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# Electromechanical delay in the tibialis anterior muscle during time-varying ankle dorsiflexion

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*Abstract*— **We evaluated the electromechanical delay (EMD) for the tibialis anterior (TA) muscle during the performance of time-varying ankle dorsiflexions. Subjects were asked to track a sinusoidal trajectory, for a range of amplitudes and frequencies. Motor unit (MU) action potential trains were identified from surface electromyography (EMG) decomposition and summed to generate the cumulative spike train (CST). CST and the exerted force were cross-correlated to identify the delay between the CST and force, which was considered as an estimate of the EMD. The results showed that the EMD decreased logarithmically with the increase in the slope of the force produced.**

## I. INTRODUCTION

Electromechanical delay (EMD) is typically described as the time lag between the electrical activation of the muscle and the onset of the exerted force [1]. This delay includes the electrochemical and mechanical events from neuromuscular activation to force transmission to the bone [1]. The delay in electromechanical coupling has been usually reported to be between 30 and 100 ms, although important variations can be found across studies [2], [3], [4], [5]. For instance, very short delays (around 8.5 ms) have been observed for the gastrocnemius muscle during the performance of isometric contractions [4]. On the other hand, large delays were measured in the biceps femoris muscle in similar conditions [5]. These large differences could be the result of many factors, such as the recording methods employed to detect muscle onset that range from dynamometers or force plates [6], mechanomyographic (MMG) sensors [7] or high-frame ultrasound (US) [2].

Research over the past two decades has provided knowledge on how EMD can be assessed to extract information of the neuromuscular behaviour. For instance, EMD has been computed to find differences between gender, age, fatigue and contraction level [4]. EMD was reported to increase with contraction level, age and fatigue, while gender did not have a significant influence on EMD [4]. This was consistent with other works, where EMD did not differ significantly between males and females, while force production (RFP) and musculotendinous stiffness (MTS) were greater in male subjects [5]. Most studies agree that EMD significantly increases with fatigue [8], [9].

Current literature on EMD pays particular attention to muscle contractions initiated from a rest state. Muscular electrical onset is generally measured through bipolar surface electrodes. When the movement is voluntary initiated, this information is compared to the onset of the exerted force usually recorded from force plates or dynamometers. However, when the performed movement is continuous it is difficult to estimate this delay due to the absence of an actual movement onset.

In this work, we assess EMD during the performance of sinusoidal ankle dorsiflexions. For this reason, we propose a new way of measuring EMD. It is based on the peak of cross-correlation between an estimate of the neural drive to the muscle and force. This technique may achieve a more precise estimation of the portion of EMD due to MU recruitment while preserving the portion related to the mechanical properties of the muscle and the tendon. A similar approach was used previously [11] but was limited to the cross-correlation of the activity of only individual MUs with force. Conversely, here we estimate the neural drive as the cumulative set of discharges of several MUs. The use of the cumulative activity better describes the behavior of the common pool of neurons in charge of force generation [12].

We applied this methodology to the tibialis anterior (TA) muscle during ankle dorsiflexion to compare our results with previous findings. Additionally, we investigated the dependence of EMD on the slope of the exerted force.

## II. MATERIALS AND METHODS

### A. Experimental Setup

8 able-bodied subjects (age: 27.9±2.8 years old, all male) were recruited for this study. Subjects were seated comfortably in a Biodex System 3 (Biodex Medical Systems) facing a screen which provided visual feedback. Isometric ankle

