Review: Selective Electrodes for Human Motoneuron Research

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Abstract

Research on human motoneurons is conducted through the analysis of single motor unit potential trains. Although human motoneurons have been investigated for years, their firing patterns at high force levels remain largely unexplored. Filling this gap requires selective electrodes, which enable the recording of distinguishable single motor unit potentials even at maximal effort. This review describes the characteristics of various electrodes used in electromyography with special attention paid to electrode selectivity. Those most suitable for this purpose are highlighted.

Keywords: Selective electrodes, Motor unit, Motoneuron, Spatial filters, Electromyography (EMG)

1. Introduction

In normal conditions, all individual fibers of a motor unit (MU) are activated by every discharge of its motoneuron (MN) [1,2]. Thus, the analysis of MU discharge patterns in voluntary contractions and their responses to nerve stimulation allows the investigation of MN firing characteristics, strategies adopted by the central nervous system to control muscle contraction, and their adaptation to conditions such as ageing, fatigue, exercise, pathology, or pain [3-13]. For such investigations to be reliable, it is important to get the full timing information on single MU potential (MUP) trains.

The essential problem in this analysis is the extraction of a single MUP train from the composite electromyographic signal (EMG). This problem is comparatively easy to solve for intramuscular EMG collected at low force levels, since such a signal contains potentials of a few MUs firing at low rates. With an increase in force level, both the number of active MUs and their firing rates increase which makes the recognition of single MUPs increasingly difficult. That is why many experienced scholars involved in human MN research perform experiments at low levels of muscle contraction and adjust the intramuscular electrode position to obtain records with clear potentials from only one or a few MUs (e.g., [14-17]). MN firing patterns at high force levels remain largely unexplored and the results of the scarce studies in this domain leave a number of unresolved questions. The major questions are listed below.

Maximum firing rates. In the earliest report, Marsden et al. [18] demonstrated several peculiar MUs from the adductor pollicis innervated by a median nerve, which after blockade of the ulnar nerve, could be recorded in the whole range of muscle forces. At maximal voluntary contraction (MVC), the firing rates of these MUs started from 60-100/s and decreased to 20/s in about 30 s. Other research reported similar behavior of MUs from the muscle extensor digitorum brevis [19]. Gydikov and Kosarov [20] and Piotrkiewicz et al. [21] found maximal firing rates at MVC in brachial biceps not exceeding 25/s. Other studies on the adductor pollicis and on the soleus reported MN firing rates at MVC not exceeding 35/s [3,22].

MU types. Gydikov and Kosarov [20] as well as Grimby and Hannerz [19] have distinguished two types of MU which differed by their fatigability and discharge properties. In the former study, these two distinct MU types were denoted as “tonic” and “phasic” according to the contemporary classification. In animal studies, the MUs were classified into three groups on the basis of their contraction times and fatigability (slow, fast fatigable, and fast fatigue-resistant) [23]. This classification was extended also to the human muscles, although no systematic studies on MU types according to their threshold are presently available. In modern human neurophysiology, the prevailing view is that the distribution of MU properties in each muscle is continuous. However, this view needs systematic verification across the entire range of force levels, including maximal ones.

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Dependency of firing rate on contraction force. In investigation of brachial biceps MUs, researchers found two types of firing-rate-force dependency [20]. The low-threshold MUs displayed rate-limiting whereas the high-threshold MUs linearly increased their firing rates until MVC. During weak contractions with a trapezoid force profile, the lower-threshold motor units fired always at greater rates than those of the higher-threshold units [24]. This behavior was later called the “onion skin” phenomenon. Recently, using selective intramuscular electrodes, researchers have shown that during contractions close to MVC, peak forces of higher-threshold MUs exceeded those of low-threshold MUs [22]. The validation of the “onion skin” scheme for a wide range of contraction force has been described in studies performed with a novel system for the acquisition and decomposition of EMG signals from 4-channel surface electrodes [26,30]. It should be noted, however, that the credibility of the results of the latter studies is hindered by the complete disappearance of the parallel fluctuations of individual MU firing rates, which were denoted “common drive” and extensively studied by the same group earlier [25,28,29,31]. The check for common drive is very useful in manual corrections of automatic decomposition, and every deflection of the common pattern indicates an error in recognition.

