Analysis of intramuscular electromyogram signals

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Intramuscular electromyographic (EMG) signals are detected with needles or wires inserted into muscles. With respect to non-invasive techniques, intramuscular electromyography has high selectivity for individual motor unit action potentials and is thus used to measure motor unit activity. Decomposition of intramuscular signals into individual motor unit action potentials consists in detection and classification, usually followed by separation of superimposed action potentials. Although intramuscular EMG signal decomposition is the primary tool for physiological investigations of motor unit properties, it is rarely applied in clinical routine, because of the need for human interaction and the difficulty in interpreting the quantitative data provided by EMG signal decomposition to support clinical decisions. The current clinical use of intramuscular EMG signals relates to the diagnosis of myopathies, of diseases of the α-motor neuron and of the neuromuscular junction through the analysis of the interference signal or of the shape of some motor unit action potentials, usually without a full decomposition of the signal.

Keywords: intramuscular electromyogram; decomposition; integration of invasive and non-invasive electromyogram analysis techniques; muscle; peripheral nervous system

1. Introduction

The motor unit consists of a motor neuron and the muscle fibres innervated by its axonal branches. The number of muscle fibres it contains ranges widely across human muscles. The muscle fibres of each motor unit are intermingled with fibres of other motor units so that fibres belonging to several motor units are close to each other. The fibres of a motor unit occupy a portion of the cross section of a muscle (motor unit territory), which has an irregular round shape (Stålberg et al. 1995; Stålberg & Falck 1997; Trontelj et al. 2004).

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One contribution of 13 to a Theme Issue ‘Signal processing in vital rhythms and signs’.

Published online 11 November 2008

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The electrical activity associated with the contraction of muscle fibres in the motor units can be recorded using either intramuscular or non-invasive (surface) detection systems. The difference between the two detection modalities consists in the effect of the volume conductor that separates the muscle fibres from the detecting electrodes; this effect can be modelled as a low-pass filter whose selectivity depends on the distance between the electrodes and the source. As compared with surface electromyographic (EMG) recording, the insertion of electrodes into muscles allows the detection of electric potentials close to the muscle fibres, thus the effect of the volume conductor is limited. For this reason the action potentials of individual motor units can be easily identified from the interference signal, at least at moderate force levels.

The invasive recording of EMG signals also allows the measure of different types of spontaneous activity, e.g. fibrillation potentials due to denervation, as well as changes in the motor unit potentials related to disorders affecting the muscle, the neuromuscular junction or the peripheral nerves. The traditional clinical use of intramuscular electromyography is in the diagnosis of myopathies, diseases of the neuromuscular junction and of the lower motor neuron through the observation of features of the detected individual action potentials (single fibre electromyography) up to the interference signal. This analysis is based on signal features (e.g. number of turns) that differ in healthy conditions and diseases. Most of these features are lost in surface recordings owing to the severe low-pass filtering and diffusion due to the volume conductor, which causes the EMG signal bandwidth to reduce from more that 1 kHz, typical of the intramuscular signals, to less than 400 Hz, typical of the surface recordings.

Needle or wire electrodes inserted in the contracting muscle record individual motor unit action potentials. Depending on the type of electrode used and its location (see figure 1), the recorded action potentials can be the result of the activity of a small (1–3), moderate (15–20) or large (more than 20) number of muscle fibres. The action potentials generated by fibres belonging to the same motor unit are not time aligned owing to the different locations of the end-plates (figure 2).

In neurophysiologic investigations, intramuscular recordings are often used to identify the discharge times of individual motor units (De Luca et al. 1982; Nawab et al. 2008; Marateb & McGill in press). In this application, the shape of the action potentials is used to identify the occurrences of the discharges of the active motor units. Although surface EMG signals can also be decomposed into individual motor unit activities (e.g. Gazzoni et al. 2004; Holobar & Zazula 2004), the analysis of motor units is usually more accurate using intramuscular systems owing to the narrower action potentials. Moreover, by contrast with surface techniques, intramuscular electromyography allows recordings from deep muscles.