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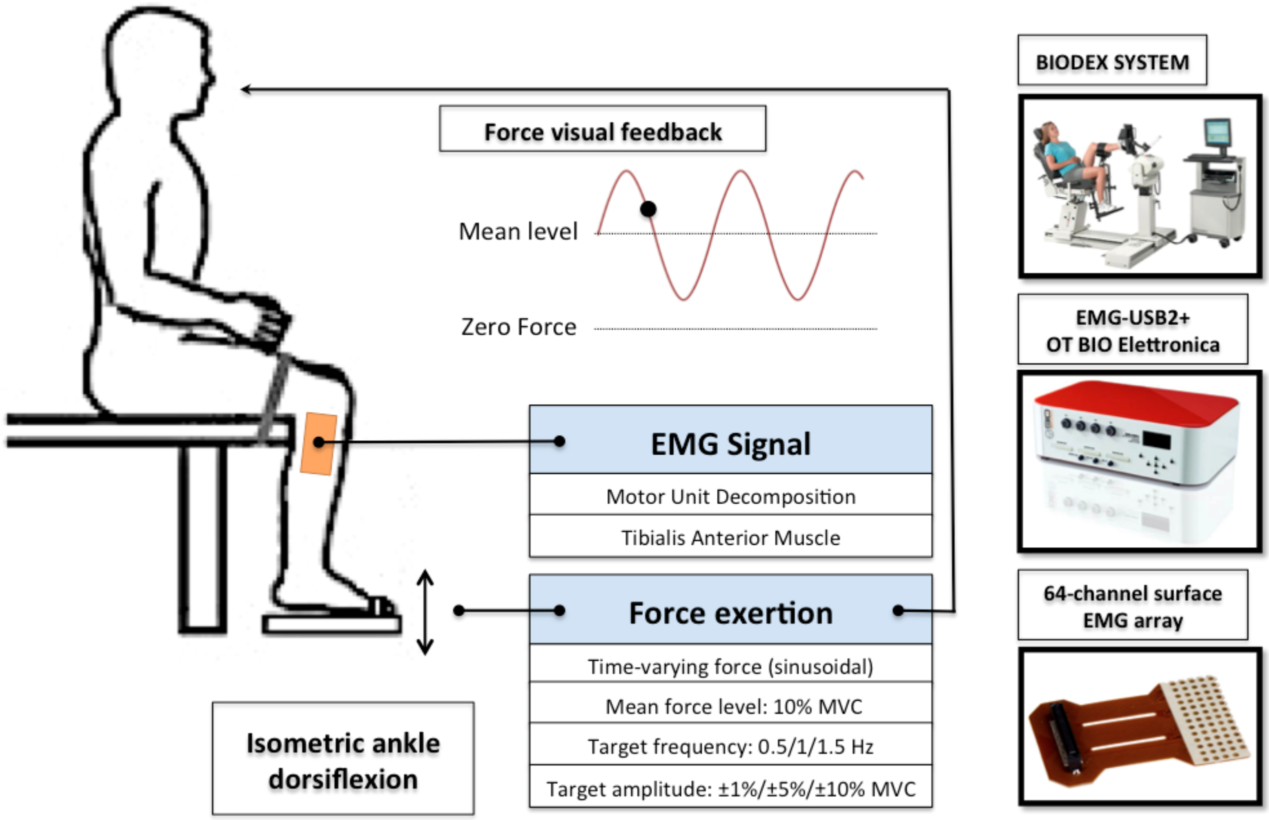



Fig. 1: Experimental setup. Subjects performed isometric ankle dorsiflexions while EMG and force signals were concurrently recorded. Subjects were asked to track the sinusoidal visual cue while controlling a black dot that moved proportionally to the exerted force.

dorsiflexion forces were exerted on a dynamometer fixed to the dominant leg (right leg in all cases). The knee was flexed at a 120° angle and the ankle at a 140° angle. For each trial, subjects were asked to track sinusoidal trajectories. The mean level of the target trajectories was fixed to a 10% of the previously measured maximum voluntary contraction (MVC). Three target frequencies (0.5, 1 and 1.5 Hz) and three amplitudes ($\pm 1$, $\pm 5$ and $\pm 10\%$ MVC) were chosen making a total of 9 different conditions. For each condition, isometric contractions were performed for 2 min each. Electromyographic (EMG) signals were measured along with the exerted force.

### B. Motor unit decomposition

Surface electromyography (EMG) signals were measured from the Tibialis Anterior (TA) muscle using a 64-channel surface EMG array (OT Bioelettronica). The rectangular array was placed over the belly of the muscle. EMG signals were band-pass filtered between 10 and 500 Hz and decomposed into motor units using a convolutive blind source separation technique [13]. This method provides a solid framework for the decomposition of multi-channel non-invasive EMG signals that allows the study of the behaviour of a relatively large number of concurrently active motor units.

Only motor units with a stable firing rate were selected (pauses of less than 2 seconds). Additionally, motor units with less than $<30$ dB pulse/noise ratio (PNR) were discarded. This metric is considered a robust and reliable indicator of accuracy in MU identification [14].

Finally, single spike trains from all extracted motor units were summed to generate the muscle Cumulative Spike Trains (CSTs) [12]. CST was an estimate of the neural drive to the muscle.

