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A ROBUST ALGORITHM FOR MUSCLE CONDUCTION VELOCITY ESTIMATION

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ABSTRACT

Muscle conduction velocity (CV) is used for detection of muscle fatigue and nerve malfunctioning, and for localization of the muscle innervation zone. Conventionally, CV is determined from the delay between two adjacent electromyogram (EMG) electrodes and the inter electrode distance (IED), i.e., CV=IED/delay. However, the accuracy of the CV estimate highly depends on the quality of the EMG signals. In this paper, a method to improve the robustness of the delay, and consequently the CV, estimate is developed and tested.

INTRODUCTION

Conduction velocity (CV) estimation from recorded surface EMG is an important indicator of muscle function [1] [2]. CV can also be used to estimate the position of the innervation zone from a multi-channel EMG recording over the length of a muscle. Merletti and Conte [3] described the mechanism by which the motor unit action potentials propagate and addressed a means to calculate CV by using the cross correlation function and calculating the difference in the latency of the signal at adjacent channels. This method is called the conventional method in the following text.

In this paper, techniques derived from digital signal [4] and image processing [5] [6] are incorporated in a novel algorithm to improve the robustness of muscle CV estimation. This algorithm is based on the assumption that the EMG propagates at a constant speed along the muscle fiber and maintains the same energy distribution. By using a weighting scheme derived, from the signal amplitude spectra, a more robust CV estimate is obtained, but at the cost of reduced SNR in the cross correlation.

METHODS

Experiment

Six healthy subjects participated in the experiment; the subjects provided informed consent prior to their participation. EMG was recorded from the long head and short head of the biceps brachii, the brachioradialis and the long head of the triceps brachii, for three force levels (20%, 35% and 50% maximum voluntary contraction (MVC)) in elbow flexion and extension. EMG signals were recorded using linear 8-electrode monopolar adhesive arrays (ELSCH008), with 5mm spacing between electrodes. The array was attached to the skin surface over each muscle, approximately aligned with the muscle fiber direction. The EMG was recorded at a sampling rate of 2048 Hz using the EMG-USB2 acquisition system (Bio Elettronica, Torino, Italy).

After the electrodes were attached, the subject placed the upper arm in a fixed bracket, with the forearm along a pivot bar, such that the arm was supported in the horizontal plane, in order to minimize the effect of gravity. The pivot bar was fixed at one of three joint angles (60^o, 90o and 120^o) at which the subject performed isometric contractions.

After data acquisition, normalization was applied to the monopolar signals by subtracting the signal mean and dividing by the standard deviation. Bipolar EMG signals were obtained by subtracting the adjacent monopolar channels.

Model

The activation signal sent from the spinal cord and carried by the motor neuron, stimulates the skeletal muscle fibers via the neuron's axonal terminals. Once the signal reaches the muscle fiber, it propagates in opposite directions

along the muscle fiber toward the tendons. In Figure 1, the signal s(t) propagates from the top to the bottom (innervation zone to tendon), and is delayed at successive electrode locations. In the ideal situation, the signal propagates at a constant speed and the inter electrode distances (IEDs) of multiple EMG recording electrodes, aligned with the muscle fibers are constant. Thus the delay, θ_T, is derived from θ_T=IED/CV, and is a constant between successive electrodes.

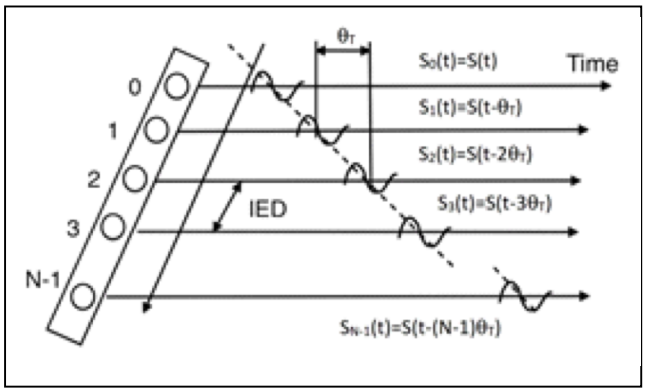

Figure 1 EMG acquisition mechanism [7]

Fourier transform

In the frequency domain, let the Fourier transform of the signal s(t), which is propagating on the muscle fiber, be denoted as S(k). Considering the time-shift property of the Fourier transform [8], the signal picked up from electrode 0 is $S_0(k) = S(k)$, while the signal from electrode 1 is $S_1(k) = S(k)e^{-i2\pi k\theta_T}$, and the signal from electrode N-1 is $S_{N-1}(k) = S(k)e^{-i2\pi k(N-1)\theta_T}$,. The delay of the signal in the frequency domain is only reflected in the phase, and ideally the signals from electrode 0 to electrode N-1 share the same amplitude distribution in the frequency domain.

