Medical Engineering & Physics 33 (2011) 427–437

Contents lists available at ScienceDirect

Medical Engineering & Physics

journal homepage: www.elsevier.com/locate/medengphy

Experimental and modelling investigation of surface EMG spike analysis

David A. Gabriel [a,*], Anita Christie [b], J. Greig Inglis [a], Gary Kamen [b]

[a] *Electromyographic Kinesiology Laboratory, Faculty of Applied Health Sciences, Brock University, 500 Glenridge Avenue, St. Catharines, ON, Canada L2S 3A1*
[b] *Exercise Neuroscience Laboratory, Department of Kinesiology, University of Massachusetts Amherst, Amherst, MA 01003, USA*

ARTICLE INFO

Article history:
Received 23 March 2010
Received in revised form 27 October 2010
Accepted 12 November 2010

Keywords:
Electromyography
Signal Processing
Surface EMG modelling
Motor unit behaviour
Biceps brachii

ABSTRACT

A pattern classification method based on five measures extracted from the surface electromyographic (sEMG) signal is used to provide a unique characterization of the interference pattern for different motor unit behaviours. This study investigated the sensitivity of the five sEMG measures during the force gradation process. Tissue and electrode filtering effects were further evaluated using a sEMG model. Subjects ($N = 8$) performed isometric elbow flexion contractions from 0 to 100% MVC. The sEMG signals from the biceps brachii were recorded simultaneously with force. The basic building block of the sEMG model was the detection of single fibre action potentials (SFAPs) through a homogeneous, equivalent isotropic, infinite volume conduction medium. The SFAPs were summed to generate single motor unit action potentials. The physiologic properties from a well-known muscle model and motor unit recruitment and firing rate schemes were combined to generate synthetic sEMG signals. The following pattern classification measures were calculated: mean spike amplitude, mean spike frequency, mean spike slope, mean spike duration, and the mean number of peaks per spike. Root-mean-square amplitude and mean power frequency were also calculated. Taken together, the experimental data and modelling analysis showed that below 50% MVC, the pattern classification measures were more sensitive to changes in force than traditional time and frequency measures. However, there are additional limitations associated with electrode distance from the source that must be explored further. Future experimental work should ensure that the inter-electrode distance is no greater than 1 cm to mitigate the effects of tissue filtering.

1. Introduction

Traditional time and frequency analyses of the surface electromyographic (sEMG) signal have had limited success in differentiating between types of motor unit behaviours because they produce similar changes in the two domains. For example, high motor unit firing rates and synchronization levels increase sEMG amplitude and decrease the frequency content due to an increased probability of temporal overlap between motor unit action potentials [1–5]. The ability to monitor changes in motor unit activity patterns from the sEMG signal is also complicated by several factors that include: (1) wave cancellation, (2) low-pass filtering by the tissues (fat, fascia, and skin), and (3) the physical characteristics of the electrode detection system [6,7].

Problems in the traditional analysis of the surface electromyographic (sEMG) signal have been well documented [8]. Previous work has described an alternative technique using a pattern classification method [9]. Five measures extracted from the sEMG signal are used to provide a unique characterization of the interference pattern for different motor unit behaviours. This method is based on the observation that the summations of a population of motor unit action potentials for one type of motor unit behaviour result in an interference pattern that is quantitatively different from one generated by another type of behaviour [10–25]. The five measures are mean spike amplitude (MSA), mean spike frequency (MSF), mean spike duration (MSD), mean spike slope (MSS), and the mean number of peaks per spike (MNPPS).

