnosis of ischemic first-ever stroke with at least 6 months of evolution (Lamontagne et al., 2002; Lamontagne et al., 2000; A. S. P. Sousa et al., 2013), involving middle cerebral artery territory, at the subcortical level, confirmed by Axial Computed Tomography (Lamontagne et al., 2002; A. S. P. Sousa et al., 2013); lower limb sensorimotor impairment (A. S. P. Sousa et al., 2013); ability to walk at least 10 meters, with close supervision, if

necessary, but without physical assistance (Lamontagne et al., 2000; A. S. P. Sousa et al., 2013). In turn, people who had any of the exclusion criteria mentioned for the healthy group, or the presence of a lesion involving the brainstem and/ or cerebellum were excluded (Lamontagne et al., 2002; Lamontagne et al., 2000; A. S. P. Sousa et al., 2013).

Table 1. Participants characteristics (sociodemographic and clinical). Data presented from the mean (standard deviation). p value reflects a comparison between the stroke and healthy groups.

	Stroke (n=11)	Healthy (n=13)	Between- groups comparison
	Mean (sd)	Mean (sd)	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/m2)	24,58 (3,38)	26,29 (3,79)	0,261
Double support time (s)			
Ipsilesional/ Dominant	0,29 (0,31)	0,20 (0,03)	0,313
Contralesional/ Non-Dominant	0,24 (0,10)	0,19 (0,03)	0,162
Post-stroke time (months)	52,18 (29,89)	---	---

Ethical approval was obtained by the institutional Ethics Committee (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 inclusion and exclusion criteria defined for the selection of participants, and to characterize the sample, namely: age, sex, dominance, and time of evolution in the stroke group. This questionnaire included the Brief Physical Activity Assessment Tool (Cruz et al., 2021) to verify the people's level of physical activity.

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 the person has cognitive deficit depending on the level of schooling: score ≥ 15 for illiterates, ≥ 22 for people with 1 to 11 years of schooling, or ≥ 27 for people with more than 11 years of schooling (Guerreiro, 1994). This instrument has been adapted and validated for the Portuguese population by (Guerreiro, 1998; Guerreiro, 1994), with a sensitivity between 63% and 73.4% and a specificity between 90 and 96.8% (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 the person has sensorimotor impairment of the lower limb if a score < 34 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 by Costa (2003) and Santos (2005) with excellent internal consistency (α Cronbach=0.99) (Costa, 2003).

To characterize the sample, anthropometric measurements were performed, such as body mass (kg) and height (m), using a seca® 760 scale (seca – Medical Scales and Measuring Systems®, United Kingdom), with a resolution of 0, 1Kg; and a seca® 222 stadiometer (seca – Medical Scales and Measuring Systems®, United Kingdom), with a 1mm resolution.

Kinematic data

The joint position of the hip, knee, and ankle in the sagittal plane, on the dominant (DOM) side in the healthy group/ IPSI side in the stroke group, and on the non-dominant (NDOM)/ CONTRA sides, respectively, and time of the double support phase (space-time variable) were assessed using an optoelectronic technique, the Qualisys Motion Capture System (Qualisys AB, Sweden). The spatial position of reflector markers, placed on the

participant, was collected using twelve infrared cameras, eight Oqus 500 and four Miquis M3, connected to the Qualisys USB Analog Acquisition interface, at a sampling frequency of 100Hz.

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 1000Hz. The capture hardware was connected to the Qualisys Motion Capture System analogue board.

Electromyographic data

Surface electromyography (EMG) was monitored using the wireless Trigno™ 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 Gluteus Maximus (GMax), on the DOM side in the healthy group/ IPSI in the stroke group, and on the NDOM/ CONTRA, respectively. 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 EMG signals, at an acquisition frequency of 1000 Hz. The EMG signal was integrated into the Qualisys Motion Capture System, connecting its analogue board to the BNC-2090a board (National Instruments, United States of America). EMGworks software was used to analyse the EMG 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, v6 x64 (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 LEAD position (initial contact and loading response) and in the TRAIL position (pre-swing) (Kuo et al., 2005).

Procedures

Data collection took place at a biomechanical laboratory, the Centre for Rehabilitation Research 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 task. 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 $BMI = \frac{weight}{height^2}$.

For the collection of kinematic data, 46 reflective markers were placed bilaterally in anatomical references (identified by manual palpation) of the participant body: apex of the head, earlobes, jugular notch, xiphoid apophysis, spinous apophysis of the 7th cervical vertebral, lateral part of the acromion, anterior iliac spines superior, posterior superior iliac spines, greater trochanters, lateral and medial epicondyles of femur, anterior tuberosities of tibia; head of the fibula, lateral and medial malleoli, posterior surface of the calcaneus, head of the 1st, 2nd and 5th metatarsals, lateral and medial epicondyle of the humerus, styloid apophysis of the ulna and radius, 2nd and 5th metacarpals (C-Motion, United States of America) (Cappozzo et al., 1995). These markers would allow the built of a full-body biomechanical model in the Visual 3D software.

To collect the EMG signal, the electrodes were placed over the muscles under analysis followed the recommendation of the Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000) and the study by A. S. P. Sousa et al. (2013) (Table 2). Electrode placements were confirmed by palpation of the muscle belly and muscle contraction with the subject in the test position. Before placing the

electrodes, the participant's hair was shaved, an abrasive cream was used to remove dead cells from the skin's surface, and the skin was cleaned with isopropyl alcohol (70%) to remove oil and any remaining dead cells. The electrode impedance checker was used to ensure that impedance levels were less than 5 K Ω , which was considered to signify the satisfactory acquisition of the EMG signal (Hermens et al., 2000). Finally, the electrodes were placed parallel to the orientation of the underlying muscle fibers, 5 minutes after skin preparation.

Table 2. Anatomical references for the electrode placement (Hermens et al., 2000; A. S. Sousa et al., 2013).

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
Gluteus maximus	50% on the line between the sacrum and the greater trochanter

Task

Participants were asked to walk for 10 meters, 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 (A. S. Sousa et al., 2013; A. S. P. Sousa et al., 2013).

The participants performed 10 valid trials with the DOM/ IPSI lower limb in the TRAIL position and 10 valid trials with the NDOM/ CONT lower limb in the TRAIL position (Fotiadou et al., 2018), with a resting time of two-minutes between trials to prevent fatigue (A. S. Sousa et al., 2013; A. S. P. Sousa et al., 2013).