The discrepancies listed above may be related to differences in muscles investigated and in details of the experimental protocols, which have to be clarified in more systematic future investigations. These investigations require electrodes that are selective enough to distinguish single MUPs even at highest force levels. Therefore, this review is concentrated mainly on the solutions that result in the high selectivity of EMG electrodes suitable for single MN studies. Special attention is paid to surface electrodes because of their noninvasiveness.

Small surface electrodes in various configurations, enabling single MUP recording at high force levels, were introduced in Gydikov’s laboratory in the early 1970s [20,32,33]. These investigations, however, did not gain due attention until the late 1990s, when the fear of infection by needle electrodes resulted in growing pressure towards introduction of noninvasive techniques of EMG recording. In response to this challenge, a more sophisticated technique was developed, so-called high-spatial-resolution [34] or high-density [35] EMG. This technique is based on two-dimensional arrays of small surface electrodes, which increase the amount of useful information in EMG signal decomposition into single MUP trains.

Two-dimensionality introduced the added value of spatial information, which is helpful in studying MU morphology, dimensions, orientation and location within the muscle, as well as propagation of excitation along the muscle fiber and its conduction velocity. However, even with these techniques, full automatic decomposition of EMG signals was possible only at weak muscle contractions [36].

Research on human MNs is based on the analysis of MUP trains. The proper recognition of each single MUP in an EMG record is therefore crucial. The result does not depend as much on the algorithm used for decomposition, as on the selectivity of the electrodes [37]. In other words, if some information is lost in the stage of EMG acquisition, it cannot be restored even by the most sophisticated algorithm. The electrodes, which would enable expansion of human MN research to the entire range of muscle contraction forces, should be convenient to use and easy to manipulate in search of the proper position to record activity of 1-3 single MUs at any contraction strength. This review describes the characteristics of various electrodes and indicates which of them are most suitable for this purpose.

2. Selectivity of EMG recording

The question of electrode selectivity is closely coupled with the properties of muscle tissue, which can be described as a volume conductor. The adjective “volume” here emphasizes that current flow is three-dimensional, in contrast to the confined one-dimensional flow within insulated wires [38]. The potential generated by a muscle fiber spreads around, which results in attenuation of its amplitude and prolongation of its duration.

The EMG signal usually contains a mixture of potentials generated by several MUs. Signal characteristics depend on the kind of applied electrodes. In order to obtain a selective recording with the activity from a small number of muscle fibers, the amplitude ratio between close and remote muscle fiber potentials should be maximal [39].

The electrode selectivity is reflected by its uptake area, i.e., the semicircle on the muscle cross section, surrounding the leading-off surface(s). The smaller the uptake area, the fewer muscle fibers it contains and the more selective is the electrode.

There are three ways to increase the selectivity of intramuscular electrodes.

1. Decreasing the electrode leading-off surface (e.g., [40,41]). A small electrode is likely to have smaller uptake area, containing fewer muscle fibers of one MU than that a bigger electrode would have. An electrode surface with a diameter of 40 µm was found to be equivalent to that of a point-shaped electrode [42]. A further decrease of the recording surface has been considered ineffective due to the substantial increase in electrode impedance, which might introduce problems with input amplifier circuits without offering any practical recording advantage [43,44]. Nevertheless, most selective recordings were obtained with smaller surfaces, down to 10 µm [45,46].

