EFFECTS OF FATIGUE INDUCED BY PROLONGED GAIT WHEN WALKING ON THE ELDERLY

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ABSTRACT

Purpose. Fatigue has been pointed as a fall risk in the elderly; however, the effects of prolonged gait on neuromuscular recruitment and on its pattern remain unknown. The aim of this study was to evaluate the effects of prolonged gait on neuromuscular recruitment levels and spatial-temporal gait variables. **Methods.** Eight healthy older women (age: 72.63 ± 6.55 years) walked at their preferred walking speed for twenty minutes on a treadmill. The Root Mean Square (RMS) from the vastus-lateralis, femoral biceps, tibialis anterior and lateral gastrocnemius muscles were determined at the first and last minute of the test during the moments of Heel Strike (HS), Terminal Stance and Terminal Swing (TS). In addition, coactivation in the knee and ankle as well as the stride cadence and length were measured in the test. The two RMS data (taken at the first and last minute) were compared by means of a Student's t-test. **Results.** Twenty minutes of walking induced fatigue in the subjects, as observed through an increase in RMS, notably during the HS and TS. Coactivation was also influenced by the prolonged gait test. The only gait phase where a risk of falling was enhanced was the HS. Nonetheless, subjects developed strategies to maintain a safe motor pattern, which was evidenced by an increase in stride length and a decrease in stride cadence. **Conclusion.** Tests lasting just twenty minutes on a treadmill were enough to induce fatigue in older adults. However, the level of fatigue was not enough to present a danger or fall risk to elderly individuals.

Key words: ageing, fatigue, walking, electromyography, kinematic

Introduction

Although walking is a human movement pattern that expends the least amount of energy to be performed [1], many aspects play important roles in its performance, such as age and fatigue [2, 3]. Aging is directly related to specific gait adaptations, such as step length decrease [4] and higher mediolateral trunk acceleration [2]. However, these adaptations give rise to greater instability [2, 5].

The instability found in those who are elderly is attributed to the loss of force [6], which is most evident when an individual's muscles are fatigued [7]. The decrease in force in this population segment is due to a reduction in the number of motor units and in neural conduction velocity [8]. The influence of such neuromuscular adaptations can be measured by surface electromyography (EMG), which also allows the evaluation of neuromuscular fatigue. Fatigue can be measured by the amplitude of the EMG parameters. Its effects can be observed through an increase in the Root Mean Square (RMS), that is brought on by the recruitment of new motor units or by a higher activation level of those already activated [9, 10].

EMG can also be used to evaluate coactivation, which is calculated by the agonist and antagonistic ratio. Coactivation is higher in the elderly when compared to younger individuals [11] and it decreases the capacity to produce force during toe-off [5] as well as also increasing heel strike velocity during loading [12]. There is no standard value for this variable; however, increased values represent higher energy consumption [3] and a decreased capacity in controlling movement [12]. In addition, lower coactivation scores have been found to represent a reduction in joint stability [13, 14].

The influence of fatigue on both elderly (an encompassing study was done by Kent-Braun [15]) and younger individuals [16, 17] has been already studied, however it was induced by means of isometric contractions [17] or other movements that are not performed in daily life, such as the seat-to-stand test [2]. Therefore, the effect of fatigue on the elderly induced by functional activities, such as prolonged gait, may better represent the routine movements and adjustments made by this population. In addition, another feature that is still not fully understood is the influence of fatigue on coactivation; recent studies have found that during isometric contractions the co-contraction magnitude is not influenced by fatigue [18]. However, this interaction in the elderly is not clear, especially during gait.

A period of about twenty minutes is recommended

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to assess the performance of many body systems of the elderly [19–21], since it represents a sufficient period for registering change without cardiorespiratory risks [20]. As such, the main objective of this study was to evaluate the effect of fatigue on the elderly, induced by gait for a period of twenty minutes. Specifically, the aim of the study was to investigate the effects of fatigue on lower limb neuromuscular recruitment levels, coactivation, and stride spatio-temporal variables. As hypothesized, it was believed that fatigue would cause higher RMS values, would not influence coactivation, and would lead to increased stride cadence and decreased stride length.

