Simple EMG Control for FES-Cycling

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Abstract

We describe preliminary results from a series of experiments investigating the use of minimally processed EMG signals as a potential control source for FES-assisted cycling. Experiments were conducted under both isometric conditions and during cycling on a static, semi recumbent tricycle. While cycling, EMG was recorded from the left quadriceps in 5 normal subjects and one subject with incomplete spinal cord injury. Stimulation pulses were applied to the same muscle. EMG activity was measured in a window preceding the stimulation pulse and correlated with that recorded after the pulse in various time windows. Significant correlations were found under both conditions and in both normal and the injured subject when a window 70-110ms after the stimulation pulse was used.

1 Introduction

Functional Electrical Stimulation (FES) can be used to provide many functions to the neurologically impaired population. One activity that has come under increasing attention in recent years and has started to become truly functional is FES-assisted cycling. We have previously reported on a single subject with incomplete spinal cord injury who has used FES-assisted cycling recreationally and achieved apparent gains in neurological function [1]. Perkins has described the cycling system that this patient uses [2]. Rushton’s hypothesis [3] concerning functional recovery and FES suggests that optimising recovery may be dependent upon providing ascending and descending drive to motor neurones concurrently. We propose to test this hypothesis in the future by using voluntary-EMG control of stimulation. In this control scheme, stimulation amplitude is modulated by voluntary drive. Therefore antidromic activation of motor fibres through the stimulation will occur when there is orthodromic descending voluntary drive.

Using EMG signals to trigger or modulate FES has a long history [5]. We want to be able to stimulate a muscle in direct relation to its level of voluntary drive. This is complicated because of many effects that occur when stimulating during voluntary activity, including collision-blocking and reflex responses. Because it is random, we measure EMG as an RMS level during a defined period or ‘window’. Normally, stimulation would be applied as trains of pulses and for FES-cycling, these would be applied during the range of crank angles that the particular muscle is required to contract. In this study, we simplified the situation and just applied one stimulus pulse per revolution while the muscle was active (Figure 1). The EMG measurement from the window just before the stimulus was used as a measure of the effort. The experiment is to investigate the effect of the time of the post-stimulus window.

The optimal window in this situation; 1) can be recorded as soon as possible after the stimulus pulse, 2) would be highly correlated to the level of voluntary effort on the part of the subject. In our results we are therefore looking for a window that gives a high correlation coefficient and the slope of the line of the pre-stimulus EMG plotted against the post-stimulus EMG should be neither too step or too flat.

In order to decide on the window length, we recorded EMG in 10 normal subjects while they were producing a steady isometric contraction. The EMG records were cut into segments of various lengths and their RMS values computed. Correlation coefficients were then calculated between successive measurements.
With 40ms windows, the correlation coefficient between pairs was >0.7 for all subjects.

In this study, the questions that we want to answer are:

- to get an accurate estimate of the voluntary drive, how soon can the window be after the stimulus pulse?
- how good is this estimate compared to the estimate if there is no stimulation?

When answered, we can apply appropriate pulse trains, perhaps with irregular inter-pulse intervals to allow a recording window.

## 2 Methods

All experiments were performed with the written, informed consent of the subject and with local ethical committee approval (joint ION/NHNN). A stimulator was modified to give a symmetrical biphasic output with a 300µs inter-phase interval. Stimulation pulses were delivered to left quadriceps through round (5cm diameter) surface electrodes (Pals, Nidd Valley). EMG electrodes were orientated as described by [4] so as to minimise stimulation artefacts. Despite this the stimulation artefacts were still large compared to the extracted EMG signal. The difference between the stimulation signal and the EMG signal was of the order 1,000:1. A commercial tricycle was instrumented with a shaft encoder that gave a voltage output proportional to the crank angle. Using a comparator, this voltage was used to trigger a stimulation pulse at a crank angle that was set near to the peak of quadriceps EMG activity. The myoelectric signal and the crank angle were recorded using the Matlab Data Acquisition Toolbox and National Instruments AD board (MIO64E). Data was sampled at 1kHz and EMG data filtered from 0.3 to 300Hz.