For the kinetic variable, the trials were considered valid if the 1st lower limb made full contact on the 1st platform – TRAIL, and the 2nd lower limb also made full contact on the 2nd platform – LEAD (A. S. Sousa et al., 2013).

Data processing

Kinematic data processing was carried out using the Qualisys Track Manager software, with the aim of recognizing the trajectory of the markers and verifying whether they were correctly identified during data collection. Path deviations/ interruptions were interpolated using the linear, polynomial, and relational calculations built into the software. Subsequently, the resulting data were 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 and toe-off events of the NDOM/ CONTRA and DOM/ IPSI limb. The double support phase was identified as the interval between heel strike of the LEAD and toe-off of the TRAIL limb. To identify these events, the methods proposed by Banks et al. (2015) for the heel strike and by Zeni et al. (2008) for the toe-off were considered. The Banks et al. (2015) method refers to the horizontal distance between the calcaneus and the ankle, and the heel strike is determined through the maximum displacement between the marker on the ipsilateral calcaneus and the marker on the contralateral lateral malleolus (Banks et al., 2015; French et al., 2020). The Zeni et al. (2008) method refers to the horizontal distance between the sacrum and the calcaneus, and the toe-off is determined through the minimum displacement between the calcaneal marker and the sacral marker (French et al., 2020; Zeni et al., 2008). Regarding the position of the hip and knee joints, in the sagittal plane during each of the events, the positive variation of the values refers to flexion and the negative variation to extension. For the position of the ankle joint, the positive variation of the values refers to plantar flexion and the negative variation to dorsal flexion. The identification of the heel strike and toe-off events also allowed

extracting the spatiotemporal parameter relative to the time of the double support phase of the gait (time difference between the two events).

Kinetic data processing was carried out by calculating the WCOM of each lower limb during the double support phase. The power generated on the body's centre of mass (external mechanical power) by a limb is equal to the dot product of the external force acting on the limb and the velocity of the centre of mass (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 Fy 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 (normalized by the mass of the participant) by the velocity of the centre of mass for Fy direction. Finally, the external work done on the centre of mass for each member was computed by calculating the mechanical energy integral of each member in the double limb support phase.

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

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

The electromyographic data processing was carried out using Matlab software, version 3.9.0. (MathWorks, United States of America). A second-order digital filter Infinite Impulse Response (Butterworth) – one at 20 Hz (high pass) and the other at 450 Hz (low pass) was applied to the EMG to remove electrical noise and sensor movement. Subsequently, the root mean square (RMS) value was calculated using a moving average window set to 100 samples (Lamontagne et al., 2000). The percentage of activation intensity of each muscle under analysis was obtained by normalizing the EMG signal by calculating the quotient between the mean RMS of each muscle during the double support phase and the maximum RMS of the same muscle during the gait cycle, multiplied by 100 (A. S. Sousa et al., 2013; Sousa & Tavares, 2012). In turn, the coactivation ratio was calculated according to the following equation (Kellis et al., 2003):

$$\text{Coactivation ratio} = \frac{\text{agonist activity}}{\text{agonist activity} + \text{antagonist activity}}$$

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 role of each agonist muscle in the double support phase was considered for each of the events. For the ankle coactivation ratio, in the LEAD position, TA was considered agonist, and in the TRAIL position, SOL and GasM were considered. In the knee ratio, RF and VM were considered agonists, in both positions, in LEAD and TRAIL. For the hip coactivation ratio, in the LEAD position, RF was considered agonist, and in the TRAIL position, GMax and BF were considered. 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's 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 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's 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 characterization (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 covariate of the 10 repetitions performed by each participant were considered for the analysis.

Results

Joint position

Through the analysis of table 1 differences between groups were observed in lower limb joints position in both TRAIL and LEAD positions. The stroke group presented lower knee flexion and ankle plantar flexion in the CONTRA limb in TRAIL position and higher knee flexion of both, IPSI and CONTRA 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 and IPSI, in TRAIL position. In LEAD position, the coefficient of variation was higher in the stroke group in the CONTRA and IPSI hip and ankle joint positions, but a lower variation coefficient was observed in knee joint position in the CONTRA limb.

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

		Stroke	Healthy	p	Stroke	Healthy	p
		CONTRA	NDOM	value	IPSI	DOM	value
TRAIL							
Hip	Mean	3,23 (6,45)	3,02 (8,64)	0,949	7,14 (6,42)	2,80 (9,17)	0,201
flexion (°)	CV	0,88 (2,91)	0,46 (1,46)	0,654	0,60 (1,57)	0,23 (0,62)	0,447
Knee	Mean	31,57 (9,21)	46,29 (4,53)	<0,001	45,92 (6,55)	45,67 (5,87)	0,922
flexion (°)	CV	0,19 (0,16)	0,04 (0,01)	0,011	0,11 (0,06)	0,04 (0,02)	0,007
Ankle	Mean	16,44 (9,63)	30,18 (6,04)	<0,001	25,47 (8,57)	29,89 (6,85)	0,174
plantar	CV	0,45 (0,65)	0,07 (0,03)	0,001*	0,18 (0,10)	0,08 (0,04)	0,007
flexion (°)							
LEAD							
Hip	Mean	31,15 (6,88)	32,15 (8,32)	0,756	36,18 (7,85)	31,71 (8,20)	0,188
flexion (°)	CV	0,10 (0,07)	0,04 (0,01)	0,019	0,08 (0,08)	0,04 (0,02)	0,041*
Knee	Mean	11,05 (4,31)	0,00 (5,56)	<0,001	16,50 (10,13)	-0,19 (5,08)	<0,001
flexion (°)	CV	0,36 (0,39)	-2,63 (6,06)	0,035*	0,08 (0,42)	0,11 (0,55)	0,890
Ankle plantar	Mean	17,73 (7,73)	14,83 (2,90)	0,262	12,40 (5,64)	14,56 (3,39)	0,259
flexion (°)	CV	0,21 (0,24)	0,09 (0,09)	0,009*	0,51 (0,78)	0,07 (0,02)	<0,001*

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

Double support time and external mechanical work on the centre of mass

Table 2 presents the values of WCOM and double support time by group, limb, and position. The results demonstrate that post-stroke group presented a lower WCOM when CONTRA limb was both in LEAD and TRAIL positions, and when IPSI limb was in LEAD position. Higher coefficient of variation was found on stroke group in respect to double support time and WCOM in TRAIL position on both limbs and higher WCOM in the IPSI limb.

