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# Muscle Activation Patterns During Level Walking and Stair Ambulation

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## 1. Introduction

Dynamic electromyographic (EMG) data recorded using surface electrodes is commonly used to measure muscular activity during the performance of motor tasks and investigate normal and pathological motor strategies (Frigo & Crenna, 2010; Ferdjallah, 1998). In clinical gait analysis, the determination of the timing of muscle activation ("on-off") is of paramount importance (C.J. De Luca, 1997; Shiavi, 1985). The evaluation of the "on-off" pattern of one or more muscles, particularly when examined together with kinematics (joint angles) and kinetics (joint moments and powers), provides an insight into the performance of muscles and their role in accomplishing a motor task (Gage, 1992; Benedetti et al., 1999; Davis, 1997). The relevance of considering the muscle activation timing is supported by several publications demonstrating its usefulness in orthopaedics (Andriacchi et al., 1982; Winter, 1984), treatment of cerebral palsy (Rose et al., 1999; Lyons et al., 1983; P.A. De Luca et al., 1987), and a number of other clinical applications (C.J. De Luca, 1997). More specifically, the timing of activation of muscles ("on-off" pattern) is the only reliable information that can be obtained from surface EMG signals recorded during dynamic contractions. In fact, the amplitude of the EMG signal is not only influenced by the actual electrical activation level of the muscle, but also by the electrical characteristics of the tissue between the active muscle fibers and the surface electrodes. Furthermore, since the relationship between electrical activity level and muscle force is non-linear, caution has to be taken in the interpretation of the EMG amplitude as an indicator of the muscle force output when data are recorded in dynamic conditions (C.J. De Luca, 1997). Unfortunately, this approach does not allow one to investigate completely the advanced control features of the central nervous systems, since muscle activation is simplified considering a muscle either "on" or "off". However, in a great number of the clinical queries, the muscle activation timing represents the information of interest for the clinician, provided that reliable data are obtained for the activation intervals.

Methods proposed in literature to determine the onset and offset of muscle activation intervals generally rely on an arbitrary choice of the threshold value that discriminates

background noise from the signal generated by the active muscles (Hodges & Bang, 1996; Winter & Yack, 1987). The main drawbacks of these methods are: (1) the estimated activation patterns strongly depend on the subjective decision of such threshold value thus resulting in unreliable estimates and large variability (Bonato et al., 1998); and (2) these methods do not allow for the appraisal of the likelihood of false positive (specificity) and false negative (sensitivity) detections. In order to overcome these limitations, a double-threshold detector of muscle activation, based on the statistical properties of the EMG signal was developed (Bonato et al., 1998).
In a recent study this method was applied to obtain a paediatric reference data-set of muscle activation patterns during gait (Agostini et al., 2010).
The aim of the present work is to provide a reliable reference data set, referred to an adult population, of the muscular activation timing during common activities of daily living - such as level walking and stair ambulation - for clinical and research use. To this purpose, raw surface EMG signals from trunk and lower limb muscles were recorded in a sample population of healthy young volunteers and processed by means of the above-mentioned double-threshold detector. The main problems encountered during data collection and processing are herein discussed to provide suggestions for further improvements to be accomplished in future work.

## 2. Materials and methods

Sixteen young subjects were recruited in this study, 9 males and 7 females, age 27.7±2.6 years (mean and standard deviation), weight 67.4±10.5 kg and height 171.6±8.8 cm. They had no history of locomotor disturbances or injuries. Subjects walked barefoot, along a 15m-walkway, at a self-selected speed of progression (Table 1). Two force platforms (Kistler Instruments, Switzerland) were mounted in the walkway to measure ground reaction force and detect the occurrence of stance and swing phases.

| | Mean | SD |
|---|---|---|
| Walking (cm/s) | 125.6 | 16.8 |
| Stair ascending (cm/s) | 43.9 | 6.6 |
| Stair descending (cm/s) | 48.1 | 1.7 |

Table 1. Speed of progression.

The same group of subjects was tested while walking up and down a flight of stairs barefoot at a self-selected speed of progression (Table 1). The staircase was made in wood, its dimensions were chosen according to the architectural normative in use in Italy. It was composed of three steps, each with a tread depth of 28 cm , a width of 86 cm, and a riser of 16 cm. The first and the third steps were connected to the two Kistler force platforms. During stair ascent, the stride cycle was defined by the right foot, beginning at foot contact on the first step and ending at subsequent contact of the foot on the third step. During stair descent the stride cycle was defined by the right foot beginning at foot contact on the third step and ending at subsequent contact of the same foot on the first step.

