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Walking and Jogging: Quantification of Muscle Activity of the Lower Extremities

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1. Introduction

Basmajian (1967) defined locomotion as "the translation of the centre of gravity through space along a path requiring the least expenditure of energy". Taking as reference the centre of gravity (CG) of the body, it has been already established that the trajectory while walking or jogging follows a sinusoidal course (Inman 1968), with two vertical summits per stride (one cycle of locomotion).

During locomotion, the movement of translation of the body is produced by the spatial angular movement of the lower extremities that shows support and non-support phases. The basic unit of measurement in gait analysis is the gait cycle (GC) or stride. The gait cycle is commonly divided into two phases: 1) stance phase, and 2) non-support phase. In walking, in the stance phase there are two periods of double support. In jogging cycle (JC) the phase of non support is composed of two float periods (when both feet are airborne) and between both periods there is one period of swing of the ipsilateral leg.

Walking and jogging present differences at: 1) the velocity pattern, which is higher in jogging, as it implies joint movements of higher amplitude and speed; 2) the duration of cycles and of their respective phases; 3) the kinematics of: a) the peaks of the body CG trajectory (its amplitude and time of appearance), and b) of the joints, as there are wider angular joints displacements and different distribution of arches movements during cycle phases (Novacheck, 1998); 4) the kinetic: a) amplitude, duration, and evolution of the ground reaction force (Mann, 1982), external moments of force, and muscle power (Winter, 1983a and 1983b); b) misalignment of CG between the segments of the leg and of the upper part of the body (head, trunk, and arms); c) inertia; and 5) the mechanisms for saving energy structures (Cavagna & Kaneko, 1977; van Ingen Schenau et al., 1987).

Several studies have compared the muscle activity of the legs during walking and jogging or running. In the works of Mann & Hagy (1980), and Mann (1982), the subjects were trained athletes, and in the work of Ounpuu (1994) they were children. In all the above-mentioned papers the muscle activity was recorded from only one leg, and the muscular activity was qualitatively described rather than quantified. Nilsson et al. (1985) and Gazendam and Hof (2007) quantified the profiles of averaged rectified EMGs at different speeds of locomotion and mode of progression in humans on motor-driven treadmill.

The work here presented was carried out to investigate and to quantify the electromyographic activity (EMG –the weak electric signal produced by contracting muscles) of the lower extremities during walking and jogging and to evaluate the changes produced in jogging. The subjects (who practice regularly physical activities) walked and jogged barefoot at ground level at spontaneous speed. These subjects had a healthy condition, but they were exposed to lesions due to their jogging practice. The aim of the present study was to quantify the changes produced in the EMG signal characteristics (as amplitude and time of occurrence of peak activity) of both legs muscles during jogging compared with walking.

The comparison carried out in this study between the EMG activity of the leg muscles during walking and during jogging can reveal the adaptability of the neuromuscular system against more demanding mechanical needs, and in this way we can learn more about the etiopathology of lesions.

2. Materials and methods

2.1 Subjects

Six male and four female University students who regularly practice jogging (heel-toe style) have participated voluntarily in this study. Five subjects were 19 years old, four of them 20, and one, 30, with an average height of 1.73±0.10m, and a body mass of 64.2±6.0 Kg. The 10 subjects were right lower extremity dominant (determined by the leg they reported to use for shooting a ball). Written informed consent was obtained from all of them. The experimental protocol got the approval of the Scientific Committee of the Basque Institute of Physical Education.

2.2 Instrumentation

The stride phases (support and non-support) were identified by four footswitches (B&L Engineering, Tustin, California) that were taped to each foot sole over the heel, on the heads of the first and fifth metatarsus, and on the tip of the toe. The footswitches (FSs) were activated when the pressure was above 1.5N, producing an on/off output voltage.

For the surface EMG (s-EMG) signals acquisition, circular, three-pole (1.0cm diameter each pole), stainless steel, bipolar surface electrodes (B&L Engineering, Tustin, California) were used. Double differential signals were obtained between each of the outer two active electrodes and a reference central electrode to improve the signal-to-noise ratio and to diminish the cross-talk from adjacent muscles. The centre-to-centre inter-pole distance was 1.5cm. An earth electrode was placed on the wrist. The electrodes had a built-in 320-gain preamplifier, which had an input impedance greater than 20MΩ, minimum common-mode rejection ratio (CMRR) of 95dB (at 50Hz), and a bandwidth from 10Hz to 30KHz (cut-offs at -3dB). The dimensions of the whole body of the electrodes were 5cm x 1.8cm x 0.7cm. An optic-fiber cable transmitted the signals from the connection box, located on the subject, to the acquisition system.

Surface electrodes were chosen for the study of the muscular activity because they offer the following advantages respect to intra-muscular electrodes: they produce no pain neither harm the tissue, they are easier to place, and as already shown by Giroux & Lamontagne (1990), they record statistically similar signals to those obtained with intramuscular wire electrodes. The potential problem of crosstalk was reduced using a double differential technique, which is based on a single amplifier fed with three electrodes (De Luca &

Merletti, 1988; Winter, 1990; Winter et al., 1994). It is already widely accepted that double-differential technique reduces the level of cross-talk (see, *e.g.*, De Luca & Merletti, 1988; and Meinecke et al., 2004). In any case, cross-talk cannot ever be fully cancelled.

In addition, as the subjects were moving during the recordings, surface electrodes gave a more robust signal (free of artifacts) than internal electrodes.

Before performing the recordings, the skin over the target muscles was shaved and then cleaned with alcohol; no gel was used between the skin and the electrodes. Subsequently, the subjects were asked to execute specific movements for each muscle to check the output of the s-EMG electrodes. In order to test the proper functioning of the FSs, the subjects were asked to step on the floor in a way that activated sequentially the FSs. Once the experimental setup was verified, the subjects were asked to walk or jog freely along the laboratory to get used to the electrodes and sensors until they felt comfortable with the equipment and the speeds of consecutive strolls were similar.

Surface EMG was recorded from six target muscles of each leg: rectus femoris (RF), vastus medialis (VM), long head of biceps femoris (BF), semitendinosus (ST), tibialis anterior (TA), and lateral gastrocnemius (LG). Those muscles are the ones normally targeted in the literature, because they are important for the locomotion and because they are superficial and hence easily detectable. In order to obtain a signal with higher amplitude, the electrodes were placed parallel to the fibers direction of each muscle, which was estimated following the work by Wickiewiz et al. (1983). Electrodes were placed on the muscles following the SENIAM recommendations (SENIAM, 1999).

