ASSOCIATION BETWEEN STRENGTH, KINEMATICS, AND THE ENERGY COST OF WALKING IN OLDER, FEMALE FALLERS AND NON-FALLERS

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INTRODUCTION

Walking is the most executed movement during daily activities and is essential to functional independence in older adults. However, aging causes changes in the neuromuscular system that result in abnormal gait during walking. Despite the documented changes in the biomechanical parameters of gait, the causes for an increased oxygen cost of walking (Cw) in older adults remain unclear [1].

Most falls occur during walking, and poor gait performance has been indicated as a major contributing factor to falling. In addition, higher energy expenditure during walking in older adults could cause clinically significant increases in task difficulty and fatigue rate [2]. Consequently, our study aimed to determine the contributions of gait biomechanics, muscle activation, and hip, knee and ankle strength to Cw in older women with and without a history of falls.

METHODS

Subjects

Data of thirty-seven older women were considered for this study. Volunteers were separated into two groups according to their report of having fallen or not fallen over the one year period before the study. This resulted in 15 volunteers in the faller group and 22 volunteers in the non-faller group. The subjects had an average age of 67.5 ± 7.1 yr, body mass of 65.9 ± 11.9 kg, body mass index of 28 ± 4.4 kg m⁻², and preferred treadmill walking speed of 0.9 ± 0.2 m s⁻¹, which were not different between groups (all p > 0.1).

Procedures

Data collection was performed on two separate days. On the first day, hip, knee and ankle maximal, voluntary, joint torques were recorded isokinetically at 120 deg s⁻¹ for both flexion and extension movements. On the second day, volunteers were familiarized with treadmill walking at their preferred speed. Then, indirect calorimetry was used to measure oxygen uptake (VO₂) first in a sitting, resting posture and second, during an 8-minute treadmill walk. Lower-extremity gait kinematic parameters were also recorded during a 1-minute period of the walk using a three-dimensional, optical motion system.

EMG signals were recorded at a sample frequency of 2000 Hz for the following muscles: internal oblique (IO), multifidus (MU), gluteus maximus (GM), biceps femoris (BF), rectus femoris (RF), tibialis anterior (TA) and gastrocnemius lateralis (GL).

Data Analysis

The average VO₂ recorded between minutes 3-6 of the walking trial was used to calculate Cw. The resting, body mass-normalized VO₂ was subtracted from the walking, body mass-normalized VO₂ and this difference was divided by the gait speed chosen by each volunteer.

For strength measures, the peak torque was normalized by the mass of each volunteer. The EMG signal was processed using full-wave rectification and a low pass filter (4th order and cut-off of 10 Hz) at 100 ms before and after heel contact and toe-off of ten consecutive strides. Then, the linear envelope values were normalized by the mean of each muscle activation.

Stride length, stride time, ankle angle at heel contact and hip angle at toe-off were analyzed using motion data and were averaged over ten consecutive strides. These measures were chosen as they have been previously shown to be related to Cw.

The statistical analysis compared strength, gait kinematic variables, and Cw between groups using the student’s t-test for independent samples. Also, Spearman correlation coefficients were computed to quantify the association between the Cw and each
gait, muscle activation, and torque measure. The significance level was set at \( p < 0.05 \) for all tests.

**RESULTS AND DISCUSSION**

When comparing 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 \( (p = 0.01; \text{Table 1}) \).

For non-fallers, only age was associated with Cw \( (r = 0.6, \ p = 0.01) \). However, for fallers, we found that a higher activation of GM during initial stance was associated with a higher Cw. For example, Cw was correlated to GM activation before heel contact \( (r = 0.5, \ p = 0.03) \) and GM activation after heel contact \( (r = 0.7, \ p = 0.01) \).

The most novel finding of this study is that in female, older fallers, impaired knee extensor strength may cause a greater reliance on the hip extensors during walking. This compensation was demonstrated by a high hip angle at toe-off and high levels of hip extensor activation at initial stance.

During initial stance, hip stabilizer muscles such as GM and gluteus medius are recruited to reduce the displacement of the center of mass in the sagittal plane [3]. We could therefore suggest that in older, female fallers the decreased knee extensor strength may reduce the capacity of quadriceps to perform negative work and stabilize the knee. In turn, a reduced capacity to absorb work at the knee may result in diminished return of energy, more positive work, and an increased Cw.

Reduced knee extensor strength could shift the burden of body support to the hip extensors, challenge hip joint stability, and increase hip stabilizer muscle activation. According to Hortobágyi et al. (2011) high levels of muscle coactivation, which is a compensatory adaptation to maintain stability, is also related to an increased Cw.

**CONCLUSIONS**

In older, female non-fallers age was the only factor that related to the Cw, while in older fallers, kinematic and muscle activation parameters were related to the Cw. We speculate that in the fallers observed in this study that poor knee extensor strength could cause compensatory adaptations at the hip, such as high levels of GM activation at initial stance. Thus, abnormal gait patterns could increase the Cw which could contribute to the onset of fatigue and increased fall risk in older people.

**REFERENCES**


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<table>
<thead>
<tr>
<th>Variable</th>
<th>Faller Group (n=15)</th>
<th>Non-Faller Group (n=22)</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cost of walking (ml·kg(^{-1})·min(^{-1})·m(^{-1})·s(^{-1}))</td>
<td>9.3 (4.3)</td>
<td>8.7 (2.9)</td>
<td>0.6</td>
</tr>
<tr>
<td>Knee flexor torque (N·m·kg(^{-1}))</td>
<td>0.54 (0.26)</td>
<td>0.55 (0.14)</td>
<td>0.8</td>
</tr>
<tr>
<td>Knee extensor torque (N·m·kg(^{-1}))</td>
<td>0.69 (0.2)</td>
<td>0.96 (0.24)</td>
<td>0.02*</td>
</tr>
<tr>
<td>Hip flexor torque (N·m·kg(^{-1}))</td>
<td>0.68 (0.26)</td>
<td>0.74 (0.19)</td>
<td>0.4</td>
</tr>
<tr>
<td>Hip extensor torque (N·m·kg(^{-1}))</td>
<td>0.78 (0.23)</td>
<td>0.89 (0.36)</td>
<td>0.2</td>
</tr>
<tr>
<td>Ankle plantarflexor torque (N·m·kg(^{-1}))</td>
<td>0.31 (0.19)</td>
<td>0.33 (0.13)</td>
<td>0.7</td>
</tr>
<tr>
<td>Ankle dorsiflexor torque (N·m·kg(^{-1}))</td>
<td>0.28 (0.11)</td>
<td>0.31 (0.13)</td>
<td>0.3</td>
</tr>
<tr>
<td>Stride time (s)</td>
<td>2.3 (0.89)</td>
<td>2.6 (0.9)</td>
<td>0.3</td>
</tr>
<tr>
<td>Stride length (mm)</td>
<td>509.3 (61.9)</td>
<td>497.2 (70.7)</td>
<td>0.6</td>
</tr>
<tr>
<td>Ankle angle at heel contact (deg)</td>
<td>6.4 (4.3)</td>
<td>5.9 (4.4)</td>
<td>0.2</td>
</tr>
<tr>
<td>Hip angle at toe-off (deg)</td>
<td>9.5 (4.8)</td>
<td>5.4 (4.7)</td>
<td>0.01*</td>
</tr>
</tbody>
</table>

* Significant difference \( (p < 0.05) \) between faller and non-faller groups. Values are mean (SD).