UNIVERSIDADE ESTADUAL PAULISTA
"JÚLIO DE MESQUITA FILHO"
INSTITUTO DE BIOCIÊNCIAS - RIO CLARO

DOUTORADO EM DESENVOLVIMENTO HUMANO E TECNOLOGIAS

NISE RIBEIRO MARQUES

EFEITO DE DOIS DIFERENTES TREINAMENTOS COM EXERCÍCIOS FÍSICOS EM VARIÁVEIS BIOMECÂNICAS E METABÓLICAS DA MARCHA DE IDOSAS

Rio Claro

2013

NISE RIBEIRO MARQUES

EFEITO DE DOIS DIFERENTES TREINAMENTOS COM EXERCÍCIOS FÍSICOS EM VARIÁVEIS BIOMECÂNICAS E METABÓLICAS DA MARCHA DE IDOSAS

Orientador: Prof. Dr. Mauro Gonçalves

Tese apresentada no Instituto de Biociências da Universidade Estadual Paulista, Campus de Rio Claro como requisito para a obtenção do título de Doutor em Desenvolvimento Humano e Tecnologias – Área: Tecnologia e Desempenho Humano.

Rio Claro

EPÍGRAFE

"Eu acredito na intuição e na inspiração. A imaginação é mais importante que o conhecimento. O conhecimento é limitado, enquanto a imaginação abraça o mundo inteiro, estimulando o progresso, dando à luz à evolução. Ela é, rigorosamente falando, um fator real na pesquisa científica."

Albert Einstein (1879-1955)

AGRADECIMENTOS

Em primeiro lugar agradeço a Deus e minha família, que sempre me proporcionaram apoio e carinho inesgotáveis. Aos meus pais, Francisco e Jacira por me ensinarem a importância do estudo e que nunca mediram esforços para darem a mim e minha irmã a melhor educação. Agradeço a Camilla pela paciência e companheirismo nessa jornada. Obrigada Camilla por ser meu porto seguro quando precisei e por dividir cada momento desses anos de doutorado.

A todos os colegas e amigos do Laboratório de Biomecânica gostaria de desejar um muito obrigada! Em destaque, aos que me acolheram no início, como a Carina, e também aos que vieram depois e se tornaram grandes amigos: Antônio, Luciano, Fernanda, Aline, Deborah, Renata, Denise, Mary e Alex.

Agradeço aos membros da banca examinadora Prof. Dr. Adalgiso Croscato Cardozo, Prof Dr. Fábio Viadanna Serrão, pelas contribuições dadas ao trabalho e as Prof^{as}. Dr^{as}. Camila Coelho Greco e Daniela Cristina Leite de Abreu pelas contribuições para a qualificação e defesa, bem como, pelo exemplo como mulheres pesquisadoras e grandes profissionais.

Agradeço a dois docentes que marcaram meus estudos, aqueles que me inspiram enquanto futura docente. Ao Prof. Dr. Marcelo Tavella Navega, um amigo, que sempre me aconselhou da melhor maneira nos estudos e também nos rumos que poderia seguir em minha carreira, uma pessoa de excelentíssimo caráter e digno de todas as suas conquistas pessoas e profissionais. E ao Prof Dr Dain LaRoche, meu supervisor de doutorado sanduíche, e a Universidade de New Hampshire, pela recepção que ele e sua universidade me ofertaram. Eu não sei se ele faz idéia de quanto o tempo que estive na Universidade de New Hampshire contribui para minha vida em diversos aspectos. Mas, sinto que essa experiência foi uma das melhores partes da minha pós-graduação e após essa experiência eu acredito ter me tornado uma melhor aluna, melhor pesquisadora e melhor futura professora. O Professor Dain LaRoche realmente me inspirou e espero um dia poder me tornar uma professora, pesquisadora e um ser humano como ele.

Em especial, gostaria de agradecer ao meu orientador Prof. Dr. Mauro Gonçalves por esses anos em que estive sob sua orientação. Professor, muito obrigada por todas as vezes em que o senhor sentou ao meu lado para me ensinar. Cada uma dessas conversas que me fizeram crescer muito. Obrigada, por cada ensinamento, por todos os por quês, todos os elogios e “puxões de orelha”, que o senhor já me deu. Espero que estes cinco anos, não somente de orientação, mas de amizade, perdurem por mais outros tantos.

Por fim, gostaria de agradecer ao apoio financeiro da Fundação de Amparo a Pesquisa do Estado de São Paulo (FAPESP), que foi fundamental para a conclusão deste estudo.

SUMÁRIO

RESUMO

Durante a marcha a atuação adequada de todos os segmentos corporais é essencial para manter a estabilidade. Contudo, ao longo do processo de envelhecimento humano, alterações no sistema neuromuscular podem reduzir a habilidade de executar tarefas motoras complexas, como a marcha, e aumentar o risco de quedas na população idosa. Nesse sentido, os objetivos do presente estudo foram: correlacionar o custo energético, ativação muscular e parâmetros biomecânicos da marcha de idosas caidoras e não caidoras; correlacionar força articular do quadril, joelho e tornozelo com parâmetros biomecânicos da marcha de idosas caidoras e não caidoras; comparar a ativação eletromiográfica dos músculos estabilizadores do tronco e de membros inferiores durante a marcha de mulheres jovens e idosas; e identificar o efeito do treinamento de oito semanas com o Método Pilates no padrão de ativação eletromiográfico dos músculos estabilizadores do tronco e de membros inferiores durante a marcha de mulheres idosas. Participaram do estudo trinta e oito idosas (68,21 anos) e quinze mulheres jovens (22,13 anos). O protocolo para coleta de dados foi realizado em duas diferentes visitas ao ambiente de coleta. No primeiro dia, foram coletados dados de força articular do quadril, joelho e tornozelo, bem como, foi realizada a familiarização dos sujeitos com a caminhada na esteira em velocidade de preferência. Durante a segunda visita, foram coletados: o consumo de oxigênio na posição sentada e durante caminhada em velocidade de preferência; e parâmetros biomecânicos (eletromiografia e cinemática) da marcha. De acordo com nossos achados, em idosas não caidoras o aumento da idade e da coativação dos músculos da coxa (reto femoral/glúteo máximo) foram os únicos fatores relacionados ao aumento do custo energético durante a marcha, enquanto que para idosas caidoras, o aumento da coativação dos músculos da perna (tibial anterior/gastrocnêmio lateral) e o aumento da ativação do glúteo máximo foram relacionados com o aumento do custo energético durante a marcha. Nossos resultados indicam que a redução da força de abdução do quadril e extensão do joelho promovem alterações biomecânicas no recrutamento dos músculos estabilizadores do tronco, quadril e joelho durante a fase de apoio da marcha em idosos caidores e não caidores. Para ambos os grupos de idosas foi encontrado que a redução da força está relacionada à redução da velocidade da marcha. Encontramos que idosas apresentam redução da ativação dos músculos estabilizadores do tronco (oblíquo interno e multífido) e de membros inferiores (reto femoral, bíceps femoral e glúteo máximo) em comparação com mulheres jovens. Além disso, nossos resultados indicaram que com a intervenção de oito semanas com o Método Pilates houve aumento da ativação dos músculos oblíquo interno e reto femoral na fase de apoio da marcha em mulheres idosas, bem como, redução do período de *onset* do músculo oblíquo interno. Portanto, identificamos que a redução da força do quadril e o aumento da ativação dos músculos estabilizadores dessa região acarretam em elevação do custo energético durante a marcha. Além disso, que o processo de envelhecimento causa redução da velocidade da marcha e da ativação dos músculos estabilizadores do tronco e membros inferiores. Nesse sentido, a intervenção com o Método Pilates pode ser indicada para aumentar a estabilidade articular do tronco e dos membros inferiores durante a marcha de idosas.

Palavras-chave: Envelhecimento, Biomecânica, Cinemática, Quedas, Reabilitação

1. APRESENTAÇÃO

Esta tese de doutorado é composta de introdução e quatro artigos científicos. Em consonância com as regras do Programa de Pós-Graduação em Desenvolvimento Humano e Tecnologias, os artigos foram redigidos de acordo com as normas dos periódicos: *Clinical Biomechanics*, *Isokinetics and Exercise Science Journal, Journal of Electromyography and Kinesiology and Physiotherapy*. Assim, este material é composto das seguintes seções:

- **Introdução**
- **Artigo I:** ***Association between energy cost of walking, muscle activation, and biomechanical parameters in older female fallers and non-fallers***

Nise Ribeiro Marques, Dain Patrick LaRoche, Camilla Zamfolini Hallal, Luciano Fernandes Crozara, Mary Hellen Morcelli, Aline Harumi Karuka, Marcelo Tavella Navega, Mauro Gonçalves

Artigo publicado no periódico *Clinical Biomechanics,* DOI: http://dx.doi.org/10.1016/j.clinbiomech.2013.01.004, em Fevereiro de 2013.

- **Artigo II:** ***Lower limb strength is associated with gait biomechanical abnormalities in older female fallers and non-fallers***

Nise Ribeiro Marques, Camilla Zamfolini Hallal, Luciano Fernandes Crozara, Mary Hellen Morcelli, Aline Harumi Karuka, Marcelo Tavella Navega, Mauro Gonçalves

Artigo publicado no periódico *Isokinetics and Exercise Science Journal*, DOI: http://10.3233/ies-130491, em Janeiro de 2013.

- **Artigo III:** ***Age-related alterations on trunk and lower limb stabilizer muscles activation during walking***

Nise Ribeiro Marques, Camilla Zamfolini Hallal, Luciano Fernandes Crozara, Mary Hellen Morcelli, Aline Harumi Karuka, Marcelo Tavella Navega, Mauro Gonçalves

Artigo submetido ao *Journal of Electromyography and Kinesiology*

- **Artigo IV: Eight-weeks of Pilates Method intervention increases trunk and lower limb stabilizer muscles activation during walking in older female adults**

Nise Ribeiro Marques, Camilla Zamfolini Hallal, Mary Hellen Morcelli, Luciano Fernandes Crozara, Aline Harumi Karuka, Marcelo Tavella Navega, Mauro Gonçalves

Artigo a ser submetido para *Physiotherapy*

2. INTRODUÇÃO

A marcha humana é um dos movimentos mais frequentemente executados durante as atividades de vida diária e promove importante independência funcional ao indivíduo (WERT et al., 2010). Durante a marcha, a atuação adequada de todos os segmentos corporais (membros inferiores, superiores e tronco) é essencial para manter a estabilidade e evitar a ocorrência de quedas (ANDERS et al., 2007a). Contudo, durante o processo de envelhecimento humano, alterações nos sistemas sensório-motor e músculo-esquelético podem reduzir a habilidade de executar tarefas motoras complexas, tal como a marcha, aumentado assim, o risco de quedas na população idosa (YEN et al., 2009).

As quedas estão entre as principais causas de morte e lesão corporal em idosos, representando cerca de 45% dos casos de óbito nesta população (SCHULZ, LlOYD e WILLIAM, 2010). As consequências das quedas são potencialmente danosas, pois contribuem para a prevalência de problemas de saúde e diminuição da qualidade de vida, além de acarretar um elevado custo ao sistema de saúde (LORD e DAYHEW, 2001; CHAMBERS e CHAN, 2007; VAN DIEEN e PIJNAPPELS, 2008). Estima-se também que 50% das quedas em idosos ocorrem durante a marcha e isto pode estar associado a modificações biomecânicas no padrão deste movimento (DEVITA e HORTOBÁGYI, 2000; HAHN e CHOU, 2004; TALBOT et al, 2005; BAIRD e RICHARD, 2009; HOLLMAN, YOUDAS e LANZINO, 2009; HORTOBÁGYI et al, 2009).

Considerando que a etiologia da queda é multifatorial, o entendimento amplo de fatores biomecânicos e fisiológicos, que podem aumentar o risco deste evento em idosos é de extrema relevância (LAROCHE et al., 2010). Nesse sentido, destacam-se entre os fatores intrínsecos, relacionados às alterações fisiológicas do envelhecimento, a redução da capacidade de gerar força muscular, perda da eficiência sensório-motora em respostas à perturbações no equilíbrio e alterações no recrutamento muscular, o que ocasionam alterações biomecânicas na realização da marcha e no aumento do custo energético durante a execução deste ato motor (SKELTON et al., 2002; BEAN et al., 2002; PETRELLA et al., 2004; PIJNAPPELS et al., 2008).

A literatura atual aponta que a redução da força muscular dos segmentos do membro inferior pode acarretar em diminuição da velocidade da marcha, da altura, cadência e do comprimento do passo, variáveis estas determinantes para o risco de quedas durante o ato de andar (PIJNAPPELS et al., 2008, van DIEEN e PIJNAPPELS, 2008). Contudo, segundo Burnfield et al. (2000), que avaliou o torque isocinético de movimentos no plano sagital das articulações do tornozelo, joelho e quadril (flexão e extensão), apenas a redução da força dos músculos extensores do quadril são determinantes para influenciar as variáveis temporais e espaciais da marcha.

Embora diversos estudos na literatura recentes tenham explorado as principais causas de quedas em idosos, o foco da maioria destes está relacionado à extremidade inferior, principalmente, no que tange as articulações do joelho e tornozelo. A contribuição biomecânica das articulações do quadril e do tronco durante a marcha e para manter ou recuperar o equilíbrio diante de uma perturbação externa ainda é pouco conhecida, embora diversos autores considerem crucial que haja estabilidade nestas regiões anatômicas para evitar a queda (ANDERS et al., 2007a; WERT et al., 2010).

Haja vista que as quedas na população idosa são eventos, amplamente, responsável pela causa de lesões e acarretam elevado custo ao sistema de saúde, a busca por intervenções terapêuticas com o objetivo preventivo é extremamente relevante. A prática de atividade física regular, com exercícios tradicionais, durante a senescência tem sido amplamente descrita como um importante fator para manutenção do equilíbrio e diminuição do risco de quedas (CARVALHO et al., 2010). Além disso, recentemente, diversos métodos de exercícios físicos, que buscam a automatização de novos padrões de recrutamento muscular, vêm sendo aplicados para a prevenção desse evento na população idosa (SIQUEIRA-RODRIGUES et al., 2010).

Entre as diversas técnicas de exercícios que podem promover essas modificações no recrutamento muscular e no controle neuromotor estão o Método Pilates e os exercícios com haste vibratória, os quais vêm sendo aplicados por diversos profissionais da fisioterapia e educação física (HALLAL et al., 2011; GONÇALVES et al., 2012; SIQUEIRA-RODRIGUES et al., 2010).

O Método Pilates é uma modalidade de exercícios, criada no início do século XX, que possui como princípios básicos: a Concentração, o Controle, a Fluidez, a Precisão, a Respiração e o Centrando (MARQUES et al., 2012a). Entre esses princípios, destaca-se o Centrando, que consiste na contração isométrica do músculo oblíquo interno, essa contração deve ocorrer em todos os exercícios, com intuito de prover estabilidade à região lombo-pélvica (MARQUES et al., 2012a). Além disso, assim como na marcha, os exercícios do Método Pilates exigem a integração entre os movimentos rítmicos de membros superiores e de membros inferiores, juntamente, com a estabilização do tronco.

Já, o treinamento com haste vibratória distingue-se dos demais treinamentos com vibração pela menor frequência alcançada e pelo comportamento passivo da haste, uma vez que, a vibração não é produzida pelo equipamento, mas pela contração muscular (HALLAL et al., 2010a). Diversos estudos apontaram que exercícios com haste vibratória promovem maior ativação dos músculos estabilizadores do tronco, o que pode acarretar em efeitos positivos na estabilização do tronco e, consequentemente, no equilíbrio (HALLAL et al., 2010b; GONÇALVES et al., 2012; MARQUES et al., 2012b).

3. ARTIGOS

3.1. Artigo I – Publicado na *Clinical Biomechanics*

Association between energy cost of walking, muscle activation, and biomechanical parameters in older female fallers and non-fallers

Nise Ribeiro Marques[a,*], Dain Patrick LaRoche[b], Camilla Zamfolini Hallal[a], Luciano Fernandes Crozara[a], Mary Hellen Morcelli[a], Aline Harumi Karuka[a], Marcelo Tavella Navega[c], Mauro Gonçalves[a]

[a] Department of Physical Education, São Paulo State University, BRA

[b] Department of Kinesiology, University of New Hampshire, USA

[c] Department of Physical Therapy and Occupational Therapy, São Paulo State University, BRA

Word count paper: 4.284 words

Word count abstract: 250 words

Tables: 4

Figures: 1

Abstract

Objective: To determine the nervous activation, muscle strength, and biomechanical parameters that influences the cost of walking in older fallers and nonfallers.

Methods: Maximal voluntary isokinetic torque was measured for the hip, knee and ankle of older women. Oxygen consumption was measured at rest and during eight minutes of walking at self-selected speed. An additional minute of walking was performed to collect kinematic variables and the electromyographic signal of trunk, hip, knee, and ankle muscles, which was analyzed by the linear envelope. Cost of walking was calculated by subtracting resting body mass-normalized oxygen consumption from walking body mass-normalized oxygen consumption. Stride time and length, and ankle and hip range of motion were calculated from kinematic data.

Findings: Older adult fallers had 28% lower knee extensor strength ($p = 0.01$), 47% lower internal oblique activation at heel contact ($p = 0.03$), and higher coactivation between tibialis anterior and gastrocnemius lateralis in each of the gait phases ($p < 0.05$). For fallers, a higher activation of gluteus maximus was associated with a higher cost of walking ($r = 0.55$, $p < 0.05$ and $r = 0.71$, $p < 0.01$, before and after heel contact, respectively). For non-fallers, an association between cost of walking and age ($r = 0.60$, $p = 0.01$) and cost of walking and thigh muscle coactivation ($r = 0.53$, $p = 0.01$) existed.

Interpretation: This study demonstrated that there may be links between lower-extremity muscle weakness, muscle activation patterns, altered gait, and increased cost of walking in older fallers.

Keywords Aging; Fall risk; EMG; Gait; Kinematic

1. Introduction

Aging causes changes in the neuromuscular system such as strength loss and alteration of muscle activation, which may result in reduced walking performance (DeVita and Hortobágyi, 2000; Malatesta et al., 2004; Schimitz et al., 2008; Wert et al., 2010; Hortobágyi et al., 2011). Since most falls occur during walking, poor gait performance has been suggested to be a major risk factor for falls in older adults (Burnfield et al., 2000; Kerrigan et al., 2000). In addition, the energy cost of walking (Cw) has been shown to be 23% higher in older adults than in young (Peterson and Martin, 2010). The higher energy expenditure during walking increases the sense of task effort, is likely to facilitate fatigue, and therefore may contribute to increased risk of falling (Hortobágyi et al., 2011). However, to the authors' knowledge no study has yet investigated whether the Cw differs between older fallers and non-fallers, and subsequently, what neuromuscular and gait factors are related to the Cw in these groups.

Age-related loss of muscles strength is considered a well-known risk factor for falls in elderly people (Skelton et al., 2002; Mian et al., 2006; Bento et al., 2010) but it may also be related to Cw. Several studies have indicated that peak magnitudes of lower-limb voluntary torque are related with balance recovery responses and history of falls (Robino et al, 2002; Pijnappeels et al, 2008; LaRoche et al, 2010). This concept is also supported by epidemiological studies that identified a greater risk for falls among older adults with lower-limb weakness (Robino et al, 2002). In addition, it is also possible that reduced strength may be related to an increased Cw in older adults because the ability to absorb work during lengthening contractions (and return energy elastically) may be compromised (LaStayo et al., 2003, Whittington et al., 2008). As such, there may be an interaction among lower limb strength, the Cw, and risk of falling in older adults that has not been adequately studied.

Slow gait speed is another factor that increases the risk for falls in older adults (Kelsey et al, 2012; Perracini et al, 2012; Callisaya et al, 2011), and, older adults with slow gait speed have higher Cw (Hortobágyi et al., 2011; Malatesta et al., 2003; Mian et al., 2007; Wert et al., 2010). According, to Mian et al (Mian et al., 2007) twelve months of aerobic training improved gait speed by 6% but did not change Cw. Consequently, the authors considered that reduced gait speed alone cannot explain the greater Cw in older people (Mian et al, 2006) and therefore other neuromuscular or biomechanical factors may help explain the increased Cw in the aged. Nevertheless, Wert et al. (Wert et al., 2010) considered slow gait as an obvious contributing factor to the increase in Cw seen in this population. Despite the expectation that older fallers have slower gait speed, and expectedly a greater Cw, a knowledge of how biomechanical gait parameters affect the Cw in older fallers and non-fallers is lacking.