The placement of the electrodes over each target muscle was as follow; RF: at 50% on the line from the anterior iliac spine superior to the superior part of patella; VM: at 80% distal on the line between the anterior iliac spine superior and the joint space of the knee in front of the anterior border of the medial ligament; BF: at 50% on the line between the ischial tuberosity superior and the lateral epicondyle of the tibia; ST: at 50% on the line ischial tuberosity superior and the medial epicondyle of the tibia; TA: at 1/3 proximal on the line between the tip of the fibula and the tip of the medial malleolus; and GL: at 1/3 proximal on the line between the head of the fibula and the heel.

The abovementioned signals (12-channel s-EMG, and two from the FSs status) were registered with a MA-200 (MotionLab Systems, LA, USA) equipped with a Pentium PC endowed with a 16-channel acquisition card CODAS DI 400 PGH (Dataq Instruments, OH, USA) featuring 12 bits of resolution, a gain of x8, a pass-band filter from 10Hz to 1KHz, a CMMR of 40dB, an input impedance above 1MΩ, and a sampling rate of 3 Ksamples/s, as indicated by Merletti (1994).

2.3 Experimental protocol

Individuals were barefoot. Five successive s-EMG recordings were made from each subject while walking and another five recordings while jogging, all of which at spontaneous speed because, as proofed by Kadaba (1989), the s-EMG activity is more reproducible when subjects walk at their spontaneous velocity.

The measurements were carried out in a 10m long laboratory, where the subjects walked at ground level. The average speed (in m/s) was calculated measuring the time required for covering the 10m. There was a one-minute break between each recording. Only the three central cycles of those recordings have been analyzed in order to avoid the acceleration and deceleration effect. All the subjects that participated in the study were athletes, whose ability to start and stop abruptly is quite higher than that of average subjects. Therefore, the effect

of acceleration and deceleration is low in this kind of subjects. In any case, the objective of this work was not to study the influence of the locomotion *speed* on the muscular activity, but of the *type* of locomotion, keeping constant the EMG signals acquisition conditions.

2.4 Data analysis
2.4.1 Spatio-temporal parameters

2.4.1.1 Velocity

Average velocity was calculated by measuring with a stopwatch the elapsed time each subject took to cross the 10m length of the laboratory.

2.4.1.2 Temporal components of the locomotion cycle

The signals from the FSs were displayed as a staircase. The time length of support and non-support phases were normalized with respect to the duration of the whole locomotion cycle and expressed as percentages of the respective cycle (gait cycle or jogging cycle), 0% being the beginning of the support phase and 100% the end of the non-support phase.

2.4.2 EMG signal

Surface EMG signals were acquired from six muscles for each leg (RF, VM, BF, ST, TA, and LG). In order to smooth the signal and to reliably determine the s-EMG peak value and its time of occurrence, the raw signals were firstly full-wave rectified and then averaged using a moving window of 50ms length (150 samples), which yielded the linear envelope (LE) corresponding to the s-EMG signal. The LE was used because it closely represents the continuous rises and falls profile of muscular activity as a function of time (Inman et al., 1952).

The intra-subject s-EMG activation profiles were calculated using ensemble averages (EAV) computed as follows: each EAV point was calculated as the average of a 2%-segment of the already averaged s-EMG signal (Winter, 1991) of the 15 selected strides (3 central strides of each stroll x 5 strolls per subject for each locomotion mode).

The s-EMG amplitude normalization of each muscle activity for each subject and for each type of locomotion was performed with respect to the maximum peak of subject's EAV in each type of locomotion (Yang & Winter, 1984). For muscles having two activity peaks, the highest of them was chosen as the normalizing peak and it will be the one referred as "activity peak" hereafter.

The grand ensemble averaged (GEAV) was obtained by averaging the EAV of all the subjects for each kind of locomotion. The electromyographic activity, expressed as GEAV, was analyzed in absolute (GEAV) and normalized (nGEAV) values. Un-normalized values allowed us to appreciate the evolution of the muscle activity when the locomotion changed from walking to jogging (hence increasing the mechanical requirements), while the electrodes remained in the same position.

2.5 Statistical analysis

The mean and the standard deviation (SD) were calculated for the spatio-temporal parameters and for the peak of the lineal envelop (EAV).

The influence of the kind of locomotion on s-EMG peak location and on spatio-temporal parameters (namely, kind of locomotion and percentage of the cycle) was determined applying a t-Student test for independent measurements using *SPSS* statistical package (SPSS Inc., IL, USA), all of them calculated at a significance level of $p < 0.05$.

The coefficient of correlation (CC) between two muscles activity profiles was defined as the ratio between the covariance of those two profiles and its associated standard deviation. The CC cannot exceed 1 in absolute value, and the closer it is to 1, the more similar are the compared profiles. Therefore, the CC provides an index of similarity, and it has been used to measure the similitude of the EMG profile of homologous muscles (those that are the same, but located in different legs).

The inter-subject variability of s-EMG signal was determined by the coefficient of variation (CV –ratio of the standard deviation to the mean) of the average across the 10-subject EAV, which generated an inter-subject EAV and its associated standard deviation.

Locomotion is a complex act that involves the integration of several variables during its learning process; hence, each subject has its own personal way of walking. Therefore, in order to increase our understanding of human locomotion it is necessary to use average values of the muscular activity, so that inter-subject differences can be diminished. Therefore, in this study, the acquired s-EMG signals have been further studied only after the inter-subject reproducibility was assessed, and always from both legs (Inman et al., 1981).

3. Results and discussion

The aim of this study was to analyze: 1) the changes in amplitude on un-normalized electromyographic signal, and 2) the time of appearance of the s-EMG peak from muscles of both legs when the locomotion style changed while the electrodes remained in the same position. All of that after having evaluated and validated the reproducibility of the s-EMG signals. We have quantified the two types of locomotion by using EMG in order to obtain a powerful and objective tool for understanding the origin of lesions that are more frequent in jogging.

3.1 Spatio-temporal parameters

The average speed during walking was 1.33±0.12m/s (mean ± standard deviation), and 2.50 ±0.31m/s in jogging, which implies an average increase of 87%.

