Motor unit distribution estimation by multichannel surface EMG

MOTOR UNIT DISTRIBUTION ESTIMATION BY MULTI-CHANNEL SURFACE EMG

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Abstract

A new method for the non invasive estimation of motor unit distribution within the muscle is proposed. It is based on the surface EMG signal detected at different contraction levels in single differential configuration from an array of electrodes placed over the considered muscle in the direction orthogonal to the fibres.

1 Introduction

Surface EMG has found applications in basic physiology and clinical studies as a non invasive technique to investigate muscle contraction [2]. Multi-channel surface EMG has been proposed to overcome the lower selectivity with respect to needle EMG and to provide information on a wide area over the considered muscle. Important information can be extracted from multi-channel surface EMG, e.g. on the anatomy of the muscle (position of innervation zone and tendon endings) and on the velocity of propagation of motor unit (MU) action potentials (MUAP). This work proposes a new method which extends the potential application of multi-channel surface EMG to the non invasive estimation of the spatial distribution of MUs within a muscle, as an alternative to biopsies [7][8].

2 Methods

2.1 The algorithm

Surface MU potential magnitude detected by an array of electrodes placed transversal to the muscle fibres has a maximum on the electrode placed over the MU and decreases for more distant recording electrodes. The steepness of the decrease depends on MU depth within the muscle (it is larger for superficial MUs) and was proposed as an estimator of MU depth [10].

Spatial and temporal MU recruitment are the strategies by which the central nervous system regulates the level of a contraction. The Henneman size principle assumes that small MUs are recruited first, and larger MUs are involved in the contraction as the required force level increases [6]. If larger MUs are preferentially distributed superficial or deep within the considered muscle, the recruitment of the MUs at different force levels is also related to the activation of different portions of the muscle. This allows the investigation of MU distribution within the muscle by comparison of surface EMG signal at low (only small MUs recruited) and high force levels (all MUs recruited) recorded by a linear array of electrodes placed orthogonal to the direction of the fibres.

As MUs are differently displaced with respect to the centre of the detection array, the separation of their contribution should be required to estimate MU depth by examining the decay of the potential distribution in the space domain. A simpler and faster global estimation (which doesn't require the separation of the contributions of different MUs) can be obtained by transforming the space domain (i.e. the set of detection points of the electrode array) into the Fourier domain, i.e. considering surface EMG signal s(x,t) into a function depending on spatial frequency and time $S(k_x,t)$. Indeed, the magnitude of the Fourier transform ($|S(k_x,t)|$) is not sensitive to the displacement of different MUs with respect to the centre of the detection array. The second order moment $w^2(t)$ of the magnitude of the Fourier transform averaged over time $< w^2(t) >$ (referred to as "averaged second order moment of magnitude") is proposed to estimate the location of the MUs

$$w^2(t) = \frac{\int (k_x - \mu_{PSD})^2 |S(k_x,t)| dk_x}{\int |S(k_x,t)| dk_x}$$

where μ_{PSD} is the mean of the magnitude of the Fourier transform. As the magnitude of the Fourier transform has a spread which is inversely related to the spread of the transformed function, a larger/smaller spread of the magnitude of the Fourier transform is related to superficial/deep MUs.

When different force levels are considered, $< w^2(t) >$ increases with increasing force level if the larger MUs

(recruited only during high level contractions) are located deeper than smaller MUs (recruited also during low level contractions).

2.2 Simulations

The plane layer model of generation of surface EMG signals proposed in [5] was considered to simulate MUAPs. Fat layer thickness ranged between 2÷6 mm (2 mm step). Finite length fibres (120 mm long) symmetrical with respect to the innervation zone were simulated with depth in the range 1÷10 mm (1 mm step) and transversal displacement with respect to the direction of the fibres in the range -50÷50 mm (step 5 mm). Each single fibre action potentials (SFAP) was used to simulate a MUAP, approximating the smoothing due to the spread of the innervation zone and tendon endings (8 mm) by a time convolution with a Gaussian function.
The distribution of muscle fibre conduction velocity (CV) of the simulated MUs was Gaussian, with mean value and standard deviation depending on the set of simulations, corresponding to the following three different distributions of MUs within the muscle:
1) *Uniform distribution of MUs*: mean CV of 4 m/s and standard deviation of 0.4 m/s.
2) *Larger MUs Deeper* in the muscle: mean CV varying linearly between 3.5 and 4.5 m/s as a function of the depth within the muscle, standard deviation of 0.2 m/s.
3) *Smaller MUs Deeper* in the muscle: mean CV varying linearly between 4.5 and 3.5 m/s as a function of the depth within the muscle, standard deviation of 0.2 m/s.

The number of fibres in the MUs was distributed as an exponential function [3], with the largest MU including a number of fibres 20 times larger than that of the smallest MU. Higher values of CV were associated to larger MUs [1].
For each set of simulation parameters, 8 signals were simulated (each referred to as a "simulated subject") considering different random choices of the sizes of the MUs and of the CV distribution.

2.3 Experimental protocol

Surface EMG signals were detected in monopolar configuration from the dominant biceps brachii of 6 healthy subjects using a two dimensional system of electrodes (interelectrode distance 8 mm) placed half way between the innervation zone and the distal tendon. The reference electrode was placed on the wrist of the dominant arm. Driven Right Leg circuit was used to reduce power line interference. Single differential detection with electrode pairs along the direction of the fibres where obtained for a linear array placed transversal to the fibres. The experimental protocol consisted of an isometric task of 5 s selective for the biceps brachii muscle at different force levels (10% - 80% of maximal voluntary contraction - MVC, with step of 10% MVC).

