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Comparison of EEG-EMG Time Delays Calculated by Phase Estimates and Inverse FFT

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Abstract: Currently, many studies have focused on the magnitude of coherence with less emphasis on the time delay, or have mostly used only one method to establish the temporal relationship between the sensorimotor cortex and the peripheral muscles. Here, the time delays using inverse Fast Fourier transformation (IFFT), least squares regression analysis (LSR), weighted least squares regression analysis (WLSR), maximum coherence (MAX-COH) and mean of significant coherences (MEAN-COH) methods in the same subjects are compared to clarify the best method(s) for electroencephalography (EEG)-electromyography (EMG) temporal analysis. EEG activity and surface EMG activity from the first dorsal interosseous (FDI) muscle of the right hand were recorded in eight normal subjects during a weak contraction task. The current source density (CSD) reference method was estimated and used in the phase and temporal analysis. For the EEG and EMG time delay in the same subjects, MAX-COH, MEAN-COH and LSR methods are found to produce time delays that were nearer to those using transcranial stimulation compared to IFFT and WLSR methods. Therefore, the former three are more suitable compare to the latter two methods in the study of time delay between the EEG and EMG signals.

Keywords: Current source density, EEG-EMG coherence, Phase, Time delay.

1. INTRODUCTION

Voluntary movements involve the cooperation of many muscles. Communications are believed to exist between the peripheral muscles and the sensorimotor cortex, which controls the movement functions in the brain. Magnetoencephalography (MEG) and electromyography (EMG) show significant coherence in the frequency band 15-35 Hz of measurements of the first dorsal interosseous (FDI) hand muscle [1]. Such coherence is also found in the work of others [2-5].

Following this, coherence between electroencephalography (EEG) and EMG also has been reported [6-7]. Generally, coherence size (magnitude) and time delay (phase shift) are investigated to reveal communication between the human motor cortex and muscle. However, many studies have focused on the magnitude of coherence with less emphasis on the time delay, and have mostly used only one method to establish the temporal relationship between the sensorimotor cortex and the peripheral muscles [1,5,8].

In the present study, the time delays using inverse Fourier transformation (IFFT) [1,5], least squares regression analysis (LSR) [9], weighted least squares regression analysis (WLSR) [8], phase estimate of maximum coherence (MAX-COH) and mean of significant coherences (MEAN-COH) methods in the same subjects are compared to clarify the best method(s) for EEG-EMG temporal analysis and investigate the physiological significance.

2. METHODS

2.1 Subjects and data acquisition

Eight normal subjects aged between 21-24 years old participated in the study. Surface EMG was recorded from the FDI muscle of the right hand. Subjects were asked to hold a device with a pressure gauge sensor at its center between the thumb and the index finger. The task was a weak contraction between 10-20% of maximum voluntary contraction. Visual feedback of contraction force level was given to subjects prior to EEG and EMG recordings to maintain the contraction.

The EEG signals were recorded from 19 electrodes, which were placed on the scalp base with the international 10-20 electrodes placement system. The EEG and EMG were recorded during the contraction task for 1 minute, repeated four times with intervening rest periods to avoid fatigue. EEG and EMG signals were recorded with passbands of 0.5-200 Hz and 5-5000 Hz, respectively, and stored in a personal computer with a sampling frequency of 1 kHz.

Signals were segmented into non-overlapping 1024 points, resulting in 232 epochs. Epochs with artifacts were rejected, yielding a mean of 224 ± 8 epochs for the analysis. Here the proposed current source density (CSD) reference method for the EEG-EMG coherence and temporal analysis is used [7]. The CSD reference was estimated using the spherical spline interpolation method as described by Perrin et al. [10]. Besides, Nunez et al.

[11] has shown that CSD could enhance spatial resolution for time series analysis of single trial EEG data.

