EMG SEGMENTATION FOR WIRELESS ACCESS

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INTRODUCTION

Access Technologies

Access technologies are devices and strategies that facilitate unconventional methods of interaction. They are utilized by individuals with a variety of motor impairments and degenerative disorders including cerebral palsy and Amyotrophic Lateral Sclerosis (ALS) and acquired spinal cord injuries (SCI) [1]. Some modern assistive technologies include mechanical switches, eye-tracking and brain-computer interfaces (BCI). EMG-based access pathways are a potentially viable alternative to these devices due to their cost-effectiveness and resolution. EMG provides an easy and effective pathway toward a variety of environmental controls and communication methods for patients with residual muscle movement. EMG methods may be customized based on the user's motor abilities.

Electromyography

Surface electromyography (EMG) is the spatial and temporal summation of motor unit action potentials during muscle contraction. EMG measures skin conductance caused by the spread of electrical impulses from the neuronal and muscular cells to the body's surface and is dominated by the high-amplitude action potentials during neuronal stimulation of myofibrils. Large signal amplitudes are therefore observed when motor-unit recruitment increases during muscle contraction.

EMG is also affected by excitatory and inhibitory post-synaptic potentials [2]. Other sources of noise include the electrode cable and signal instability. These noise sources make it difficult to classify muscle contractions as the skin exhibits a voltage change induced by all sources in the vicinity.

There are several characteristics of EMG signals that must be considered during experimentation. The amplitude range of EMG is typically 20-2000µV and thus significant amplification must be provided. In addition the EMG frequency is in the range of 20-200Hz [2] and is easily affected by environmental noise such as ambient electromagnetic radiation. Signal clarity is dependent on electrode structure, placement and surface characteristics. Gelled electrodes with an AgCl surface act as a chemical interface between the skin and electrode; this enables an oxidation-reduction reaction at the skin surface and allows current flow to the electrode. The electrodes should be placed 20mm apart along the muscle longitudinally for optimal signal detection [4].

EMG Segmentation

EMG data is complex and carries a large amount of information in its time and frequency spectra. Several methods have been implemented in the past for single motor unit action potential (MUAP) recognition. Gazzoni et al. (2004) gathered information using an array of small sensors aligned in rows and columns along the muscle in a process called multi-channel surface EMG signal detection [5]. Huang et

al. applied the continuous wavelet transform (CWT) method to distinguish the EMG signal from surrounding noise [6]. Huang's work required digitization of the signal by using a threshold, which was determined by finding the maximum voltage value M of a test function for a given period of rest multiplied by a constant γ>1.

Threshold = γM	(1)

If the signal to noise ratio is high, the choice of γ is not critical, since the difference between a MUAP and the noise is large. However, if the ratio is low, a higher γ is selected to differentiate the signal from the noise [6]. The approach taken in this study is similar, but applies a linear smoothing function to the raw EMG instead of the CWT.

Segmentation algorithms that output a binary signal in place of the complex analog EMG may facilitate the implementation of switch-based access pathways for patients with disabilities.

Objective

The objective of this project was to develop a wireless system that provides a muscle-controlled access pathway to an individual with limited motor function. Raw EMG signals recorded in real time are processed in hardware and software and transmitted wirelessly (via Bluetooth) to a computer interface where a single muscle contraction is recognized as a left mouse click input. This will provide individuals with severe motor disabilities the opportunity and means to operate computers, which may have been previously inaccessible. This is a proof-of-concept prototype and its demonstration is anticipated to promote further research and development of a variety of wireless myoelectric access applications.

