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Motor readiness and joint torque production in lower limbs of older women fallers and non-fallers

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This study aimed to evaluate the motor response time and ability to develop joint torque at the knee and ankle in older women with and without a history of falls, in addition to investigating the effect of aging on these capacities. We assessed 18 young females, 21 older female fallers and 22 older female non-fallers. The peak torque, rate of torque development, rate of electromyography (EMG) rise, reaction time, premotor time and motor time were obtained through a dynamometric assessment and simultaneous electromyography. Surface EMGs of the rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), gastrocnemius lateralis (GL) and tibialis anterior (TA) muscles were recorded. Knee extension and flexion peak torques were lower in older fallers than in non-fallers. Knee extension and flexion and ankle plantarflexion and dorsiflexion peak torques were lower in both older groups than in the younger group. The rate of EMG rise of the BF and the motor time of the TA were lower and higher, respectively, in older fallers than in the younger adults. The time to reach peak torque in knee extension/flexion and ankle plantarflexion/dorsiflexion and the motor times of the RF, VL, BF and GL were higher in both older groups than in the younger groups. The motor time of the TA during ankle dorsiflexion and the knee extension peak torque were the major predictors of falls in older women, accounting for approximately 28% of the number of falls. Thus, these results further reveal the biomechanical parameters that affect the risk of falls and provide initial findings to support the prescription of exercises in fall prevention programs.

1. Introduction

Age-related changes in vestibular, visual, proprioceptive and neuromuscular systems cause impaired balance and weakness, increasing the risk of falls in older adults (Stelmach et al., 1990; Tang and Woollacott, 1998; Kerrigan et al., 1998; Shaffer and Harrison, 2007). Falls are considered a major health problem, affecting 30–60% of the elderly, and 50% of older adults who experience a fall will have multiple falls each year (Perracini and Ramos, 2002; Reyes-Ortiz et al., 2005). In addition, 10% of falls in older adults result in serious injuries, such as hipbone fracture and head concussion, and sometimes cause death (CDC, 2008). Thus, an investigation into the cause of falls in older adults may enhance the clinical approach to the prevention of falls in the elderly.

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Lower limb weakness is a common age-related condition, and it is reported as a main cause of falling in older adults (AGS, 2001; Hughes et al., 2001; Aagaard et al., 2010). The strength loss associated with aging was previously attributed to the progressive loss of muscle fibers in size and number (Vandervoort, 2002; Edstrom et al., 2007; Aagaard et al., 2010); however, recent studies have shown that the loss of muscle mass only explains 6–10% of the strength loss observed in older adults (Clark and Fielding, 2012; Manini and Clark, 2012). The remainder of age-related weakness can be attributed to an impaired intrinsic force generation capacity and abnormalities in muscle fiber contractile properties, metabolic properties, excitation–contraction coupling and muscle activation patterns (Clark and Fielding, 2012; Manini and Clark, 2012). These neuromuscular changes reduce the ability of older adults to generate muscle forces quickly, which is particularly important in the recovery of balance after tripping (Skelton et al., 1994; Pijnappels et al., 2005b; Thelen et al., 1996; Robinovitch et al., 2002). Thus, various biomechanical parameters, such as peak torque, rate of torque development and motor response time, may be important clinical tools for assessing the risk of falls (Pijnappels et al., 2005b; Thelen et al., 1996; LaRoche et al., 2010; Bento et al., 2010).

Results from previous studies support the clinical application of the rate of torque development, the peak torque and the motor response time. According to Izquierdo et al. (1999), the rate of torque development had a moderate association with balance variables, suggesting that the relationship between balance and strength could linearly decline with age. Pijnappels et al. (2008) found that the isometric push off peak force during a leg press can discriminate between fallers and non-fallers with 86% of sensitivity and 90% of specificity. LaRoche et al. (2010) found that older female fallers had a 19% lower peak torque in lower limb muscles (combination of knee extensor/flexor and ankle plantarflexor/dorsiflexor peak torque) and 29% longer motor time. These authors also suggested that lower limb weakness should be considered a contributor to the risk of falls in older adults.