The greater selectivity of intramuscular recordings compared with surface recordings also poses some limitations. Owing to its selectivity, intramuscular EMG recordings reflect the activity of only a small number of active motor units whose fibres are close to the position of the detection site. In addition, the identified intramuscular action potentials are not representative of all the fibres in the motor unit, and it is difficult to extract parameters related to the membrane fibre properties by using intramuscular recordings only. Finally, intramuscular recording systems usually have a very small detection volume and thus it is difficult to reposition a needle to detect the same motor units in repeated insertions. Surface electromyography provides a more global assessment of muscle properties and thus overcomes
some of these limitations. For example, muscle fibre conduction velocity can be estimated during voluntary contractions with surface electromyography (Farina & Merletti 2004), whereas it is usually assessed only with electrical stimulation of the muscle fibres with intramuscular systems (Troni et al. 1983).

This review describes the methods used for recording and processing intramuscular EMG signals.

2. Electrodes for EMG signal detection

Since 1929 when Adrian & Bronk (1929) proposed the concentric needle electrode, a number of systems for the detection of intramuscular EMG signals have been developed. The concentric needle electrode detects signals between the
tip of a wire insulated in the cannula and the cannula. Other needle electrodes have been proposed in more recent times (figure 1) and have been adapted to the specific algorithms for information extraction. For example, Buchthal et al. (1957) extended the concentric needle by placing 12 insulated wires in a slot in the cannula. With this system it was possible to record the motor unit electrical activity from many points in the muscle, thus analysing the decrease in signal amplitude with distance and measuring the size of the motor unit territory.

De Luca & Forrest (1972) described the quadripolar needle, consisting of four detecting surfaces that provide different representations of the same motor unit activity, thus adding information for the automatic decomposition of the EMG signal.

The size of needles used to record intramuscular EMG signals is larger than that of the nearby muscle fibres (a concentric needle, for example, has a diameter of 0.3–0.5 mm), thus the insertion of the needle damages some muscle fibres generating a small local oedema. In addition, in some cases, the insertion of needles may unintentionally damage important structures. For example, in a study on cadavers, Haig et al. (2003) showed that when a needle was inserted into a muscle according to standard clinical procedures, the tip was in the intended location in only 45 per cent of the cases and within 5 mm of important structures, such as nerves, arteries or veins, in 17 per cent of the cases.

The potential distribution in space, and therefore the signal features, is affected by the presence of the equipotential needle surface (Gootzen et al. 1989; Theeuwen et al. 1993; Stegeman et al. 1994). The recorded signal amplitude depends on whether the fibre is located at the same or the opposite side of the

Figure 2. (a) Single fibre action potentials do not sum synchronously to form motor unit action potentials, which thus present multiple phases (P) and turns (T) (reproduced from Stålberg & Falck 1997, with permission). (b) A motor unit action potential showing an abnormal ‘jiggle’. Eight traces are superimposed. Some of the components show an abnormal jitter, also resulting in vertical shifts. Two components (marked with B) display intermittent blocking. A late component, ‘satellite’ (S), is generated by a muscle fibre that is either innervated by a slowly conducting terminal axon or has a slower conduction velocity due to atrophy (reproduced from Trontelj et al. 2004, with permission © (2004) IEEE).
electrode’s leading-off surface(s). Simulations indicated that a general decrease in EMG signal amplitude (whose magnitude depends on the thickness of the fluid layer) should be expected. However, these changes do not seem to be relevant for diagnostic purposes (Ekstedt & Stålberg 1973; Gootzen et al. 1989; Brownell & Bromberg 2007).

Needle size, type and insertion technique determine the pain or discomfort of the patient during the insertion (Strommen & Daube 2001). Slight movements of the concentric needle not only cause pain but modify the action potential amplitude and area, which vary markedly when the position of the needle changes within the motor unit territory. The ratio of motor unit action potential area to amplitude is less affected by changes in electrode position and measures a ‘duration’ or ‘thickness’ of the waveform. This ratio is also less sensitive to the signal-to-noise ratio and inter-operator differences than other signal features (Nandedkar et al. 1988).