Material and methods

Of a total of 82 invited subjects, 50 agreed to participate in the study. From this subtotal, 19 were not eligible due to the inclusion criteria and another 19 were not considered physically active. Four of the remaining 12 subjects were unable to adapt to walking on the treadmill. Therefore, total participation in this study composed of eight female subjects aged 72.63 ± 6.55 years weighing 63.43 ± 7.98 kg and 1.53 ± 0.05 m in height. All of the volunteers were considered physically inactive (with a score of 6.78 ± 4.50 in a physical activity questionnaire proposed by Voorrips et al. [22], the Modified Baecke Questionnaire for Elderly People). The exclusion criteria were any orthopedic, neurological, visual, vestibular or cardiovascular conditions that would not allow the subject to perform all of the proposed activities. The subjects visited the laboratory on two different occasions, with a minimum of 24 h and maximum of 72 h between the two visits. All of volunteers signed a consent form, approved by the Local Ethics Committee.

During the first visit, the volunteers were familiarized with a Millennium Super ATL motorized treadmill (Ibramed, Brazil), where velocities varying between 1 km/h and 4 km/h were used (with a 0.5 km/h step increase) for 15 min of activity [23].

During the second day of testing, the preferred walking velocity was determined as proposed by Malatesta et al. [24]. The volunteers began to walk at the lowest velocity (1 km/h) which was then progressively increased (0.1 km/h/second) until the subject found a comfortable walking speed. After a 5 min rest period, the same procedure was used, but with an initial walking speed of 4 km/h with progressive speed reduction. Finally, the volunteers' preferred walking speed was determined as the mean of the two tests.

After at least 10 min of rest, the volunteers walked at their preferred walking speed for 20 min or until they voluntarily stopped. During all gait tests, the volunteers used their preferred walking shoes. As a safety procedure, during the entire test the volunteers wore a pelvic belt attached to a special rope that was fixed to the ceiling, with a support capacity of 15 kN. Data was acquired after 3 min of walking in order to take into account for any possible adaptation.

Kinemetry

Volunteers were filmed with a Panasonic digital camcorder (60 Hz frame rate) positioned perpendicularly to their right sagittal plane. MyoResearch XP software (Noraxon, USA) was used for image acquisition and to simultaneously record EMG data. To determine the gait phases (heel strike), a 2 cm reflexive marker was positioned over the volunteers' right calcaneus, which was then tracked by Peak Motus software (Vicon, USA). The raw digital kinemetric data were filtered by a forth-order Butterworth filter, with a 6 Hz cut-off frequency. The kinematic variables that were determined were stride cadence (strides/min) and stride length (cm).

Electromyography

Passive Ag/AgCl electrodes (MediTrace, USA), with a 1 cm diameter, were used with the subject's skin prepared (abrasion and cleaning) according to SENIAM. The electrodes were positioned over the vastus-lateralis (VL), femoral biceps (BF), tibiailis anterior (TA) and lateral gastrocnemius (GL). EMG data were acquired using a Telemyo 900 telemetric surface EMG system (Noraxon, USA) with 2000x gain and a 1000 Hz sample rate. Raw data were filtered off-line with a 20–500 Hz band-pass filter.

To identify the Root Mean Square (RMS) of specific gait phases, the entire cycle period (from one heel strike to the next) was normalized at 100% and the raw signal was averaged over 2% periods. Thereafter, for each muscle, the RMS was determined at three different gait phases i) Loading Response (0–10% of the gait cycle); ii) Terminal Stance (30–60%) and iii) Terminal Swing (87–100%) [25], according to the following formula.

$$x_{\mathrm{rms}} = \sqrt{\frac{1}{T_2 - T_1} \int_{T_1}^{T_2} [f(t)]^2 \, dt}$$

In order to calculate a normalized RMS, all data were normalized by a standard value: the mean RMS of the last five strides (recorded during the entire cycle) at the first minute (MIN1) when data were collected. A schematic of this calculation is shown in Figure 1.