### 2.1 Muscles investigated

Eight muscles were considered for the right side of each subject: ipsilateral and contralateral erector spinae at lumbar site, gluteus medius, rectus femoris, medial hamstrings, lateral hamstrings (biceps femoris, long head), gastrocnemius (medial head), and tibialis anterior.

### 2.2 Electromyographic instrumentation and methodology

The surface EMG signal was detected using active clip-type probes connected to adhesive disposable electrodes made of Ag/AgCl for paediatric electrocardiographic application. The inter-electrode distance was 20 mm, center to center. The signal was amplified using an eight-channel system (TELEMG, BTS, Milan, Italy). Sampling rate was 500 Hz, low and high cut-off frequencies of the amplifier were 40 Hz and 200 Hz, respectively. Three repetitions were gathered for each subject.

The location and orientation of the electrodes for each muscle were identified according to the literature (Delagi & Perotto, 1981). The skin under the electrodes was shaved and cleansed with alcohol.

### 2.3 Statistical double-threshold detector for muscle on-off timing

EMG signals were processed using a statistical detector designed and implemented as briefly summarized in the following. The complete description and full characterization of the algorithm is provided in a previous work (Bonato, 1998).

The raw EMG signal was first processed using an adaptive whitening filter. Then an auxiliary time-series was derived by squaring and summing adjacent samples. The amplitude of the auxiliary time-series was therefore $\chi^2$ distributed. Based on a background noise sample, we determined the characteristics of the $\chi^2$ distribution of the data for intervals when muscles were silent. A window of 5 samples was slid through the auxiliary time-series. A double-threshold strategy was applied to the samples within the window to determine whether the signal reflected a silent or active status of the monitored muscle. A first threshold was set on the amplitude of the samples of the auxiliary time-series. A second threshold was set on the number of samples within the sliding window that needed to be above threshold to result in a detection of muscular activity. Moreover, through the definition of receiver operating characteristics derived via simulation, the user is provided with means to choose the operating point defined by sensitivity and false alarm probability with more degrees of freedom than what traditional approaches would provide. This allows the user to adapt the behaviour of the detector as a function of the signal characteristics (e.g. signal-to-noise ratio). This strategy was applied to the EMG signal together with a post-processor that rejected transitions shorter than 30 ms. The combination of the two techniques provided us with means to detect muscle activation intervals with probability of false alarm smaller than 2 % and sensitivity higher than 85 % for a signal-to-ratio (SNR) as low as 6 dB. Fig. 1 shows an example of application of the double-threshold detection algorithm to an EMG signal. As suggested in the literature (Perry, 1992; Bogey et al., 1992) we defined the minimum period of muscle inactivity between contractions as 5% of the gait cycle. When the duration of "on" intervals was less than 5% of the gait cycle, the muscle was considered as silent, i.e. the short detected activation between two silent periods was considered a false detection and corrected. Similarly, when "off" intervals shorter than 5 % were detected, the muscle was considered as active, i.e. the short silent period between intervals of muscle activity was considered a false negative and corrected.

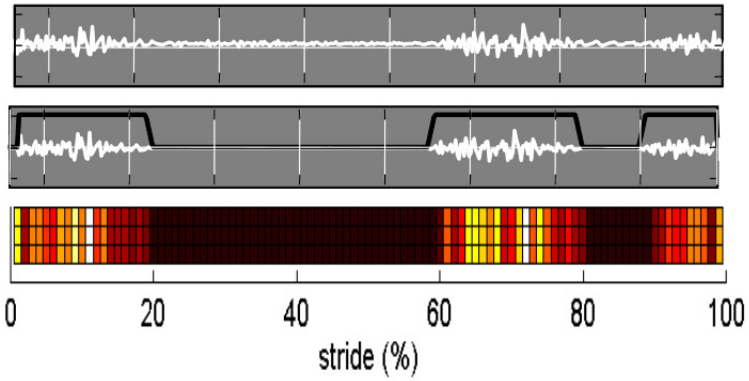


Fig. 1. Example of raw EMG signal (upper plot), whitened EMG signal and on-off detection of muscle activations (middle plot), and representation of the amplitude of the EMG signal using five discrete levels of intensity represented by different colors (lower plot).

## 3. Results

In order to give a global representation of the muscle activation intervals on the entire population we considered, for each percent of the gait cycle, the percentage of subjects in which a specific muscle is active. Fig. 2 shows the on-off muscle activation intervals of the population during level walking, Fig. 3 represents the on-off intervals during stair ascending, while Fig. 4 represents the on-off intervals during stair descending. In Figs. 2-3-4, vertical bars reaching 100% mean that all the subjects have a specific muscle active at the percent of the stride considered. On the contrary, null vertical bars (0%) mean that all the subjects have their muscle silent at the specific percent of the stride considered.