The sEMG signal is assumed to be a composite of discrete spikes and peaks, each shaped by a pair of upward and downward deflections. Spikes are differentiated from peaks as both deflections of a spike cross the 99% confidence interval for baseline noise (see Fig. 1). A peak is any pair of upward and downward deflections within a spike that do not altogether constitute a spike. Spike #2 has a single peak. In the case of a spike with a single peak, the deflections of the spike and the peak are one and the same (spike #1). Any deflections that do not constitute discrete spikes as previously defined and are found before, between, or after identified spikes are assumed to be background noise and not sEMG signal. The deflections between spikes #1 and #2 do not constitute a complete spike. The peak-to-peak amplitude of each spike is used to calculate MSA. The number of spikes that occur within a 1-s time period is used to calculate MSF. The duration of the base of each spike is used to calculate MSD. Spike slope is calculated from the onset of the base to the peak of the signal, and is used to determine

* Corresponding author. Tel.: +1 905 688 5550x4362; fax: +1 905 688 8364.
E-mail address: dgabriel@brocku.ca (D.A. Gabriel).

1350-4533/$ – see front matter © 2010 IPEM. Published by Elsevier Ltd. All rights reserved.
doi:10.1016/j.medengphy.2010.11.010

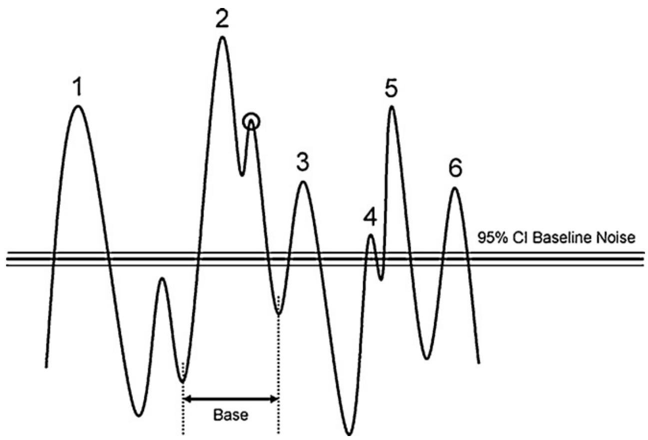

Fig. 1. Surface electromyographic (sEMG) interference pattern to illustrate the difference between a spike and a peak. A spike is composed of a single upward and downward deflection that are greater than the 95% confidence (CI) interval for baseline activity. There are six spikes and spike #2 one peak (circled).

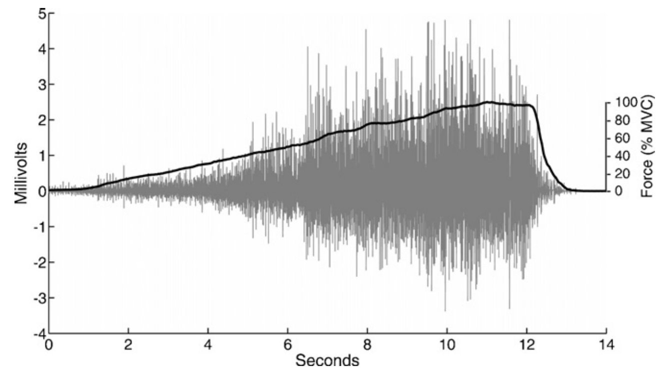

Fig. 2. Representative force (dark line) and biceps brachii (BB) surface electromyographic (sEMG) activity (grey) for one subject who performed a ramp increase in force from 0 to 100% of maximal voluntary contraction (MVC). Since subjects performed ramp contractions at 10% MVC per second, the *x*-axis in seconds also corresponds to percent MVC.

MSS. The number of peaks for each spike used to calculate MNPPS.

A pattern classification table is then used to evaluate changes in the sEMG signal with increases in muscle force and its underlying motor unit activity. Each type of motor unit behaviour (rate coding, recruitment, or synchronization) is associated with a unique pattern of change in the five measures of the sEMG signal. The pattern classification table was constructed based on direct observation of both indwelling [14,17,18,25,26] and surface [20–24] recordings. Luttmann et al. [27] proposed a similar approach by advocating joint analysis of the sEMG spectrum and amplitude (JASA) simultaneously to determine the onset of muscle fatigue based on known relationships between muscle force and these two sEMG measures. Simultaneous analysis of multiple measures allows for greater specificity in identifying changes in the sEMG interference pattern [22–24].