### C. Data analysis

To compute MU/force delays a narrow band-pass filter (bandwidth of 0.2 Hz) centred on the target frequency was applied to CST and force signals for each condition (4th order zero-phase Butterworth filter). This was done to ensure that activity in other frequency bands does not affect the correct computation of the delay. After filtering, paired data (CST and force) were divided into one-cycle epochs and the cross-correlation was computed for each pair and then averaged across all epochs. Cross-correlation plots were computed for each subject and condition showing the behavior of the correlation between CST and force in terms of correlation amplitude and EMD.

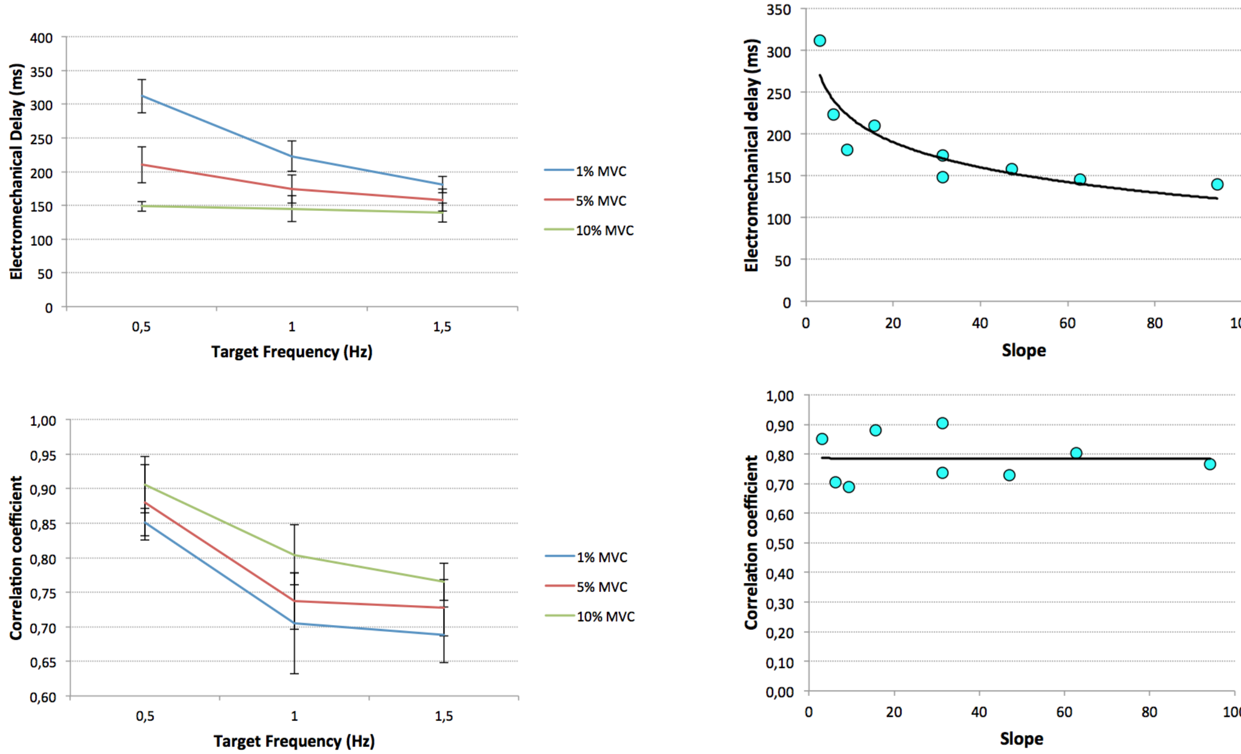
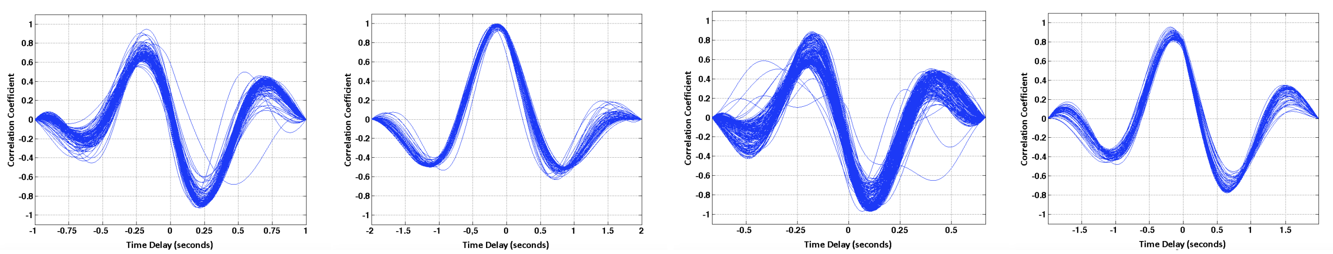