A neuromuscular factor which has been associated with greater Cw in older adults is muscular coactivation, that is, the cocontraction of agonist and antagonist muscles. Previous studies demonstrated an age-related adaptation in the recruitment pattern of lower limb muscles during gait such that increased antagonist coactivation was related to a significantly higher Cw in old adults (Hortobágyi et al., 2011, Peterson and Martin, 2010). However, Peterson et al. (Peterson and Martin, 2010) demonstrated that this association was more clinically significant for the coactivation of thigh muscles than for those of the shank (Peterson and Martin, 2010). To the authors' knowledge no study has simultaneously examined the contribution of trunk and lower limb activation to the Cw in older adults. Though, Saha et al. (Saha et al., 2007) indicated that trunk flexion postures, commonly adopted by older adults, increase the oxygen consumption rate by 30-60%.

It is likely that in older adults declining neuromuscular performance leads to alterations in gait biomechanics that subsequently result in increased Cw, and, it is likely that this relationship may differ between older fallers and nonfallers. Thus the aim of this study was to provide a comprehensive assessment of the nervous activation, muscle strength, and biomechanical parameters that influence the Cw in older fallers and nonfallers. A second aim was to determine the association between age, preferred treadmill walking speed (PTWS), and Cw. We hypothesized that those individuals with altered gait biomechanics, including greater muscle activation, decreased stride time, stride length, and joint range of motion (hip and ankle) would have an increased Cw at PTWS. It was also hypothesized that poor hip, knee, and ankle strength, slow PTWS, and older age would be associated with higher Cw.

2. Methods

2.1. Study Design and Participants

Gait biomechanics, hip, knee and ankle maximal voluntary torques, and the Cw in the current cross-sectional study were evaluated as part of the baseline assessment for a randomized controlled trial of two interventions for improving walking in older adults. Gait measurements were collected during two baseline clinic visits.

Data of thirty-seven older, adult women between the ages of 60-85 yr were considered for this study (Table 1). The participants were divided in two groups based on having fallen or not in the period of one year before the evaluation. A fall was defined as any balance perturbation which caused the person's body to have significant contact with the floor. All participants signed a consent form approved by the Institutional Ethics Committee. People who had uncontrolled cardiovascular disease, diagnosed dementia or cognitive impairment (defined as a Mini-Mental State Examination score < 20), balance disturbance (defined as a BERG balance score < 36), hemiparesis, pain in the lower limbs or trunk, or a progressive motor disorder were excluded.

Table 1 Subject characteristics.

Variable	**Older Faller (n=15)**	**Older Non-Faller (n=22)**	**P**
Age (years)	69.6 (8.1)	66.1 (6.2)	0.1
Body Mass (kg)	66.8 (9.2)	65.3 (13.6)	0.6
Height (m)	1.51 (0.06)	1.54 (0.05)	0.2
Body mass index ($kg \cdot m^{-2}$)	28.9 (3.2)	27.4 (4.9)	0.2
BERG Balance Scale	54.6 (1.6)	54.7 (1.5)	0.8
Mini-Mental Examination Score	22.9 (4.8)	24.1 (6.05)	0.5
Number of falls (per year)	1.8 (1.3)	-	-
Basal metabolic rate ($ml.kg^{-1}.min^{-1}$)	4.2 (0.7)	3.9 (0.5)	0.3
Metabolic rate during walking ($ml.kg^{-1}.min^{-1}$)	13.3 (3.1)	11.8 (1.9)	0.1

2.2. Procedures

The volunteers visited the laboratory on two separate occasions within 24-72 hours. On the first day of data collection the preferred overground and treadmill walking speeds (POWS and PTWS), and the hip, knee and ankle maximal voluntary joint torques were measured. During the second day of data collection, the volunteers were familiarized with treadmill walking at PTWS for 10 minutes and then the energy expenditure during seated rest was measured via indirect calorimetry. Next the Cw at PTWS was measured followed by kinematic and electromyographic assessment of gait (Figure 1).

Kinematic gait parameters, Cw, electromyograms (EMG), and joint torques were assessed by: a motion analysis system with seven infrared cameras (Vicon Motus, Oxford, UK®); a metabolic measurement system (Quark PFT, Cosmed®, Rome, ITA); an 8-channel, telemetered electromyogram (Noraxon®, Phoenix, USA); and an isokinetic dynamometer (Biodex®, New York, USA).

For the kinematic gait parameters, 39 markers were fixed on the head, trunk, upper and lower limbs, according to the manufacturer's Plugin Gait Full Body model. The kinematic parameters were recorded at a sample frequency of 100 Hz. Also, heel contact and toe-off were detected by two footswitches (Noraxon®, Phoenix, USA), composed of four sensors attached on both feet at the heel, first and fifth metatarsals, and toe, which were synchronized with the EMG and motion analysis system.

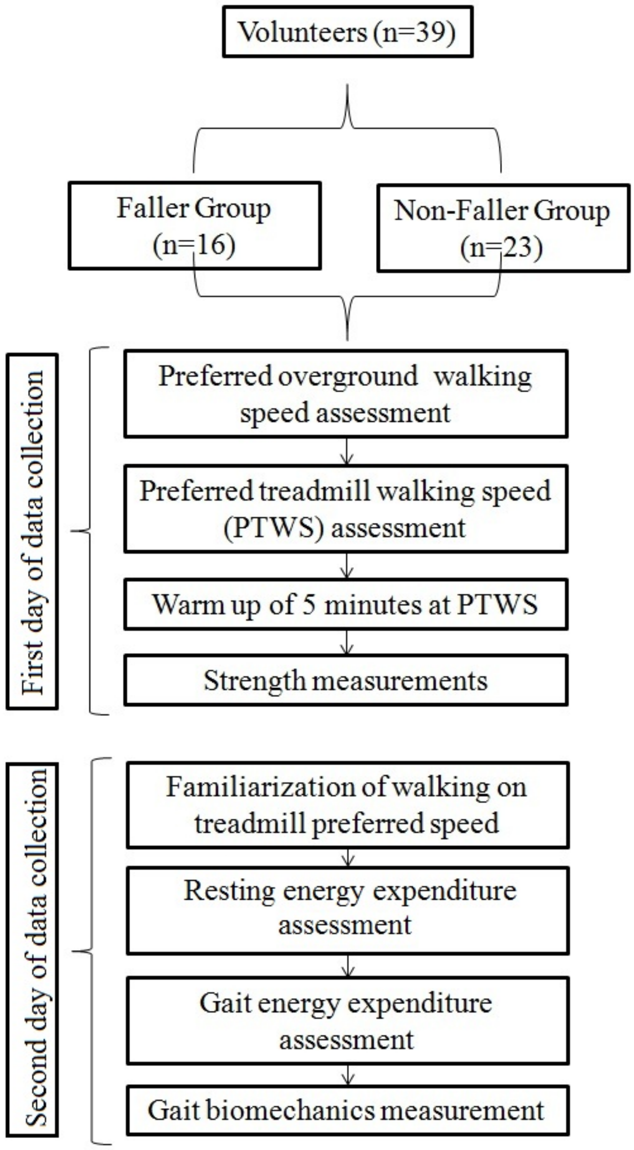

Figure 1 Procedures of data collection.

EMG signals were collected at sample frequency of 2000 Hz, using Ag/AgCl disc electrodes (Miotec®, Porto Alegre, Brazil) with an active area of 1 cm^2 and inter-electrode distance of 2 cm arranged in bipolar configuration. The electrodes were positioned on the participants' right side on the internal oblique muscles (IO) according to Marshal and Murphy 2003, and on multifidus (MU), gluteus maximus (GM), biceps femoris (BF), rectus femoris (RF), tibiallis anterior (TA), and gastrocnemius lateralis

(GL) muscles according to Hermens et al. 2000. Before placing the electrodes, the subject's skin was shaved and cleaned with alcohol to reduce impedance (Hermens et al. 2000).

2.3. Measures

2.3.1. Measurement of hip, knee, and ankle strength

Before strength assessment the volunteers performed a warm up of five minutes walking on the treadmill at PTWS (Burnfield et al., 2000). Maximal joint torques were then measured isokinetically at 120 $deg^{-1}.s^{-1}$ for the hip, knee, and ankle for both flexion and extension movements. The order of joint testing was randomized, and the procedure and desired movement pattern were explained to the volunteers before testing each joint. Five submaximal repetitions were performed for familiarization purposes, and five maximal repetitions were collected for analysis. Strong verbal encouragement to obtain maximal effort was provided.

Hip extension and hip flexion torque were measured with each subject in a supine position. A resistance pad was secured with a strap around the thigh of the tested limb, and each subject's pelvis and trunk were stabilized on the table using belts. The dynamometer was aligned to the approximate axis of rotation of the tested hip joint (superior and anterior to the greater trochanter). The test was initiated with the hip flexed at 10º and hip flexion torque was assessed as the subject moved from this position toward 60º of hip flexion, while hip extension torque was assessed during movement from the flexed position toward 10º of hip flexion (Pavol et al., 2002).

Isokinetic knee flexion and extension torques were measured with the volunteer seated and the hip flexed at 90º. The dynamometer was aligned to the approximate axis of rotation of the knee joint (a line traversing the femoral epicondyles), and the resistance pad was placed on the tibia (just proximal to the superior border of the medial malleolus). The subject's thigh, trunk, and pelvis were stabilized with straps, and subjects crossed their arms in front of the chest throughout the test (Burnfield et al., 2000). Testing was initiated at 90º of knee flexion with the initial movement toward 60º of extension (Hartman et al., 2009).

Isokinetic plantar flexion and dorsiflexion torques were measured with each subject lying supine. The knees and hips were flexed and the ankle was in neutral inversion-eversion. The dynamometer was aligned to approximate the axis of rotation of the tested ankle joint (the projection of a line passing obliquely through the distal tip of the tibia and fibula), and the foot was strapped securely to the foot plate. Proximal thigh and trunk stabilization (using belts) was provided to prevent unwanted movement. The test was initiated with subjects in 10º of dorsiflexion with the initial movement toward 30º of plantar flexion (Hartman et al., 2009).

2.3.2. Gait analysis, EMG activity, and cost of walking

Preferred overground walking speed (POWS) was determined on the first day of data collection. Volunteers were instructed to walk at their self-selected speed at a natural pace over 20 m and the duration for each trial was determined with two infrared timing gates placed at 5 and 15 m. POWS was calculated by dividing distance walked (10 m) by the time to cover this distance (s).

PTWS was determined by starting treadmill walking at 50% of the POWS. Then, speed was increased until the subject reported that the current speed was faster than preferred gait speed and slowly decreased until the subject reported that the current speed was slower than preferred. This procedure was repeated three times and the average of the 3 ‘‘faster’’ and 3 ‘‘slower’’ than preferred speeds was taken as the subject’s PTWS.

In the second day of data collection, volunteers were familiarized to the PTWS for ten minutes, followed by an additional eight minutes of walking during which expired gases were collected to determine the Cw. Energy cost was determined via indirect calorimetry and analysis of expired gases. Oxygen uptake (VO_2) was recorded during the final eight minutes of walking at PTWS. After the VO_2 data were collected, and while still walking at PTWS, one minute of EMG and kinematic data were recorded.

2.4. Data Analysis

A measure of mean VO_2 was determined during the physiological steady state collected between minutes three and six of the eight-minute assessment. The energy cost of walking was calculated by subtracting the resting body mass-normalized VO_2, collected in a seated and relaxed position, from the body mass-normalized VO_2 during walking, and this difference was divided by the chosen gait speed of each volunteer (Parvataneni et al., 2009).

The EMG signal was processed in specific routines developed in Matlab (Mathworks®, Natick, USA) using full-wave rectification and a low pass, fourth order filter with a cut-off frequency of 10 Hz. Then, the mean of the linear envelope of the EMG signal was obtained 100 ms before and after heel contact, and before and after toe-off of the first ten strides. All the linear envelope values were normalized to the peak activation obtained during the gait. Additionally, to measure the antagonist co-contraction we calculated the ratio of activation between trunk (IO/MU), thigh (RF/GM and RF/BF) and shank (TA/GL) muscles.

Gait speed, stride length, stride time, ankle angle at heel contact and hip angle at toe-off were analyzed during the first ten strides of the one-minute kinematic analysis (Shiavi et al., 1998). A fourth-order Butterworth filter with a cut-off frequency of 6 Hz was used to smooth marker trajectories. We prioritized these kinematic variables because gait speed, stride length, and stride time represent temporal and spatial gait adjustments, and ankle dorsiflexion angle at heel contact and hip extension angle at toe-off have been shown to be related to the mechanical efficiency of gait (Wert et al., 2010).

PASW 18.0 (SPSS inc.) was used for all statistical analyses and means and standard deviations were used to summarize participant characteristics. Then, the Shapiro-Wilk test was used to determine if the data were normally distributed, and t-tests for independent samples were used to compare the electromyographic and kinematic gait variables, and joint torques between elderly fallers and nonfallers. Pearson and Spearman correlation coefficients were computed, as appropriate, to quantify the association between age, PTWS, and Cw and each of the gait and torque measures. The significance level for all statistical tests was set at $p<0.05$.

3. Results

When comparing older female fallers and non-fallers, we found that knee extensor maximal voluntary torque was 28% higher in non-fallers than in fallers ($p = 0.01$) and the hip angle at toe-off was four degrees higher in non-fallers than in fallers ($p = 0.01$; Table 2). IO activation before heel contact was 47% higher in the non-faller group and BF activation before and after heel contact was 31% and 33% higher (respectively) in fallers, while GM activation before toe-off was 39% higher in fallers. Antagonist co-contraction ratio between shank muscles (TA/GL) was higher in fallers in all gait phases (before heel contact, $p = 0.001$; after heel contact, $p = 0.002$; before toe-off, $p = 0.005$; and after toe-off, $p = 0.036$; Table 3), whereas coactivation of trunk (IO/MU) and thigh (RF/GM) did not differ between groups.

Table 2 Preferred treadmill walking speed, energy cost of walking, isokinetic torque of hip and ankle joints, and biomechanical variables of gait. Values are mean (standard deviation).

Variable	**Older faller (n=15)**	**Older Non-Faller (n=22)**	**P**
Cost of walking ($ml.kg^{-1}.min^{-1}.m.s^{-1}$)	9.3 (4.3)	8.7 (2.9)	0.6
Overground walking speed ($m{\cdot}s^{-1}$)	1.1 (0.1)	1.3 (0.2)	0.06
Treadmill walking speed ($m{\cdot}s^{-1}$)	0.9 (0.1)	0.9 (0.1)	0.6
Knee flexor torque ($N{\cdot}m{\cdot}kg^{-1}$)	0.54 (0.26)	0.55 (0.14)	0.8
Knee extensor torque ($N{\cdot}m{\cdot}kg^{-1}$)	**0.69 (0.2)**	**0.96 (0.24)**	**0.02***
Hip flexor torque ($N{\cdot}m{\cdot}kg^{-1}$)	0.68 (0.26)	0.74 (0.19)	0.4
Hip extensor torque ($N{\cdot}m{\cdot}kg^{-1}$)	0.78 (0.23)	0.89 (0.36)	0.2
Ankle flexor plantar torque ($N{\cdot}m{\cdot}kg^{-1}$)	0.31 (0.19)	0.33 (0.13)	0.7
Ankle dorsiflexor torque ($N{\cdot}m{\cdot}kg^{-1}$)	0.28 (0.11)	0.31 (0.13)	0.3
Stride time (second)	2.3 (0.89)	2.6 (0.9)	0.3
Stride length (mm)	509.3 (61.9)	497.2 (70.7)	0.6

Ankle angle at heel contact (degrees)	6.4 (4.3)	5.9 (4.4)	0.2
Hip angle at toe-off (degrees)	**9.5 (4.8)**	**5.4 (4.7)**	**0.01***
IO activation before heel contact (% mean)	**8.3 (4.8)**	**15.7 (12.4)**	**0.03***
RF activation before heel contact (% mean)	12.7 (5.5)	15.2 (12.1)	0.4
TA activation before heel contact (% mean)	40.01 (24.6)	30.05 (13.6)	0.1
MU activation before heel contact(% mean)	16.4 (10.4)	18.4 (12.3)	0.6
GM activation before heel contact (% mean)	12.03 (8.07)	16.8 (17.7)	0.3
BF activation before heel contact (% mean)	**45.5 (21.4)**	**31.3 (17.1)**	**0.03***
GL activation before heel contact (% mean)	7.2 (3.8)	14.9 (16.4)	0.08
IO activation after heel contact (% mean)	97.2 (38.1)	100.3 (29.4)	0.7
RF activation after heel contact (% mean)	143.8 (46.03)	130.6 (33.5)	0.3
TA activation after heel contact (% mean)	106.7 (21.4)	122.8 (31.6)	0.1
MU activation after heel contact(% mean)	150.5 (38.8)	147.7 (40.8)	0.8
GM activation after heel contact (% mean)	154.7 (44.3)	179.9 (52.2)	0.1
BF activation after heel contact (% mean)	**36.4 (22.8)**	**24.1 (8.3)**	**0.02***
IO activation before toe-off (% mean)	117.5 (38.7)	105.09 (29.5)	0.2
RF activation before toe-off (% mean)	89.5 (41.2)	80.09 (28.4)	0.4
MU activation before toe-off (% mean)	76.1 (27.3)	82.8 (37.3)	0.5
GM activation before toe-off (% mean)	**86.4 (39.6)**	**52.3 (28.4)**	**0.005***
BF activation before toe-off (% mean)	43.8 (47.1)	50.1 (41.03)	0.6
GL activation before toe-off (% mean)	91.7 (28.3)	75.8 (24.8)	0.08
IO activation after toe-off (% mean)	21.6 (25.4)	20.5 (12.9)	0.8
RF activation after toe-off (% mean)	10.9 (5.4)	15.9 (20.9)	0.3
TA activation after toe-off (% mean)	35.1 (19.8)	31.9 (13.9)	0.8
MU activation before toe-off (% mean)	12.2 (13.7)	11.6 (14.07)	0.8

GM activation after toe-off (% mean)	6.8 (6.4)	10.1 (17.6)	0.4
BF activation after toe-off (% mean)	16.2 (16.8)	13.08 (9.4)	0.4
GL activation after toe-off (% mean)	7.9 (3.6)	12.2 (9.6)	0.1

* p<0.05 significant difference between faller and non-fallers groups.

Table 3 Antagonist cocontraction ratio. Values are mean (standard deviation).

Antagonist Co-contraction ratio	**Older faller (n=15)**	**Older Non-Faller (n=22)**	**P**
IO/MU before heel contact	0.63 (0.46)	1.16 (1,15)	0.104
RF/GM before heel contact	3.84 (0.761)	1.93 (2.94)	0.291
RF/BF before heel contact	0.34 (0.18)	0.5 (0.39)	0.144
TA/GL before heel contact	**6.23 (3.5)**	**3.11 (1.92)**	**0.001***
IO/MU after heel contact	0.61 (0.39)	1.25 (1.89)	0.281
RF/GM after heel contact	0.92 (0.46)	1.22 (1.89)	0.557
RF/BF after heel contact	0.84 (0.59)	0.83 (0.56)	0.998
TA/GL after heel contact	**6.21 (3.19)**	**3.48 (1.59)**	**0.002***
IO/MU before toe-off	1.61 (1.12)	2.32 (2.23)	0.266
RF/GM before toe-off	2.01 (1.59)	2.46 (1.95)	0.469
RF/BF before toe-off	1.68 (1.75)	2.49 (4.33)	0.499
TA/GL before toe-off	**5.28 (2.85)**	**2.73 (2.25)**	**0.005***
IO/MU after toe-off	1.9 (1.7)	2.67 (2.02)	0.245
RF/GM after toe-off	2.25 (1.57)	3.23 (3.1)	0.268
RF/BF after toe-off	1.09 (0.65)	1.6 (1.51)	0.235
TA/GL after toe-off	**6.3 (2.78)**	**3.91 (3.54)**	**0.036**

* p < 0.05 significant difference between faller and non-fallers groups.