### 3.1 Level walking

*Right Erector Spinae*. In all subjects, the ipsilateral erector spinae muscle was active around initial contact and between terminal stance and pre-swing (Fig. 2).
*Left Erector Spinae*. The activation pattern of the contralateral muscle is very similar to that of the ipsilateral one. Synergic lumbar paraspinal muscles function consists of decelerating the forward trunk rotation at heel strike and toe off. A further activity of this muscle is present, in a certain number of patients, during midstance, aimed to sustain the swinging limb on the coronal plane and tilting the trunk on the support limb.
*Gluteus Medius*. In all the subjects the activity begins at terminal swing, in preparation to heel strike, and it continues throughout the first half of the stance phase.
*Rectus Femoris*. The rectus femoris resulted to be active, in all the subjects, during the terminal swing and during the first part of the gait cycle, thus stabilizing the knee under loading. In a small percentage of subjects, in accordance with previous findings (Nene, 1999), we detected activity at toe off.
*Lateral Hamstrings*. The long head of the biceps femoris begins to be active in the last part of swing phase - decelerating hip flexion and knee extension, and preparing the lower limb to heel strike and load absorption - and continues during initial stance phase, extending the

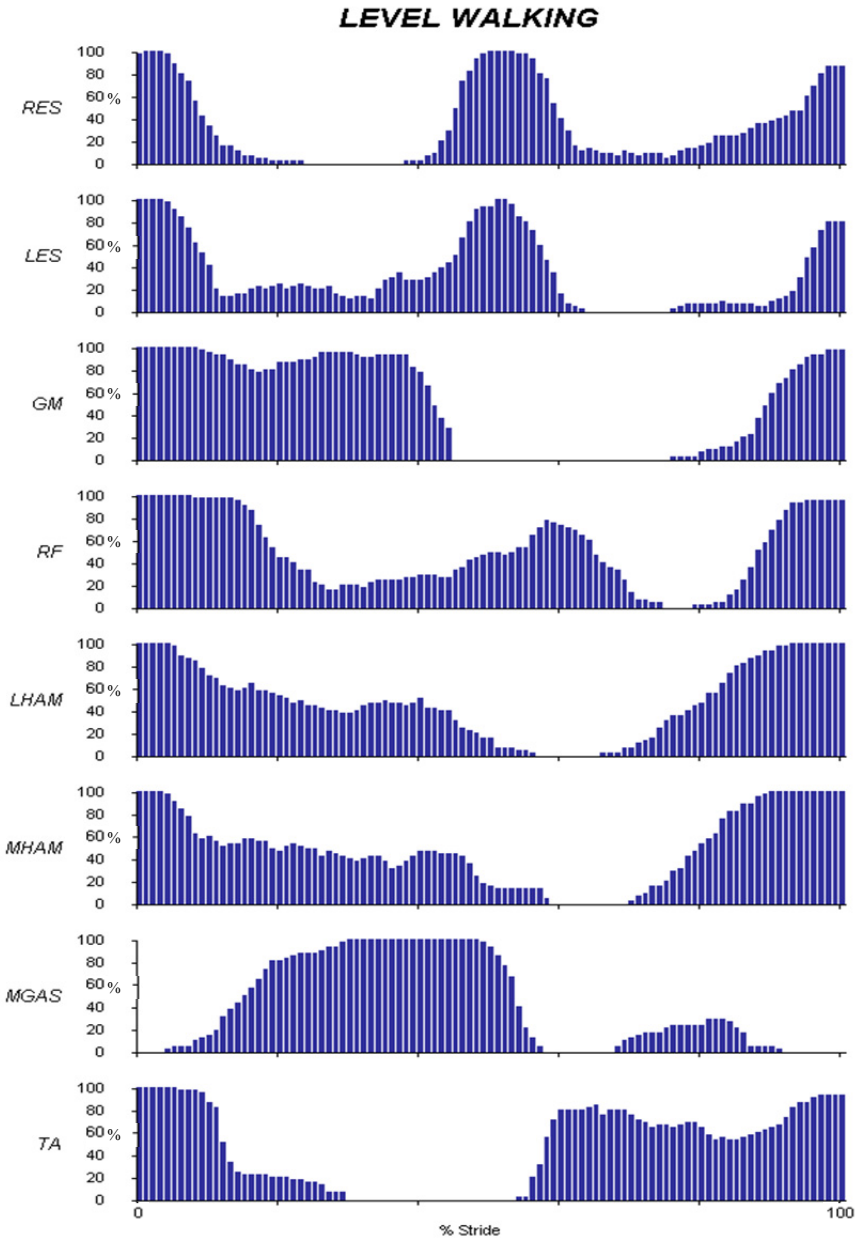


Fig. 2. On-off muscle timing during level walking, as a function of the percentage of the gait cycle. For each muscle, the muscular activity is presented as percentage of subjects whose muscle is active at the specific percent of the stride considered. RES: right erector spinae, LES: left erector spinae, GM: gluteus medius, RF: rectus femoris, MHAM: medial hamstrings, LHAM: lateral hamstrings, MGAS: medial gastrocnemius, TA: tibialis anterior.

