DELAY ESTIMATION BETWEEN EEG AND EMG VIA COHERENCE WITH TIME LAG

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ABSTRACT

The traditional way to estimate the time delay between the motor cortex and the periphery is based on the estimation of the slope of the phase of the cross spectral density between motor cortex electroencephalogram (EEG) and electromyography (EMG) signals recorded synchronously during a motor control task. There are several issues that could make the delay estimation using this method subject to errors, leading frequently to estimates which are in disagreement with underlying physiology. This study introduces cortico-muscular coherence with time lag (CMCTL) function and proposes a method for estimating the delay based on finding its local maxima. We further address the issue of the interpretation of such time delay in multi-path propagation systems. Delay estimates obtained using the proposed method are more consistent compared with results obtained using the phase method and in a better agreement with physiological facts.

***Index Terms*—** Cortico-muscular coherence, EEG, EMG, time delay.

1. INTRODUCTION

Cortico-muscular coherence (CMC) has been used extensively as tool for studying the functional coupling between the motor cortex and muscle activity [1–5], since the initial evidence was found of significant coherence between the motor cortex EEG and surface electromyography (sEMG) during constant isometric contractions [6]. There is, however, a time delay between coupled EEG and EMG signals, which if not accounted for, may decrease the level of coherence [7], and thus make the cortico-muscular coupling difficult or even impossible to detect. This time delay that needs to be compensated for is challenging to estimate. In addition to its relevance to enhancing the CMC, knowing the time delay between the motor cortex and the periphery can reveal important information about the communication between motor cortex and muscles by characterising the direction of information propagation and/or by differentiating the cortico-spinal pathways via which the activity is transmitted.

A method which is widely used for identification of time delays in biological systems is based on the estimation of the slope of the phase of the cross spectral density of considered processes [8–11]. However, it has produced conflicting results [2, 12–14]. There are several issues that could make this method to lead to erroneous results. Firstly, the slope of the phase spectrum is well defined only if the system is a linear-phase system, which is generally not satisfied by neural systems. Secondly, there could be more than one event in the observation period and the delay of each event could be different. Besides, the results are sensitive to the frequency range chosen for the linear regression and the degree of linearity in that frequency range [4]. Furthermore, until recently it has been common practice in CMC analysis to perform EMG rectification before the estimation of the delay between EEG and EMG events. However, that approach has recently been challenged by some groups [15–18] who point out that nonlinear processing generates spurious frequencies.

Govindan *et al.* proposed estimating the delay as the time offset between EEG and EMG signal which maximises their coherence [19]. Their method involved band-pass filtering and rectification of EMG signals and Fourier analysis over relatively long segments. When applied to stationary tremor events it produced some results that could not be supported by underlying physiology. The authors conclude that further work is needed to make the method applicable to non-stationary events. To address the non-stationarity, we build on their work and propose cortico-muscular coherence with time lag (CMCTL) in the domain of short-time Fourier transform [20] which uses much shorter (an order of magnitude) analysis windows, and involves all pairs of offsets of EEG and EMG signals rather than pairs where only of the signals is shifted, while the other is kept at a reference location [19]. Further, we propose to remove rectification and band-pass filtering as preprocessing steps. Finally, we address the interpretation of the notion of time delay in multi-path propagation scenarios. The method is then applied to data collected in a motor control task, and the results are mutually consistent as well as consistent with physiological findings.

2. METHODS

2.1. Simplified Model of Motor Control System

Cortical events propagate to the periphery [1, 4], and vice versa, sensori-motor cortex receives peripheral input [21].

Cortical activity is transmitted to the motor neurons within the spinal cord via the corticospinal tract, which contains nerve fibres that introduce different delays and attenuations. Each motorneurone innervates multiple fibres within the muscle comprising a motor unit. The response $y_i(t)$ of a motor unit i can thus be represented as a linear combination of delayed and attenuated versions of the cortical signal $x(t)$, that is

$$y_i(t) = \sum_{k=1}^{K_i} \alpha_{i,k} x(t - T_i - \tau_{i,k}) \; , \tag{1}$$

where $a_{i,k}$ are attenuations, while T_i and $\tau_{i,k}$ are delays of individual nerve fibres, defined so that T_i is equal to the minimal delay within the motor unit. Within the pick up area of an electrode, there are several motor units that would be recruited by the same cortical activity [22–25]. Therefore, sEMG signal $y(t)$ is a linear combination of several motor unit signals, as well as signals unrelated to the considered cortical activity, which we will collectively refer to as noise. Surface EMG signal thus has the form

$$y(t) = \sum_{i=1}^{I} \sum_{k=1}^{K_i} \beta_i \alpha_{i,k} x(t - T_i - \tau_{i,k}) + n(t) \; , \tag{2}$$

where β_i factors represent the attenuations of the pathways between particular motor units and the electrode, while $n(t)$ is the noise. To simplify the notation, in the following we will express the above model as

$$y(t) = \sum_{i=1}^{N} b_i x(t - D_i) + n(t) \; . \tag{3}$$

where b_i and D_i are the attenuations and time delays along each pathway, respectively.

