Effect of Gait Speed on Muscle Activity Patterns and Magnitude During Stance

Andreia S. P. Sousa and João Manuel R. S. Tavares

This study aims to assess the influence of gait speed (manipulated through cadence) on muscle activity patterns and activation degree during stance. Methods: Thirty-five healthy individuals participated in this study. Surface electromyographic activity from the gastrocnemius medialis (GM), gluteus maximus (GMax), biceps femoris (BF) and rectus femoris (RF) was acquired with subjects walking at three different speeds. Results: Speed influenced: (1) relative motor activity patterns at heel strike, midstance-propulsion transition and propulsion; (2) the activity level of RF, GMax, GM and BF, in decreasing order, with higher activity at the fastest and slowest speeds. Conclusions: In general, muscle activity was higher at the fastest and slowest speeds than at the self-selected speed and only the activity of the main actions in each subphase remained stable. These findings suggest that gait speeds different from the self-selected speed influence not only activity levels but also relative muscle activity patterns. As a result, caution is advised when choosing standard speeds in gait studies, as this can lead to increased variability in relative muscle activity patterns.

Keywords: electromyography; gait; muscle function; motor control; motor behavior; motion analysis.

It is well known that there are some factors which account for metabolic energy expenditure during gait, such as the need to redirect the center of mass (COM) (Kuo, Donelan & Ruina, 2005), body weight support, leg movement or stability control (Liu, Anderson, Pandy & Delp, 2006; Neptune, Kautz & Zajac, 2001; Zajac, Neptune & Kautz, 2003). This metabolic energy needed to walk is explained by the mechanical power generated by muscles (Woledge, Curtin & Homsher, 1985), whose activity patterns and intensity of activation change with speed (Crowe, Schiereck, de Boer & Keessen, 1993, 1995; den Otter, Geurts, Mulder & Duysens, 2004; Ivanenko, Poppele, Macellari & Lacquaniti, 2004) to control the accelerating and decelerating forces of individual body segments needed to establish safe forward progression (Yang & Winter, 1985). Muscle activity during locomotion occurs in bursts that change in both amplitude and duration as a function of locomotion...
speed (den Otter et al. 2004; Ivanenko et al. 2002; Nilsson et al. 1985). According to (Cappellini, Ivanenko, Poppele & Lacquaniti, 2006), changes in walking speed are associated mainly with modifications in the intensity of muscle activation and only minor changes are observed in their relative timings during stance (Cappellini, et al., 2006; den Otter, et al., 2004). Also the temporal EMG profiles of functionally related muscles can show considerable similarity at different cadences (Davis & Vaughan, 1993; Yang & Winter, 1985).

The simple model of gait described in (Kuo, 1999) predicts that collision costs increase with step length and frequency, accounting for a significant part of the energy consumed during walking. The amplitude of muscle activity increases with walking speed because of the need for larger muscular force output. This normal positive relationship may be challenged if walking speed is markedly reduced, due to changes in the underlying locomotor task demands. Extreme reductions in walking speed will prolong substantially the time spent in double support, and one may expect a switch from locomotor to merely postural muscular synergies. In addition, the larger horizontal excursions of the COM associated with slow walking may necessitate more explicit muscular efforts to maintain frontal plane balance during walking (Babý & Kuo, 2000). Moreover, at self-selected walking speeds energy expenditure is diminished, as the gait system is adapted to execute walking at the usual speed (Masani, Kousaki & Fukunaga, 2002).

In general, muscles are most active around the time of foot contact (Singh, 1970), as there is a distribution of hip and knee extensor muscle forces in early stance and ankle plantar flexor and rectus femoris forces in late stance to provide support and forward propulsion (Liu, et al., 2006). However, how the muscle contributions change with walking speed is not well understood. Intuitively, walking at faster speeds would require increased activity from muscles contributing to forward propulsion. However, fast walking speeds are associated with longer stride lengths, which may require increased activity from those muscles contributing to vertical support, as there is an increased vertical excursion of the body’s COM (Orendurff et al., 2004). On the other hand, walking at slower speeds may be mechanically less efficient and less conducive to storage and recovery of elastic energy in the musculotendon complex (Neptune, Sasaki & Kautz, 2008).