Table 4 shows Pearson and Spearman correlation coefficients between age, PTWS, Cw, gait parameters, and maximal voluntary joint torques of the hip, knee and ankle for fallers and non-fallers. For fallers, we found associations between age and BF activation after heel contact (r = 0.56, p = 0.03), age and IO/MU before contact (r = 0.57, p = 0.02), age and RF/GM before toe-off (r = 0.54, p = 0.03), age and TA/GL before toe-off (r = 0.52, p = 0.04). Also in fallers, correlations existed begtween PTWS and POWS (r = 0.55, p = 0.03), and PTWS and stride length (r = 0.79, p = 0.006). Fallers also had associations between gait and muscle activations measures including PTWS and BF activation after toe-off (r = -0.54, p = 0.04), PTWS and RF/BF before toe-off (r = 0.58, p = 0.02), Cw and GM activation before heel contact (r = 0.55, p = 0.3) and, Cw and GM activation after heel contact (r = 0.71, p = 0.004). Non fallers had associations between age and Cw (r = 0.60, p = 0.004), and age and IO/MU in all gait phases (before heel contact, r = 0.47, p = 0.02; after heel contact, r = 0.64, p = 0.001; before toe-off, r = 0.44, p = 0.03; and after toe-off, r = 0.53, p = 0.01). Non-fallers also had significant relationships between PTWS and POWS (r = 0.47, p = 0.02), PTWS and stride length (r = 0.52, p = 0.02), PTWS and cadence (r = 0.43, p = 0.04), PTWS and hip angle at toe-off (r = 0.52, p = 0.01). In non-fallers significant relationships between gait and muscle activation were found for PTWS and GM activation before heel contact (r = 0.48, p = 0.02), and PTWS and RF/GM before and after heel contact (r = -0.54, p = 0.009 and r = -0.45, p = 0.03, respectively). In non-fallers, Cw and RF/GM coactivation before toe-off (r = 0.53, p = 0.01) were also related.

Table 4 Correlations between age, preferred treadmill walking speed, knee, hip and ankle isokinetic torques, and biomechanical gait characteristics of elderly, female, fallers and non-fallers.

	Fallers			Non-Fallers		
	Age[(a)]	PTWS[(a)]	Cw[(b)]	Age[(a)]	PTWS[(a)]	Cw[(b)]
Age	1	-0.2	-0.1	1	-0.2	0.6**
PTWS	-0.3	1	-0.3	-0.3	1	-0.3
Cw	-0.1	-0.2	1	0.6**	-0.3	1
POWS	-0.4	0.5*	0.09	-0.4*	0.3	-0.1
Knee flexor joint torque	-0.3	-0.05	0.1	-0.1	0.1	-0.3
Knee extensor joint torque	0.2	0.02	-0.1	-0.1	0.2	-0.01
Hip flexor joint torque	-0.2	-0.1	0.1	-0.1	0.3	-0.1
Hip extensor joint torque	0.02	-0.2	-0.1	-0.4	0.3	-0.3
Ankle flexor plantar joint torque	-0.3	0.02	-0.1	-0.1	-0.1	-0.1
Ankle dorsiflexor joint torque	-0.3	0.09	-0.04	-0.3	0.3	-0.4
Stride time	0.1	-0.2	-0.1	-0.02	-0.3	0.08

Stride length	-0.07	0.7**	0.01	0.08	0.5*	-0.1
Cadence	-0.3	0.09	0.1	0.01	0.4*	-0.09
Ankle angle at heel contact	0.3	0.4	-0.3	0.01	0.1	-0.2
Hip extensor angle at toe-off	0.1	-0.1	0.07	0.1	0.5*	0.09
IO activation before heel contact	-0.1	-0.09	-0.3	-0.1	0.1	0.1
RF activation before heel contact	-0.002	0.2	-0.02	0.08	0.07	0.2
TA activation before heel contact	0.3	0.2	-0.09	0.2	-0.03	0.2
MU activation before heel contact	0.4	0.09	-0.3	-0.3	0.1	-0.06
GM activation before heel contact	0.1	-0.2	0.5*	0.02	0.4*	-0.1
BF activation before heel contact	0.3	0.02	-0.1	0.07	0.1	0.1
GL activation before heel contact	0.06	-0.1	0.03	-0.06	0.09	-0.1
IO activation after heel contact	-0.1	-0.1	-0.3	0.003	0.2	-0.06
RF activation after heel contact	-0.1	0.2	0.02	-0.01	0.02	0.01
TA activation after heel contact	0.1	0.2	-0.1	0.2	-0.1	0.06
MU activation after heel contact	-0.2	0.1	-0.3	-0.3	0.2	-0.1

GM activation after heel contact	0.2	-0.4	0.7**	0.08	0.3	-0.2
BF activation after heel contact	0.5*	-0.1	-0.1	0.02	-0.05	-0.1
GL activation after heel contact	0.07	-0.2	-0.03	-0.1	0.01	-0.2
IO activation before toe-off	-0.08	0.4	0.05	-0.06	-0.1	0.1
RF activation before toe-off	-0.1	0.2	0.07	0.1	0.1	0.2
TA activation before toe-off	0.2	0.2	0.1	0.2	0.07	0.2
MU activation before toe-off	-0.3	-0.1	-0.08	-0.1	0.07	0.01
GM activation before toe-off	-0.1	0.1	0.2	-0.01	0.2	-0.2
BF activation before toe-off	0.1	-0.3	-0.04	-0.2	-0.09	0.1
GL activation before toe-off	-0.1	-0.2	0.2	-0.2	0.05	-0.06
IO activation after toe-off	-0.1	-0.1	-0.05	0.07	0.03	0.1
RF activation after toe-off	0.02	-0.05	0.3	0.2	0.1	0.2
TA activation after toe-off	0.3	0.1	0.1	0.3	0.1	0.2
MU activation after toe-off	-0.2	-0.1	0.01	-0.1	0.1	-0.1
GM activation after toe-off	-0.1	-0.04	0.2	0.00	0.2	-0.1

BF activation after toe-off	0.07	-0.5*	0.1	-0.1	-0.04	0.2
GL activation after toe-off	0.09	-0.2	0.02	-0.03	0.1	0.1
IO/MU before heel contact	0.5*	-0.05	0.1	0.4*	-0.1	0.1
RF/GM before heel contact	-0.3	0.2	-0.2	-0.05	-0.5**	0.2
RF/BF before heel contact	-0.1	0.1	0.05	-0.08	0.009	0.01
TA/GL before heel contact	0.04	0.1	0.1	0.03	-0.1	0.1
IO/MU after heel contact	0.4	-0.2	-0.3	0.6**	-0.06	0.1
RF/GM after heel contact	-0.1	0.3	0.01	-0.09	-0.4	0.07
RF/BF after heel contact	-0.4	0.4	-0.03	-0.1	0.1	-0.1
TA/GL after heel contact	0.3	0.4	0.06	0.3	-0.2	0.2
IO/MU before toe-off	0.3	0.1	0.09	0.4*	-0.02	0.2
RF/GM before toe-off	0.5*	-0.05	0.02	0.3	-0.05	0.5*
RF/BF before toe-off	-0.2	0.5*	0.1	0.02	0.1	0.004
TA/GL before toe-off	0.5*	0.4	0.02	0.3	-0.003	0.3
IO/MU after toe-off	0.3	0.3	0.1	0.5	-0.05	0.3

RF/GM after toe-off	0.2	0.06	-0.01	0.01	-0.3	0.2
RF/BF after toe-off	0.07	0.5	-0.04	0.01	0.1	-0.04
TA/GL after toe-off	0.3	0.3	-0.1	0.1	-0.09	0.06

*p<0.05; **p<0.01, [(a)] Pearson coefficient, and [(b)] Spearman Coefficient. PTWS: preferred treadmill walking speed; POWS: preferred overground walking speed; Cw: cost of walking; IO: internal oblique; RF: rectus femoris; TA: tibialis anterior; MU: multifidus; GM: gluteus maximus; BF: biceps femoris; GL: gastrocnemius lateralis.

4. Discussion

The most novel aspect of this study is that it demonstrates that in older, female non-fallers age and coactivation of the thigh muscles were the only factors that influenced the Cw, while in older, female fallers a greater reliance on the hip extensors was associated with a higher Cw. This may have occurred as compensation for low knee extensor strength in female fallers that resulted in a high hip angle at toe-off and high levels of hip extensor activation at initial and final stance. These gait changes may predispose older fallers to impaired walking and increased energy cost of movement, possibly increasing the risk of recurrent falls. With respect to these findings, our results partially agree with our first hypothesis that higher muscle activation is associated with higher Cw in older women. However, our findings did not agree with our second hypothesis, that poor lower-extremity strength would be associated with a higher Cw.

4.1. Contributions of gait biomechanics to speed

Across the range of walking speeds displayed by non-fallers, increased treadmill walking speed was associated with increased cadence, stride length, and hip angle during toe-off as expected. The positive association between PTWS and GM activation at heel strike in the non-faller group is an anticipated neuromuscular response as high speeds require a greater deceleration of the center of mass during the initial stance phase (Liu et al., 2006; Anderson et al., 2003). The relationship between spatial, temporal, and kinematic gait parameters and PTWS was different in the older faller group, which only had a significant, positive association between PTWS and stride length. It is possible that older female fallers had a reduced capacity to increase cadence to obtain faster walking speeds and thus relied solely on greater stride lengths.

4.2. Biomechanical gait alterations and cost of walking

Older non-fallers demonstrated higher energy expenditure during walking with an increase in age. This finding is consistent with the literature, which shows that both walking economy and maximal oxygen consumption (VO_{2max}) decline with age. The increased Cw is probably due to both cardiovascular and neuromuscular changes, which respectively reduce muscle blood flow and maximal cardiac output and cause biomechanical gait abnormalities (Betik and Hepple, 2008; Holt et al., 1995; Malatesta et al., 2003; Mian et al., 2006, 2007; Martin et al., 1992).

Because VO_{2max} declines at a rate of approximately 8% per decade (Wilson and Tanaka, 2000), older adults perform their daily activities at higher relative intensities (% VO_{2max}) than young people. As a result of a reduced maximal aerobic capacity, small increases in the Cw may profoundly increase relative exercise intensity in older adults, and subsequently task difficulty. For example, maximal aerobic capacity for women near 70 yr is approximately 20 $ml.kg^{-1}.min^{-1}$ (Stathokostas et al., 2004). The standard deviation for the Cw in our older faller sample was ± 4 $ml.kg^{-1}.min^{-1}$. Therefore a shift of 1 SD (e.g. from 8 to 12 ml/kg/min) in the Cw would represent an increase in relative intensity from 40% VO_{2max} to 60% VO_{2max}. Thus the decline of aerobic capacity with aging, and the concurrent increase in Cw, likely contribute to a reduced ability to maintain activities of daily living, and potentially increase fatigue and fall risk while performing them. It is therefore important to determine the underlying kinematic gait and muscle activation patterns that contribute to the increased Cw in older women.

In older fallers increased Cw was associated with higher activation of GM muscles and a higher coactivation of shank muscles (TA/GL). In addition, coactivation of RF/GM in older non-fallers was positively correlated with an increased Cw. Coactivation of these lower-extremity muscles may contribute to increased stability at the ankle, knee or hip, but may also elicit a greater metabolic cost. These results are in accordance with Peterson and Martin’s (2010) findings, which showed that older adults had 40% higher activation in thigh muscles than young adults and this elevated activation was positively associated with Cw. Similarly, Hortobágyi et al. (2011) showed that coactivation of lower-limb muscles explained 43% of the variance in Cw in older adults, yet only explained 18% of the variance in Cw in young.

During walking, the quadriceps muscles, principally RF, exhibit similar phasic activity as the hip stabilizer muscles; gluteus medius, BF and GM. During initial stance, these muscles display a burst of activation as they stabilize the pelvis and the thigh to reduce the displacement of the center of mass in the sagittal plane (Liu et al., 2006; Anderson et al., 2003; Vaughan et al., 1998). Also, during terminal stance, when the anterior acceleration of the center of mass occurs, these muscles exhibit a second burst of activation, again to stabilize the lumbopelvic and thigh regions (Liu et al., 2006; Anderson et al., 2003; Vaughan et al., 1998). We theorize that older women with a history of falls increased the antagonist cocontraction of shank muscles and recruited BF and GM during stance to a greater extent than older non-fallers in an effort to maintain stabilization of the ankle and hip, as a consequence of lower knee extensor strength.

Aging causes a reduction in the number and cross-sectional area of muscle fibers, and a denervation of type II muscle fibers, which reduce the capacity to generate torque (Kinkerdall et al., 1996). In our study, knee extensor maximal voluntary torque was the only joint action that was significantly weaker in older fallers. According to Ikezoe et al. (Ikezoe et al., 2011), antigravity muscles, such as quadriceps, are the most affected by inactivity. Considering that falls usually induce a fear of falling and reduce the level of functionality, older women who have experienced a fall may be likely to develop quadriceps atrophy and decreased maximal knee extensor torque. In addition, knee extensor weakness predisposes older adults to falling (Moreland et al., 2004), but may also impact gait and Cw as has been proposed in this study.

Older female fallers also had lower activation of IO before heel contact, which also could have contributed to a higher demand for hip stabilizer muscle activation at initial stance. According to Anders et al. (Anders et al., 2007), the IO acts to stabilize the lumbopelvic region and, generally, the recruitment of this muscle occurs prior to heel contact in the attempt to provide stability. Thus, impaired lumbopelvic stability, in conjunction with lower knee extensor strength, could contribute to the greater reliance on the hip and ankle musculature in fallers. This idea is supported by the higher antagonist coactivation of shank muscles, higher hip angle at toe-off, and the higher activation of hip stabilizer muscles during stance seen in fallers in this study. The higher levels of active motor units required to stabilize the ankle and hip might therefore elicit a higher metabolic cost (Mian et al., 2006). Hence, the greater reliance on hip musculature by older fallers, and greater antagonist cocontraction at the shank, may be a suboptimal motor control strategy that increased Cw in this sample.

4.3. Limitations

The fact that our study was conducted on the treadmill limits the comparison of Cw and gait biomechanics between fallers and non-fallers, because these variables are directly related to gait velocity, which was not equal to POWS. The use of treadmill walking was necessitated by the technical challenge of concurrently conducting indirect calorimetry, EMG, and optical motion analysis measurements, which should be viewed as a strength of the study. Considering that both groups chose slower walking speeds on the treadmill than overground, our older volunteers probably did not walk at their most economical speed. To minimize the potential impact of treadmill walking on our analysis, we used ten minutes of familiarization, which according to Matsas et al. (2000) and Taylor et al. (1996) is sufficient to assure the reliability of lumbar, hip and knee kinematics during treadmill walking.

The statistical analysis used to compare faller and non-faller groups (student’s t-test for independent samples) could have increased the occurrence of type I error despite the fact that the majority of statistical differences between groups had p-values lower than 0.01. Therefore, future studies should be conducted to corroborate these initial findings.

5. Conclusion

This study provides new knowledge of the association between muscle activation, biomechanical gait alterations and metabolic cost of walking in older, female fallers and non-fallers. According to our findings, in older, female non-fallers, increased age and increased coactivation of the thigh muscles were the only factors related to a higher Cw, while in fallers, greater coactivation of shank muscles and hip muscle activation were related to an increased Cw. We speculate that in the fallers observed in this study that poor knee extensor strength, and lower activation of the lumbopelvic stabilizer muscles (IO), resulted in compensatory adaptations at the ankle and hip including high levels of antagonist coactivation in the shank muscles and greater GM activation at initial stance. Thus, this study has demonstrated that there may be a link between lower-extremity muscle weakness, muscle activation patterns, gait alterations, and increased Cw in older fallers. The elevated Cw could consequently contribute to the onset of fatigue and increased risk of falling in older people.

Conflict of interest statements

None of the authors had any conflict of interest which could bias the results of this study.

Acknowledgement

This project was funded by grants from Fundação de Amparo à Pesquisa do Estado de São Paulo (FAPESP), Conselho Nacional de Desenvolvimento Científico e Tecnológico (CNPq), Fundação para o Desenvolvimento da UNESP (FUNDUNESP) and Coordenação de Aperfeiçoamento de Pessoal de Nível Superior (CAPES).

References

Bento, P.C.B., Pereira, G., Ugrinowitsch, C., Rodacki, A.L.F. 2010. Peak torque and rate of torque development in elderly with and without fall history. Clin Biomech. 25, 450-454.

Berryman, N., Gayda, M., Nigam, A., Juneau, M., Bherer, L., Bosquet, L. 2012. Comparison of metabolic energy cost of overground and treadmill walking in older adults. Eur J Appl Physiol. 112, 1613-1620.

Betik, A.C., Hepple, R.T. 2008. Determinants of VO2max decline with aging: an integrated perspective. App Physiol Nutr Metab. 33, 130-140.

Brujin, S.M., van Dieen, J.H., Meijer, O.G., Beeke, P. 2009. Is slow walk more stable? J Biomech. 42, 1506-1512.

Burnfield, J.M., Josephson, K.R., Powers, C.M., Rubenstein, L.Z. 2000. The influence of lower extremity joint torque on gait characteristics in elderly men. Arch Phys Med Rehabil. 81, 1153-1157.

Dal, U., Erdogan, T., Resitoglu, B., Beydagi, H. 2010. Determination of preferred walking speed on treadmill lead to high cost on treadmill walking. Gait Post. 31, 366-369.

DeVita, P., Hortobágyi, T. 2000. Age causes a redistribution of joint torques and powers during gait. J Appl Physiol. 88, 1804-1811.

Hartmann, A., Knolls, R., Murer, K., Bruin, E.D. 2009. Reproducibility of an isokinetic strength-testing protocol of the knee and ankle in older adults. J Gerontol A Biol Sci Med Sci. 55, 259-268.

Hermens, J.H., Freriks, B., Disselhorst-Klug, C., Rau, G. 2000. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyog Kinesiol. 10, 361-374.

Holt, K.J., Jeng, S.F., Rr, R.R., Hamill, J. 1995. Energetic cost and stability during human walking at the preferred stride velocity. J Mot Behav. 27, 164-178.

Hortobágyi, T., Finch, A., Solnik, S., Rider, P., DeVita, P. 2011. Association between muscle activation and metabolic cost of walking in young and old adults. J Gerontol A Biol Sci Med Sci. 66A, 541-547.

Ikezoe, T., Mori, N., Nakamura, M., Ichihashi, N. 2011. Atrophy of the lower limb in elderly women: is it related to walking ability? Eur J Appl. 111, 989-995.

Kang, H., Dingwell, J.B. 2008. Effects of walking speed, strength and range of motion on gait stability in healthy older adults. J Biomech. 41, 2899-2905.

Kerrigan, D.C., Lee, L.W., Nieto, T.J. 2000. Kinetic alterations independent of walking speed in elderly faller. Arch Phys Med Rehabil. 81, 730-735.

Kirkendall, D.T., Garret, W.E. (1998). The effects of aging and training on skeletal muscle. Am J Sport Med. 26, 698-602.

LaStayo, P.C., Ewy, G.A., Pierotti, D.D., Johns, R.K., Lindstet, S. 2003. The positive effects of negative work: increased muscle strength and decreased fall risk in a frail elderly population. J Gerontol A Biol Sci Med Sci. 58, M419-424.

Lee, S.J., Jidler, J. 2008. Biomechanics of overground vs. treadmill walking in healthy subjects. J Appl Physiol. 104, 747-755.

LaRoche DP, Cremin KA, Greenleaf B, Croce RV. 2010. Rapid torque development in older female fallers and nonfallers: a comparison across lower-extremity muscles. J Electromyogr Kinesiol. 20, 482-488.

Liu, M.Q., Anderson, F.C., Pandy, M.C., Delp, S.L. 2006. Muscles that support the body also modulate forward progression during walking. J Biomech. 39, 2623-2630.

Malatesta, D., Simar, D., Dauvilliers, Y., Candau, R., Borrani, F., Prefaut, C, Caillaud, C. 2003. Energy cost of walking and gait instability in healthy 65- and 80-yr-olds. J Appl Physiol. 95, 2248-2256.