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

		Stroke	Healthy	p	Stroke	Healthy	p
		CONTRA	NDOM	value	IPSI	DOM	value
Double support time							
	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

Significant differences in bold. CV: coefficient of variation

Electromyographic activity

The results presented in table 3 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 and GAS muscle activity of CONTRA limb and higher percentage of SOL activity and ankle ratio of the IPSI limb, but lower percentage of VM activity and knee ratio of CONTRA limb. In the LEAD position, the stroke group presented higher percentage of TA, RF, VM, BF and GMax activity in both limbs and higher percentage of SOL activity also in the IPSI limb.

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

	Stroke	Healthy	p value	Stroke	Healthy	p value
	CONTRA	NDOM		IPSI	DOM	
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
LEAD						
Ankle						
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

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

When analysing the variation coefficient of the electromyographic variables expressed in table 4, a distinct pattern was observed compared to the one observed for kinematic variables, since, globally, the variation coefficient was lower in the stroke group than in the healthy group. In the TRAIL position, significant differences were observed in SOL, GAS, and ankle ratio for both limbs and in TA of CONTRA limb. In the LEAD position the

differences were observed in TA, SOL, RF and BF and hip ratio of both limbs, in GasM of the CONTRA and in ankle and knee ratio of the IPSI limb.

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 Gluteus Maximus (GMax) muscles and the coactivation ratio of hip, knee, and ankle joints. 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.

	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*	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*	0,10 (0,04)	0,18 (0,10)	0,031

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

Discussion

This study aimed to identify possible differences between participants with and without stroke sequelae in respect to IPSI and CONTRA lower limbs kinematics, kinetics, and muscle activity during double support phase of gait in terms of mean and covariation. In human movement the variability is a natural, often beneficial, feature that can be optimized 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.

The results obtained demonstrate that the mean time of double support was similar 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 (TRAIL) 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 centre of mass) 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., 1995; Olney & Richards, 1996; A. S. P. Sousa et al., 2013). 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 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; A. S. P. Sousa et al., 2013) (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 hypothesize that SOL and VM activity may influence the ratio. These results contradict previous studies that demonstrate, during double support phase, different results with different normalization procedures, namely mean dynamic method, showing a decreased muscle activity (A. S. P. Sousa et al., 2013) or a decreased coactivation ratio (Silva et al., 2015); and a higher coactivation ratio, using the absolute EMG value (Lamontagne et al., 2000). The normalization 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, normalized 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 normalization 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; A. S. P. Sousa et al., 2013). 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; A. S. P. Sousa et al., 2013). 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; A. S. P. Sousa et al., 2013).

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 TRAIL 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.

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 CONTRA knee and ankle flexion in TRAIL position and increased IPSI 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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CAPÍTULO V – DISCUSSÃO GERAL

Neste capítulo, encontra-se apresentada uma discussão integrada dos principais resultados, orientada para as três questões de investigação da presente Tese. A metodologia e os resultados de cada estudo foram discutidos individualmente no Capítulo IV.

A presente tese apresentou como objetivo principal caracterizar a coordenação entre os membros inferiores na marcha, especificamente na fase de duplo apoio, em pessoas com e sem sequelas de AVC, através da análise cinemática, cinética e eletromiográfica. É de salientar que a fase de duplo apoio da marcha, corresponde à fase de transição entre passos, em que ambos os membros inferiores estão em contacto com o solo. Nesta fase, a coordenação entre os membros assume um papel decisivo na resposta aos requisitos do controlo postural e da progressão do corpo em constante movimento. Este é, assim, um aspeto relevante do controlo motor da marcha, e tem sido apontado como um fator determinante na sua eficiência (Donelan et al., 2002a; Donelan et al., 2004; Krasovsky et al., 2012; Kuo, 2002, 2007; Kuo et al., 2005; Sousa & Tavares, 2015). A ocorrência de um AVC, pelo atingimento de estruturas neurais com relevância para a modulação bilateral da marcha, associado às adaptações compensatórias pós-lesão, expressa o comprometimento bilateral assimétrico na inter-relação dos membros inferiores. A importância de explorar a coordenação entre membros, na fase de duplo apoio da marcha, em pessoas com sequelas de AVC, não só depende do impacto da lesão do membro contralesional (CONTRA) sobre o membro ipsilesional (IPSI), mas também da disfunção do IPSI no desempenho do CONTRA.

Avaliação da coordenação entre membros na marcha

Previamente à recolha de dados biomecânicos, foi sentida a necessidade de se aprofundar, refletir e fundamentar a temática, de forma a desenvolver a metodologia mais adequada e especialmente consolidada pela comunidade científica. Neste sentido, foi desenvolvida a *Scoping Review* que permitiu uma melhor compreensão das considerações metodológicas para a avaliação da coordenação entre membros na

marcha e promoveu uma reflexão sobre as informações-chave a considerar para pesquisas futuras. Esta revisão viabilizou a decisão sobre as condições de análise da marcha, a definição dos instrumentos a utilizar e as diferentes variáveis a estudar.

No que diz respeito às suas condições de avaliação, a marcha no solo estável, a uma velocidade auto-selecionada, tem uma menor interferência nas características naturais, minimizando a variabilidade dos parâmetros da marcha das pessoas com sequelas de AVC (Masani et al., 2002; Reisman et al., 2009; Sekiya et al., 1997; Sousa & Tavares, 2012).

