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DETECTION VOLUME OF SIMULATED ELECTRODE SYSTEMS FOR RECORDING SPHINCTER MUSCLE ELECTROMYOGRAM

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Keywords: electromyography, EMG modeling, spatial filters, selectivity, end-of-fiber components

Running title: Detection volume of EMG electrodes in a sphincter muscle model

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ABSTRACT

Background. Recording surface electromyogram (EMG) signals from sphincter muscles has become of increasing interest due to potential applications in diagnosis or investigation of the mechanisms of incontinence. Recently developed probes allow high-resolution detection of EMG signal from the external anal sphincter. One of the main issues in the interpretation of experimental signals is the detection volume of the recording electrodes.

Methods. An analytical model of generation of EMG from the anal sphincter was applied to simulate single fiber action potentials varying the fiber length, conductivity and thickness of the mucosa, shape of the electrodes, and detection system (monopolar, bipolar, double differential).

Results. The new probe is more selective than classical probes with big electrodes and large interelectrode distance. The decay of the recorded surface EMG action potentials with distance of the fiber was faster for increasing depth than axial distance (this is due to the geometry of the volume conductor, as indicated by comparison with simulations with a planar volume conductor). Point electrodes led to smaller detection volumes than rectangular electrodes, and the double differential system was the most selective recording configuration. The detection volume increased for increasing thickness and conductivity of the mucosa.

Conclusions. The detection volume of EMG recording systems from sphincter muscles extends more in the axial than in the depth direction and is affected by many parameters that cannot be estimated in practical situations, thus introducing a rather large variability in the muscle portion investigated among subjects. End-of-fiber components are enhanced by the circular geometry of sphincter muscles with respect to the case of a planar volume conductor. More selective information can be obtained using the recently developed multi-

channel probe with respect to classical probes. Selectivity is increased with small electrodes and double differential recordings.

INTRODUCTION

Surface electromyographic (EMG) recordings have poorer spatial selectivity than intramuscular recordings. The tissues interposed between the signal sources and the detection electrodes act as low-pass filters, causing a blurring effect on the surface potentials. As a consequence, a relatively large number of sources contribute to the signal.

The selectivity of surface EMG recordings depends on many factors that vary among subjects, muscles, and recording conditions. Subcutaneous tissue thickness [1], skin conductivity [2], fiber length [3], spatial filter applied to detect the signal [4], and interelectrode distance are some of the relevant factors that influence selectivity. Prediction of selectivity of a surface EMG system can be done by varying these parameters with the use of models of EMG generation (e.g., [2][4]).

Simulation (e.g., [4]) and experimental studies (e.g., [5]) have proven that spatial filtering may increase selectivity and thus reduce the detection volume, i.e., the portion of the muscle from which fibers significantly contribute to the detected signal. Increased selectivity may be useful for reducing crosstalk, i.e., the EMG contribution of active muscles other than the muscle under study [6]. Less selective systems however may better represent the global muscle activity.

All previous simulation studies on selectivity of EMG recordings are based on volume conductors that may well characterize cylindrical (limb) or planar (back) muscles. However, recent applications of surface EMG include muscles with more complex anatomy. For example, recording of surface EMG from sphincter muscles has become of increasing interest due to potential applications in diagnosis or investigation of the mechanisms of

incontinence [7]. Multi-channel EMG systems have been recently proposed for high-resolution detection of EMG from the external anal sphincter [8].

Theoretical prediction of the detection volume of EMG recordings from sphincter muscles is not possible due to the complexity of signal generation, anatomy, and the difficulty in predicting the relative weight of end-of-fiber components, i.e. the signal components originating when the intramuscular action potentials extinguish at the fiber endings. On the other hand, there are currently no studies that investigated by modeling the muscle portion from which sphincter muscle fibers contribute to the surface EMG. This information is relevant in the application of multi-channel EMG [8] in sphincter muscles.

This study applies a model of generation of EMG signals from a cylindrical volume conductor that represents in a simplified way the sphincter anatomy and compares the detection volumes of recording systems that can be implemented in practical applications.

