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Electromechanical delay in the vastus lateralis muscle during dynamic isometric contractions

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Summary. Electromechanical delay (EMD) values were obtained using a cross-correlation technique for a series of 14 repetitive submaximal dynamic isometric contractions of the vastus lateralis performed by five subjects. To avoid a phase lag, which is introduced with one-way filtering, the EMG was processed with a bi-directional application of a second-order Butterworth filter. A mean EMD value of 86 ms (SD = 5.1 ms) was found. Moreover, contraction and relaxation delays were computed and compared. There was a significant difference between the contraction and relaxation delays ($P < 0.005$). The mean contraction delay was 81.9 ms and the mean relaxation delay was 88.8 ms. Despite this significant difference, the computed contraction and relaxation delay values lie in the same range as the total phase lag, calculated with the cross-correlation technique. The magnitude of EMD values found supports the need to account for this delay when interpreting temporal aspects of patterns of intermuscular coordination.

Key words: Electromechanical delay — Contraction delay — Relaxation delay — Cross-correlation technique

Introduction

At various levels of observation electromyographic (EMG) data are used for analyses of human movement and to achieve insight into the control of movement. Particularly when EMG data are used as an indication of the activation patterns of separate muscles in relation to profiles of net moments in joints and joint powers, as obtained, for example, by means of inverse dynamics, it is imperative to have an impression of the temporal relation between an EMG pattern and the mechanical response of the muscle (e.g. Simonsen et al. 1985; de Koning et al. 1988). For example in studies of the actions of bi-articular muscles and their obvious highly functional coordination with mono-articular muscles (Ingen Schenau et al. 1987; Ingen Schenau 1989; Bobbert and Ingen Schenau 1988), the applied mechanical reasoning is to a large extent dependent on the reliability of assumptions concerning the phase shift between EMG and mechanical response. In publications on this subject the delay between the onset of EMG and the onset of mechanical response is usually referred to as electro-mechanical delay (EMD) (e.g. Cavanagh and Komi 1979; Norman and Komi 1979; Bell and Jacobs 1986; Grabiner 1986). Large discrepancies exist in the magnitude of EMD values reported. The EMD has been stated to lie between 30 ms and 100 ms (Cavanagh and Komi 1979), but for leg muscles higher (Morris 1977; Horita and Ishiko 1987) as well as lower EMD values (Muro and Nagata 1985; Moritani et al. 1987) are reported. Norman and Komi (1979) stated that the length of the EMD is primarily affected by the time necessary to stretch the series elastic component of a muscle to a point where muscle force can be detected and that factors such as initial muscle length, muscle fibre type and type of contraction influence this phase lag. However, the variety of reported EMD values can not totally be explained in the light of the above-mentioned factors. For instance, there are discrepancies reported between EMD values of the order of 300% for the same muscle and the same initial muscle length and type of contraction. Houston et al. (1988) found an EMD of 41 ms in the vastus lateralis muscle preceding isokinetic concentric contractions, while Horita and Ishiko (1987) found an EMD of 117.9 ms in the vastus lateralis under the same exercise conditions. In the above-mentioned studies the EMD value was defined as the time lag between the onset of EMG and the onset of mechanical response. The disadvantage of this procedure is that one should define the threshold levels of both signals. The studies were also predominantly focused on the rising limb of force pro-

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duction. When interpreting a temporally ordered time sequence of muscle activations and deactivations, one also needs information about the phase shift between both signals during relaxation. To achieve a global measure for the phase shift between EMG and force, including both the rising and falling limbs, Olney and Winter (1985) proposed using a low-pass filter procedure with a cut-off frequency adapted to the force response. The phase shift of the linear envelope thus obtained is dependent on the order of the filter and on the cut-off frequency. Olney and Winter proposed a second-order filter with cut-off frequencies adapted to the EMG data of individual muscles. For leg muscles the cut-off frequencies varied from 1.25 Hz for the soleus to 2.15 Hz for the rectus femoris. Since a second-order low-pass filter introduces a phase shift of 90 degrees at the -3 dB point (Papoulis 1984), the cut-off frequencies correspond to phase shifts that vary from 116 ms to 200 ms. The only disadvantage of this procedure is that the location of the EMG patterns in real time is lost. For the interpretation of EMG data in the light of control of human movement, more insight into the magnitude of EMD is necessary.

