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Comparison by EMG of Running Barefoot and Running Shod

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1. Introduction

Footwear is the "interface" between the locomotive system and the athlete's physical environment through which all the forces that act and react between the legs and the ground are transmitted.

According to Nigg (1986), the choice of footwear is based on the price, durability, comfort, colour, safety, weight, and performance; but, how footwear affects the human-ground interaction?

During jogging, running shoes have to reduce foot shock impact with the ground, to hold the foot, to control the pronation-supination, to direct the force at the time of propulsion for takeoff (Luethi & Stacoff 1987), and to increase efficiency.

Nowadays, more and more people practice sport, being jogging an activity common to many sports, besides being a sport specialty itself. Jogging is a complex activity requiring an exact timing of muscle activation and a precise control of movement. Many injuries occurred during jogging affect the musculoskeletal system: (1) tendonitis of: tibialis anterior, peroneus brevis, tibialis posterior, quadriceps, and Achilles tendon; (2) calcaneal apophysitis; (3) chronic syndromes: anterior and posterior compartments; (4) stress fractures; (5) plantar fasciitis; and (6) rupture of the hamstrings (Reber et al. 1993). Certain types of contractions may predispose the runner to a particular injury (Vaughan 1984).

The displacement speed is determined by the frequency and length of the stride. These variables are the result of the successful integration of many mechanical and neuromuscular processes.

The human body is a biological system that has many possibilities of action and reaction to external environmental influences. During the movement, the neuromuscular system is

involved not only in the production of force to move the segments, but through different mechanisms sensory feedback is capable of reacting to small changes. Facilitation or reduction of these feedback mechanisms enable the human motor system for a wide variety of functions, such as: (1) control of the position and stiffness of joints, (2) shock absorption, (3) dynamic stability during the support, and (4) propulsion, facilitating that the involved muscles perform with suitable elastic and contractile characteristics (Gollhofer and Komi 1987).

During one cycle of jogging, lower extremities undergoes a phase of unipodal support (in which only of the feet is in contact with the ground), one of swinging, and two phases of flight (in which none of the feet are in contact with the ground): the first takes place before the swing phase and the second, after it. Support phase lasts less than 50% of the stride. Slocum & James (1968) divided the jogging stride in the following phases:

1. Support (or stand) phase: begins when one foot contacts the ground and ends when the first finger of the same foot is no longer in contact with the ground. This phase can be subdivided into three: (1.a) impact phase; (1.b) phase of medium support or of absorption: time when the whole foot is resting on the ground; and (1.c) propulsion or push phase, which begins when the foot is lifted off the ground and ends when the toes leave the ground.
2. Phase of no support or of recovery, which comprising three phases: (2.a) initial flight phase: begins when the first toe of the support foot is no longer in contact with the ground and ends when the heel of the contralateral foot (opposite one) touches the ground; (2.b) half swing phase: begins when the heel of the contralateral foot contacts the ground and ends when its first finger is no longer in contact with the ground; and (2.c) final flight phase: begins when the first finger of the opposite foot is no longer in contact with the ground and ends when the heel of the ipsilateral foot rests again on the floor.

Another way of dividing the non-phase support is as follows: period of follow-through (after leg takeoff, the hip stretches); forward period (the ipsilateral leg moves forward while the hip is flexed); and period of descend of the foot.

1.1. Movement of the joints

The displacement of the centre of gravity is due to the angular movement of the joints caused by the resultant of different forces: muscular force (caused by the neuromuscular system), ground reaction force, weight of the segments, misalignment of body weight, and the inertia of the moving segments. During jogging, the path of body's centre of gravity is sinusoidal, moving twice in the vertical direction, so there are two peaks for each stride. At the same time, when the centre of gravity loses height, it loses also horizontal speed, and the kinetic and potential energies are in phase, so large changes occur in the resultant of both forms of energy at each step. However, a significant amount of mechanical energy is conserved stored as potential elastic energy in the tissues. Another mechanism to save energy is its transferring between segments by two-joint muscles.

In jogging, the movements of the joints are larger in the sagittal (or anteroposterior) plane, even though the movements in the coronal and transverse planes facilitate the stability and progress in the sagittal plane, respectively. The movements are as follows:

1. Sagittal plane: the axis of rotation is medial-lateral and the movements are of flexion-extension.
 - 1.a. In the ankle: *support phase* –dorsiflexion (10°), plantar flexion (45°); *phase of non support* –dorsiflexion (45°).
 - 1.b. In the knee: *support phase* –flexion (20°), extension (18°); *phase of non support* – flexion (90°), extension (90°).
 - 1.c. In the hip: *support phase* –extension (35°); *phase of non support* – flexion (40°), extension (10°) (Milliron & Cavanagh, 1990; Mann et al. 1986).
 - 1.d. The pelvis has a rocking motion, respect to the anterior-superior iliac spine: forward or down and back or up. In the stance phase there are up and down movements; and in the no-support phase, down, up, and down.
2. Transverse plane: the lower limb segments rotate around a vertical axis, the direction of rotation is outward or inward and its development is the same in the components of the lower extremity: pelvis, thigh, and leg, but their magnitudes vary. The swing phase is characterized by the internal rotation of each segment; the more distal segments rotate faster and in a greater degree than the proximal ones. The average rotation of the pelvis is 5° and that of the tibia and femur 9° , summing up a total of 23° . In support phase the external rotation continues and then, halfway, its direction changes.
3. Coronal or frontal plane: the axis of rotation is anterior-posterior and the movements are of ab-adduction. The displacements of the pelvis and lower extremity in the frontal plane are not large but are very important to maintain balance. In the knee and ankle the movements are ab-adduction and are limited by the characteristics of the joints and the presence of lateral ligaments. During jogging, the hip is adduced during the support phase until half of it. In the swing phase occurs continuous abduction of the supporting leg and at half of the way, it changes its direction to adduction.
4. The movements of the ankle with the tibiofibular and subtalar joints produce the movement of supination-pronation.

1.2. Anatomy of the joints

From a biomechanical perspective, the factors that dictate the movement are our inherent structure and alignment, the joint range of motion, and the muscle strength available. The joint range is partly defined by the anatomical structure. In the following, the peculiarities of the joints of the legs and their angular displacements during locomotion are described (Testut 1971, Inman 1981, Perry 1992, Behnke 2001).

Each of the lower extremities is a system of articulated segments, with its own mechanical characteristics. The different joints involved are: (1) lumbosacral, (2) the two hips, (3) the two knees, (4) the two ankles, (5) the subtalar joints, and (6) the midtarsal joints. In studies on locomotion the foot is considered as a rigid segment (although it is formed by 26 bones) serving for the transmission of force between the body and the ground. During the movement, the body segments serve as levers (Perry 1992).

According to Arsenault et al. (1987) it seems clear that the kinematics of locomotion does not show high variability. From the data found in the literature, we describe the angular

movements of different segments: pelvis, thigh, leg, and foot of the lower extremity in three planes: sagittal, coronal, and transverse during the different phases of the cycle locomotion.

In the sagittal plane, the movements are wider. In the other two planes the movements are small but are involved in the magnitude of the displacement of the center of gravity in the sagittal plane and also provide stability.

1.2.1. Ankle

The ankle is the only anatomical area where the vertical forces are transmitted to the horizontal support system, in this case the foot. The ankle includes the tibiofibular and the subtalar joints. The tibiofibular joint allows the movements of dorsiflexion and plantar flexion or extension in the sagittal plane. The lateral movement of the foot (or eversion) around the anterior-posterior and the medial movement (or inversion) are made in the subtalar joint in the frontal plane. Adduction and abduction movements occur around the vertical axis and the transverse plane. The combination of the movements of the tarsus with those of the ankle allows complex movements, such as: (1) supination, or inversion of the ankle with adduction of the foot and plantar flexion, and (2) pronation, or ankle eversion with abduction of the foot and dorsiflexion.

1.2.2. Knee

The knee is the binding site of two long bones, femur and tibia, which are the major body segments. Small range of motion produces significant changes in the foot or in the body.

The knee joint is very complex, bicondylar, characterized by a very wide range of motion in the sagittal plane and small arcs of motion in the coronal and transverse planes. During the movement of the knee in the sagittal plane, the tibia slides around the distal end of the femur so that the mediolateral axis of rotation displaces with movement.

The patella is the largest sesamoid bone in the human body. It modifies the thrust angle of the quadriceps femoris, affecting the production of muscle force components, so that the rotational component is greater (Nordin & Frankel, 2004).

During support, the knee is essential for the stability of the leg. During swinging, the flexibility of the knee is the main factor that determines the progress of the leg. The number of bicondylar muscles involved in controlling the knee indicates a great functional coordination with the hip and ankle.

The movement in the sagittal plane is used for progression through the support and during swing. During the phase of non support, the knee makes use of a range of movement widest than that of the any other joint. The rotation in the transverse plane accommodates changes in alignment when the body oscillates back and forth of the supporting leg. In walking, when the knee extends, the leg rotates externally; when the knee is flexed, the leg rotates internally. In jogging, the knee is flexed at the beginning of the stance phase while the leg is externally rotated (Novacheck 1998).

The movement in the coronal plane facilitates vertical balance on the leg, particularly during leg stance. In each cycle the knee moves into abduction and adduction. In support phase, the movement is of abduction. During oscillation, the knee returns to a more neutral position in adduction. In the transverse plane of a position of maximum external rotation at the end of the stance phase, the lower limb starts an internal rotation in the take-off of the leg and continues during the oscillation and the load response (initial part of the support phase).

1.2.3. Hip

The hip function differs from the other two joints in the following respects: (1) represents the junction between the passenger and the motor system, (2) allows movements in three planes of space with a specific control in each plane, although in the coronal plane movement is limited, but the mechanical demands are substantial.

In the sagittal plane hip extends in the phase of support and flexes in the non-support one. The hip has small arcs of motion in adduction and in abduction. At the initial contact of the heel with the ground, the hip is in adducted position. At the beginning of the swing phase, the hip is in a relative abduction of 5°. In the transverse plane, the internal rotation peak occurs at the end of the loading phase and maximum external rotation occurs at the end of the pre-swing phase.

1.2.4. Pelvis

During the stride, the pelvis moves in three directions asynchronously. The point of support is the hip of the leg that is in support. All its ranges of motion are small: in the sagittal plane it is a rocking motion of 4°; in the frontal plane, 7°; and in the transverse plane, 10°.

1.3. Differences in locomotion due to shoes

De Wit et al. (2000) describe different angular displacements of the knee and ankle when subjects ran barefoot and when running shod. During running, the body reacts to the external environment which produces the ground reaction force (GRF) that occurs in response to the force action transmitted by the leg in contact with the ground. The GRF reflects the net effect of the muscle action and the accelerations of the segments while the foot is in contact with the ground (Martin & Morgan 1992). All segments contribute to the total acceleration of the body in proportion to the acceleration of its centre of gravity and its relative mass.

The three components of the GRF (vertical, anterior-posterior, and medial-lateral) change their size when using footwear (Nigg 1983). The GRF reflects the acceleration and deceleration of the centre of gravity. The gravity eases the contact of the foot with the ground.

1.3.1. Vertical component (GRF_v)

During movement, the GRF_v varies above and below the body weight due to the positive and negative accelerations undergone by the body. The difference between the vertical

component and the body weight is due to acceleration of the body. The direction of displacement of centre of gravity and acceleration influences also the magnitude of the vertical component.

