SECTION – SPORT SCIENCES

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MUSCLE ACTIVATION PATTERN OF THE LOWER LIMBS IN A FEMALE RACE WALKER – CASE STUDY

Authors' contribution:

A. Study design/planning
B. Data collection/entry
C. Data analysis/statistics
D. Data interpretation
E. Preparation of manuscript
F. Literature analysis/search
G. Funds collection

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Abstract

Study aim: During race walking, the athlete executes specific patterns of movement to meet the judge's requirements. Thus, maintaining continuous foot contact with the ground and keeping the supporting leg straight may put specific demands on muscle activation patterns.

The objective of this research was to identify the changes in the lower limb muscle activity when the walker increased the speed of walking.

Basic procedures: The study was carried out at the indoor running track of the University of Physical Education in Krakow in 2015. The task of the subject was to walk a 60 m distance at different speeds. The subject walked with wearing shoes. The electromyographic (EMG) activity of the muscles was recorded using the telemetric EMG system. Raw signals of all the muscles were recorded simultaneously with foot-floor contact signals from foot switches fixed to the plantar surface of the foot.

Main findings: The study revealed that muscle leg activity had similar patterns at all speeds, however, a higher speed of walking put higher demands on the muscles. Due to that, we observed increased activity of the muscles when the subject walked faster. In general, the muscles were more active at the time of foot contact. In terms of kinematic data, we found that the number of steps for the chosen unit of time (10 s) increased from 13 strides at 9 km/h to 16 strides at 15.5 km/h. At the same time, the stride was lenghtened from 1.9 to 2.7 m.

Conclusions: The presence of an ‚atypical' burst of activity in the proximal muscles in late stance during racewalking requires special attention from the coaches during the training process of athletes.

Introduction

Race Walking is a progression of steps made in such a manner that the walker has contact with the ground, so that no visible (to the human eye) loss of contact occurs. The advancing leg must be straightened (i.e. not bent at the knee) from the moment of the first contact with the Grodnu until achieving the vertical upright position according to the rules of I.A.A.F. During race walking, the athlete executes specific patterns of movement to minimize oscillations of the center of gravity to reduce mechanical energy demands [1]. That is why it requires great technical ability. Optimal positioning of the supporting leg requires optimal hip, knee and ankle joint angles [2] under muscular control. Walking and running are the two most common forms of human locomotion. Although some similarities in both forms of gait exist, there are numerous changes reffered to by increased speed, including increased step length and cycle duration [3] or increased intensity of muscle activation [4,5]. Even though the muscle activity in locomotion is well recognized [6,7,8], there are only a few studies on muscle activity during race walking [2,9].

The speed of locomotion is determined by the length of the step and the step frequency. Thus, the natural strategy used to increase velocity of the race walker's movement may be provided by: either lengtening the step, or increasing the cadency of steps, or both [10]. Investigators report that there are some differences in the control of speed resulting from sexual dimorphism [11]. In their view, women more often use the frequency of steps to change the speed and men use both: step length and frequency to improve the speed. The strategy of controling the speed of a walking man is based on the optimization of the duration of the swing phase: it should not be too short (optimization due to the efficiency of locomotion) or too long (optimization due to the stability of the body) [12].

Walking is a cycling movement in which we can distinguish two phases: stance and swing. Stance phase is recognised when any part of the ipsilateral foot is in contact with the walking surface, while swing phase is the time period when that foot is not in contact [13]. Considering the movement of two legs, there are two subphases: single suport when one leg is maintained on the ground and double support when two legs are in contact with the ground. A fast walk maintaining continuous foot contact with the ground and keeping the supporting leg straight (Race Walking) may also require specific demands on muscle activity patterns.

The objective of this research was to identify the changes in the lower limb muscle activity when the walker increased the speed during race walking.