2.2 Data Analysis

The linear correlation between two signals was investigated with a coherence function. The coherence function was calculated using the fast Fourier transformation of 1024 points with frequency resolution of 0.98 Hz. The coherence function is expressed as:

$$r_{xy}^2(f) = \frac{|G_{xy}(f)|^2}{G_{xx}(f)G_{yy}(f)} \quad \le \quad 1 \qquad (1)$$

where $G_{xy}(f)$ is the estimated cross-spectral density function, and $G_{xx}(f)$ and $G_{yy}(f)$ are the estimated auto-spectral density functions for signal $x(t)$ and $y(t)$, respectively. Signal $x(t)$ represents the EEG whereas $y(t)$ represents the EMG signal. The method previously proposed for the calculation of the estimated auto-spectral density and cross-spectral density functions is used [12]. The argument of the cross-spectral density function is equal to the phase angle $\theta_{xy}(f)$, which is important to determine the time delay (t) between the two signals.

The coherence value can be any real number between 0 and 1, with 1 indicating perfect linear correlation between the two signals and 0 showing perfect independence. Coherence was considered to be significant above the 95% confidence limit [13]. The confidence limit (CL) was estimated by

$$CL_{(\alpha\%)} = 1 - \left(1 - \frac{\alpha}{100}\right)^{\frac{1}{(n-1)}} \qquad (2)$$

where $\alpha = 95$ and n is the number of epochs used in the estimation of auto- and cross-spectra. The confidence limit of coherence value obtained was 0.013 ± 0.001 (mean ± standard deviation).

The time delay between EEG and rectified EMG signals was investigated by applying IFFT to the averaged normalized cross-spectra ($G_{xy}(f)$) to get the cross correlation of the two signals, defined as the time of the largest peak in the cross-correlogram.

Time delay can also be determined by phase angle measurement since the phase is a linear function: of frequency with a slope equal to $2\pi t$.

$$t = \frac{\theta_{xy}(f)}{2\pi f} \qquad (3)$$

Time delay at peak coherence (MAX-COH) was calculated using above equation and in the case of multiple significant coherences (P<0.05), their time delays were calculated with the same equation and the mean is measured (MEAN-COH).

Time delay was also determined by the LSR method. Here, the phase consists of two factors, constant time lag and constant phase shift [7]. Since the result is sensitive to the width of the frequency region chosen for the linear fit and the degree of linearity in that region [14], procedures to overcome the problem are as describe. First, instead of using an arbitrary low-cut-off criteria's, the first frequency that showed significant coherence in the beta band (13 – 50 Hz) is selected as the starting point of the frequency band. The beta and higher frequency waves are selected since it is a well-known fact that the cortico-muscular coherence is significantly represented in the beta and higher frequency waves during isometric contractions [1], [15]-[16]. Secondly, the analysis of the correlation coefficient is used to determine the linear relationship between the frequency and phase, e.g. if the first frequency that showed significant coherence is 15 Hz, then the correlation coefficient is calculated for frequency bands 15-18 Hz, 15-19 Hz, …, 15-50 Hz. Thirdly, only frequency bands that were statistically correlated with the phase are chosen for the time delay analysis on condition that the frequency bands must include all the frequencies at which the signals are significantly correlated. Finally, the mean value of these time delays is defined as the time delay calculated from the LSR method.

Following the above steps, the analysis is extended by applying weightings to the LSR analysis. The weighting is given as the inverse of variance (σ_j^2) of the phase shift estimated as

$$\sigma_j^2 = \frac{1}{2n}\left[\frac{1}{r_{xy}^2(f)} - 1\right] \qquad (4)$$

where n is the number of epochs and $r_{xy}^2(f)$ is the coherence value.

3. RESULT

Figure 1 shows an example of 1 s segment of EEG and EMG signals (upper) and the cortico-muscular coherence spectra (lower) for weak contraction of right hand FDI muscle in a subject. Maximally coherences over the contralateral hemisphere were found with C_3 location showed the highest one. Significant coherences were also present at F_3 and temporal area of the contralateral side. Similarly, maximally coherences at C_3 location for the other subjects were found as well. For these subjects, smaller significant coherences than at C_3 location were presents at F_3 (1 subject), F_3, P_3 and F_Z (1 subject), C_Z (1 subject), and P_3 (1 subject) locations. These results show that significant coherences were localized mainly at contralateral sensorimotor area.