The vertical component is biphasic and has a first peak of early impact (at 20ms after the impact), representing between 140% and 160% (and up to 200%) of body weight (BW), in the runners who touch the ground first with the heel, and a second peak in the stance phase, which appears at 80 ms and can almost triple the body weight. The two peaks have different slopes, the first very fast and the second more gradual. Contact time is about 0.25 s. The first peak is associated with heel strike and indicates pronation. It is surprising that this peak is smaller in magnitude than the second peak that is associated with the propulsion.

1.3.2. Antero-posterior component (GRF a-p)

When the foot contacts the ground, it is pushed forward and suffers a reaction force that slows it down. At the time the body passes over the foot that is resting on the ground, the horizontal component is zero. When the body moves over the foot which is resting, the foot is pushed against the ground and the antero-posterior component becomes positive, facilitating the forward propulsion. Its magnitude represents 50% of the BW during jogging.

1.3.3. Medial-lateral component (GRF m-l)

The medial-lateral component is the smallest of all components. It has two polarities, the first in reaction to the force transmitted by the foot on the medial direction, and the second in reaction to the force transmitted by the foot in the lateral direction. The polarity of this component of one leg is opposite to the polarity of this component in the contralateral leg. Thus the sign of the lateral component of the right foot would be first positive and then negative, and for the left foot would be reversed. The variation in magnitude of the vertical, antero-posterior, and medial-lateral components means that during the displacement, the speed is not constant, as the body moves faster in one point and slower in another.

The parameters related to the vertical component are the peak impact and the rate of increase of force (obtained with a force plate), and were used to examine the load under which the locomotor system is during locomotion (Nigg 1983). One of the main functions of the footwear is to cushion the strength of the action the subject exerts on the ground and to absorb the reaction force in order to protect the musculoskeletal system. In jogging, running shoes nullifies the impact peak and is involved in delaying the onset of the support vertical force by changing the gradient loading (Nigg 1983, De Wit et al. 2000). The anteroposterior force, which has two phases (braking and propulsion), is influenced by the friction introduced by the shoe sole. The medial-lateral component that guides us on the pronosupination movement can be modified by the shoes as they change the distance between the point of application of the GRF and the subtalar or calcaneo-talar joint.

Wakeling et al. (2001) speculated that the muscle activation levels in the lower extremities are adjusted depending on the loading speed of impact forces. Nigg & Wakeling (2001)

suggested that the repetitive impact forces during physical activities are not responsible for possible injury but are the cause of changes in myoelectric activity (activation time and amplitude), and these changes are responsible for the injury. Gollhofer & Komi (1987) found differences in the electrical activity of muscles when subjects ran barefoot first, then shod; Gavilanes & Goiriena-de-Gandarias (2004) found changes in myoelectric activity throughout the gait cycle when the subjects walked barefoot or with two different types of footwear, with no differences due to types of footwear used. Wakeling et al. (2002) found that the muscle activity concomitant to the impact can be altered by changing the hardness of the shoe. Frederick (1986) concluded that footwear induces adjustments in the movement of the legs, which in turn have secondary effects on the kinetics.

As found in the literature, footwear induces adaptations in the motion of the joints of the lower extremities, changes in the reaction force, and modifications in the myoelectric activity. When designing and making shoes, different types of considerations are taken into account: (1) reduce excessive burden, by absorbing the impact, (2) improve the dynamic stability, (3) increase the performance, and (4) feel comfortable (Ramiro et al. 1988, Segesser & Nigg 1993). The impact absorption is carried out through the midsole of sport shoes that acts as a filter by changing the impact forces (Luethi & Stacoff 1987). The increase in performance with the use of the shoe has been an argument used by athletes and shoe manufacturers. The midsole of athletic shoes is a layer of resilient, deformable material that is interposed between the upper shoe and the outside. The main function is to provide a protective layer between the foot and the ground and soften the shock of impact. During the first stage of the stride of jogging, the midsole is compressed by the pressure of the foot and the forces acting on it do some work on the viscoelastic material of the sole. Part of this work becomes stored as deformation energy in the material (elastically deformed). When the load on the midsole is reduced, the material undergoes an elastic recovery to its original shape (Shorteen 1993).

The literature review reveals as well that the ability of sport shoes to mitigate the impact forces between the ground and the body has been examined by different researchers (Denoth et al. 1981, Bates et al. 1983, Nigg et al. 1986, Gollhofer & Komi 1987, Dufek JS et al. 1991, Forner et al. 1995, De Wit et al. 2000). Less studied are the effects of footwear on kinematics (Frederick 1986, Nigg et al. 1986), or muscle activity (Gollhofer & Komi 1987, Wakeling et al. 2001, Nigg & Wakeling 2001, Wakeling et al. 2002, Gavilanes & Goiriena de Gandarias 2004).

The design of sport shoes and the elasticity of the materials used in their sole influence, respectively, the location of the application point of the GRF and its magnitude. These parameters influence the ability to produce an angular movement of the joints. The elasticity of the materials can be characterized based on the concepts of elasticity, rigidity, deformability, hysteresis, resilience, and viscosity. Elasticity: ability of a body to recover its original shape once the force that has deformed it has disappeared. Stiffness: a body resistance to deformation. Deformability: the inverse of the stiffness, requiring little force

per unit area to produce large deformation. Hysteresis: represents the energy dissipated between the deformation and recovery of the original shape. Resilience: the amount of energy returned by the deformed material during the discharge phase. Viscosity of a liquid or semiliquid substance is the resistance of a body to deformation in response to a load.

The duration of each phase of the jogging stride (support, flight, and swing) depends on the control of the muscles of each leg executed by the Nervous System: suprasegmental centres, spinal networks, and afferent information from the different senses and from the osteoarticular system. If the information from the feet changes due to modifications on the interface between the foot and the floor, it is expected that the muscular activity will be also modified.

The recording of electrical activity obtained during muscle contraction or electromyogram (EMG) reflects the muscle involvement in the movement of the joints and therefore in the kinetic response of the ground or reaction force. The EMG amplitudes are related both to the nervous system and to muscle tension, although the response of the mechanical system is not directly related to the nervous system signal (Bouisset 1973), as the mechanical response depends on more variables than the muscle activity (such as the length of the muscle, the rate of change of length, time of contraction, and the lever arm magnitude). The relationship between muscle activity and force is not straightforward; however, EMG amplitude, duration, and coordination among different muscles can provide information about the neural and mechanical systems. Therefore, the electromyography is a powerful tool in the study of the neuromuscular control of movement. The EMG signal is not easily recorded, as it is very susceptible to interferences and cable movement, and it is quite small, varying its amplitude between microvolts and millivolts (Kleissen et al. 1998) being its maximum amplitude (peak-to-peak) only 5 mV when using surface electrodes (Winter 1979).

During locomotion, the muscles of the legs are used to meet the following mechanical demands: progression, dynamic stability, and improving the impact and energy conservation (Inman et al. 1981); if any of these tasks is altered, the record of the muscles electrical activity will provide information about its contribution.

The aim of our present study was to evaluate the influence of footwear on the electrical activity of muscles of both legs when running barefoot and running with two different types of sports footwear, in order to assess the effects on: the extent of muscle electrical signal, the profile of muscular electrical activity, the order of muscle involvement, and the coactivation of antagonist muscles.

2. Material and methods

2.1. Subjects

Ten (six male and four female) healthy subjects, 19 and 20 years old, with an average height of 1.73 ± 0.10 m, with no history of neurological or musculoskeletal dysfunction, voluntarily participated in the study. All of them gave their written informed consent before participating in this research.

2.2. Experimental conditions

We have classified the hardness of the sole of each footwear type during jogging on the basis of their subjective hardness: barefoot condition was interpreted as the maximum hardness, as the outer protection and reduction of the shock at the beginning of the stance phase was minimal. The own athletic shoes of each subject (typically used to run) was the condition interpreted as the softest. The standard shoes were harder than the athletic shoes, and therefore feature less cushioning of the impact.

2.3. Proceeding

Prior to obtaining EMG recordings, subjects got used to carrying the electrodes and contact sensors (foot switches –FS) by walking freely in the laboratory until obtaining a normal gait.

For each individual there have been five successive records barefoot at spontaneous speed (no specific speed requested), five with a standard sports shoes and five with his/her own shoes. Subjects ran at ground level at their preferred speed in both the first registration and in the remaining four. When subjects were shod, they were given also some time to get habituated to the shoes, and moved at a freely chosen speed.

Each record registered the EMG corresponding to the cycles required to cross a distance of 10 m. From the cycles registered, clearly identified by the FS signal, only the central 2 were further analyzed; thus avoiding the effect of acceleration and deceleration on muscle activity. In order to calculate the average speed (in ms^{-1}), the time taken to cross the 10 m has been timed. Between each of the five records made for each condition, a pause of one minute was given.

The eight FSs (B & L, U.S.A.) facilitated the identification of the phases of the stride for each of the lower extremities. These sensors are flexible disks of two sizes: 18mm and 30mm. They were placed under the heel (30mm FS) and on the heads of the first and fifth metatarsal and toe tip (18mm FS). When subjects were shod, the FSs were placed on the bottom of the shoes at the sites corresponding to the outer edge of the heel, first and fifth metatarsal, and toe tip. A FS is activated when a pressure greater than 150 g is applied on it.

2.4. EMG

Visual monitoring of the signal from the FS eased the removal of stride records with deficiencies. The simultaneous recording of signals originating in the FS and the EMG has also allowed identifying the cycle phases (support, swing, or flight) in which the muscles were active. Prior to the start of the records, we checked the signals obtained through the electrodes and the FSs.

The electrode characteristics are presented in **Table 1**. The surface electrodes used were active, equipped with pre-amplifiers providing a gain of x320. EMG signal thus obtained is better than that achieved with passive electrodes respect to the level of noise. They are composed of three stainless steel electrodes, acting two as active electrodes and one as a common ground.

The leads used were bipolar, recording the difference signal between the two active electrodes ends. A general reference electrode was located on the right forearm of each subject.

Body size	50mm x 18mm x 7mm
Inter-electrode distance	Three 1/2" (1.27 cm) disks on 11/16" (0.16 cm) centres
Weight	30 grams (including connector)
Connector	LEMO, 5 pin male style
Cable Length	60 inches (1.5 metres)
Input Impedance	Greater than 100 MΩ
CMRR	>100dB at 60Hz
Bandwidth	10 Hz to 30KHz (-3dB)
Gain at 1kHz	320
Power Requirements	±4V to ±14V at 200μA

Table 1. Technical specifications of the active, surface electrodes used in this study.

The electrodes chosen for the study were of surface type, because they have the following advantages over intramuscular ones: do not cause pain or bleeding, are easier to apply, and as shown by Kadaba et al. (1985), they provide a more reproducible signal than that obtained with intramuscular electrodes.