METHODS

EMG Data Acquisition

Several samples of muscle activity were taken using EMG through the MATLAB data acquisition tool. Three EMG AgCl-coated sensors were placed on the participant's forearm, oriented axially above the Flexor Carpi Radialis 2cm apart. The outer sensors were inputs to a difference amplifier and the central sensor was grounded. The microvolt signals were amplified with a gain of 10000 to bring them into the Volt range. Several samples of EMG during muscle rest and contraction were recorded. A dynamometer was used to measure the squeezing force of the contraction. The conversion from Newtons to volts was given by:

V_{out} = F (2.37mV / 100N) *gain, gain = 100000 Ω /R$_{gain}$.	(2)

where R_{gain} may be adjusted according to the input force of the participant. It was selected as 274 Ω in this experiment.

The EMG signals and dynamometer output were sampled at 10 kHz using a multi-function I/O card. The simultaneous recording of the dynamometer output and the EMG signals allows for temporal comparison between the EMG signal and the actual force output. This facilitated the design of the segmentation algorithm through visualization of a switch-event.

Segmentation Algorithm Establishment

The EMG data was analyzed and the segmentation algorithm was developed in MATLAB. The algorithm involved a two-part smoothing function of the absolute values of the signal, followed by a threshold-detection method to produce a binary output. The segmentation algorithm was then re-programmed in C in order to facilitate real-time implementation on a microcontroller.

Real-Time Segmentation

A Texas Instruments TMS270r1b1m microcontroller development board was used to implement the EMG segmentation algorithm in real time. The EMG sensors were connected to the input pins on the development board through a Grass Instrument AC Preamplifier (Model P55). An output pin of the microcontroller was set to HIGH when a switch event was recognized. This triggered a mouse click on the computer interface.

RESULTS AND DISCUSSION

The EMG and dynamometer data collected was analyzed in MATLAB, where an initial segmentation algorithm was developed. The segmentation program in C was tested in Visual Studio 6.0 with a five second EMG and dynamometer segment. One such example is shown in Figure 1.
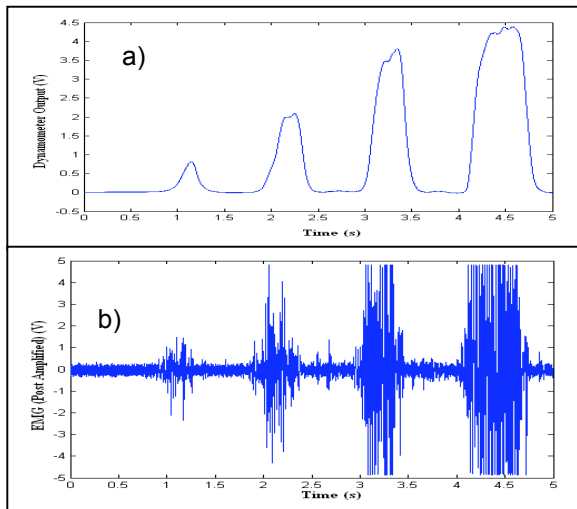

Figure 1: a) Dynamometer b) Corresponding EMG

The raw EMG signal was stored in a data array. From this signal, maximum and smooth arrays were constructed, the former consisting of the absolute maxima of a specified EMG window size (x=200) and the latter containing smoothed values over a consecutive number (y=9) of those maxima. This is shown in Figure 2.
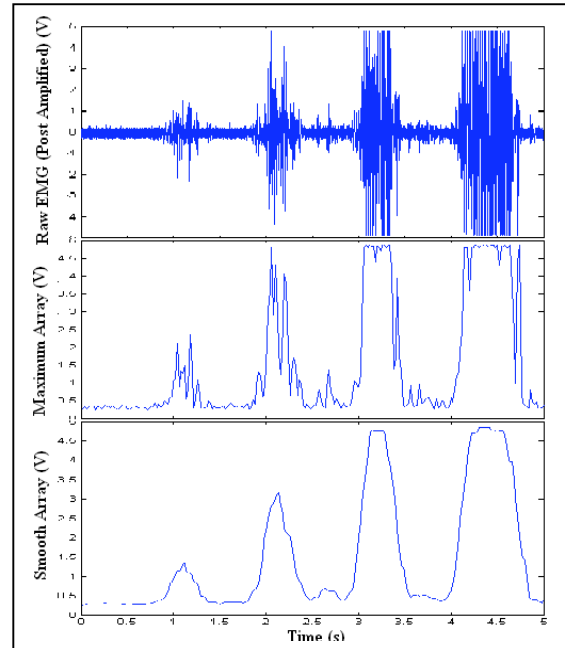