Por conseguinte, foram identificados os instrumentos a usar para a avaliação da marcha, nomeadamente para a recolha de dados cinemáticos, cinéticos e de EMG. Os sistemas optoelectrónicos foram os mais referenciados no que diz respeito à recolha dos dados cinemáticos (captura de movimento) (Corazza et al., 2010). É importante salientar que a precisão destes sistemas é dependente de diversos fatores, como a localização relativa entre as câmaras, da distância entre estas e os marcadores; da posição, do número e do tipo de marcadores e do movimento dos marcadores no volume de captura (Maletsky et al., 2007); sendo que este último é diretamente dependente do número máximo de câmaras e do campo de visão de cada uma (van der Kruk & Reijne, 2018). Por outro lado, para a recolha dos dados cinéticos, tem vindo a ser recomendado o recurso a plataformas de forças, como o instrumento de alta precisão para o registo das forças de reação ao solo e respetivos torques (Garza-Ulloa, 2018). Por fim, no que diz respeito à EMG, foi recomendada a utilização de sensores para medir de forma global, e não invasiva, a atividade muscular (Agostini et al., 2020; McManus et al., 2020; Papagiannis et al., 2019).

Perante as ferramentas supracitadas, tornou-se necessário identificar as variáveis que permitem a avaliação dos dados cinemáticos, cinéticos e da EMG. Das variáveis reportadas, as mais frequentes foram os parâmetros espaço-temporais. Porém, para responder aos objetivos desta tese, considerou-se dar destaque apenas ao tempo de duplo apoio, por expressar especificamente a fase em estudo. Esta fase foi definida de acordo com a identificação dos eventos *heel-strike* e *toe-off*. Nestes, como variáveis cinemáticas foram consideradas a posição das articulações dos membros inferiores, nomeadamente, da coxo-femoral, do joelho e do tornozelo, bilateralmente. Das variáveis cinéticas, foi considerado apenas o trabalho mecânico externo sobre o centro de massa

(WCOM), realizado por cada membro inferior na fase de duplo apoio, considerando a contribuição de cada membro em cada posição (*leading* (LEAD) e *trailing* (TRAIL)). De facto, a posição de cada membro inferior traduz a relação energética entre ambos na fase de duplo apoio, e de acordo com o modelo teórico, aproximadamente 75% da perda energética ocorre durante o contacto inicial do calcanhar ao solo, do membro LEAD, sendo compensada pelo membro TRAIL, durante a propulsão na fase de duplo apoio (Kuo, 2002, 2007). Nas variáveis eletromiográficas dos diversos músculos dos membros inferiores, foi identificado o *root mean square* (RMS) para expressar a potência da atividade muscular. Foi ainda considerada outra variável, o rácio de coativação muscular, de acordo com a posição funcional de cada membro.

Apesar da importância de sumariar as principais considerações metodológicas a serem incluídas na coordenação entre membros, a identificação das variáveis a ter em conta pode não ser suficiente. Parece também ser importante compreender o comportamento dessas variáveis em termos de estabilidade de desempenho a nível temporal (intra-sessão e inter-sessão). Assim, e conhecendo a consistência do seu comportamento, será possível identificar as que melhor possibilitam estudar a coordenação entre membros, permitindo também acompanhar as melhorias clínicas no contexto da reabilitação da marcha.

Estabilidade de desempenho das variáveis biomecânicas que expressam a coordenação entre os membros na marcha – intra-sessão e inter-sessão

A avaliação da coordenação entre membros inferiores na fase de duplo apoio da marcha, em pessoas com sequelas de AVC, carece de abordagens de avaliação reprodutíveis para a compreensão dos mecanismos subjacentes à reabilitação. A definição de parâmetros de avaliação repetíveis (intra-sessão) e consistentes temporalmente (inter-sessão) pode ser analisada através do número mínimo de repetições para alcançar um bom ICC (>0.75), com o intuito de garantir uma boa estabilidade de desempenho e também permitir uma melhor compreensão da variabilidade motora (Fotiadou et al., 2018). Neste sentido, no Artigo II foi possível, de forma global, concluir que as variáveis cinemáticas e cinéticas estudadas demonstraram uma boa estabilidade de desempenho, contrariamente ao

observado nas variáveis eletromiográficas, em pessoas com e sem sequelas de AVC. Não obstante, o número mínimo de repetições intra-sessão traduziu uma menor variabilidade comparativamente com o inter-sessões, nas diferentes variáveis analisadas e em ambos os grupos.

As condições da avaliação da marcha, nomeadamente ter sido realizada numa superfície estável e a uma velocidade confortável auto-selecionada, podem justificar a necessidade de um número de repetições baixo para atingir uma boa estabilidade de desempenho ($ICC > 0,75$), nas variáveis cinemáticas e cinética estudadas numa única sessão de avaliação (Carson et al., 2014; Terrier et al., 2005). Independentemente das estratégias de movimento adotadas pelas pessoas com sequelas de AVC se traduzirem ou não em diferenças e/ ou na variabilidade nos parâmetros biomecânicos estudados, este resultado poderá estar relacionado com o desenvolvimento de padrões de fixação posicional, estereotipado, que se poderão associar ao processo de compensação na recuperação pós-AVC (Jones, 2017; Perry et al., 1995).

O plano sagital foi o único considerado para a avaliação da marcha, por ser o principal plano de movimento dos segmentos durante esta atividade, e no qual é expectável uma menor variabilidade (Fotiadou et al., 2018; Kadaba et al., 1989; Yavuzer et al., 2008). A seleção destes critérios para a avaliação da coordenação entre membros foi relevante para permitir uma interpretação mais criteriosa dos resultados.

Apesar de a EMG de superfície ser considerada uma medida de ouro na avaliação da atividade muscular, esta parece precisar de um maior número de repetições em comparação com a cinemática e a cinética, em pessoas com e sem sequelas de AVC. O uso de diversas estratégias de ativação muscular – sinergias musculares – podem dar origem a padrões cinemáticos semelhantes (Grasso et al., 1998; Ivanenko et al., 2004; Lacquaniti et al., 1999; Lacquaniti et al., 2002). A redundância motora pode assim suportar esta interpretação, uma vez que o mesmo movimento pode ser realizado através de diferentes combinações de atividade muscular (Lacquaniti et al., 1999). Tal facto, pode também explicar, em certa medida, a escassez de estudos na repetibilidade de variáveis eletromiográficas em pessoas saudáveis durante a marcha (Rutherford et al.,

2020; Taborri et al., 2018), e até onde se sabe, não existem estudos sobre este tipo de avaliação em pessoas pós-AVC.