Table 1 shows the average time length of each phase in absolute and relative values (respect

Phase	Jogging		Walking	
	average [s]	JC [%]	average [s]	GC [%]
Left leg support	0.26 ±0.02	37.8	0.60 ±0.06	61.0
Left leg non-support	0.45 ±0.03	62.8	0.38 ±0.03	39.0
Right leg support	0.25 ±0.10	35.3	0.59 ±0.07	60.5
Right leg non-support	0.46 ±0.12	64.3	0.38 ±0.04	38.6
Left stride	0.71 ±0.03	100.0	0.99 ±0.08	100.0
Right stride	0.72 ±0.03	100.0	0.97 ±0.09	100.0

Table 1. Comparison of time length of each phase and its corresponding cycle percentage for the jogging cycle (JC) and for the gait cycle (GC). All the values showed significant statistical difference ($p<0.05$)

to the duration of the whole stride) detected for walking and for jogging. On average, the cycle time for walking was 0.99s for the left leg and 0.97s for the right leg. For jogging, the cycle time decreased to 0.71s and 0.72s respectively. In jogging, the support phase decreased a 50% with respect to the GC and hence, in that short time, a bigger ground reaction force had to be attenuated and bigger and faster displacement of the segments had to be achieved (Mann, 1982). All the above mentioned changes are significant at a certainty level of 95% ($p<0.05$).

3.2 Reliability of the s-EMG signal

For the muscles studied, the obtained functional patterns for walking and for jogging were similar to those already published (Gazendam & Hof, 2007; Mann & Hagy, 1980; Mann, 1982; and Ounpuu, 1994).

Inter-subject variability was determined by averaging the un-normalized and normalized EAV of the 10 subjects (GEAV). Table 2 shows the CVs calculated from those un-normalized (GEAV) and normalized (nGEAV) grand ensemble averages. A high CV corresponds to a higher inter-subject variability in the s-EMG of the analyzed muscles, and hence of their activity levels (Kadaba, 1989; and Winter & Yack, 1987).

Leg	Muscle	n	Number of observations	CV of GEAV [%]			
				Jogging		Walking	
				uN	N	uN	N
Left Leg	Rectus Femoris	10	50	79	17	41	35
	Vastus Medialis	9	45	49	25	51	34
	Tibialis Anterior	10	50	39	31	30	24
	Biceps Femoris	10	50	67	26	63	37
	Semitendinosus	10	50	56	27	51	25
	Lateral Gastrocnemius	10	50	66	24	47	25
Right Leg	Rectus Femoris	10	50	67	20	51	39
	Vastus Medialis	9	45	34	69	44	59
	Tibialis Anterior	10	50	38	27	31	27
	Biceps Femoris	10	50	61	26	77	28
	Semitendinosus	10	50	59	28	71	25
	Lateral Gastrocnemius	10	50	66	25	55	29

Table 2. Inter-subject (n: number of subjects) coefficients of variation (CV) of un-normalized (uN) and normalized (N) grand ensemble average (GEAV)

In jogging, when the mechanical demand increased, there was a higher inter-subject variability in the LE of the s-EMG of the greater part of the muscles, as showed by the higher values of CV for the un-normalized s-EMG.

The CV of the GEAV obtained from the normalized EAV, was reduced of about 50% in the greater part of the muscles. That CV was in general not too different in both types of locomotion, except for the RF muscle, which showed a higher variability in walking than in jogging, although its muscular activity was smaller in walking. The variability of the EMG has been studied deeper in walking (*e.g.*, Kadaba et al., 1989; and Winter & Yack, 1987) than in jogging (*e.g.*, Karamanidis et al., 2004).

In the present study, the nCV values obtained during walking were smaller than those reported by Kadaba et al. (1989); and by Winter & Yack (1987). The differences found between our CV results and those of the abovementioned studies could be explained by the different computations used in the normalization of the EMG activity: setting the mean value of each subject's EMG over the whole stride period (Winter & Yack, 1987), or smoothing the EMG data using a filter with a low pass cut-off frequency of 12-14Hz (Kadaba et al., 1989).

Other studies normalized the EMG signal of each muscle by the value corresponding to its maximum voluntary contraction (*e.g.*, Arsenault et al., 1986a; and Perry et al., 1993). In the present study we did not use that normalization because it is not always possible to apply it; *e.g.*, in the case of injured subjects and hence it could not be used for the sake of comparison in future works. Instead, we normalized the EMG recordings corresponding to walking and to jogging trials by their respective peak values during locomotion at spontaneous speed because it is a simple and reliable way of obtaining comparable muscular activity profiles (Yang & Winter, 1984) .

In the present work, the nCV of the 10 analyzed subjects were not significantly different (below 35%) in walking and jogging, which points out that there are more similarities than differences in the muscular activity between subjects in both ways of locomotion.

The differences detected on the CV of muscles from a same muscle group can indicate that a muscle has more than one function and a more variable activity, as already pointed out by Winter & Yack (1987).

3.3 Amplitude of myoelectric activity

Figure 1 shows that the muscular activity profiles of five out of the six studied muscles for each leg were similar in walking and in jogging, although there were differences in the sequence of movement arches of hip and knee corresponding to the phases of support and non-support, and also in the direction of the movement arches of the ankle.

The above-mentioned findings support other studies, such as those of Gazendam & Hof (2007); Mann & Hagy (1980); Mann (1982); and Ounpuu (1994).

Figure 1 also shows that there was a similar pattern (nearly identical) between the grand ensemble averages of muscular activity in both legs.

The idea of registering the activity of both legs was to ensure the validity of the assumption made in other works about the symmetry of the activity of both legs.