2. Bipolar recording, which also restricts the electrode uptake area [39,43,44,47-50]. In the bipolar configuration, the activity from fibers approximately equidistant from both recording surfaces is cancelled. In order to obtain a selective recording with the bipolar technique, the inter-electrode distance must be considerably smaller than the effective uptake radius of each electrode surface. This electrode configuration additionally reduces the noise caused by remote sources such as electrical appliances or electrocardiogram signals.

3. High-pass filtering and/or differentiation of EMG signals. These two methods for increasing selectivity have essentially the same effect. The effect of high-pass filtering is based upon the fact that frequency content of muscle fiber
action potential depends on the electrode-fiber distance \[43,51,52\]. Thus, action potentials from distant fibers are suppressed and the difference in amplitude between the action potentials generated by close and remote fibers is enhanced. Similarly, differentiation enhances the amplitudes of the MUPs with the shortest rise times, i.e., those representing the highest frequencies \[53,54\]. These effects are more prominent when the frequency contents of the signals from close and remote fibers differ considerably, as is the case in recording with small electrodes. Hannerz \[46\] indicated that another effect obtained with this technique is shortening of the action potential duration, which facilitates signal decomposition.

These three methods, intended to increase the selectivity of recordings, are basically different. By using a small electrode surface and bipolar recording, activity from distant fibers is suppressed even before the amplification stage \[43\]. With high-pass filtering and/or differentiation, the obtained recording could be further treated to increase the amplitude ratio between action potentials of different frequency contents, i.e., those located at different distances from the electrode.

All of the above-mentioned methods for increasing selectivity are applicable also to surface electrodes \[55-57\], although they are not equally effective. Surface electrodes are separated from the muscle fibers not only by the much thicker layer of muscle tissue, but also by skin and fat layers, so the attenuation of the potential is much more pronounced than in the case of recordings from inside of the muscle. The number of muscle fibers from each MU contributing to surface MUP is much bigger, but the differences in electrode-fiber distances as well as the differences in frequency content between potentials from different MUs are much smaller. Consequently, the shapes of single MUPs picked up by surface electrodes are more stereotyped than those recorded by intramuscular ones and potentials from different MUs can be distinguished mostly by their amplitude. Due to the much longer electrode-fiber distances, the duration of a single MUP substantially increases, which evokes frequent phase cancellations and adds to the difficulties encountered during decomposition. Moreover, surface electrodes, even those of high selectivity, are not always able to suppress crosstalk, i.e., the electrical activity of distant muscles \[58\]. This review concentrates on electrode selectivity; readers interested in crosstalk may refer to several papers on this subject \[59-69\].

As mentioned above, electrode selectivity is reflected by its uptake area, which is measured in terms of the uptake radius, usually defined as the distance over which the potential amplitude decreases to a certain value. This value is given either as a percentage of maximum (10% \[39,70\] or 25% \[45\]) or as an absolute value (200 µm \[71\]). Unfortunately, the uptake radius values are rarely reported in the relevant literature. One of the reasons is presumably the dependency of selectivity not only on the electrode construction but also on the source properties and the mutual configuration of both. Among the reviewed literature, mean values of uptake radius were found for only two electrode types. More often, the researchers defined electrode selectivity in terms of the maximal effort at which single MUPs could be recorded.

Table 1 presents a summary of the characteristics of all electrodes described in this review with the comments on the selectivity provided, wherever possible.

### 3. Electrodes

The electrical activity associated with the contraction of muscle fibers in MUs can be recorded using either invasive (intramuscular) or noninvasive (surface) detection.

#### 3.1 Intramuscular electrodes

Intramuscular EMG signals are detected by the electrodes inserted into muscles. During weak muscle contractions, the signal contains distinguishable individual MUPs. The range of the measured potential difference for intramuscular electrodes is 0.1-20 mV. There are two types of intramuscular electrode: needle and wire electrodes.