2.2. Cortico-Muscular Coherence with Time Lag

The coherence between non-stationary processes can be calculated in the domain of short-time Fourier transform (STFT) [20], where the power spectral densities and their cross spectral density are estimated using windowed processes, which separates the nonstationary processes into a number of shorter time segments within which the statistical properties stay fairly constant. Usually, CMC is calculated between EEG and EMG signals which have been recorded simultaneously. To address the misalignment of coupled events, in this study the coherence patterns are considered between time-shifted version of EEG and EMG signals. Thus we propose the following cortico-muscular coherence with time lag (CMCTL) function

$$C_{xy}(t_c, \tau_1, \tau_2, \omega) = \frac{|\hat{S}_{xy}(t_c + \tau_1, t_c + \tau_2, \omega)|^2}{\hat{S}_{xx}(t_c + \tau_1, \omega) \hat{S}_{yy}(t_c + \tau_2, \omega)} \; . \tag{4}$$

Here the power spectral density of $x(t)$ is estimated as:

$$\hat{S}_{xx}(t_c + \tau_1, \omega) = \frac{1}{L} \sum_{n=1}^{L} |X_n(t_c + \tau_1, \omega)|^2 \tag{5}$$

where $X_n(t_c + \tau_1, \omega)$ is the STFT of the n-th trial of $x(t)$, shifted in time by τ_1, corresponding to the STFT window centred at t_c, and L is the number of trials. The power spectral density of $y(t)$ is estimated analogously, and finally their cross-spectral density is estimated as

$$\hat{S}_{xy}(t_c{+}\tau_1, t_c{+}\tau_2, \omega) = \frac{1}{L} \sum_{n=1}^{L} X_n(t_c{+}\tau_1, \omega) Y_n^*(t_c{+}\tau_2, \omega) \tag{6}$$

The time lag between these two processes introduced in CMCTL is thus $\tau = \tau_2 - \tau_1$.

2.3. Delay Estimation and Its Interpretation

To illustrate the effect of time delay on coherence estimation consider the scenario where $y(t) = bx(t - D) + n(t)$. If the coherence between $x(t)$ and $y(t)$ is estimated over a finite window of duration T, then the delay will cause a bias, and the estimated coherence $E[\hat{C}_{xy}(\omega)]$ will be related to the maximal coherence $C_{max}(\omega)$ as [7]

$$\frac{E[\hat{C}_{xy}(\omega)] - C_{max}(\omega)}{C_{max}(\omega)} \approx -\frac{2|D|}{T} + \left(\frac{|D|}{T}\right)^2 \; . \tag{7}$$

Apparently, the coherence will be maximised if in the estimation a time lag D_s is introduced (with appropriate sign) in one of the signals which is equal to the delay D. Systems which involve signalling over multiple paths as given in (3) blur the notion of delay. For such systems we propose to introduce the notion of the *global delay*, D_g, and define it, in analogy with the single-path case, as the time lag between the two processes corresponding to a local maximum of $C_{xy}(t_c, \tau_1, \tau_2, \omega)$.

Towards gaining the intuition about the physical meaning of the delay defined in this manner, consider introducing a shift D_s in $y(t)$ as given by (3). It can be shown that the estimated cross-spectral density between the shifted version of $y(t)$, $y(t+D_s) = \sum_{i=1}^{N} b_i x(t - D_i + D_s) + n(t)$ and $x(t)$ has the form

$$\hat{S}_{xy}(\omega) \approx \left(\sum_{i=1}^{N} b_i \big(1 - \frac{|D_i - D_s|}{T}\big) e^{j\omega(D_i - D_s)} \right) S_{xx}(\omega) \; . \tag{8}$$

In this scenario, the estimated coherence is related to the maximum coherence according to

$$\frac{E[\hat{C}_{xy}(\omega)] - C_{max}(\omega)}{C_{max}(\omega)} \approx \frac{\left| \sum_{i=1}^{N} b_i \big(1 - \frac{|D_i - D_s|}{T}\big) e^{j\omega D_i} \right|^2}{\left| \sum_{i=1}^{N} b_i \big(1 - \frac{|D_i - D_g|}{T}\big) e^{j\omega D_i} \right|^2} - 1 \; . \tag{9}$$

As this formula does not appear analytically tractable, we resort to simulations. Fig. 1 shows results of simulations of the above formula when D_i are generated according to a Gaussian distribution. We find that D_g which maximises the coherence corresponds to the mean of the distribution.