METHODS

Simulation model

The generation model is based on the analytical solution for a cylindrical volume conductor, electrostatics problem [9], i.e. modeling the physiological tissues with the equations of electrostatics (hypothesis of quasi-static sources). In this study, the probe conductivity was neglected. The mucosa was infinite in the longitudinal direction, and placed between the probe and the muscle layer. The muscle was infinite in both the longitudinal and the radial direction (for the radial direction, the muscle layer extended from the muscle/mucosa interface to infinity; Figure 1). The muscle layer accounts for both internal and external anal sphincter (IAS and EAS, respectively), assuming for simplicity that the conductivity is the same for these two muscles.

The model accounts for the generation, propagation, and extinction of the intracellular action potentials at the end-plate, along the fiber, and at the fiber endings, respectively. These phenomena are described by the progressive appearing, propagation, and disappearing of the first derivative of the intracellular action potential [9]. The intracellular action potential shape was described as proposed by Rosenfalck [10]. Single fiber action potentials were simulated. The conductivity of the mucosa varied in the range 0.5 - 1 S/m and the conductivity of the muscle was 0.5 S/m in the fiber direction and 0.1 S/m in the directions orthogonal to the fiber. The thickness of the mucosa layer varied in the range 1 - 2 mm [7][11]. The thickness and conductivity of the mucosa depends on the specific anatomy of the subject and on pathological conditions [11]. The thickness of the IAS is of the order of 1-2 mm [7]. This muscle was considered to have the same conductivity as EAS, but no active fibers (this smooth muscle contributes with surface action potentials of long duration, which have bandwidth outside the range of frequencies considered). The muscle fiber end-plates were in the center of the fibers. Two fiber lengths were considered (Figure 1).

Figure 1 about here

In order to assess the effect of the geometry of the (cylindrical) volume conductor on the simulated surface EMG signals, some simulations were also performed using a plane layer model, described in [12]. Two layers were simulated, with the same conductivities as the mucosa and muscle layers of the sphincter model, respectively. Thickness of mucosa was also the same for the two models and muscle layer was semi-infinite. The simulated fibers were located at the same depth and transversal distance from the recording electrodes. The simulated fibers had constant length for the planar model (differently from the sphincter model where the angles are preserved). Innervation zone and end-of-fiber had the same

orthogonal projection on the mucosa surface. Considering the potential recorded at the detection point, the plane layer model can be considered as a first order approximation of the cylindrical model (in which the geometry is approximated, but all other physical parameters are the same).

Recording systems

The simulated recording systems were monopolar, bipolar and double differential filters in the angular direction. The probe had a radius of 7 mm with 16 electrodes equi-spaced along its circumference [8], see Figure 1d. Both point and rectangular electrodes (with size 1 mm in the angular direction and 10 mm in the longitudinal direction) were considered. The detection volume was assessed with respect to a specific recording location (Figure 1), which was at the middle angle between the innervation zone and one fiber ending.

In order to compare the selective probe used for research studies (shown in Figure 1d) with surface EMG probes used in the clinical practice, some simulations were performed with larger electrodes [13] (rectangular, 1.5 mm in the angular direction and 20 mm in the axial direction) and larger interelectrode distance (two electrodes, one on the opposite side of the probe with respect to the other), [14]. In the following, we refer to such a probe as the "classical probe". Monopolar and bipolar systems were simulated. In order to allow a simple comparison between the results of the two recording systems, the same geometry of the probe was used for both (even though different geometries are possible; for example a sponge can be used as a support for the electrodes, [14]).

Signals were simulated as detected at axial distances (with respect to the detection point) in the range 0 - 10 mm (1 mm increment) with fibers placed 1 - 10 mm (1 mm increment) deep into the muscle. The most superficial fiber considered was at 1 mm depth.

The detection volume was assessed by computing the average rectified value (ARV) of the simulated signal relative to all the simulated fibers and expressed as a percentage of the maximum, which corresponds to the fiber closest to the detection point. The following measures were defined:

$$dd = \sqrt{\frac{\int y_0^2 ARV(y_0, z_0) dy_0 dz_0}{\int ARV(y_0, z_0) dy_0 dz_0}} \qquad ad = \sqrt{\frac{\int z_0^2 ARV(y_0, z_0) dy_0 dz_0}{\int ARV(y_0, z_0) dy_0 dz_0}}$$

where y_0 and z_0 are the depth and axial distance respectively (see Figure 1). The indexes *dd* and *ad* are the standard deviations of ARV normalized in order to have integral 1 in the direction of depth and axial distance, respectively. *dd* will be referred to as the decay distance in the depth direction and *ad* as the decay distance in the axial direction.