##### 3.1.1 Needle electrodes for clinical diagnosis

The traditional use of intramuscular EMG signals is in clinical diagnosis through the analysis of changes in the signal characteristics related to disorders affecting MNs, muscles, neuromuscular junctions, or peripheral nerves. Basically, two levels of muscle contraction are analyzed: one corresponding to the weak contraction, which allows the analysis of single MUP shapes, and the other corresponding to strong contraction, resulting in an interference signal. The diagnosis relies on the features that differ in healthy conditions and diseases, i.e., statistics of the parameters characterizing single MUP shapes such as amplitude, duration, and number of phases on one hand, and turn-amplitude dependency, describing the interference pattern, on the other.

It is obvious that the electrodes applied in clinical diagnosis should be standardized to avoid ambiguities related to different electrode characteristics. Generally, these electrodes have an insulated wire(s) mounted firmly inside the metallic cannula.

The monopolar needle electrode in Fig. 1(a) has to be used with a surface reference disk electrode applied close to the needle. This configuration may be simplified if the needle cannula is taken as the reference (concentric needle electrode). It has been shown that both configurations are basically equivalent \[72\] from the point of view of clinical practice. The main difference concerns electrode uptake area, which is about two times larger for the monopolar electrode.

Nowadays, the most common needle electrode used in clinical EMG diagnosis is the concentric needle electrode with one detection surface formed by the beveled cross section of a centrally located wire typically 200 µm in diameter, as shown in Fig. 1(b). The tip of the wire is bare and acts as a detection surface. The potentials are picked up between the detection surface and the cannula. The radius of the uptake area of the standard concentric needle electrode is around 500 µm \[73\]. Such an area usually contains 1-3 muscle fibers belonging to the given MU.
Table 1. The summary of basic electrode properties.

<table>
<thead>
<tr>
<th>Electrode</th>
<th>Electrode parameters</th>
<th>Selectivity</th>
<th>References</th>
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<tr>
<td><strong>Intramuscular electrodes</strong></td>
<td></td>
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<tr>
<td>for clinical diagnosis</td>
<td></td>
<td></td>
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<tr>
<td>area of single recording surface [mm²]</td>
<td></td>
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<td></td>
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<tr>
<td>Needle electrodes</td>
<td></td>
<td></td>
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<tr>
<td>monopolar</td>
<td>0.12</td>
<td>about 750 µm</td>
<td>[72]</td>
</tr>
<tr>
<td>concentric needle</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>standard (monopolar)</td>
<td>0.07</td>
<td>500 µm</td>
<td>[6,72,75]</td>
</tr>
<tr>
<td>ocular (bipolar)</td>
<td>0.015</td>
<td>&lt; 30%</td>
<td>[6,7,74,75]</td>
</tr>
<tr>
<td>single fiber</td>
<td>0.002</td>
<td>300 µm</td>
<td>[42,76,77]</td>
</tr>
<tr>
<td>macro</td>
<td>1.5 mm²</td>
<td></td>
<td>[78-80]</td>
</tr>
<tr>
<td>Wire electrodes</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>bipolar</td>
<td>25</td>
<td>100%</td>
<td>[46]</td>
</tr>
<tr>
<td>1-2 mm of insulation stripped off the tips</td>
<td>comparable to standard concentric electrode</td>
<td>[47]</td>
<td></td>
</tr>
<tr>
<td>50</td>
<td>100%</td>
<td>[95]</td>
<td></td>
</tr>
<tr>
<td>70</td>
<td>weak contractions</td>
<td>[14,93,94]</td>
<td></td>
</tr>
<tr>
<td>90</td>
<td>100%</td>
<td>[10,16,98]</td>
<td></td>
</tr>
<tr>
<td>Quadrilateral</td>
<td>25</td>
<td>100%</td>
<td>[25,29,31,99]</td>
</tr>
<tr>
<td>50</td>
<td>100%</td>
<td>[100-104]</td>
<td></td>
</tr>
<tr>
<td>Branched</td>
<td>100%</td>
<td>[13,22,94,107,110,111]</td>
<td></td>
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<tr>
<td><strong>Surface electrodes</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>DSE [mm]</td>
<td>0.5</td>
<td>5</td>
<td>[34,35,117-120,131,155]</td>
</tr>
<tr>
<td>IED [mm]</td>
<td>1-10</td>
<td>5 &amp; 10</td>
<td>Better than inverse rectangle filter</td>
</tr>
<tr>
<td>Selective</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Single differential</td>
<td>0.5</td>
<td>1.0</td>
<td>6.5</td>
</tr>
<tr>
<td>double differential</td>
<td>0.5</td>
<td>1.0</td>
<td>5 &amp; 10</td>
</tr>
<tr>
<td>Normal double differential</td>
<td>0.5</td>
<td>1.0</td>
<td>6.5</td>
</tr>
<tr>
<td>2x double differential</td>
<td>0.5</td>
<td>1.0</td>
<td>6.5</td>
</tr>
<tr>
<td>Inverse rectangle</td>
<td>0.5</td>
<td>1.0</td>
<td>6.5</td>
</tr>
</tbody>
</table>