Material and methods

The subject of our study was a female race walker of the national class, age 23 (body mass 48 kg, body height 161 cm). The study was carried out at the indoor running track of the University of Physical Education in Krakow in June 2015. The task of the subject was to walk a distance of 60 m at different velocities. The subject walked wearing shoes. The electromyographic (EMG) activity of the muscles was recorded using the telemetric EMG system (Telemyo 900, Noraxon, USA, Inc., Arizona, USA). First, the subject was prepared for the placement of the electrodes on 8 muscles of the right leg. Before the electrodes were applied, the skin was shaved and cleaned with alcohol to minimize impedance. The placement and location of the electrodes were according to the recommendations by SENIAM (Surface EMG for noninvasive Assessment of Muscles) [14]. Therefore, two electrodes were carefully placed on the belly of each muscle, parallel to the muscle fibers with an inter-electrode distance of 20 mm. We registered the activity of the following lower limb muscles: *Vastus Medialis* (VM), *Vastus Lateralis* (VL), *Rectus Femoris* (RF), *Biceps Femoris* (BF) and *Medial* and *Lateral Gastrocnemius* (MG and LG, respectively) and *Tensor Fascia Latae* (TFL). The electrodes were not moved during the test. Raw signals of all the muscles were recorded simultaneously with foot-floor contact signals from foot switches fixed to the plantar surface. Foot switch data were used to determine inictial contact and toe-off of each recorded stride. The gait cycle was defined as the time between two successive foot contacts of the right leg. Stride lenght was recognised as the distance between two legs, while step frequency was defined as the number of steps during the unit of time.

EMG analysis

The raw EMG signal was processed using Noraxon's Myoresearch software (MR-XP 1.07. Master Edition). A linear envelope was created by applying full wave rectification and RMS based smoothing to the raw signal. The raw EMG data was filtered using a 50 Hz notch filter to remove any electrical interference from external sources. The signal was then filtered a second time using a 15–500 Hz band pass filter. This allowed noise or movement interference below 15 Hz and other nonphysiological signals above 500 Hz to be removed. The data were smoothed using root mean squared analysis (RMS), which was calculated for a 50 ms window. All gait strides were averaged in normalized time cycles. The analysis was based on the 10-second interval selected for each velocity, that allowed to achieve mean values of amplitude. To normalize our data, we used the trial of the highest velocity (15.5 km/h), where we referred the data to the peak value of the amplitude. This test normalization makes the average EMG patterns highly reproducible and enables test comparisons.

Results

We recorded EMG activity from 8 unilateral muscles in one subject who performed race walking as she walked on the ground at different speeds. We studied

EMG signals of the muscles in order to identify the overall pattern and characteristics of the muscle activity at different race walking speeds, changes in their amplitude and duration, as well as to compare muscle activation under loaded (stance) and unloaded (swing) conditions. Apart from this, our experiment allowed to observe some kinematic changes.

Our observation revealed changes in the proportion of the stance and swing phase. As velocity increased, the duration of the stance phase regularly decreased (see: Table 1).

Since velocity is the product of step lenght and frequency, we found that the number of steps for the chosen unit time (10 s) increased from 13 strides at 9 km/h to 16 strides at 15.5 km/h. At the same time, the stride was lengthened from 1.9 to 2.7 m. The 60 m distance was covered 10 seconds faster from the slowest (23.8 s) to the fastest velocity (13.9 s).

The average EMG patterns (see: Fig. 1) show the activity characteristics of the muscles during race walking when the speed of walking trials gradually increased.

The study revealed that muscle leg activity had similar patterns at all speeds, however, a higher speed of walking put higher demands on the muscles. Due to that, we observedincreased activity of the muscles when the subject walked faster. The largest activation of the muscles was recorded for the trials at the highest speed (15.5 km/h).

At the moment of heel strike, when the stance phase is initiated, we observed activation of the VL, BF and TA which continued from the limb advancement in the late swing phase. Thereafter, when the supportive leg bears the load, the EMG amplitudes of G (both heads) increased considerably. Moving into late stance, their activity gradually decreased, but at the same time, we observed a burst of activity in proximal muscles (RF, VM, VL and TFL), which is typically not present during normal walking [4,5]. TFL activity continued to be present for half of the swing phase, while the rest was diminished until the moment of the late swing when the activity of BF, VL and TA gradually increased towards heel strike (as mentioned above).

The increase in walking speeds is accompanied by the increment activation of all the muscles, especially for the highest speed. While for most of the muscles we can see gradual growth of the amplitude, in the case of the *Quadriceps* and *Tensor FL*, the highest speed required a rapid burst of activation.

The results presented in Table 2 allow to quantify the differences of EMG activity within the phases of the gait cycle in terms of the gradually increased speed as well as between the phases (stance and swing).