Most authors studying the influence of speed on gait muscle activity have used treadmills (den Otter, et al., 2004; Masani, et al., 2002; Nilsson, Thorstensson & Halbertsma, 1985; van Hedel, Tomatis & Muller, 2006), but there are some differences between treadmill and overground walking. During treadmill walking, increases in the hip range of motion, maximum hip flexion, joint angle and cadence occur, while the stance time decreases (Alton, Baldey, Caplan & Morrissey, 1998). Knee kinematics appear to be comparable (Matsas, Taylor & McBurney, 2000), while vertical ground reaction forces show similar patterns, but differ slightly in force magnitude during mid and late stance (White, Yack, Tucker & Lin, 1998). Leg muscle EMG activity is somewhat smaller in overground versus treadmill walking (Arsenault, Winter & Marteniuk, 1986; Murray, Spurr, Sepic, Gardner & Mollinger, 1985). The effect of speed on overground walking has already been studied, but only addressing the muscle recruitment timing with the individuals walking at the same standard speed (Murray, et al., 1984). However, as already mentioned, the gait system is adapted to walking at different speeds, and therefore it is important to study the influence of different gait speeds
on muscle recruitment patterns compared with the self-selected speed adopted by each subject.

Therefore, the purpose of this study is to examine how muscle activation depends on locomotion speed during overground walking. Specifically, this study aims to address the influence of speed variation, based on self-selected speed, on intensity and muscle activity patterns during stance phase of walking. To achieve this goal, we tested the following hypotheses: (1) there will be differences in muscle activity level between locomotion at self-selected speed and at slower and faster speeds, (2) there will be no differences in relative muscle activity patterns between the different speeds. The findings should provide information on the neuromuscular strategies related to locomotion at different speeds.

Methods

Subjects

Thirty-five healthy female subjects were recruited (age = 21.6 ± 3.17 years, height = 1.65 ± 0.045 m, body weight = 58.8 ± 7.72 kg, left Q angle = 14.57 ± 0.85 degrees; right Q angle = 14.7 ± 0.96° degrees; mean ± SD); possible candidates with at least one of the following criteria were excluded: history of recent osteoarticular or musculotendon injury of the lower limb or signs of neurological dysfunction which could affect lower limb motor performance; history of lower limb surgery; lower limb anatomical deformities, Q angle below 14° or above 17° (Nguyen, Boling, Levine & Shultz, 2009), due to the possibility that biomechanical changes resulting from abnormal alignment might influence joint loads, mechanical efficiency of muscles, and proprioceptive orientation and feedback from the hip and knee, resulting in altered musculoskeletal function and control of lower extremities (Shultz, Garcia, Gansneder & Perrin, 2006). All subjects were right-leg dominant.

The study was conducted according to the ethical norms of the Institutions involved and conformed to the Declaration of Helsinki, 1964. Informed consent was obtained from all participants.

Instrumentation

Ground reaction force (GRF) values were obtained from a force plate, model FP4060–10 from Bertec Corporation (USA), connected to a Bertec AM 6300 amplifier, with default gains and a 1000 Hz sampling rate. The amplifier was connected to a Biopac 16 bit analogical-digital (A/D) converter from Biopac Systems, Inc. (USA).

According to (Neptune, Zajac & Kautz, 2004), the muscle mechanical (and most likely metabolic) energetic cost is dominated not only by the need to redirect the COM in double support but also by the need to raise the COM in single support. The authors showed that the muscles responsible for most of the work on the COM were the gluteus maximus (GMax), hamstrings, rectus femoris (RF), gastrocnemius medialis (GM), soleus and vasti muscles. Hence, we selected a representative muscle for each subphase. As such, the electromyographic activity (EMGa) of GMax, BF, RF and GM muscles was monitored using the model MP 100 Workstation from Biopac Systems, Inc. (USA), with a sampling rate of 1000 Hz
and an amplified band-pass filter between 10–500 Hz (common mode rejection ratio (CMRR)>110 dB, gain = 1000) and analog-to-digital converted (12 bit). Data were collected using steel surface electrodes, model TSD150 from Biopac Systems, Inc. (USA), bipolar configuration, with a 11.4 mm contact area and an interelectrode distance of 20 mm, and a ground electrode. Skin impedance was measured with an Electrode Impedance Checker (Noraxon USA, Inc.)

