Sensitivity of surface EMG-based conduction velocity estimates to local tissue in-homogeneities – influence of the number of channels and inter-channel distance

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SENSITIVITY OF SURFACE EMG-BASED CONDUCTION VELOCITY ESTIMATES TO LOCAL TISSUE IN-HOMOGENEITIES – INFLUENCE OF THE NUMBER OF CHANNELS AND INTER-CHANNEL DISTANCE

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ABSTRACT

The aim of this simulation study was to investigate the influence of local tissue in-homogeneities on the estimates of muscle fiber conduction velocity (CV) from surface EMG signals. A recently developed analytical surface EMG model was used to generate simulated surface EMG signals from a planar layered volume conductor, comprised of the muscle tissue and fat layer, with spheres (1 mm radius) in the fat layer of conductivity different from the surrounding tissue. CV was estimated with a maximum likelihood multi-channel approach, varying the number of channels and the inter-channel distance used for the estimate. The action potentials detected along the muscle fiber direction changed shape due to the presence of the in-homogeneities, thus affecting CV estimates. CV estimates were influenced by the location of the in-homogeneities with respect to the fiber and detection electrodes. The maximum percent variation of CV estimates due to the presence of in-homogeneities decreased with increasing number of channels and inter-channel distance: 19.6% (2 channels), 12.1% (3 channels), 6.4% (4 channels), for 5 mm inter-channel distance, and 12.0% (2 channels), 5.2% (3 channels), 2.4% (4 channels), for 10 mm inter-channel distance (for double differential detection). The results were in agreement and explained previous experimental findings. It was concluded that multi-channel methods for CV estimation significantly reduce the sensitivity of CV estimates to tissue in-homogeneities.

1. INTRODUCTION

The estimation of muscle fiber conduction velocity (CV) is one of the most relevant issues in surface EMG signal detection and processing. Muscle fiber CV is an important physiological parameter which allows an insight into the muscle fiber membrane properties (Arendt-Nielsen & Zwarts, 1989). Although the basic requirements for CV estimation are rather simple, the issue is made complex by the deviations from ideal conditions of the practical cases (Arabadzhiev et al., 2003). Surface EMG signals detected along the muscle fiber direction are ideally delayed versions of the same waveform. However, this is never the case in practice. As a consequence, there is not a unique definition of delay, but rather many definitions are possible and each of them represents an estimation method (Farina & Merletti, 2004).

Many methods for estimating CV have been proposed in the past (for a recent review refer to Farina & Merletti, 2004). In particular, CV can be estimated from two or more surface EMG signals detected along the direction of muscle fibers with a linear array of electrodes (Farina et al., 2001). The selection of the number of signals and distance between detection points should be based on the analysis of performance accounting for many possible sources of error. Desirable characteristics of a CV estimation method are, for example, small estimation variance, repeatability of the results, small sensitivity to electrode displacements, to end-plate and end-of-fiber components (Dimitrova, 1974), to fiber inclination with respect to the detection system. Some of these characteristics can be compared among different CV estimation methods with relatively simple models. Estimation variance can be for example determined by simulations with phenomenological EMG models (Farina & Merletti, 2000). Other characteristics of CV estimation methods are more complex to be analysed. The experimentally observed variability of CV estimates for different electrode locations in case of long fibers and superficial motor units can not be easily interpreted by models of signal generation. Models assuming space-invariant systems (i.e., volume conductors which present the same geometrical and physical characteristics along the direction of muscle fibers) (Farina et al., 2004d) do not predict shape changes in the detected potential along the direction of propagation if

the end-plate and end-of-fiber effects are negligible. Experimentally, the variation of the estimates with small electrode displacements may be as large as 25% (Farina et al., 2004a). These variations may be due to many factors, including the inclination of the fibers with respect to the detection systems or local tissue in-homogeneities (Schneider et al., 1991).

From experimental analyses, it was shown that CV estimates present significantly different reproducibility depending on the number of EMG channels and inter-channel distance adopted (Farina et al., 2004a). These results were interpreted in the light of a different sensitivity of methods for CV estimation to the location of the electrodes over the muscle. In particular, increasing the number of channels used for CV estimation was effective in reducing the sensitivity to electrode displacements (Farina et al., 2004a). To some extent, this sensitivity was reduced also by increasing the distance between detection points for CV estimation. The increasing of the number of surface EMG channels results in similar sensitivity of CV estimates to end-plate and end-of-fiber components and to fiber misalignment (Farina et al., 2001). Thus, it was speculated that the reduction of sensitivity to small electrode displacements, associated to increasing the number of channels, reported by Farina et al. (2004a), could be due to a decreased influence of the action potential shape changes due to local tissue in-homogeneities. However, the unavailability of models accounting for small tissue in-homogeneities did not allow a clear model-based interpretation of the experimental results.