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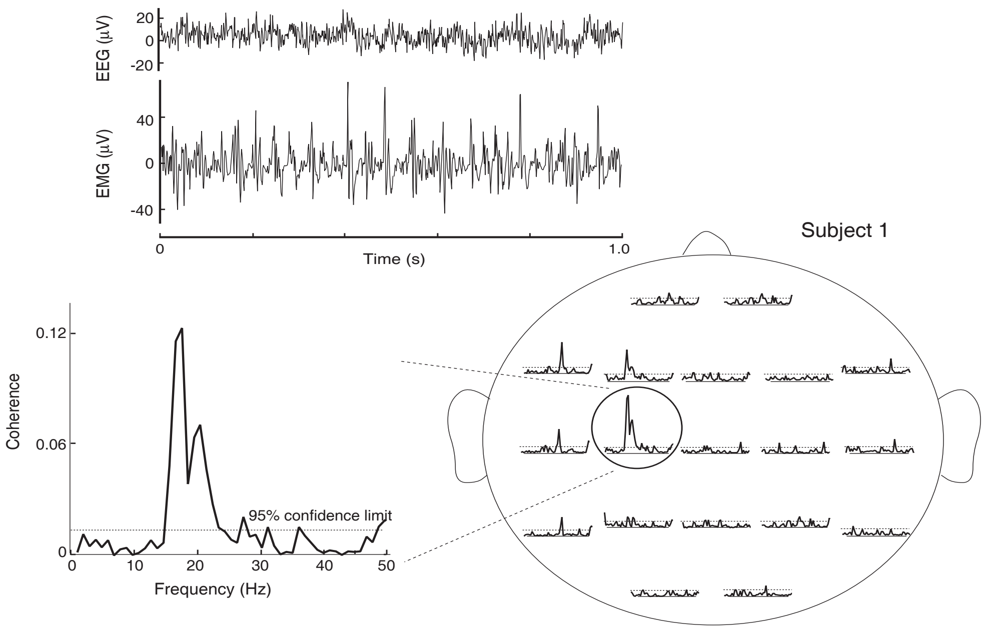

Figure 1. An example of 1 s segment of EEG (C3 electrode) and EMG signals (upper) and the EEG-EMG coherence spectra (lower) for isometric contraction of the first dorsal interosseous muscle in Subject 1. Maximally coherences are located over the contralateral hemisphere with location at C3 showed the highest one.

Table 1 summarizes the main peak coherence, frequency, significant frequency range and the scalp electrodes location of main peak coherence for all the subjects. The means ± SD for the peak coherence and frequency were 0.07 ± 0.04 and 21 ± 4 Hz, respectively. Cortico-muscular peak coherences were found to present at beta waves that range from 14 to 27 Hz for all the subjects.

Figure 2(a) shows the estimated coherence between the C_3 scalp electrode and the FDI muscle by CSD derivation method in Subject 5 and Subject 8. Subject 5 showed higher peak coherence value (0.13, 18 Hz) compared to Subject 8 (0.06, 22 Hz). However, looking at the figures of cross-correlograms of IFFT (Figure 2(b)), maximum peak cross-correlograms was easily distinguishable in Subject 8 than Subject 5. The time delays were 10 ms and 8 ms, in Subject 5 and Subject 8, respectively, as shown by the maximum peak at the cross-correlograms. Both subjects showed maximum peak, at which the EEG (C_3) signal lead the EMG signal.

Figure 2(c) shows the phase estimates between the EEG (C_3) and EMG signals. For both subjects, almost constant phase shifts were found over the range of frequencies, at which the EEG and EMG signals were significantly correlated. In addition, both subjects also showed phase shifts that varied remarkably in dispersive-like behavior, at which both signals were significantly uncorrelated. Similar phase shifts patterns were found for the other subjects. The black dots shown in the figures are the maximum coherence points used in MAX-COH method. From MAX-COH method, the time delays were 15 ms and 14 ms for Subject 5 and Subject 8, respectively. The time delays for frequencies, at which both signals were correlated were also calculated and the mean time delay (MEAN-COH) was estimated. From the MEAN-COH method, the time delays were 13 ms and 16 ms, for Subject 5 and Subject 8, respectively.