Table 3 shows the coefficients of correlation (CC) obtained when comparing the GEAVs of homologous muscles (those same muscles in different legs). These findings have provided some support to the assumption of symmetry made in the literature and are on line with the studies de Arseanult et al. (1986b), and by Ounpuu & Winter (1989), who showed the symmetry existing in walking; and by Raibert (1986), who found out that symmetry is desirable because it would simplify control strategies.
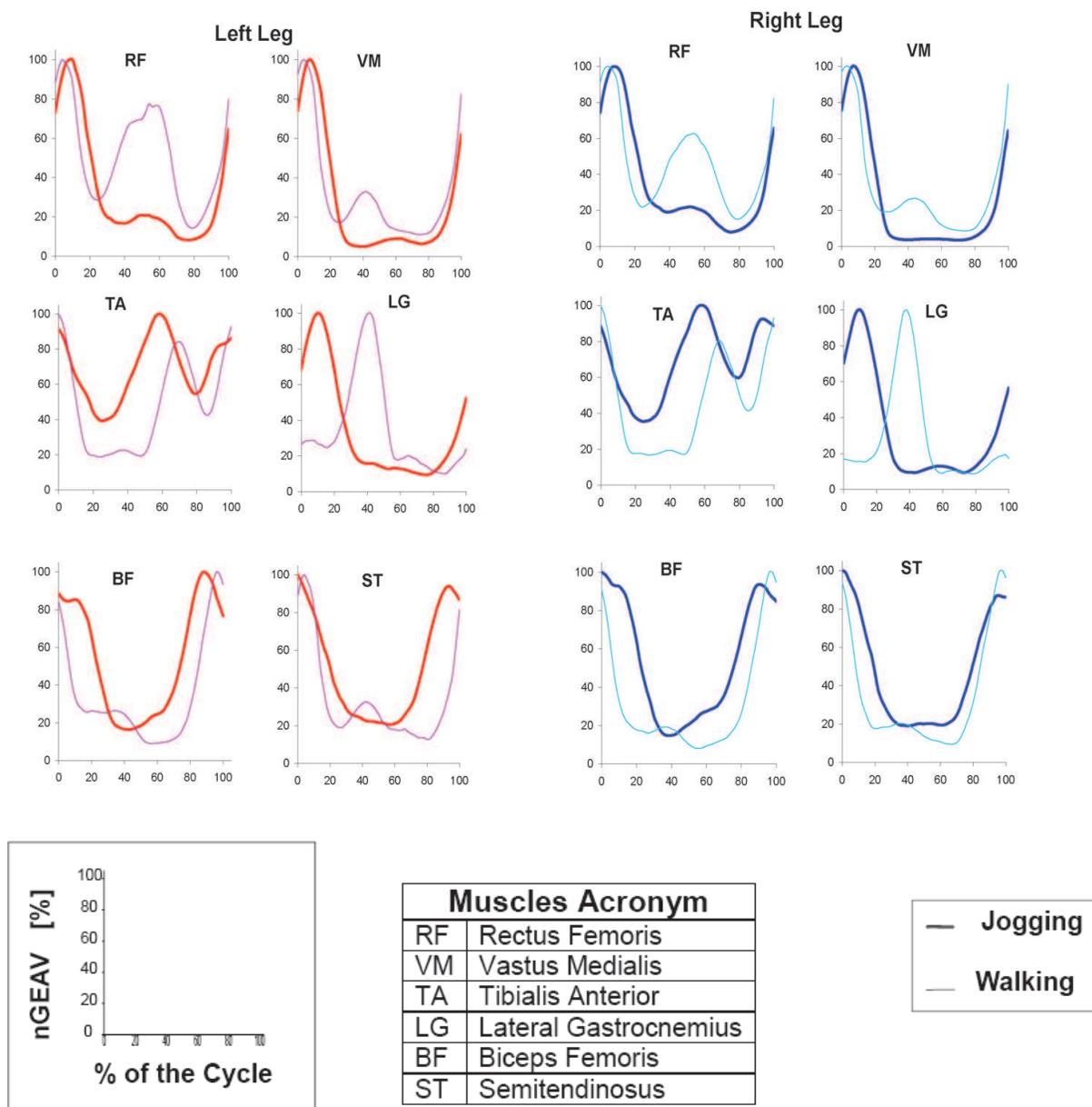
Fig. 1. Normalized grand ensemble average (nGEAV) for left and right leg muscles s-EMG activity. On each graph have been overlapped the activity profile of each muscle in walking (thin line) and in jogging (thick line). It could be appreciated how the peak of each muscle shifted in time in jogging with respect to walking. There was a similar pattern (nearly identical) between the nGEAV of both legs.

Table 4 shows the average of EAV maximum, minimum, and mean amplitudes from ensemble average of all subjects, and their respective standard deviation. Across the different subjects there was a wide range of peak amplitudes as can be seen from the high standard deviation values, which were bigger during jogging. In both types of locomotion, each muscle showed a characteristic level of maximal amplitude. In particular, the peak of the RF was smaller than that of the TA; even though the peak of electrical activity of both muscles was different between subjects. It can be also noted that the TA muscle is always active during jogging, as its minimum values are quite high: 0.39mV and 0.36 mV, which are

similar to the average activation levels of TA during walking (0.47mV and 0.51mV). As a matter of fact, 0.39mV and 0.36 mV represent the 40% of its maximal amplitude during jogging cycle.

Muscle	Walking	Jogging
Rectus Femoris	0.96	1.00
Vastus Medialis	0.99	1.00
Tibialis Anterior	1.00	0.96
Biceps Femoris	0.99	0.98
Semitendinosus	0.64	0.99
Lateral Gastrocnemius	0.95	0.99

Table 3. Correlation coefficients between the s-EMG activity grand ensemble averages (GEAV) of homologous muscles

Leg	Muscles	Jogging [mV]						Walking [mV]					
		max	SD	min	SD	mean	SD	max	SD	min	SD	mean	SD
Left Leg	Rectus Femoris	0.52	0.33	0.04	0.02	0.18	0.12	0.16	0.04	0.02	0.01	0.07	0.02
	Vastus Medialis	1.67	0.47	0.07	0.02	0.46	0.13	0.40	0.17	0.04	0.00	0.13	0.05
	Tibialis Anterior	1.50	0.22	0.39	0.19	0.88	0.13	1.01	0.18	0.13	0.04	0.47	0.05
	Biceps Femoris	1.07	0.47	0.12	0.06	0.52	0.19	0.60	0.38	0.04	0.01	0.20	0.09
	Semitendinosus	0.83	0.49	0.11	0.05	0.38	0.18	0.39	0.15	0.04	0.00	0.13	0.05
	Lateral Gastroc.	1.47	0.66	0.12	0.10	0.50	0.25	0.62	0.24	0.05	0.02	0.20	0.07
Right Leg	Rectus Femoris	0.60	0.30	0.04	0.02	0.21	0.09	0.19	0.08	0.02	0.01	0.08	0.03
	Vastus Medialis	1.77	0.37	0.05	0.01	0.46	0.10	0.44	0.19	0.04	0.00	0.14	0.05
	Tibialis Anterior	1.35	0.24	0.36	0.15	0.81	0.17	1.12	0.31	0.16	0.05	0.51	0.10
	Biceps Femoris	1.11	0.62	0.12	0.05	0.52	0.23	0.61	0.43	0.04	0.02	0.19	0.11
	Semitendinosus	1.03	0.66	0.13	0.09	0.46	0.26	0.54	0.40	0.04	0.01	0.18	0.11
	Lateral Gastroc.	1.42	0.59	0.09	0.03	0.47	0.18	0.76	0.34	0.05	0.02	0.21	0.08

Table 4. Walking and jogging: average value of maximal (max), minimal (min), and mean amplitudes from the ensemble average (EAV) of all subjects, and their respective standard deviation (SD)