Fig. 2: EMD obtained for each condition (mean±STD) (top-left). Relationship between EMD and force slope (top-right). MU/Force correlation for each condition (mean±STD) (bottom-left). Relationship between MU/Force correlation and force slope (bottom-right)




(a) A02, 1 Hz, ±1% MVC (b) A02, 0.5 Hz, ±10% MVC (c) A05,1.5 Hz, ±1% MVC (d) A05, 0.5 Hz, ±10% MVC

Fig. 3: MU/Force cross-correlation plots of two representative subjects (A02 and A05). Cross-correlation plots shown for the best and the worst average correlation coefficient.

Additionally, force slope has been studied to assess its relationship with EMD and correlation coefficients. Force slope can be defined as the slope of the sinusoidal curve at the beginning of each cycle. For each condition, the tracked sinusoidal can be defined as:

$$x = A \cdot sin(2\pi f t) \tag{1}$$

where $A$ is the amplitude of the sinusoidal, $f$ is the target frequency and $t$ is the time.

Force slope was then computed as the value of the derivative at time zero:

$$s = A \cdot 2\pi f \cdot cos(2\pi f t) \underset{t=0}{\rightarrow} s = A \cdot 2\pi f \tag{2}$$

From this equation it can be inferred that force slope is proportional to both target amplitude and frequency.

## III. Results and Discussion

EMD ranged from 112 ms to 361 ms (Figure 2, top-left). On average, maximum and minimum EMD were approximately 312 ms and 139 ms, respectively. In contrast to earlier findings, these values are greater than what is obtained with standard EMD computation [2], [3], [4], [5], but more similar to [11]. These differences may partly be explained by the methodology we employed to compute the MU/Force delay but also by the nature of the performed movement, as continuous time-varying movements differ in the way motor units are recruited. It is also noticeable that standard deviations were very small (<30 ms in all cases), suggesting that the EMD estimates were consistent across different subjects.

Another interesting finding is how EMD behaved regarding the two main parameters that are changed in the force tracking feedback. Figure 2 (top-left) shows that EMD decreased with both amplitude and frequency. This is partly consistent with previous findings [11] where gain and delay dropped between EMG modulation and force oscillation for increasing target frequency. Here, we also demonstrate that the delay decreased with increasing target amplitude. As a consequence, we showed a dependency on the force slope rather than on the two individual parameters. Both parameters, although tuned independently, proportionally influenced the force slope. Figure 2 (top-right) indicates that the EMD dependence on force slope followed a logarithmic curve ($R^2 = 0.81$).

EMD may be dependent on the slope of the force exerted rather than amplitude or frequency per se due to the relation between muscular torque and neural drive. During in-vivo voluntary force contractions MU discharge rate increases with force [15] and speed of contraction [16] thus the shorter EMD may be expected due to concurrent increase in MU discharge rate and MU recruitment. Indeed, MU recruitment is proportional to MU contraction time, so the recruitment of larger MU will shorten the EMD.

When analyzing the values for correlation between CST and force, we observed that they were better correlated for high amplitudes and low frequencies. These findings suggest that an easier tracking of the sinusoidal cue led to a better correlation (Figure 2, bottom-left). This can be observed in Figure 3. The cross-correlation plots for two representative subjects (A02 and A05) are shown. The variability of the amplitude of the correlation curves increased from conditions of higher target amplitudes and lower target frequencies (Figures 3b and 3d) to conditions of lower target amplitudes and higher target frequencies (Figures 3a and 3c). However, the correlation values did not depend on force slope (see Figure 2, bottom-right).

## IV. Conclusion

In this paper, we have analyzed how electromechanical delay on the tibialis anterior muscle changes during the performance of ankle dorsiflexion while tracking a sinusoidal trajectory that changes in amplitude and frequency. The results showed that EMD was dependent on the force slope. This may indicate that EMD is related to the increase of MU discharge rate and MU recruitment. A limitation of this study is that it does not discriminate between the inner components of EMD. Further work could assess how much of the EMD is due to the mechanical properties of the musculotendon and how much is due to MU recruitment itself.

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