Gait timing was measured using a photovoltaic system, model IRD-T175 from Brower Timing Systems (USA), and a metronome, model TempoPerfect Metronome Software from NCH Software (USA), was used to help subjects control gait cadence.

The signals acquired were processed with Acqknowledge version 3.9 for MP150 system (Biopac Systems, Inc. (USA)).

**Procedures**

**Self-Selected Speed Definition**  All subjects walked along a 10 m walkway (Chen, Kuo & Andriacchi, 1997; Whitle, 2007). They were instructed to walk along a force plate located in the middle of the walkway and to keep walking past the reference point without stopping. Self-selected cadence was recorded for each subject, from which two different cadences were defined—one 25% higher and the other 25% lower—and each subject was studied while walking at these three cadences. Average walking speed was assessed by measuring the time interval between two infrared beams at either end of the walkway, 8 m apart (Whitle, 2007). To diminish possible speed variations resulting from acceleration and deceleration, subjects walked a distance greater than the distance being monitored.

**Skin and Instrument Preparation**  The subjects’ lower limb skin surfaces were prepared to reduce electrical resistance to less than 5000 Ω (Basmajian & De Luca, 1985) by (1) shaving the skin surface of the muscle belly area; (2) removing dead cells with alcohol; and (3) removing nonconductor elements between electrode and muscle with an abrasive pad (Hermens, Freriks, Disselhorst-Klug & Rau, 2000).

Measurement electrodes were placed at GM, BF, RF and GMax midbelly according to anatomical references and fixed with adhesive tape (Basmajian & De Luca, 1985; Freriks, Hermens, Disselhorst-Klug & Rau, 1999). Measurements began 5 min after electrode placement as evidence suggests that there is a reduction of 20–30% in impedance values during the first 5 min after electrode placement (Vredenbregt & Rau, 1973).

**Measurement**  Subjects were required to walk at three different speeds (based on self selected, fast and slow cadence) after adequate practice (10–15 trials) to reach a steady pace in which the dominant foot stepped onto the plate. The subjects were barefoot and were asked to look straight ahead and walk, as naturally as possible, for a minimum of 8 steps (James, Herman, Dufek & Bates, 2007). They performed each speed three times, with metronome feedback. Only one foot at a time had full contact with the plate, and there was no extra load of any kind on the plate. Measurements were randomized to reduce the order effect, which can be caused by fatigue and previous muscle activation, and all procedures and verbal commands were given equally to all subjects.

The EMGa for each muscle was collected by a four-channel unit at 1000 Hz. The signals were preamplified at the electrode site and then fed into a dif-
ferential amplifier with an adjustable gain setting (12–500 Hz; CMRR: 95 dB at 60 Hz and input impedance of 100 MΩ). Raw energy signals were digitized and stored on computer disks for subsequent analysis by the Acqknowledge software. Then, the signal was filtered (20–500 Hz) and the root mean square (RMS) values for each muscle were calculated for each subphase (Basmajian & De Luca, 1985; Medved, 2001). They were also normalized according to the peak of the subject ensemble average (Yang & Winter, 1984). GRF data were filtered using a low-pass filter with a 20 Hz cutoff frequency and normalized according to weight (Mullineaux, 2006). EMG and GRF data were collected on the same A/D board and the subjects’ weight was measured during static posture on the plate.

The stance phase was defined as the interval during which the vertical reaction force exceeded 7% of body weight. This phase was then segmented into four subphases recognized in the trace of the vertical component of GRF. The time from heel strike to the first main peak of force, due to loading, was designated in this study as heel strike (HS). Midstance (MS) spans the interval between the two main peaks in the vertical force. The time between the first main peak of force and the local minimum during midstance was designated as transition between heel strike and midstance (HS-MS). The time between the local minimum during midstance and the second main peak was designated as transition between midstance and propulsion (MS-P). Late stance matches the final push-off phase of the ipsilateral limb; it lasts from the second peak of the vertical force to toe-off and was designated as propulsion (P).