The subject carried on his back a box with 14 channels (12 for EMG signal input from the 12 target muscles) and 2 for the signal from the FSs. Each channel had an additional gain range of 1 to 8. The 14 signals were transmitted through a optic fibre cable from the junction box to the electromyograph, a Motion Lab MA200 system equipped with a Pentium PC 64 MB of RAM, a CODAS acquisition card PGH DI 400, with 16 channel and 12 bit resolution. Through the Motion Lab software 900, reports of the actual electrical activity have been obtained and the linear envelop (LE) has been calculated. Afterwards, each LE has been expressed with respect to the normalized gait cycle. The display of the EMG signal in real time while being recorded has allowed us to assess the quality of the recording during movement, ensuring a good contact between the electrode and the skin and the absence of artefacts. The acquisition system (200 Motion Lab) has the following characteristics: level of signal output ± 5 V, bandwidth of 20 Hz to 1000 Hz, CMMR of 40 dB, and input impedance of 100 Ω.

The 12 muscles whose activity has been recorded have been forcibly superficial, as we were using surface electrodes. Four muscles were proximal: rectus femoris (RF), vastus medialis (VM), biceps femoris (BF) long portion, semitendinosus (ST); and two distal: tibialis anterior (TA) and lateral gastrocnemius (GN, or calf). These muscles were selected for their synergistic action and agonist-antagonist relationships. Biarticular muscles: rectus femoris, long head of biceps femoris, semitendinosus, and gastrocnemius; monoarticular: vastus medialis; through one of the joints examined: tibialis anterior. Architectural features of the muscles under study are shown in **Table 2**, and their function and innervation are shown in **Table 3**.

Muscle	Mass (g)	Muscular Length (ML) [mm]	Fibre Length (FL) [mm]	Pennation Angle [°]	Cross section area [cm ²]	FL/ML
Rectus Femoris	84,3±14	316±5.7	66.0±1.5	5.0±0.0	12.7±1.9	0.209±0.002
Vastus Medialis	175±41	335±15	70.3±3.3	5.0±0.0	21.1±4.3	0.210±0.005
Tibialis Ant.	65,7±10	298±12	77.3±7.8	5.0±0.1	9.9±1.5	0.258±0.015
Biceps Femoris	128±28	342±14	85.3±5	0.0±0.0	12.8±2.8	0.251±0.022
Semitendinosus	76.9±7.7	317±4	158±2	5.0±0.0	5.4±1.0	0.498±0.0
Gastrocnemius	150±14	248±9.9	35.3±2	16.7±4.4	32.4±3.1	0.143±0.010

Table 2. Architectural characteristics of the studied muscles (from Wickiewiz et al. 1983).

The location of the electrodes was done by orienting the surfaces of the electrodes with respect to the direction of muscle fibres (Testut 1971, Wickiewiz et al. 1983, Lieber 1992), in order to obtain a signal of greater amplitude and frequency (De Luca 1997). The electrodes were placed following the recommendations of SENIAM (1999). Electrodes location was verified by performing specific muscular contractions before carrying out the records.

Muscles	Joint	Function (at each Plane)			Innervation
		Sagittal	Coronal	Transversal	
Rectus Femoris	hip	flexor	abductor		Crural Nerve L2-L4
	knee	extensor			
Vastus Medialis	knee	extensor			Crural Nerve L2-L4
Biceps Femoris	hip	extensor	adductor	external rotator	Sciatic Nerve L4-S2
	knee	flexor		external rotator	
Semitendinosus	hip	extensor	adductor	internal rotator	Sciatic Nerve L4-S2
	knee	flexor		internal rotator	
Tibialis Anterior	tibiofibular-talar	flexor			Tibial Nerve L4-S1
	subtalar		inverter	adductor	
Gastrocnemius	knee	flexor			Tibial Nerve L4-S3
	ankle	extensor			

Table 3. Muscles targeted in this study: function and innervation (Kendall 2000).

The optimum recording of action potentials require excellent preparation of the skin before placing on it the electrodes; i.e., waxing the area, removing debris with alcohol, and letting it to air dry. Each electrode was attached to the skin via hypoallergenic tape and bandage.

2.5. Signals analysis

The EMG signals and those from the pressure sensors were recorded digitally at a frequency of 3000 samples per second using an analog-to-digital card CODAS (DataQ Instruments, OH, USA). The records were afterwards selected for further processing based on the signal obtained with the pressure sensors.

In order to determine the intensity of the signal, peak amplitude (peak activity), and time of its appearance, it is necessary to use quantitative methods on the EMG signal such as the LE. The steps to obtain it are: (1) complete (full-wave) rectification of the raw EMG, and (2) obtaining the linear envelop window (LE window) by calculating the average amplitude values contained in a moving window of 50 points.

Further processing was carried out with the signals of six subjects (the records of four subjects were excluded because their signals were not fully valid). From each record, the activity corresponding to 2 (out of 5) cycles were used ($5 \times 2 = 10$), for 6 subjects ($10 \times 6 = 60$), so 60 cycles were used in each condition. Since there are 3 different conditions, we analyzed a total of 180 (60×3) cycles. Therefore, the database consisted of 180 files; each file containing the 51 values of the LE corresponding to each of the 12 muscles and the time length of the phases of the cycle for both feet.

Each file has been then processes in a spreadsheet to obtain time-space parameters and the 51 values of each of the 12 muscles' EMG signals expressed in mV and corresponding to each 2% of the normalized length of jogging cycle.

For each subject, the LE corresponding to the 2% for each consecutive cycle locomotion were averaged across the 10 selected strides resulting in a pattern "ensemble average" of EMG (EAV), which represents an average pattern of the intra-subject LE. Using the EAV of all subjects, the Great Ensemble Average (GEAV) was obtained.

Muscle activity, represented by its LE, was expressed between the 0% and the 100% of the duration of the cycle. The maximum amplitude and time of peak onset were obtained from the GEAV.

2.6. Statistical analysis

The influence of footwear on speed, the phases of locomotion, right-left legs symmetry, general effort, and on the maximum amplitude of the EMG activity of the 12 target muscles was evaluated by analysis of variance (ANOVA) for one factor with the statistical package of Matlab (The MathWorks, Massachusetts, USA), using the following factorial model: footwear, with three levels (barefoot, shod with standard shoes, and shod with athletic shoes). The statistic F has been analyzed and the significance level considered was $p < 0.05$.

3. Results

The aim of our study was to evaluate the influence of footwear on the electrical activity of muscles of both legs when jogging barefoot and jogging with two types of footwear

featuring different geometry and damping characteristics. The barefoot condition was interpreted as the one in which the external protection and shock reduction at the beginning of the stance phase were minimal. The standard shoe was harder than the athletic one, cushioning less the impact.

For the realization of this goal we propose more specific objectives: to determine the profile of the muscle electrical activity and the order of muscle participation, to detect changes in the amplitude of the electrical signal muscles, and to measure the level of coactivation of antagonistic muscles. The different phases of the gait cycle were measured based on the records obtained by the pressure sensors (see **Figure 1**) located under the foot or shoe sole.



Figure 1. Bilateral support and non-support patterns from each foot obtained with pressure sensors.

The speed was higher when jogging with the own athletic shoes, and of a similar magnitude when jogging barefoot than when using standard sneakers. **Figure 2** shows the mean values of the speed and, for both legs, the mean values of the phases of: support, non support, and double flight. When running shod, the duration of the phases of the stride was different with respect the barefoot condition. The stance phase was shorter and the non-support phase, longer. The time length of the double flight increased. Between the two types of footwear used (standard and athletic) there were differences in speed but not in phases.

3.1. Muscular activity: EMG

Figure 3 shows the EMG signal, processed and expressed as GEAV for the three conditions. To assess the influence of footwear, we have divided the stride into two phases: (1) support phase and (2) non-support phase, each of which exhibited its own characteristics in muscle activity.

In order to check whether there are differences in the muscular effort depending on the locomotion condition, we calculated the area under each muscle's GEAV for each condition and used it as an estimation of that effort. The ANOVA found that there is no statistically significant difference between the three conditions with respect to the general effort required for the locomotion.

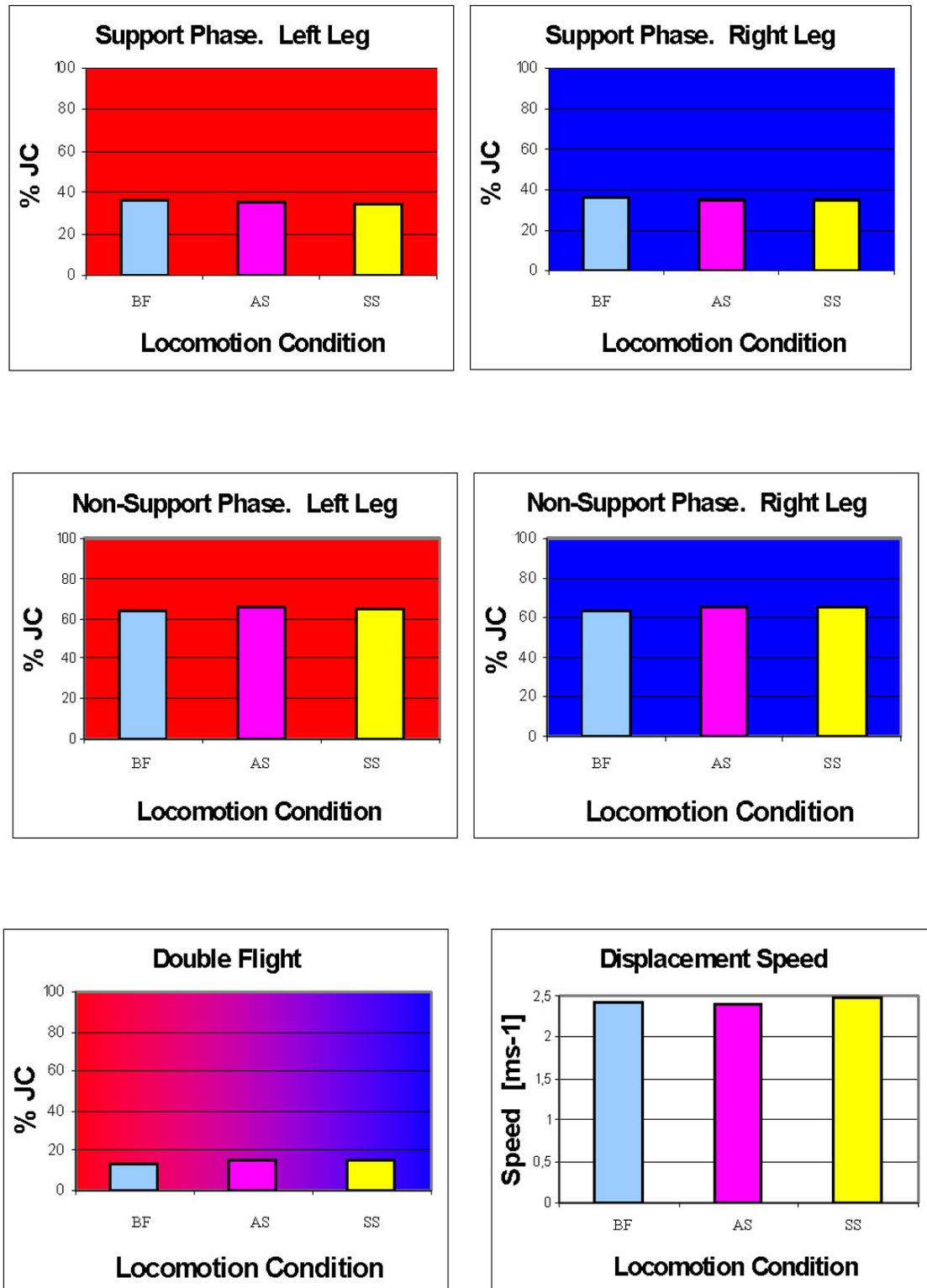


Figure 2. Space-temporal parameters expressed respect to the percentage of the jogging cycle (JC). Locomotion Conditions: BF –barefoot, AS –wearing athletic shoes, SS –wearing standard shoes.