We recently developed an analytical model for surface EMG signal simulation in volume conductors of general shape with local, spherical in-homogeneities (Mesin et al., 2004; Mesin & Farina, 2004). The system described is non-space invariant along the direction of source propagation. The main aim of this simulation study is to assess the sensitivity of multi-channel algorithms for CV estimation to local tissue in-homogeneities, in order to interpret experimental results obtained in previous work (Farina et al., 2004a).

2. METHODS

Surface EMG signal simulation

The surface EMG model proposed by Mesin et al. (2004; Mesin & Farina, 2004) was used to simulate single fiber action potentials in a two-layer planar volume conductor (Figure 1). The volume conductor describes an anisotropic muscle tissue, where the fibers are located, and an isotropic fat layer. Local spherical in-homogeneities were located in the isotropic layer, at random locations. The model derivation is detailed elsewhere (Mesin & Farina, 2004). Briefly, the effect of a local spherical in-homogeneity in the isotropic layer is described adding a perturbation term to the in-homogeneity free solution (i.e., the potential distribution generated in the volume conductor without the in-homogeneity). The perturbation term is a series of harmonic functions decaying at infinity (Sneddon, 1966). In the model, only the first two terms of this series are considered, thus obtaining an approximate solution (Landau & Lifshitz, 1975; Mesin & Farina, 2004). Moreover, in the case of more than one in-homogeneity, the mutual effects between the perturbation terms of the in-homogeneities are neglected. The approximations introduced can all be evaluated analytically (Mesin & Farina, 2004) and imply constraints in the selection of the geometrical relations between the source and the in-homogeneities. This selection was performed so that the worst case approximation error was smaller than 5% of the perturbation term (distance between two spheres larger than $3 \cdot R$, where R is the radius of the in-homogeneity). In this study, the fibers were assumed of infinite length, in order to exclude end-plate and end-of-fiber effects, whose influence on CV estimates has been analysed in previous work (Arabadzhiev et al., 2003; Farina et al., 2001; Farina et al., 2002a).

Three spherical in-homogeneities were considered as randomly distributed in the fat layer (Figure 1). Twenty-five sets of random locations of the three spheres were considered for each set of simulation parameters. The maximum distance between the centers of the spheres was set to $6 \cdot R$. The locations of the centers of the spheres over the 25 simulations covered uniformly a line 60 mm long, above the fiber, 2 mm deep in the fat layer (Figure 1). The sensitivity of CV estimates to the

presence of the in-homogeneities was assessed by the standard deviation of CV estimates across the 25 random locations of the spheres. The lower the standard deviation, the smaller the effect of the in-homogeneities on the estimates. The parameters varied in the simulations were: 1) the distance between detection points (5, 10, 15 mm); 2) the spatial filter applied for signal detection (monopolar, single differential, double differential, and normal double differential) (Reucher et al., 1987a; 1987b; Disselhorst-Klug et al., 2000); and 3) the conductivity of the spheres (ratio with the conductivity of fat 2.5-10, 2.5 increments). The simulated CV was in all cases 4 m/s. The transmembrane current was described as a tripole, with parameters taken from Merletti et al. (1999).

Figure 1 about here

Conduction velocity estimation

CV was estimated as the ratio of the distance between the centres of the two spatial filters and the delay of propagation of the potentials detected by the spatial filters (referred in the following as signals). The delay was estimated by a maximum likelihood approach which consists in minimizing the following mean square error function (Farina et al., 2001):

$$e^2(\theta) = \sum_{k=1}^{K}\sum_{n=1}^{N}\left[x_k(n) - \frac{1}{K}\sum_{m=1}^{K} x_m(n+(m-k)\theta) \right]^2 \qquad (1)$$

where K is the number of signals used for the estimate, $x_k(n)$ are the signals, and θ is the delay between adjacent signals. In case $K = 2$, the minimization of the mean square error (1) is equivalent to the spectral matching estimator (McGill & Dorfman, 1984) or to the cross-correlation function method (Naeije & Zorn, 1983; Parker & Scott, 1973), often used in the applications. For K larger than two, the estimator is a multi-channel maximum likelihood estimator, recently used in basic and applied studies (Farina et al., 2002b; Farina et al., 2004b; Farina et al., 2004c). CV will be estimated from the simulated signals using 2, 3, or 4 channels.