Statistics

Data were analyzed with the Statistical Package Social Science software version 16.0 from SPSS Inc. (USA) using a significance level of $\alpha<0.01$.

As speed was not controlled directly but through step cadence, the ANOVA (Analysis of Variance) test was used to analyze speed differences. The Friedman and Wilcoxon tests were used to compare muscle activation levels at different stance subphases and speeds and to analyze the influence of speed on muscle activity patterns.

Results

Gait Speeds Adopted

There were statistically significant differences between the walking speeds adopted (Table 1), and as such our results can be further discussed under this premise.

<table>
<thead>
<tr>
<th>Speed (m/s)</th>
<th>N</th>
<th>Mean ± SD</th>
<th>p values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow</td>
<td>35</td>
<td>1.32 ± 0.29</td>
<td>&lt;0.0001</td>
</tr>
<tr>
<td>Self-selected</td>
<td></td>
<td>1.56 ± 0.36</td>
<td></td>
</tr>
<tr>
<td>Fast</td>
<td></td>
<td>1.81 ± 0.43</td>
<td></td>
</tr>
</tbody>
</table>
**GM, GMax, BF and RF Activity Patterns During Stance at Different Speeds**

At all speeds and stance subphases there were significant differences in relative muscle activity recruitment for GM, GMax, BF and RF \( (p < .0001) \).

At HS, GMax exhibited the highest activity at all speeds, followed by RF and then by BF and GM \( (\text{fast: GM-Gmax/BF/RF, } p < .0001; \text{BF/RF-GMax, } p = .001; \text{self-selected: GM-BF/RF/GMax, } p < .0001; \text{GMax-BF/RF, } p = .004; \text{slow: BF-RF/GMax, } p = .001; \text{RF-BF, } p = .005; \text{GMax-GM/RF, } p < .0001; \text{GM-RF, } p < .0001) \).

For values obtained during HS-MS, a similar pattern was observed at all speeds, with GMax and GM showing the highest activity, followed by BF and RF, although at the fastest walking speed there were no significant differences between RF and GMax and GM \( (\text{fast: RF-BF, } p = .003; \text{BF-GMax, } p = .001; \text{GM-BF, } p < .0001; \text{self-selected: RF-GMax, } p = .005; \text{GMax-BF, } p = .001; \text{GF-BF, } p < .0001; \text{slow: RF-BF, } p = .002; \text{GMax-BF/RF, } p < .0001; \text{GM-BF, } p < .0001) \).

During MS-P, the GM exhibited the highest activity, followed by GMax and RF, and lastly by BF \( (\text{RF-BF, } p = .005; \text{GMax-BF, } p < .0001; \text{BM-BF, } p = .001; \text{GM-BF, } p < .0001) \). At self-selected speed the GM showed the highest activity, when compared with RF and BF, and GMax exhibited the same activity level as GM \( (\text{RF-GM, } p = .001; \text{GMax-BF, } p = .003; \text{BF-BF, } p < .0001) \). At self-selected speed the GM and GMax showed the highest activity, followed by RF and then by BF \( (\text{RF-BF, } p = .005; \text{GMax-BF, } p < .0001; \text{GM-BF, } p = .001; \text{GM-BF, } p < .0001) \).

During P, the RF presented the highest activity at the fastest and slowest speeds, followed by BF and GMax, and finally by GM, while at self-selected speed the highest activity was exhibited by RF and BF, followed by GMax and GM; these last two muscles showing no significant differences \( (\text{GM-BG/RF, } p = .001; \text{BF-GMax, } p = .001; \text{GMax-GM-RF, } p < .0001; \text{BF-GM, } p < .0001; \text{GM-GM-RF, } p = .007; \text{GM-RF/RF, } p < .0001; \text{RF-BF, } p = .001) \).