hip and stabilizing the knee in co-contraction with the rectus femoris. In agreement with Shiavi (Shiavi, 1981) a further period of activity was present, in 50 % of the subjects, during terminal stance, to lead into the unloading response of preswing when inertia is not sufficient to flex the knee.
*Medial Hamstrings*. Semitendinosus and semimembranosus activity during gait is generally similar to the lateral hamstrings activity.
*Medial Gastrocnemius*. The gastrocnemius muscle begins its activity during initial stance and ends its activity at terminal stance in accordance with literature (Shiavi et al., 1983). In a small percentage of the examined subjects we detected also a short burst of activity in the middle of the swing phase, as previously already observed (Basmajian, 1978).
*Tibialis Anterior*. The tibialis anterior activity begins at the toe off and continues throughout the swing phase with decreased amplitude during mid swing in most of the subjects. This muscle is active at heel strike and continues to be active during the loading response phase, until the foot is completely in contact with the floor.

### 3.2 Stair ascending

*Erectors Spinae*. No comparable data are available in the literature for these muscles during stair climbing. The muscle contralateral to the supporting limb is active throughout the stance phase. Its activity helps the climbing limb to swing over the next step thus elevating the pelvis on this side. Vice versa, the ipsilateral erector spinae is generally active at initial stance, controlling the limb loading, and during midstance, as fixators of pelvis during contralateral swing (Fig. 3).
*Gluteus Medius*. As for walking, the gluteus medius is active in the first half of the stance phase of the supporting limb.
*Rectus Femoris*. Results obtained in our subjects are consistent with data previously published by others (Andriacchi et al., 1980), reporting activity of this muscle in the first half of the stance phase. Nevertheless, there is only partial agreement with findings by Perry (Perry, 1992), who documented a pattern marked by two activation intervals during initial stance and mid swing.
*Hamstrings*. The activity of these muscles starts at the end of the swing phase to continue during initial stance, extending the hip. The activity continues, in most subjects, during midstance, fixing the hip under loading. Furthermore all subjects, approximately at toe off, activate the hamstrings to flex the knee for the next step over. Data previously published (Lyons et al., 1983) reports similar activation patterns.
*Medial Gastrocnemius*. Gastrocnemius is active for most of the stance phase. This finding is not in agreement with what reported in the literature (Andriacchi et al., 1980). In fact, in our recordings, the onset of the gastrocnemius activity occurs earlier, at heel strike. However, this pattern can be justified, from a kinesiological point of view, since the initial support during stair ascending is on the forefoot.
*Tibialis Anterior*. The tibialis anterior muscle is active during the swing phase, providing adequate foot clearance. Its activity continues, in most of the subjects, after the ground contact in order to control the foot landing. Moreover, a certain activity is present in many subjects throughout stance. It is worthwhile to emphasize the role of the tibialis anterior in the control of foot inversion-eversion, a very important motion for balance control in the coronal plane during single limb support.