In reality, it is common that a loose electrode connection or other noise artifact causes signal distortion, and the signal picked up from the electrode would be:

$$S_{N-1}(k) = S(k)e^{-i2\pi k(N-1)\theta_T} + A_{N-1}(k)$$

Where S(k) is the propagating signal, $A_{N-1}(k)$ is the artifact signal generated at electrode N-1, and (N-1) θ_T is the delay at electrode N-1 with respect to electrode 0.

If the frequency content of the artifact overlaps with the signal frequency content, it is not possible to denoise the signal simply by applying a low pass filter.

As shown in figure 2, the first signal (from the top to the bottom) is noisy, and has a noise spike in the frequency domain. The most dominant EMG frequency range is 10-500 Hz, which is (0.01-0.5) π in Figure 3.

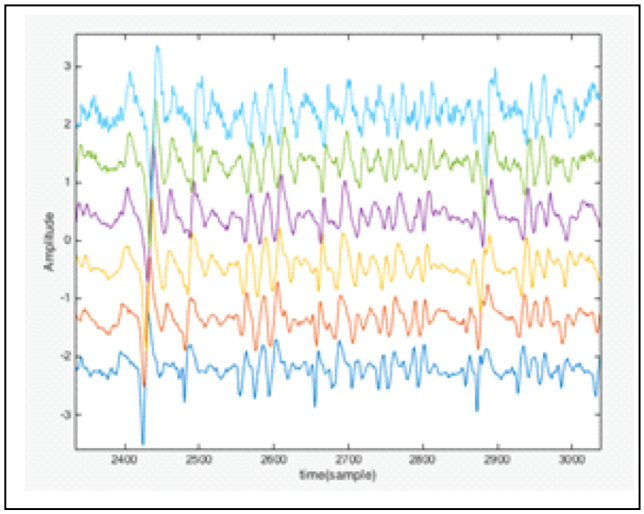

Figure 2 the time domain of 6 channels

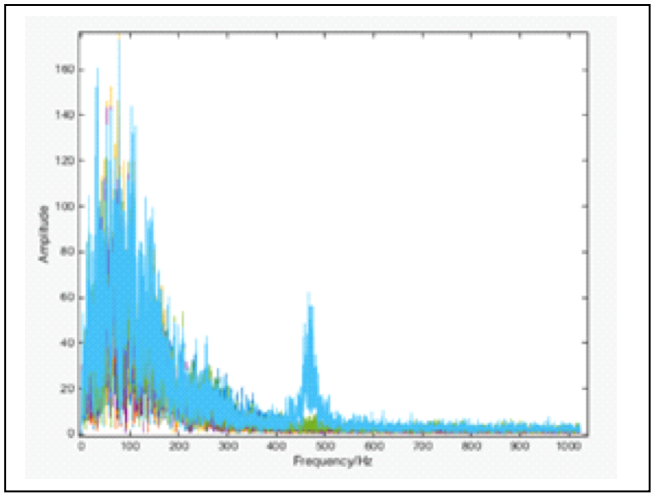

Figure 1 the frequency domain of 6 channels

Method

The bipolar signals are transformed into the frequency domain using the FFT function in Matlab. The signals are compared at each frequency, and the minimum amplitude is selected.

$$|S(k)|min = min \begin{cases} |S_0(k)| \\ |S_1(k)| \\ \vdots \\ |S_{N-1}(k)| \end{cases}$$

In the frequency domain, there is an amplitude and phase component at each frequency. The amplitude represents the signal energy at each frequency. The common shape of the signals over all channels can be preserved by keeping $|S(k)|min$ for each frequency. Thus the frequency amplitudes for all channels are replaced by the minimum amplitude at each frequency. After this process, all signal channels have the same energy and energy distribution, but the original phases, which represent the delay at each frequency, are retained.