**Speed Influence on Muscle Recruitment Level in Each Stance Subphase**

Figures 1–4 show speed influence on muscle activity in the different stance subphases. BF activity showed speed-related changes only during HS, decreasing at self-selected speed \( (p = .001) \). RF activity is most prominent at the fastest and slowest speeds \( (p < .0001) \) at HS, while in the other subphases it exhibits higher activity at the fastest speed \( (\text{HS-MS, } p < .0001; \text{MS-P, } p < .0001; \text{P, } p < .0001) \). GMax activity was higher at the fastest and slowest speeds than at self-selected speed \( (p = .002; \text{MS-P, } p < .0001) \), except during HS-MS and P. During HS-MS, GMax showed no speed-related changes, and during P the highest GMax activity occurred at the fastest walking speed \( (p < .0001) \). GM activity increased with speed at the fastest and slowest walking speeds (in decreasing order) only during MS-P \( (p < .0001) \). A comparison of each muscle activity during the different stance subphases shows statistically significant differences, as shown in Figures 1–4. At the fastest speed, GM showed differences in all subphases \( (p < .0001) \). At all speeds this muscle exhibited the highest activity during MS-P, followed by HS-MS, HS, and finally
by P. GMax presented the same pattern at all speeds, the highest activity occurring during HS, followed by HS-MS and MS-P, and finally by P ((HS vs HS-MS, $p < .0001$; MS-P vs P, $p = .001$ (fast) and $p < .0001$ (self-selected and slow); HS vs MS-P, $p < .0001$ (fast and slow) and $p = .001$ (self-selected); HS-MS vs P, $p = .005$ (fast) and $p < .0001$ (self-selected and slow); HS-P, $p = .001$ (fast) and $p < .0001$ (self-selected and slow)). RF, at all speeds, exhibited the highest activity during P, followed by HS, with no difference at the other subphases ((HS vs HS-MS, $p < .0001$; MS-P vs P, $p < .0001$ (self-selected); HS vs MS-P, $p < .0001$; HS-MS vs P, $p = .006$ (self-selected); HS vs P, $p = .001$ (fast), $p = .002$ (self-selected) and $p = .004$ (fast)). BF contribution was higher during HS for the fastest and slowest speeds. At self-selected speed its activity was higher at HS and P ((HS vs HS-MS,
Effect of Gait Speed on Muscle Activity in Stance

**Discussion**

Influence of Gait Speed on Muscle Activation Recruitment Patterns During Stance

During HS, GMax exhibited the highest activity, which illustrates its importance during this subphase, as the passive support of the bones alone is not enough to
prevent collapse. As a stance-side muscle, GMax plays an important role in the body’s fore-aft deceleration during the first half of stance (Liu, et al., 2006). The lowest activity of this muscle at self-selected speed could be explained by the fact that total negative work is reduced during loading response at the self-selected speed and increases at faster and slower speeds, as a result of changes in stride length and activation-deactivation dynamics, which limit the rate at which the muscle force can deactivate (Neptune & Kautz, 2001; Neptune, et al., 2008). The second highest activity was exhibited by RF, although according to (Nene, Byrne & Hermens, 2004) during swing to stand transition RF activity is probably due to crosstalk, which is consistent with the observation that the vasti muscles are the most active group (Liu, et al., 2006; Neptune, Kautz & Zajac, 2004; Zajac, et al., 2003). The importance of hamstrings in preventing knee hyperextension during late swing has been mentioned, suggesting the importance of this muscle on loading response (Whitle, 2007). The results of this study demonstrate a significant activity of this muscle during HS. The lowest activity was exhibited by GM, whose activity was not affected by speed changes. This can be explained by GM role in this subphase, as plantar flexors begin to support the trunk in early single-leg stance, because of their individual contributions to the hip intersegmental force, which accelerates the trunk upwards before MS (Neptune, et al., 2001).

During HS-MS, the higher activity of GMax and GM is supported by the importance of hip extensors in the development of positive work by concentric activity and the importance of GM contribution to trunk support, acting almost isometrically, and in the execution of negative work, when the tibia rotates over the foot (Neptune, et al., 2001; Norkin & Levangie, 1992). A similar pattern for GMax and GM at all speeds seems to be explained by the role of these muscles in body support during this subphase (Neptune, et al., 2001). It is important to note that at the fastest walking speed, RF activity matched GM and GMax levels which is consistent with RF activity being higher at the fastest speed. In fact, during MS knee extensors execute negative work, acting eccentrically to control knee flexion (Norkin & Levangie, 1992).