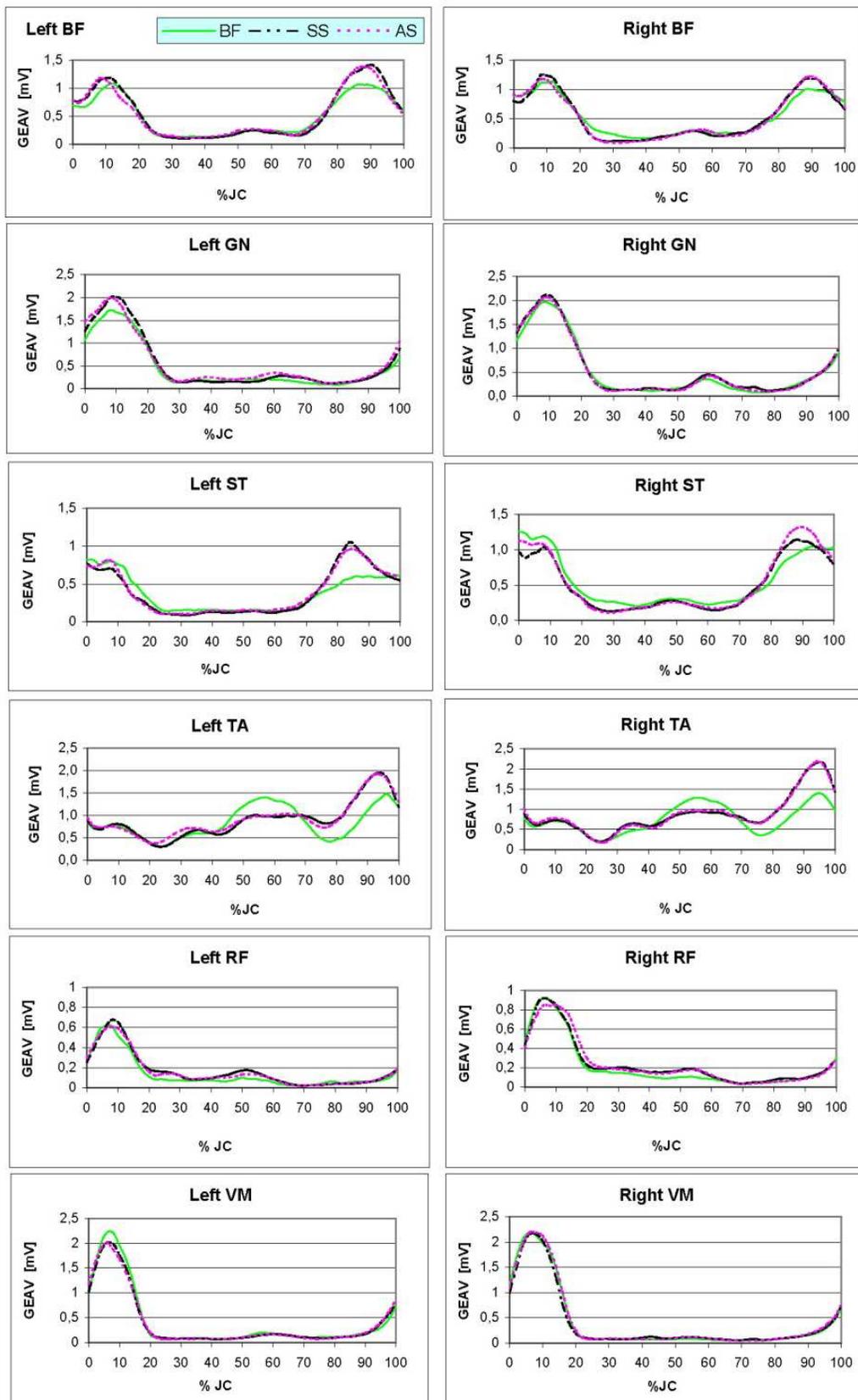


Figure 3. Profile of the GEAV for each muscle and for each locomotion condition (BF: barefoot, SS: standard shoes, and AS: athletic shoes).

3.2. Order of muscle participation

From the GEAV signal, the order of participation of each muscle can be seen. This order remained constant in all three conditions: tibialis anterior, hamstrings, gastrocnemius, and quadriceps. The AT muscle was the first one to be activated during the stride; it was activated before the flight of the ipsilateral foot, in the last part of unipodal support. The BF and ST were activated after the TA, but in the swing phase, i.e., when the contralateral leg was resting on the floor. The GN was activated in the final flight. Vastus medialis and RF were also activated in the final flight, but after the GN.

3.3. Profile of muscular electrical activity

The profile of the electrical activity is characterized by the length of time of muscle activity, the peak (maximum amplitude), and the time of its appearance. Changes in muscle activity can occur in any of the variables: amplitude, time, and frequency. In all three conditions muscle electrical activity was present for short periods of time.

The functions of the muscles varied through the cycle of jogging. There is no statistically significant difference between the three conditions with respect to the activity profile of the jogging cycle.

Rectus femoris, VM, and GN had a peak at the beginning of the stance phase. Tibialis anterior showed two peaks: one in the swing phase and another in the final flight. Biceps femoris and ST showed two peaks, one at the end of the non-support phase, and one in the initial support.

3.3.1. Time of onset of maximum intensity

Table 4 gives the exact percentages of jogging cycle at which the maximum amplitude occurred. For five muscles there were no differences in the time of peak onset in their GEAV, maintaining their patterns of activity in the three conditions. As shown in Figure 3, the peak of the RF, VM, and GN appeared in the support phase; the peak of the BF, ST, TA, in the non-support phase.

In both legs, the peak appears in very similar times. Depending on the footwear, the time of occurrence of the peak is slightly modified. In general, muscles that have their peak after the impact when shod may appear a little later; and muscles that have their peak before impact when shod can occur earlier.

The peak of the RF, VM, and GN appeared between 6% and 10% in the three conditions and in both legs. The TA peaks between 94-96% in the final flight; the highest peak of the BF appeared between 88% and 90%. Semitendinosus peak appeared about 84%-90% in the shod condition, ahead respect to the barefoot condition, which appeared between 94-98%.

3.4. Maximum amplitude

The changes of amplitude or intensity of muscle activity can be evaluated considering: (1) the entire cycle, (2) the stance phase after impact, and (3) the non-support, before impact. We will describe the changes in that order.

Locomotion condition	Left leg muscles					
	RF	VM	GN	TA	BF	ST
Barefoot	8	6	8	96	90	98
Standard shoes	8	6	10	94	90	84
Athletic shoes	8	6	8	94	88	84
Locomotion condition	Right leg muscles					
	RF	VM	GN	TA	BF	ST
Barefoot	6	6	8	94	88	94
Standard shoes	6	8	10	94	90	88
Athletic shoes	10	6	10	94	90	90

Table 4. Time of occurrence (in % of the jogging cycle) of the GEAV peak in each of the three different locomotion conditions.

3.4.1. Maximum amplitude along the entire cycle

Figure 4 shows the maximum activity displayed during the entire cycle by each of the twelve muscles. In all three conditions the muscles that had the highest amplitude were the VM, GN and TA. The muscles that showed less activity were the BF, ST and RF.

The maximum amplitude of homologous muscles was of the same order of magnitude for both legs, as shown in **Figure 5**.

There is no statistically significant difference between the three conditions with respect to the value of the amplitude peak.

3.4.2. Maximum amplitude during the support phase after impact

Figure 6a shows the maximum activity during the support phase produced after the impact of the heel. The TA, GN, and BF muscles in both legs increased their activity with both types of shoes. The behaviour of RF, VM, and ST varied positively and negatively.

The evolution of electrical activity in buffer muscles (RF and VM) during the loading phase was not the same in both legs, neither with both types of shoes.

Figure 7 shows the increase (not normalized, in mV) of EMG activity with respect to the barefoot condition. When subjects wore standard shoes with hard soles, the RF of the left leg increased its maximum amplitude after impact, and the left VM decreased their activity.

When subjects wore their own shoes, the RF of the left leg decreased its peak amplitude after impact. The left VM decreased its activity; and the one of the right, increased.

When the shoe was harder, the left RF increased its activity and the VM of both legs decreased. When the shoe was softer, the RF of both legs and the VM of the left leg decreased its activity, and the right VM slightly increased its activity.

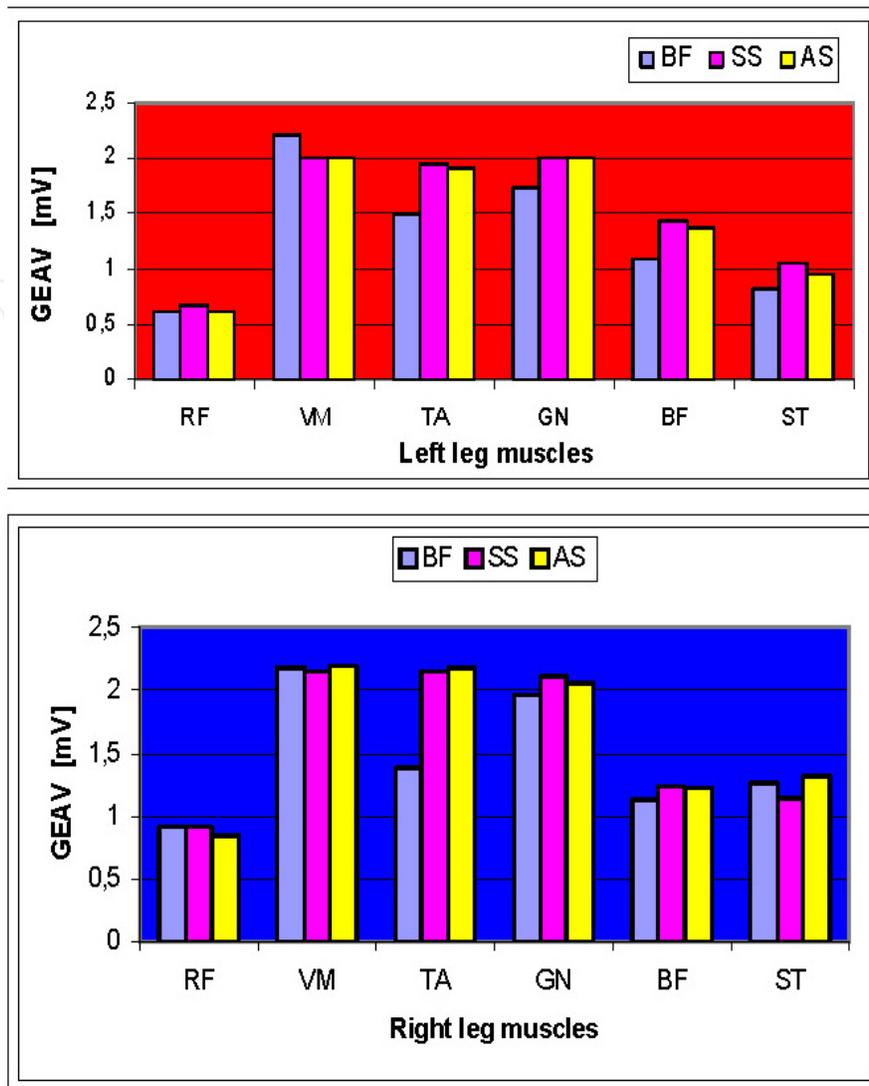


Figure 4. Peak of EMG during the whole cycle, for the left (upper graph) and right (bottom) legs muscles for each Locomotion Condition (BF: barefoot, SS: standard shoes, and AS: athletic shoes).