Figure 2: a) Original EMG b) Array of maxima c) Smoothed array

The purpose of the smooth array is to reduce the frequency in the processed EMG signals, so as to improve the accuracy of threshold methods. The threshold used to determine the binary value of the EMG at any given time was established by modeling noise at 1000 Hz sampling when the muscle was at rest. Noise was modeled by a normal distribution, to yield $\mu = -0.0172$ V and $\sigma = 0.1524$ V, as shown in Figure 3. The chosen segmentation threshold is a function of the standard deviation of this noise model. However, the sensitivity may be adjusted based on the individual user's needs including residual muscle capability.
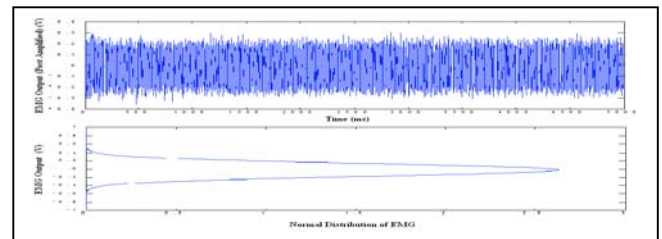

Figure 3: a) EMG at Rest b) Normal Distribution of Noise

$$\text{Threshold} = \mu + \text{Sensitivity} * \sigma \qquad (3)$$

The program for the real-time segmentation initiates with a calibration period of 5 seconds, in which the participant is asked to remain at rest while the resting EMG is recorded. The parameters of the noise model (μ and σ) are obtained from this resting signal. A threshold value of 5σ was chosen to digitize the smoothed signal shown in Figure 4.
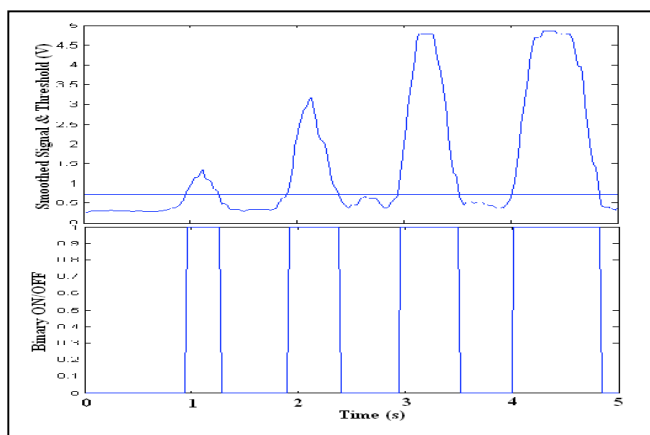

Figure 4: Binary output from smoothed signal

One important aspect of the real-time system is the latency between muscle input and switch output. In the real-time interface developed, the sampling frequency is 1 kHz. The latency may be estimated by:

$$L = x/\text{Sampling Frequency} + D_0 \qquad (4)$$

where x is the window size used to find the maxima and D_0 is the computational delay that includes microcontroller calculations and iterations, and transmission time between the wireless signal transmitter and the computer interface. Since computers operate in the MHz range, it can be assumed that D_0 is significantly less than the data recording time. Thus the latency for an EMG recording of 200ms at 1 kHz is approximately 0.2s.

CONCLUSION

The EMG segmentation system is now ready for subject testing and the smoothing parameters and sensitivity will be modified based on user feedback. The interface developed for this project is a practical prototype system with many potential applications in the field of Rehabilitation Engineering in academia and industry. This will provide individuals with severe motor disabilities the opportunity and means to operate computers.

ACKNOWLEDGEMENTS

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