Martin, P.E., Rothstein, D.E., Larish, D.D. 1992. Effects of age and physical activity status on the speed-aerobic demand relationship of walking. J Appl Physiol. 73, 200-206.

Marshall, P., Murphy, B. 2003. The validity and reliability of surface EMG to assess the neuromuscular response of the abdominal muscles to rapid limb movement. J Electromyogr Kinesiol. 13, 477-489.

Mian, O.S., Thom, J.M., Ardigo, L.P., Narici, M.V., Minetti, A.E. 2006. Metabolic cost, mechanical work, and efficiency during walking in young and older men. Acta Physiol. 186, 127-139.

Mian, O.S., Thom, J.M., Ardigo, L.P., Morse, C.I., Narici, M.V., Minetti, A.E. 2007. Effect of a 12-month physical conditioning programme on the metabolic cost of walking in healthy older adults. Eur J Appl Physiol. 100, 499-505.

Moreland, J.D., Richardson, J.A., Goldsmith, C.H., Clase, C.M. 2004. Muscle weakness and falls in older adults: a systematic review and meta-analysis. J Am Ger Soc. 52, 1121-1129.

Ortega, J.D., Farley, C.T. 2007. Individual limb work does not explain the greater metabolic cost of walking in elderly adults. J Appl Physiol. 102, 2266-2273.

Parvataneni, K., Ploeg, L., Olney, S.J., Browmer, B. 2009. Kinematic, kinetic and metabolic parameters of treadmill versus walking in healthy older adults. Clin Biomech. 24, 95-100.

Pavol, M.J., Owings, T.M., Foley, K.T., Grabiner, M.D. 2002. Influence of lower extremity strength of healthy older adults on the outcome of an induced trip. J Am Geriatr Soc. 50, 256-262.

Pijnappels, M., Van Der Burg, P.J., Reeves, N.D., Van Dieen, J.H. 2008. Identification of elderly fallers by muscle strength measures. Eur J App Physiol. 10, 585-92.

Peterson, D.S., Martin, P.E. 2011. Effects of age and walking speed on coactivation and cost of walking in healthy adults. Gait Post. 31, 355-359.

Riley, P.O., Paolinin, G., Croce, U.D., Paylo, K.W., Kerrigan, D.C. 2007. A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. Gait Post. 27, 17-24.

Robinovitch, S.N., Heller, B., Lui, A., Cortez, J. 2002. Effect of strength and speed of torque development on balance recovery with the ankle strategy. J Neurophysiol. 88, 613-620.

Saha, D., Gard, S., Fatone, S., Ondra, S. 2007. The effect of trunk-flexed postures on balance and metabolic energy expenditure during standing. Spine. 32, 1605-1611.

Schmitz, A., Silder, A., Heiderscheit, B., Mahoney, J., Thelen, D.G. 2008. Differences in lower-extremity muscular activation during walking between healthy older and young adults. J Electromyogr Kinesiol. 19, 1085-1091.

Shiavi, R., Frigo, C., Pedotti, A. 1998. Electromyographic signals during walking: criteria for envelop filtering and number of strides. Med Biol Eng Comput. 35, 171-178.

Skelton, D.A., Kennedy, J., Rutherford, O.M. 2002. Explosive power and asymmetry in leg muscle function in frequent fallers and non-faller aged over 65. Age Aging. 31, 119-125.

Stathokostas, L., Jacob-Johnson, S., Petrella, R.J., Paterson, D.H. 2004. Longitudinal changes in aerobic power in older men and women. J Appl Physiol. 97, 781-789.

Verhoeff, L.L., Horlings, C.G.C., Janssen, L.J.F. 2009. Effects of biofeedback on trunk sway during dual tasking in the healthy young and elderly. Gait Post. 30, 76-81.

Watt, J.R., Franz, J.R., Jackson, K., Dicharry, J., Riley, P.O., Kerrigan, D.C. 2010. A tree-dimensional kinematic and kinetic comparison of overground and treadmill walking in healthy elderly subjects. Clin Biomech. 25, 444-449.

Wert, D.M., Brach, J., Perera, S., VanSwearingen, J.M. 2010. Gait biomechanics, spatial and temporal characteristics, and the energy cost of walking in older adults with impaired mobility. Phys Ther. 90, 977-985.

Whittington, B., Silder, A., Thelen, D.G. 2008. The contribution of passive-elastic mechanisms to lower extremity joint kinetics during human walking. Gait Post. 27, 628-634.

Wilson, T.M., Tanaka, H. 2000. Meta-analysis of the age-associated decline in maximal aerobic capacity in men: relation to training status. Am J Physiol Heart Circ Physiol. 278, H829-34.

3.2. Artigo II – Artigo publicado na *Isokinetic and Exercise Science*

Lower limb strength is associated with gait biomechanical abnormalities in older female fallers and non-fallers

Nise Ribeiro Marques[a,*], Camilla Zamfolini Hallal[a], Luciano Fernandes Crozara[a], Mary Hellen Morcelli[a], Aline Harumi Karuka[b], Marcelo Tavella Navega[c], Mauro Gonçalves[a]

[a] Department of Physical Education, São Paulo State University, Rio Claro, Brazil.

[b] Department of Physical Therapy, São Paulo State University, Presidente Prudente, Brazil

[c] Department of Physical Therapy and Occupational Therapy, São Paulo State University, Marília, Brazil.

Abstract

Background: Age-related loss in lower limb strength is related with impaired mobility. However, the association between decreased lower limb strength and gait biomechanical abnormalities is unclear. In line with this, our study aimed to compare the maximum isokinetic voluntary strength (MIVS) of hip, knee and ankle of older women with and without history of falls. Also, we correlate the strength of each group with gait biomechanics.

Methods: The MIVS were assessed during concentric/concentric movements performed for hip, knee and ankle joints. Gait biomechanics (kinematic and electromyography) were assessed during 1-minute recorded during the volunteers walking on the treadmill at self-selected speed. Electromyographic signal was analyzed by the linear envelop after heel strike and before toe-off. The kinematic data were analyzed using the variables: step time, length and step width and ankle angle at heel strike, and hip angle at toe-off.

Results: In faller group, we found that a decreased hip abduction and adduction MIVS is associated with a higher tibialis anterior activation at initial stance ($p = 0.04$ and $r = -0.53$ and $p = 0.04$ and $r = -0.52$).

Conclusion: Therefore, an impaired strength of hip could causes compensation in ankle stabilizer muscles activation at initial stance in older female fallers.

Key words: Aging. Fall risk · Torque · EMG · Kinematic · Walking

1. Introduction

Muscle strength in the lower limbs decreases with age and is frequently considered a risk factor for falls in older adults [1,2,3]. Falls are a main cause of death and injury in older adults; fear of subsequent falls may lead to decreased activity, which may aggravate other morbidities in the orthopedic, cardiovascular and respiratory systems [4,5]. Considering that approximately 50% of falls occur while walking, adequate biomechanical performance during this movement is essential to prevent falls [6,7,8].

There are several causes for age-related muscle strength impairments. Among these are muscle atrophy, decreased capacity to generate torque, decreased velocity of muscle shortening and decreased velocity in nerve impulse conduction [3]. Recent studies have shown that older fallers and non-fallers differ in the relative amount of lower limb strength and in the rate of torque development [2,3,9]. However, there is no consensus as to which muscle group makes the greatest contribution to fall risk. According to LaRoche et al. [3], fallers and non-fallers differ in ankle plantar flexors and dorsiflexors as well as in the rate of torque development; these factors are the main contributors to fall risk. Similarly, Bento et al. [2] reported that the higher rate of torque development in the isometric knee flexors indicate that older female non-fallers should have a greater ability to recover balance after tripping.

Though decreased muscle strength is identified as a factor for impaired mobility, a small number of studies have correlated hip, knee and ankle strength with gait biomechanics [1,10]. According to Burnfield et al. [10], who investigated associations between hip, knee and ankle isokinetic strength and gait kinematic parameters, only maximum voluntary torque of the hip extensor was correlated with walking speed, stride length and cadence in older male non-fallers. However, Suzuki et al. [1] showed that the maximum isometric torque and power of ankle plantar flexors and dorsiflexors were associated with walking speed in older adults with normal mobility. There are no studies known to the authors that have analyzed the contribution of maximal voluntary strength of the hip, knee and ankle in lower-limb angular displacement and in electromyographic activation during walking in older female fallers and non-fallers.

Training exercises to increase muscle strength are widely used for older adults to prevent falls. Determining the contribution of lower limb strength in gait biomechanics (via electromyography and kinematic analysis) is an important clinical issue. In light of this, our study aimed to compare the maximal isokinetic voluntary strength (MIVS) of the hip, knee and ankle in older women with and without a history of falls and to correlate the lower limb strength of each group with gait biomechanical parameters (kinematic and electromyography). Our main hypothesis is that impaired lower limb strength is strongly related to gait abnormalities in older female fallers.

2. Methods

2.1. Study Design and Participant

Data collected from thirty-seven older adult women who completed a baseline assessment of a clinical trial were considered for this study. The participants were divided into two groups (fallers, n = 15; and non-fallers, n = 22) based on whether they had fallen in the one-year period preceding the evaluation. Older female fallers and non-fallers had similar personal and anthropometric characteristics (age, mass, height, body mass index, BERG balance scale and Mini-mental examination scores; Table 1). We defined a fall as any balance disturbance that caused significant contact with the floor. All participants signed a consent form, which was approved by an Institutional Ethics Committee. People who had a history of untreated cardiovascular disease, dementia or cognitive impairment (defined as a Mini-Mental State Examination score < 20), balance disturbance (defined as a BERG balance score < 36), hemiparesis, pain of the lower limbs or trunk, or a progressive motor disorder were excluded (one subject with heart failure).

Table 1 Subjects characteristic.

Variable	**Elderly Faller (n=15)**	**Elderly Non-Faller (n=22)**	**P**
Age (years)	69.6 (8.09)	66.1 (6.2)	0.1
Weight (kg)	66.8 (9.2)	65.3 (13.6)	0.6
Height (mm)	1.51 (0.06)	1.54 (0.05)	0.2
Body mass index ($kg \cdot m^{-2}$)	28.9 (3.2)	27.4 (4.9)	0.2
BERG balance scale (score)	50.7 (6.8)	44.8 (21.6)	0.2
Mini-Mental Examination Score	22.9 (4.8)	24.1 (6.05)	0.5
Number of falls	1.8 (1.3)	-	-

2.2. Procedure

The data were collected from participants during two sessions separated by 24-72 hours. During the first data collection visit, preferred overground walking speed (POWS) and preferred treadmill walking speed (PTWS) as well as the maximal voluntary joint torques of the hip, knee and ankle were measured. During the second day of data collection, participants used a treadmill for 10 minutes to familiarize themselves with gait at their PTWS; then, one minute of gait biomechanics (kinematic and electromyographic variables) at PTWS was recorded.

Gait kinematic parameters, EMG activity and joint torques were assessed using a motion analysis system with seven infrared cameras (Vicon Motus, Oxford, UK®); an 8-channel telemetry electromyography system (Noraxon®, Phoenix, USA); and an isokinetic dynamometer (Biodex®, New York, USA).

To measure gait kinematic parameters, 39 markers fixed on the head, trunk, and upper and lower limbs were used according to the manufacturer's specifications. The kinematic parameters were recorded at a sampling frequency of 100 Hz. Heel contact and toe-off were identified using two footswitches (Noraxon®, Phoenix, USA), which were composed of 4 sensors attached to each foot. One sensor was attached to the heel, first and fifth metatarsus and toe, and they were synchronized with the EMG and motion analysis system.

EMG signals were collected at a sampling frequency of 2000 Hz using Ag/AgCl (Miotec®, Porto Alegre, Brazil) disc electrodes with an active area of 1 cm^2 and an inter-electrode distance of 2 cm in a bipolar configuration. The electrodes were positioned on the following muscles of the right side of the participants' body: the internal oblique (IO), 2 cm medial and inferior to the anterior superior iliac spine; the multifidus (MU), 2 cm lateral to the space between the spinous processes of L4-L5; the gluteus maximus (GM), on the midpoint of the line between the sacral vertebrae and the greater trochanter; the biceps femoris (BF), on the midpoint of the line between the ischial tuberosity and the lateral epicondyle of the tibia; the rectus femoris (RF), on the midpoint of the line from the anterior spine iliac superior to the superior part of the patella; the tibialis anterior (TA), at a distance of 1/3 of the line between the proximal head the fibula and the tip of the medial malleolus; and the gastrocnemius lateralis (GL), at a distance of 1/3 of the line between the proximal head of the fibula and the heel [11,12]. A reference electrode was placed on right malleolus medialis. Subjects' skin was shaved and cleaned with alcohol before placing the electrodes [11].

2.3. Measures

2.3.1. Measurement of hip, knee and ankle strength

Maximal joint torques were measured isokinetically at 120°/sec for the hip, knee and ankle for both flexion and extension. Joint testing order was randomized and the procedure and desired movement pattern were explained to participants before testing each joint. Five submaximal repetitions were performed so that participants were familiarized with the movements, and five maximal repetitions were collected for analysis. Strong verbal encouragement was provided to obtain maximal effort during data acquisition of each joint, and before the strength assessment, the volunteers performed a five-minute warm-up that involved walking on the treadmill at PTWS [10].

Hip extension and hip flexion torque were measured with each subject in a supine position. A resistance pad was secured with a strap around the thigh of the limb being tested, and each subject's pelvis and trunk were stabilized on the table using belts. The dynamometer was aligned to the approximate axis of rotation of the hip joint being tested (superior and anterior to the greater trochanter). The test was initiated with the hip flexed at 10º; hip flexion torque was assessed as the subject moved from this position to 60º of hip flexion. Hip extension torque was assessed during movement from the flexed 60º position back to 10º of hip flexion[13].

To assess hip adduction and abduction joint torque, participants were asked to lie down in a lateral posture with the dominant lower limb positioned on top of the non-dominant lower limb. The limb being tested was strapped to the Biodex input arm with the thigh pad just proximal to the knee (10 cm above superior border patella). The dynamometer was aligned to the approximate axis of rotation of the hip joint being tested (superior and anterior to the greater trochanter). The test was initiated with the hip in a neutral position (anterior iliac spine aligned to the patella superior border) and hip abduction torque was assessed as the subject moved from this position to 30º of hip abduction. Hip adduction torque was assessed during movement from the 30º abducted position toward the neutral hip position.

Isokinetic knee flexion and extension torques were measured with participants seated with the hip flexed at 90º. The dynamometer was aligned to the approximate axis of rotation of the knee joint (a line traversing the femoral epicondyles) and the resistance pad was placed on the tibia, just proximal to the superior border of the medial malleolus. The subjects' thigh, trunk, and pelvis were stabilized with straps, and subjects crossed their arms in front of the chest throughout the test [10]. Testing was initiated at 90º of knee flexion with the initial movement toward 60º of extension [14].

Isokinetic plantar flexion and dorsiflexion torques were measured with each subject lying in a supine position. The knees and hips were flexed while the ankle was in neutral inversion-eversion. The dynamometer was aligned to approximate the axis of rotation of the ankle joint being tested (the projection of a line passing obliquely through the distal tip of the tibia and fibula), and the foot was strapped securely to the foot plate. Proximal thigh and trunk stabilization was provided using belts to prevent unwanted movement. The test was initiated with subjects at 10º of dorsiflexion, with the initial movement toward 30º of plantar flexion [14]. Figure 1a, 1b and 1c show the subject position during hip, knee and ankle joint torques measurement.

Fig 1. Isokinetic positioning to assess the maximal joint torque during (a) hip extension and flexion, (b) hip adduction and abduction, (c) ankle dorsiflexion and plantar flexion and (d) knee flexion and extension.

2.3.2. Gait analysis: gait speed, step length, step width, cadence, ankle and hip angle and EMG activity

On the first day of data collection, PTWS was determined by starting at 50% of the POWS. To determine POWS, participants were instructed to walk at a self-selected natural pace for 20 m. The duration of each trial was determined with two infrared timing gates placed at 5 m and 15 m distances. POWS was calculated by dividing the distance walked (10 m) by the amount of time required to walk this distance. Velocity was obtained in units of $m \cdot s^{-1}$.

Participants were asked to walk on the treadmill; its speed was increased until the subject reported that it was faster than her preferred gait speed. Treadmill speed was then slowly decreased until the subject reported that the current speed was slower than preferred. This procedure was repeated three times and the average of the 3 ‘‘faster’’ and 3 ‘‘slower’’ than preferred speeds was taken as the subject’s PTWS.

On the second day of data collection, participants were familiarized with the PTWS for 10 minutes. Then, 1 minute of EMG and kinematic data were recorded while the participants walked at the PTWS (Figure 2).

2.4. Data Analysis

Maximal voluntary joint torque of the hip, knee and ankle were processed with a fourth order Butterworth filter set to a 15-Hz cut-off frequency. Maximal joint torque values were obtained and normalized using each participants’ mass.

EMG signals were processed using specific routines developed for the MATLAB environment (Mathworks®, Natick, USA). Full-wave rectification, a fourth order low pass filter and a 10 Hz cut-off frequency were the parameters used. A 100 ms linear envelope of EMG signal was recorded after heel contact (initial stance) and before toe-off (final stance) of the first ten strides. Linear envelope values were normalized with the mean of each muscle activation recorded. All the linear envelope values were normalized to the mean activation obtained during the gait.

Fig 2. Procedures of data collection for gait biomechanics.

Gait speed, stride length, stride time, angle of ankle dorsiflexion during heel contact and angle of hip extension during toe-off were analyzed during the ten first strides. A fourth order Butterworth filter with a 6-Hz cut-off frequency was used to smooth marker trajectory. We chose these kinematic variables because gait speed, stride length and time should give more information about the mechanical efficiency of gait. These parameters represent adjustments that must be made to the temporal and spatial components of gait as well as to ankle dorsiflexion at heel contact and the angle of hip extension at toe-off.

PASW 18.0 (SPSS Inc., Chicago, USA) was used for all statistical analyses. We used the appropriate descriptive statistics (mean and standard deviation) to summarize participant characteristics. The Shapiro-Wilk test was used to test the normality of the data and MANOVA was used to compare the dependent variables between groups. We computed the Pearson and Spearman correlation coefficients, as appropriate, to quantify the association between the maximal voluntary isokinetic torque and age as well as each of the gait biomechanical variables. The significance level was set to $p < 0.05$.

3. Results

Our MANOVA analysis yielded significant differences ($F = 4.47$ and $p = 0.01$) between the faller and non-faller groups. Maximal knee extensor torque was 28% greater in non-fallers ($p = 0.001$). However, the faller group had 4º more hip extension at toe-off ($p = 0.02$), 33% more activation of the BF at the initial stance ($p = 0.02$) and 39% greater activation of the GM at the final stance when compared to the non-faller group ($p = 0.005$).

Table 2 Values of mean and standard deviation of BERG balance scale, preferred overground and treadmill walking speed, isokinetic torque of hip and ankle joints and biomechanical (kinematic and EMG) variables of gait.