The Coactivation Index (CCI) was determined for the same gait phases as mentioned above and was calculated as follows [26, 27]:

$$\mathrm{CCI} = \frac{2 \times \mathrm{Ant}}{\mathrm{AtTot}} \times 100$$
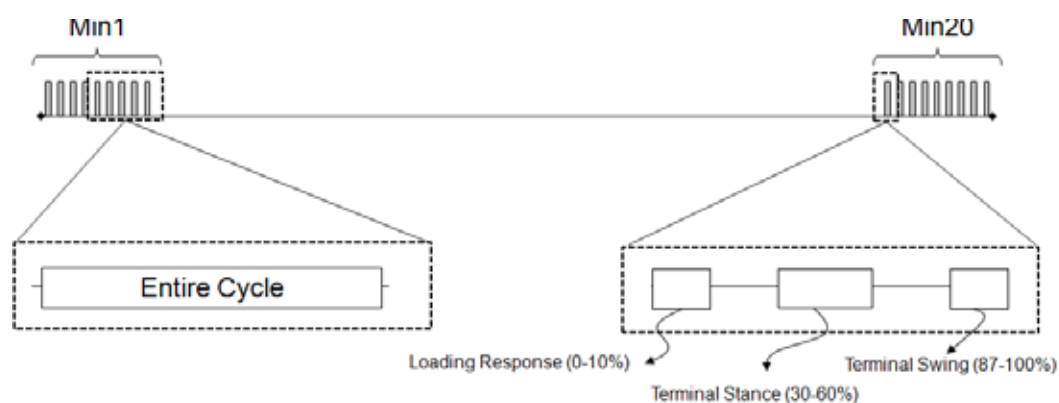
243

Figure 1. A schematic showing EMG acquisition and processing. Each gray bar represents one step. During the first minute, or MIN1, the RMS value (for each muscle) was determined for the entire stride cycle and during the last five strides (left corner). This value was used for normalizing the RMS values. In addition, three RMS values were determined during the last ten strides at the first (MIN1) and twentieth minute (MIN20): the Loading Response, Terminal Stance and Terminal Swing (right corner). Thereafter, a mean RMS value for each step phase from these last ten steps was determined and normalized as mentioned previously.

where 2 x Ant is double of the antagonistic RMS and *AtTot* is the total muscular activation at a joint (the antagonistic plus agonist RMS values). Since one muscle can act as agonist in a specific gait phase and as antagonistic in another, the muscle that presented the higher RMS value during each analyzed period, when compared to its pair (BF vs. VL; TA vs. GL), was considered as the agonist one. The CCI was determined between VL and BF, and TA and GL.

All of the EMG variables were calculated using specific algorithms in MatLab software (Mathworks, USA). All of the variables were determined from an average of the last ten strides of the first minute (MIN1) and last minute (MIN20) of data collection.

To calculate the activation pattern, EMG data were also filtered with a fourth-order Buterworth filter with a 10 Hz cut-off frequency. The last ten strides of the two test samples, at MIN1 and MIN20, were then averaged and a new average was found for all the subjects.

Statistical analysis

To ensure normal data distribution a Shaprio-Wilk test was used, where the influence of each test stage (MIN1 vs. MIN20) was analyzed by a paired Student's T-test. A *p* value of 0.05 was considered statistically significant. All data are expressed as mean ± standard deviation.

Results

The mean preferred walking speed was found to be 2.64 ± 0.32 km/h. Just one subject was unable to fulfill the entire 20 min test, voluntarily stopping at the twelfth minute of the test. Therefore, for this volunteer, the variables recorded at MIN1 were compared to those recorded at MIN12.