After the steps above, the signals are transformed back into the time domain, using the inverse Fourier transform. Channel 0 ($S_0(k)$) and all channels ($S_0(k)$ - $S_{N-1}(k)$) are computed. After the delays are obtained, since each IED is known, the conduction velocity can be calculated, CV=IED x (N-1)/(N-1) θ_T for N>1.

RESULTS

A single bipolar EMG recording (50% MVC and 90o elbow joint angle), shown in figure 2, is tested to illustrate the robustness of the proposed method. Six bipolar channels which lie on one side of the innervation zone (as determined by the relative offsets) were extracted. The re-constructed signals obtained after applying the proposed method, are shown in Figure 4 and the amplitude spectra are shown in Figure 5. It is apparent that the reconstructed signals across channels are more similar, and that the noise in the top channel has been reduced.

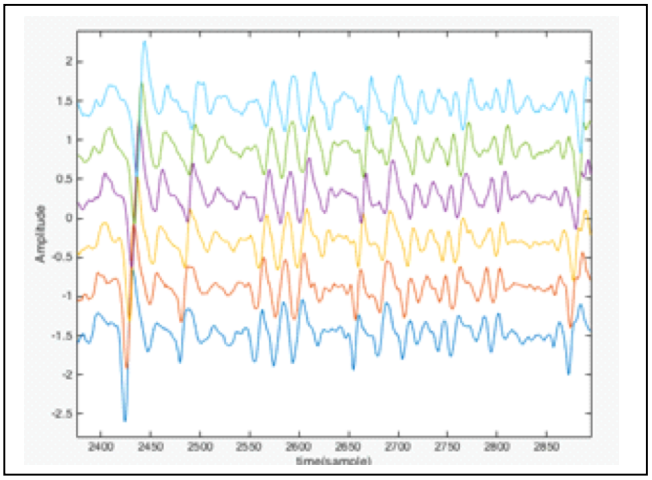

Figure 4 the time domain of the 6 channels after the proposed algorithm

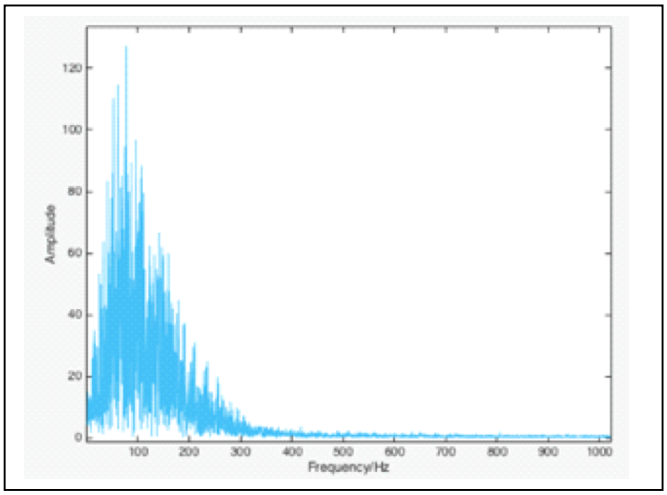

Figure 5 the amplitude spectra of 6 channels after the proposed algorithm

Figure 6 shows the cross-correlation SNRs, defined as the ratio of the highest peak to the second peak after taking the absolute value, for the proposed method and the conventional method. Although the SNR is lower for the proposed method, this is a traded-off for a better time delay estimate. However, in this case, the worst SNR is still greater than 2, permitting easy determination of the delay.

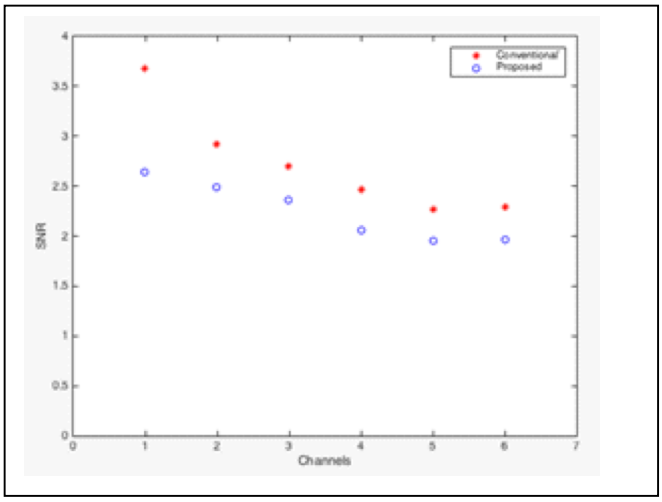

Figure 6 the SNR in the cross correlation after conventional method and proposed algorithm