3. RESULTS

Figure 2 shows surface EMG signals detected by three double differential systems located along fiber direction and generated by an infinite muscle fiber in a layered volume conductor with three spherical in-homogeneities (see also Figure 1). The difference between the potentials detected with and without the in-homogeneities is also shown. The in-homogeneities introduce signal components which do not travel along the direction of propagation of the intra-cellular action potential, thus making the shapes of the three detected potentials different.

Figure 2 about here

Figure 3 reports the potential distributions as generated by an impulsive current and detected by three spatial filters in case of a local in-homogeneity placed over the source. The sphere introduces a perturbation effect on the potential distribution, whose shape and relative amplitude depends on the spatial filter applied.

Figure 4 shows the standard deviation of CV estimation across the 25 simulations with random location of the spheres, for the four detection systems, the different numbers of channels and inter-channel distances. The variability of CV estimates was significantly reduced by 1) increasing the inter-channel distance, and 2) increasing the number of channels used for the estimates. In particular, for the longitudinal double differential filter, CV standard deviation with 5 mm inter-electrode distance was reduced from approximately 0.4 m/s to approximately 0.1 m/s when the number of channels increased from 2 to 4. The standard deviation was further reduced increasing the distance between detection points. CV can be over- or under-estimated in the presence of in-homogeneities, depending on the relative location of the in-homogeneities with respect to the detection electrodes (Figure 5).

Different filters led to different sensitivities of CV estimates to the in-homogeneities. Longitudinal and normal double differential filters performed worse than single differential and monopolar

configurations, which is probably related to the number of electrodes used for the spatial filtering in the direction of source propagation.

In case of maximum difference between the fat and the sphere conductivities and double differential detection, the maximum percent variation of CV estimates over the 25 conditions was 19.6% (2 channels), 12.1% (3 channels), 6.4% (4 channels), for 5 mm inter-channel distance, and 12.0% (2 channels), 5.2% (3 channels), 2.4% (4 channels), for 10 mm inter-channel distance.

Figures 4 and 5 about here

4. DISCUSSION & CONCLUSION

CV estimates are affected by many factors related to the generation and detection of the surface EMG signal. These factors influence the estimates to a different extent depending on the estimation method. Since the factors to be considered are many, the selection of a specific estimation method is not trivial and should be based on a systematic evaluation of the advantages and drawbacks of the different methods. In this study we evaluated, by modelling, maximum likelihood methods for delay (and thus CV) estimation. The study provides for the first time the assessment of the influence of the number of channels and distance between channels on the sensitivity of CV estimates to local tissue in-homogeneities.

Multi-channel CV estimation methods have been recently used in applied studies and proved to significantly reduce the estimation variance due to noise with respect to classic two-channel methods (Farina et al., 2002b). Moreover, it has been recently shown that the use of more than two channels for CV estimation increases the repeatability of the measure of CV (Farina et al., 2004a) and this was associated to the experimental observation that increasing the number of channels decreases the sensitivity of the measure to small electrode displacements. Farina et al. (2004a) showed that sensitivity of CV estimates to electrode displacements of ±5 mm can be as large as 25% using two channels for the estimate, with a reduction to approximately 8% using four channels.

These experimental results can now be interpreted in the light of the simulations performed. The simulations showed that the number of channels and the distance between detection points have a large influence on the sensitivity of the estimates to local in-homogeneities (Figure 5). The analysis performed was possible with an advanced surface EMG model, which considers a non-space invariant generation system, i.e., a system which changes its properties (in this case the conductivity) along the direction of propagation of the source (Farina et al., 2004d; Mesin & Farina, 2004). This model allowed the simulation of changes in the shape of the surface detected action potentials generated by infinite muscle fibers. Moreover, the model permitted to explain the large variability of CV estimates obtained in different locations along the muscle fibers observed experimentally and presumably not due to action potential generation and/or extinction.

The results presented indicated also that different spatial filters applied for signal detection may result in different sensitivity of the estimates to the in-homogeneities. Indeed, different spatial filters reduce to a different extent the perturbation of the potential caused by the presence of the in-homogeneities (Figure 3).