It is worth noting that, within each mode of locomotion, the amplitude peak was very similar for homologous muscles (see Figure 2). These findings complement the findings of Arsenault et al. (1986b); and Ounpuu & Winter (1989).
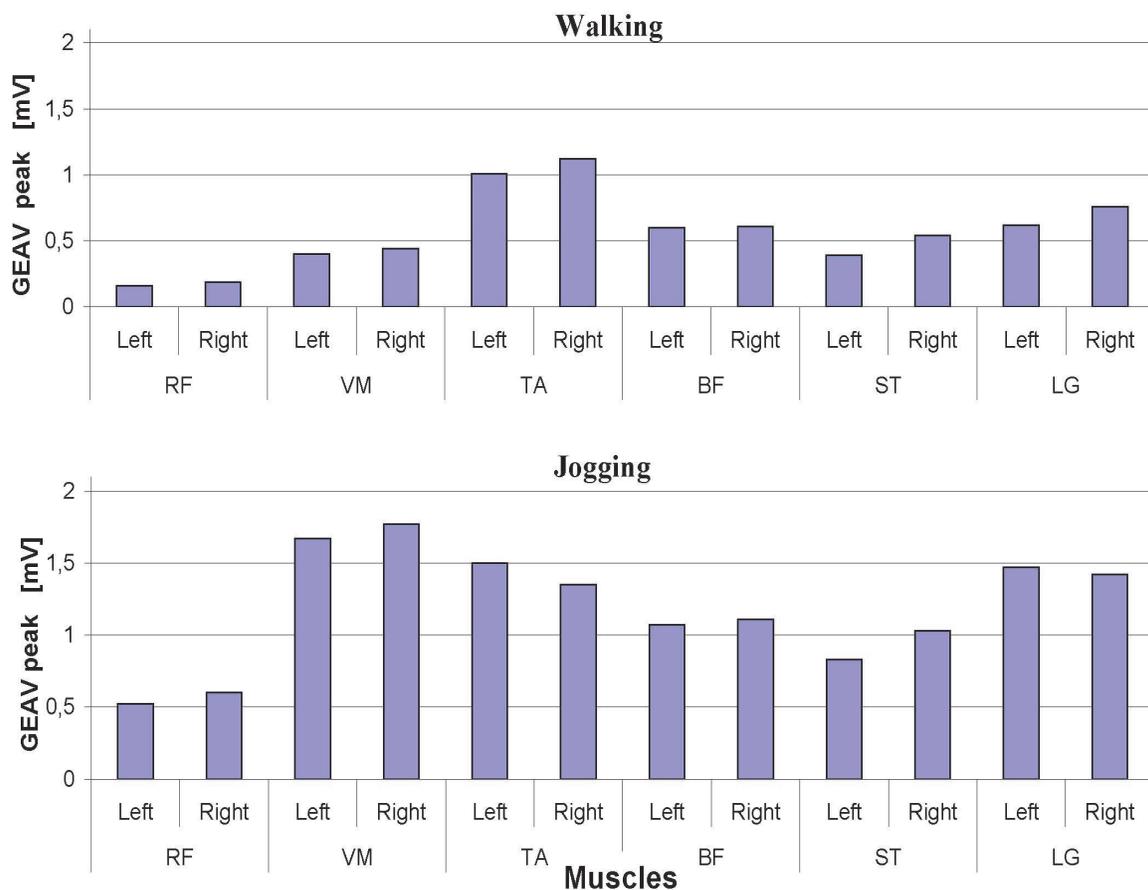

Fig. 2. Peak values of the grand ensemble average (GEAV) for each of the studied muscles for each leg during walking and during jogging. The amplitude peak was very similar for homologous muscles in the two legs for both ways of locomotion.

During jogging, all muscles increased significantly their maximal activity ($p < 0.05$). It can be seen from peak amplitudes that in walking distal muscles were more active than proximal muscles, being TA the muscle showing the biggest peak, and that in jogging it was the proximal muscle VM the most active. The higher sustained level of activity found in the TA agrees with the results of Reber et al. (1993), and indicates that the TA is more susceptible to fatigue and to related injuries.

The electrodes have been left placed on the same locations for each subject while walking than while jogging; therefore, as there was no other changes, the differences observed in the amplitude of the EMG signals of a muscle are due to a different participation of that muscle on the specific type of locomotion.

We have used the absolute (not normalized) signal to precisely quantify the participation of each muscle in each type of locomotion. If the signal was normalized, those changes would have been not detected (as shown in Figure 1), because the bigger EMG signal amplitude obtained while jogging would have been divided by a bigger peak value. However, the use

of the non-normalized values reveals the adaptability of the neuromuscular system against more demanding mechanical needs during jogging.

The recordings of each subject have been averaged because locomotion is a complex act that involves the integration of several variables during its learning process; hence, each subject has its own personal way of walking. Therefore, in order to increase our understanding of human locomotion it is necessary to use average values of the muscular activity, so that inter-subject differences can be diminished, as suggested by Inman et al. (1981).

3.4 Peak amplitude of myoelectric activity and relative increment

In order to find the activity increment during jogging in comparison with walking, the activity peak during jogging was normalized with respect to the peak found during walking.

Table 5 shows the increase of the maximum amplitude expressed in absolute values and in percentage of the maximum amplitude registered during walking. In both legs, the muscles that showed the biggest relative activity increment were the VM and the RF. The minimum relative increase occurred in the TA muscle.

It can be seen that there were different increases of activity of the muscles belonging to the same muscular group. This finding is in line with Gazendam & Hof (2007), who found that the relationship between speed and muscular effort was generally different between muscles of the same group.

The activity peak of muscles belonging to the same group could be very different (RF and VM) or quite similar (BF and ST). In all the cases, the difference in the peak intensity was bigger in jogging (see Table 4), which is in agreement with the results of Gazendam & Hof (2007); and of Montgomery et al. (1994).

The muscles belonging to the same muscular groups RF and VM are innervated by muscular branches of the crural nerve (L2-L4). The BF and ST muscles are innervated by the muscular branches of the sciatic nerve (L4-S2); however, for both legs, the peak and the increase of activity peak between walking and jogging of those muscles belonging to the same group was different. In particular (see Figure 2 and Table 5), the BF showed more activity than the ST in walking, and showed less activity increase in jogging than the ST, even though the BF was more active. The VM showed a higher activity in walking than RF and its activity increased more in jogging than RF. This difference in the activity increment may be due to: 1) their architectural characteristics: mass, muscle fiber length, transversal section area, and pennation angle (Wickiewiz, 1983); 2) the functions they perform in the three-dimensional space; and 3) the amount of joints that they cross. For example, the bigger s-EMG amplitude peak found in the VM in the support phase might be due to its mono-articular and unipennate nature. In addition, this muscle has the roles of: 1) preventing the genu valgum (Perry, 1992), the occurrence of which is more probable in jogging at the beginning of the support phase due to the concomitant dorsal flexion; 2) opposing the flexor action of the LG, which was active and presented, in some subjects, its activity peak at the same time than that of the VM; and 3) controlling the external rotation produced by the activation of the BF. The lesser increment in the peak of RF could be explained by the fact that it is biarticular and bipennate (which facilitates the force production; Gans, 1982). In both walking and jogging, the RF and VM muscles showed the maximum peak at the beginning of the support phase, when there was an extension movement of the hip and a simultaneous flexion of the knee (Mann & Hagy, 1980). If the RF showed high activity during the hip extension, it would oppose that hip movement.