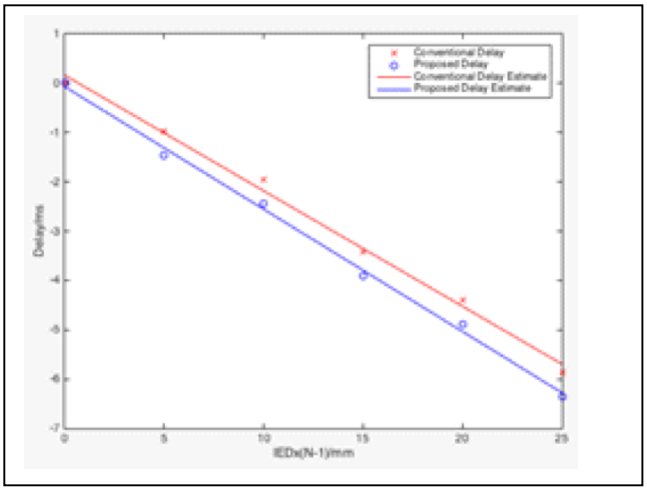

Figure 7 the time delay and the conduction velocity estimation

Time delays were obtained from the cross-correlations for the proposed and the conventional methods, as shown in Figure 7. A best-line fit minimum root mean square (rms) error of delay versus distance was determined for each set of delays, and conduction velocity is estimated as the inverse slope of the lines. In terms of the first order linear estimation, the rms error for the proposed method is e=0.3429, and for the conventional method is e=0.5333.

DISCUSSION

The underlying assumptions in developing the proposed processing method are that the same EMG signal propagates across all channels at a constant speed, and that a clean EMG signal is recorded on at least one channel. If these assumptions hold true, more accurate delay estimates are obtained from the cross correlations of the processed signals. Although the cross-correlation SNR decreased after processing, the highest peak was still the central lobe giving the correct delay estimate. This method also requires that the spectral content of the noise does not overlap the signal spectral content. If this is not the case, the proposed method will give the same results as the conventional method.

In the proposed method, selecting a reference channel for cross correlation is less challenging, since the signals across channels are more similar after processing.

Lastly, even though the results indicate that the proposed method gives a better delay estimate, the resolution of the estimate is limited by the sampling rate. In future work, the sampling rate will be increased to obtain a more precise delay estimate.

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REFERENCES

[1] S. Arjunan, D. Kumar, K. Wheeler, H. Shimada, G. Naik, "Spectral properties of surface EMG and muscle conduction velocity: A study based on sEMG model," Biosignals and Biorobotics Conference, 2011

[2] K. Pires, L. Pimenta, M. Andrade, D. Domingues, C. Mendes, A. Rocha, "Influence of neuromuscular electrical stimulation on the conduction velocity measured using EMG signals provided by linear arrays of electrodes," Engineering in Medicine and Biology Society, EMBC, Annual International Conference of the IEEE, pp. 3865-3868, 2011

[3] R. Merletti, L. Lo Conte, "Advances in processing of surface myoelectric signals: Part 1 and 2," Med. Biol. Eng. Comput., vol. 33, pp. 362-372 and 373-384, 1995

[4] C. Knapp, G. Carter, "The generalized correlation method for estimation of time delay," IEEE Trans Acous Speech Sig Proc., 1976.24(4):p.320-327, 1976

[5] G. Carter, "Coherence and time delay estimation," Proc. IEEE, 1987. 75:p.236-255, 1987

[6] A. Eckstein, "Development of robust correlation algorithms for image velocimetry using advanced filtering" master thesis of the Virginia Polytechnic Institute and State University, 2007

[7] D. Farina, E. Fortunato, R. Merletti, "Noninvasive estimation of motor unit conduction velocity distribution using linear electrode arrays", IEEE Transactions on Biomedical Engineering, vol. 47, no. 3, pp. 380-388, 2000

[8] E. Arthur, "Tables of integral transforms", vol. 1, McGraw-Hill, 1954

[9] K. Kapinchev, A. Bradu, F. Barnes, A. Podoleanu, "GPU implementation of cross-correlation for image generation in real time", Signal Processing and Communication Systems (ICSPCS), 9th International Conference, 2015