OS-9-2

A New Surface Electromyography Amplitude Estimation Method

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Abstract

Surface electromyography (sEMG) has been widely used to estimate muscle activity. To date, the estimators have been used such as MAV (mean absolute value) and RMS (root mean squre), and they are designed based on the statistical characteristics (probabilistic distribution) of raw sEMG amplitudes in a time window interval. However, their processing methods have no relevance to muscle physiology. We propose a new surface sEMG amplitude estimator, inspired by mechanical behavior of the muscle and characteristics of sEMG recordings, and they have more relevance to muscle physiology. This estimator might provide a better analysis tool for quantitative measurement of muscle activities in a large of research areas such as functional electrical stimulation, physiology, neuroscience, ergonomics, and biomedical engineering.

1. Introduction

The electrophysiological phenomena at the cell membrane reflect the active state of living cells. In this sense, the surface electromyography (sEMG) is connected to the complex activation of skeletal muscles which results in static and dynamic active force exertion and movement control. To date, with refined electrode and amplifier technologies, it is easier to obtain the signals from a muscle on the skin surface. However, it is still a big challenge to utilize those signals in order to obtain detailed, repeatable, reliable and meaningful information which represent an indirect assessment of muscular force without using invasive measurement techniques. The information should represent quantitative values which correspond to the degree of activation of skeletal muscles and be highly correlated to the muscle force. In this paper, we discuss a new sEMG amplitude estimator inspired by mechanical behavior of the muscle and sEMG generating mechanism. The purpose of this study is to design the estimator which provides a envelop signal with better qualities compared to the conventional ones such as moving average value (MAV) and root-mean squire (RMS). This purpose must be of great importance for the successful implementation of sEMG-based human-machine interaction or FES-based neural prosthesis. When the estimated envelop has low SNR, it produces the large variation (noise) which results in significant instability during the machine operation or FES. In addition, because low SNR represents low power of the signal, the signal resolution is also low, so that it could be difficult to delicately extract human movement intents. Part of this work has been presented at the World Congress 2009 Medical Physics and Biomedical Engineering [1].

2. Methods

2.1 SEMG amplitude estimator

SEMG is the electrical activity summation from a number of muscle fibers during muscle contraction by the electrode placed on the skin overlying the muscle. When motor unit action potentials (MUAPs) occur within close proximity, they make a signal with longer duration and greater magnitude, and it resembles a larger, single MUAP as shown in Fig. 1. In this sense, the number of activated muscle fibers can be approximated as the magnitude of raw sEMG at a turning point (where the gradient is zero). Also, the moment of the turning point could approximate the time of muscle contracting initiation.

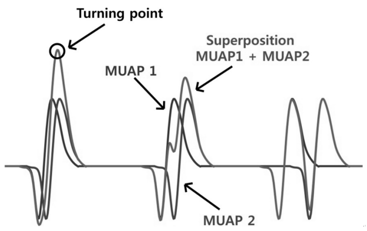

Fig. 1. A schematic example of sEMG generating mechanism when action potentials from two different MUAPs are summed together. For simplicity, only two MUAPs are described.

Extraction of two parameters, the number of the activated muscle fibers and the time of muscle contracting initiation, is important as far as the muscle recruitment properties and its firing rate modulation are concerned. According to Henneman's size principle, motor units are recruited in the order of their size, from smallest (low-threshold motor units, type I) to largest (high-threshold motor units. Type II) depending on the muscle force they must produce. Therefore, as the muscle force is increased, the number of recruited MUAPs is also increased. The magnitude of muscle contraction depends on not only the number of recruited muscle fibers, but also on the frequency of stimulation.

When two parameters, the number of the activated muscle fibers and the time of muscle contracting initiation, are achieved from raw sEMG, they can be represented as a unit impulse train. This unit impulse train in this study is analogous to a series of action potentials in the muscle. The unit impulse can be approximated by a pulse centered at the time i of muscle contracting initiation with a signal amplitude of the raw sEMG at the time i.

$$\begin{array}{ll} if\left(x_{i-1}-x_{i-2}\right)\times\left(x_i-x_{i-1}\right)<0, & y_i=x_{i-1} \\ else, & y_i=0 \end{array}$$

In the equation, x_i indicates raw sEMG signal at time i, and y_i indicates the sampled unit impulse at time i.

When an action potential arrives at muscle fibers, the smallest contractile response appears and this is called the "twitch". In this sense, we designed a sEMG amplitude estimator in which the twitch is developed when an impulse is created. The twitch can be approximated as the impulse response of a critically damped, second-order system [2], and the force is given as an equation of time by $f(t)$ as follows:

$$f(t)=a\cdot t\cdot\exp\left(-\frac{t}{T}\right),\quad a=P\cdot\frac{\exp(1)}{T}$$

where exp, P, and T are the natural logarithm, peak force, and contraction time, respectively. The impulse magnitude P represents the number of muscle fibers which are recruited, so that the impulse with a high P value creates a large peak force. Peak force P is determined by x_i because x_i indicates the number of the recruited muscle fibers. Contraction time T can be determined depending on the tasks and/or the muscle properties.