The purpose of this study is to present an alternative method, based on cross-correlation techniques, where a linear envelope is constructed without phase shift with respect to the raw EMG data. Subsequently the phase difference between this linear envelope and the responding force is established by cross-correlating the linear envelope with the force in repetitive exercises. Moreover, the time shifts of the rising and falling limbs were determined separately in order to assess differences between activation and relaxation.

Methods

Five healthy subjects, two female and three male, volunteered to participate in this study. Their mean age was 25.0 (SD = 2.5) years. The subjects were seated in a chair with their back supported and with straps across the trunk and thigh for fixation of the upper leg. The hip and knee angles were both 90°. The left lower leg was attached at its distal end (just above the malleolus medialis) with an inextensible ankle-tape to a force transducer (Load-cell, LM-100ka). After a warming-up, maximum voluntary contraction (MVC) was defined according to the Caldwell protocol (1974). After a recovery period of 2 min, the subjects performed, guided by the sound of a metronome (1.5 Hz), dynamic submaximal (approximately 50% MVC) isometric left knee extensor contractions for 25 s (approximately 16 contractions). Two such trials were analysed per subject. The subjects received visual feedback from an oscilloscope, on which the required as well as the performed forces were shown. The force signal was stored on tape (AMPEX) with a portable data recorder (TEAC SR-70, bandwidth 0–625 Hz), through a digital voltmeter (3465A digital multimeter, Hewlett Packard). EMG activity during contractions was recorded from the vastus lateralis. Surface electrodes (silver chloride, Sentry medical products, diameter: 0.13 m) were positioned at the approximate geometrical centre of the muscle belly with an inter-electrode distance of 3 cm. The EMG signal was also stored on tape, in parallel with the force signal. Simultaneously with the recording, the raw EMG and the force signal were displayed on a writer (Gould, ES1000, paper speed 25 mm/s) for control. Raw EMG data were filtered (bandpass 5–100 Hz, 24 dB/oct), rectified and AD-converted (12 bits, sample frequency 400 Hz). The force signal was AD-converted with the same sample frequency. Both signals were filtered with a bi-directional application of a second-order Butterworth digital five-point low-pass filter with a cut-off frequency of 3 Hz (net: fourth-order, zero-lag, $f_c = 2.4$ Hz). The first contraction of a series of contractions was removed because of the start-up transient of the filter. Using a series of 14 contractions, the total phase shift between EMG and force signals was calculated by cross-correlating force and the EMG linear envelope on a Cyper 995 E main-frame computer. The EMD was defined as the delay at which the highest correlation was found. In order to evaluate possible differences in the time shift between EMG and force data between periods of increasing and decreasing activity, the average-amplitude points on the rising and falling limbs of the EMG and force signals were computed and compared. The time lag between the average-amplitude points of the rising limbs of the EMG and force signals was called the ‘activation delay’, that of the falling limbs the ‘relaxation delay’ (Fig. 1). Conventional statistical methods were employed to calculate the mean and standard deviation and the standard error of the mean. Differences between the time-lag values of the rising and falling limbs of the EMG and force were tested using the paired t-test. We considered $P < 0.05$ to be significant.

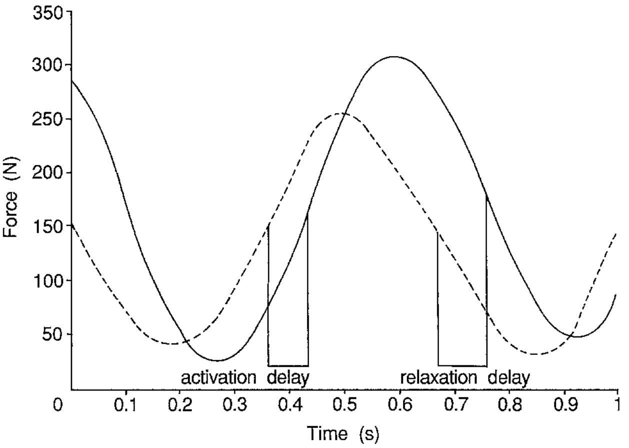

Fig. 1. A time plot of bi-directional filtered EMG with the corresponding force. The *vertical lines* indicate the average-amplitude points of the rising and falling limbs of force and EMG. The time lag between the average-amplitude points of the rising limbs of force and EMG was called the activation delay, that of the falling limbs the relaxation delay

