An Investigation of Task Related Changes in EMG Activity for use in Prosthetic Control

E.M. Mulcahy, V. Stamatopoulos, K. Bray, D.M Halliday, B.A. Conway
Bioengineering Unit, University of Strathclyde, Glasgow, G4 0NW, UK

Abstract
Changes in the frequency content of EMG signals during normal voluntary movement correlate with different features of motor behaviour and may offer a source of signals that may be used in the activation and control of neural prostheses. In normal subjects it has been shown that oscillations occurring in the 8-12Hz range dominate EMG activity during voluntary movements, while 15-35Hz rhythms in cortical and muscle activity are enhanced during maintained postural tasks. In this study EMG activity from flexor and extensor muscles on the amputated and intact forearm of upper-limb amputees were recorded along with the EEG activity from the left and right motor cortices while subjects performed a series of movements of the real and phantom hands. An 8-12Hz feature frequently dominated the stump EMG signal regardless of the task condition. There was little evidence of functional coupling between the EMG and contralateral EEG signals on either the amputated side or intact side.

Keywords: EEG, EMG, intention detection, motor control, neuroprosthetics

1. Introduction

Rhythmicity and synchronisation are believed to have functional significance for communication between neural systems [1]. In the motor system, oscillations in the 15-30Hz (beta) frequency range are commonly observed in the activity of the motor cortex [2,3,4], and of motor unit firing in muscles [5,6] active during maintained contractions. In synergistic muscles, inter-muscle coupling requiring co-activation is also reported [7,8]. Significantly, recent studies have revealed a functional coupling in the beta band between the EEG or MEG recorded from the motor cortex and contralateral EMG activities during postural contractions [9,10].

During movements, the 15-35Hz rhythms in the EEG and EMG tend to decrease. In the EMG, as well as a reduction in beta band power, there is also increased oscillatory activity at 8-12Hz [11]. During normal movement it is common to see an interplay between opposing muscles, with alternating bursts of EMG activity in the muscles, giving rise to small amplitude 8-12Hz discontinuities observed during slow movements. This feature of movement has been seen to accompany finger, wrist, elbow and leg movements. Vallbo and Wessberg [11] suggest the 8-12Hz rhythms are the result of a “pulsatile motor command” descending from supraspinal centres, however the origin of the 8-12Hz activity that generates these pulses has yet to be identified.

Figure 1. Autospectra of rectified EMG from active forearm muscle. A1 and A2 represent typical spectra for normal subjects. B1 and B2 show spectra for the same task conditions for an amputee subject. An 8-12Hz component is present in both. The vertical bar in the upper right hand corner of each plot represents the 95% confidence limit.

Different elements of the motor system therefore tend to display task-specific characteristics. Activity of the motor cortex in the beta-band appears to modulate with task performance, with 15-35Hz rhythms increasing during posture and decreasing during movement tasks. While, in the EMG, the transition from a movement to a postural task involves a switch from a dominant 8-12Hz rhythmic component to a 15-35Hz dominating feature (figure 1A) [12].
These task dependencies present in the EEG and EMG may have important functional implications for the future of prosthetic control. The task specific features in the neural signals could provide a means of predicting motor intention in people who suffer from paralysis or who have lost a limb. In paralysis it might prove possible to use some of the remaining active muscles or even the signals directly from the brain to predict motor intention, while in the case where a person has lost a limb much of the musculature that once controlled the limb’s movement will still remain intact, and therefore still maintain the ability to receive information from brain centres.

Recent studies have demonstrated the possibility of communicating with a computer directly with brain signals [13,14]. In these brain-computer interfaces (BCI) the brain’s changing alpha rhythms are used to control a cursor on a computer monitor. However, these systems are slow to work and the user usually requires a long learning period before they can control the alpha rhythms. Because beta rhythms recorded from the motor cortex occur naturally and display task-specific features, they may be more efficient to use in BCIs, or as intention detection indicators in people who suffer from paralysis. An important feature of these rhythms is that they can be modulated during imagined motor tasks [15] and this implies that very little learning would be required for a patient to reproduce the rhythms.

Presently, EMG is used in the control of FES and prosthetic devices. However, these mechanisms use the changing amplitude of the EMG as the intention indicator, requiring the user to produce and maintain a specified contraction strength in the recording muscle before the device can be activated. We believe that a more sophisticated method of predicting intention may be achieved by monitoring changes in the frequency content of the signal along with the changes in amplitude. By recognising changes in the frequencies of the neural signal more information about the intended task can be obtained, without the need for strong muscle contractions. Also, because these rhythms are produced naturally, the prosthetic control would also, in effect, become more natural.

In order to determine whether neural rhythms can be used for neuroprosthetic control, it is first necessary to establish how robust the neural signals are and to identify whether similar rhythmicities exist in the neural signals obtained from people with neurological disorders or who have lost a limb.

The aim of this study is to identify task specific features of the EEG and EMG signals recorded from upper-limb amputees.