Table1 : Features of coherence for cortex-FDI muscle during isometric contraction

Subject No.	Peak Coherence	Peak Frequency (Hz)	Significant frequency range (Hz)	Location
1	0.12	17	15 - 22	C_3
2	0.06	18	15 - 25	C_3
3	0.10	14	13 - 23	C_3
4	0.03	25	19 - 26	C_3
5	0.13	18	14 - 26	C_3
6	0.04	27	19 - 31	C_3
7	0.04	24	24 - 29	C_3
8	0.06	22	13 - 27	C_3
Range	0.03 - 0.13	14 - 27	13 - 29	C_3
Mean ± SD	0.07 ± 0.04	21 ± 4		

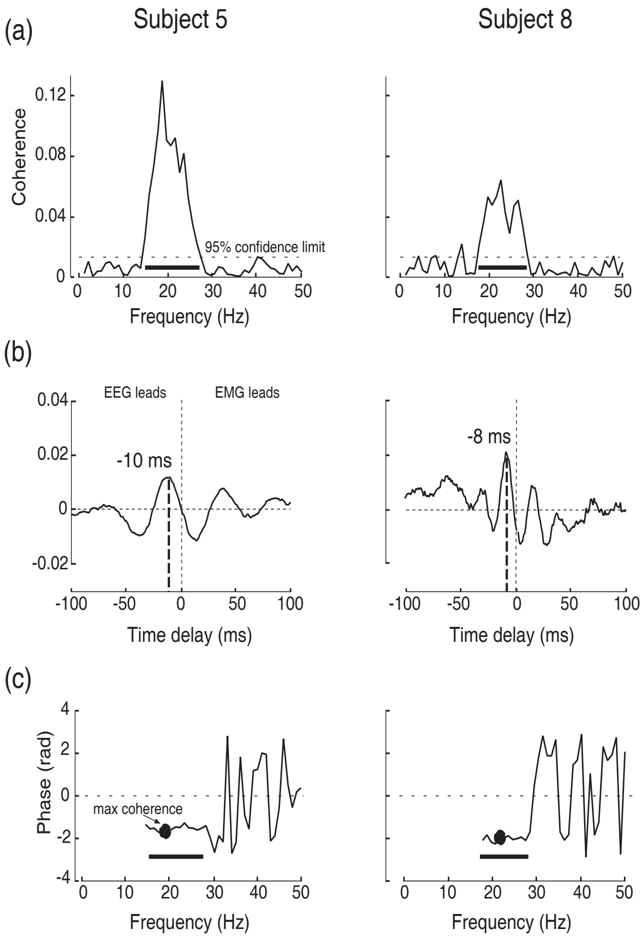

Figure 2. (a) Coherence estimates between the EEG (C_3 electrode) and EMG signals of the first dorsal interosseous muscle in Subject 5 and Subject 8. Significant coherence was found in the range of 13–30 Hz. The dashed horizontal line represents the 95% confidence limit. (b) Cross-correlograms between the EEG (C_3) signal and the rectified EMG signal. (c) Phase estimate between the two signals.

Separate experiment was done to support these findings, i.e. the almost constant phase shifts trend. EEG and EMG signals from a subject were recorded during resting and 10-20% contraction from maximum strength of FDI muscle for 1 minute, repeated 4 times with intervening rest periods. Figure 3(a) shows the 1 s segment of EEG and EMG signals for both resting and contraction state. No significant coherences and constant phase shifts were found during resting of FDI muscle as opposed to contraction state of FDI muscle (Figure 2(b) and Figure 2(c)). In resting state, phase shifts varied remarkably for all the frequencies. On the other hand, in 10-20% contraction state, phase shifts varied remarkably only at which both signals were significantly uncorrelated while producing constant phase shifts, at which both signals were significantly correlated.