Table 1 presents the RMS and CCI results. During Loading, all muscles with the exception of the GL were higher at MIN20 with a reduction of coactivation in the ankle. During Terminal Stance, TA presented lower RMS values at MIN20 but increased RMS values for VL and GL. During Terminal-Swing, all RMS values of all the muscles as well as the CCI at the knee were

Table 1. The RMS values (% Mean) of the vastus-lateralis (VL), biceps femoris (BF), tibialis anterior (TA) and gastrocnemius lateralis (GL) obtained during MIN1 and MIN20 of the gait test during three different gait phases

	MIN1	MIN20	*p* – value
Loading			
VL	225.45 (58.94)	252.84 (59.29)	**0.002**
BF	171.66 (88.09)	189.72 (65.90)	**0.006**
TA	131.91 (63.13)	169.90 (63.38)	**< 0.001**
GL	112.797 (55.93)	99.13 (62.00)	0.780
CCI Knee	73.22 (13.16)	76.42 (13.63)	0.076
CCI Ankle	72.06 (19.99)	63.03 (22.14)	**0.004**
Terminal-Stance			
VL	38.84 (21.86)	51.01 (27.91)	**< 0.001**
BF	52.28 (34.63)	58.37 (44.33)	0.240
TA	76.17 (33.81)	60.92 (35.67)	**< 0.001**
GL	123.21 (40.32)	146.51 (38.55)	**< 0.001**
CCI Knee	77.01 (12.41)	79.72 (14.02)	0.195
CCI Ankle	66.17 (18.21)	53.76 (19.86)	**< 0.001**
Terminal-Swing			
VL	219.98 (67.23)	142.22 (70.94)	**< 0.001**
BF	197.34 (61.54)	195.08 (82.94)	**< 0.001**
TA	161.66 (48.19)	123.60 (48.45)	**< 0.001**
GL	73.69 (49.97)	48.57 (32.23)	**< 0.001**
CCI Knee	84.60 (11.22)	69.62 (17.14)	**< 0.001**
CCI Ankle	57.07 (24.80)	56.35 (25.51)	0.735

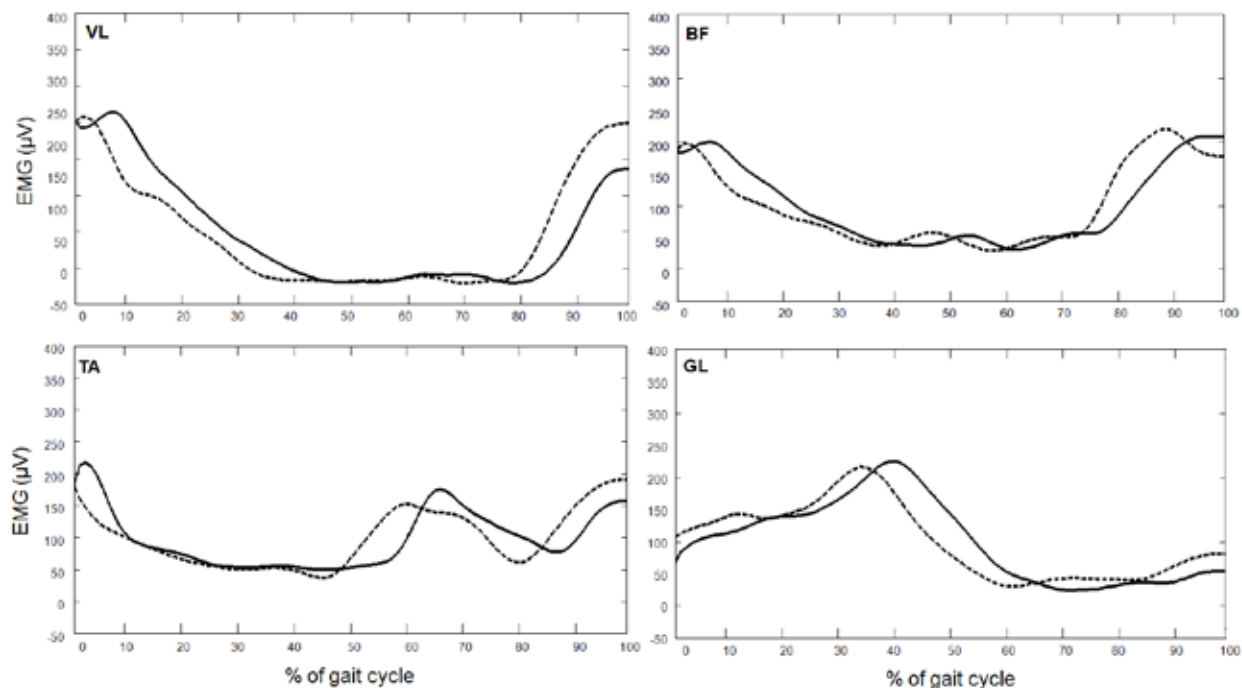The values in bold denote statistical significant differences (*p* < 0.05)