Variable	**Older faller (n=16)**	**Older Non-Faller (n=23)**	**P**
POWS ($m \cdot s^{-1}$)	1.1 (0.1)	1.3 (0.2)	0.06
PTWS ($m \cdot s^{-1}$)	0.9 (0.1)	0.9 (0.1)	0.6
Knee flexor torque ($N \cdot m \cdot kg^{-1}$)	0.54 (0.26)	0.55 (0.14)	0.8
Knee extensor torque ($N \cdot m \cdot kg^{-1}$)	0.69 (0.2)	0.96 (0.24)	0.02*
Hip flexor torque ($N \cdot m \cdot kg^{-1}$)	0.68 (0.26)	0.74 (0.19)	0.4
Hip extensor torque ($N \cdot m \cdot kg^{-1}$)	0.78 (0.23)	0.89 (0.36)	0.2

Ankle flexorplantar torque ($N^{.}m^{.}kg^{-1}$)	0.31 (0.19)	0.33 (0.13)	0.7
Ankle dorsiflexor torque ($N^{.}m^{.}kg^{-1}$)	0.28 (0.11)	0.31 (0.13)	0.3
Step time (second)	2.3 (0.89)	2.6 (0.9)	0.3
Step length (mm)	509.3 (61.9)	497.2 (70.7)	0.6
Step width (mm)	140.4 (31.1)	170.9 (28.8)	0.06
Ankle angle at heel contact (degrees)	6.4 (4.3)	5.9 (4.4)	0.2
Hip angle at toe-off (degrees)	5.4 (4.7)	9.5 (4.8)	0.01*
IO activation at initial stance (% mean)	97.2 (38.1)	100.3 (29.4)	0.7
RF activation at initial stance (% mean)	143.8 (46.03)	130.6 (33.5)	0.3
TA activation at initial stance (% mean)	106.7 (21.4)	122.8 (31.6)	0.1
MU activation at initial stance (% mean)	150.5 (38.8)	147.7 (40.8)	0.8
GM activation at initial stance (% mean)	154.7 (44.3)	179.9 (52.2)	0.1
BF activation at initial stance (% mean)	36.4 (22.8)	24.1 (8.3)	0.02*
IO activation at final stance (% mean)	117.5 (38.7)	105.09 (29.5)	0.2
RF activation at final stance (% mean)	89.5 (41.2)	80.09 (28.4)	0.4
MU activation at final stance (% mean)	76.1 (27.3)	82.8 (37.3)	0.5
GM activation at final stance (% mean)	86.4 (39.6)	52.3 (28.4)	0.005**
BF activation at final stance (% mean)	43.8 (47.1)	50.1 (41.03)	0.6
GL activation at final stance (% mean)	91.7 (28.3)	75.8 (24.8)	0.08

*$p<0.05$; **$p<0.01$. PTWS: preferred treadmill walking speed; POWS: preferred overground walking speed; IO: internal oblique; RF: rectus femoris; TA: tibialis anterior; MU: multifidus; GM: gluteus maximus; BF: biceps femoris; GL: gastrocnemius lateralis.

For older female non-fallers, we found a significant association between hip abduction maximal torque and preferred treadmill walking speed ($p = 0.02$ and $r = 0.49$). Hip extensor maximal torque and hip extensor angle at toe-off were also significantly correlated ($p = 0.03$ and $r = 0.45$), as were hip extensor maximal torque and RF activation at the initial stance ($p = 0.04$ and $r = -0.43$).

In older female fallers, we found a significant association between the following 8 pairs of gait biomechanical factors: 1) hip abduction maximal torque and TA activation at the initial stance ($p = 0.04$ and $r = -0.53$); 2) hip adduction maximal torque and TA activation at the initial stance ($p = 0.04$ and $r = -0.52$); 3) hip adduction maximal torque and GM activation at the final stance ($p = 0.02$ and $r = -0.58$); 4) knee extensor maximal torque and RF activation at the initial stance ($p = 0.02$ and $r = -0.58$); 5) knee extensor maximal torque and RF activation at the final stance ($p = 0.04$ and $r = 0.53$); 6) hip flexor maximal torque and hip extensor angle at toe-off ($p = 0.01$ and $r = 0.61$); 7) hip extensor maximal torque and hip extensor angle at toe-off ($p = 0.03$ and $r = 0.54$); and 8) hip extensor maximal torque and MU activation at the final stance ($p = 0.04$ and $r = 0.52$).

Table 3 Pearson correlation coefficients between hip maximum voluntary torque and biomechanical gait parameters of older female fallers and non-fallers.

	Hip Abduction Torque		**Hip Adduction Torque**		**Hip Flexor Torque**		**Hip Extensor Torque**	
	Fallers	Non-Fallers	Fallers	Non-Fallers	Fallers	Non-Fallers	Fallers	Non-Fallers
PTWS	0.21	0.49*	-0.27	0.04	-0.07	0.37	-0.1	0.37
Step time	0.12	-0.14	-0.03	-0.04	0.18	0.003	-0.23	0.09
Step length	0.08	0.29	-0.2	0.29	-0.07	0.1	0.04	0.1
Step width	0.01	-0.08	-0.11	-0.41	0.24	-0.24	-0.09	-0.19
Ankle angle at heel contact	-0.18	0.29	-0.35	0.29	0.26	0.16	-0.23	0.04
Hip angle at toe-off	-0.22	0.21	0.08	-0.27	0.61*	0.34	0.54*	0.45*
IO activation at initial stance	-0.05	0.14	-0.24	0.2	-0.05	0.1	-0.03	0.09
RF activation at initial stance	0.09	-0.1	-0.38	-0.01	0.12	-0.4	-0.36	-0.43*
TA activation at initial stance	-0.53*	-0.02	-0.52*	0.04	-0.03	-0.05	-0.33	-0.24
MU activation at initial stance	0.29	0.02	-0.17	0.007	0.03	0.02	-0.38	-0.01
GM activation at initial stance	0.08	0.27	0.37	0.02	-0.06	-0.2	0.06	-0.35

BF activation at initial stance	-0.47	0.07	0.23	0.18	-0.45	0.1	0.16	-0.01
IO activation at final stance	-0.32	-0.33	-0.07	-0.3	-0.04	-0.18	0.02	-0.37
RF activation at final stance	-0.14	-0.19	0.1	-0.1	-0.13	0.36	-0.007	0.38
MU activation at final stance	-0.37	-0.05	0.08	-0.04	0.11	0.01	0.52*	-0.08
GM activation at final stance	-0.16	-0.19	-0.58*	0.09	-0.003	0.11	-0.28	0.14
BF activation at final stance	-0.28	-0.09	0.49	-0.04	-0.49	-0.09	-0.02	0.01
GL activation at final stance	-0.34	-0.24	-0.08	-0.4	0.08	0.12	-0.03	0.09

*p<0.05. PTWS: preferred treadmill walking speed; POWS: preferred overground walking speed; IO: internal oblique; RF: rectus femoris; TA: tibialis anterior; MU: multifidus; GM: gluteus maximus; BF: biceps femoris; GL: gastrocnemius lateralis.

Table 4 Pearson correlation coefficients between knee and ankle maximum voluntary torque and biomechanical gait parameters of older female fallers and non-fallers.

	Knee Flexor Torque		**Knee Extensor Torque**		**Ankle Flexor Plantar Torque**		**Ankle Dorsiflexor Torque**	
	Fallers	Non-Fallers	Fallers	Non-Fallers	Fallers	Non-Fallers	Fallers	Non-Fallers
PTWS	-0.06	0.17	0.12	0.2	0.05	-0.19	0.21	0.35
Step time	0.13	0.22	-0.29	-0.16	-0.29	0.04	-0.46	-0.33
Step length	0.27	0.39	-0.21	-0.06	0.32	-0.29	-0.03	0.15
Step width	0.48	-0.32	-0.11	0.12	-0.04	0.15	-0.19	-0.1
Ankle angle at heel contact	-0.42	-0.03	0.02	0.1	0.2	0.18	0.06	0.33
Hip angle at toe-off	-0.02	-0.04	0.28	0.34	-0.2	0.09	0.12	0.34
IO activation at initial stance	-0.007	-0.03	-0.21	0.09	-0.03	0.25	-0.15	0.02
RF activation at initial stance	0.18	-0.03	-0.58*	-0.4	0.49	-0.19	-0.26	-0.16
TA activation at initial stance	0.19	-0.2	-0.12	-0.05	0.14	0.13	-0.11	0.21
MU activation at initial stance	0.32	0.11	-0.08	0.3	0.33	0.06	0.32	0.03

GM activation at initial stance	-0.15	-0.3	-0.17	0.11	0.29	-0.13	-0.27	0.25
BF activation at initial stance	0.22	-0.09	0.38	0.18	-0.13	0.03	-0.17	-0.16
IO activation at final stance	-0.06	-0.19	-0.07	-0.16	0.4	-0.25	0.06	-0.16
RF activation at final stance	-0.2	0.18	-0.53*	0.28	-0.37	0.16	0.26	0.02
MU activation at final stance	-0.32	-0.15	-0.08	-0.25	-0.31	-0.04	-0.32	0.12
GM activation at final stance	0.16	0.17	-0.08	-0.13	-0.3	0.22	0.18	-0.19
BF activation at final stance	0.29	-0.02	0.38	0.03	-0.15	-0.05	-0.36	0.1
GL activation at final stance	0.08	-0.15	0.08	-0.05	-0.18	0.05	-0.21	0.16

*p<0.05; **p<0.01. PTWS: preferred treadmill walking speed; POWS: preferred overground walking speed; IO: internal oblique; RF: rectus femoris; TA: tibialis anterior; MU: multifidus; GM: gluteus maximus; BF: biceps femoris; GL: gastrocnemius lateralis.

4. Discussion

Impaired lower limb strength is considered an important risk factor for falls in older adults [1,2,3]. Therefore, our study aimed to identify associations between lower limb strength and gait biomechanics in older female fallers and non-fallers. We found an association between decreased hip strength and greater activation of the ankle stabilizer muscles in older female fallers; this may reflect a compensatory mechanism to provide stability during the initial stance. Therefore, our findings support our initial hypothesis, which states that the impairment of lower limb strength generates abnormal gait biomechanics, principally, in older female fallers.

4.1. Association between strength and walking speed

Our findings showed that for older female non-fallers, hip abduction strength is moderately associated with PTWS. This association is most likely related to greater ground reaction force during heel contact, which is equivalent to approximately twice the body weight carried in the hip joint and causes perturbations in hip stability [15,16]. Similarly, the muscles around the hip generate the highest compressive force on the acetabulum to provide stability at the initial stance [16]. Therefore, we suggest that greater hip abduction strength could provide more hip stability and better conservation of mechanic energy during this gait phase, potentially contributing to higher PTWS.

We also found that hip flexor and extensor strength are associated with hip range of motion at toe-off. This finding supports our hypothesis that greater hip strength contributes to hip stability, which conserves more mechanic energy for subsequent steps; this is reflected in the changes in kinematic parameters such as hip angle at toe-off. In line with this finding, we hypothesized that a wider range of motion of the hip could contribute to greater PTWSs; however, we did not identify an association between these variables. Hip angle at toe-off is an important sensory input to control the next step. According to Wert et al. [17], impaired hip range of motion at toe-off interrupts sensory feedback, which is provided by the stretching of muscle spindles in the anterior thigh; this interruption increases both the variability in muscle activation and the cost of walking while reducing gait speed.

4.2. Deficits in lower limb strength and muscles activation abnormalities

According to our findings, hip weakness is the major cause of muscle recruitment abnormalities during gait. Our most novel finding is that impaired hip abduction and adduction strength is associated with higher TA activation at the initial stance. We believe that this finding is related to ankle postural strategies to maintain balance in the sagittal plane. Ankle postural strategies are characterized by the activation of ankle stabilizer muscles, which adapt when there are minor perturbations in the center of mass. We also found that frontal hip weakness is associated with using this strategy to maintain balance during gait.

Another biomechanical variable that was moderately associated with hip strength is hip angle at toe-off. According to Wert et al. [17], this variable represents an important sensory input, which regulates the next step and signals the transition from the stance phase to the balance phase during gait. Therefore, older adults with a narrower range of hip motion at toe-off may experience a deficit in sensory feedback, which increases muscle activation and decreases walking speed.

Hip stabilizer muscles, such as the GM and RF, have similar activation patterns during gait. These muscles have two peaks of activation, one at the initial stance and one at the final stance [16]. During the initial stance, the GM and RF contract eccentrically to maintain hip and knee stability; these muscles also contribute to decelerate displacement of the anterior center of mass [18,19]. In the final stance, these two muscles stabilize hip and knee joints, while the anterior center of mass accelerates [18,19]. Thus, strength loss in hip adduction and knee extension could result in greater perturbations of dynamic balance in these joints, during both the initial and final stance phases. These perturbations increase the demand of RF and GM activation as a compensatory responses to maintain stability.

In the female faller group, we found an association between decreased strength in hip adduction and knee extension and higher activation of the RF at the initial stance and the GM at the final stance. Similarly, strength loss in these muscles groups had an important role in muscle activation patterns during walking. Greater activation of these muscles may increase the sensation of effort and cause fatigue, potentially increasing the risk of recurrent falls.

In the faller group, we found a moderate association between hip extension strength loss and decreased activation of the MU at the final stance. Increased hip extension strength is associated with wider hip angles at toe-off in fallers. We hypothesize that increased range of motion of the hip in this gait phase promotes greater trunk extension, which could increase MU activation at toe-off and provide spinal stability [20].

Though walking is a complex task requiring postural control, strength and aerobic capacity, our study only focused on strength to understand gait biomechanical abnormalities. Walking rehabilitation programs should consider postural control and aerobic capacity to prescribe exercises.

5. Conclusion

To our knowledge, this study is the first to be conducted to identify associations between biomechanical abnormalities and lower limb strength in older female fallers and non-fallers. We found that reduced strength of hip adduction, hip extension and knee extension leads to biomechanical gait abnormalities in the recruitment of trunk, hip and knee stabilizer muscles during the initial and final stance in older female non-fallers. For both fallers and non-fallers, knee flexor strength was positively associated with walking speed. Thus, decreased strength of the lower limbs is associated with abnormal gait patterns, which may cause poor gait performance; this performance is the major risk factor for falls in older adults.

Acknoledgement

This project was funded by grants from Fundação de Amparo à Pesquisa do Estado de São Paulo (FAPESP), Conselho Nacional de Desenvolvimento Científico e Tecnológico (CNPq), Fundação para o Desenvolvimento da UNESP (FUNDUNESP) and Coordenação de Aperfeiçoamento de Pessoal de Nível Superior (CAPES).

References

[1] Suzuki T, Bean JF, Fielding RA. Muscle power of the ankle flexors predicts functional performance in community-dwelling older women. J Am Geriatr Soc. 2001; 49: 1161-1167.

[2] Bento PCB, Pereira G, Ugrinowitsch C, Rodacki ALF. Peak torque and rate of torque development in elderly with and without fall history. Clin Biomech. 2010; 25: 450-454.

[3] LaRoche DP, Cremin KA, Greenleaf B, Croce RV. Rapid torque development in older female fallers and nonfallers: a comparison across lower-extremity muscles. J Electromyogr Kinesiol. 2010; 20: 482-488.

[4] van Dieen JH, Pijnappells M. Falls in older people. J Electromyogr Kinesiol. 2008; 18: 169-171, 2008.

[5] Schulz BW, Lloyd JD, William E. The effects of everyday concurrent task on overground minimum toe clearance and gait parameters. Gait Post. 2010; 32: 18-22.

[6] Baird JL, Richard EA. Young and older adults use different strategies to perform a standing turning task. Clin Biomech. 2009; 24: 826-832.

[7] Hollman JH, Youdas JW, Lanzinho DJ. Gender differences in dual task gait performance in older adults. Am J Men Health. 2009; 22: 131-134.

[8] Hortobágyi T, Finch A, Solnik S, Rider P, DeVita P. Association between muscle activation and metabolic cost of walking in young and old adults. J Gerontol A Biol Sci Med Sci. 2011; 66A: 541-547.

[9] Skelton DA, Kennedy J, Rutherford OM. Explosive power and asymmetry in leg muscle function in frequent fallers and non-faller aged over 65. Age Aging. 2002; 31: 119-125.

[10] Burnfield JM, Josephson KR, Powers,CM, Rubenstein LZ. The influence of lower extremity joint torque on gait characteristics in elderly men. Arch Phys Med Rehabil. 2000; 81: 1153-1157.

[11] Hermens JH, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyog Kinesiol. 2000; 10: 361-374.

[12] Marques NR, Morcelli MH, Hallal CZ, Gonçalves M. EMG activity of trunk stabilizer muscles during Centering Principle of Pilates Method. J B Mov Ther. 2012; *impress*.

[13] Pavol MJ, Owings TM, Foley KT, Grabiner MD. Influence of lower extremity strength of healthy older adults on the outcome of an induced trip. J Am Geriatr Soc. 2002; 50: 256-262.

[14] Hartmann A, Knolls R, Murer K, Bruin ED. Reproducibility of an isokinetic strength-testing protocol of the knee and ankle in older adults. J Gerontol A Biol Sci Med Sci. 2009; 55: 259-268.

[15] Vaughan CL, Davis BL, O’Connor JC. Dynamics of human gait. 2nd ed. Cape Town: Kiboho Publishers; 1998.

[16] Bowman KF, Fox J, Sekiya JK. A clinically relevant review of hip biomechanics. J Arthroscop Rel Surg. 2010; 26: 1118-1129.

[17] Wert DM, Brach J, Perera S, VanSwearingen JM. Gait biomechanics, spatial and temporal characteristics, and the energy cost of walking in older adults with impaired mobility. Phys Ther. 2010; 90: 977-985.

[18] Liu MQ, Anderson FC, Pandy MC, Delp SL. Muscles that support the body also modulate forward progression during walking. J Biomech. 2006; 39: 2623-2630.

[19] Anderson FC, Pandy MG. Individual muscle contribution to support in normal walking. Gait Post. 2003; 17: 159-169.

[20] Anders C, Wagner H, Puta C, Grassme R, Petrovich A, Scholle HC. Trunk muscle activation patterns during walking in different speeds. J Electromyogr Kinesiol. 2007; 17: 245-252.

3.3. Artigo III

Age-related alterations on trunk and lower limb muscles activation during walking

Nise Ribeiro Marques[a,*], Camilla Zamfolini Hallal[a], Luciano Fernandez Crozara[a], Mary Hellen Morcelli[a], Aline Harumi Karuka[b], Marcelo Tavella Navega[c], Mauro Gonçalves[a]

[a] Department of Physical Education, São Paulo State University, Rio Claro, Brazil

[b] Department of Physical Therapy, São Paulo State University, Presidente Prudente, Brazil

[c] Department of Physical Therapy and Occupational Therapy, São Paulo State University, Marília, Brazil

* Corresponding Author. Address: Universidade Estadual Paulista, Departamento de Educação Física, Avenida 24-A, 1515, Bela Vista, CEP: 13506-900, Rio Claro, SP, Brazil. Email address: nisermarques@yahoo.com.br (N.R. Marques).

1. Introduction

Falls in older adults result in significant costs to healthcare system and are responsible for about 50% of injuries that cause death in people with more than 65 years (Callisaya et al., 2011; Schulz, 2012). Up to 70% of falls in older adults occur during walking (Menz et al., 2003). Thus, differences on gait biomechanics in older fallers and non-fallers were previously investigated (Gabell et al., 2006; Hausdorff, 2007, Lord et al., 2011; Marques et al., 2013). However, the majority of these studies were focused at kinematics of lower limbs (Gabell et al., 2006; Hausdorff, 2007, Lord et al., 2011).

Walking is a complex motor task that requires an integrated coordination of trunk, lower and upper limbs movements (Winter, 2005). Furthermore, head, arms and trunk represent more than 50% of the body mass and greatly influences the dynamic stability of the rest of the body (Kang and Dingwell, 2009). Thus, it is suggested that the neuromuscular control of trunk motion may be prioritized over other inferior segments to maintain stability during movements, such as walking (Kang and Dingwell, 2009).

Aging reduces the dynamic stability across all segments, especially on trunk motion (Kang and Dingwell, 2009). According to Kang and Dingwell (2009), the greater inertia of the trunk may attenuate the effect of a given perturbation on the trunk motion, which may makes the feedback control less effective and reduces the ability to retake the stability after a balance perturbation. Further, Menz et al. (2003) demonstrated that older people at risk of falling have difficulty to control the rhythmic displacements of the trunk during walking. However, to the author’s knowledge no known study has investigated how aging affects the trunk muscles recruitment.

Few studies described the trunk and lower limb muscles activation pattern during walking (Anders et al, 2007; Anders et al., 2009). Anders et al. (2007) and Anders et al

(2009) verified the amount of trunk muscles activation during walking in different speeds in younger healthy adults. The main findings of these studies showed that the trunk muscles had two functions during the gait: perform the inter-vertebral and pelvic movements, which are required to maintain the balance and reduces the metabolic cost of walking; and stabilize the spine and pelvic joints at the stance phase (Anders et al, 2007; Anders et al., 2009). Schmitz et al (2009), investigated the age-related differences on lower limb muscles activation and demonstrated that older adults have greater coactivation of ankle muscles during mid-stance.

Considering that trunk and lower limb muscles may have an important role to maintain the postural control and balance, which have an important clinical appliance to prevent falls in older adults. Our study aimed to compare the trunk (multifidus and internal oblique) and lower limb (gluteus maximus, rectus femoris, biceps femoris, gastrocnemius lateralis and tibialis anterior) muscles activation between younger and older women at stance phase of gait. We hypothesized that older female adults would have lower activation of lower limb and trunk muscles, which may be related with age-related weakness and balance impairments.