The TA of both legs increased their activity. The right TA increased more than the left one. In both legs, the TA increases more its maximum amplitude in the condition of own jogging shoes; perhaps because this type of shoe offers more cushioning. The GM activity in the condition of standard jogging shoe increased more than in the own shoe condition. When the hardness of the sole was higher, the translational velocity was lower and higher GM activity. Perhaps this trend has to do with balance, because when the speed is less, dynamic stability decreases.

The BF of both legs and both conditions increased their activity. In the standard shoe condition, it increased more. The ST in both conditions decreased its activity.

The response of the agonist muscles from the same muscle group is not the same when the shoe condition changes. Thus, with the standard shoe RF activity increased and decreased the one of VM, the BF increased, and ST decreased.

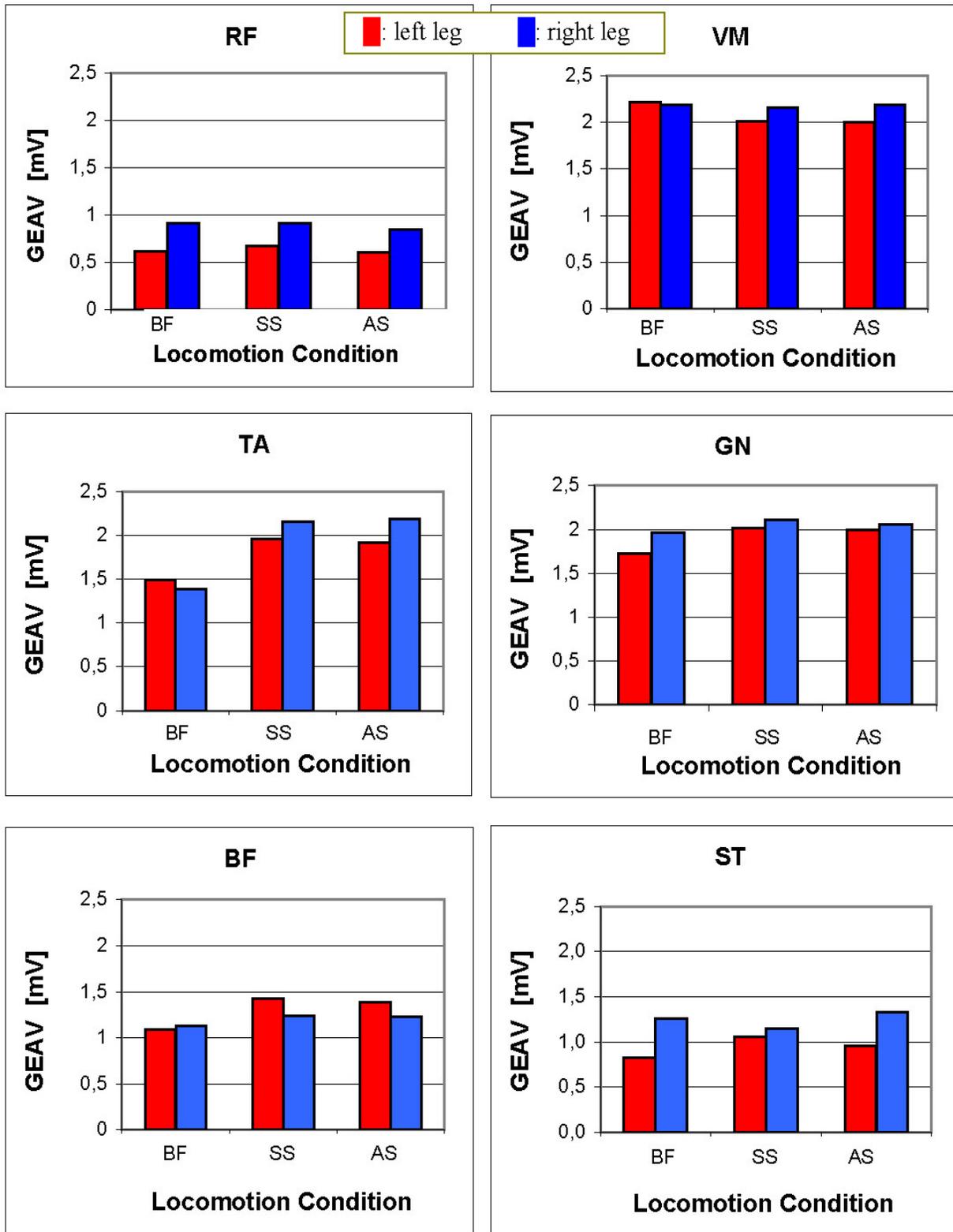


Figure 5. Maximum amplitude of the EMG from homologous leg muscles (red: left, blue: right; RF: rectus femoris, VM: vastus medialis, TA: tibialis anterior, GN: lateral gastrocnemius, BF: biceps femoris, ST: semitendinosus) for each locomotion condition (BF: barefoot, SS: standard shoes, and AS: athletic shoes).

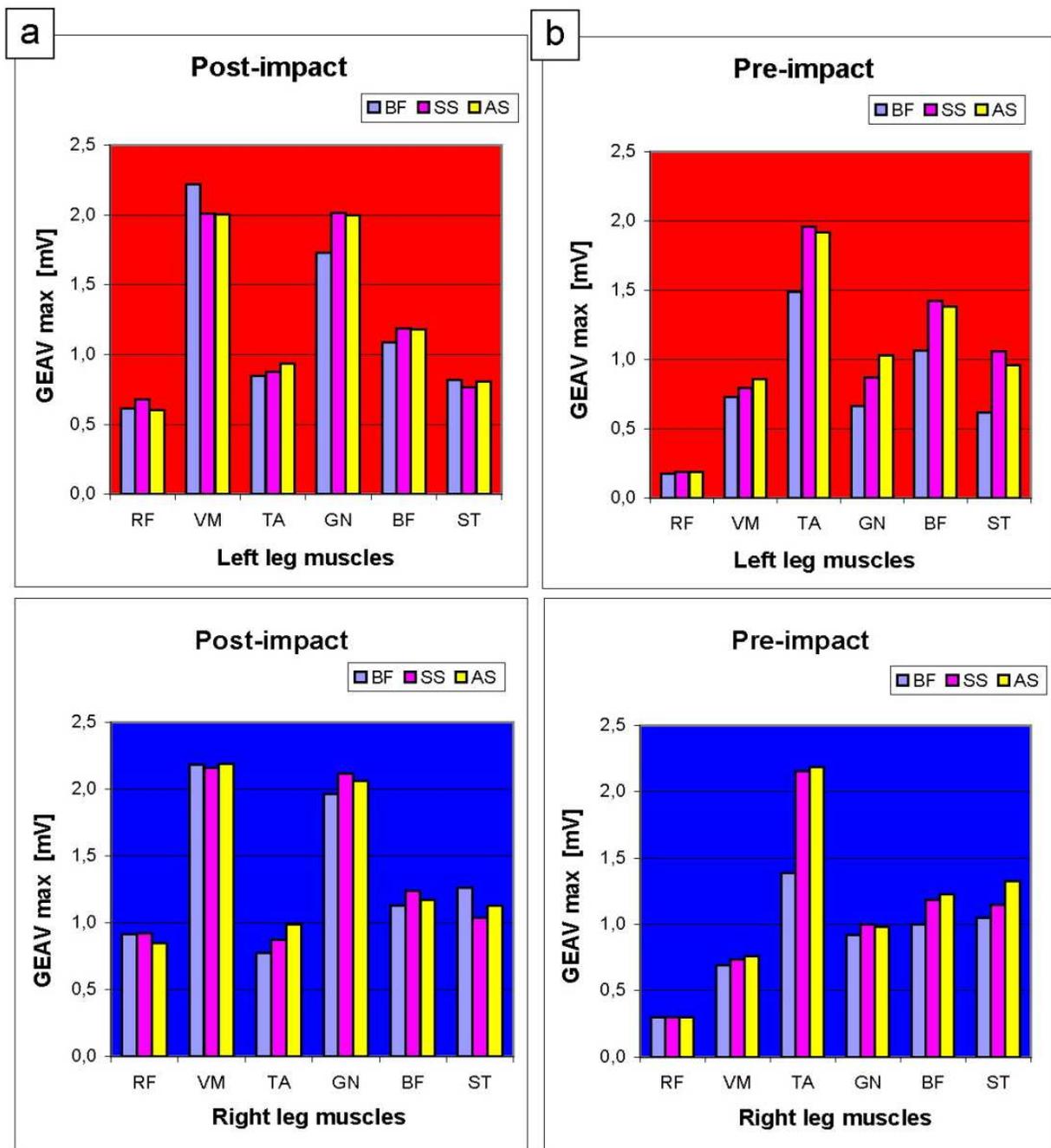


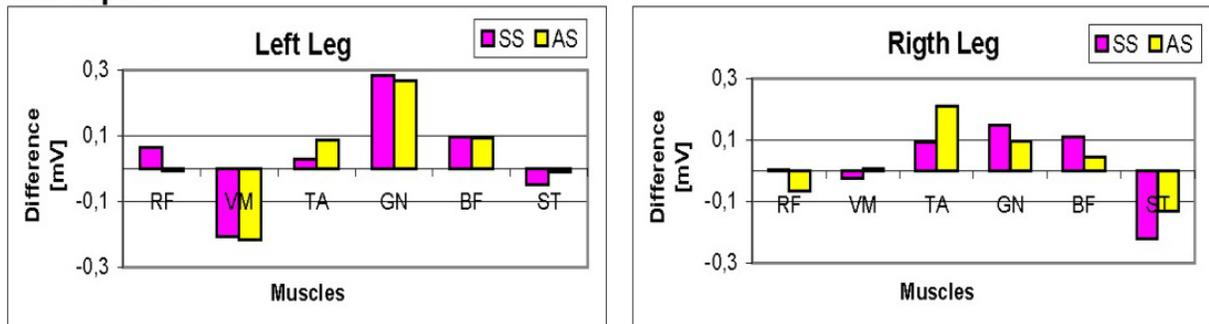
Figure 6. Maximum amplitude of the GEAV post - and pre-impact (after and before initial contact) for the 12 studied muscles for each locomotion condition (BF: barefoot, SS: standard shoes, and AS: athletic shoes).

3.4.3. Maximum amplitude before impact

As shown in **Figure 6b**, for both legs, five out of the six muscles (TA, GN, BF, ST, and VM) increased their activity in the non-support phase in shod condition. The RF maintained its level of activity; unlike what happened in the stance phase, in which the RF, VM, and ST decreased their activity.