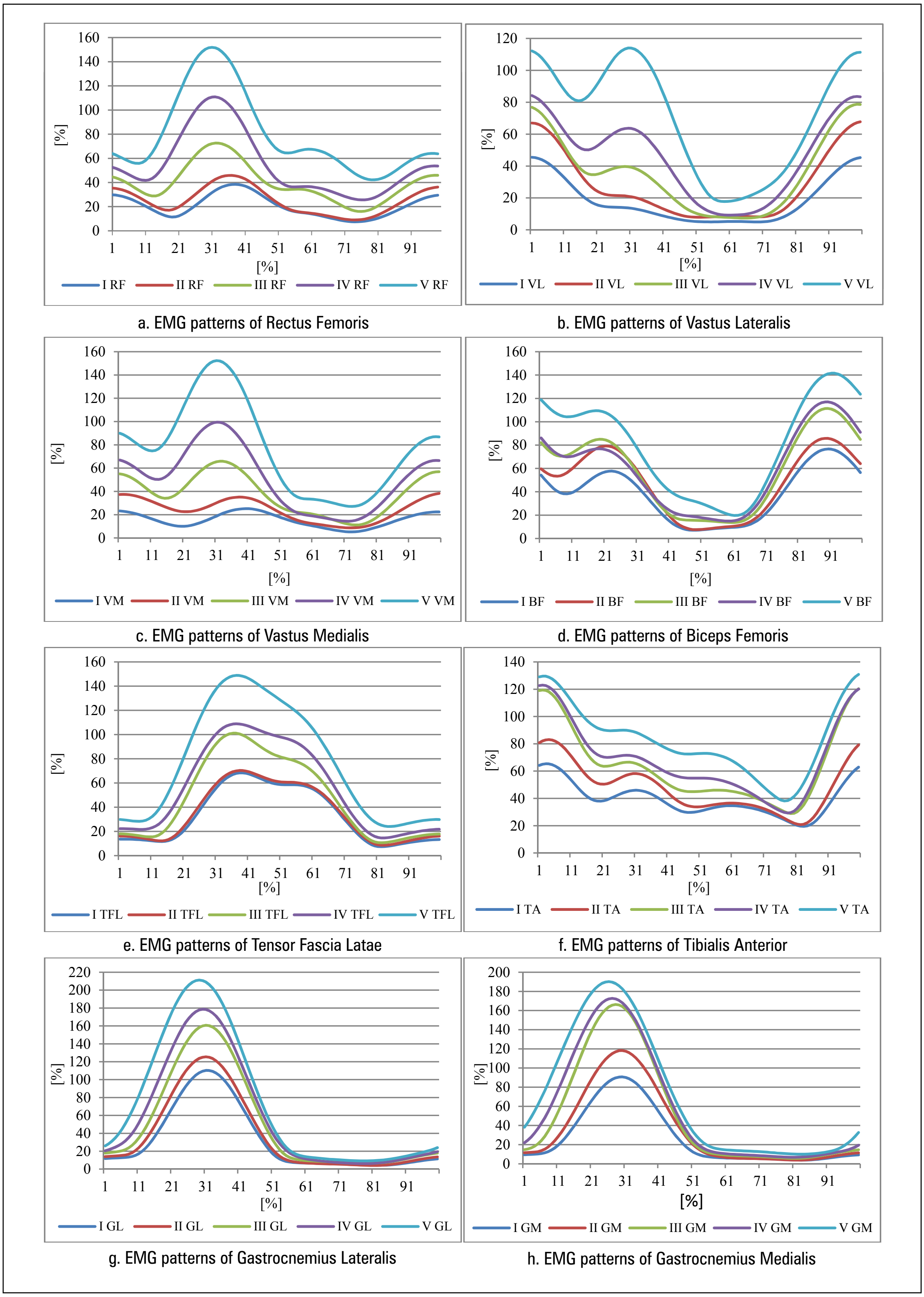
Observing the muscles' involvement between the walking phases, in general, muscles were more active at the time of foot contact (RF, VM, TA, GL, GM). For VL, the differences were significantly higher during the stance phase when the subject was walking faster (above 12.1 km/h). The activity appeared to be significantly different for the TFL at 12.1 km/h. As analysis revealed, the BF activity was stastistically lower for the swing phase for BF in the fastest walking.

Considering the level of amplitude of the registered muscles, the post hoc Tuckey test revealed significant differences during stance phase among all the walking trials in the following muscles: GL, GM, VM, VL, TA. We did not observe significant differences for BF for walking at 12.1 and 13.6 km/h, and for RF and TFL between trials at the velocity of 9 and 10.9 km/h.