From the mean peak amplitudes it can be seen that, for both legs, during walking the proximal muscles (VM and RF) showed an amplitude peak smaller than that of the distal muscles. When jogging, the proximal muscles showed greater relative increase than the other muscles, which can be due to the importance of the knee for both the movement and the stability. The VM was the muscle showing the bigger peak amplitude; namely, its average peak level was 4 times bigger during jogging (1.67mV and 1.77 mV) than during walking (0.40 mV and 0.44mV).

RF and VM muscles have an important role in the absorption of the mechanical shock resulting from the impact of the heel against the floor. In jogging, the ground reaction force (GRF) is bigger than in walking, and in addition, that shock has to be absorbed in a shorter time, as the contact with the floor passes from being 0,60s to just 0,26s (a reduction to less than a half). Furthermore, the repetition of wrong alignments of the lower limbs can facilitate the production of lesions.

This result complements the studies of Gazendam & Hof (2007); and of Montgomery et al. (1994), and provides support to the results of Taunton et al. (1988) on the prevalence of the patella femoral pain syndrome and of the patellar tendinitis in athletes.

Of the distal muscles, the LG absolute and relative activity during jogging increased more than that of the TA, showing both muscles a similar maximal peak during jogging. Mizrahi et al. (2000b) demonstrated that with the progressing of fatigue in long-distance running, an imbalance in the activities between the ankle plantar and dorsi flexor muscles develops and the muscles that span the tensile surface of the bone becomes less active than those of the opposite side, the result is a decrease in the protection abilities of the muscles.

Leg	Muscle	peak [mV]		Increase [mV]	Increase [%]
		Walking	Jogging		
Left Leg	Rectus Femoris	0.16	0.52	0.37	233
	Vastus Medialis	0.40	1.67	1.27	314
	Tibialis Anterior	1.01	1.50	0.49	49
	Biceps Femoris	0.60	1.07	0.47	79
	Semitendinosus	0.39	0.83	0.44	112
	Lateral Gastrocnemius	0.62	1.47	0.85	137
Right Leg	Rectus Femoris	0.19	0.60	0.41	216
	Vastus Medialis	0.44	1.77	1.33	302
	Tibialis Anterior	1.12	1.35	0.22	20
	Biceps Femoris	0.61	1.11	0.50	81
	Semitendinosus	0.54	1.03	0.49	90
	Lateral Gastrocnemius	0.76	1.42	0.66	86

Table 5. Increase of the s-EMG amplitude peak expressed in absolute values and in percentages respect to the maximum amplitude of the walking cycle

3.5 Appearance time of the activity peak

Figure 1 shows how the peak of each muscle shifted in time in jogging with respect to walking.

Subject	Time of occurrence for the right leg [% of cycle]											
	RF		VM		BF		ST		TA		LG	
	JC	GC	JC	GC	JC	GC	JC	GC	JC	GC	JC	GC
s1	8→12	6	12	N/A	12→14	100	8	4	62	0	8→12	42
s2	8→12	4	10	4	14	0	12	4	94	0	14	40
s3	6→8	46	8	4	84→86	90	0	6	4	0	8→10	38
s4	6→10	4	6→8	4	12	100	94	4	56	0	10	40
s5	8	2	8	2	88	96	2	2	0	0	10	38
s6	8	4	8	4	6	96	4	6	60	N/A	10	32
s7	8	6	8	6	12	100	4→8	4	60	0	12	36
s8	10	42	4	2	0	94	92	2	58	0	6	40
s9	6→10	62	6→8	4	86	98	88	4	56	70	8	36
s10	8	58	2	2	92	98	92	4	4	0	8	44

Subject	Time of occurrence for the left leg [% of cycle]											
	RF		VM		BF		ST		TA		LG	
	JC	GC	JC	GC	JC	GC	JC	GC	JC	GC	JC	GC
s1	16	6	N/A	N/A	90	96	2	96	60	0	10	40
s2	8	4	8	2	16	98	20	98	92	0	10	38
s3	10	56	6	2	14	0	0	98	92	0	8	36
s4	8	6	8	4	14	0	0	0	54	68	12	38
s5	8	2	6→8	0	0	96	2	96	94→98	0	8	34
s6	8	4	8	8	8→10	98	4	98	58→60	0	10	36
s7	12	6	8→10	6	86	98	2	100	58	68	12	36
s8	6	4	6	2	0	96	0	0	56	0	6	34
s9	8	56	8	2	14	36	88	96	58	68	8	36
s10	8	4	6	0	88	98	0	0	60	0	10	36

Table 6. Time of occurrence of the amplitude peak in the ensemble average (EAV) for the muscles of both legs, expressed in percentage of the cycle of walking (GC) and of jogging (JC). (N/A: Not Available)

Table 6 shows, for each subject and for each muscle, the time of EAV peak, which corresponds to the moment of maximum activity of each muscle. In this case, it would be misleading to use the GEAV, as it would yield average values that may actually not appear in any of the real signals. In walking and in jogging the time of appearance of the activity peak of RF, VM, and LG muscles was consistent and always occurred in eccentric contraction; in addition, there was a higher inter-subject variability in the time of occurrence of the IT and TA muscles peak activity.

3.5.1 Rectus femoris

During walking, the peak activity of the RF appeared at different times for each subject; in the majority of the subjects, the peak activity occurred at the loading phase (2-6%GC). In other subjects the peak appeared either at the end of the final stance phase (42%GC), at the pre-swinging phase (58%GC), or at the beginning of the oscillation phase (62%GC).
In jogging, for all the subjects, the peak appeared at the beginning of the support phase (6-12%JC), a bit later than in walking.

3.5.2 Vastus medialis

In walking, the VM muscle of both legs showed its highest activity peak in the loading phase (2-8%GC). In jogging, the peak appeared always at the beginning of stance phase (2-12%JC), but in some subjects it appeared later than in walking. In none of the subjects the peak appeared at the end of the loading phase because the knee extends during the propulsion; while during walking, the knee flexes.