2. Methods

2.1. Participants

Data of thirty-four female adults were considered for this study (Table 1). Fifteen younger volunteers were recruited from a university setting and nineteen older female adults were recruited from community-based physical activity groups. Older women were heavier ($p = 0.016$), shorter ($p < 0.001$) and had higher body mass index ($p < 0.001$). All participants signed a consent form approved by the Institutional Ethics

Committee. People who had uncontrolled cardiovascular disease, diagnosed dementia or cognitive impairment (defined as a Mini-Mental State Examination score < 20), balance disturbance (defined as a BERG balance score < 36), hemiparesis, pain in the lower limbs or trunk, or a progressive motor disorder were excluded.

Table 1 Subject characteristics.

Variables	Pilates group (n = 10)	Control group (n = 9)	*P*
Age (y)	**64.2 ± 3.08**	**72.6 ± 8.4**	**0.008***
Mass (kg)	66.5 ± 6.7	63.1 ± 8.7	0.2
Height (m)	1.53 ± 0.05	1.53 ± 0.04	0.6
Body mass index ($kg \cdot m^{-2}$)	28.3 ± 2.6	26.9 ± 4.7	0.07
BERG Balance Scale	55.1 ± 0.9	55.2 ± 1.9	0.8
Mini Mental Examination Score	24.1 ± 2.9	24.6 ± 3.6	0.8
Preferred overground walking speed ($m \cdot s^{-1}$)	1.2 ± 0.2	1.1 ± 0.1	0.5
Preferred treadmill walking speed ($m \cdot s^{-1}$)	1 ± 0.2	1.01 ± 0.1	0.7

* Significant difference ($p < 0.05$).

2.2. Procedure

The volunteers visited the laboratory on two separate occasions within 24-72 hours. On the first day of data collection, the preferred overground and treadmill walking speeds (PTWS) were measured. During the second day of data collection, the volunteers were familiarized with treadmill walking at PTWS (10 minutes) and after that surface electromyography signals (EMG) were recorded.

EMG activity (Figure 1) was assessed using an 8-channel, telemetered electromyogram (Noraxon®, Phoenix, USA). EMG signals were collected at sample frequency of 2000 Hz, using Ag/AgCl disc electrodes (Miotec®, Porto Alegre, Brazil) with an active area of 1 cm^2 and inter-electrode distance of 2 cm arranged in bipolar configuration. The electrodes were positioned on the participants' right side on the muscles: internal oblique (IO) - 2 cm medial and inferior to the anterior superior iliac spine (Marshall and Murphy, 2003); multifidus (MU) - 2 cm lateral to space between the spinous processes of L4-L5; gluteus maximus (GM), at 50% of the line between the sacral vertebrae and the greater trochanter; biceps femoris (BF), at 50% of the line between the ischial tuberosity and the lateral epicondyle of the tibia; rectus femoris (RF), at 50% of the line from the anterior superior iliac spine to the superior border of the patella; tibiallis anterior (TA), at 1/3 of the line between the proximal head the fibula and the tip of the medial malleolus; and gastrocnemius lateralis (GL), at 1/3 of the line between the proximal head of the fibula and the calcaneus (Hermens et al., 2000). A reference electrode was placed on the right medial maleolus. Before placing the electrodes, the subject's skin was shaved and cleaned with alcohol to reduce impedance (Hermens et al., 2000). Hz. Heel contact and toe-off were detected by one footswitch (Noraxon®, Phoenix, USA), composed of four sensors attached on right foot at the heel, first and fifth metatarsals, and toe, which was synchronized with the EMG system.

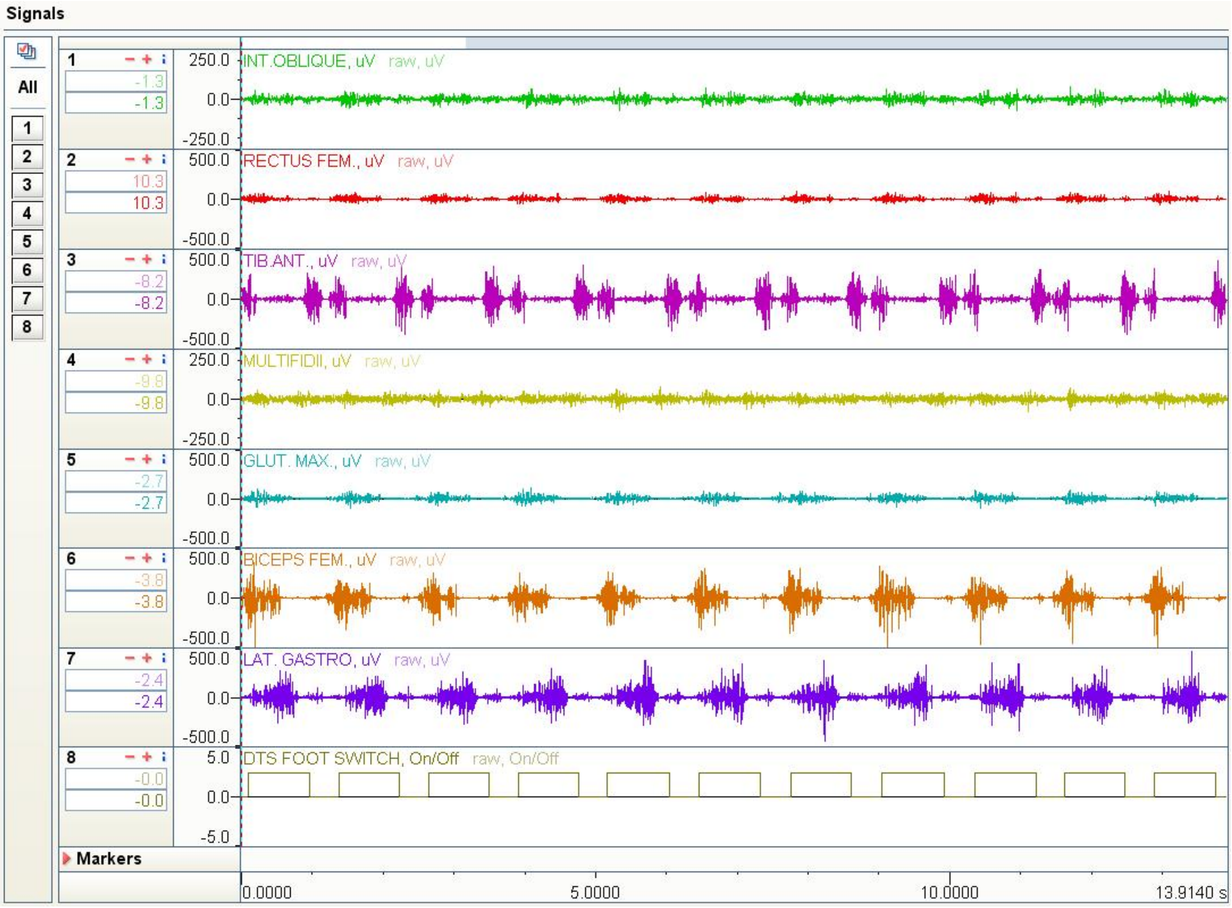

Fig. 1. EMG signals and footswitch data from an older female subject (69 yr) recorded during walking.

2.3. Gait analysis

Preferred overground walking speed was determined on the first day of data collection. Volunteers were instructed to walk at their self-selected speed at a natural pace over 20 m and the duration for each trial was determined with two infrared timing gates placed at 5 and 15 m. Preferred overground walking speed was calculated by dividing distance walked (10 m) by the time to cover this distance (s).

PTWS was determined by starting treadmill walking at 50% of the preferred overground speed. Then, speed was increased until the subject reported that the current speed was faster than preferred gait speed and slowly decreased until the subject reported that the current speed was slower than preferred. This procedure was repeated three times and the average of the 3 ‘‘faster’’ and 3 ‘‘slower’’ than preferred speeds was taken as the subject’s PTWS. Then, the volunteers performed a habituation trial of 5 minutes walking on the treadmill at the self-selected speed.

In the second day of data collection, volunteers were familiarized to the PTWS for 10 minutes. After that while still walking at PTWS, 1 minute of EMG data was recorded

2.4. Data Analysis

The EMG signal was processed in specific routines developed in Matlab (Mathworks®, Natick, USA) using a band pass filter with cut-off frequency of 20-500 Hz, full-wave rectification and a low pass, fourth order filter with a cut-off frequency of 10 Hz. Then, the mean of the linear envelope of EMG signal was obtained 50 ms after heel contact (initial stance phase), and before toe-off (final stance phase) of the first ten strides. All the linear envelope values were normalized to the mean activation obtained during the gait.

PASW 18.0 (SPSS inc.) was used for all statistical analyses. We used appropriate descriptive statistics (mean and standard deviation) to summarize participant characteristics. The Shapiro-Wilk test was used to test the normality of the data. Covariance multivariate analysis (MANCOVA) was used to compare the dependent variables (muscles activation) between groups, using as covariate gait speed. Also, in order to evaluate the influence of age on the muscles activation we computed the partial correlation coefficients to quantify the association between age and EMG activity and gait velocity was entered as a covariate. The significance level was set to $p < 0.05$.

3. Results

MANCOVA analysis showed significant main group effect (F = 2.623 and p = 0.026). Younger female adults had 23.96% faster PTWS (p < 0.001; Figure 2), 52.32% higher activation of IO (p = 0.001; Figure 3) and 39.95% higher activation of RF (p = 0.004; Figure 3) at initial stance. Also, at final stance, younger women had 57.01% higher activation of BF (p = 0.008; Figure 3) and older women had 39.82% higher activation of TA (p = 0.34; Figure 3). In addition, partial correlations showed that RF activation at initial stance and BF activation at final stance were negatively associated with age (r = - 0.422 and p = 0.013; and r = - 0.341 and p = 0.041).

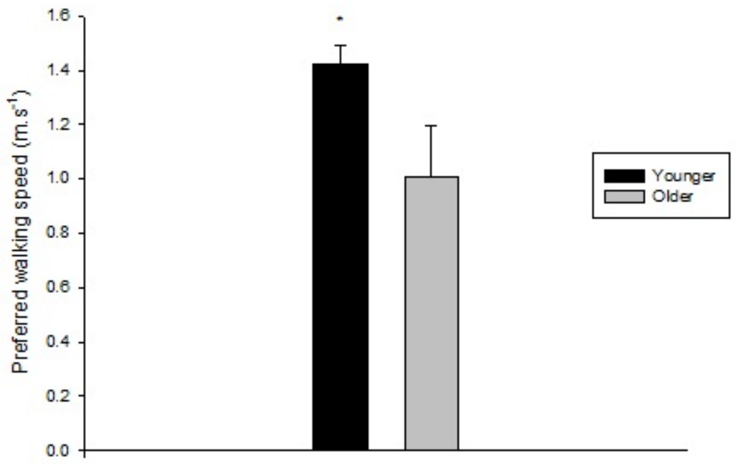

Fig. 2. Comparisons of preferred walking speed between younger and older adults. * p < 0.05.

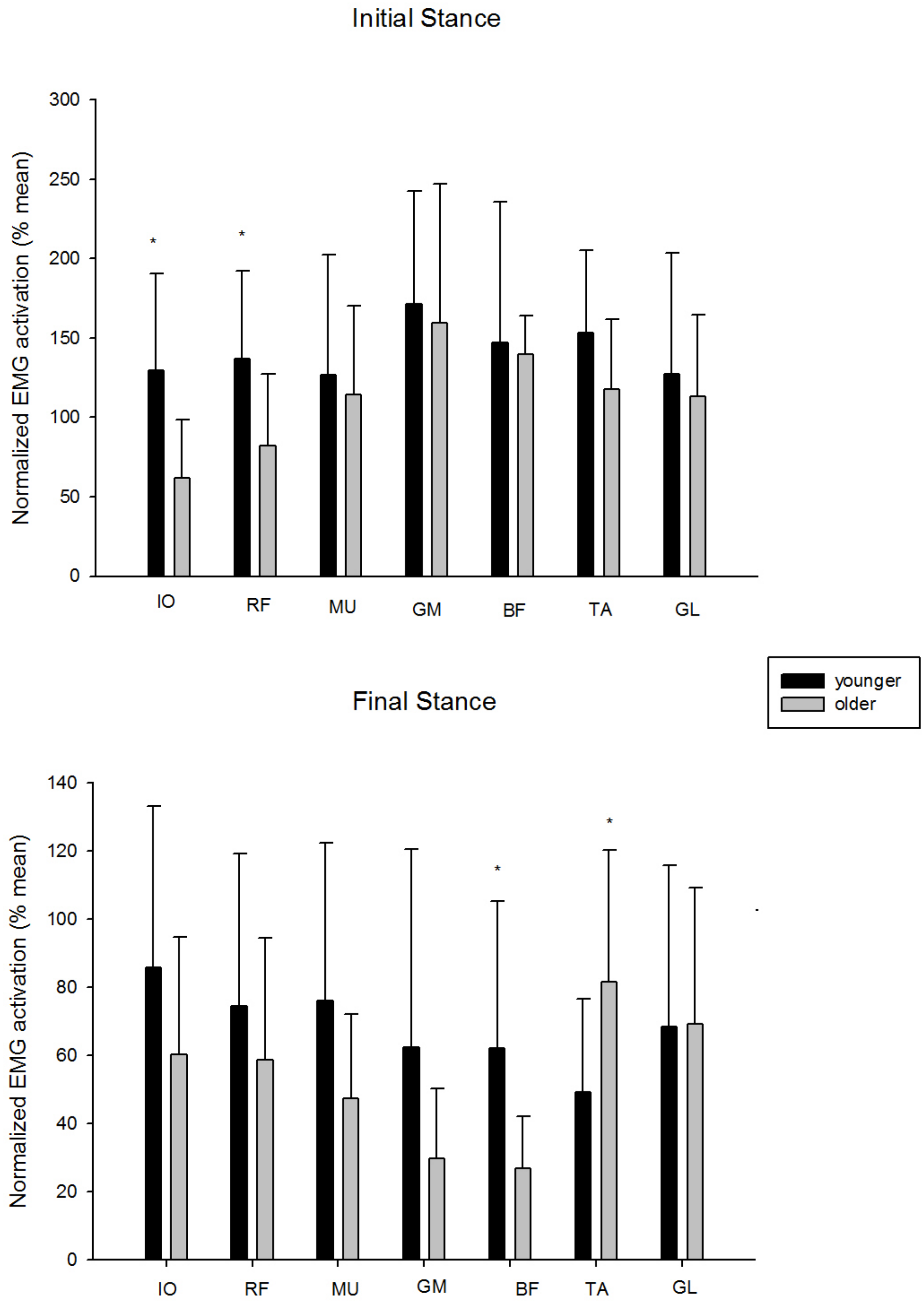

Fig. 3. Comparisons of trunk and lower limb muscles activation between younger and older adults. * $p < 0.05$.

4. Discussion

The present study compared the activation of trunk (IO and MU) and lower limb (GM, RF, TA, GL, and BF) muscles activation between younger and older adults during the stance phase of the gait. Only during the initial stance phase for IO and RF muscles were found significant differences between groups. Thus, our results partially agree with the previous hypothesis that older female adults would have lower trunk and lower limb muscles activation during stance phase of the gait.

During the gait, IO and RF have similar activation pattern, which is characterized by two peaks of activation, the first at initial stance (heel strike) and the second at final stance (push-off phase). At the initial stance, IO is recruited to provide stability to trunk segment, while the eccentric contraction of RF contributes to decelerate the anterior displacement of the center mass (Vaughan et al., 1998). Thus, according to our findings older adults may have the impaired trunk and lower limb stability during the initial stance phase, which could increase the risk of falling.

Considering that the amount of active motor units is indirectly related with the force generated (DeLuca, 1997). Older adults have lower limb weakness and aging particularly influences quadriceps muscles strength (LaRoche et al, 2010; Crozara et al., 2013). According to Ikezoe et al. (2011), quadriceps muscles are the lower limb group muscle that most lost muscle mass during aging. Also, these authors showed that the muscle mass loss in quadriceps has impact on the mobility status.

For IO and RF muscles the walking speed has an important contribution on the amount of EMG activation during this phase. According to Anders et al (2007), IO activation pattern is mixed by a continuous activation at low walking speed (0.55-0.83 $m \cdot s^{-1}$) and phasic activation at high speeds (1.11-1.66 $m \cdot s^{-1}$) (Anders et al., 2007). In addition, there is a positive association between walking speed and weight acceptance force at heel strike, which leads to increase the amount of the RF activation (LaRoche et al., 2011).

Despite our findings that demonstrated that there is a negative correlation between age and walking speed (Novaes et al., 2011; Bohannon et al., 1996; Bohannon et al., 1997) are in concordance with the majority studies. This difference on the walking speed between younger and older female adults is the main limitation of this study. Thus, to attenuate the effect of this limitation on the comparison of muscles activation between younger and older adults we considered the walking speed as covariate in the statistical analysis. In addition, with respect to our results, we found that younger adults had greater activation of BF and older adults had greater activation of TA at the final stance, which may leads to the difference on gait speeds between groups.

During the final stance, the plantarflexors are highly activated to push the foot against the ground while knee flexors and hip extensors are also activated to promote anterior acceleration of the body (Vaughan et al., 1998). Thus, the higher activation of BF in younger women may contribute to increase their walking speed. On the other hand, the greater co-activation between TA and GL in older adults during final stance may compromise the capacity to generate torque and accelerate the body to perform the next step (Schmitz et al., 2012).

Surface EMG is frequently used to describe muscles activation pattern during human gait (Anders et al., 2007). Recently studies used surface EMG to measure and describe trunk muscles activation during walking (Anders et al., 2007; Anders et al., 2009). Despite the amount of fat tissues above the electrodes and the cross talk between the muscles (IO and transversus abdominis; and MU and erectors spinae) surface EMG is a non-invasive and safe method to gives a clue about spinal stability (Anders et al., 2007).

Considering that body mass index has a strong correlation with the amount of fat tissue, in our sample, older female adults should had higher concentration of fat tissue, specially, above the areas where the electrodes were placed. Thus, the amount of EMG activation may be reduced in this group. However, to try to avoid this limitation for the use of surface EMG we amplified and normalized the signal.

5. Conclusion

This is an initial study about how age affects the EMG activation pattern of trunk and lower limb muscles during walking. The main findings of this study demonstrated that older female adults had lower activation of trunk and knee muscles during initial stance, which may be related with weakness and balance impairments caused by aging and predispose this population to increased risk of falling. Thus, we suggest that future studies should be performed to identify the effect of aging on lower limb joint and trunk stability and loading distribution during walking.

Acknoledgement

This project was funded by grants from Fundação de Amparo à Pesquisa do Estado de São Paulo (FAPESP), Conselho Nacional de Desenvolvimento Científico e Tecnológico (CNPq), Fundação para o Desenvolvimento da UNESP (FUNDUNESP) and Coordenação de Aperfeiçoamento de Pessoal de Nível Superior (CAPES).

References

Anders C, Wagner H, Puta C, Grassme R, Petrovitch A, Scholle HC. Trunk muscle activation patterns during walking in different speeds. J Electromyogr Kinesiol 2007; 17: 245-252.

Anders C, Wagner H, Puta C, Grassme R, Petrovitch A, Scholle HC. Healthy humans use sex-specific co-ordination of trunk muscles during gait. Eur J App Physiol 2009; 105: 585-594.

Bohannon RW, Andrews A W, Thomas MW. Walking speed: reference values and correlates for older adults. J Orthop and Sports Phys Ther 1996; 24(2): 86-90.

Bohannon RW. Comfortable and maximum walking speed of adults aged 20-79 years: reference values and determinants. Age Aging 1997; 26(1): 15-19. Callisaya ML, Blizard L, Schmidt MD, Martin KL, McGinley JL, Sanders LM, Srikanth VK. Gait, gait variability and the risk of multiple incident falls in older people: a population-based. Age and Ageing 2011; 40:481-487.