No que diz respeito à consistência temporal (inter-sessão), o número mínimo de repetições para atingir uma boa estabilidade de desempenho ($ICC > 0,75$) nas variáveis cinemáticas e eletromiográficas foi maior em comparação com a da intra-sessão, demonstrando menor reprodutibilidade entre os dois momentos de análise. Esta, nos dois grupos, poderá refletir a influência de parâmetros de variabilidade extrínseca, associada a fatores metodológicos durante a avaliação em estudo (Schwartz et al., 2004; Yavuzer et al., 2008). Não obstante, o tempo de duplo apoio e o WCOM inter-sessão parecem ser as variáveis mais robustas das estudadas, levantando a possibilidade de estas serem consistentes temporalmente para a monitorização da evolução de um plano de intervenção.

A procura de metodologias fiáveis para avaliar a marcha em pessoas com sequelas de AVC é crucial para definir o planeamento e a monitorização dos programas de reabilitação. Das implicações clínicas dos resultados deste estudo de estabilidade de desempenho das variáveis, para pessoas com sequelas de AVC, podem destacar-se a análise da marcha no plano sagital com boa e excelente repetibilidade, na maioria das variáveis cinemáticas e cinéticas, e a sua aplicação para avaliação da marcha na fase crónica da recuperação pós-AVC. No entanto, na avaliação inter-sessão, há necessidade de procurar formas para minimizar a interferência dos fatores metodológicos, permitindo assim a viabilidade de monitorizar a intervenção e a evolução neuromotora.

Para além de perceber se as diversas variáveis biomecânicas que expressam a coordenação entre membros apresentam uma boa estabilidade de desempenho, intra-sessão e inter-sessão, é crucial identificar se estas identificam as diferenças nos membros das pessoas com sequelas de AVC, não só do CONTRA, mas também do IPSI.

Coordenação entre membros na marcha em pessoas com sequelas de Acidente Vascular Cerebral

A compreensão do comprometimento bilateral dos membros inferiores pós-AVC tornou necessário para identificar as possíveis diferenças das variáveis biomecânicas que expressam a coordenação entre os membros inferiores das pessoas com e sem sequelas de AVC relativamente à cinemática, cinética e atividade muscular durante a fase de duplo apoio da marcha. Reconhecendo a quantificação da variabilidade como uma estratégia adaptativa pós-AVC, o seu estudo no movimento é relevante para identificar alterações em ambos os membros inferiores, durante a fase de duplo apoio. De facto, de acordo com as características da marcha humana, rítmica, automática e integrada (Carson et al., 2014; Terrier et al., 2005), e pelas características metodológicas assumidas (Masani et al., 2002; Reisman et al., 2009; Sekiya et al., 1997; Sousa & Tavares, 2012), não seria expectável uma alteração significativa da variabilidade dos parâmetros biomecânicos em pessoas com sequelas de AVC. No entanto, a literatura aponta para uma variabilidade superior na condição patológica (Rosano et al., 2007).

Os resultados observados no Artigo III demonstraram as diferenças entre grupos nos parâmetros cinemáticos, cinéticos e eletromiográficos, sendo estas mais notórias na quantificação da variabilidade dos mesmos, entre as pessoas com e sem sequelas de AVC, nomeadamente nos membros contralesional (CONTRA)/ não dominante (NDOM) e ipsilesional (IPSI)/ dominante (DOM), em ambas as posições (TRAIL e LEAD).

Especificamente, na posição TRAIL, as pessoas com sequelas de AVC apresentaram uma menor flexão do joelho e flexão plantar do tornozelo no membro CONTRA durante o *toe-off* (TRAIL), corroborando resultados de outros autores (Chen et al., 2005; de Quervain et al., 1996; Goldberg et al., 2006; Matsuda et al., 2016; Moseley et al., 1993; Nadeau et al., 2013; Perry & Burnfield, 2010). Estes resultados podem relacionar-se com o menor WCOM (trabalho positivo sobre o COM) observado no mesmo membro, e de acordo com estudos prévios (Beyaert et al., 2015; Bowden et al., 2006; Kim & Eng, 2003; Morita et al., 1995; Olney & Richards, 1996; Sousa et al., 2013). No entanto, nestas variáveis foi observada uma variabilidade superior, possivelmente relacionada com a instabilidade postural na transição do membro inferior de uma cadeia cinética fechada para uma

aberta, na fase de propulsão (Fotiadou et al., 2018). Por outro lado, o resultado do WCOM CONTRA em TRAIL foi menor do que o WCOM do IPSI em LEAD, demonstrando que o trabalho negativo do IPSI no *heel-strike* não é compensado pelo trabalho positivo do CONTRA no *toe-off*. Este achado para além de estar de acordo com estudos prévios (Balasubramanian et al., 2007; Sousa et al., 2013), pode estar associado a uma capacidade inferior do membro CONTRA para se ajustar ao IPSI nesta posição (Kim & Eng, 2004; Olney & Richards, 1996; Sousa et al., 2013).

Na posição LEAD, o grupo pós-AVC apresentou maior flexão do joelho bilateral durante o contacto inicial com o solo, de acordo com resultados anteriores (Levine et al., 2012; Moore et al., 1993; Nadeau et al., 2013), podendo explicar o baixo WCOM dos membros inferiores CONTRA e IPSI, uma vez que a flexão do joelho tem um papel relevante na absorção das forças de reação ao solo (Mazuquin et al., 2014). Tal como foi mencionado acima para a posição TRAIL, foi observada uma variabilidade significativamente maior na globalidade das posições articulares e no WCOM. Esta elevada variabilidade poderá estar relacionada com a precisão da colocação do pé no solo. Seguindo esta suposição, Donelan and Pearson (2004) declarou que a colocação do pé no contacto inicial implica ajustes que se traduzem num custo metabólico mais elevado na marcha.

À semelhança da variabilidade da cinemática e cinética, a variabilidade no tempo de duplo apoio foi superior nas pessoas com sequelas de AVC. Esta variabilidade superior é apontada como resultado da alteração da coordenação entre os membros (Balasubramanian et al., 2009; Wang et al., 2020; Zukowski et al., 2019). Sabe-se que a variabilidade temporal pode ser considerada um indicador robusto da marcha, bem como um alvo para a intervenção neuromotora de pessoas com sequelas de AVC (Patel et al., 2022).

Relativamente à atividade muscular, a variabilidade foi menor nos membros do grupo AVC podendo este achado estar relacionado com um padrão de recrutamento muscular estereotipado, que traduz uma menor variação e ajuste da atividade muscular no desempenho motor da marcha (Di Nardo et al., 2020; Lord et al., 2011).