Although age-related lower limb strength loss has been investigated in the literature, the decline of lower limb muscle recruitment and the biomechanical parameters that may better predict impaired functional performance and increased risk of falling remain uncertain. Thus, this study aimed to compare motor response time, lower-limb strength, and rate of torque development of the knee and ankle in younger female adults, older female fallers and older female non-fallers. The second aim was to evaluate the ability of the women in these three groups to activate lower limb muscles in a short period of time to produce maximal force. We hypothesized that younger female adults would be stronger and have faster muscle activation than older female adults. We also hypothesized that older female fallers would have lower knee strength, ankle strength and rate of torque development, as well as longer motor time responses. Lastly, we hypothesized that older female fallers would have the slowest lower limb muscle activation when producing maximal force.

2. Methods

2.1. Subjects

Forty-three adult females between the ages of 60 and 85 years, recruited from community-based physical activity groups (minimum 1 year of experience), and eighteen adult females between the ages of 18 and 25 years, recruited from a university student population, were considered in this study (Table 1). The older participants were divided into two groups: those that had fallen during the year prior to the evaluation ($n = 21$) and those that had not ($n = 22$). A fall was defined as any balance perturbation that 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.

2.2. Procedure

Data collection was performed over a period of 2 days. On the first day, the subjects' physical characteristics (body mass, stature and age), cognitive statuses [Mini-mental Examination (Brucki et al., 2003)], physical activity levels [International Physical Activity Questionnaire (Benedetti et al., 2004)], fall risks [BERG scale (Miyamoto et al., 2004)], lower limb dominance statuses (Hoffman et al., 1998) and fall histories were collected. Lower limb dominance was determined using three functional tests (ball kick, stair climbing, and balance recovery); three trials of each test were conducted. The leg used in most of the individual tests was categorized as the dominant leg. All of the young participants were considered physically active because they were physical education undergraduate students who practiced physical activities regularly. On the second day of data collection, the joint torque evaluation and electromyography (EMG) were performed.

Table 1
Subject characteristics.

	Young	Non-fallers	Fallers	p[a]
Age (years)	21.79 ± 2.12	66.14 ± 6.1	69.62 ± 7.16	0.06
Body mass (kg)	60.72 ± 7.92	65.04 ± 12.89	65.91 ± 10.03	0.512
Stature (m)	1.62 ± 0.06	1.55 ± 0.06	1.52 ± 0.05	0.119
BMI (kg m^{-2})	23.29 ± 3.23	27.13 ± 4.78	28.53 ± 3.95	0.301

BMI: Body mass index.
[a] Comparisons between older faller and non-fallers.

2.3. Measurement of knee and ankle strength

Joint torque was assessed using an isokinetic dynamometer (Biodex®, New York, USA). Before the strength assessment, the subjects performed a warm up, walking 5 min on a treadmill at the preferred walking speed. After the warm up, three maximal isometric voluntary contractions (MIVCs) were performed for 5 s each and with 30 s of rest between each trial. A minimum of 2 min rest was given between each joint assessment to avoid fatigue. The subjects were instructed and strongly encouraged to perform the MIVC as fast and hard as possible in response to a visual stimulus (simple on/off light, synchronized with EMG and torque data). Before data collection, a familiarization period was performed using the same procedure. The sequence of evaluations was randomized and gravitational correction was performed.

Isometric knee flexion and extension torques were measured while the subject was seated with their hip flexed at 90° and their knee flexed at 30° (considering 0° to be full extension). 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 (slightly 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 their chests throughout the test. Isometric plantarflexion and dorsiflexion torques were measured with the subject seated with their hip flexed at 70°, their knee flexed at 45°, and their ankle 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 a foot plate. Proximal thigh and trunk stabilization (using belts) was provided to prevent extraneous movement.

2.4. Measurement of EMG signal

EMG was performed using a Telemyo 900 (Noraxon®, Phoenix, USA) with a common mode rejection ratio >100 dB (60 Hz), total gain of 2000× (20× in the preamplifier and 100× in the equipment), baseline noise <1 μV root mean square, and differential input impedance >10 MΩ. Torque and EMG data were synchronized using an external analog/digital board (NorBNC, Noraxon®, Phoenix, USA), sampled at 2000 Hz, and stored on the computer's hard drive using data acquisition software (Noraxon®, Phoenix, USA). Bipolar disk silver/silver chloride surface electrodes (Miotec®, Porto Alegre, Brazil) with a diameter of 1 cm and inter-electrode distance of 2 cm were positioned on the participant's dominant side, over the belly of the rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA) and gastrocnemius lateralis (GL) muscles, longitudinally parallel to the underlying muscle fiber arrangement in accordance with the approach described by Hermens et al. (2000). A reference electrode was placed on the lateral malleolus. Before placing the electrodes, the subject's skin was