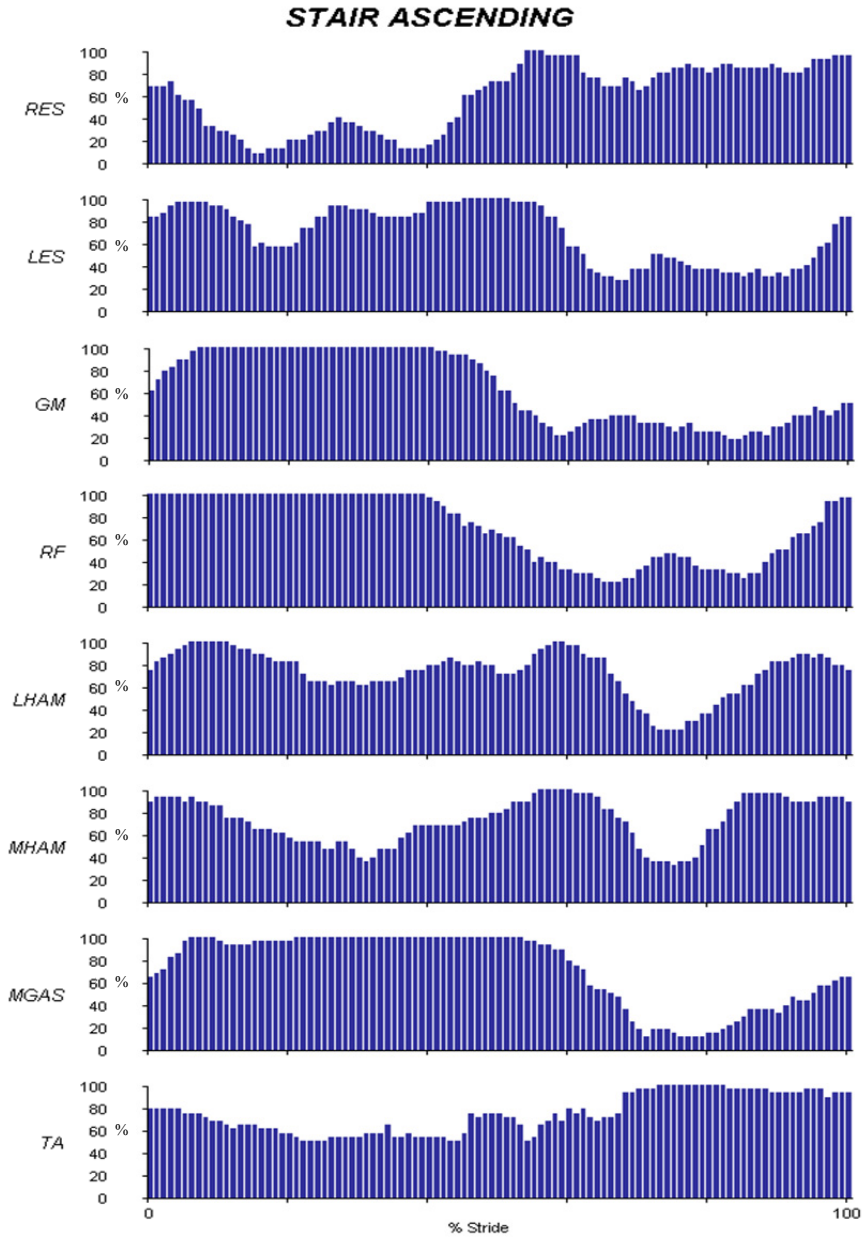


Fig. 3. On-off muscle timing during stair ascending, as a function of the percentage of the gait cycle. For each muscle, the muscular activity is presented as percentage of subjects whose muscle is active at the specific percent of the stride considered. RES: right erector spinae, LES: left erector spinae, GM: gluteus medius, RF: rectus femoris, MHAM: medial hamstrings, LHAM: lateral hamstrings, MGAS: medial gastrocnemius, TA: tibialis anterior.

### 3.3 Stair descending

*Right Erector Spinae*. In the large majority of subjects, the onset of this muscle is found twice, at heel strike and toe off. These two activations are likely to be related to the control of the trunk forward momentum. Furthermore, this muscle presents activity, in many subjects, during the swing phase, probably controlling the fall down of the pelvis due to the descending limb. No references are available for this muscle in the literature (Fig. 4).
*Left Erector Spinae*. In the examined group of subjects this muscle shows a continuous activity during the stance phase, aimed at controlling the obliquity of the pelvis, while subjects swing the limb to reach a lower step of the stair.
*Gluteus Medius*. In the majority of the subjects, this muscle is active throughout the stance phase. All subjects show a first muscle activation interval during the loading response, likely to be associated with the stabilization of the pelvis in the coronal plane while the weight is transferred to the ipsilateral limb. The large majority of subjects show also a second activation interval associated to the contralateral foot contact. No reference data are available for this muscle.
*Rectus Femoris*. This muscle is active during stance, controlling knee flexion, and in mid and terminal swing, contributing to knee extension and preparing the landing limb to load transfer.
*Hamstrings*. There is hamstrings activity at the beginning of the stance phase, to extend the hip controlling inertia, and at toe off, to flex the knee. Furthermore, this muscle shows activity, in the majority of the population, at mid swing, in opposite phase with respect to the activity of rectus femoris, aimed to reduce knee extension and to prepare the limb loading. The hamstrings are also active at the end of the swing phase. This activity is observed in all the subjects in preparation to limb loading.
*Medial Gastrocnemius*. Similarly to what was observed for stair ascending, we detected an activation onset of medial gastrocnemius around heel strike, that is, hence, anticipated in comparison to what was found by others in the past (Andriacchi et al., 1980). This finding is consistent with the forefoot initial contact during stair descending. The activity of the gastrocnemius muscle lasts until terminal stance. This muscle is active again at the end of the swing phase preparing foot landing.
*Tibialis Anterior*. The activity observed in the population for this muscle is spread throughout the stance phase. Subjects typically show muscle activation at initial stance, probably controlling foot inversion-eversion. Furthermore, it can be observed muscle activity, common to most of the subjects, at initial swing, to sustain the foot during the landing on the lower step, and before initial contact.