Os resultados supracitados apresentam uma visão pertinente do comprometimento neuromotor após um AVC comparativamente com pessoas sem sequelas de AVC, uma

vez que, para além das alterações identificadas no membro CONTRA, os resultados apontam para alterações também no membro IPSI. Estas foram encontradas não só na quantificação média das variáveis, mas também na sua variabilidade, na fase de duplo apoio. De forma global, o desempenho neuro-motor do membro IPSI pode expressar adaptações compensatórias associadas ao comprometimento do CONTRA, mas é importante ter em consideração que podem expressar a própria lesão, suportada pelas bases neurofisiológicas.

Considerações metodológicas

A decisão de comparar as variáveis entre os membros inferiores, dos grupos AVC e saudável, em que se optou por comparar o membro CONTRA com o NDOM e o IPSI com o DOM teve por base o estudo de Krasovsky et al. (2013). Nesse estudo, o autor refere que a maior parte da variabilidade da duração do duplo apoio deriva da assimetria entre a duração das 2 subfases do duplo apoio no ciclo de marcha, ou seja, quando cada membro está em TRAIL e em LEAD, e não da variabilidade da duração dessas subfases de duplo apoio de cada lado. Assim, as durações da subfase de duplo apoio foram calculadas separadamente, considerando como membro CONTRA/ NDOM e o IPSI/ DOM na posição LEAD (Krasovsky et al., 2013).

A recolha dos dados cinemáticos teve em consideração a utilização de modelos de captura de movimento apropriados de acordo, especificamente, com o objetivo de pesquisa delineado, para garantir a definição dos ângulos cinemáticos em análise. Para a exatidão e precisão das medições, apenas um investigador esteve na preparação e colocação criteriosa dos marcadores (Della Croce et al., 2005).

Na recolha dos dados eletromiográficos, a preparação cuidada da pele e a escolha do tipo de sensores, a colocação destes e as configurações de gravação, incluindo as características do filtro e ganhos do amplificador foram detalhadamente definidos para o registo dos sinais de EMG com elevada qualidade e baixo ruído (Agostini et al., 2020; McManus et al., 2020; Papagiannis et al., 2019).

De forma a aumentar a robustez dos dados e acompanhar a evolução tecnológica, os investigadores consideraram a realização de um estudo piloto, para aumentar a familiarização com o protocolo experimental e melhorar as configurações de modo mais preciso.

Apesar de, nos resultados da revisão incluída nesta tese, ter sido referido o índice de simetria, de parâmetros espaço-temporais ou amplitudes articulares, não foi uma escolha na metodologia do protocolo experimental, dado que o cálculo deste índice feito, maioritariamente, em função da referência do IPSI, pode não corresponder à real alteração da expressão do comprometimento, uma vez que não é possível garantir que a aproximação à simetria deverá ser refletida no CONTRA a aproximar-se do IPSI e não o oposto. Para tal, foi levado em consideração que, embora uma melhoria na coordenação entre membros seja desejável e necessária na reabilitação, a simetria é um descritor incompleto (Krasovsky & Levin, 2010). Os estudos que avaliaram o WCOM dos membros inferiores, LEAD e TRAIL, mostraram que este último precisa produzir mais trabalho do que o LEAD (Kuo, 2002, 2007; Kuo et al., 2005; Soo & Donelan, 2010, 2012; Sousa et al., 2012). Por outro lado, considerar o valor das variáveis do membro IPSI reconhecendo, à partida, que este também apresenta alterações, é subestimar o potencial de recuperação do CONTRA. Mais se acrescenta que, apesar de ser objetivo comum da reabilitação a procura da simetria, alguns investigadores sugerem que a assimetria da marcha não deve ser totalmente alterada, particularmente na fase crónica pós-AVC, sendo pouco provável que resulte num desempenho ótimo (Balasubramanian et al., 2007). De acordo com este ponto de vista, a assimetria da marcha pode ser uma adaptação ou compensação positiva face aos défices neurológicos pós-AVC, que proporciona às pessoas um certo nível de independência e autonomia, de acordo com o domínio das atividades da Classificação de Independência Funcional.

A coativação muscular na fase de duplo apoio da marcha foi um dos *outcomes* da presente tese com o propósito de avaliar a: 1) capacidade de inibir a atividade muscular não relacionada com a tarefa; 2) estabilidade articular; 3) ineficiência do movimento humano. Diversos métodos têm sido usados para calcular o rácio de coativação, com o intuito de mensurar esta ativação simultânea dos vários músculos que atuam numa articulação. Por conseguinte, toda e qualquer conclusão sobre a coativação muscular

dependerá da equação usada para estimar o seu rácio (Kellis et al., 2003). Não existindo, então, um consenso na literatura científica sobre o método de cálculo do rácio de coativação, a equação usada na presente tese foi baseada na relevância da função de cada um dos músculos na fase de duplo de apoio da marcha, tendo em consideração a posição do membro inferior (LEAD *versus* TRAIL). Tendo em conta o referido, os resultados deste *outcome* foram interpretados de acordo com o método de cálculo do rácio de coativação.

Aplicações da avaliação da coordenação entre membros na marcha

Dos resultados da presente Tese podem ser destacadas algumas aplicações da avaliação da coordenação entre membros na fase de duplo apoio da marcha para a prática clínica.

As metodologias precisas e adequadamente definidas podem permitir a identificação de características específicas do padrão de marcha, auxiliando na avaliação, na deteção de disfunções de movimento e na intervenção, de forma individualizada (Horst et al., 2019). Estes resultados contribuíram para refinar a definição e o processo de avaliação da coordenação entre membros na marcha em pessoas com sequelas de AVC, possibilitando a seleção dos instrumentos mais indicados e das variáveis mais relevantes.

A identificação de parâmetros com uma boa estabilidade de desempenho entre várias repetições (intra-sessão) e entre sessões (inter-sessão), pode auxiliar no planeamento, estruturação e monitorização dos programas de intervenção da marcha. O baixo número de repetições observado, especialmente intra-sessão, nos parâmetros cinemáticos (posição articular da coxo-femoral, joelho e tornozelo; tempo de duplo apoio) e cinéticos (WCOM) permite refletir sobre a possibilidade da sua aplicação na avaliação e decisão clínica de uma única avaliação da marcha.