Figure 7 shows that the EMG signal before contact of the heel exhibited greater increase than in the signal after contact. The intensity of the myoelectric activity showed differences between both types of shoes. In both shod conditions the TA muscle activity increased more in absolute terms. The ST, BF and GN increased their activity more than the VM, the activity of which increased very little.

Post-impact



Pre-impact

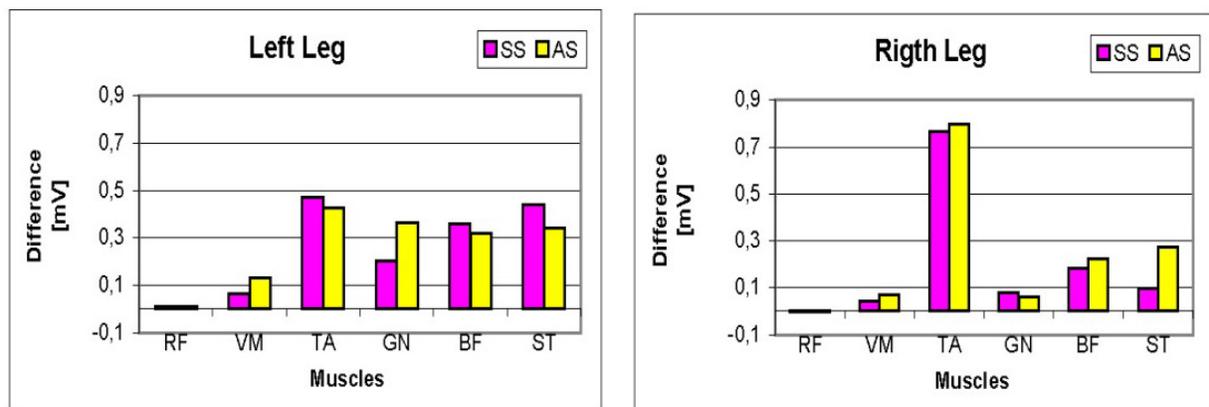


Figure 7. Absolute GEAV peaks amplitude increments respect to barefoot condition for the other two locomotion conditions (SS: standard shoes and AS: athletic shoes) for both legs muscles after impact (upper graph) and before impact (bottom graph).

In the left leg, the TA, BF, and ST were more active in the standard shoe condition. The increased activity of the semitendinosus was more important than that of the biceps femoris. In the right leg, the TA, BF, ST, and VM activity increased more in the athletic shoe condition than in the standard shoe condition. The electrical activity of the muscles of both legs showed different evolution depending on the type of shoe.

3.5. Activity of antagonist muscles

In order to evaluate the coactivity of agonist-antagonist muscles, we have established an index obtained as the ratio between the peak of the agonist muscle activity and the activity of the oppose muscle or muscles (antagonists). That index can be higher (agonist's peak is bigger than that of the antagonist) or below (opposite case) one (see **Figure 8**). To establish the role of a muscle as agonist or antagonist, we rely on the movement of joints:

- In the support phase, between 6% and 10% of the jogging cycle, the peaks of the following muscles occur: GN, RF, and VM; there is also a second peak of lesser magnitude in the activity of the BF and ST muscles. Concomitant to the important activity of these muscles, ankle dorsiflexion, knee flexion, and hip extension occur.
- In the phase of non support, the peak of the TA, BF, and ST are concomitant with an ankle dorsiflexion, knee extension, and hip extension.

There is no statistically significant difference between the three conditions with respect to the level of coactivation.

3.5.1. Coactivation during support phase

3.5.1.1. Barefoot

- Coactivation of tibialis anterior (agonist) / gastrocnemius (antagonist): the peak activity of GN occurs between 8% and 10% of the jogging cycle. The TA shows no peak in the stance phase. The index is below unity. The LG shows greater activity during dorsiflexion.
- Coactivation of hamstring (agonist) / rectus femoris (antagonist): index above unity. The BF (hip extensor) shows increased activity.
- Coactivation of hamstring (agonist) BF / vastus medialis (antagonist): the index is less than unity and therefore smaller than that showed by the antagonist muscles BF and RF because the activity of VM is much higher.
- Coactivation of semitendinosus (agonist) / rectus femoris (antagonist): index greater than one, but less than that found between the BF and RF because the ST has less activity than the BF.
- Coactivation of semitendinosus (agonist) / vastus medialis (antagonist): the index is less than unity and therefore is less than that presented by the antagonist muscles BF and RF because the ST activity was smaller than that of the BF, and the activity of the VM is much higher.

The peaks of RF and VM were not taken into account because they occur while acting as antagonists.

3.5.1.2. Shod

- Coactivation of tibialis anterior (agonist) / gastrocnemius (antagonist): both muscles increase their activity, so the index remained similar.
- Coactivation of semitendinosus (agonist) / vastus medialis (antagonist): the index is maintained because the two muscles decrease their respective activity.
- Coactivation of hamstring (agonist) / vastus medialis (antagonist): the index increases slightly because the BF activity increases and that of the VM decreases.
- Coactivation of semitendinosus (agonist) / rectus femoris (antagonist): the index decreases because the ST activity decreases.
- Coactivation of biceps femoris (agonist) / rectus femoris (antagonist): the index decreases because the RF activity increases.

The antagonist muscles showed more activity than agonist ones in the ankle and knee joints, for all the three conditions. In the hip, agonist muscles were more active, mainly in shod condition.

3.5.1.3. Coactivation during non-support phase

In the ankle and knee joints, agonist muscles showed a slightly higher activity than the antagonists. In the hip and knee, agonist muscles activity was higher.

The peaks of TA, BF, and ST muscles occur in the non-support phase (see Figure 3), concomitant with the dorsiflexion, the extension of the knee, and the hip extension. In all three conditions, the agonist muscles show greater activity than the antagonist muscles, as shown by the coactivity index being higher than one (see Figure 8). The results about the muscular coactivity during jogging barefoot and shod are just preliminary, but indicate that it is worth a study in greater depth on the subject.

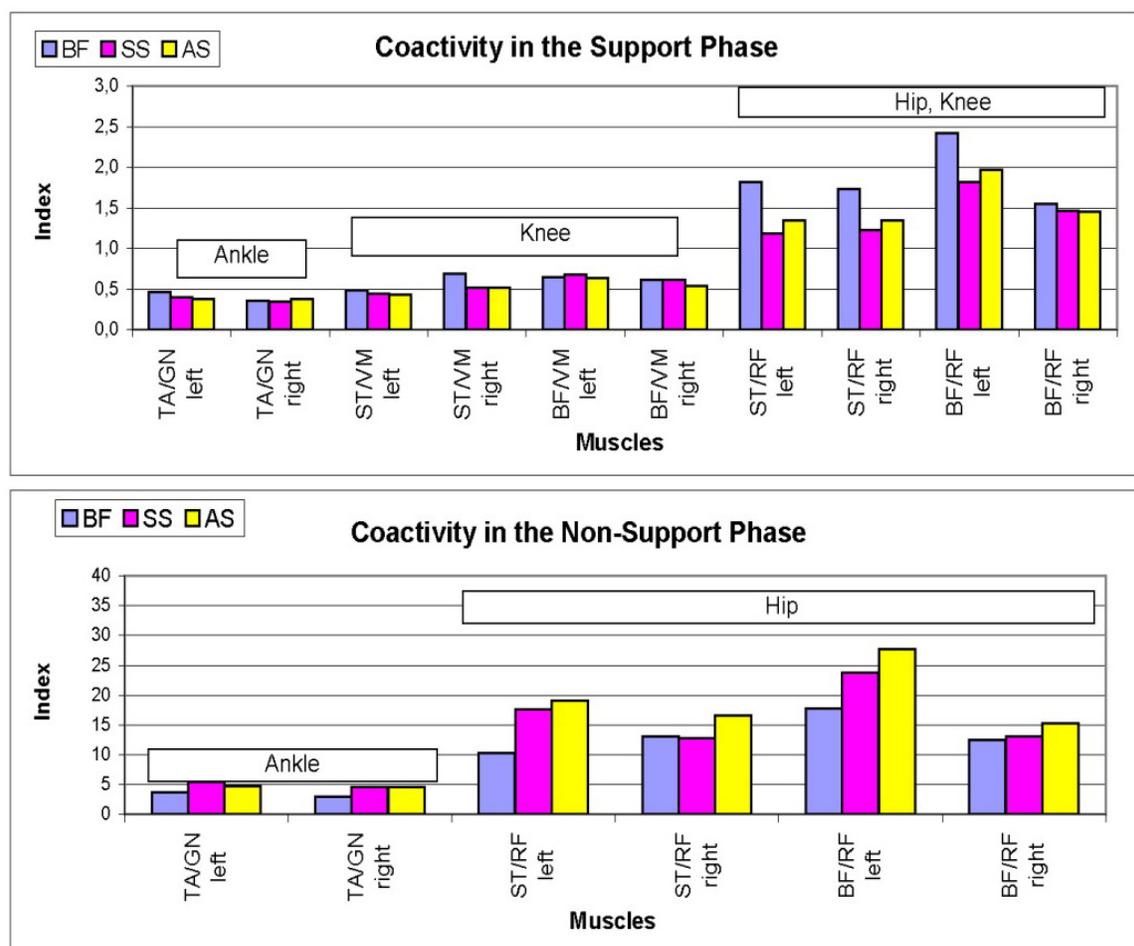


Figure 8. Coactivity of ago- and antagonist muscles during support (upper graph) and non-support (bottom graph) phases.

4. Discussion

The aim of our study was to evaluate the influence of footwear on the electrical activity of muscles of both legs when jogging barefoot and shod with two types of shoes, with different

geometry and damping features. Our findings show that shoes increase the intensity of muscle activity, especially in the non-support phase. These results are consistent with those of Nigg et al. (2003) that showed specific changes in the intensity of muscle activation before heel strike.

The speed increase is concomitant with decreased activity of the VM and ST in the stance phase. The increase in the intensity of TA in the stance phase is less than in the non-support phase, a result consistent with Komi et al. (1987).

The speed of jogging at which the EMG activity was recorded in the three conditions was freely chosen by the subjects because previous research (e.g., Kadaba et al. 1985) have shown that EMG activity is more reproducible when the speed is chosen freely than when it is imposed.

We note that the footwear influences the speed of locomotion, increasing it in the athletic shoe condition. This result suggests that the speed has to be one more parameter to consider when assessing the influence of footwear on the interaction between man-shoe-surface. The speed in similar condition of hardness (barefoot and standard shoe) was similar, but muscle activity was different in both conditions, being more similar that of the standard shoe and athletic shoe conditions.

In the different conditions under study, the pattern of muscle activity kept its profile. The starting and ending times of the activity were constant, but the amplitude of the EMG signal changed when using footwear. Also, the duration of the phases in the shod condition was reversed with respect to the bare condition, decreasing the stance phase and increasing the non-support one.

A limitation of this work was that has considered only 6 muscles out of the 57 having each of the lower extremities. The advantage over previous studies is that it has analyzed those six muscles from both legs simultaneously. The fact that the homologous muscles of both legs have similar amplitude allows us to use their absolute values.