In the early 1960s, Basmajian proposed wire electrodes for the detection of intramuscular EMG signals (for review, see Basmajian & De Luca 1985). Wire electrodes may be made from small diameter, non-oxidizing wires with insulation. They are placed in the cannula of a needle and bent at the tip; the needle is inserted into the muscle and then removed, with the wires left in the muscle. The advantage with respect to needle electrodes is that the wires can be hardly felt when the needle is removed, thus allowing strong contractions without discomfort or pain for the subject. However, their position cannot be adjusted after removal of the needle, whereas a needle electrode can be moved in the muscle to search for a suitable position. Wire electrodes are usually preferred in studies in which the EMG signals are recorded over long periods or during movement, since they are more stable than needle electrodes.

Gydikov et al. (1986) proposed subcutaneous branched electrodes, which consisted of two isolated parallel wires connected together to prevent relative displacement between the wires. The isolation of one of the wires was removed at one point and the isolation of the other wire at two points spaced by 1–2 mm. The points of removal of the isolation formed the leading-off areas. The signals recorded by the first wire were subtracted to the signals detected by the second wire (the two uninsulated points of the second wire detect a signal that is the average of the potentials in the two points). These wires were inserted subcutaneously by means of a needle and were proven to detect EMG signals with good selectivity for single motor unit analysis (Gydikov et al. 1986).

The main limitation of wire electrodes is the difficulty in placing many detection sites with consistent site geometry into a single system. Thus, multi-channel recordings are usually performed by many wires inserted into the cannula of a single needle. Recently, indwelling systems made of thin-film technology have also been proposed for intramuscular recordings, following similar technological developments as for neural interfaces. Longitudinal intrafascicular electrodes (LIFEs) are fine wire electrodes designed to be implanted into peripheral nerves to serve as neural interfaces (Malagodi et al. 1989). Conventional LIFEs, usually handmade, are constructed using Teflon insulated Pt–Ir wires (Lefurge et al. 1991) or metallized and insulated polyaramid filaments (McNaughton & Horch 1996). The new generation of LIFE systems consists in microfabricated thin-film LIFE. Thin-film technology allows the design of multiple detection sites with consistent site geometry on an
extremely flexible substrate (Farina et al. 2008). These systems have the advantage of allowing multi-channel intramuscular electromyography with a single insertion and with a well-defined and reproducible spatial arrangement of the detection sites.

A few techniques have been proposed to solve the problem of the very local nature of the information extracted from intramuscular recordings. The macro EMG electrode (Stålberg 1980; Stålberg & Fawcett 1982; fig. 1D) consists in a modified single fibre EMG needle where a portion of the isolated cannula is exposed and acts as a macro electrode. The signal recorded from the macro electrode is averaged using single motor unit discharges as triggers, as extracted from the single fibre recording. Although the averaged action potentials recorded using a macro needle electrode (MA–EMG), should better represent the electrical activity of a motor unit, there is a relatively high correlation between features of the actions potentials detected from concentric needle electrodes (CN–EMG) and MA–EMG (Finsterera & Fuglsang-Frederiksen 2000).

Scanning electromyography is another technique based on triggered averaging. This method provides an electrophysiological cross section of a motor unit (Stålberg & Antoni 1980). Two intramuscular electrodes are used: a single fibre electrode for triggering and a concentric electrode for acquisition, inserted at least 2 cm away and in the direction of the fibres. The muscle is voluntarily activated and the concentric electrode is connected to a mechanical linear actuator, controlled by the acquisition system; the concentric electrode is pulled out in steps of 50 µm or multiples, so that a corridor of approximately 20 mm is investigated. The signal from the concentric electrode in each position is averaged using the single fibre action potentials as trigger and therefore outlines the thickness of the motor unit territory in each location and direction of pulling.