1. Selectivity is given either as the radius of uptake area [µm] or as the maximum contraction level for single MU recording [% of MVC].
2. The radius of uptake area was estimated as the radius of MU territory [2], since macro electrode is supposed to record activity of all fibers belonging to this MU.
3. Diameter of single electrode.
4. Inter electrode distance.
5. Central pole diameter.
6. Ring width.
7. Ring radius.

For the recording from ocular muscles, smaller concentric electrodes are used, with the recording surface created by a wire typically 100 µm in diameter. Such an electrode is presented in bipolar configuration with two wires exposed in cross section in Fig. 1(c). Here, the potentials are picked up between two detection surfaces, and the cannula is used as a shield. As explained above, the bipolar configuration is more selective and thus more suitable for recording from small muscles. Because of its selectivity, this electrode is often used in MN research (e.g., [6,7,74,75]).

Among the electrodes used in clinical diagnosis, the most selective is the single-fiber electrode developed by Ekstedt and Stålberg [76]. The detection surface is formed by a 25-µm diameter wire, usually made of tungsten, protruding perpendicularly from the cannula wall, as shown in Fig. 1(d). The mean uptake area is around 300 µm [42]. This electrode is used to detect the activity of individual muscle fibers.

With the invention of this electrode, a new branch of clinical EMG emerged. Single-fiber EMG, developed by Stålberg [77], has proven to be especially useful for...
examinations of denervated and reinnervated muscles, where neuromuscular jitter and fiber density measurements provide additional diagnostic information that is not readily available with concentric electrodes.

The major disadvantage of the single-fiber electrode, which hinders its application in MN research, is its recording instability: even minimal displacement of the electrode dramatically changes the amplitude of the recorded fiber potential, so long recordings from one MU are practically impossible.

Stålberg also developed the macro electrode [78] as the inverse of the single-fiber electrode (Fig. 1(c)). The single-fiber electrode is used as a trigger, and the cannula of the needle as a detection surface. The potential difference is measured between the cannula surface and the reference electrode, which is placed remotely. To obtain proper MU potential, macro EMG is averaged. The averaging is triggered by the single-fiber potentials.

Macro-MUP reflects activity generated from all muscle fibers within a single MU rather than from a small number of fibers as in conventional needle EMG [78-84]. It is analyzed for maximal peak-to-peak amplitude and for the area under the signal between the 10th and 70th ms of an 80-ms trace. These two values reflect the total electrical size of a MU, comprising the number and size of its muscle fibers. Since the macro electrode contains a single-fiber electrode, it enables also the analysis of fiber density and neuromuscular jitter [85]. With further assumptions, it is also possible to estimate the number of MUs from macro-MUPs [86]. However, despite all its advantages, this electrode at present is not widely used in routine EMG diagnosis [85].