Results

The mean maximal isometric force from all persons was 600 N (SD = 38.1 N). The two series of approximately 14 contractions each, appeared to be performed at 48.3% (SD = 2.9%) of the MVC. A typical time plot of the force-rectified raw EMG relationship resulting from dynamic isometric contractions is shown in Fig. 2. Note the phase lag between the EMG and the force signal. Figure 3 shows an example of the cross-correlation function between the filtered EMG and force signals against delay time. Results of the cross-correlation between the filtered signals are presented in Table 1. The mean EMD value appears to be 86 ms (SD = 5.1 ms). Table 2 gives the results of the rectified raw EMG signals directly cross-correlated with the forces. Approximately the same EMD values show that the bi-directional EMG filtering procedure indeed does not intro-

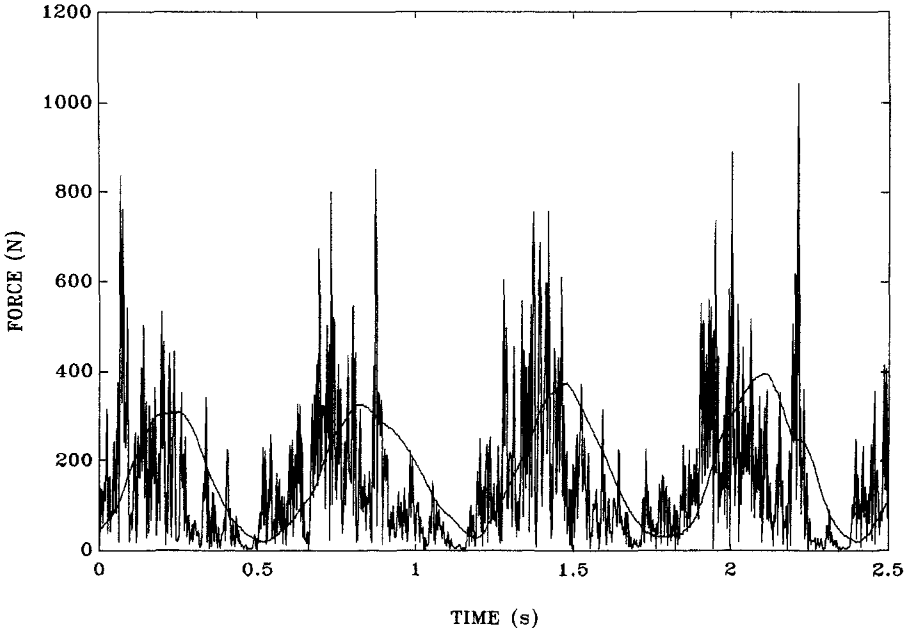

Fig. 2. A typical example of raw, rectified EMG of the vastus lateralis during repeated isometric contractions. The *solid curve* represents the corresponding force. Note the time shift between the EMG and force signals

duce a phase lag. Table 3 shows the mean time lag between the average-amplitude points on the rising and falling limbs of the filtered EMG and force signals in the first series of contractions. The relaxation delays have slightly higher values than the contraction delays. The mean contraction delay is 81.9 ms (SD = 14.8 ms) and the mean relaxation delay is 88.8 ms (SD = 18.7 ms). With the paired *t*-test, the contraction and relaxation delays for each contraction from all subjects were compared. There was a significant difference between mean contraction and relaxation delays [$t(66) = -3.09$, $P = 0.0032$]. The mean relaxation delay being 8.4% longer than the mean contraction delay.

Discussion

In this study the phase shift between EMG and mechanical response of the vastus lateralis was calculated by a cross-correlation technique. It should be noted that there is a limitation of the cross-correlation technique: it can only be used if it can be assumed that, in the experimental setting used, the measured force/torque is produced by the measured muscle; antagonistic coactivation must be absent.

The philosophy behind this study was that for the interpretation of EMG data in the light of control of human movement, one needs an impression of the temporal relationship between EMG pattern and the me-

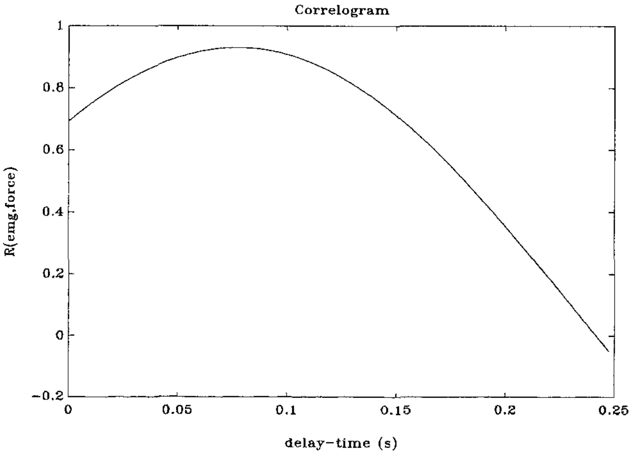

Fig. 3. A typical example of the cross-correlation function between the bi-directional filtered EMG and the force signal against delay time