2.4 Statistical Analysis

Analysis of variance (ANOVA) was applied on $< w^2(t) >$ estimated from simulated signals to assess possible statistical dependence on the following 4 factors: MU distribution within the muscle, fat layer thickness, contraction level, simulated subject. Significance was set to $p<0.01$. Wilcoxon signed rank test for zero median was applied to the slopes of the regression lines of $< w^2(t) >$ estimated from experimental signals.

3 Results

When tested on simulated signals, $< w^2(t) >$ was statistically dependent on the contraction level and on MU distribution within the muscle, but it was not dependent on the simulated subject and on the fat layer thickness. Figure 1 shows $< w^2(t) >$ in the case of simulated signals. Averaged second order moment of magnitude $< w^2(t) >$ increases with increasing force level in the case in which smaller MUs are placed deep in the muscle, vice versa if smaller MUs are superficial.

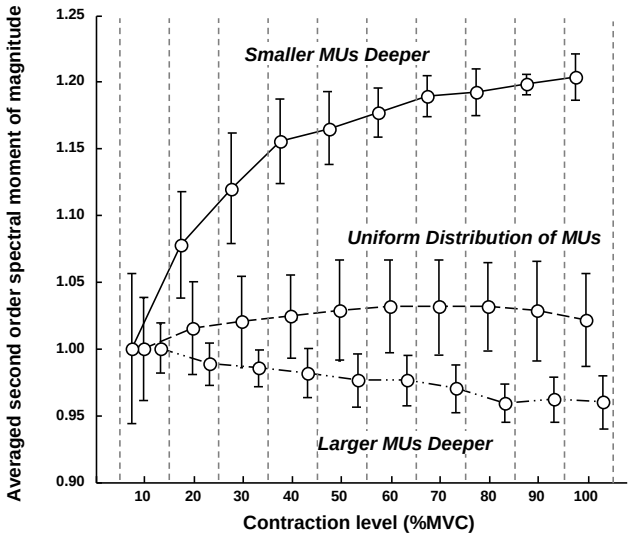

Figure 1. Mean and standard deviation across 8 "simulated subjects" of averaged second order moment of magnitude $< w^2(t) >$ normalised with respect to the mean value corresponding to 10 % MVC (fat layer thickness 2 mm).

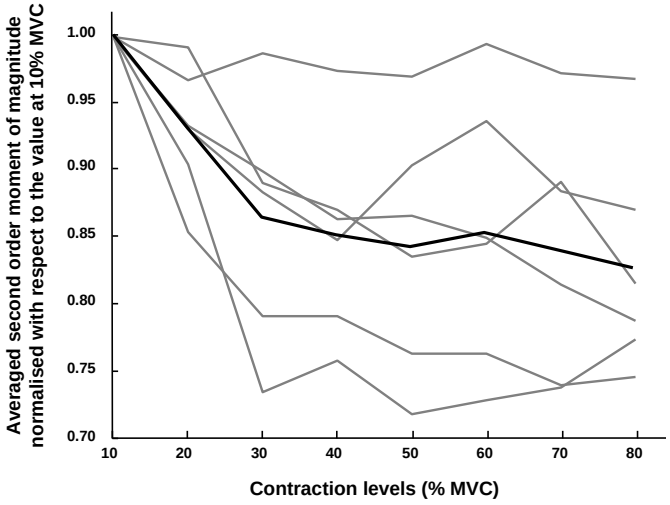

Figure 2. Application to experimental signals recorded over the biceps brachii muscle of three healthy subjects. Averaged second order moment of magnitude $< w^2(t) >$ normalised with respect to the value corresponding to 10 % MVC.

Figure 2 shows the application of the method to experimental signals (averaging on a portion of the signal 2 s long). As $< w^2(t) >$ decreases when force level is increased (statistically significant changes indicated by the Wilcoxon signed rank test for zero median), it is argued that biceps brachii has superficial MUs which are small.

4 Conclusions

A new method for the non invasive estimation of MU distribution within a muscle is proposed. It is based on multi-channel detection of surface EMG signal during voluntary contractions at different force levels. The method shows good performances when applied to simulated signals (Figure 1).

A preliminary experimental application to signals recorded over the biceps brachii muscle suggests that a large variability of MU distribution within the same muscle of different subjects can be expected (Figure 2). Nevertheless, MU distribution within biceps brachii of all the three subjects investigated in this study is in line with the simulation results only if larger MUs are distributed preferentially deeper in the muscle.

This partly contrasts with the larger percentage of type II fibres in the superficial than in the deep muscle layers of the biceps brachii muscle [9]. On the other hand, by comparing experimental and simulated signals, it was suggested in [4] that, with increasing transcutaneous electrical stimulation, MUs in biceps brachii are recruited from low to high CV and from superficial to deep muscle layer. This indicates the possibility that a fixed relation between fibre type and size (and hence CV) could be incorrect. For example, both type I and II muscle fibres have a larger diameter in the deep than in the superficial muscle layers of the tibialis anterior muscle [7]. Furthermore, in some muscles, an inverse association between muscle-fibre diameter and fibre type occurs [8].

Finally, this work proposes a method which opens new potential applications of multi-channel surface EMG signal. Further experimental investigation on different superficial muscles is suggested.

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