3.1.2 Needle multielectrodes

Two special, custom-made needle multielectrodes were used in classical experimental studies of volume conduction in muscle tissue. These multifilar electrodes contain several thin wires in one needle, whose ends are bent at an angle of 90° and arranged in one or more rows in a window made in the cannula side (Fig. 2).

Buchthal et al. constructed a multifilar electrode containing 12 insulated wires with diameters of 50 or 100 μm embedded in a window in the cannula with a 1-mm diameter. Their leading-off surfaces were distributed over a length of 2.5 mm in one row. This multifilar electrode was used for exploring MU territory (i.e., the area on the muscle cross-section occupied by muscle fibers belonging to the given MU) and measuring the voltage change as a function of electrode-source distance in muscle tissue [87, 88].

Ekstedt and Stålberg constructed multielectrodes for investigations of single fiber potential propagation [89-91]. Fourteen isolated platinum wires with a diameter of 25 μm were introduced into a 0.6-mm hypodermic needle. Two millimeters from the point opposite to the bevel surface, they were bent into a window made in the cannula. The wires were arranged into two rows with 13 electrodes in one row and 1 electrode in the other, and fixed using epoxy resin. The distance between electrode centers was 30 μm, and that between rows was 240 μm. A few other variants of multielectrode construction were also used.

These electrodes were invented and applied exclusively for the studies mentioned above. Although very selective, they are not suitable for MN studies and therefore are not included in Table 1.

3.1.3 Wire electrodes for research in motor control

Needle electrodes have two main advantages. One is that their relatively small uptake area enables the electrodes to detect individual MUPs during low force contractions. The other is that they may be conveniently repositioned within the muscle after insertion, so that the signal quality may be improved and new tissue territories may be explored. In contrast, repositioning of wire electrodes is problematic, but they have much smaller uptake areas and their drift within the contracting muscle is minimized.

The most popular wire electrodes are as a principle custom-made in research laboratories from small-diameter insulated wires according to the procedure described by Basmajian and Stecko [47]. Two perpendicularly cut fragments of wire are introduced into a hypodermic needle, and their ends are hooked (Fig. 3(a)). The needle is inserted into a muscle and then withdrawn, whereupon the hooked ends fix the wires in the muscle. The recording surfaces are determined by the diameter of the wire; those most commonly used are between 25-90 μm [4,10,14,16,92-98].

Hannnerz obtained an even smaller recording surface of a wire electrode by making a small hole (diameter: 10-20 μm) in the insulation coat by means of a spark. With such a small surface, the shape of single fiber potentials is liable to...
distortion but selectivity is high (due to high-pass filtering). This high-impedance electrode allowed recording of chosen MUPs up to 100% MVC [45,46].

The NeuroMuscular Research Center at Boston University developed the quadrifilar needle electrode [99]. The electrode has four 25-µm wires fixed in a 25-gauge disposable needle by epoxy cement (Fig. 3(b)). The recording surfaces are located on the corners of a square, 50 µm apart (center to center). This arrangement allows a choice of four monopolar electrodes and/or six bipolar electrodes. The selectivity of this electrode is enhanced by use of multiple channel recordings, since not only the differences in MUP amplitudes but also different combinations of their single-channel representations can be taken into account during EMG signal decomposition. This custom-made electrode, as with commercial needle electrodes, allows easily repositioning, but it does not prevent electrode drift during repeated muscle contractions.

More recently, a new method for quadrifilar electrode construction was proposed by Ahmed and Roark [100]. Instead of fixing the four wire surfaces within the needle, the authors twisted four insulated wires with a diameter of 50 µm and glued them together with a droplet of surgical-quality adhesive. This allowed them to form a hook, securing the electrode position in the muscle, like with other wire electrodes.

Various quadrifilar fine-wire electrodes have been applied by other researchers [101-104].