## 4. Discussion

Procedures previously adopted for the on-off pattern detection were generally based on single-threshold methods (Bekey et al., 1985). These methods have been shown to be strongly dependent on the arbitrary choice of the threshold value. Different threshold values may lead to dramatically different estimates of the EMG activation intervals (Winter & Yack, 1987). The double-threshold statistical detector used in the present study outperforms single-threshold methods (Staude, 2001). For a given value of false alarm probability, it leads to higher detection probability. The increase in sensitivity provided by the double-threshold approach depends on the SNR and it is dramatic for relatively low values of SNR and low values of false alarm probability.

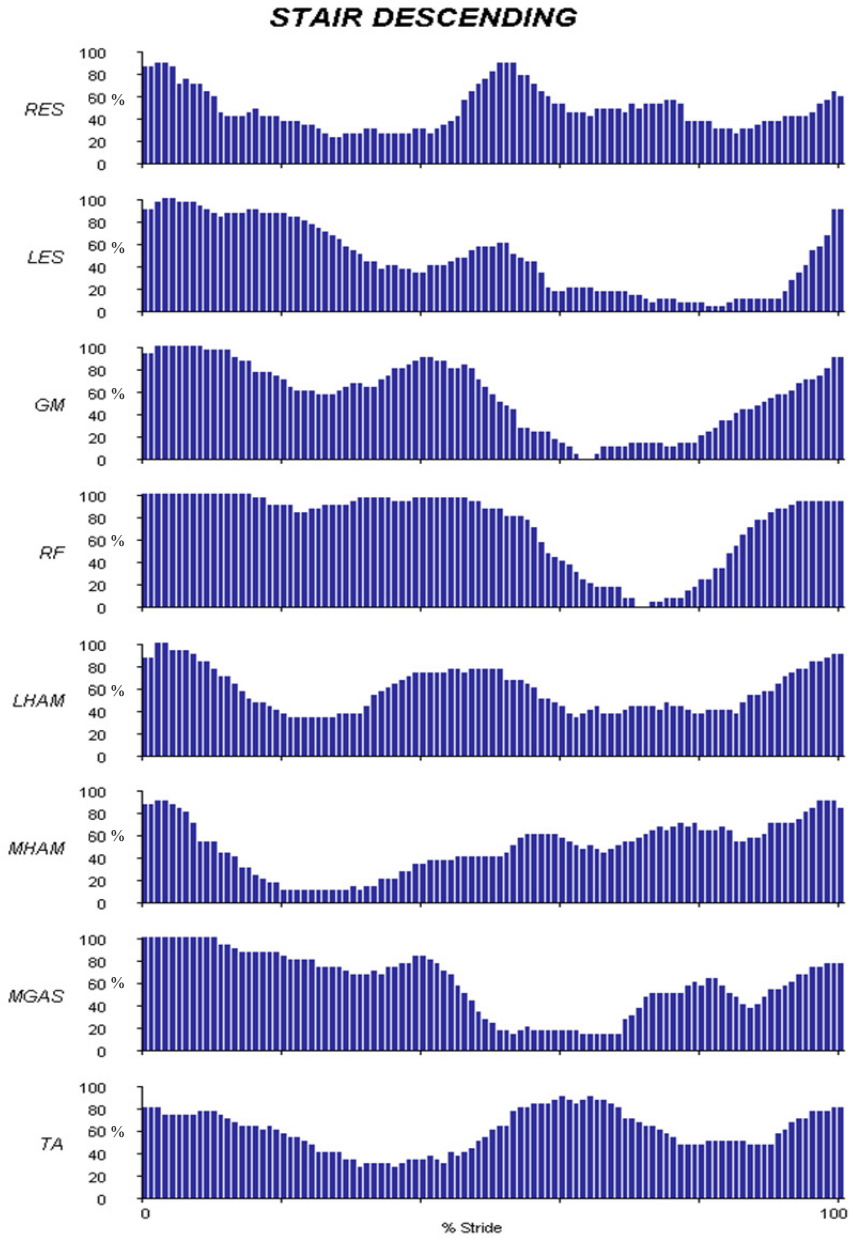


Fig. 4. On-off muscle timing during stair descending, as a function of the percentage of the gait cycle. For each muscle, the muscular activity is presented as percentage of subjects whose muscle is active at the specific percent of the stride considered. RES: right erector spinae, LES: left erector spinae, GM: gluteus medius, RF: rectus femoris, MHAM: medial hamstrings, LHAM: lateral hamstrings, MGAS: medial gastrocnemius, TA: tibialis anterior.