The simulated single fiber action potential was divided into propagating and non-propagating signal components. The propagating component corresponded to the generation and propagation of the intracellular action potential, the non-propagating component corresponded to the interval of time during which the intracellular action potential extinguished at the fiber endings [15].

Figure 2 about here

RESULTS

Figure 2 shows representative single fiber action potentials (cylindrical volume conductor, simulated probe resembling that in Figure 1d) for one simulated anatomy varying transverse distance and depth of the fiber. Signals are shown as detected by monopolar, bipolar, and double differential systems. The signal amplitude decreases by increasing distance from the detection point. The rate of decrease is related to filter selectivity and electrode shape. Fibers with different depths have also different lengths, as the angles are

preserved (Figure 1). This determines different temporal supports of the simulated signals, with support increasing with distance.

Figure 3 shows the level curve plots for ARV as a function of depth and transverse distance from the fiber. Both the selective probe in Figure 1d and the "classical probe" (see Methods section) were simulated. In the case of the selective probe, the double differential filter was the most selective system in all cases and selectivity improved with bipolar with respect to monopolar recording. The detection volume was larger when simulating short with respect to long fibers. This was due to the larger end-of-fiber components generated by short fibers. End-of-fiber components are far field potentials that decay slower than propagating components. The rectangular electrodes were less selective than point electrodes for the axial direction (Table 1). The detection volume of the "classical probe" is much larger than that of the selective probe. As shown in Figure 3c and 3f, the decay in depth and (mainly) in axial directions are much less in the case of "classical probe" with respect to the selective probe.

Figure 3 and Table 1 about here

Figure 4 shows the ratio between the peak-to-peak amplitude of the non-propagating signal over that of the propagating component, in the case of selective probe. The anatomy and electrode shape did not affect consistently the selectivity to non-propagating components. For short fibers the ratio was ~3 times larger than for long fibers.

Figure 4 about here

Figure 5 shows a comparison between cylindrical and planar model for a representative anatomy. Detection volume of different spatial filters is studied. The detection

volume is larger for the planar model with respect to the sphincter mode. Furthermore, comparing the decay in depth and axial direction, we observe that the second one is lower. This finding is more evident in the case of the model of sphincter muscle with respect to the plane layer model. Selectivity to non-propagating components is also considered. Non-propagating components have larger relative amplitude (with respect to propagating component) in the case of the model of sphincter muscle with respect to the plane layer volume conductor.

Figure 5 about here

DISCUSSION

The detection volume of three widely used recording systems for EMG detection was investigated in simulated EMG signals generated from a sphincter muscle. The model allowed the assessment of the sensitivity of the detection volume to anatomical and physical parameters.

Simulated detection model

The estimate of the volume that contains the muscle fibers that mainly contribute to the surface EMG is very complicated experimentally. It implies concomitant knowledge on fiber position into the muscle and the respective surface detected action potentials. The use of mathematical models of EMG generation provides an alternative approach for the characterization of the detection volume.

The use of models has provided the basis for theoretical relationships between detection volume and detection and generation system parameters, e.g., effect of interelectrode distance and electrode size [16]. However, most of the knowledge is related to

limb muscles and based on models with planar geometry or of infinite extent [2][3][4][9][16].

Detection of surface EMG from sphincter muscles is classically performed with bipolar detection systems with large interelectrode distance [13][14]. Recently, circular arrays of many electrodes have been developed for better spatial resolution in sphincter EMG [8]. The improvement in selectivity of this new selective probe with respect to a probe with large electrodes and big interelectrode distance (resembling those used in clinical studies) is assessed in Figure 3.

The model used in this study has been previously proposed for the simulation of signals detected from sphincter muscles [9]. The volume conductor comprises an internal cylindrical layer, carrying the electrodes on its surface, surrounded by mucosa and muscle layers. This model is the only one currently available that allows the description of this geometry and no previous studies addressed the problem of assessing the EMG detection volume from sphincter muscles.