One of the most selective types of wire electrode is branched electrodes, developed by Gydikov’s group from the Institute of Biophysics, Bulgarian Academy of Sciences [70,105]. The subcutaneous branched electrode consists of two wires containing a set of three leading-off surfaces, two on one wire (separated by 2-3 mm) and one directly across from the midpoint between these two on the other wire (Fig. 3(c)). Each leading-off surface is created by circumferential stripping of the insulation on the length of approximately 0.25-0.50 mm. The relative positions of the leading-off surfaces are secured by knots or glue [106]. The electrode is inserted into the hypodermic needle and positioned subcutaneously over the belly of the muscle, perpendicularly to the direction of the muscle fibers. By adjusting the position of the leading-off surfaces relative to the muscle, the recording site of the highest selectivity can be found. The subcutaneous location of the electrode allows reasonable stability. With this type of electrode, investigations of single MU activity during strong muscle contractions were conducted [22,107-111].

3.2 Surface electrodes

3.2.1 Conventional surface electrodes

Conventional surface electrodes enable noninvasive EMG recording of the global muscle electrical activity. The range of potential difference measured on the muscle surface is 0.05-1 mV. These electrodes are typically used in bipolar configuration (Fig. 4) and consist of conductive (usually metal, e.g., silver-silver chloride) detection surfaces with a diameter of less than 10 mm that sense the current on the skin through its skin-electrode interface.

Surface electrodes are preferred over needle electrodes when the assessment of the global muscle activity is required (e.g., during gait analysis, biofeedback therapy, fatigue studies, prostheses control [59,112-114]). They are also used in cases in which the insertion of the needle electrodes is difficult or not possible, e.g., for examination of children, long-term recordings, ergonomics, sports, or space medicine.

However, conventional surface electrodes are much less selective than intramuscular electrodes and record the activity of a large proportion of muscle fibers. Therefore, the interference signal obtained by this type of electrode is not suitable for studying firing characteristics of single MNs (Fig. 5).

Figure 4. Surface electrode: (a) transverse section and (b) recording surfaces.

Figure 5. Comparison of signals picked up by (a) wire and (b) surface electrodes. Both signals were recorded from the soleus muscle during weak muscle contraction and sampled at 20 and 2 kHz, respectively. More details on experimental methods can be found elsewhere [16].

3.2.2 Selective surface electrodes

As mentioned in Section 2, selective recording of muscle activity from the skin surface is difficult, since the differences in distances to electrode and in frequency content of the signal for different MUs are much less pronounced in this case. However, despite these difficulties, the first successful attempts at selective recording from single MU potentials by small surface electrodes were performed in Gydikov’s laboratory as far back as at the early 1970s [32,33,115]. The same research group later developed more selective, branched electrodes, whose subcutaneous version was described above. The surface version (see Fig. 6(b)) is not as selective as the subcutaneous one, but it still enables the recording of MU activity at high force levels [107,108,116].
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The further evolution of selective surface recording of MUPs resulted in the invention of high-spatial-resolution [34,117], high-density [35,118,119], and multichannel [120,121] surface EMG systems. In these systems, two-dimensional small electrode arrays are placed on the muscle surface, which increases the amount of information and enhances the capacity to discriminate the potentials of separate MUs [122].

The introduction of electrode arrays enabled several new studies of MU characteristics such as dimensions, location, and orientation [123,124], spatial recruitment [125,126], conduction velocity [127,128] and its changes during fatiguing contractions [129], spatial selectivity of transcranial magnetic stimulation [130], and the excitation spread of the innervation pattern [131]. Multichannel EMG is also considered as a useful solution for bioelectric prosthesis control [132,133]. The differences in location of single MUs with respect to the multielectrode and the information about the delay of the given MUP resulting from its propagation along the muscle fibers create so-called MU signatures [134] or fingerprints, which are utilized in the algorithms of surface EMG decomposition. This technique also enables so-called MU tracking [135]. Such a procedure is supposed to make possible longitudinal studies of morphological changes in chosen diseased MUs.