In terms of the muscle activity during swing phase, we observed significant differences for all speeds in the case of VM. The differences were not significant between the first and second speeds for RF and TFL. Amplitide of BF and TA was not significantly different between the triasl at the speed of 9 and 10.9 km/h or 12.1 and 13.6 km/h. The level of muscle activation was not significantly different for both heads of the Gastrocnemius while walking at the speed of 9 vs. 10.9 km/h, 10.9 vs. 12.1 km/h, and finally 12.1 vs. 13.6 km/h. For the remaining cases, the amplitude was significantly different.

Table 1. Kinematic parameters of the walking trials: v - velocity, ST - stance phase in percentage of gait cycle (GC), T- time, SL – stride lenght, f - frequency = strides per 10 s

Kinematic parameters	velocity				
	9 km/h	10.9 km/h	12.1 km/h	13.6 km/h	15.5 km/h
v [m/s]	2.52	3.03	3.35	3.77	4.31
ST [% GC]	52	50	48	46	44
f [strides per 10 s]	13	14	14	15	16
T [s]	23.8	19.8	17.9	15.9	13.9
SL [m]	1.9	2.2	2.4	2.5	2.7



a. EMG patterns of Rectus Femoris

b. EMG patterns of Vastus Lateralis

c. EMG patterns of Vastus Medialis

d. EMG patterns of Biceps Femoris

e. EMG patterns of Tensor Fascia Latae

f. EMG patterns of Tibialis Anterior

g. EMG patterns of Gastrocnemius Lateralis

h. EMG patterns of Gastrocnemius Medialis

[*] I, II, III, IV, V – values of the race walking velocity

Figure 1. The average EMG patterns of the muscles during the gait cycle

Table. 2. Average EMG amplitude [%] in the phases for the muscles during walking at increasing velocity (significant differences between phases are marked in red, p≤ 0.001)

Walking Cycle	velocity									
	9 km/h		10.9 km/h		12.1 km/h		13.6 km/h		15.5 km/h	
	$\bar{x}$	SD	$\bar{x}$	SD	$\bar{x}$	SD	$\bar{x}$	SD	$\bar{x}$	SD
Tensor Fascia Latae [%]										
Stance	39.00	23.05	40.59	23.45	58.65	34.46	66.32	35.80	89.65	48.64
Swing	26.77	19.52	29.79	20.12	38.01	27.03	48.19	32.77	67.84	42.03
Rectus Femoris [%]										
Stance	25.64	8.36	31.99	9.06	49.72	14.58	74.28	25.02	103.35	36.11
Swing	15.32	6.56	18.37	8.30	30.49	8.98	38.28	9.31	60.18	11.88
Vastus Lateralis [%]										
Stance	18.85	12.63	29.35	18.71	40.55	16.44	58.00	12.83	96.92	12.73
Swing	16.44	14.24	24.73	21.17	29.17	25.97	34.28	26.68	49.87	31.28
Vastus Medialis [%]										
Stance	18.31	4.94	29.97	4.84	49.31	10.63	72.59	17.02	111.36	27.81
Swing	12.37	5.55	18.09	9.29	26.93	14.25	32.72	17.61	49.97	21.12
Tibialis Anterior [%]										
Stance	43.95	10.68	56.76	14.46	72.91	22.92	80.16	21.31	96.85	17.23
Swing	33.37	10.82	37.73	14.71	54.00	26.04	57.40	25.27	70.83	25.01
Gastrocnemius Lateralis [%]										
Stance	56.09	35.91	68.00	39.93	91.67	50.47	108.82	54.24	137.33	61.10
Swing	6.74	1.99	7.73	3.74	12.33	7.87	15.65	13.80	21.97	20.72
Gastrocnemius Medialis [%]										
Stance	46.99	29.29	65.28	37.39	95.82	51.43	109.54	48.44	132.53	46.66
Swing	5.87	1.54	7.17	3.42	10.12	4.82	12.85	8.26	19.49	13.89
Biceps Femoris [%]										
Stance	35.99	17.66	49.92	23.96	58.40	25.09	58.34	21.06	87.88	26.25
Swing	40.95	26.75	45.09	30.36	57.75	39.13	60.85	40.80	73.79	48.04

Discussion

We recorded over ground race walking trials at different speeds (range 9-15.5 km/h).