Factors that affect the detection volume

The detection volume extended more in the axial than in the depth direction, both in the case of point or rectangular electrodes. This is true also for a plane layer volume conductor, but with lower difference between the decay of the potential in the two directions (in the case of point electrodes the detection volume is almost equally extended in depth and transversal directions for the planar model, not shown results). The different behavior of the two models is due to the different geometries of the volume conductors. The thickness of the tissue interposed between the EAS and the electrodes comprises the IAS and the mucosa. Considering that the thickness of the IAS is approximately 1-2 mm and that of the EAS is 3

mm [7], the surface EMG recording provides information on almost all muscle fibers in the depth direction.

The detection volume of a single channel placed between the innervation zone and a fiber ending depended on the fiber length, indicating an effect of end-of-fiber components on the far-field potentials, as previously observed for limb muscles [3]. Attenuation of end-of-fiber components with distance is smaller than attenuation of propagating components [3], as also observed in this study on sphincter muscles (Figure 4). However, it is worth noticing that the end-of-fiber component is the same (up to a rotation) when detected by all channels in an array of electrodes in a probe. End-of-fiber components have larger amplitude (with respect to the amplitude of the propagating component) in the case of the model of sphincter muscle with respect to the plane layer model. A large amount of non propagating components is observed in experimental signals detected from EAS by the selective probe in Figure 1d with respect to those present in signals from muscles with parallel fibers. Our simulations suggest that end-of-fiber components can be an important contribution to non propagating components in signals from EAS (together with other possible contributions, like the crosstalk from other muscles, or the complex interference of different propagating potentials corresponding to motor units innervated in different locations).

In the case of the selective probe, the double differential filter resulted more selective than the bipolar system and the bipolar system was more selective than the monopolar recording in all conditions. This result was expected from theoretical results on selectivity of the detection systems [5] and modeling analyses in other muscles [4].

Implications

The results presented have practical implications in the design of recording systems for sphincter muscle EMG detection and for the interpretation of signals. The results on

selectivity suggest that bipolar recording is more selective than monopolar and that double differential provides a further improvement. Thus, for enhancing selectivity, the bipolar or double differential recording should be preferred over the monopolar. However, reducing electrode size is also an effective way to obtain selective recordings and may have higher impact on selectivity than spatial filtering. For example, monopolar detection with a point electrode is more selective in the axial direction than both bipolar and double differential recordings with the simulated rectangular electrodes.

The detection volume extends more in the axial than in the depth direction, thus the muscle portion investigated is larger in the direction transversal to the fibers. The detection volume, however, largely varies with anatomical and physical parameters that cannot be controlled or estimated in practical application (Table 1), thus the portion of muscle investigated substantially varies among subjects. This makes direct comparison of absolute amplitude values not appropriate and suggests the use of normalization methods.

Limitations

The volume conductor model used within this study is very simple. The muscle fibers of the EAS are assumed to be circular and their support plane is orthogonal to the axis of the anal canal. The mucosa is approximated with a cylindrical layer and only two layers are considered, with constant conductivity (using the same conductivity values for EAS and IAS). However, the simulation model has the advantage of an analytical solution.

More complex models of volume conductor would be difficult to develop at this stage since the geometry in real cases is not completely known. For example, the direction of the fibers of the EAS is not reported in anatomical studies and, in addition to circularly structured fibers, some studies [17] indicate that there are also longitudinal fibers in the EAS. Furthermore, it is not known if the fibers follow an elliptic, rather than circular, path,

changing depth within the muscle. It is also not known if the paths of the fibers of the EAS are orthogonal to the axis of the anal canal or if they are in a plane with an oblique angle with respect to the axis of the anal canal. In addition to anatomical information, physical properties of the volume conductor are also not fully known. For example, the values of conductivity of the mucosa and its changes in case of pathologies are not available.

The limited knowledge of physical, anatomical, and geometrical characteristics of the anal sphincter suggests that simulation studies cannot be accurate with respect to all the hypotheses. For this reason, a precise indication of the absolute amplitude of the simulated potentials expressed in microvolts would not be reliable. The inverse problem of associating experimental signals to simulated recordings cannot be addressed yet due to the oversimplification of the modeling approach. Further work should be done in this direction, probably by developing a numerical model of the anal sphincter (e.g., by finite elements method).