The application of the statistical detector to the large number of EMG trials collected in this study was completely automatic, once the parameters of the algorithm were tuned on the characteristics of the EMG signals. Detected phases of activity of the examined muscles were in most of the cases comparable with previous work for level walking at self-selected speed (Arsenault et al., 1986; Basmajian, 1978; Battye & Joseph, 1996; Kadaba et al., 1989; Yang & Winter, 1985; Perry, 1992; Wootten et al., 1990) and, when available, for stair ascending /descending (Basmajian, 1978; Joseph & Watson, 1967; Powers et al., 1997; Shiavi, 1981; Shinno, 1971; Townsed et al., 1978). Only in a few cases our results appeared to be different from previous findings by others (Andriacchi et al., 1980). The detection of the rectus femoris activity is problematic when one uses surface electrodes, as the activity of other heads of quadriceps muscle appears to affect the signal recorded from the rectus femoris (Perry, 1992; Nene et al., 1999).

Results for the gastrocnemius muscle appear to be consistent with the biomechanics of motion even if different from results previously reported in the literature. The activity that we detected at heel strike during stair ambulation is consistent with the need to control the forward momentum of the trunk. In interpreting the results for stair ambulation, it must be taken into account that differences with respect to previous studies may depend on external factors such as the staircase set-up.

Problems related to the presence of crosstalk were encountered in some cases. In the EMG signals recorded for the present study, cross-talk affected particularly the signals relative to the gastrocnemius and the tibialis anterior muscles at the beginning of the stride, during mid stance, and during the swing phase. If detected during the acquisition session, cross-talk could be reduced by decreasing the inter-electrode distance or by using electrode arrays such as the double differential electrode (C.J. De Luca & Merletti, 1988). When cross-talk is unavoidable during the signal acquisition, and evident from the shape and the timing of the burst, it is manually rejected from the on-off detection.

The clinical relevance of the analysis of muscle activation intervals has been established. Nevertheless, the use of a detector limited to the determination of the muscle activation onset and offset, disregarding the relative magnitude of the EMG signal, fails to recognize as pathological a muscle activation with correct timing but inappropriate intensity. In order to overcome this restriction, it was decided to perform also a gross amplitude quantification in five levels of intensity. Hence Fig. 5 shows, for various muscles, the mean amplitude values over the population. In this representation, amplitude values are color-coded. The black color represents high-intensity muscle activation, while the white color represents a condition in which the muscle is silent.

The purpose of this added information is mainly to provide a qualitative evaluation of the location of the peak of the EMG activity and of its sharpness in the gait cycle. According to Perry (Perry, 1992), this information could be of relevance for clinical applications.

A limit of this study is that the results presented here refer only to the most frequent pattern of activation of each muscle, leaving out less frequent patterns that may be observed. This is a point often disregarded by traditional approaches, though it may be of relevance in interpreting clinical data. In fact, during a continuous and cyclical motor task - such as gait or stair ambulation – each subject typically shows different muscular patterns of activation. This means that, during subsequent cycles of the same motor task, a specific muscle shows a different number of activations, possibly with different timing. This was demonstrated for level walking in a population of 100 children (Agostini, 2010). Recording and processing a considerable amount of cycles (tens to hundreds) for each subject makes it possible having a