Implicações da avaliação da coordenação entre membros na marcha para a prática clínica

A partir dos resultados da presente Tese podem ser destacadas algumas implicações da avaliação da coordenação entre membros na fase de duplo apoio da marcha para a prática clínica.

O tema explorado nesta Tese pode trazer implicações importantes para a neuro-reabilitação de pessoas com sequelas de AVC. Numa primeira instância, os resultados da presente Tese demonstram que o número adequado de repetições deve considerar não só o parâmetro a avaliar, mas também a pessoa em análise, pelo que Fotiadou et al. (2018) sugere que o número de repetições possa ser determinado individualmente antes de cada avaliação clínica.

Para além de fornecer benefícios substanciais para o conhecimento e compreensão da aplicação na avaliação da coordenação entre membros na marcha nesta condição patológica, permitiu aprofundar a caracterização da expressão motora da disfunção do IPSI, para além da lesão do CONTRA pós-AVC. As implicações deste comprometimento assimétrico bilateral devem ser analisadas quando se refere o membro inferior IPSI como "não afetado", e se usar o seu comportamento como medida de referência e monitorização para reabilitação e investigação, dado que pode subestimar o potencial de reabilitação do membro CONTRA (Raja et al., 2012).

Reconhecendo que as alterações identificadas no membro IPSI podem ser resultantes da combinação dos défices neurológicos associados à lesão estrutural primária e aos mecanismos de recuperação neuronal, mas também aos componentes biomecânicos, relacionados com a função residual e com o desenvolvimento de movimentos e estratégias compensatórias, sugere-se que ambos os membros inferiores devam ser considerados numa abordagem de neuro-reabilitação bilateral, para favorecer a integração e inter-relação dos dois membros inferiores. Dietz and Schrafl-Altermatt (2016) referem que a influência do lado IPSI sobre o lado CONTRA parece estar preservada, conduzindo a uma ativação muscular quase normal do lado CONTRA, mas o mesmo não ocorre no sentido oposto (Dietz & Schrafl-Altermatt, 2016). Assim, pode pensar-se que, para a neuro-reabilitação de pessoas pós-AVC, esta estratégia de

intervenção pode trazer implicações importantes, sendo necessário o desenvolvimento de mais investigação neste sentido, sugerindo o acompanhamento e monitorização de casos clínicos.

Investigação futura

Da presente Tese podem ser mencionadas algumas sugestões para a investigação futura, nomeadamente:

- Incluir outras variáveis espaço-temporais para além das estudadas, como o comprimento do passo e da passada, o tempo do passo e da passada, a velocidade e a cadência; bem como outras medidas que expressem a coordenação entre membros como o índice de coordenação; e analisar outros planos de movimento (frontal e transversal);
- Investigar a repetibilidade dos parâmetros de marcha em pessoas com sequelas de AVC, mas em diferentes velocidades de marcha, em diferentes fases de recuperação motora, bem como também a comparação entre pessoas com sequelas de AVC em diferentes áreas de lesão;
- Estudar a influencia do membro superior na coordenação dos membros inferiores nesta fase da marcha;
- Estudar o comportamento motor dos membros inferiores com o recurso a medidas não lineares na análise biomecânica, que poderá permitir uma melhor compreensão da qualidade do movimento e dos seus parâmetros cinéticos, cinemáticos e de atividade muscular, e do seu comportamento ao longo do tempo;
- Incluir medidas de fiabilidade absoluta como o erro padrão de medição (SEM), a diferença mínima detetável a 95% (MDC95) e a diferença minimal clinicamente importante (MCIC), de modo a permitir minimizar a sobrevalorização das pequenas diferenças como significativas.

CAPÍTULO VI – CONCLUSÕES

A presente Tese contribuiu com evidência científica sobre a caracterização da coordenação entre membros, contralesional (CONTRA) e ipsilesional (IPSI), na fase de duplo apoio da marcha em pessoas com sequelas de Acidente Vascular Cerebral (AVC).

Na *Scoping Review*, foi possível sumariar as orientações para a metodologia de avaliação da coordenação entre membros na marcha em pessoas com sequelas de AVC, nomeadamente: as condições da marcha (solo estável e velocidade), os instrumentos para a avaliação (cinemática, cinética e eletromiografia), e as respetivas variáveis biomecânicas a estudar, (por exemplo os parâmetros espaço-temporais, a posição articular, a força de reação ao solo ântero-posterior e a magnitude da atividade eletromiográfica).

O estudo II permitiu concluir que, intra-sessão, para a maioria das variáveis cinemáticas no plano sagital (posição da articulação da anca, joelho e tornozelo) e para a variável cinética (trabalho mecânico externo no centro de massa (WCOM)) um baixo número de repetições (2 a 3) parece ser suficiente para atingir uma boa estabilidade temporal; já as variáveis eletromiográficas do membro inferior revelaram ser necessário um maior número de repetições. Relativamente à consistência inter-sessões, para a variável tempo de duplo apoio, 1 repetição foi suficiente para atingir uma boa estabilidade de desempenho e para as restantes variáveis foi identificada uma maior variabilidade na quantidade de repetições necessárias.

Por sua vez, no estudo III, as pessoas com sequelas de AVC parecem diferentes dos saudáveis em ambos os membros e em ambas as posições. Especificamente, parece existir uma diminuição do WCOM; uma diminuição da flexão do joelho e do tornozelo do CONTRA em *trailing* (TRAIL); um aumento da flexão do joelho IPSI em *leading* (LEAD); uma maior da variabilidade bilateral nestas variáveis; uma maior percentagem relativa da atividade muscular e uma menor variabilidade da mesma, bilateralmente. Globalmente, a posição articular, o WCOM e a percentagem de atividade muscular bem como a variabilidade nestas variáveis parecem diferenciar pessoas com sequelas de AVC de pessoas sem sequelas de AVC.