As mentioned previously, the first frequency that showed significant coherence in the beta band (13 – 50 Hz) was selected as the starting point of the frequency band for the LSR and WLSR analyses. The first and last frequencies that showed significant coherence for all subjects were shown in table 1. Since beta band was wide in number, many linear regression lines between phase and selected frequency bands can be obtained as illustrated in figure 4. Figure 4(a) shows the linear relationship between phase and frequency band 14 – 36 Hz (23 points) using LSR analysis while Figure 4(b) shows another linear relationship but at frequency band 14 – 50 Hz (37 points) for Subject 5. Both exhibit differences in slope, for frequency band 14-36 Hz, slope value was 0.04 while for frequencies that ranges from 14 to 50 Hz, 0.07. This could relate to the difference in time delays since time delay is calculated by dividing the slope with 2π.

Because of the difference in data points, and the fact that calculated linear regression line must indeed represents significant linear relationship between the phase and frequency band of interest, standardized normal variable (z) was calculated to distinguish frequency bands that are significantly correlated with phase from those that are non-correlated. Figure 5(a) shows the z value of the normal distribution function for each frequency regions with 95% confidence limits for Subject 5 and Subject 8. Values inside the region bounded by ± 1.96 (gray zone) indicated that the phase and frequency bands of interest were not statistically correlated (white circles). Examples are frequency bands 14-26 Hz and 14-36 Hz for Subject 5 that were indicated by arrows in the figure. Another example is frequency band 13-27 Hz for Subject 8. Values inside the region

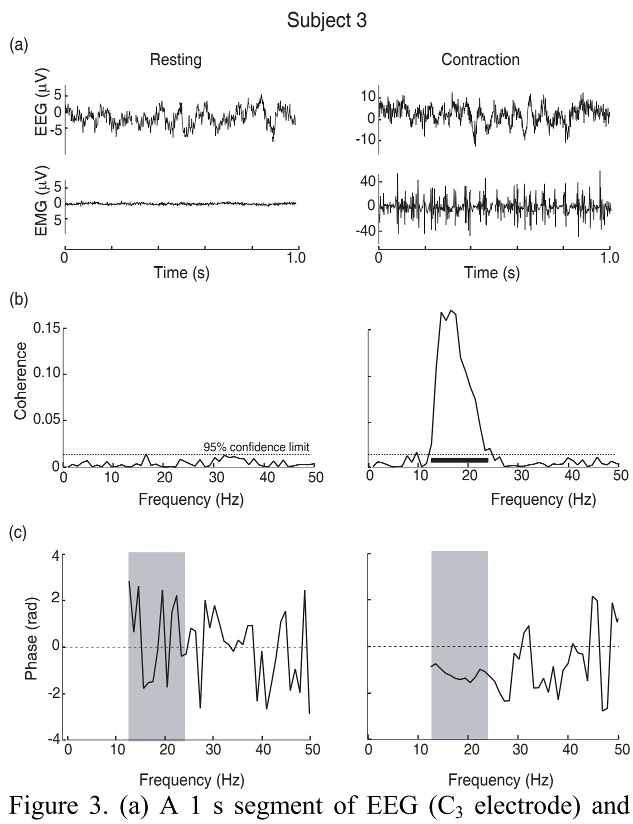

Figure 3. (a) A 1 s segment of EEG (C_3 electrode) and EMG signals during resting and 10-20% from maximum contraction of the first dorsal interosseous muscle in Subject 3. (b) Coherence estimate between the two signals for non-contraction and 10-20% contraction trials. (c) Phase estimate between the two signals.

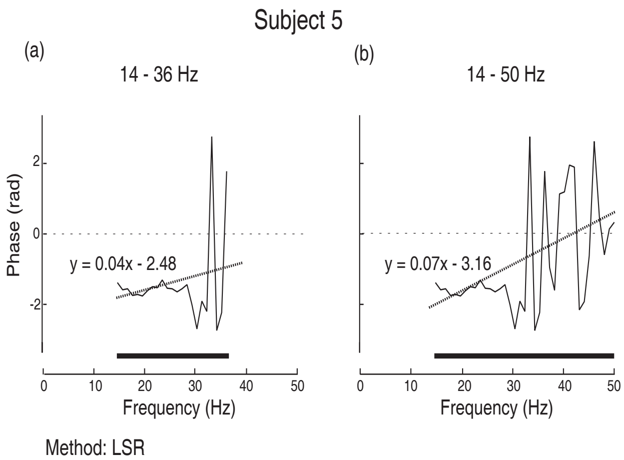

Figure 4. (a) Linear relationship between phase and frequency band 14 – 36 Hz (23 points) using least squares regression analysis for Subject 5. The slope value was 0.04 with intercept value, -2.48 radian. (b) Linear relationship between phase and frequency band 14 – 50 Hz (37 points) using least squares regression analysis for the same subject as (a). The slope value was 0.07 with intercept value, -3.16 radian.