Crozara LF, Morcelli MH, Marques NR, Hallal CZ, Spinoso DH, Almeida Neto AF, Cardozo AC, Gonçalves M. Motor readiness and joint torque production in lower limbs of older women fallers and non-fallers. J Electromyogr Kinesiol 2013, impress.

De Luca CJ. The use of surface electromyography in biomechanics. J App Biomech 1997;13: 135-163.

Gabell A, Nayak U, Wada M, Sasaki S, Kawahara H, Uchida K. Effect of compensation procedures for velocity on repeatability ans variability of gait parameters in normal subjects. Clin Rehabil 2006; 20:239-245.

Hausdorff J. Gait dynamics, fractals and falls: finding meaning in the stride-to-stride fluctuations of human walking. Hum Mov Sci 2007; 26:557-589

Hermens JH, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol 2000; 10: 361-374.

Ikezoe T, Mori N, Nakamura M, Ichihashi N. Atrophy of the lower limb in elderly women: is it related to walking ability? Eur J Appl 2011;111: 989-995.

Kang HG, Dingwell JB. Dynamic stability of superior vs. inferior segments during walking in young and older adults. Gait Posture 2009; 30: 260-263.

LaRoche DL, Millet ED, Kralian RJ. Low strength is related with diminished ground reaction forces and walking performance in older women. Gait Post 2011; 33: 668-672.

LaRoche DP, Cremin KA, Greenleaf B, Croce RV. Rapid torque development in older female fallers and non-fallers: a comparison across lower-extremity muscles. J Electromyogr Kinesiol 2010; 20(3):482-8.

Lord S, Howe T, Greenland J, Simpson L, Rochester L (2011) Gait variability in older adults: A structured review of testing protocol and clinimetrics properties. Gait & Posture 2011;34:433-450.

Marques NR, LaRoche DP, Hallal CZ, Morcelli MH, Crozara LF, Karuka AH, Navega MT, Gonçalves M. Association between energy cost of walking, muscle activation, and biomechanical parameters in older female fallers and non-fallers. Clin Biomech 2013, *impress*.

Marshall P, Murphy B. The validity and reliability of surface EMG to assess the neuromuscular response of the abdominal muscles to rapid limb movement. J Electromyogr Kinesiol 2003; 13: 477-489.

Menz HB, Lord RS, Fritzpatrick RC. Age-related in walking stability. Age and Ageing 2003; 32: 137-142.

Novaes RD, Miranda AS, Dourado VZ. Usual gait speed in middle-aged and elderly Brazilian subjects. Br J Phys Ther 2011; 15(2): 117-122.

Schmitz A, Silder A, Heiderscheit B, Mahoney J, Thelen DG. Differences in lower-extermity muscular activation during walking between healthy older and young adults. J Electromyogr Kinesiol 2008; 19: 1085-1091.

Schulz BW. Healthy younger and older adults control foot placement to avoid small obstacles during gait primarily by modulating step width. J NeuroEng Rehab 2012; 9:1-9.

Vaughan CL, Davis BL, O'Connor JC. Dynamics of human gait. 2nd ed. Cape Town: Kiboho Publishers; 1998.

Winter DA. Biomechanics and motor control of human movement. 3rd ed. United Kingdom: Hardcover, 2005.

3.4. Artigo IV – Artigo submetido para *Journal of Applied Biomechanics*

EIGHT-WEEKS OF PILATES METHOD INTERVENTION INCREASES TRUNK AND LOWER LIMB STABILIZER MUSCLES ACTIVATION DURING WALKING IN OLDER FEMALE ADULTS

Nise R. Marques,[1] Camilla Z. Hallal,[1] Mary H. Morcelli,[1] Luciano F. Crozara,[1] Aline Harumi Karuka,[2] Marcelo T. Navega,[3] Mauro Gonçalves[1]

[1] Department of Physical Education, Institute of Bioscience, São Paulo State University, Rio Claro, SP, BRA

[2] Department of Physical Therapy, Faculty of Science and Technology, Presidente Prudente, SP, BRA

[3] Department of Physical Therapy and Occupational Therapy, Faculty of Phylosophy and Science, São Paulo State University, Marília, SP, BRA

Funding: This study was funding by grants from Fundação de Âmparo à Pesquisa do Estado de São Paulo (FAPESP), Conselho Nacional de Desenvolvimento à Pesquisa (CNPq), Fundação de Desenvolvimento da UNESP (FUNDUNESP), and Coordenação de Aperfeiçoamento de Pessoal de Nível Superior (CAPES).

Conflict of Interest Disclosure: None of the authors had any conflict of interest which could bias the results of this study.

Abstract

Falls in older adults may be prevented with regular practice of exercise. Among the exercise prescribed for older population is the Pilates Method. However, little is known about the neuromuscular responses of Pilates intervention. This study aimed to identify the effect of eight-weeks of Pilates intervention on trunk and lower limb stabilizer muscles activation pattern during the gait in older female adults. Eighteen older female adults were separated into two groups: Pilates (n = 10) and control (n = 8). Electromyographic (EMG) activation was recorded during one-minute of walking on treadmill at preferred speed on the muscles: internal oblique (IO), multifidus (MU), rectus femoris (RF), gluteus maximum (GM) and biceps femoris (BF). EMG activation was analyzed at initial and final stance. EMG onset time for IO and MU muscles was identified before heel strike. After Pilates intervention IO, MU and RF activation increased ($p = 0.009$, $p = 0.001$, and $p = 0.04$). Additionally, Pilates intervention reduced in 40% IO onset time before heel strike ($p = 0.04$). According to our results, eight-weeks of Pilates intervention increased IO, MU and RF activation at initial stance of gait, and promote earlier IO activation response before heel strike in older female adults.

Keywords: gait, falls, aging, physical therapy, exercise, biomechanics

Word count: 2,785.

1. Introduction

Falls in older adults are potentially damage and cause injuries, loss of functional independence and death.[1,2] Among the causes of falls in older adults are the age-related alterations in neuromuscular system, which results in weakness and balance loss. [1-3] Thus, the regular practice of physical activity is considered the major intervention to prevent falls and the decline of physical function. [4] Recently, some methods of physical exercises, such as Pilates Method and vibration training, which aim to automatize core muscles recruitment, have been widely prescribed for physical therapists those would improve balance and stability during walking and prevent falls in older adults. [5,6]

Pilates Method is an exercise technique that has the following principles: Concentration, Control, Flow, Precision, Breath and Centering.[8] Among these principles, Centering deserves special attention because it requires the voluntary isometric contraction of core muscles while the exercises are performed.[8] Furthermore, Pilates Method exercises require trunk stabilization while rhythmic movements of lower and upper limbs are performed, such as occurs during the gait.

Despite Pilates Method has been prescribe for older population[9], to the authors knowledge no study has yet investigated the effect of Pilates intervention on trunk and lower limb stabilizer muscles activation during walking in older adults. Also, just few studies had investigated the effect of Pilates intervention on lower limb strength and balance in older population.[7,9] Bird et al. (2012)[9] found that five-weeks of Pilates intervention did not promote significant increments on strength of knee extensor and ankle dorsiflexors, and did not change the static (center of mass displacement) and dynamic (Four Square Step Test e Time up and Go Test) balance. In opposite of Bird et al. (2012)[9] study, Rodrigues et al. (2010)[7] found that eight-weeks of Pilates training promoted significant increment on balance (assessed by Tinetti scale).

Considering that little is known about the effect of Pilates intervention on neuromuscular system, our study aimed to identify the effect of eight-weeks of Pilates Method intervention on trunk and lower limb stabilizer muscles (internal oblique, IO; rectus femoris, RF; multifidus, MU; gluteus maximum, GM; e biceps femoris, BF) activation pattern during the gait in older female adults. Our hypothesis is that eight-weeks of Pilates Method intervention can promote significant increment on trunk and lower limb stabilizer muscles activation and can reduce the onset time of trunk stabilizer muscles (IO and MU) before heel strike.

2. Methods

2.1. Participants

Independently living and ambulating older female adults (n = 37; Table 1) were recruited through local community physical activity groups in an urban area. Participants were included if they did not currently have or had not recently had an acute medical condition. Volunteers who had controlled chronic conditions such as arthritis or stable chronic cardiovascular or metabolic conditions (e.g., hypertension, diabetes mellitus) were included in the study. The institutional human research ethics committee gave ethical approval for this study, and written informed consent was obtained from all participants before participation.

Table 1 Subject characteristics.

Variables	Pilates group (n = 10)	Control group (n = 8)	*P*
Age (y)	**64.2 ± 3.08**	**72.6 ± 8.4**	**0.008***
Mass (kg)	66.5 ± 6.7	63.1 ± 8.7	0.2
Height (m)	1.53 ± 0.05	1.53 ± 0.04	0.6
Body mass index ($kg{\cdot}m^{-2}$)	28.3 ± 2.6	26.9 ± 4.7	0.07
BERG Balance Scale	55.1 ± 0.9	55.2 ± 1.9	0.8
Mini Mental Examination Score	24.1 ± 2.9	24.6 ± 3.6	0.8
Preferred overground walking speed ($m{\cdot}s^{-1}$)	1.2 ± 0.2	1.1 ± 0.1	0.5
Preferred treadmill walking speed ($m{\cdot}s^{-1}$)	1 ± 0.2	1.01 ± 0.1	0.7

* Significant difference ($p < 0.05$).

2.2. Study design

A randomized clinical design (Figure 1) was used to test the effect of Pilates intervention on lower limb and trunk muscle activation pattern during the gait in older female adults. Dependent variables were measured at 2 times: baseline (t1), and immediately after the intervention (t2). Each group completed eight-weeks of the Pilates or control intervention.

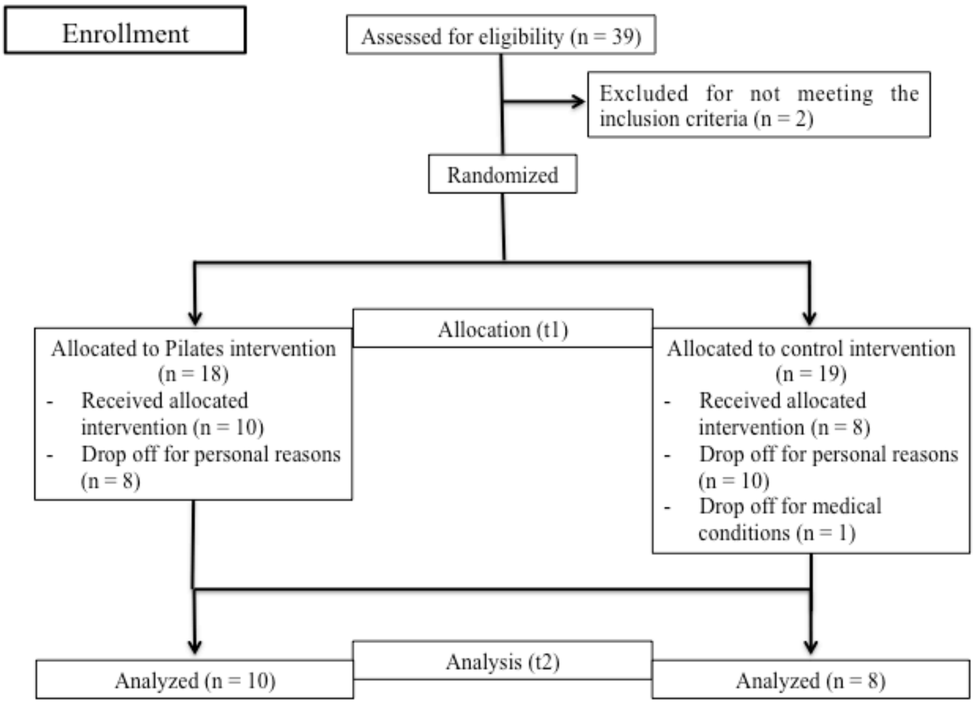

Figure 1 – Consolidated standards of reporting trials flow diagram of study design.

2.3. Outcome Measures

Data were collected from participants during two sessions separated by 24–72 hours. During the first data collection visit, preferred overground walking speed (POWS) and preferred treadmill walking speed (PTWS) were measured as well as an habituation section of ten minutes walking on the treadmill at PTWS were performed. During the second day of data collection, participants used a treadmill for 10 minutes to familiarize themselves with gait at their PTWS; then, one minute of gait biomechanics (electromyographic, EMG, signal) at PTWS was recorded.

EMG signals were collected at a sampling frequency of 2000 Hz using Ag/AgCl (Miotec, Porto Alegre, Brazil) disc electrodes with an active area of 1 cm^2 and an inter-electrode distance of 2 cm in a bipolar configuration. The electrodes were positioned on the following muscles of the right side of the participants' body: the internal oblique (IO), 2 cm medial and inferior to the anterior superior iliac spine; the multifidus (MU), 2 cm lateral to the space between the spinous processes of L4-L5[8]; the gluteus maximus (GM), on the midpoint of the line between the sacral vertebrae and the greater trochanter; the biceps femoris (BF), on the midpoint of the line between the ischial tuberosity and the lateral epicondyle of the tibia; and the rectus femoris (RF), on the midpoint of the line from the anterior spine iliac superior to the superior part of the patella.[10] A reference electrode was placed on right malleolus medialis. Subjects' skin was shaved and cleaned with alcohol before placing the electrodes.[10] Also, four footswitches sensors (Noraxon, Phoenix, USA) were attached to the heel, first and fifth metatarsus and toe, and they signal were synchronized with the EMG signal.

On the first day of data collection, PTWS was determined by starting at 50% of the POWS. To determine POWS, participants were instructed to walk at a self-selected natural pace for 20 m. The duration of each trial was determined with two infrared timing gates placed at 5 m and 15 m distances. POWS was calculated by dividing the distance walked (10 m) by the amount of time required to walk this distance. Velocity was obtained in units of $m \cdot s^{-1}$. Participants were asked to walk on the treadmill; its speed was increased until the subject reported that it was faster than her preferred gait speed. Treadmill speed was then slowly decreased until the subject reported that the current speed was slower than preferred. This procedure was repeated three times and the average of the 3 "faster" and 3 "slower" than preferred speeds was taken as the subject's PTWS. On the second day of data collection, participants were familiarized with the PTWS for 10 minutes. Then, 1 minute of EMG data were recorded while the participants walked at the PTWS.

2.4. Interventions

Pilates and control interventions were performed in two classes per week, one hour each for eight consecutive weeks. Both interventions were composed by aerobic, strength, stretching and lumbar spine stabilization exercises (Pilates Method and oscilatory pole). Classes were held in small groups of no more than eight people and were conducted by three physical therapists and one physical educator.

Participants of both groups completed aerobic, strength, stretching and lumbo-pelvic stabilization exercises two days per week. Each aerobic session involved a 10 min of low-intensity warm-up (heart rate < 60% of maximum, dance and walking), five minutes of flexibility exercises were performed seated (5–7 different exercises, held for 15 seconds once on each side). The strength training involved 15 min of 5-8 different exercises performed with free weights and elastic bands. The muscles and muscle groups targeted during strength training were: quadriceps, calves, hip abductors and adductors, hamstrings, latissimus dorsi, pectorals, abdominals, triceps, deltoids, and biceps. Thirty minutes of lumbo-pelvic stabilization exercises (Pilates or vibration training) were performed on each group.

Pilates Method intervention was composed by mat exercises with progressive order of difficult. Also, exercise balls and elastic band were used to challenging the exercises. An initial class was performed to habituate the participants with Pilates Method principles. In the following classes, lower and upper limb and trunk exercises were introduced.

The control condition was also composed by lumbo-pelvic stabilization exercises. Thus, we used an oscilatory pole (LiveUp®, São Paulo, BRA), with natural frequency of 5 Hz, 1.6 m of length and 0.63 kg of weight. According to the literature, oscilatory pole exercises promote increase on IO and MU activation also may automatize core muscles recruitment while lower and upper limbs movements are performed.[11,12] An initial class for subjects's familiarization with the use of oscillatory pole was made and the exercises had a progressive difficult order. Also, during all class a metronome set at 5 Hz was used to control the frequency of the pole. For both interventions, we included in the study's sample only the subjects that were at more than 75% of the classes (presence in more that 12 classes).

2.5. Data analysis

The EMG signal was processed in specific routines developed in Matlab (Mathworks®, Natick, USA) using full-wave rectification and a low pass, fourth order filter with a cut-off frequency of 10 Hz. Then, the mean of the linear envelope of the EMG signal was obtained 50 ms after heel strike and before toe-off of the first ten strides. All the linear envelope values were normalized to the mean activation obtained during the gait.

The IO and MU muscles onset before heel contact was detected considering the muscle active when the signal magnitude surpassed two standard deviations of the minimal magnitude of the average signal per individual.[13]

PASW 18.0 (SPSS inc.) was used for all statistical analyses and means and standard deviations were used to summarize participant characteristics. Then, the Shapiro-Wilk test was used to determine if the data were normally distributed. ANOVA two-way repeated measures were performed for each muscle, with conditions (Pilates Method and control) and sessions (before and after eight-weeks of intervention) as independent variables and normalized EMG activation as dependent variable. For posthoc comparisons, paired t-tests with Bonferroni correction were used. Also, we used ANOVA one-way test to identify significant differences when ANOVA repeated measures showed interactions. Non-parametric t-test of Wilcoxon was used to compare IO and MU onset before and after eight-weeks of intervention for each condition (Pilates and control interventions). The significance level for all statistical tests was set at $p < 0.05$.

3. Results

At the initial stance phase, ANOVA two-way repeated measures showed non-significant main effect between groups ($p = 0.434$; Table 2) and sessions (before and after intervention; $p = 0.104$; Table 2). However, a significant interaction between groups and sessions was found ($p = 0.01$; Table 2). Thus, IO, MU and RF activation were, respectively, 61.41, 36.03 and 61.11% higher after Pilates intervention ($p < 0.001$, $p = 0.047$ and $p = 0.008$; Figure 2).

Table 2 Repeated measures ANOVA results on the main effect of groups (Pilates, control) and sessions (before and after interventions) on EMG activity at initial stance (50 ms after heel strike).

	F	*P*
Groups	1.225	0.434
Sessions	3.942	0.104
Groups*Sessions	**15.489**	**0.01***

* Significant interaction between groups and sessions ($p < 0.05$).

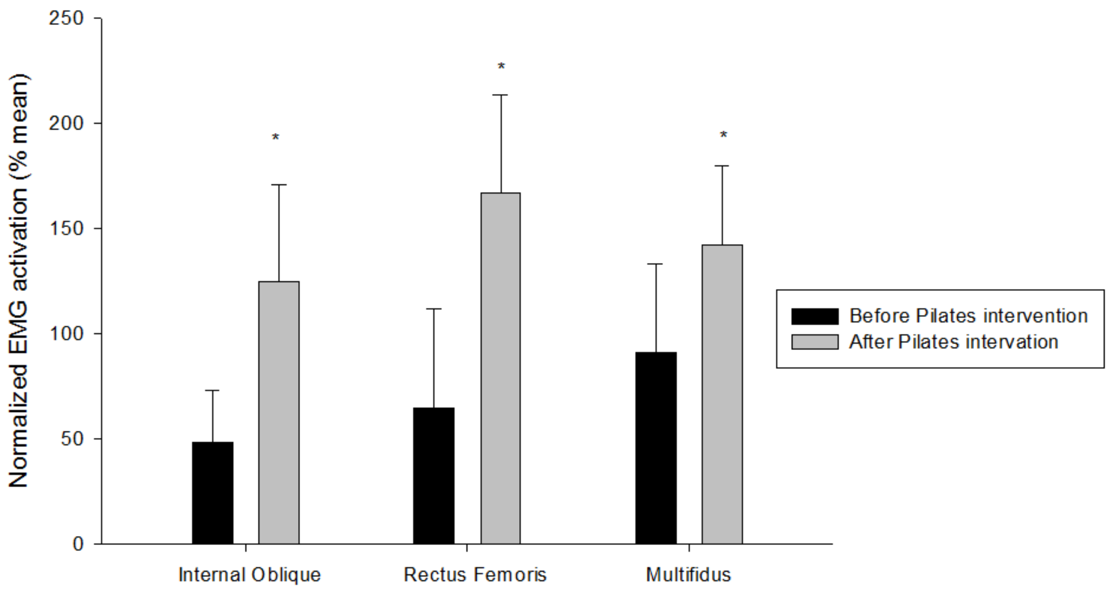

Figure 2 – Mean and standard deviation of internal oblique, rectus femoris and multifidus EMG activation at initial stance before and after Pilates intervention. * Significant difference ($p < 0.05$).