The timing of muscle activity was assessed using visual inspection of the actual records, which is the detection method used to interpret the real surface EMG signal. Usually, visual inspection provides a high level of accuracy because all the details of the signal can be measured. Detecting the beginning and end of the activity by using different types of algorithms is based on establishing a threshold, usually based on intuitive criteria. In the end, both visual interpretation and detection using algorithms are based on subjective criteria. The advantage of the subjectivity of the experimenter is based on the acquisition of the ability to interpret the EMG signal and experience that allows the evaluator to use the capabilities of the human brain to perform a more sophisticated interpretation of highly variable data (Staudte & Wolf 1999).

The order of participation of the muscles studied did not vary with the type of shoes. First, the TA muscles activated, afterwards the muscles of the posterior part of the thigh (BF and ST), followed by the activation of the GN of the back of the leg, and finally the activation of the RF and VM of the anterior part of the thigh.

The length of time of muscle activity in the three conditions did not vary with respect to the normalized jogging cycle. In all conditions, muscle activity was present in the first half of the stance phase, disappearing in the second, when the body weight fell on the forefoot. In

absolute values, the time of activity would be different because the stance phase was shorter in the condition of athletic shoes and the phase of non support was longer. Thus, the time of muscle activity would be greater in the stance phase and lower in the non-support phase.

It is interesting to find that during the jogging, as happens in walking, there are phases in which no muscle activity is taking place, although there are movements that would make one expect some muscular activity. This result is consistent with other studies that have shown that the main goal of locomotion is to transport the body from one position to another using lower limb locomotor coordination and effective mechanisms. The nervous system integrates the movement of different body segments and controls the activity of the muscles to decrease energy expenditure (Inman 1968).

This lack of muscle activity is not surprising. In jogging, the decrease of kinetic and potential energies in the first part of the support is concomitant with the accumulation of elastic energy in the tendons of the muscles; this energy is later released during the remainder of the stance phase to facilitate the progression and stability dynamics. Margaria and Cavagna (1965) found that during jogging, elasticity accounts for 50% of the work. During the unipodal support, the contralateral leg swing provides a second pushing force generated by the acceleration of the leg. Once the centre of gravity has shifted over the foot on the ground, the weight of the body becomes a driving force. When the contralateral leg is slowed by the hamstring muscles, the ipsilateral leg acquires kinetic energy, which is used to start the double flight. Inman (1968) argues that the slowdown of the swing leg during walking can contribute more to the forward movement of the body than the thrust of the ipsilateral leg. Once the leg is in motion, as the first law of Newton states, a force is needed to stop it, and that is the role of the hamstrings when they become active at the end of the swing phase. While the hamstrings are active, the RF and VM become activated to improve the accuracy of the landing of the foot.

The amplitude of the myoelectric activity, expressed in absolute values, varied when using the two different types of footwear; these results are consistent with those found by Gollhofer & Komi (1987), who found differences in the electrical activity of muscles when subjects were at first barefoot and then shod. The effects of athletic shoes on the amplitude of muscle activity were significant. Changes in hardness and shape of the shoe did not affect the muscle activity.

Different considerations have led us to not normalize the EMG signal amplitude: (1) to assess absolute muscle activity when a sole is interposed between the foot and the ground, and (2) to maintain the electrodes in place when the geometry of the footwear and the elasticity of the sole varied.

The subjects ran faster in the athletic shoe condition and muscle activity was similar in the two footwear conditions: higher in the non-support phase. There is no statistically significant difference between the three conditions with respect to the speed of locomotion.

There was similar muscle activity in both shod conditions, which was different from the barefoot condition. In this paper we have classified the hardness of footwear subjectively. Although the interaction subject-shoe-ground occurs only during the stance phase, the intensity of the EMG activity was modified both during the stance phase and during the phase of non support. At first, one might think that the presence of the sole alters the physical

characteristics of the force of action and reaction and influences the feedback mechanisms of muscle activity. The results show that although the interaction subject-shoe-ground occurs only during the stance phase, the amplitude of the EMG activity was modified both in the stance and in the swing phases, the major changes occurring in the non-support phase.

Before the foot contacts the ground, there is no information available from the ground; therefore, no feedback mechanism is in use and muscle activity would be controlled by forward mechanisms. The changes in the amplitude of muscle activity in the two footwear conditions were more homogeneous in both legs during the post-impact than during the pre-impact. This finding is in agreement with those of Nigg (2003), who suggests that the EMG activity before and after impact corresponds to two different events. Before the foot contact with the ground, there is no information available from the ground, then no feedback mechanism is in use. It is speculated that this would be some pre-programmed muscle activity based on the expected impact and would be related to the "muscle tuning" and controlled by a forward mechanism. Thus, the configuration of the leg when the heel hits the floor is prepared beforehand during the non-support phase, following the strategy adopted (De Wit et al. 2000). In the other hand, the activity that appears after heel contact is due to a reflex effect. In this context, one can understand the variability of electrical activity in the pre-impact and its consistency in the post-impact.

The changes in the impact force described under the influence of footwear may be a consequence of changes in muscle activity in the non-support phase to reduce soft tissue vibrations during locomotion. Wakeling et al. (2002) explains how the frequencies of the impact force and the soft tissues may overlap during walking and running. It can be expected that the impact forces will cause vibrations within the range of vibration of the soft tissues. However, observations show that such vibrations are minimal. Wakeling et al. (2002) proposed that muscle activity is adjusted before impact in order to reduce possible vibrations resulting from the impact.

The reflex effect has to be understood in the context that the heel contact with the ground is a stimulus that triggers feedback mechanisms. This stimulus does not trigger the onset of muscle activity, which had already begun in all muscles during the swing phase or during the final flight phase (or even during the initial flight phase), but could just change it. The reflex time is around 110 ms in the normal population (Kroll & Clarson 1978), and is divided into a latency time of 20 ms and a motor time of 90 ms. A peak muscle activity for most of the muscles studied appears on the 8% of 700ms-long jogging cycle, i.e. at 56 ms, which correspond well with the reflex motor time.

The maximum activity peak appears in the three conditions in the same phase. The increase in mean activity increased recruitment of motor units and increased force production. The decrease means the opposite. The observation of decreased activity of the VM in the post-impact is consistent with previous findings (De Wit et al. 2000) about the influence of shoes on the ground reaction force. In the shod condition, impact peak appears later perhaps due to decreased activity of the damper (shock absorber) muscles. The loading speed is influenced by muscle activity, which affects the acceleration of the segments.

When wearing footwear, the right and left BF increased their activity at the end of the swing phase, in the final flight, and in the onset of the stance phase. At the end of the swing phase, those muscles act as antagonists of the extension of the knee and of the hip flexion in the swing phase; from the final flight phase, they act as agonists of the extension of the hip.

The increased activity of BF in the shod condition at the end of the swing phase could be justified by the higher speed and greater participation of these muscles to slow down the flexion of the thigh. In principle, its action would facilitate the advancement of the leg by inertia, and then its action would stop the leg being extended to facilitate the support of the foot on the floor. Perhaps, the increased activity in the phase of non-support has an effect on increasing the speed of the contralateral leg, as suggested by the studies of Inman (1968). At the beginning of the stance phase, the BF and ST intervene as agonists of the extension of the hip and knee flexion, cooperating synergistically with the torque that has the lower extremity and maintained as a result of inertia when the foot has been slowed by friction with the ground. When the sole absorbs more impact, shock muscles (RF and VM) decreased its activity. When the shoe was harder, the left RF increased its activity and that of the VM of both legs decreased.

The fact that the muscles of both legs do not modify its maximum amplitude in the same percentage is understood in the context that the activity of a leg is adapted to the activity of the other. However, in the left leg there are significant differences in the activity of some of its muscles when wearing different types of footwear (see for example O'Connor 2005).

The lower activity of the RF and of the VM observed at the beginning of the stance phase may facilitate hip extension and knee flexion caused by increased activity of the hamstrings. This finding is consistent with the results of De Wit et al. (2000) who found that the knee is more flexed in the medium supporting phase in the shod condition. Also in that condition, the TA increased its activity in the final flight and at the beginning of the stance phase, being its activity lower during part of the swing. The role of the TA in the final flight and at the beginning of the stance phase is to promote dorsiflexion. The increased activity in the final flight can be explained based on segmental geometry: the hip extends and the foot flexes dorsally to avoid bumping into the ground until the proper stride length has been obtained. When jogging without shoes, De Wit et al. (2000) reported increased plantar flexion at the beginning of the support phase. Our results are consistent with their findings: the lower activity of the TA muscle in this condition would favour plantar flexion, apparently intended to reduce the pressure, which is higher in the barefoot condition because the impact force is increased and occurs in a shorter time.

One explanation for the decreased activity of ST in the stance phase in the shod condition could be the following: this muscle is extensor and adductor of the hip, and also flexor and rotator of the knee. Shod, the movement from supine-pronation can be affected by the change in the distance between the point of application of the GRF and the subtalar or calcaneus-talus joint. The shoes would favour a faster decrease of the inversion with a faster rotation of the knee in a lower limb that is externally rotated, favouring a twist in the knee. If the ST does not decrease its activity, it would increase that twisting torque affecting the knee. A more active GN in the stance phase could influence the stability of the ankle. In the

shod condition, there is more damping of the impact force. Cushioning and stability are conflicting (Perry 1992): more cushioning brings less stability.

Nigg et al. (2003) speculated that the body reacts to changes in input signal to adapt the muscle activity to reduce the vibrations of the soft tissues. In principle, the proximal muscles would be less required than the distal ones. These authors found that the TA activity increased more than that of the proximal ones (BF and ST). Our results are consistent with those findings respect to the TA and the ST; but not for the BF.

According to Mundermann et al. (2003), changes in the intensity of leg muscles activity predict better the differences in comfort than changes in the impact force. As the muscle activity decreased in the stance phase, the kinematic differences between the conditions disappeared (De Wit et al. 2000). This result is consistent with the absence of muscle activity. The signal amplitude indicates the level of activation of the muscle: there is a relationship between activity and strength, implying that increased activity is the source of increased muscle strength. The change in muscle electrical activity may be due to the role of the muscles to adapt to the characteristics of elasticity and friction of the shoes used.

According to Arsenault et al. (1987), it seems clear that the kinematics of locomotion is not very variable. From the scientific literature one can obtain the parameters describing the movement and explain muscle activity during jogging in different conditions. When the peak of the RF, VM, GN, BF, and ST muscles appears, the hip stretches, the knee flexes, and the ankle flexes dorsally. Rectus femoris and VM undergo an eccentric contraction, BF and ST concentric, and GN concentric in the knee and eccentric in the ankle. Biceps femoris and GN promote the flexing of the knee and the GN transfers energy from the ankle to the knee.

The activation of the muscles acting as antagonists (i.e., contracting eccentrically) promote the absorption of energy recovered in the immediate concentric contraction. In the condition barefoot, the BF presented electrical activity at the end of the swing phase, in the final flight, and at the onset of stance phase, acting first as antagonist and then as agonist. With shoes, the activity increased in both phases, increasing the energy absorbed in the non-support phase. During the concentric contraction at the beginning of the stance phase, the increased activity means more power generation. The increase of the absorbed energy and the energy produced could be interpreted as an effort at the beginning of the stance phase in order to do hip extension, perhaps to counterbalance the increased frictional force that opposes the advance in the shod condition. At the beginning of the support phase, the GN, RF, and VM are active in eccentric contraction. The reduction of the duration of the support phase in the shod condition may improve the efficacy of the stretching-shortening cycle.

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.