bounded by ±1.96 (gray zone) indicated that the phase and frequency bands of interest were not statistically correlated (white circles). Examples are frequency bands 14-26 Hz and 14-36 Hz for Subject 5 that were indicated by arrows in the figure. Another example is frequency band 13-27 Hz for Subject 8. Similarly, many frequency bands that significantly correlated with phase were identified (black circles). Subject 5 showed 11 frequency bands that were statistically correlated with phase. For example, frequency band 14-50 Hz. Meanwhile, Subject 8 showed 19 frequency bands that were statistically correlated with phase such as frequency bands 13-36 Hz and 13-50 Hz as indicated by the arrows in the figure. These frequency bands that correlated with phase were chosen for the time delay analysis using the LSR (Figure 5(b)) and WLSR (Figure 5(c)) analyses. Positive time latency indicates that the EEG signal leads the EMG signal and negative time latency indicates the reverse. The means time delay calculated from these chosen frequency bands were selected as the time delays for the LSR and WLSR methods. From Figure 5(b) (LSR), the mean time latencies were 11 ms, and 18 ms for Subject 5, and Subject 8, respectively, as shown by the horizontal lines. Similarly, from the WLSR analysis (Figure 5(c)), the mean latencies were 11 ms, and 17 ms for Subject 5, and Subject 8, respectively.

Figure 6 summarizes the time delays in all subjects (Figure 6(a)) and the mean time delays (Figure 6(b)) for the different analysis methods. For individual time delays, the MAX-COH and MEAN-COH methods produced almost similar time delays. Compared to these methods, the LSR method produced higher time delays in 3 subjects, while for the other 5, unchanged or lower time delays. Lower time delays were apparent for WLSR and IFFT methods compared to the other 3 methods. IFFT of the cross-spectra for all subjects found that the mean time latency was 10 ± 4 ms with the EEG signal leading the EMG signal in all subjects. From the MAX-COH method and MEAN-COH methods, the mean time delays were 14 ± 4 ms and 14 ± 3 ms, respectively. The average (n = 8 subjects) mean time delays calculated from LSR and WLSR analyses were 15 ± 5 ms and 10 ± 4 ms, respectively.

The time delays in the same subjects were statistically higher using the MAX-COH, MEAN-COH and LSR methods compared to those using IFFT and WLSR methods ($P<0.05$). The order of cortex-FDI muscle mean time lag was IFFT < WLSR < MEAN-COH < MAX-COH < LSR.