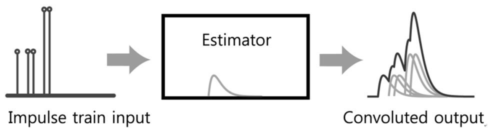

Fig. 2. Proposed sEMG amplitude estimator.

When a signal impulse is applied, an envelope is developed. When another impulse is applied and the applied time before the envelope had completely relaxed from the first impulse, a second envelope was added on top of the fist twitch. The resultant output can be expressed as a mathematical form of a convolution process as followings:

$$I(t)*f(t)$$

In the equations, $I(t)$ is the unit impulse train sampled from sEMG and $f(t)$ is the envelope.

2.2 Experiment

In order to verify the proposed method, it is required to collect sEMG data and force of the activated muscle. Palmar pinch, produced by the thumb and index finger tips together, was measured, because it was easy for a subject to control the volitional force by using their fingers. SMEG was measured from the first dorsal interosseous (FDI) muscle which mainly contributes the force for the pinch force. Then, the experimental task was performed by simultaneously measuring a pinch force and sEMG from the FDI muscle as shown in Fig. 3.

The maximum pinch force was limited to 5N to avoid a phenomenon muscle fatigue. When fatigue occurs in the muscle, the sEMG increases continuously during a constantly exerted force in

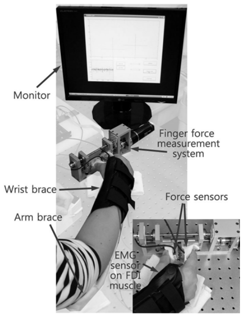

Fig. 3. Experimental setup.

isometric condition; the rate of increase is more pronounced when the force level is kept higher. A predefined pinch force trajectory was displayed, and the trajectory consisted of both ramp and hold. The duration of the hold level for both 0N and 5N was 5 seconds and the duration of the incremental was from 0N to 5N and decremental level from 5N to 0N is also 5 seconds. Subjects were requested to produce as the same pinch force as the displayed force level on the monitor and sEMG and force were recorded. This experiment was repeated 25 times for one subject.

3. Results

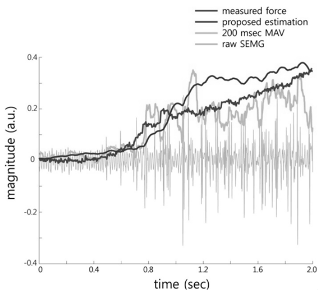

Fig. 4. Sample plot of the results.

Fig. 4 shows sample plots of the raw sEMG signal, the measured pinch force, the proposed sEMG amplitude estimation, and MAV estimation with 200 msec window interval, when the pinch force is initially developed. When the turning point is detected in sEMG (a gray line shown in Fig. 4), the smoothed is developed as a gray line shown in Fig. 4. This plot of the envelope can be compared it of the MAV estimation with a 200 msec window interval which is commonly used for the sEMG applications.

Figures from Fig. 5 shows simulation results of SNR. SNR of the proposed estimator is significantly higher compared to it of RMS with all window lengths (10~500 msec) used in this study and it of MAV with 10~300 msec window lengths ($P < 0.01$). SNR of the proposed estimator is no different than that of MAV with 400~500 msec.

4. Discussion and Conclusions

The observation, in which the estimation stabilities of the proposed estimator and MAV with large window lengths (>400 msec) are comparable, is an encouraging aspect we achieved. There is a trade-off between time delay and signal stability, and use of larger window length produces a significant delay leading to failure to detect rapid onset or offset (responsiveness) of muscle activities. In general, a 300 msec window length is regarded as a threshold of the delay that is perceivable by a subject [3]. The

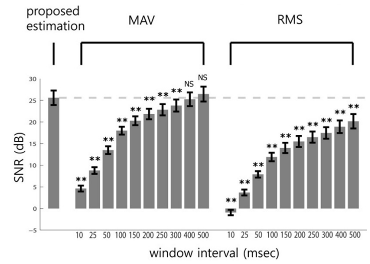

Fig. 5. SNR comparison between the proposed estimation and standard amplitude estimations (MAV and RMS with various window intervals). Significant differences from the proposed estimator are indicated as: **P<0.01; *P<0.1; NS: not significant.

time delay of the proposed estimator is 100 msec (contraction time T) required to reach the peak value from an impulse. Therefore, we could conclude that the proposed estimator relatively introduces a short delay which is not perceivable by a subject, but produce good SNR estimation values of sEMG.

Future works will progress on more experimental tests to prove the effectiveness of the proposed estimator. When the muscle force is changing, SNR is no longer meaningful. An alternative measure of performance is to display a real-time amplitude estimate to the subject as a form of biofeedback. Target amplitude will be displayed on a monitor and a subject will be requested to track it. The tracking error serves as a performance measure, with better EMG amplitude estimators presumably providing lower error.

The proposed estimator has more relevance to the physiology and may offer further insight into the physiological connections between sEMG and muscle activities. This estimator might provide a better analysis tool for quantitative measurement of muscle activities in a large of research areas such as functional electrical stimulation, physiology, neuroscience, ergonomics, and biomedical engineering.

References

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Acknowledgements

This work was supported by the Korea Science and Engineering Foundation (KOSEF) grant funded by the Korea government (MEST) (No. R01-2007-000-11659-0).