3. Intramuscular EMG signal decomposition

Indwelling EMG signal decomposition is the process of identification and classification of action potentials of individual motor units from the interference pattern (figure 3). This process is complex owing to the variability in shape of the action potentials generated by the same motor unit and the overlapping in time and frequency of action potentials generated by different motor units. The changes in shape of action potentials are usually limited for isometric constant force contractions of short duration. For longer recordings, however, it is usually necessary to update the templates by averaging the last $N$ identified action potentials to account for changes due to fatigue.

Pioneer work in EMG signal decomposition was accomplished by De Luca and co-workers (De Luca et al. 1982; LeFever & De Luca 1982; LeFever et al. 1982). In the last two decades, these and other researchers worked towards a robust solution to the problem of intramuscular EMG signal decomposition (McGill et al. 1985, 2005; Stashuk & De Luca 1989; Wellig et al. 1998; Stashuk & Qu 1998; Stashuk 2001; Katsis et al. 2007; Nawab et al. 2008; Marateb & McGill in press). Although the solutions vary greatly for the need of operator’s assistance, complexity and accuracy of the results, and computational time, some general considerations can be drawn. In general, it is not possible to accurately decompose EMG signals detected during very high-force contractions (more than

Phil. Trans. R. Soc. A (2009)
50% of the maximal force). Moreover, the number of identified motor units is limited to 3–8, with peaks of 11–12 with recent algorithms (Nawab et al. 2008) or multi-channel recordings (McGill et al. 2005). Despite this limitation, intramuscular EMG signal decomposition currently offers the possibility of accurate in vivo analysis of the neural drive to muscles and has allowed the construction of conceptual models of motor unit control strategies.

In classic approaches, the decomposition of a composite EMG signal requires two steps: (i) detecting action potentials (segmentation) and (ii) recognizing detected action potentials as members of a class (classification). For segmentation, the characteristics that differentiate the action potential shapes from the background noise should be explored. Various manifestation variables can be adopted for solving this detection problem, such as amplitude, signal derivatives, coefficients of a transformation (e.g. wavelet transform). To classify the detected action potentials, it is essential that the action potentials produced by the same motor unit are more similar in shape than action potentials generated by different motor units. The differences in shape should be quantified (feature space) in order to compare action potentials. Various features have been used for this purpose, including morphological statistics (e.g. peak-to-peak voltage, number of phases, duration, number of turns), Fourier transformation coefficients, coefficients obtained from other transformations (e.g. a wavelet basis), the time samples of the band-pass filtered signal and the time samples of the low-pass differentiated signal (for review, see Stashuk 2001).

In most decomposition methods, the first classification step is performed only on action potentials that do not overlap in time. The overlapping potentials, which form compound waveforms that do not repeat (therefore forming classes with single elements), should be decomposed into the basic single unit potentials. This operation is usually the most difficult in EMG signal decomposition and

Figure 3. Schematic representation of the detection and decomposition of intramuscular EMG signals. The quadrifilar needle proposed by De Luca (Basmajian & De Luca 1985) is indicated in the figure. Three raw interference signals are detected between electrode pairs (only one is indicated in the figure). The aim of the EMG decomposition algorithm is to identify the motor unit action potential trains and therefore the discharge times of the motor units which contribute to the interference signals. Such discharge times represent the control signals provided by the individual motor neurons.
algorithms often differ in their performance owing to their different sensitivity, accuracy and ability to resolve superimposed action potentials (Nawab et al. 2008; Marateb & McGill in press).

There are two basic strategies for resolving superimposed action potentials. The first is based on matching the action potentials identified in the classification phase, one at a time, with the superimposed waveform or a residual form of it until finding the ‘best’ match. Once the best match is found, a residual waveform is created, which is the difference between the initial waveform and the identified action potential. The procedure is repeated until the residual is below an energy threshold. This strategy is called peel-off or sequential approach. The second strategy synthesizes superimposed waveform models by adding up combinations of templates shifted in time by different amounts and searches for an optimal or acceptable match between a model-superimposed waveform and the actual combination of action potentials. This method is called modelling approach and, although it is theoretically capable of resolving all types of superimposed waveforms, it is computationally demanding. More details can be found in Sörnmo & Laguna (2005), LeFever et al. (LeFever & De Luca 1982; LeFever et al. 1982) and Stashuk et al. (Stashuk & Qu 1998; Stashuk et al. 2004).