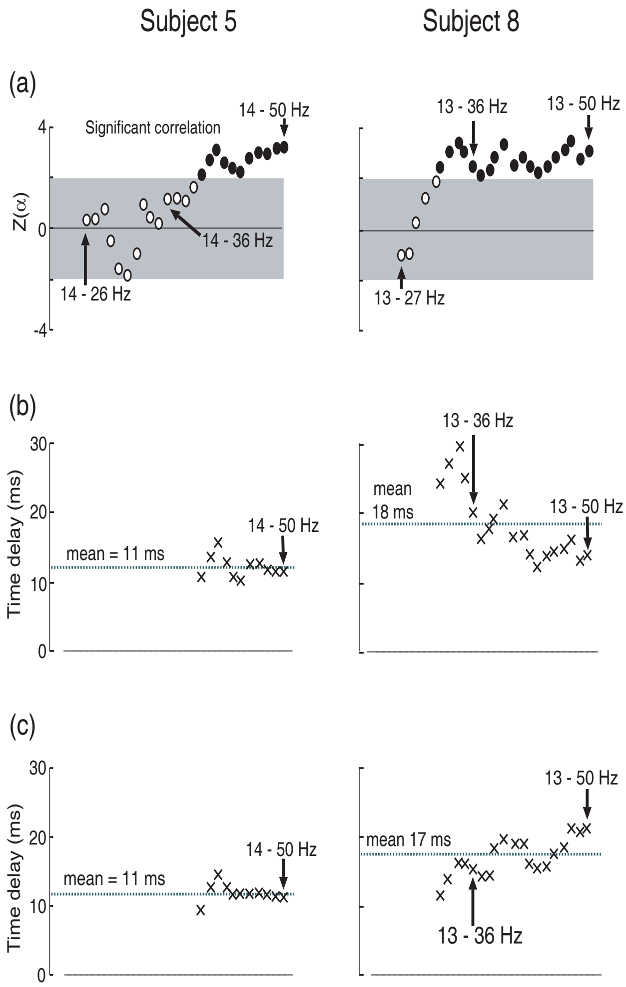

Figure 5. (a) Z value of the normal distribution function for frequency starting at point 14 Hz, and 13 Hz in Subject 5, and Subject 8, respectively. Horizontal dashed lines indicate the 95% confidence level ($z = \pm 1.96$) (gray area). Shown in black circles are the frequency ranges that were significantly correlated with phase. (b) Time delay calculated from least squares regression analysis in the 2 subjects. Only frequency bands that were significantly correlated with phase were examined. The horizontal dashed line represents the mean time delay. (c) Time delay calculated from weighted least squares regression analysis with the same procedures as (b).

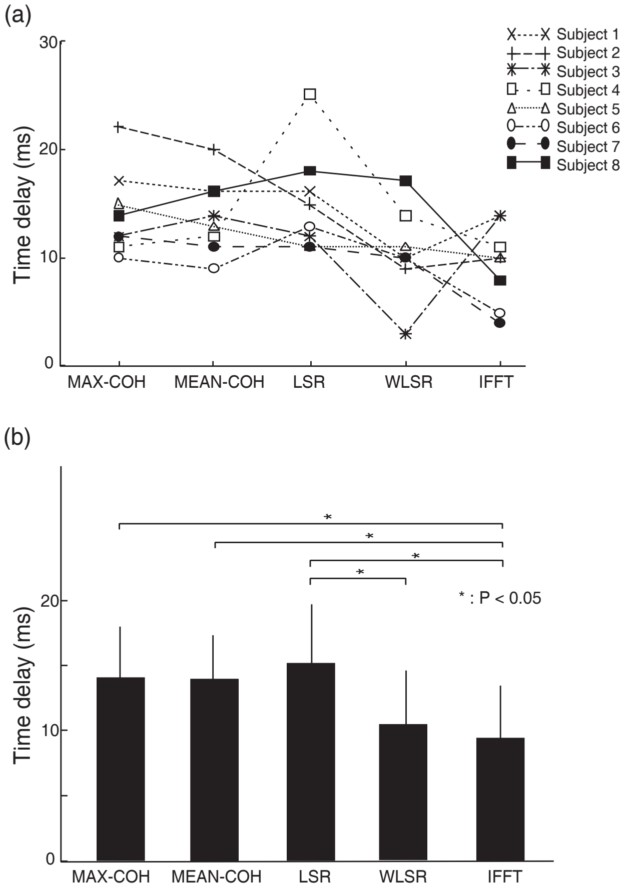

Figure 6. (a) Comparison of time delay between different analysis methods in eight subjects. (b) Mean ± standard deviation of the time delays.