Because the five sEMG measures of the interference pattern essentially describe the average shape of individual spikes, the pattern classification method has been termed “spike shape analysis” [18]. Spike shape analysis is directed at clinical and kinesiological applications wherein forces can range from 0 to 100% MVC. For example, patients generally perform only moderate muscular contractions (<50% MVC) as part of the clinical exam associated with electrodiagnostic testing [28–30]. In contrast, functional assessment following physical medicine and rehabilitation or other interventions can require muscle contractions much greater than 50% MVC [31,32]. The purpose of this paper was to evaluate the sensitivity of spike shape measures throughout the force gradation process, and how the measures are affected by both tissue and electrode spatial filtering. An assessment of sensitivity is important because the pattern classification method for spike shape analysis depends on observing changes with increasing force levels. To this end, subjects performed a ramp increase in isometric elbow flexion force from 0 to 100% MVC. The effects of tissue and electrode spatial filtering were further evaluated using a sEMG model.

2. Methods

2.1. Experimental methods

2.1.1. Testing procedures

All subjects ($N = 8$) signed an informed consent document prior to participation in accordance with the university guidelines and ethical treatment of human subjects. Subjects visited the lab on one occasion for testing. Surface EMG recordings were obtained from the medial head of the biceps brachii while subjects performed isometric elbow flexion to 100% of maximal voluntary contraction (MVC). Subjects were seated in a chair with the elbow resting on a platform directly in front of them. The shoulder and elbow were flexed to approximately 90° and the wrist was placed at 180°. A custom-molded fiberglass splint was worn on the anterior side of the right wrist and forearm. The splint fit firmly against a force transducer and the transducer was aligned with the wrist. Isometric elbow flexion was performed by pulling against the force transducer.

The testing protocol involved the use of ramp contractions according to the procedures outlined by Sbriccoli et al. [33]. The maximal voluntary contraction (MVC) force of each individual was determined by having subjects perform a minimum of three MVC contractions, with 2 min of rest between contractions. The highest value obtained by each subject was taken as the MVC. Subjects then performed 5, 10-s contractions from 0 to 100% MVC, increasing force at a rate of 10% MVC per second. The 100% MVC at the end of each ramp was held for an additional 1 s to ensure that subjects achieved their maximum. These ramp maximal contractions were accomplished by having subjects follow a force trajectory with real-time feedback of their force presented on a computer screen. At least 5 min rest was given between each contraction.

2.1.2. Measurement of force and sEMG

The force exerted during elbow flexion was measured with a strain gauge force transducer (Interface Model MB-250, Scottsdale, AZ), secured to a metal plate. The force signal was amplified and low-pass filtered at 10 Hz with a hardware amplifier/filter (DataQ PM 1000, DataQ Instruments, Akron, OH). The force signal was sampled at 50 Hz using a 16-bit A/D converter (DT-322, Data Translation, Marlboro, MA), and was displayed on a computer screen for real-time feedback to subjects using DasyLab software (Data Acquisition System Laboratory, DasyTec, USA, Inc, Amherst, NH). The force signal was also sent to a second computer where it was sampled at 25.6 kHz using a 16 bit A/D converter (NI-DAQmx PCI-6251, National Instruments, Austin, TX), and was stored for offline analysis. Surface EMG was recorded with 3 mm diameter Ag/AgCl electrodes placed in a bipolar configuration with an inter-electrode (IED) of 2 cm. The electrodes were placed between the center of the muscle and the distal tendon, away from the motor point. The sEMG signal was amplified and band-pass filtered at 10–2000 Hz, using a Dantec Counterpoint Electromyograph (Dantec Electronik Medicinsk, Sklovlunde, Denmark), before being sampled at 25.6 kHz.

Representative force and sEMG traces from a subject are presented in Fig. 2. Surface EMG data were assessed over 500 ms