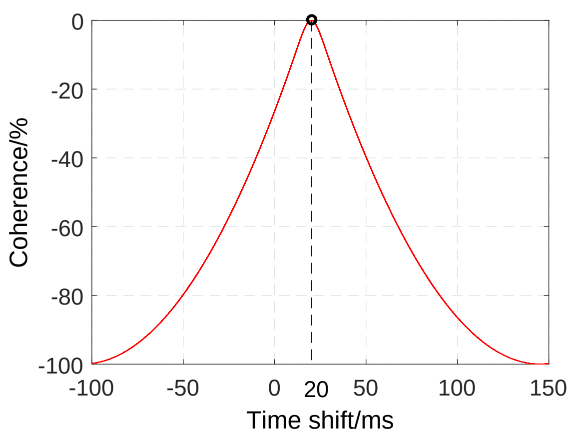


Fig. 1. *The curve shown in the figure consists of 1000 separate curves each of which represents a different simulation of equation* (9) *with D_i generated according to Gaussian distribution with mean* $20\ ms$ *and the standard deviation of* $4\ ms$. *The attenuation factors b_i also follow Gaussian distribution, ω corresponds to* $24\ Hz$ *and T is set to* $125\ ms$.

3. RESULTS

In this section we apply the CMCTL and the proposed delay estimation algorithm to data collected in a neurophysiological experiment.

3.1. Experiment

EEG and EMG signals were collected from 5 healthy subjects performing a simple motor task. The task was to hold a plastic ruler in a key grip parallel to the table surface. The stylus of an electromechanical tapper was placed horizontally to the ruler. The tapper provided pulses of lateral displacement at defined times, giving the subject the sensation that their grip on the ruler may be lost. The subject was asked to hold the ruler gently against the stylus of the tapper to maintain the position of the ruler. A single trial lasted $5\ s$, with the stimulus delivered $1.1\ s$ after the start of the data collection period. The stimuli were delivered at pseudorandom intervals varying between $5.6\ s$ and $8.4\ s$ so that subjects could not anticipate the arrival of the next stimulus. Stimuli were delivered and corresponding data epochs collected in blocks of 25 with a short rest between blocks. Up to 8 blocks of data (200 epochs) were collected for each stimulation condition. EMG and EEG signals were amplified and bandpass filtered ($0.5 - 100\ Hz$ for EEG; $5 - 500\ Hz$ for EMG) and sampled at $1024\ Hz$.

3.2. Estimation Period

The STFT [20] was performed using the 128-sample ($125\ ms$) Hanning window, the bandwidth of which is around $11\ Hz$. The window is shifted in 10-sample increments (about $9.8\ ms$). Significant coherence in the β band ($14 - 34\ Hz$) disappears in the immediate post-stimulus period ($1\ s$ to $1.5\ s$) and reappears soon after $1.5\ s$. Although prominent peaks of coherence were observed in the early post-stimulus period, we initially concentrated our analysis on the late post-stimulus period, since the motor control would be more stable and there would be less bidirectional coupling as time goes on after the stimulation. We therefore identified for each subject the most prominent coherence peak in the late post-stimulus period around which to estimate the time delay (see Table 1).

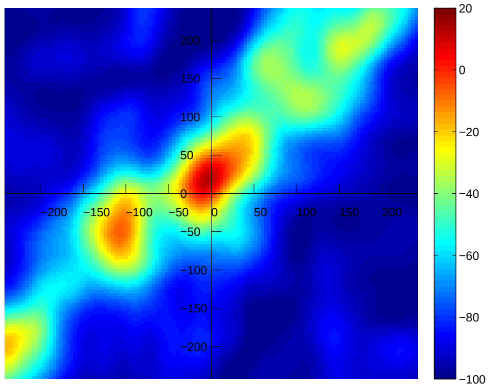
Table 1. *Prominent peaks of each subject*

