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A new method for the localization of the innervation zone based on monopolar surface-detected potentials

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ABSTRACT

In monopolar electrode configuration, the shape of the motor unit potential (MUP) undergoes essential morphologic changes with increasing distance from the innervation zone. In particular, the rising phase of the MUP slows down and becomes longer as the electrode is moved further from the innervation zone. Based on this, it is hypothesized that the maximum slope of the rising phase of a monopolar MUP ($Slope_{MAX}$) would reach its highest value at the innervation zone. Herein, we sought to determine whether the location of the innervation zone coincides with the position in the muscle of the monopolar MUP with highest $Slope_{MAX}$ and whether new criteria for the localization of the innervation region can be proposed based on the $Slope_{MAX}$ of monopolar potentials. Multichannel surface (13×5 electrode array) and intramuscular (wire electrodes inserted with needles of lengths 15 and 25 mm) EMG signals were concurrently recorded in monopolar configuration from the biceps brachii muscle of 10 healthy subjects. The spatial distribution of monopolar and bipolar MUPs along the fibers' direction were obtained by spike-triggered averaging of the surface EMG. We found that the monopolar MUP with steepest rising phase (i.e., highest $Slope_{MAX}$) was located either just above or at half of the inter-electrode distance from the innervation zone. High levels of agreement (94–98%) were found between the position of the innervation zone assessed with the $Slope_{MAX}$ criteria of monopolar potentials and the position identified using the phase reversal criteria of bipolar potentials. The present results legitimize the use of compound muscle action potentials (M waves) to localize the innervation zone: this location would correspond to the electrode position which yielded the monopolar M wave with highest $Slope_{MAX}$.

1. Introduction

The axons of a motor nerve divide into a number of smaller branches as they approach the muscle. Each of these terminal branches contacts the muscle fiber at a specific site, referred to as neuromuscular junction. These junctions tend to cluster in a narrow region, often in the center of the muscle belly, which is termed the *innervation zone*. The location of the innervation zone can be determined by surface electromyography (sEMG) using linear arrays of electrodes oriented along the muscle fibers' direction (Masuda et al., 1983). Classical methods of identification of the innervation region are based on the fact that action potentials propagate in opposite directions from the neuromuscular junctions toward the tendons (Masuda et al., 1985). Due to this oppositely directed propagation, if signals are recorded in single differential (bipolar) configuration, the innervation zone can be identified as the point of sign inversion or minimum signal amplitude of the bipolar recordings (Merletti et al., 1999, 2003). Knowledge on the location of muscle innervation zones has special relevance in the clinical practice: for example, to reduce the quantity of injected botulinum toxin for the treatment of spasticity and to guide episiotomy during child delivery. In this context, several automatic methods have been proposed for the estimation of the innervation zone distribution in the external anal sphincter from surface EMG (Mesin et al., 2009; Ullah et al., 2014). Moreover, precise identification of the innervation zone is essential in the sports field since it has been demonstrated that the biceps brachii innervation zone may shift several millimiters as a result of prolonged passive static stretching (Ye et al., 2015).

Surface electrode arrays (Fig. 1) have been proven successful in identifying the innervation zone position in a large number of lower and upper limb muscles (Saitou et al., 2000; Beretta Piccoli et al., 2014), with studies comparing such identification between men and women (DeFreitas et al., 2010), and efforts directed to standarize the electrode placement (Rainoldi et al., 2004). Recently, a number of advanced methods have been developed to enhance

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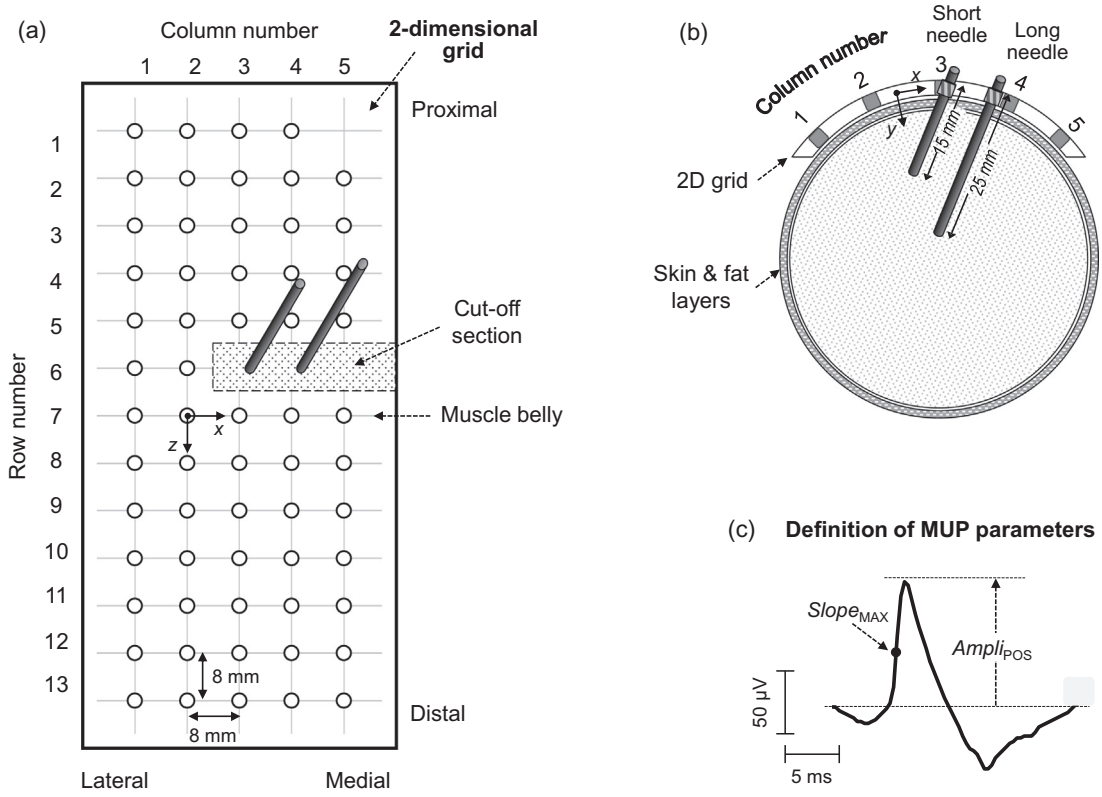