3.5.3 Lateral gastrocnemius

Interestingly, the peak of LG occurred during unipodal support both in walking and in jogging, but in walking at the end and in jogging at the beginning of the unipodal support. The LG peak appeared in eccentric contraction, when it was acting as antagonist (Mann & Hagy, 1980; and Mann, 1982), while dorsal flexion is performed, increasing in this way its efficiency.
In jogging, its activity peak appeared at the same time than that of the VM. This study has confirmed the results of Gazendam et al. (2007), who indicated that in running quadriceps and calf muscles work together in absorbing and generating energy.

3.5.4 Tibialis anterior

The TA muscle presented two peaks. Its maximal amplitude peak in walking, in the greatest part of the subjects, occurred at the beginning of the support phase (0%GC), and –only in a few subjects- in the swinging phase (68-70%GC). In jogging, the peak appeared either at the loading phase (0%JC), in the swing phase (58-62%JC), or in the final flight (92-98%JC).
In any case, the time of appearance of the TA activity peak showed less variability in walking than during jogging. The role of the TA muscle during walking changes depending on the moment in which it becomes active: it acts as antagonist of the plantar flexion if active at the beginning of the support phase, or as agonist of the dorsal flexion if it is active at the beginning of the swinging phase. In jogging, when the TA activity peaked at the very beginning of the loading phase, it facilitated the forward movement of the tibia and hence eased the flexion of the knee. When it peaked in the swing phase and in the final fly, it facilitated the dorsal flexion. Therefore, in jogging, the peak of TA always occurred when it was acting as agonist.

3.5.5 Ischiotibial muscles

The peak of the ischiotibial (ITI) muscles (BF and ST) appeared in the same or in different phases (final non support or initial support). In walking, the maximal amplitude of the BF muscle appeared for nearly all the subjects at the end of the swinging phase (94-100%GC). The peak of the ST occurred in the loading phase (2-6%GC) for the left leg and at the end of the swinging phase (96-100%GC) for the right leg.

The role of the ITI muscles during the swinging phase is to stop the leg in order to ease its standing. Inman (1968) proposed that the deceleration of the oscillating leg by the ITI muscles can contribute even more to the forward movement of the body than the push of the ipsilateral leg.

In jogging, the occurrence of the BF and ST activity peak varied among subjects, appearing indistinctly at the beginning (BF, 0-14%JC; ST, 0-4%JC) or at the end (BF, 86-90%JC; ST, 88-94%JC) of the cycle.

The variability in the time of the peak appearance for the ST and BF muscles supports the theory that forces are optimally distributed between the several muscles that cross a joint (Chao & Rim, 1973; and Crowninshield, 1978).

3.6 Activity pattern of the studied muscles in relation to their mechanical functions during locomotion

The mechanical functions required for the locomotion are: 1) landing impact absorption, 2) dynamic stability, 3) propulsion, and 4) energy conservation (Perry, 1992). The execution of each function depends on a distinct motion pattern.

3.6.1 Landing impact absorption

Muscles amplitude peaks were higher in jogging that in walking and in five or in six (depending on the subject) of the muscles studied for each leg (VM, RF, BF, ST, and LG) the peak occurred mainly in the support phase. These results were as expected because the mechanical requirements for the absorption of the landing impact (the vertical component of the GRF is in the order of 1.5 times each subject's body weight), and the stability requirements (the support during jogging is only on one leg) are bigger in jogging than in walking.

Both in walking and in jogging, the RF and VM muscles showed their peak activity when they were activated eccentrically. The TA muscle controls the plantar flexion in walking, helping to absorb the shock of the landing. As the foot comes in contact with the ground during jogging, dorsiflexion of the ankle takes place. The other mechanism helping to absorb this impact is controlled pronation of foot, which provides flexibility within the foot.

For a deeper study about the influence of the mechanical impact on the lesions, see the works by Mizhrahi et al. (2000a) and (2000b).

3.6.2 Dynamic stability

The stability during walking is improved by the double support, when between two and five (depending on the subject) of the six muscles studied were active and showed their activity peak (TA, VM, and/or RF, and/or BF, and/or ST).

In jogging, the weight is supported when only one leg is on the floor, while the six studied muscle were active. The peak of the VM and of the LG muscles occurred simultaneously; they play an important role in the stabilization of the knee, which is more flexed at the beginning of the support phase in jogging (Man & Hagy, 1980).

The peak of the ITI muscles (flexor muscles of the knee) could also appear at the beginning of the support phase. The co-contraction of VM and BF balances the knee in both sagittal and transversal planes.

In jogging, the support starts with a dorsal flexion (Man & Hagy, 1980); the co-contraction of TA (acting as agonist) and LG (acting as antagonist), facilitates the stability of the foot, which is decreased by the fast changing from supination to pronation (Cavanagh 1987).

Our results, which are in agreement with those of Prilutsky et al. (1998), show that the coordination of the tight muscles does not depend on the type of locomotion, as they are co-activated during both walking and jogging. However, the coordination of the leg muscles is different in walking than in jogging; for example, the TA and LG muscles are co-activated only during jogging.

The co-contraction existing in both ways of locomotion facilitates the movement of the segments, as it has been already established by Falconer & Winter (1985).

3.6.3 Energy conservation, progression, and propulsion

In our study, during jogging, the average speed increment was of 87% respect to walking. In addition, the peak of muscular activity was significantly different (Table 5); but not the peak time, except for the LG (Table 6), the activity of which started earlier in the non-support phase and it was present during the biggest part of the support phase (Figure 1).

It is worth noting that 70% of the energy produced by the muscle is lost as heat (Astrand & Rodhahl, 1980). Furthermore, in jogging the fluctuations of kinetic and gravitational energies are in phase, which produces bigger changes in their sum during stride (McMahon, 1990). In the other hand, in walking the transformation of kinetic energy into potential energy, and vice versa, decreases the work needed to be carried out by the muscles (Eberhart et al., 1954),

The co-contraction of the antagonist muscles (Table 6) plays an important role in the accumulation of elastic energy in the muscles and tendons. During jogging, the wider movements of the joints (Mann 1982) allow active muscles to absorb more energy, which is stored in form of elastic energy. Due to a higher muscular activity (Table 5), there is a bigger amount of elastic energy stored; and, in addition, the speed of the movements facilitates its recovery more efficiently. That energy is released to progress and propel the body up and forward. This result is in line with the study of Cavagna et al. (1964) who established that during jogging tendons elasticity contributes to the 50% of the performed work.