Table 1. Delay between the filtered EMG and force signals at the highest cross-correlation in a series of 14 dynamic isometric contractions

Subject	Cross-correlation coefficient	EMD (ms)
1	0.955	87.5
2	0.957	77.5
3	0.959	90.0
4	0.938	90.0
5	0.953	85.0
$\bar{X}$ (SD)	0.952 (0.0075)	86.0 (5.1)

EMD, Electromechanical delay

Table 2. Delay between rectified raw electromyograph (EMG) data and force signals at the highest cross-correlation

Subject	Cross-correlation coefficient	EMD (ms)
1	0.504	82.5
2	0.531	77.7
3	0.593	90.0
4	0.575	90.0
5	0.568	82.5
$\bar{X}$ (SD)	0.554 (0.032)	84.5 (4.8)

Table 3. Mean time lag and standard deviation between the average-amplitude point on the rising and falling limbs for each contraction from each subject of the filtered EMG and force signals; the total mean and standard deviation for each contraction from all subjects is also given

Type of delay	Subject					Total mean
	1	2	3	4	5	
Mean contraction delay (ms) SD	80.6 14.1 ($n=13$)	74.3 14.9 ($n=14$)	82.9 13.6 ($n=13$)	89.6 16.8 ($n=13$)	82.3 12.7 ($n=14$)	81.9 14.8 ($n=67$)
Mean relaxation delay (ms) SD	84.2 21.9 ($n=13$)	85.7 13.7 ($n=14$)	99.0 14.6 ($n=13$)	87.7 24.2 ($n=13$)	87.3 16.7 ($n=14$)	88.8 18.7 ($n=67$)

chanical response of the muscle. As such, one should construct a measure for muscle activation in real time. This means that one has to smooth the rectified raw EMG without introducing any phase lag. To establish this a bi-directional application of a second-order filter was used. When cross-correlating the curves, as obtained with the filtering, with the corresponding forces, a mean phase shift of 86 ms was found (Table 1). The magnitude of this phase shift is in agreement with the studies of Inman et al. (1952) and Sherif et al. (1983). The same time lag is obtained if the unsmoothed, rectified EMG signals were directly correlated with the forces (Table 2). This implies that our filtering method did not introduce a time lag. Olney and Winter (1985) have shown that certain muscle-specific cut-off frequencies of a second-order Butterworth low-pass filter can lead to linear envelope profiles, which are close to each muscle's predicted force profile. For the vastus lateralis the reported cut-off frequency is 2 Hz, which correspond to a time shift of 125 ms (Papoulis 1984). As a result of this EMG-processing method, the location of the EMG patterns in real time is lost. Application of Olney and Winter's method (one-way filtering with a second-order Butterworth, $f_c = 2$ Hz) to the present rectified raw EMG data leads to linear envelope curves that can be compared to ours. When cross-correlating these curves with the corresponding forces we found a mean shift of -39 ms (Table 4), which means that the total mean time lag equals 125 ms (Tables 1 and 4). As such, the one-way-filtered EMG as proposed by Olney and Winter (1985) lies behind the corresponding forces (Fig. 4). The 39-ms overestimation of Olney and Winter's method does not disqualify their approach, at most it demonstrates that our subjects have faster muscles than the subjects of Olney and Winter.

Table 4. Delay between one-way filtered EMG (second-order Butterworth low-pass filter, cut-off frequency 2 Hz) and force signals at the highest cross-correlation

Subject	Cross-correlation coefficient	EMD (ms)
1	0.952	−37.5
2	0.941	−45.0
3	0.959	−35.0
4	0.929	−35.0
5	0.949	−42.5
$\bar{X}$ (SD)	0.946 (0.001)	−39.0 (4.1)

The large difference in the EMD time obtained, in comparison to those reported by Muro and Nagata (1985) and Moritani et al. (1987) (10.9 ms and 18.8 ms, respectively), is in our opinion attributable to dissimilarity in the methods used. The EMD recorded by Muro and Nagata and Moritani et al. was determined as the time difference between the onset of the M-wave and the gastrocnemius muscle twitch force, obtained during supramaximal electrical stimulation of the posterior tibial nerve. Possibly the definition of the threshold levels of both signals and the fact that supramaximal electrical stimulation is a non-physiological input to the muscle explain the great discrepancy between their result and ours.