Fig. 1. Schematic representation of the adhesive 2D array placed over the biceps brachii under a frontal (a) and cross section (b) view. Surface electrodes were arranged in a 13 × 5 array with an inter-electrode distance of 8 mm. Columns of the array were oriented along the direction of the muscle fibers. The 7th row of the array was lined up with the center of the muscle belly, where the innervation zone of the biceps brachii is normally found. Two intramuscular electrodes, one short and one long, were inserted at the level of the 6th row of the array. The estimated depths of the short and long needles were 10 and 20 mm, respectively, relative to the skin surface. (c) Representation of the maximum slope of the rising phase of the MUP ($Slope_{MAX}$) and the amplitude of the positive peak.

the performance in the identification of the innervation zone. Indeed, Beck et al. (2012) showed that the cross-correlation-based approach is more accurate than the lowest amplitude and highest mean frequency criteria (Merletti et al., 1999, 2003). Identification of the innervation zone has been also addressed by estimating the motion of the potential distribution using an optical-flow-based technique (Ostlund et al., 2007). This method was proven to be superior than the lowest amplitude criteria for low signal-to-noise ratios. Interestingly, Marateb et al. (2016) developed a fully-automatic algorithm to identify multiple innervation zones in multi-channel sEMG signals with low signal-to-noise ratios. The proposed algorithm is characterized by its high robustness and accuracy as it correctly and precisely identifies multiple innervation zones in a wide range of signal quality.

Despite the advantages of multichannel surface EMG for the identification of the innervation zone, the use of this technique is still limited due to technical difficulties posed by the experimental setup. Moreover, preparation of the experimental setup is time consuming, rendering it less viable for use in the clinical and sports practice. Therefore, it would be valuable to have an alternative method that was less time-consuming and more experimentally accessible (and efficient) and clinically applicable than multichannel surface EMG.

In the sport science field, the innervation zone is often identified by means of the muscle compound action potential (M wave) recorded in monopolar mode. In particular, it is assumed that the innervation zone is located at the electrode position which "yielded the monopolar M wave with the steepest rate of rise of the initial positivity" (Brown et al., 1996). While this assumption is widely spread in the scientific community, it remains empirically unsupported (Lateva et al., 1996; Nandedkar and Barkhaus, 2007). We propose that an easy straightforward way to experimentally validate this assumption would be to verify whether the location of the innervation zone coincides with the position in the array of the monopolar MUP with the steepest rising phase. To do so, it would be necessary to extract the spatial distribution of monopolar MUPs along the fibers' direction.

In monopolar configuration, the propagation of the action potentials along the fibers can be recognized because the shape of the MUP undergoes essential morphologic changes with increasing distance from the innervation zone (Gydikov and Kosarov, 1972; Kleine et al., 2007). Specifically, the latency of the MUP positive peak increases and its amplitude decreases as the electrode is moved away from the innervation zone (Fig. 2). Based on this shape transformation, it is hypothesized that the maximum slope of the rising phase of the monopolar MUP, hereafter referred to as $Slope_{MAX}$ [Fig. 1(c)], would be at its highest at the center of the innervation zone, and this slope would markedly decrease as the MUP is detected further from the innervation zone (Fig. 2). This hypothesis, however, may be less applicable for motor units located deep in the muscle. The reason is that the waveform shape of *deep* MUPs is relatively unaltered in the fibers' direction, and hence the amplitude and slope characteristics of these MUPs may change very little as the electrode moves away from the innervation region [for an example see Fig. 2(b)].

One critical problem of surface-detected MUPs is that these potentials are affected by a number of uncontrollable factors, such as inhomogeneities of the subcutaneous layers, defectuous contact of the electrode, and local properties of the volume conductor (Farina and Rainoldi, 1999). These factors can greatly influence