High-density surface EMG has been shown to be useful in several novel clinical studies [8,12,136-138]. However, it has still not been introduced into routine clinical examinations, [139] although the usefulness of this technique has been confirmed [34,120,140,141]. It was also applied to study muscle fasciculations [11,142]. However, Jahanniri-Nezhad et al. [143] have recently shown that their high selectivity may be also a drawback, since a large number of fasciculations remain undetected.

The selectivity of small surface electrodes can be further improved by application of spatial filtering [34,37,58,117,144-150]. Spatial filtering is in principle based on the weighted summation of signals detected by several electrodes. The simplest example of a spatial (single differential) filter is the bipolar electrode (Fig. 6(a)). A double differential filter contains three channels [123,145]. This configuration is virtually identical to that of the surface version of the branched electrode (Fig. 6(b)), where external channels are short-circuited [109].

More effective are two-dimensional configurations, i.e., Laplacian filters: the normal double differential filter (Fig. 6(c)), consisting of 5 channels, two-dimensional double differential filter (6 channels, Fig. 6(d)), and inverse rectangle filter (9 channels, Fig. 6(e)). The performance of different combinations of spatial filters was analyzed in experimental and theoretical studies [37,58,146,149]. Farina et al. [37] showed that the normal double differential filter increases the percentage of discriminated MUPs from 22% to 58%, compared to that obtained using a single monopolar electrode, and with the combination of three such filters (9 electrodes), this percentage further increases to 90%. Dimitrov et al. [58] found the two-dimensional double differential filter to be the most effective. It is important to note, however, that selectivity is not entirely an intrinsic property of the electrode, but depends also on the properties of the electrical source and on electrode-source configuration [42,58]. Therefore the conclusions of theoretical and experimental studies are not unanimous.

The bull’s-eye or concentric ring electrode (Fig. 6(f)) may also be considered as a special version of the branched electrode. This electrode system, proposed by Bhullar [151], has been shown by Farina [152] to be more selective than a number of one- and two-dimensional spatial filters. Higher spatial selectivity is obtained by increasing the number of rings reflected in the smaller duration of the MUAPs and better separation of the sources. New spatial filters for surface EMG detection based on multiple concentric ring electrodes seem to be well suited for the investigation of single MN firing characteristics. Their application is not particularly more complex than that of traditional systems such as single- or double-differential recordings.

4. Concluding remarks

A selective electrode should enable good recognition of individual MUPS in the interference EMG pattern [153]. If this ability is taken as the definition of selectivity, it becomes obvious that it is too subjective to be quantified. This is reflected in the data collected in Table 1, which contains few numbers that characterize the radius of the uptake area. Moreover, it should be stressed that these quantitative data are only rough estimates of the real uptake radius, which depends not only on the electrode construction, but also on the electrode placement and the properties of muscle fiber ensembles generating the MUPS. Other experimental conditions, such as researcher experience, also influence the quality of the signal picked up by a given electrode. Therefore, it is difficult to choose one electrode that is most suitable for recording single MN activity at high levels of muscle contractions. Nevertheless, after thorough analysis of the results of all studies presented above, a few solutions that seem to be the most promising in this respect can be highlighted. In the category of intramuscular electrodes, we would suggest Hannerz [46] or branched
electrodes [106,109], which are comparatively easy to make and manipulate. In the category of noninvasive surface electrodes, the electrode arrays with 3-9 poles and Laplacian filtering as well as the concentric ring electrode appear the most promising [151,152]. The final choice belongs to the prospective users, who should experimentally test the existing options and decide which solution is most suitable for their needs. We hope that our review will be helpful in this choice.

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References

Detecting, 86-

dition potential compared with in situ:
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Action potentials recorded intra-

or muscle fibre in Ringer’s-

[57x129]

[58]

[57x194]

[56]

[57x267]

[54]

[57x368]

[51]

[57x396]

[50]

[57x488]

[46]

[57x580]

[43]

[57x607]

[42]

[57x690]

[39]