Because of the limitations stated above, in this study only comparisons of relative decay distances are provided. With the limited information available, it is not possible to give indication of the dimension of the detection volume in absolute terms, i.e. comparing the amplitude of the action potential of a motor unit of a specific size and in a specific position with the root mean square of the instrumentation noise. However, relative comparisons can be performed.

Conclusions

The detection volume of EMG recording systems from sphincter muscles extends more in the axial than in the depth direction and depends on anatomical (e.g., fiber length, thickness of the mucosa), physical (e.g., conductivity of the mucosa), and detection system parameters (e.g., spatial filter). Many of these parameters cannot be estimated in practical

measures, thus introducing a rather large variability in the muscle portion investigated among subjects. End-of-fiber components are enhanced by the circular geometry of sphincter muscles with respect to the case of a planar volume conductor. For increasing selectivity, small electrodes and bipolar or double differential recordings are suggested.

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TABLE CAPTION

Table 1: Detection distances in depth (*dd*) (within the muscle) and axial (*ad*) direction for three detection systems varying anatomical (thickness of the mucosa, fiber length) and physical (conductivity of the mucosa) parameters (*dd* and *ad* defined in Eq. (1), measured in mm; simulated configurations shown in Figure 1).

FIGURE CAPTIONS

Fig. 1 Cylindrical model of the volume conductor. The muscle extends to infinity in the radial direction and has longitudinal and transverse conductivity equal to 0.5 and 0.1 S/m, respectively. The mucosa layer has conductivity 0.5 and 1 S/m, and thickness h = 1 mm and 2 mm in different simulation sets. Longitudinal section for a long A) and short B) fiber. Three dimensional view C) with the definition of the axial distance z_0. Anal sphincter probe, with one array of rectangular electrodes in the angular direction D).

Fig. 2 Simulated single fiber action potentials detected at different transverse distances from the source (A, C, E) and different depths of the fiber within the muscle (B, D, F). The mucosa layer has conductivity 0.5 S/m and thickness h = 1 mm. Monopolar (A, B), bipolar (C, D), and double differential (E, F) signals are shown. A.U. = arbitrary units.

Fig. 3 Comparison between a probe resembling those used in clinical practice (large rectangular electrodes, 1.5 mm per 20 mm in the angular and axial direction, respectively; large interelectrode distance, about 21 mm) c, f, and the selective probe shown in Figure 1d. In the latter case both point (a, d) and rectangular (1 mm per 10 mm, in the angular and axial direction, respectively - b, e) electrodes are shown. Level curves plots (with levels 50% and 10% of the maximum) for EMG average rectified value (ARV) as a function of the position of a test fiber within the muscle (depth – y_0; transverse distance - z_0) are shown. The conductivity and thickness of the mucosa are 0.5 S/m and 1 mm, respectively. Monopolar (MP), single differential (SD), and double differential (DD) filters (DD only for the selective probe), short and long fibers are considered.

Fig. 4 Ratio between the peak-to-peak amplitude of the non-propagating signal component over the propagating component for long fibers. The conductivity and thickness of the mucosa are 0.5 S/m and 1 mm, respectively. Monopolar, single differential and double differential filters (interelectrode distance 2.74 mm as in experimental recordings [8]), point and rectangular electrodes with dimensions 2 mm in the angular direction (along the fiber direction) and 10 mm in the axial direction (transversal to the fibers) are considered.

Fig. 5 Comparison between sphincter and plane layer model. Level curves plots (with levels 50% and 10% of the maximum) for EMG average rectified value (ARV) as a function of the position of the fiber within the muscle (a, b, c). Level curves plot (with levels 40% and 20%) for the ratio between the non-propagating and the propagating signal component (d, e, f) as a function of the position of the fiber. The conductivity and thickness of the mucosa are 0.5 S/m and 1 mm, respectively. Monopolar (MP), single differential (SD), and double differential (DD) filters (interelectrode distance 2.74 mm), short and long fibers, point and rectangular electrodes (1 mm per 10 mm, in the angular and axial direction, respectively) are considered.