Four out of the six muscles studied are biarticular (RF, BF, ST, and LG) and they perform the energy transfer from one segment to the other (Van Ingen et al., 1987). That is, when a biarticular muscle acts as agonist over one joint and as antagonist over the other joint, the energy accumulated when it operates as antagonist is transferred from the segment to which it operates as antagonist to the segment in which it operates as agonist.

The higher activity showed by the ITI muscles during jogging (Table 4) in the non-support phase (Table 6) decreased the speed of the contralateral leg in the swinging phase and allows the bigger transfer of momentum between the contralateral and ipsilateral legs facilitating the propulsion of the support leg, finding congruent with that of Inman (1968) in walking.

3.7 Injuries in jogging

The comparison carried out in this study between the EMG activity of the leg muscles during walking and during jogging reveals the adaptability of the neuromuscular system against more demanding mechanical needs, and in this way we can learn more about the etiopathogeny of lesions.

An insight on EMG motor-control strategies of individuals without lesions is that the peak activity of the RF, VM, and LG muscles occurs in the eccentric phase. The probability of lesions is higher in eccentric contractions (Lieber, 1992); in addition, as the muscular activity is much higher in jogging, the probability of lesions increases even more. This could be one of the reasons why the musculotendinous lesions are so frequent in runners. Besides, the activity of those muscles occurs when the ITI and TA muscles are highly active; that is, there is a co-contraction of the muscles that cross the hip, knee, and ankle joints. In the case of the knee, the higher activity corresponds to the antagonist muscles. In addition, the higher speed of the movement performed during jogging increases as well the risk of lesions. The continuous activity of the TA at a level greater that 40% of its peak could make it susceptible to fatigue and thus of muscle injury.

Our working hypothesis about the bigger incidence of lesions during jogging is focused on muscular mechanics. However, Mizrhahi et al. (2000a) studied the role of fatigue in the etiology of lesions, and found out that the change due to fatigue exposes the shank to substantially higher impact accelerations, hence increasing the risk of overload injuries. In another study, Mizrhahi et al. (2000b) found that the fatigue-related imbalance in the contraction of the shank muscles develops in parallel to an increase in shank shock acceleration, increasing impact loading. The combination of these two conditions may hamper the loading balance on the tibia and higher risk of stress injury.

In any case, in the conditions in which this study has been carried out (average speeds: walking, 1.33±0.12m/s, and jogging 2.50±0.31m/s; and one-minute break between each recording), fatigue is not probable to appear.

4. Conclusions

The most relevant findings of this work are: (1) In both types of locomotion, each muscle showed a characteristic level of maximal amplitude. (2) In addition, the LE amplitude was very similar for homologous muscles in both legs in walking and in jogging.

Against the more demanding mechanical needs of jogging (when the average velocity increased 87%), the observed adaptations of the neuromuscular system were: 1) significantly bigger activity peak respect to walking of the six studied muscles (the highest peak activity increment was found in VM), which were active also for longer periods of time, even though the support phase was shorter and the non-support phase longer; 2) co-contraction in the three joints of the leg of agonist and antagonist muscles belonging to the same or different segments (*e.g.*, VM and LG); 3) continuous activity of TA; 4) consistent time of appearance of the peak activity during the support phase in the so-called anti-gravitational muscles (RF, VM, and LG), and more variable in the ITI and TA muscles; 5) bigger activity in eccentric contraction of the RF, VM, and LG muscles; and 6) bigger absolute inter-subject variability in the muscle activity for both distal and proximal muscles.

The abovementioned points (1) to (5) could explain the musculotendinous lesions so frequent in runners and provide the basis for the design of specific exercises to prevent those lesions.

5. Future research

As the s-EMG amplitudes obtained from homologous muscles for each subject have been similar, one could be tempted to perform this type of studies on only one leg. However, in this work we have not carried out postural balance studies, which are necessary in order to ensure whether the segment alignment has an influence on muscle activity. That is, it is necessary to

study, for example, lumbar lordosis with weakened abdominal musculature, alignment of the knee, heel, foot, configuration of the forefoot, femoral ante- or retro-version, and symmetry of legs length , as all those factors have an influence on muscle activity.

As future research, the study of the muscular activity and coordination of injured athletes could provide a better insight on the neuromuscular system adaptations during their recovery. The information so gathered could be helpful when designing rehabilitation treatments and for the foreseeing of the probable outcome of a lesion.

In jogging, the frequency of lesions in VM and LG can be due to their higher activity. The analytic training of those muscles could benefit athletes. That analytic training will take into account the contractile and elastic components of the muscles, the intensity of the contraction and of the co-contraction of agonist and antagonist muscles during jogging. The information thus collected from athletes in recovery will bring useful insight into how to avoid lesions.

Long-term studies on several athletes could provide important information about which kind of styles, muscular activation patterns, etc. are more prompt to produce future lesions, so the athletes can train to avoid them.

6. Acknowledgment

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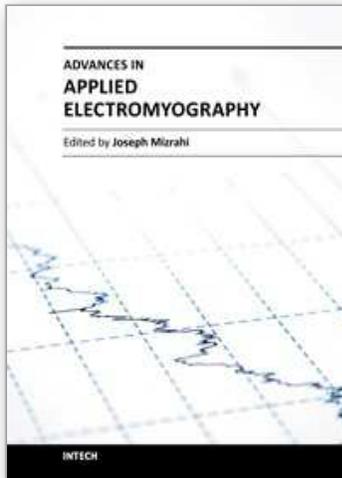
Advances in Applied Electromyography

Edited by Prof. Joseph Mizrahi

ISBN 978-953-307-382-8

Hard cover, 212 pages

Publisher InTech

Published online 29, August, 2011

Published in print edition August, 2011

The electrical activity of the muscles, as measured by means of electromyography (EMG), is a major expression of muscle contraction. This book aims at providing an updated overview of the recent developments in electromyography from diverse aspects and various applications in clinical and experimental research. It consists of ten chapters arranged in four sections. The first section deals with EMG signals from skeletal muscles and their significance in assessing biomechanical and physiologic function and in applications in neuro-musculo-skeletal rehabilitation. The second section addresses methodologies for the treatment of the signal itself: noise removal and pattern recognition for the activation of artificial limbs. The third section deals with utilizing the EMG signals for inferring on the mechanical action of the muscle, such as force, e.g., pinching force in humans or sucking pressure in the cibarial pump during feeding of the hematophagous hemiptera insect. The fourth and last section deals with the clinical role of electromyograms in studying the pelvic floor muscle function.

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