As the human body is a biological system that has many possibilities of action and reaction, it would be advisable to evaluate the electrical activity of other additional muscles when

changing the subject-shoe-ground interface. According to Ferris et al. (1999), persons adjust the degree of stiffness of the legs when they run on different surfaces. The stiffness is given by the coactivation of the ago-antagonists muscles that cross the joints. In the support phase, the coactivation in the shod condition was greater than in the barefoot condition. At the stage of non support, the coactivation in the shod condition was not homogeneous, neither for both legs nor for all muscles, as it may decrease or increase respect to the barefoot condition.

A major function of footwear is to cushion both the strength of action the subject performs on the ground and absorb its reaction force, in order to protect the musculoskeletal system. Athletic shoes diminish –or even nullify- the impact peak of the vertical component and are involved in delaying the onset of the vertical support force, by changing the load gradient (Nigg 1983, De Wit et al. 2000).

The impact force acts as an input signal in the body and influences the vibration of the soft tissues (Nigg et al. 2000). The vertical component is biphasic and has two peaks: the impact peak appears after 20 ms of the impact, representing 140-160% -200% of BW, in those runners that support the heel first in the ground. A second peak, in the stance phase, appears at 80 ms and can almost triple the BW. Previous works (De Wit et al. 2000, Nigg 1983) have shown that in jogging, the magnitude of the passive and active peaks of the vertical component of the GRF does not vary with the shoes. The load gradient was lower in the shod condition due to the later occurrence of the impact force (33 ms against 11 ms).

In our study, peak muscle activity appeared in the following cycle time: 2-6% for the ST, 6-8% for the VM and RF, 8-10% for GN. In absolute values, the 2% amounts to 14 ms, 6% to 42 ms, and 10% to 70 ms. The activity peak of four muscles occur between the peaks of the vertical component of the reaction force. Despite finding a second peak of greater magnitude in the shod condition, muscle activity was higher in only three of the six muscles studied. In two of the other muscles (VM and ST) in the shod condition the activity was lower. In the RF, the activity response was not constant.

The increased activity of the TA in the shod condition could lead to the appearance of the anterior compartment syndrome of the lower leg. The results here presented about the increased EMG signal amplitude suggest further studies to corroborate or refute the argument used by athletes and shoe manufacturers on increased performance with the use of athletic footwear.

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5. References

- Arsenault AB, Winter DA, Marteniuk RG (1987). Is there a 'normal' profile of EMG activity in gait?. *Medical & Biological Engineering & Computing* 337-343.
- Bates BT, Osterninig JA, Sawhill JA (1983). An assessment of subject variability, subject inter-action, and the evaluation of running shoes using ground reaction force data. *J Biomech* 16: 181-191.
- Behnke R (2001). *Kinetic Anatomy*. Human Kinetics, Champaign,IL.
- Bouisset S (1973). EMG and force in normal motor activities. In: Desmedt JE ed. *New developments in electromyography and clinical neurology*. Base: Karger: 547-583.
- De Luca CJ (1997). The use of surface electromyography in biomechanics. *Journal of Applied Biomechanics* 13: 135-167.
- De Luca CJ & Merletti R. (1988). Surface myoelectric signal cross-talk among muscles of the leg. *Electromyography and Clinical Neurophysiology*; 1988, 69:568-75.
- De Wit B, De Clercq D, Aerts P (2000). Biomechanical analysis of the stance phase during barefoot and shod running. *J Biomech* Mar;33(3):269-78.
- Dufek JS, Bates BT, Davis HP, Malone LA (1991). Dynamic performance assessment of selected sport shoes on impact forces. *Medicine and Science in Sports and Exercise* 23 (9): 1062-1067.
- Denoth J and Nigg B (1981). The influence of various sport floors on the load on the lower extremities. In : *Biomechanics VII-B,A*. Morecki, K Fidelus ,K Kedzior and A Wit (eds). Baltimore: University Park Press, 100-106.
- Ferris DP, Liang K, Farley CT (1999). Runners adjust leg stiffness for their first step on a new running surface. *J Biomech*. 32: 787-794.
- Forner A, Garcia AC, Alcantara E, Ramiro J, Hoyos JV, Vera P (1995). Properties of shoe insert materials related to shock wave transmission during gait. *Foot Ankle Int* Dec;16(12):778-86
- Frederick EC (1986). Kinematically mediated effects of sport shoe design: a review. *J Sports Sci* 4 (3): 169-184.

- Gavilanes MB, Goirierna de Gandarias JJ (2004). Muscle activity in shod and barefoot healthy young subjects during walking. In : International proceedings of XVth Congress of the International Society of Electrophysiology & Kinesiology. Boston U.S.A. 2004:105
- Gollhofer A, Komi PV (1987). Measurement of Man-Shoe-Surface interaction during locomotion. *Medicine Sport Sci*, 26: 187-199.
- Inman VT (1968). Conservation of energy in ambulation. In: *Bull Procc Res BPR* 1968; 10:(9):26-35.
- Inman VT, Ralston HJ, Todd F (1981). *Human walking*. Baltimore, MD, Williams and Wilkins Company.
- Kadaba MP, Wootten ME, Gainey J, Cochran GV (1985). Repeatability of phasic muscle activity: performance of surface and intramuscular wire electrodes in gait analysis. *J Orthop Res*. 1985;3(3):350-9.
- Kendall FP, McCreary EK, Provance PG (2000). *Músculos: pruebas funcionales, postura y dolor*. Willinams & Wilkins. Baltimore.
- Kleissen RFM, Buurke JH, Harlaar J, Hof AL, Zilvold G. C (1998). Electromyography in the biomechanical analysis of human movement and its clinical application. *Gait & Posture* 8 : 143-158.
- Komi PV, Gollhofer A, Schmidtbleicher D, Frick U. (1987) Interaction between man and shoe in running: considerations for a more comprehensive measurement approach. *Int J Sports Med*. 1987 Jun;8(3):196-202
- Kroll W & Clarson PM (1978). Fractionated reflex time, resisted and unresisted fractionated reaction time under normal and fatigued conditions. In DM Landrers & RW. Christina (Eds), *Psychology of Motor Behaviour and Sport* 1977 (pp 106-129) Champaign ,IL: Human Kinetics.
- Lieber RL (1992). Skeletal muscle response to injury. In: *Skeletal muscle structure and function*. John P Butler, Ed. Baltimore/Williams & Wilkins, 1992: 260-292
- Luethi S, Stacoff A (1987). The influence of shoes on foot mechanics in running. *Med. Sport Sci.*, 25:72-85.
- Mann RA, Moran GT, Doherty SE (1986). Comparative electromyography of the lower extremity in jogging, running and sprinting. *The American Journal of sports Medicine* 14 (6): 501-510.
- Margaria R, Cavagna GA (1965). The mechanics of walking. *J Physiol (Paris)*. Sep-Oct;57(5):655-6.
- Martín PE, Morgan DW (1992). Biomechanical considerations for economical walking and running. *Med. Sci. Sports Exerc*. 24 (4):467-474.
- Meinecke L., Disselhorst-Klug C., & Rau G. (2004): Crosstalk and Coactivation in Bipolar Surface EMG Data: A New Methodology for Detection, Discrimination and Quantification; in the 25th Congress of the International Society of Electrophysiological Kinesiology (ISEK), Boston, Massachusetts, June 2004, p. 87.
- Milliron M, Cavanagh PR (1990). Sagittal plane kinematics of the lower extremity during distance running. In *Biomechanics of distance running*. Human Kinetics Books, Champaign Illinois.
- Mundermann A, Nigg BM, Humblew RN, Stefanyshyn DJ (2003). Foot orthotics affect lower extremity kinematics and kinetics during running. *Cl Biom*18: 254-262.

- Nigg BM (1983). External force measurements with sport shoes and playing surfaces; in Nigg Kerr, Biomechanical aspects of sport shoes and playing surfaces 11-23 (University Printing, Calgary).
- Nigg BM, Bahlsen HA, Denoth J, Luethi SM, Stacoff A (1986). Factors influencing kinetic and kinematic variables in tuning. In: Biomechanics of running shoes. BM Nigg (ed.). Pp 139-159. Champaign, IL: Human Kinetics Publishers.
- Nigg BM, Wakeling JM (2001). Impact forces and muscle tuning: a new paradigm. *Exerc Sport Sci Rev* 29(1):37-41.
- Nigg BM, Mundeermann A, Stefanyshyn DJ, Cole G, Stergiou P, Miller J (2003). The effect of material characteristics of shoe soles on muscle activation and energy aspects during running. *J Biom* 36: 569-575.
- Novacheck TF (1998). The biomechanics of running. *Gait & Posture* 7: 77-95.
- Perry J (1992). Gait analysis systems in gait analysis: normal and pathological function. Thorofare, NJ: Slack 353-411.
- Nordin M and Frankel VH (2004). Biomechanica de la rodilla. In: Biomechanica basica del sistema musculoesqueletico. McGraw Hill/Inteamericana de Espana, S.A.U.
- O'Connor KM, Hamill J (2005). The role of selected extrinsic foot muscles during running. *Clin Biomech* (2004) Jan;19(1):71-7.
- Ramiro J, Ferranids R, Sánchez J, Alepuz R, Latorre P, Dejoz R, Candela F (1988). Evaluación de la técnica del calzado deportivo. *Archivos Medicina del Deporte V* (18): 161-168.
- Reber L, Perry J, Pink M (1993). Muscular control of the ankle in running. *Am J Sports Med* 21 (6):805-810.
- SENIAM (1999). European recommendations for surface electromyography.
- Segesser B, Nigg BM (1993). Orthopedic and biomechanical concepts of sports shoe construction *Sportverletz Sportschaden* 7 (4) 150-162.
- Shortein MR (1993). The energetics of running and running shoes. *J Biom* 26 (1): 41-51.
- Slocum DB, James SL (1968). Biomechanics of running. *JAMA*. 1968 Sep 9;205(11):721-8.
- Stauder G, Wolf W (1999). Objective motor response onset detection in surface myoelectric signals. *Medical Engineering & Physics* 21:449-467.
- Testut L, Latarjet A (1971). *Tratado de Anatomía Humana*. Barcelona: Salvat editores.
- Vaughan, CL (1984). Biomechanics of running. *Crit Rev Biomed Eng* 12:1-48
- Wakeling JM, Von Tschanner V, Nigg BM, Stergiou P (2001). Muscle activity in the leg is tuned in response to ground reaction forces. *J Appl Physiol* Sep;91(3):1307-17.
- Wakeling JM, Pascual S, Nigg B M (2002). Altering muscle activity in lower extremities by running with different shoes. *Med. Sci. Sports Exerc.* 34 (9): 1529-1532.
- Winter DA (1979). Mechanical work, energy and power. In: Biomechanics of human movement. Pp 84-107. Toronto: John Wiley and Sons.
- Winter DA (1990). Biomechanical and motor control of human movement, 2nd edn. New York:Wiley, 1990.
- Winter DA, Fugelvand AJ, & Archer SE (1994). Crosstalk in surface electromyography: theoretical and practical estimates. *Journal of Electromyography and Kinesiology* 1994;4:15-26.
- Wickiewicz TL, Roland R, Perry L, Powell BS, Edgerton R (1983). Muscle architecture of the human lower limb. *Clinical Orthopaedics and Related Research* 179: 275-283.