Fig 1

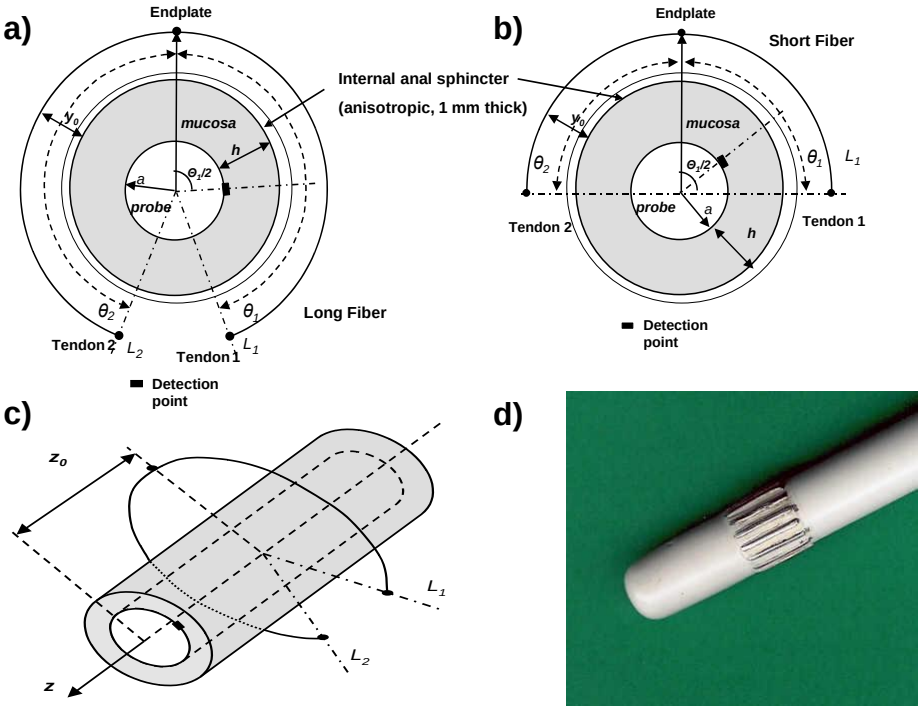

a)
Endplate
Internal anal sphincter
(anisotropic, 1 mm thick)
mucosa
h
a
probe
Long Fiber
Tendon 2
Tendon 1
Detection point
b)
Endplate
Short Fiber
mucosa
probe
Tendon 2
Tendon 1
Detection point
c)
d)