4. DISCUSSION

The peak coherences value have been demonstrated and the significant coherences between the human brain and FDI muscle in the beta waves agree very well with previous reports [1]-[5]. It is found that the time delays in the same subjects were statistically higher using the MAX-COH, MEAN-COH, and LSR methods compared to those using IFFT and WLSR methods. Even-though the time delays in the same subject were higher in the former compared to the latter, they were still shorter than those predicted from transcranial electric stimulation [17] and transcranial magnetic stimulation [18]. The possible reasons behind this have been discussed by Lopez da Silva et al. [19] and Grosse et al. [20]. They predicted that the skull and scalp acts as low pass filters that may introduce phase shifts, and thus, may underestimate real conduction delays. They also predicted that there may be a possibility that more than one coherence activity may overlap in the same frequency band, in which case the phase estimate will be a mixture of the different phases. From our observation, the inconsistency in the production of the time delays between the transcranial stimulation and the EEG-EMG temporal analysis may be due to the differences in the execution of the experimental task such as extend and maintain fingers in a horizontal position in the transcranial stimulation as contradicted to the isometric contraction of FDI muscle in our study. Signals from the brain most probably conduct faster to the peripheral muscle during maintain isometric contraction compared to when the muscle is in 'idle' state, which could explain the time delay inconsistency between the transcranial stimulation and the EEG-EMG temporal analyses. However, further study need to be done to clarify this hypothesis.

One interesting fact that came from our results was that the MAX-COH method produced time delay that was consistent with the LSR analysis method when it is common believe that to calculate time delay from only one point is ambiguous and that measuring phase over a band of frequencies reduces this ambiguity, hence the LSR analysis was introduced [20]. However, the result showed that a reliable time delay can be obtained from EEG-EMG signals by considering only the frequency and phase at maximum coherence. If we examined more closely, nearly constant phase over the range of frequencies can be found at which the signals are significantly correlated (Figure 1(c)). This was observed previously but with little explanation on its meaning [6]. Taking the mean time delay of these significant coherences, it is found that it was similar to the MAX-COH method. Therefore, considering only maximum coherence point might gives justification for the other significant coherence points and could be use in the study of EEG-EMG temporal relationship.

Furthermore, a separate experiment to support the significant of MAX-COH and MEAN-COH methods have been done. Both significant coherences and constant phase shifts were not found during resting of FDI muscle as opposed to significant coherences and almost constant phase shifts found at beta waves during isometric contraction of FDI muscle. The result shows that phase-locked synchronization between the cortex and FDI muscle was present only during isometric contraction. This might indicates that considering other points besides where the two signals were significantly correlated might have no justification and could be misleading. This makes the LSR and WLSR methods un-fit for studying the time delay between the two signals. On the other hand, different phase shifts pattern other than constant phase over the range of frequencies at which the signals are significantly correlated was found using MEG [4]. They found linear relationship between the phase and the frequencies, at which the two signals were correlated and this contradicted to our results. Further investigation need to be done to delineate this discrepancy especially on the effects of reference method use for EEG recording to those phases. It is interesting to know if similar phase pattern over significantly correlated EEG-EMG signals could be found using other than CSD method such as ear-lobe reference and averaging methods.

This study also shows that the time delay from WLSR analysis method was shorter compared to previous finding using the same analysis method [7]. The fact that the coherences value determines the weighting (refer to the weighting equation) might be the reason behind this since the significant coherences produced almost constant phase which effect the regression line steepness, and consequently contribute to low time delay. Beside, Mima et al. [7] used one frequency band only, 14-50Hz as opposed to ours, i.e. we used all the frequency bands that were correlated with phase and calculated their time delay and produced the mean. Another reason that might contributes to the difference in the result is that we used constrained form of the phase estimate (within 2π range) as opposed to unconstrained one used by Mima et al. [7]. However, we found that the unconstrained phase has no affect on time delay produced by LSR method since among the four methods, it was the nearest to the time delay from transcranial stimulation. We used the constrained form so as not to lose the information that comes with the phase-spectra; i.e. phase between 0 and –pi means the EEG signal was leading the EEG signal and phase between 0 and π means the opposite.

Between the two methods, EEG-EMG and MEG-EMG may produce similar coherence results but their phase delay might be different as the result of different techniques in capturing the brain signals. Hence, it is interesting to know if similar comparison result of the time delay in same subjects could be found using the five techniques mentioned here in MEG study.

As conclusion, for the EEG and EMG time delays in the same subjects, MAX-COH, MEAN-COH, and LSR methods produced time delays that were nearer to those using transcranial stimulations compared to IFFT and WLSR methods. In that sense, the former three are more suitable compare to the latter two methods in the study of time delay between the EEG and EMG signals.

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