Subject	Prominent peak	
	Observation time (s)	Frequency (Hz)
B	3.441	24
J	2.963	24
K	2.689	24
L	2.680	16
N	3.256	32

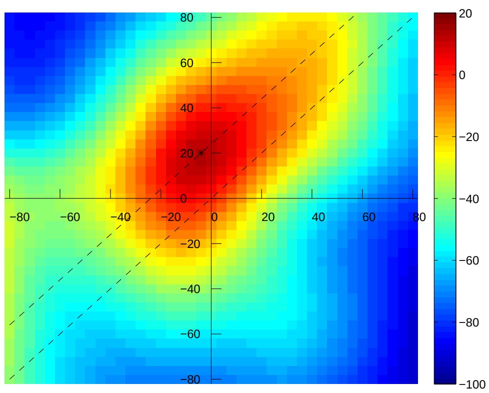
3.3. Delay Estimation via CMCTL

The CMCTL and the proposed method for delay estimation are illustrated in Fig. 2. Fig. 2(a) shows the CMCTL $C_{xy}(t_c, \tau_1, \tau_2, \omega)$ of the data collected from subject B at $t_c = 3.441\ s$ and ω corresponding to $24\ Hz$. Note that since the bandwidth of the STFT wndow is around $11\ Hz$, this CMCTL reflects events in the $18 - 30\ Hz$ range. The selected t_c and ω correspond to the prominent coherence peak observed for $\tau_1 = \tau_2 = 0$, *i.e.* no time lags. The variation of $C_{xy}(t_c, \tau_1, \tau_2, \omega)$ with time lags is evident in Fig. 2(a). The increment of time lags used for the plot in this figure is 4 sampling points ($3.9\ ms$). The maximum of the CMCTL shown in this figure is achieved for $\tau_1 = -3.9\ ms$ and $\tau_2 = 19.5\ ms$ (a zoomed version of the relevant section is shown in Fig. 2(b)), so we conclude that the delay between the coupled EEG and EMG events corresponding to this peak is $\tau = 23.4\ ms$.

Table 2 shows the results of delay estimation obtained using this method, along with the results obtained using the phase method [8] and the method of maximising the coherence as proposed in [19]. The coherence peaks we selected are all situated in the $2.5 - 3.5\ s$ interval, and we apply the



(a)



(b)

Fig. 2. *Example of time delay estimation using the CMCTL. The x and y axes represent shifts of EEG and EMG signals, respectively. The colours represent the increase ratio of coherence compared to that of the original position which corresponds to the origin. The bottom plot is the zoomed version of the top plot, and asterisk marks the position of the maximum coherence.*

Table 2. *Delay estimates obtained using the phase model [8], maximum coherence method [19] and the proposed method.*

Subject	Time delay (ms) from different methods		
	Phase model	Maximum coherence	Proposed method
B	30.7	57.6	23.4
J	2.3	30.3	23.4
K	8.0	-10.7	15.6
L	52.6	89.8	19.5
N	4.4	-127.0	15.6

two other algorithms to those intervals too. According to [8], we applied a weighted least squares regression in the frequency range of significant coherence to generate a straight line. In [19] the authors also propose maximising the coherence by shifting either EEG or EMG signal. There are, however, important differences between their method and our work. The algorithm in [19] involves bandpass filtering of EMG between $30 - 200\ Hz$, followed by its rectification, while we avoid these preprocessing steps since motor control evolves over events in the $14 - 34\ Hz$ range, while rectification causes nonlinear distortions. Further, in [19] relatively long observation segments are used, $1\ s$, while we consider much shorter windows, $125\ ms$, and finally to account for non-stationarity we propose time lags in both EEG and EMG independently, while in [19] only one of the two signals is shifted. We can observe from the results shown in the table that the new method yields much more consistent results across patients, and furthermore all estimated delays are in closer agreement with underlying physiology than the results obtained using the other two algorithms [26].

4. CONCLUSION

This study introduced the notion of cortico-muscular coherence with time lag (CMCTL) and proposed a method for estimating the delay between coupled EEG and EMG events based on local maxima of the CMCTL. The method gave results which are more consistent across healthy subjects and in closer agreement with physiological facts than state-of-the-art methods. The issue of the interpretation of the delay between EEG and EMG events in multi-path propagation scenarios was addressed and it was demonstrated using simulations that the prosed method gives an estimate of the mean delay. Considerations pertaining to bi-directional coupling are a subject of our ongoing research.

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