Subjects cycled at a fixed cadence with the aid of a visual and audio metronome. A servo-brake attached to the rear wheel of the tricycle allowed the experimenter to set the power output. The stimulus intensity was also set at a multiple of motor threshold (i.e. 1, 2, 3 and in some subjects 4 times). In each test, recording lasted 30s.

Recordings were made from 5 neurologically normal subjects and from one subject with a chronic incomplete spinal cord injury (iSCI) who was an experienced FES-assisted cyclist [1].

## 3 Results

### 3.1 No-stimulus EMG

We recorded EMG in 2 neurologically-normal subjects during cycling with no stimulation (a total of 15 revolutions at each of 4 effort levels). We calculated the RMS EMG in the window prior to the trigger pulse, and at windows following the trigger pulse. The correlation coefficients were highest (0.88) for the windows closest together (starting 0ms after the trigger), but a 40ms window starting 70ms after the ‘stimulus pulse angle’ gave a correlation coefficient of 0.86 (correlation coefficients were averaged for each subject across all conditions).

![Figure 1: A schematic representation of the sampling paradigm used in this study showing the pre-stimulus baseline period and the consecutive sample windows after the stimulation pulse.](image)

### 3.2 With-stimulus EMG

In this section we plot the results from all 5 neurologically normal subjects and from the one subject with incomplete spinal cord injury for comparison. For each of five post-stimulus windows, EMG levels before were plotted against EMG levels after the stimulus, and regression lines fitted to each. Initially, results were analysed independently for each stimulus intensity and effort level. As the results were similar, all the points for all revolutions (10 per test), both effort levels, and all stimulus intensities, were plotted on one graph for each subject (i.e. 60 or 80 points: see fig. 2). This was repeated for each post-stimulus window, the correlation coefficients calculated, and these are plotted in figure 3.

The results are primarily assessed by comparing the correlation coefficient between the pre and post stimulus estimate of voluntary drive and secondarily by plotting the slope of the regression line between pre- and post-stimulus measurements.

Figure 3 shows that the correlation coefficient rises rapidly after the stimulus (windows 1 and 2), and then more slowly (windows 3-5). At
window 3 (starting 70ms after the stimulus), the average correlation coefficient is 0.75, (for subjects, respectively: 0.94, 0.85, 0.73, 0.56, 0.85, 0.78).

Figure 2: An example of the relationship between the pre- and post- stimulus EMG levels for one subject whilst cycling. The correlation coefficient here is 0.78.

Figure 3: The correlation coefficients measured for each of the five neurologically normal subjects (a-e) and the one subject with incomplete spinal cord injury (i) during cycling. Data were combined for all speeds and power levels and stimulation intensities.

4 Discussion and Conclusions

This preliminary study has demonstrated that using a simple measure of the EMG level in a window after the stimulation may provide an adequate measure of the subject's voluntary effort. We believe that this offers a new, simple method of providing EMG control of FES for a cycling system.

Estimating the voluntary effort from the EMG can only be approximate because it is a random signal measured in a finite window. We have quantified this effect by the correlation coefficient without stimulation, which at 70ms delay (window 3) was 0.86. The correlation coefficient with stimulation should be compared to this number. The small decrease seen when stimulation is added (0.86 vs. 0.75) shows that this is also a good estimate of the pre-stimulus EMG. A correlation coefficient of around zero, as recorded immediately after the stimulus pulse, indicates that there is no relationship between the two samples, presumably in this case because the stimulus artefact is so large.

In this study we have used only single stimulation pulses and are currently investigating the effects of applying multiple stimulation pulses.

An important feature of the time window we have chosen is that the amplitude of the measured response is not dependent upon the strength of the stimulation pulse, thereby reducing the possibility of “lock-up”, a situation in which the system becomes unstable due to the feed-forward nature of the controller.

The results from these experiments carried out during cycling are in line with our results obtained during isometric conditions. Furthermore we have performed a single experiment on a subject with incomplete spinal cord injury. Aside from the generally lower EMG levels his data was similar to that seen in the neurologically normal subjects.

During fatigue the increased randomness of the signal lead to decreased correlation coefficients in both the subject with incomplete spinal cord injury and a neurologically normal subject.

In conclusion, we believe that for FES-assisted cycling, this method of control may be advantageous and warrants further development and testing.

References


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