Figure 2. Coherence between EMG and contralateral EEG during repeated movement and hold tasks. The upper row of figures represents the coherence estimates for the amputee recordings. Clearly there is no coupling evident between the EEG and EMG. In contrast, the coherence estimates for recordings from a non-amputee (lower figures) show 20Hz coupling during the hold tasks. The 95% confidence interval is represented by the horizontal line in each estimate.
2. Methods

The EEG from the left and right hemispheres was recorded simultaneously with the EMG from the left and right wrist flexor and extensor muscles in seven healthy trans-radial amputees. Written informed consent was obtained in accordance with the approval of a local ethics committee.

Bipolar EEG was recorded from over the hand area of the right and left motor cortices. The surface EMG was recorded from both the flexor and extensor muscles (where possible) of the normal forearm and the residual limb on the amputated side. The EEG and EMG signals were amplified and filtered at 3-500Hz and recorded for subsequent off-line analysis.

The subjects were instructed to perform simple flexion, hold flexed, extension and hold extended tasks of the wrist (or phantom hand) in response to auditory cues. The subjects were first allowed to work both hands simultaneously and then rested the intact hand and repeated the experiment, moving the phantom hand only. The EEG and EMG were recorded throughout the experiments.

Figure 3. Coherence between co-active muscles. The upper two plots show typical inter-muscle coherence patterns for a normal subject, with strong 20Hz coupling during posture. The lower plots show the same coherences observed in an upper-limb amputee. The horizontal line represents the 95% confidence interval for each estimate.

The data was processed using the techniques outlined in Halliday et al. [16]. It was first necessary to section the data so that corresponding task conditions were analysed together. In the frequency domain estimates of the autospectrum of the EEG and rectified EMGs were constructed along with estimates of coherence between the EEG and ipsilateral and contralateral EMG. Measures of coherence between the flexor and extensor EMGs from the same limb were also obtained.

These processes enable us to determine the frequencies of the oscillations present in the neural signals under different task conditions (autospectra), as well as giving an informative measure of association between individual signals in the frequency domain (coherence).

3. Results

In normal motor performance it is typical to see a clear difference in the EMG spectra from different task conditions (fig 1A), with an 8-12Hz feature present during movement phases and a 15-30Hz component present during hold phases. Fig 1B shows typical spectra for EMG recorded from a stump muscle of an upper-limb amputee. There is little evidence of any shift in the dominant frequency component of the EMG with a change in motor task. Instead, a feature centred about 10Hz is present regardless of the task condition and there is little evidence of a dominant 15-30Hz component similar to that seen in normal posture. This was a common feature among the amputee subjects. Interestingly, it was also noted that, on the amputee’s intact side, it was also difficult to distinguish any significant changes in the spectra from one task condition to another.

The amputee’s EMG signals did display 15-30Hz features in some cases when the task was to maintain a postural contraction for a prolonged period. However, despite the presence of a 15-30Hz component in the EMG, the spectra were always dominated by an 8-12Hz feature (not shown). This feature probably reflects single unit firing from the atrophied muscles.

It is believed that muscles involved in voluntary tasks are under the control of the motor cortex. This functional coupling is evident from the presence of coherence at 15-30Hz between the muscle and contralateral cortical activity during maintained postural contractions, which is then lost when movement starts again (fig 2). The coherence estimates for stump EMG and contralateral EEG activity showed little evidence of any functional coupling between the two signals (fig. 2). Coherence between the stump EMGs and ipsilateral EEG were also estimated and again, there was rarely any coupling observed. The same was true for the coherence between EMG activity recorded from the normal, intact side, and the ipsilateral and contralateral EEG signals.

It was also noted that there was little evidence of coupling between different muscles on the stump, in contrast to the inter-muscle coherence patterns seen in
normal motor control (fig. 3). This is most likely due to the fact that recordings were taken from stump flexor and extensor muscles and that for these subjects there were few instances of co-contraction between these muscles. In the transition from a hold extended task to a flexion movement, for example, the EMG from the extensor switched off and the flexor switched on in a reciprocal manner. Thus the characteristic agonist-antagonist bursting patterns observed in normal movement was not apparent in the amputee’s stump muscles.

During the experimental sessions the subjects were first instructed to perform the movement-hold tasks moving both sides at the same time. The experiment was then repeated with the amputees moving the stump side only. There were no observable differences between the results obtained when there was guidance from the normal hand or not. Also, five of the amputees have attended two separate experimental sessions. The results from individual subjects have not shown any significant changes between recording sessions.

4. Discussion

From these experiments it appears that, despite the ability of a trans-radial amputee to simulate movement of a phantom hand, the processes driving the muscle are not the same as those that drive the muscles involved in normal movement. Although, these findings are not encouraging for the use of frequency tracking EMG signals for intention detection in neuroprosthetics, they do provide important information for our understanding of motor control.

Although the signals obtained from the amputees did not appear to show task specificities typical of normal EEG and EMG, a significant 8-12Hz rhythm was frequently observed, and it was present in the EMG signals recorded from all amputees. It is necessary to establish whether the 8-12Hz rhythms in the stump EMG signal does itself display any modulation.

It is also likely that part of the loss of the normal pattern of EMG modulation reflects atrophy in the stump musculature, altered sensory feedback and plastic changes in cortical representation following limb loss. In this respect, it would be important to examine amputees in the period directly following limb loss.

Future work should involve the assessment of cortical and muscle signals in people who suffer from paralysis. It is important to establish whether the task specific features evident in normal movement are still present, or lost, following paralysis.

References


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