The preponderance of GM during MS-P at all speeds seems to suggest its importance in forward propulsion (Liu, et al., 2006). It is important to note that GM activity was only affected by speed during MS-P, which is supported by findings obtained by (Neptune & Sasaki, 2005) demonstrating that the ability of plantar flexors to produce force during propulsion is clearly affected as walking speed increases. As plantar flexors have been considered important contributors to support forward progression and initiation of swing during walking (Liu, et al., 2006; Neptune, et al., 2001; Zajac, et al., 2003), it might be necessary to increase the activity of other muscles to compensate for the decrease in plantar flexor activity. This can support the proximity of GMax to GM activity level. At the slowest speed, this can be explained by the temporally lower stability at slow speeds than at fast speed (England & Granata, 2007). RF and GM were the only muscles affected by speed, exhibiting higher activity at the fastest speed, as these two muscles work in synergy to provide trunk forward propulsion in walking (Jonkers, Stewart & Spaepen, 2003; Neptune, et al., 2004; Neptune, et al., 2008; Zajac, et al., 2003).

During P, the RF showed the highest activity at all speeds, which may be explained by its contribution to trunk forward propulsion in late stance (Neptune, et al., 2001). This muscle was the most affected by speed, showing higher muscle
Effect of Gait Speed on Muscle Activity in Stance

The higher activity of GM during MS-P at all speeds seems to be consistent with other studies (Liu, et al., 2006; Neptune, et al., 2008). Energy generation by plantar flexors during late stance corresponds to the greater work executed in the gait cycle and accounts for vertical and horizontal accelerations (Winter, 1991). It is important to note that during HS-MS GM exhibited the second highest activity at all speeds, corroborating dynamic simulations that demonstrate that the sum of the effects of plantar flexors during MS guarantees body support, allowing a consistent forward movement and thus preventing the collapse of the member (Neptune, et al., 2001; Simon, Mann, Hagy & Larsen, 1978). In this study, the contribution of GM to P was not predominant, which supports the statement that in this subphase there is higher contribution from the soleus muscle to increase speed (Zajac, et al., 2003).

RF exhibited the highest activity during P at all speeds, promoting trunk forward acceleration in synergy with the soleus muscle (Neptune, et al., 2004). During this subphase RF acts as the antagonist of the gastrocnemius, supporting the low GM activity observed in this study. The higher activity of GMax at HS at all speeds is consistent with other studies (Neptune, et al., 2008), as already mentioned.

The results of this study corroborate the idea that the fundamental regions of muscle activity remain relatively stable during the gait cycle (den Otter, et al., 2004; Hof, Elzinga, Grimmius & Halbetsma, 2002).

Influence of Gait speed on Muscle Recruitment Level

A global analysis of the results shows that muscles controlling the most proximal joints are more affected by speed changes than the more distal ones, as there was more variation in GMax, RF and BF than in GM. This is explained by a transition between work generated at the ankle, knee and hip (Chen, et al., 1997), suggesting a transfer of work to larger muscle groups when walking at faster speeds, which would allow muscles to work at a lower percentage of their maximum capacity and therefore optimize energy consumption during gait.

These results also show that fast and slow speeds are associated with increased muscle activity. Bipedal walking analysis shows that there is a biomechanical resonance associated with the inverted pendulum-like behavior of the skeletal structure and with muscle stiffness (Holt, Basdogan & Stoianovici, 1990), which may contribute to stability in normal walking (McGeer, 1990). Walking speeds not corresponding to this resonance frequency value require more neuromuscular active control to maintain a periodic stable movement (Ralston, 1958). In other words, faster walking speeds increase the segmental momentum, thereby requiring greater effort from the neuro-controller to attenuate kinematic disturbances. Short stride durations limit the allowable time for neuromuscular corrections to compensate for mechanical disturbances or controller errors. Slow walking speeds require active control that is out-of-phase with movement to slow the natural dynamics of the passive system (Cavagna & Kaneko, 1977). Mean walking speeds obtained in this study range from 1.32 to 1.81 m/s. It is argued that, for a moderate walking speed (0.5–1.5 m/s), costs in the swing phase are reduced; the metabolic cost is explained...
by muscle force generation during the stance phase (Winter, 1991). As fast speeds used in this study exceeded the 1.5 m/s indicated it would be relevant to execute the same analysis during the swing phase.

References