Figure 2. Mean (n = 80 for each minute: 8 volunteers, ten strides) rectified EMG activity (µV) of VL, BF, TA and GL during the entire gait cycle at MIN1 (dot lines) and MIN20 (full lines)

influenced by the gait test, with RMS decrease noted for all muscles.

Figure 2 shows the rectified neuromuscular activation of VL, BF, TA and GL. Besides the magnitude differences already shown in Table 1, a delay in the activation peaks during MIN20 can be observed when comparing it to MIN1 for all muscles. For example, the first peak of VL, during heel strike, occurs at around 5% of the gait cycle at MIN1, while at MIN20 it occurs at around 8% of the gait cycle. Also, for GL, the EMG peak at MIN1 occurs at around 35% of the gait cycle and around 40% at MIN20. However, this analysis is only descriptive, since no statistical evaluations were conducted on this data.

AT MIN20, higher stride length was found when compared to MIN1 (MIN1: 84.42 ± 12.23 cm; MIN20: 87.12 ± 12.35 cm; $p < 0.001$), while stride cadence was lower at MIN20 (MIN1: 53.00 ± 5.23 strides/min; MIN20: 51.50 ± 6.50 strides/min; $p = 0.003$).

Discussion

The main objective of this study was to investigate the influence of fatigue induced by a functional activity, such as prolonged gait, on neuromuscular recruitment levels and on coactivation levels in the knees and ankles of elderly subjects. What was found was a general increase in muscular activity after fatigue during the landing gait phase with a reduction during the swing phase. Also, a reduction in the coactivation levels was observed in the ankle during the landing phase.

The study's subjects (healthy, elderly adults) usually walked on the treadmill with a preferred walking speed of around 4.64 km/h [28]. However, the mean velocity recorded was 2.64 ± 0.32 km/h. This could be attributed to the physical fitness level of our subjects, since they received overall low scores on the Modified Baecke Questionnaire for Elderly People.

It could be surmised that such a test would not be able to induce fatigue, since twenty minutes of treadmill walking at the preferred gait speed is not an exhaustive activity. However, since the subjects had a low physical fitness level, it was expected that this activity would be enough to induce fatigue. This was confirmed by the increased RMS values for almost all muscles, indicating the presence of neuromuscular fatigue [29]. The increase in the RMS values indicates that new motor units are recruited to maintain muscular tension or a synchronous activation of those motor units already activated [10].

Fatigue also promotes conduction velocity reduction, mainly by the recruitment of type I fibers and by the accumulation of inorganic phosphate and H^+ ions [9]. This could explain the EMG peak delays observed in Figure 2. This reduction of conduction velocity could prevent muscles from contracting rapidly, and explain the delay at the observed peaks. However, this analysis was only descriptive and the induction of fatigue can only be properly discussed through RMS data.