Fig 2

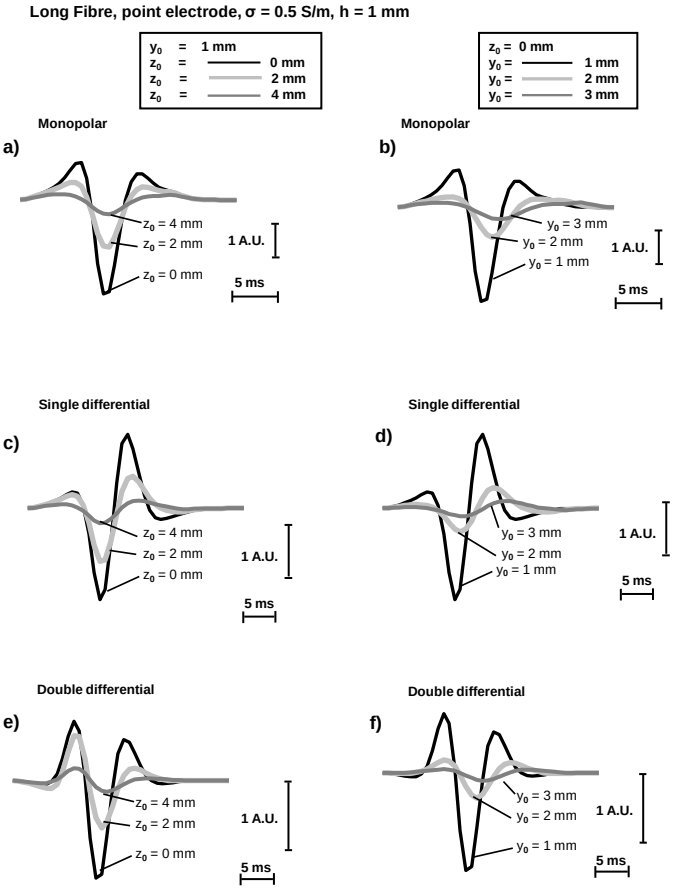

Long Fibre, point electrode, σ = 0.5 S/m, h = 1 mm
y0 = 1 mm
z0 = 0 mm
z0 = 2 mm
z0 = 4 mm
z0 = 0 mm
y0 = 1 mm
y0 = 2 mm
y0 = 3 mm
Monopolar
a)
z0 = 4 mm
z0 = 2 mm
z0 = 0 mm
1 A.U.
5 ms
Monopolar
b)
y0 = 3 mm
y0 = 2 mm
y0 = 1 mm
1 A.U.
5 ms
Single differential
c)
z0 = 4 mm
z0 = 2 mm
z0 = 0 mm
1 A.U.
5 ms
Single differential
d)
y0 = 3 mm
y0 = 2 mm
y0 = 1 mm
1 A.U.
5 ms
Double differential
e)
z0 = 4 mm
z0 = 2 mm
z0 = 0 mm
1 A.U.
5 ms
Double differential
f)
y0 = 3 mm
y0 = 2 mm
y0 = 1 mm
1 A.U.
5 ms

Fig 3

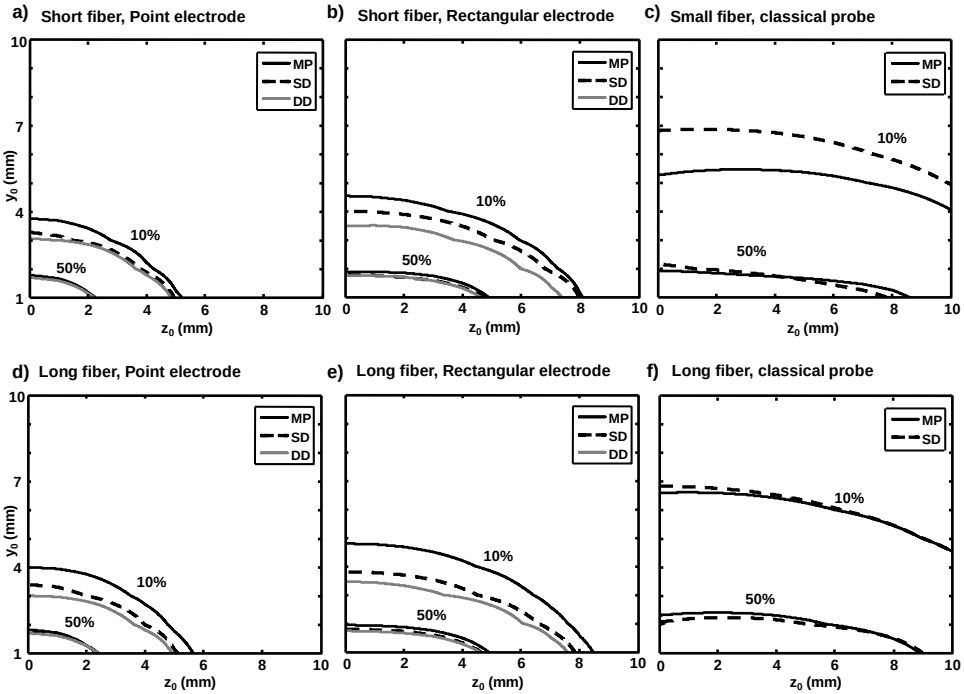

a) Short fiber, Point electrode
b) Short fiber, Rectangular electrode
c) Small fiber, classical probe
d) Long fiber, Point electrode
e) Long fiber, Rectangular electrode
f) Long fiber, classical probe
MP
SD
DD
10%
50%
y_0 (mm)
z_0 (mm)