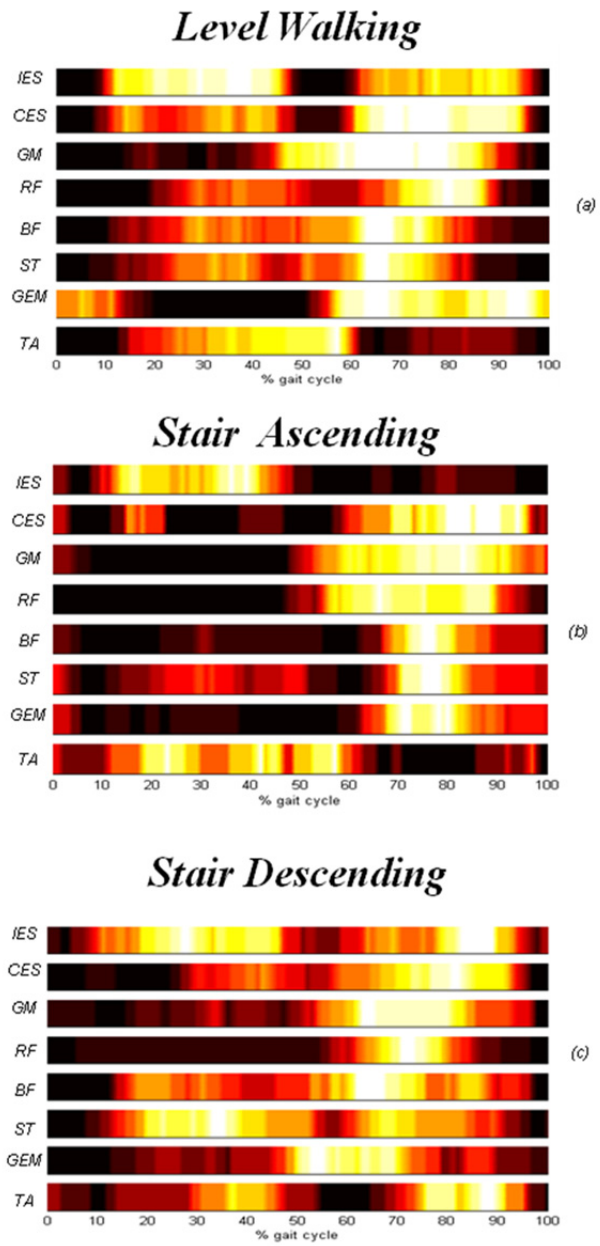


Fig. 5. Muscle timing during (a) level walking, (b) stair ascending and (c) stair descending. The EMG amplitude is color-coded with five colors that roughly codify the amplitude of the envelope of the EMG signal, with the purpose of allowing for the location of the peak amplitude occurring within each signal burst and of providing a rough estimate of its shape.

statistical description of the muscle activation patterns during performance of a motor task. Hence, one can calculate in which percentage of cycles a certain muscle pattern is observed. This allows one to capture more accurately the characteristics of the subject motor control and abilities. As an example, Fig. 6 reports the results obtained from the gait analysis of an adult volunteer who walked for 160 s. The figure refers to the EMG patterns obtained from the rectus femoris muscle during walking. The left plot of Fig. 6 reports the histogram of the number of activations observed during all the observed strides. In 54% of the strides there are three activations (most frequent pattern), in 22% of the strides there are only two activations, while in 21% of the strides there are four activations. The right plot of Fig. 6 shows the activation patterns for the three main activation modalities. When the muscle is active, the amplitude of the signal is color-coded using three levels: red means high activation amplitude (greater than 66% of the maximum root-mean-square value observed during walking), green means medium activation amplitude (between 33% and 66%) and yellow represents low activation amplitude (lower than 33%). The root-mean-square value of the signal is computed over subsequent windows lasting 50 ms.

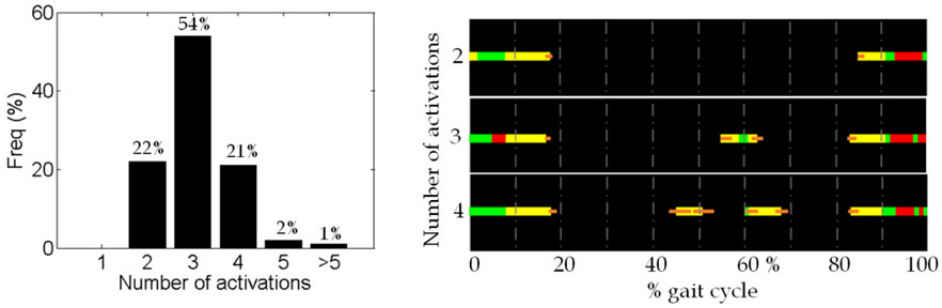


Fig. 6. Analysis of 80 strides recorded during level walking in a healthy adult subject. Left plot: histogram of the number of activations observed in the rectus femoris. Right plot: muscle activation patterns for each activation modality.

## 5. Conclusion

In order to define what is "pathological" it is necessary to have previously defined what is "normal". This was the aim of the study and the reason why we collected a reference dataset from healthy subjects. Due to the statistical properties of the EMG signal, detecting "on" and "off" timing of muscle activity is a challenging task, particularly when dealing with signals having poor SNR, as those often seen in clinical practice. The application of the double-threshold statistical algorithm has provided a reliable documentation of "normal" muscle timing during the motor tasks explored. The adaptive properties of this technique with respect to the variations of the SNR allowed us to estimate muscle activation intervals from all the EMG recordings.

Results obtained from this study can be relevant both to basic research, as reference for motor control pattern studies, and to clinical research, in order to document pathological patterns of muscle activation and their consistencies with kinematics and kinetics findings.

In future studies, the methodology herein explored may become a valuable tool for investigating the causes of motor impairment and for allowing an objective evaluation of the outcome of therapeutic approaches.

## 6. Acknowledgment

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