During Terminal Swing, the BF muscle presented lower RMS values at MIN20 than at MIN1; therefore a higher heel strike velocity was expected [12], since

this muscle is responsible for decelerating the knee extension during this gait phase [12]. This study did not investigate ground reaction forces or heel strike velocity; however, this could explain why the VL and TA RMS values increase during Loading. At this gait phase, with higher velocities of heel strike, these muscles should exert a higher workload to counteract the increased knee flexion and ankle plantarflexion momentum [25]. Since BF and VL presented the same tendency (RMS increase) on Loading, no differences in coactivation were observed in this joint.

However, a coactivation increase was found in the ankle. This can indicate a higher fall risk at MIN20. This coactivation increase can be due to the maintenance of the GL's activation level, contrary to what was observed at the TA. During Loading, the ankle should act as rigid block to support and transfer ground reaction forces to the entire lower limb. Therefore, since lower levels of coactivation implies stiffness in the lower joint [13, 14], we can suggest that at MIN20 these subjects have a less stable joint in comparison to MIN1, indicating a higher fall risk. Hence, even with an increase of VL activation during Loading (maintaining joint preparation to counteract ground reaction forces), fatigue can become dangerous, since the coactivation in the ankle results in a reduction of joint stiffness.

Nonetheless, during the Final-Stance, the subjects were able to adapt enough to maintain a safe gait. Both the TA and GL worked as expected to guarantee enough impulsion: as the GL activation level increased (maintaining muscular tension to counteract fatigue), the subjects reduced TA activity, which reduced coactivation levels in the ankle. A sufficient activation of the GL is necessary during this gait phase to ensure sufficient forward body impulsion and to maintain gait performance [5].

Similarly, during Terminal-Swing, the neuromuscular behavior in the knee at the end of the test showed safe motor strategy. At MIN20, subjects had reduced the recruitment level of the BF, which could increase coactivation in the knee and represent an increased fall risk [30]. However, this reduction accounted for only 2%, far less than the 35% reduction of the VL, which resulted in a reduction of the coactivation level in the knee. The reduction of both the TA and GL also represents a strategy to maintain gait performance, since these muscles are not necessary during this gait phase [25]. This behavior also maintained coactivation levels in the ankle.

Hence, it can be affirmed that such a test, composed of walking for at least twenty minutes, can be used in training or evaluating older subjects and is enough to induce lower limb neuromuscular adaptations but not enough to reduce balance. As mentioned previously, neuromuscular adaptations, such as those observed during the Loading phase, did increase the fall risk. However, the subjects, under more demanding conditions or when they felt unsafe, reduced their step length and increased the step cadence and width [4, 5, 31] in order to increase balance. Therefore, the subject assumed a new motor pattern when they are unbalanced or in hazardous conditions [4, 5, 32]. However, since the subjects increased their step length and reduced their step cadence, it can be affirmed that the neuromuscular adaptations observed at the end of the test were not demanding enough to make the subjects feel unsafe.

Some limitations of this study were the absence of kinetic analysis and the small sample size. However, since the focus of this study was to analyze neuromuscular adaptations, information on ground reaction forces would not produce any new information needed in the discussion. However, additional studies with a larger number of subjects should be conducted in the future.

Conclusion

It can be concluded that tests involving prolonged gait, such as those which involve at least twenty minutes of walking on a treadmill, are enough to induce fatigue in older subjects. However, in healthy older adults, generally speaking, the level of fatigue is not enough to create a hazardous condition due to the neuromuscular adaptations that were observed which in turn did not lead to an increased fall risk. However, special attention should be given to the Loading phase, as the muscular adaptations that were observed could induce unstable situations. Nonetheless, we can affirm that such a period of walking is safe for elderly subjects and can be used in physical training or for evaluation purposes.

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