At the final stance phase, ANOVA two-way repeated measures showed non-significant main effect between groups ($p = 0.432$; Table 3) and non-significant interaction between groups and sessions ($p = 0.263$; Table 3). However, a significant main effect of sessions (before and after intervention) was found ($p = 0.01$; Table3). Thus, IO, RF, MU, GM and BF activation were, respectively, 69.39 ($p < 0.001$; Figure 3), 57.42 ($p = 0.004$; Figure 3), 48.63 ($p = 0.012$; Figure 3), 67.42 ($p = 0.011$; Figure 3) and 75.8% ($p < 0.001$; Figure 3) higher after the interventions. Also, a significant decrease of 39.14% on IO onset was found after Pilates intervention ($p = 0.04$).

Table 3 Repeated measures ANOVA results on main effect of groups (Pilates, control) and sessions (before and after interventions) on EMG activity at final stance (50 ms before toe-off).

	F	*P*
Groups	1.233	0.432
Sessions	**15.135**	**0.01***
Groups*Sessions	1.984	0.263

* Significant main effect of sessions ($p < 0.05$).

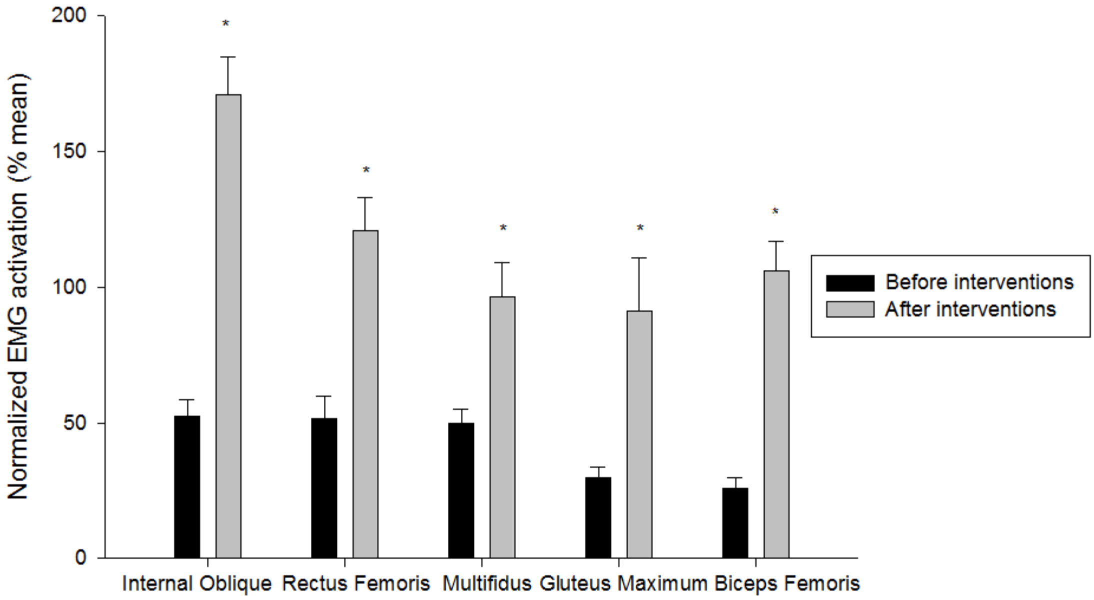

Figure 3 – Mean and standard error of internal oblique, rectus femoris, multifidus, gluteus maximum and biceps femoris EMG activation at final stance before and after interventions. * Significant difference ($p < 0.05$).

4. Discussion

This study aimed to identify the effect of eight-weeks of Pilates Method intervention on trunk and lower limb stabilizer muscles activation pattern during the gait in older female adults. With respect to our results, the most novel aspect of this study is that it demonstrated that eight-weeks of Pilates Method intervention increased trunk (IO and MU) and lower limb (RF) stabilizer muscles activation in average 52% during the initial stance (100 ms after heel strike). Also, eight-weeks of Pilates Method intervention reduced in 40% IO onset time before heel contact. Thus, these findings partially agree with our initial hypothesis that eight-weeks of Pilates Method intervention can promote significant alterations on trunk and lower limb stabilizer muscles activation pattern during walking.

4.1. Effect of training conditions on trunk and lower limb stabilizer muscles activation pattern during the gait

Lumbar spine stability is primarily provided by trunk muscles recruitment.[14] Thus, trunk muscles are classified according to their functions into two different groups: local and global systems muscles. Local system muscles (i.e. transversus abdominis) are permanently active to ensure trunk stability, while global system muscles have phasic activation pattern and are recruited to initiate trunk movements.[15,16] Global system muscles are, recently, classified into two subgroups: global mobilizer muscles (i.e. rectus abdominis and erectus spinae), which act to provide trunk motion; and global stabilizer muscles (i.e. IO and MU), which complement the function of local system muscles by controlling and limiting trunk movements. [15,16]

According to Anders et al. (2007)[15] IO and MU act like global stabilizer muscles during the gait because it preserve the phasic recruitment, shown two peaks of activation at heel strike and at propulsion phase, and limits the intervertebral joint range of motion. The two peaks of IO and MU activation occur because, during initial and final stance, trunk and lower limb muscles are recruited to stabilize the joints and dissipate the vertical ground reaction forces (VGRF).[15] With respect to our results, the increased activation of trunk stabilizer muscles at initial stance increased in 39.67% the simultaneous cocontraction of IO and MU (ratio IO/MU) after Pilates intervention, which must improved the trunk stability.[17] Thus, improved trunk stability at initial stance may reduce the kinetic perturbation caused by VGRF at lumbo-pelvic region and also improve the balance resulting in a stable gait.

Also, we found earlier IO onset after Pilates intervention, which may demonstrates that this intervention was able to modify the anticipatory postural adjustment. Few studies investigated the effect of exercise intervention on anticipatory postural adjustment.[18,19] According to Tsao and Hodges (2008)[18] four-weeks training of repeated isolated voluntary transversus abdominis (TrA) contraction in young adults with low back pain reduced the onset time of this muscle. Also, these authors found that reduced TrA onset time can be retained for six months after training cessation.[18] To our knowledge no known study investigated the effect of earlier trunk stabilizer muscle onset on trunk stability during walking in older adults. However, we suggest that earlier activation of IO may improve the balance, reduce gait variability and the risk of falls.

Our findings also showed that RF activation at initial stance increased after Pilates intervention. During the initial stance, RF acts eccentrically to reduce the anterior displacement of center of mass, while stabilizes knee and hip joints.[20] Several studies identified that age-related weakness of knee extensors is related with impaired mobility and abnormal biomechanical gait pattern.[21-23] Thus, the increased capacity to recruit more motor units during initial stance may be related with improved knee and hip joint stability.

Eight-weeks of Pilates and control interventions increased trunk and lower limb stabilizer muscles activation at final stance. Both, trunk (IO and MU) and lower limb (RF, GM, and BF) stabilizer muscles, show a second peak of activation during the final stance.[15,24] At final stance, these muscles provide joint stability while the center of mass is accelerated forward to perform the next step. Thus, we suggest that lumbo-pelvic stabilization exercises (Pilates and vibration training) can automatize trunk and lower limb stabilizer muscles activation during the propulsion gait phase. Also, improved trunk and lower limb stability during final stance may result in stable gait and aid to increase gait speed.

This study is an initial investigation about the effect of Pilates intervention on trunk and lower limb stabilizer muscles during the gait in older female adults. Despite we found a significant increment on trunk (IO and MU) activation after Pilates intervention, which may have increased the trunk stability, this study did not direct measure spinal stability. Also, little is known about the role of trunk and hip stabilizer muscles during walking. Thus, we suggest that future investigations should be conducted to identify the effect of Pilates intervention on trunk stability and balance during walking in older adults.

4.2. Limitations

This study has some limitations that must be considered on the results interpretation. The major limitation of this study is that the control group was older than

Pilates group. Thus, the adaptation of older adults to training responses may be compromised. Also, despite surface EMG is widely used to assess trunk muscles activation, the EMG signal may be disturbed by some factors, such as electrode placement and the amount of subcutaneous fat.

5. Conclusion

The present study was conducted as an initial investigation to identify the effect of Pilates Method intervention on trunk and lower limb stabilizer muscles activation pattern during the gait in older female adults. With respect to our results, eight-weeks of Pilates Method intervention increased IO, MU and RF activation at initial stance gait phase. Also, we found earlier IO onset time before heel strike after Pilates intervention. Pilates and control conditions increased trunk (IO and MU) and lower limb (RF, BF and GM) stabilizer muscles activation at final stance gait phase. Thus, we suggest that eight-weeks of Pilates Method intervention may improve trunk stabilization and anticipatory postural adjustments during walking in older female adults.

Acknowledgement

This project was funded by grants from Fundação de Amparo à Pesquisa do Estado de São Paulo (FAPESP), Conselho Nacional de Desenvolvimento Científico e Tecnológico (CNPq), Fundação para o Desenvolvimento da UNESP (FUNDUNESP) and Coordenação de Aperfeiçoamento de Pessoal de Nível Superior (CAPES).

References

1. Schimitz A, Silder A, Heiderscheit B, Mahoney J, Thelen DG. Differences in lower-extremity muscular activation during walking between healthy and young adults. J Electromyogr Kinesiol. 2009; 19: 1085-1091.

2. Van Dieen JH, Pijnappels M. Falls in older people. J Electromyogr Kinesiol. 2008;18:169-71.

3. Wert DM, Brach J, Perera S, Van Swearinger JM. Gait biomechanics, spatial and temporal characteristics, and the energy cost of walking in older adults with impaired mobility. Phys Ther. 2010;90(7):977-85.

5. Carvalho R, Vasconcelos O, Gonçalves P, Conceição F, Vilas-Boas JP. The effects of physical activity in the anticipatory postural adjustments in elderly people. Motor Control. 2010;14:371-79.

6. Hallal CZ, Marques NR, Gonçalves M. Avaliação do risco de quedas de idosas ativas submetidas a treinamento de equilíbrio com haste vibratória. Ter Man. 2010;8:192-97.

7. Siqueira-Rodrigues BG, Samaria AC, Torres NVOB, Oliveira EM, Dantas EHM. Pilates method in personal autonomy, static balance and quality of life of elderly females. J Bodyw Mov Ther. 2010;14:195-202.

8. Marques NR, Morcelli MH, Hallal CZ, Gonçalves M. EMG activity of trunk stabilizer muscles during Centering Principle of Pilates Method. J Bodw Mov Ther. http://www.sciencedirect.com/science/article/pii/S1360859212001635

9. Bird ML, Hill KD, Fell JD. A randomized controlled study investigating static and dynamic balance in older adults after training with Pilates. Arch Phys Med Rehab. 2012;93:43-47.

10. Hermens JH, Freriks B, Disselhorst-Klug C, Rau G. Development of

recommendations for SEMG sensors and sensor placement procedures. J Electromyog Kinesiol. 2000;10:361-74.

11. Moreside JM, Vera-Garcia FJ, McGill SM. Trunk muscle activation patterns, lumbar compressive forces, an spine stability when using the Bodyblade. Phys Ther. 2007;87:153-64.

12. Gonçalves M, Marques NR, Hallal CZ, Van Dieen JH. Electromyographic activity of trunk muscles during exercises with flexible and non-flexible poles. J Back Musculoskel Rehab. 2012;24:209-214.

13. Kirkwood RN, Trede RG, Moreira BS, Kirkwood SA, Pereira LSM. Decreased gastrocnemius temporal muscle activation during gait in elderly women with history of recurrent falls. Gait Posture. 2011;34:60-64.

14. Panjabi MM. The stabilizing system of the spine. Part 1. function, dysfunction, adaptation, and enhancement. J Spinal Dis. 1992;5(4):383-89.

15. Anders C, Wagner H, Puta C, Grassme R, Petrovitch A, Scholle HC. Trunk muscle activation patterns during walking in different speeds. J Electromyogr Kinesiol. 2007;17:245-252.

16. Lee PJ, Rogers EL, Granata KP. Active trunk stiffness increases with cocontraction. J Electromyogr Kinesiol. 2006;16:51-57.

17. Anders C, Wagner H, Puta C, Grassme R, Scholle HC. Healthy humans use sex-specific co-ordination patterns of trunk muscles during gait. Eur J Appl Physiol. 2009;105:585-94.

18. Tsao H, Hodges PW. Persistence of improvements in postural strategies following motor control training in people with recurrent low back pain. J Electromyogr Kinesiol. 2008;18:559-567.

19. Tsao H, Hodges PW. Specific motor control training improves postural control of

trunk muscles in people with recurrent low back pain. In: 14th Biennial conference of musculoskeletal physiotherapy, Brisbane, Australia; 2005.

20. Vaughan CL, Davis BL, O’Connor JC. Dynamics of human gait. 2nd ed. Cape Town: Kiboho Publishers; 1998.

21. Ikezoe, T., Mori, N., Nakamura, M., Ichihashi, N., 2011. Atrophy of the lower limb in elderly women: is it related to walking ability? Eur J Appl. 111, 989–995.

22. Ploutz-Snyder LL, Manini T, Ploutz-Snyder RJ, Wolf DA. Functionally relevant thresholds of quadriceps femoris strength. J Ger. 2002;57:B144-B152.

23. Marques NR, Hallal CZ, Crozara LF, Morcelli MH, Karuka AH, Navega MT, Gonçalves M. Lower limb strength is associated with gait biomechanical abnormalities in older female fallers and non-fallers. Isok Ex Sci J.

24. Liu, M.Q., Anderson, F.C., Pandy, M.C., Delp, S.L., 2006. Muscles that support the body also modulate forward progression during walking. J. Biomech. 39, 2623–2630.

REFERÊNCIAS

ANDERS, C.; WAGNER, H.; PUTA, C.; GRASSME, R.; PETROVITCH, A.; SCHOLLE, H.C. Trunk muscle activation patterns during walking in different speeds. **Journal of Electromyography and Kinesiology**, v. 17, p. 245-252, 2007.

BAIRD, J.L.; RICHARD, E.A. Young and older adults use different strategies to perform a standing turning task. **Clinical Biomechanics**, v. 24, p. 826-832, 2009.

BEAN, J.F.; KIELY D.K.; HERMAN S.; LEVEILLE S.G.; MIZER K.; FRONTERA, W.R., et al. The relationship between leg power and physical performance in mobility-limited older people**. Journal of American Geriatric Society**, v. 50, n. 3, p. 461-7, 2002.

BURNFIELD, J.M;. JOSEPHSON, K.R.; POWERS, C.M.; RUBENSTEIN, L.Z. The influence of lower extremity joint torque on gait characteristics in elderly men. **Archives of Physical Medicine and Rehabilitation**, v. 81, 2000.

CARVALHO, R.; VASCONCELOS, O.; GONÇALVES, P.; CONCEIÇÃO, F.; VILAS-BOAS J.P. The effects of physical activity in the anticipatory postural adjustments in elderly people. **Motor Control**, v. 14, p. 371-379, 2010.

CHAMBERS, A.J.; CHAN, R. Slip-related muscle activation patterns in the stance leg during walking. **Gait & Posture**, v. 25, p. 565-572, 2007.

DEVITA, P.; HORTOBÁGYI, T. Age causes a redistribution of joint torques and powers during gait. **Journal of Applied Physiology**, v. 88, p. 1804-1811, 2000.

GONÇALVES, M.; MARQUES, N.R.; HALLAL, C.Z, van DIEEN, J.H. Electromyographic activity of trunk muscles during exercises with flexible and non-flexible poles. **Journal of Back and Musculoskeletal Rehabilitation**, v. 24, p. 209-214, 2012.

HALLAL, C.Z.; MARQUES, N.R.; GONÇALVES, M. O uso da vibração como método auxiliar no treinamento das capacidades físicas: uma revisão de literatura. **Motriz**, v. 16, p. 527-533, 2010a.

HALLAL, C.Z.; MARQUES, N.R.; GONÇALVES, M. Avaliação do risco de quedas de idosas ativas submetidas a treinamento de equilíbrio com haste vibratória. **Terapia Manual**, v. 8, p. 192-197, 2010b.

HALLAL, C.Z.; MARQUES, N.R.; SILVA, S.R.D.; van DIEEN, J.H.; GONÇALVES, M. Electromyographic activity of shoulder muscles during exercises performed with oscillatory and non-oscillatory poles. **Revista Brasileira de Fisioterapia**, v. 15, p. 89-94, 2011a.

HALLAL, C.Z.; MARQUES, N.R.; GONÇALVES, M. Razão eletromiográfica dos músculos estabilizadores do ombro durante exercícios com haste oscilatória. **Revista Brasileira de Medicina do Esporte**, v. 17, p. 31-35, 2011b.

HAHN, M.E.; CHOU, L.S. Age-related reduction in sagittal plane Center of mass motion during obstacle crossing. **Journal of Biomechanics**, v. 37, p. 837-844, 2004.

HARTMAN, A.; KNOLLS, R.; MURER, K. et al. Reproducibility of an isokinetic strength-testing protocol of the knee and ankle in older adults. **Journal of Gerontology Society of America**, v. 55, p. 259-268, 2009.

HORTOBÁGYI, T.; FINCH, A.; SOLNIK, S. et al. Association between muscle activation and metabolic cost of walking in young and old adults. **Journal of Gerontology Society of America**, v. 66A, p. 541-547, 2011.

LAROCHE, D.P.; CREMIN, K.A.; GREENLEAF, B.; CROCE, R.V. Rapid torque development in older female fallers and nonfallers: A comparison across lower-extremity muscles. **Journal of Electromyography and Kinesiology**, v. 20, p. 482-488, 2010.

MARQUES, N.R.; MORCELLI, M.H.; HALLAL, N.R.; GONÇALVES, M. EMG activity of trunk stabilizer muscles during Centering Principle of Pilates Method. **Journal of Bodywork and Movements Therapies**, *impress*, 2012a.

MARQUES, N.R.; HALLAL, N.R.; GONÇALVES, M. Padrão de co-ativação dos músculos do tronco durante exercícios com haste oscilatória. **Motriz**, impress, 2012b.

PIJNAPPELS, M.; VAN DER BURG, P.J.; REEVES, N.D.; VAN DIEEN, J.H. Identification of elderly fallers by muscle strength measures. **European Journal of Applied Physiology**, v. 10, n. 5, p. 585-92, 2008.

SIQUEIRA-RODRIGUES, B.G.; SAMARIA, A.C.; TORRES, N.V.O.B.; OLIVEIRA, E.M.; DANTAS, E.H.M. Pilates method in personal autonomy, static balance and quality of life of elderly females. **Journal of Bodywork and Movement Therapies**, v. 14, p. 195-202, 2010.

SCHULZ, B.W; LlOYD, J.D; WILLIAM E. The effects of everyday concurrent task on overground minimum toe clearance and gait parameters. **Gait and Posture**, v. 32, n. 1, 18-22, 2010.

SKELTON, D.A.; KENNEDY, J.; RUTHERFORD, O.M. Explosive power and asymmetry in leg muscle function in frequent fallers and non-fallers aged over 65. **Age and Ageing**, v. 31, n. 2, p. 119-25, 2002.

TALBOT, L.A., MUSIOL, R.J., WITHAM, E.K.; METTER, E.J. Falls in young, middle-aged and older community dwelling adults: perceived cause, environmental factors and injury. **BCM Public Health**, v. 5, p. 86-90, 2005.

VAN DIEEN, J.H.; PIJNAPPELS, M. Falls in older people. **Journal of Electromyography and Kinesiology,** v. 18, p. 169-171, 2008.

WERT, D.M.; BRACH, J.; PERERA, S.; VAN SWEARINGER, J.M. Gait biomechanics, spatial and temporal characteristics, and the energy cost of walking in older adults with impaired mobility. **Physical Therapy,** v. 90, n. 7, p. 977-985, 2010.

YEN, H.C.; CHEN, H.L.; LIU, M.W.; LIU, H.C.; LU, T.W. Age effects on the inter-joint coordination during obstacle-crossing. **Journal of Biomechanics**, v. 42, p. 2501-2506, 2009.