A significant difference was found between the contraction and relaxation delays (Table 3). Ralston et al. (1976) and Hof (1984) also reported longer relaxation delays compared with contraction delays within one contraction. They found lags between the onset of EMG and force of the order of 40–50 ms, and between force decay and cessation of EMG of the order of 160–250 ms. According to Viitasalo and Komi (1981) the relaxation delays reflect the rate of calcium removal from actomyosin cross-bridges by the sarcoplasmic reticulum. With this explanation it is not suprising that in the present study only small differences in contraction and relaxation delays were found. In the studies of Ralston et al. (1976) and Hof (1984) a sudden contraction of a muscle was followed by a relatively long time for complete relaxation of that muscle. In our study a repeated exercise protocol (1.5 Hz) was used. In such a protocol the muscle has no time to relax completely and full calcium removal is not possible.

If one realizes that the push-off phase of a vertical jump or a push-off in speed skating lasts about 200 ms, it is clear that one should incorporate EMD values when interpreting the temporal structure in patterns of muscle activation in the light of kinematic and kinetic data, as performed by example Bobbert and Ingen Schenau (1988) and Koning et al. (1988). The same, of course, is true for cycling. At a cycling speed of 90 rpm

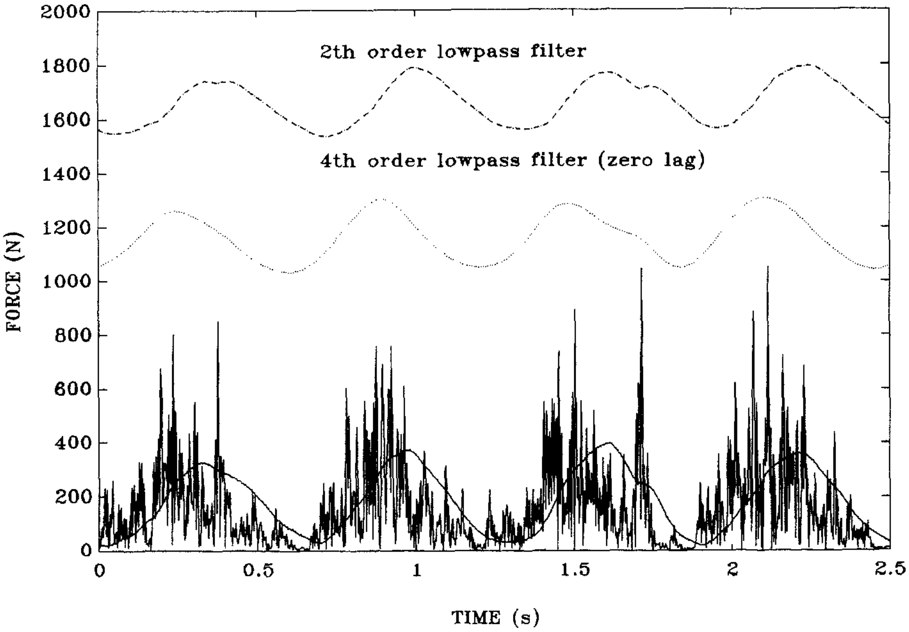

Fig. 4. An example of the raw, rectified EMG of the vastus lateralis during repeated isometric contractions. The *solid curve* represents the corresponding force. —·—, Results of the application of a 3-Hz bi-directional filter (no phase lag with respect to the raw EMG); ..., the result of the filter procedure as proposed by Olney and Winter (1985)

and EMD of 86 ms corresponds to a crank angle of 46°. When incorporating this value in the EMG linear envelopes it is striking that mono-articular muscles appear to be only active in the periods when they are in the position to shorten (Ingen Schenau 1989). This might point to an important principle in the organisation of movement. It was hypothesized that mono-articular muscles might predominantly be controlled (or their control might be learned) on the basis of muscle length information, while bi-articular muscles have to warrant the demands of a specific task (see Ingen Schenau 1989, for more details). Clearly we will need more information about the nature of the EMD and its possible dependence on, for example, muscle contraction, velocity and fatigue for these types of analyses, and to test such hypotheses with respect to the organization of movement.

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