Fig 4

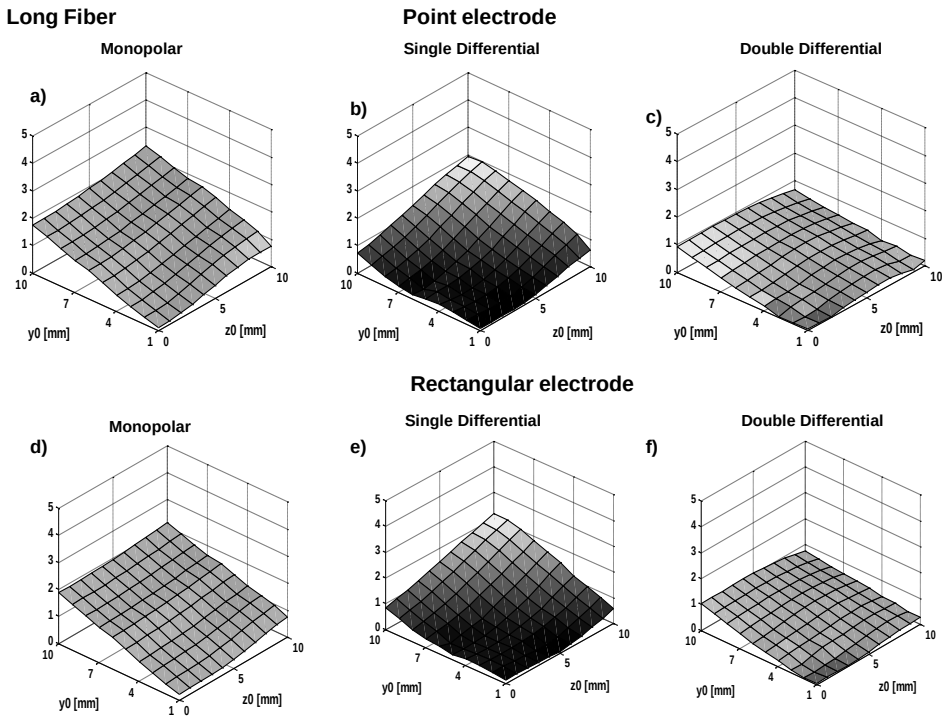
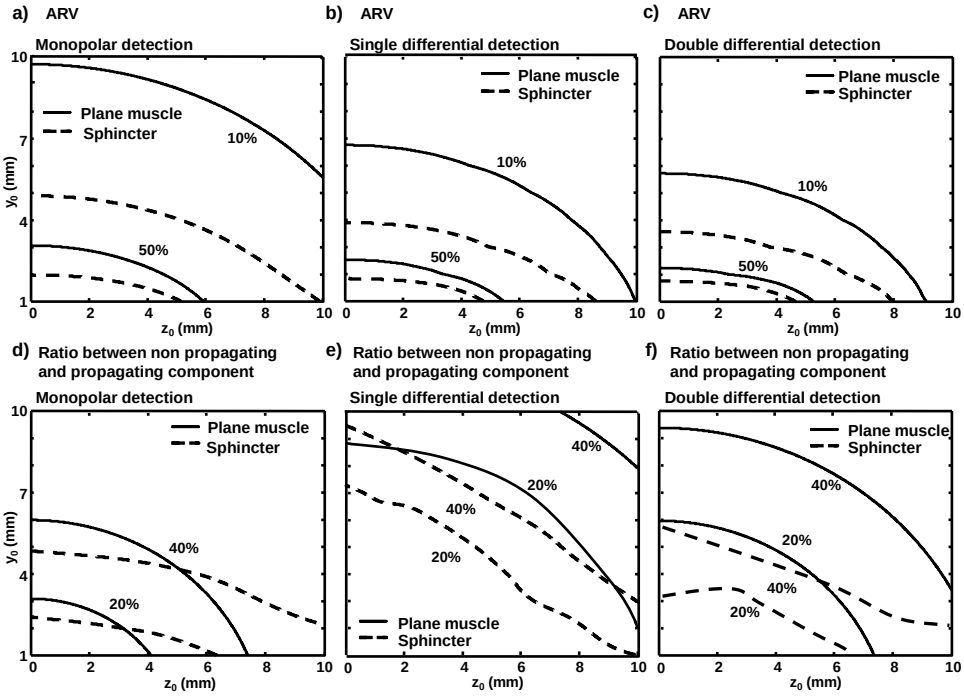

Long Fiber
Point electrode
Monopolar
Single Differential
Double Differential
a)
b)
c)
Rectangular electrode
Monopolar
Single Differential
Double Differential
d)
e)
f)
y0 [mm]
z0 [mm]

Fig 5


a) ARV
Monopolar detection
b) ARV
Single differential detection
c) ARV
Double differential detection
Plane muscle
Sphincter
10%
50%
d) Ratio between non propagating and propagating component
Monopolar detection
e) Ratio between non propagating and propagating component
Single differential detection
f) Ratio between non propagating and propagating component
Double differential detection
40%
20%
y_0 (mm)
z_0 (mm)