The verification of the accuracy of an intramuscular EMG signal decomposition requires the availability of a ‘gold standard’ for which the decomposition result is known ‘a priori’ and the definition of quantitative indexes that allow comparison of performance (Farina et al. 2001a,b). The reference result can be obtained by manual decomposition of a number of experimental signals by expert operators, if it can be assumed that these decompositions are (almost) error free. Alternatively, it is possible to independently decompose two signals that comprise the activity of the same motor units but are detected from different locations in the muscle. The results of the two independent decompositions can then be compared. The probability of incorrectly decomposing the two signals and yet obtaining the same discharge pattern for a given motor unit is very small, thus when the two decomposition results agree the decomposition is considered correct. Finally, the reference signal for testing performance can be obtained from a mathematical model that includes all the relevant characteristics of experimental signals, either by generating action potentials or by combining experimental action potentials. Each of these approaches has limitations that currently cause difficulties in comparing different algorithms and thus in selecting a standard decomposition algorithm.

4. Spike-triggered averaging

Once the signal has been decomposed, all control properties of the identified motor units can be directly measured, such as recruitment and derecruitment thresholds, discharge rate and interspike interval variability. In addition, with the joint recording of other signals, such as force and surface EMG signals, contractile and fibre membrane properties can be measured.

The joint recording of intramuscular and surface EMG signals may be useful to investigate peripheral and control properties of the neuromuscular system.

The discharge times identified from the intramuscular EMG signals are used to trigger the averaging of the surface EMG signal and thus to estimate the surface representation of the single unit action potentials. From the surface
Action potentials, motor unit properties, such as muscle fibre conduction velocity, can be measured (Farina et al. 2002; figures 4 and 5). Surface recordings can also be used jointly with intramuscular recordings to enlarge the number of motor units that are concurrently investigated (Holobar & Zazula 2004). Spike-triggered averaging is also applied to the joint torque for estimating the torque contribution of individual units (Stein et al. 1972), although the approach has several limitations (Taylor et al. 2002).

5. Interference signal analysis

In clinical practice the precise decomposition of intramuscular EMG signals has still limited applications, owing to the difficulties in accurate electrode placement and proper interpretation of the results by the clinician. In the diagnosis of neuromuscular disorders, the shape of motor unit action potentials recorded with intramuscular electrodes is often used to differentiate pathological versus healthy conditions. In most cases, the signal is not fully decomposed but rather analysed as an interference signal and characterized by global descriptors, or only a few representative action potentials are extracted. The turns and amplitude analysis is an example of these methods (Fuglsang-Frederiksen 1981). A turn in the EMG signal is defined as a change in the slope sign, with some special features (e.g. the
maximum and minimum slope, the separation between turns, etc.) to distinguish a ‘true’ turn from noise. The portion of signal included between two subsequent turns is referred to as a segment and described by its duration and amplitude. Other parameters, such as the number of turns per second, mean segment duration and amplitude, spectral variables (e.g. mean or median frequency and spectral moments), and coefficients of wavelet expansions are also applied. Number and amplitude of turns as well as other variables are often reported in scatter plots versus time or force and compared with ‘typical’ plots of healthy or pathological subjects for diagnostic inference and classification of disorders.

This work was supported by grants provided by the European Community (CyberManS, contract no. 016712), the European Space Agency (MESM, contract no. 15097/01/NL/SH), the Italian Space Agency (OSMA), and the Bank Foundations Compagnia di San Paolo and Fondazione CRT.

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Phil. Trans. R. Soc. A (2009)