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APÊNDICES

Apêndice 1. Parecer da Comissão de Ética

E-ESS/EXP-1808/2019-DATA: 05-06-2019

P.PORTO

**ESCOLA
SUPERIOR
DE SAÚDE
POLITÉCNICO
DO PORTO**

PARECER DA COMISSÃO DE ÉTICA

1372

Número de Registo da Comissão de Ética

17/05/2019

Data receção do Documento

Sim (09/04/2019)

Existência de entradas anteriores

TÍTULO DO TRABALHO

Interlimb neural coupling: avaliação, implicações e aplicações clínicas em indivíduos com sequelas de acidente vascular cerebral

INVESTIGADOR RESPONSÁVEL

Ana Couto

DATA PREVISTA PARA A REALIZAÇÃO DO TRABALHO

Início a 01/03/2019 e término 30/09/2019

RESUMO DO ESTUDO

OBJETIVOS

Nada a referir.

AMOSTRA

Indivíduos saudáveis contactados por convite directo e indivíduos com patologia recrutados através das clínicas, com apresentação presencial do projecto, via profissionais.

FORMULÁRIO DE DADOS A RECOLHER

Apresentados. O anexo B foi ajustado de acordo com as indicações dadas por esta comissão.

MATERIAL

Nada a referir.

MÉTODOS

Explicitados e conformes.

RISCOS

Referidos e de que forma vão ser acautelados caso aconteçam.

CONSENTIMENTO INFORMADO

Apresentado e conforme.

AUTORIZAÇÃO PELOS RESPONSÁVEIS LOCAIS

Apresenta autorizações das diferentes clínicas.

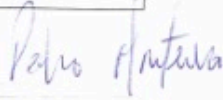
APRECIÇÃO DA COMISSÃO DE ÉTICA

As questões foram esclarecidas, o documento em falta entregue e o questionário ajustado.

PARECER FINAL DA COMISSÃO DE ÉTICA

De acordo com os dados analisados, o parecer é favorável, mas salienta-se a necessidade de cumprimento de todas as diretrizes submetidas a esta Comissão, com prejuízo de a decisão ser suspensa caso haja algum incumprimento grave.

 DATA: 27/06/2019
ESS004.MD.318.01

ASSINATURAS 

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Apêndice 2. Consentimento Informado

ESCOLA
SUPERIOR
DE SAÚDE
POLITÉCNICO
DO PORTO

P.PORTO

TERMO DE CONSENTIMENTO INFORMADO

O termo de consentimento informado deve ser específico do Estudo de Investigação (o modelo deve ser adaptado ao estudo em causa, acrescentando outros dados considerados pertinentes ou eliminando partes não aplicáveis).
Compete ao Investigador Principal, prestar aos Participantes do estudo as informações necessárias ao consentimento livre e esclarecido.

Declaração de Consentimento Informado

Conforme a lei 67/98 de 26 de Outubro e a "Declaração de Helsínquia" da Associação Médica Mundial (Helsínquia 1964; Tóquio 1975; Veneza 1983; Hong Kong 1989; Somerset West 1996, Edinburgo 2000; Washington 2002, Tóquio 2004, Seul 2008, Fortaleza 2013) - quando se aplicar

DESIGNAÇÃO DO ESTUDO

Eu, abaixo-assinado (NOME COMPLETO DO INDIVÍDUO PARTICIPANTE DO ESTUDO) ou Eu, abaixo-assinado (nome completo do representante legal do indivíduo Participante do estudo), na qualidade de representante legal de (NOME COMPLETO DO INDIVÍDUO PARTICIPANTE DO ESTUDO) [conforme o caso]:

Fui informado de que o Estudo de Investigação acima mencionado se destina a [mencionar de uma forma clara e perceptível para o doente quais são os objetivos do estudo].

Sei que neste estudo está prevista a realização de [mencionar exames, análises, tratamentos, questionários, entrevistas, etc., conforme o caso] tendo-me sido explicado em que consistem e quais os seus possíveis efeitos. [se for o caso]

Sei que uma parte do sangue vai ser utilizada de imediato para fazer algumas análises e que outra parte vai ser armazenada para ser utilizada posteriormente. [se for o caso]

Também sei que algumas análises / exames [conforme o caso] não podem ser efetuados neste hospital e que, por isso, têm que ser realizadas noutras instituições. [se for o caso]

Sei ainda que alguns dos estudos que vão ser efetuados são estudos genéticos. [se for o caso]

Foi-me garantido que todos os dados relativos à identificação dos Participantes neste estudo são confidenciais e que será mantido o anonimato. [se for o caso]

Sei que posso recusar-me a participar ou interromper a qualquer momento a participação no estudo, sem nenhum tipo de penalização por este facto ou Sei que posso recusar-me a autorizar a participação [conforme o caso] ou interromper a qualquer momento a participação no estudo, sem nenhum tipo de penalização por este facto.

Compreendi a informação que me foi dada, tive oportunidade de fazer perguntas e as minhas dúvidas foram esclarecidas.

Aceito participar de livre vontade no estudo acima mencionado ou Autorizo de livre vontade a participação daquele que legalmente represento no estudo acima mencionado. [conforme o caso]

Concordo que sejam efetuados os exames e a colheita de amostras de sangue para realizar as análises que fazem parte deste estudo. [se for o caso]

Também autorizo a divulgação dos resultados obtidos no meio científico, garantindo o anonimato.

Nome do Investigador e Contacto: [ou do seu representante legal, se for o caso].



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DATA

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ASSINATURA

**Coordenação entre Membros na
Marcha - Avaliação, Aplicações e
Implicações Clínicas em Pessoas com
Sequelas de Acidente Vascular
Cerebral**

Ana Sofia Garcez Barbosa do Couto

SEDE ADMINISTRATIVA

- FACULDADE DE ENGENHARIA**
- FACULDADE DE ARQUITECTURA**
- FACULDADE DE BELAS ARTES**
- FACULDADE DE CIÊNCIAS DA NUTRIÇÃO E ALIMENTAÇÃO**
- FACULDADE DE CIÊNCIAS**
- FACULDADE DE DESPORTO**
- FACULDADE DE DIREITO**
- FACULDADE DE FARMÁCIA**
- FACULDADE DE LETRAS**
- FACULDADE DE MEDICINA**
- FACULDADE DE MEDICINA DENTÁRIA**
- FACULDADE DE PSICOLOGIA E DE CIÊNCIAS DA EDUCAÇÃO**
- INSTITUTO DE CIÊNCIAS BIOMÉDICAS ABEL SALAZAR**