The walker can adopt a strategy mostly based on antropometry and level of training [15]. Adjusting the length of step and frequency relation is an effective way to improve the speed [16]. In our study, gradual changes in locomotion speed resulted in changes in the step length and frequency, as well as the proportion of the stance and swing phase in the gait cycle. This was also shown in other studies [10, 3]. However, observing the extent of the changes, we noticed that the speed basically inreased by prolonged step rather than increased frequency of steps. It seems that this is more typical for the strategy used by females as we mentioned previously, however, it may not be that efficient. Larger changes in cadency suggest easier control of this parameter compared to the length of step. Most likely, the mechanism responsible for the selection of the speed ranges is associated with the body's energy expenditure [17]. At faster walking speeds, a maximum limit on stride length is reached; the fastest walking speeds are achievable by increasing only stride frequency [8]. Consequently, the time of foot contact with the ground was shorter, which was also found in the study by Nilsson [3]. It seems to be a very practical premise for the coach. Knowing the values of indicators of body composition and spatiotemporal parameters, each coach can determine whether the player has the optimal values of time and space parameters in relation to body construction [18].

Our study revealed that muscle leg activity had similar patterns at all speeds although mean amplitude of the lower extremity muscle activity tends to increase with speed of movement. Also worth mentioning is that the differences in the characteristics of muscle activation mainly accompanied the stance phase.

The main role of the muscles in the regulation of walking speed is to control the accelerating and decelerating forces of individual body segments to establish safe forward progression [19]. As a result, the amplitude of muscle activity increases with walking speed because of the need for larger muscular force output [10].

In walking or running, the *Quadriceps* mainly serve to support the subject's body weight. Hamner [15], analysing the contribution of muscles in propulsive and supportive function during running on the base of an EMG noticed that during the braking of phase stance, the *Quadriceps* muscle group was the main contributor to both braking and support. However, in race walking, because of the abnormal movement of the knee, its role is also to help keep the knee extended. The requirement is that the athlete's knee must be extended from the first contact with the ground until the vertical upright position (IAAF 2014). Thus, at least one of its heads: *Vastus Lateralis, Vastus Medialis* and *Rectus Femoris,* is involved during the stance phase also supported by *Biceps Femoris*. As Murray [2] suggests, the early stance phase activity of the *Vastus Lateralis* probably serves to rapidly lock the knee in hyperextension with the hamstrings acting as dynamic ligaments to prevent stretching of the posterior capsule of the knee. Our observation is confirmed by Ivanenko [6] who studied muscle activation at different speeds in walking and running and noticed that in addition to the activation of the distal muscles seen in late stance, there is 'atypical' temporal activation of the proximal muscles (namely VL, VM, RF) at a higher (non-preferred) walking speed (9 km/h). Thus, it seems that this ‚burst' may play an important role in the biomechanics of race walking (see: Figures RF, VL, TFL).

Considering the contribution of the calf muscles (*Gastrocnemius* and *Tibialis Anterior*), Murray [2] explained that their activity during stance controls the forward rotation of the leg segment over the fixed foot (ankle dorsiflexion), and thus, also indirectly helps to keep the knee extended by exerting a backward pull on the leg segment. Besides, the *Triceps Surae* is the main contributor in propulsion [15], so also in our study, we noted that the EMG amplitudes of the *Gastrocnemius* (*Medialis* and *Lateralis*) increased considerably. Nevertheless, it is worth noting significant differences in the LG vs. MG activity between race walking and normal walking: in the former case, the activity of both muscles increase nearly to the same extent with increasing walking speed (Figure 1e,f), while in the latter case, the activity of LG increases much more than that of the MG [20]. Moving into late stance, the activity of the *Triceps Surae* gradually decreased and the preparation for toe-off occured [9]. Lower limb muscle activity during stance in race walking suggests that the muscles involved, i.e. the ankle extensors and the femoral extensors, all participate in generating propulsive shear forces and limb stiffness. In addition, lateral trunk stabilization may also be concerned. This is supported by activation of the following muscles: *TFL* and *Gluteus Medialis* [7].

Increase of the activity of the *Rectus Femoris* at the largest speed of walking during early swing probably serves to accelerate flexion of the hip of the swinging limb. In addition, this hip abductor muscle activity may also help abduct the swinging limb past the supportive limb. This action is also accompanied by continuous activation of the *Tensor Fascia Flexor* (as a hip abductor). The late swing activity of the hamstrings (*Biceps Femoris*) decelerates the knee extension of the swinging limb. The increase in the amplitude of swing-phase activity of the *Tibialis Anterior* most likely relates to the increased ankle dorsiflexion measured during race walking, as compared to normal fast walking [2].

In a further study, it would be interesting to clarify the functional significance and reorganization of muscle activity (in particular, the proximal extensor muscles) during race walking with respect to normal walking and running.

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