We focused on noise-free signals and on infinite length muscle fibers. This allowed to investigate the effects due to tissue in-homogeneities only, avoiding confounding factors such as the end-of-fiber components. The sensitivity of CV estimates to these components and to noise has been analyzed in previous work (Farina & Merletti, 2000; Farina et al., 2001; Farina et al., 2002a; Farina et al., 2002b).

The geometry of the in-homogeneities (spheres) was imposed in this simulation study by the mathematical derivations (Mesin & Farina, 2004). Real in-homogeneities are represented by blood vessels, sweat glands and may have more complex geometries. Despite this approximation, we provided an assessment of the impact of small perturbations in the conductivity tensor of the volume conductor upon the estimate of muscle fiber CV. Moreover, we provided results for different conductivities of the in-homogeneities in order to cover different practical situations.

It is concluded that increasing the number of channels and the distance between channels for estimating muscle fiber CV from surface EMG signals leads to an improvement of the stability of the estimate in case of shape changes of the surface detected action potentials. The increased stability of the estimates increases repeatability of the results (Farina et al., 2004a), which is particularly relevant in clinical applications. This study thus provides evidence that the use of multi-channel methods for CV estimation is advantageous not only for decreasing standard deviation of estimation due to additive noise (Farina et al., 2001), but also for reducing the sensitivity of the estimates to variations of the tissue properties along the path of intra-cellular action potential propagation.

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Figure captions

Fig. 1 Parameters of the simulations and notations. The volume conductor is made of two planar layers (fat: thickness 4 mm, conductivity 0.05 S/m; muscle: infinite depth, conductivity along the fibers 0.5 S/m, transversal 0.1 S/m). The fiber is infinite, 4 mm deep in the muscle. The transmembrane current is modeled by a tripole approximation (parameters from Merletti et al., 1999). Three aligned spherical in-homogeneities (1 mm radius) are randomly located in the fat layer (25 uniformly distributed locations of the three spheres along a 60 mm length) along fiber direction. The sphere are located above the fiber in all cases. All spheres have the same conductivity, which varies among 0.125 S/m, 0.25 S/m, 0.375 S/m, 0.5 S/m. The detection system is a matrix of electrodes (12×3 point electrodes) which allows to study multi-channel (2, 3, and 4 channels, inter-channel distance d_p) monopolar, longitudinal and normal double differential detection. The detection point indicates the center of the detection system. The same matrix but shifted in the longitudinal direction by 2.5 mm was used for the single differential detections, so that all the spatial filters investigated had the same detection points. The inter-electrode distance of the spatial filters (distance between the electrodes forming each filter) was 5 mm in all cases. LSD stands for longitudinal single differential filter, LDD for longitudinal double differential, and NDD for normal double differential.

Fig. 2 Simulated potentials detected by a double differential system and generated by a current tripole (parameters from Merletti et al., 1999) which travels along an infinite fiber. a) The volume conductor and the detection points (see also Figure 1). The three in-homogeneities have a radius of 1 mm and a conductivity of 0.5 S/m. b) The simulated potentials with and without the in-homogeneities. c) The perturbation term due to the in-homogeneities (difference between the plots in solid and dashed lines in b). A.U. stands for arbitrary units.

Fig. 3 Surface potentials generated by an impulsive current source, as detected by longitudinal single and double differential, and normal double differential systems. The perturbation term is shown on the left. One sphere (1 mm radius, 0.5 S/m conductivity), located above the impulsive current source, is included in the volume conductor shown in Figure 1. A.U. stands for arbitrary units.

Fig. 4 Standard deviation (across 25 simulations) of CV estimates (logarithmic scale) as a function of the conductivity of the three spheres (1: 0.125 S/m; 2: 0.25 S/m; 3: 0.375 S/m; 4: 0.5 S/m), for the four detection systems, the three sets of channels, and the three inter-channel distances. The 25 simulations were obtained by randomly positioning each of the three spheres under the electrode matrix in the direction of the fiber with uniform probability density in a range of 60 mm (see also Figure 1).

Fig. 5 CV estimates for the 25 simulations with random locations of the three spheres (1 mm radius, 0.5 S/m conductivity). The spatial filter used is the single differential. Note the significantly reduced variability of the estimates with increasing the number of channels and inter-channel distance. LSD stands for longitudinal single differential, d_p indicates the inter-channel distance.

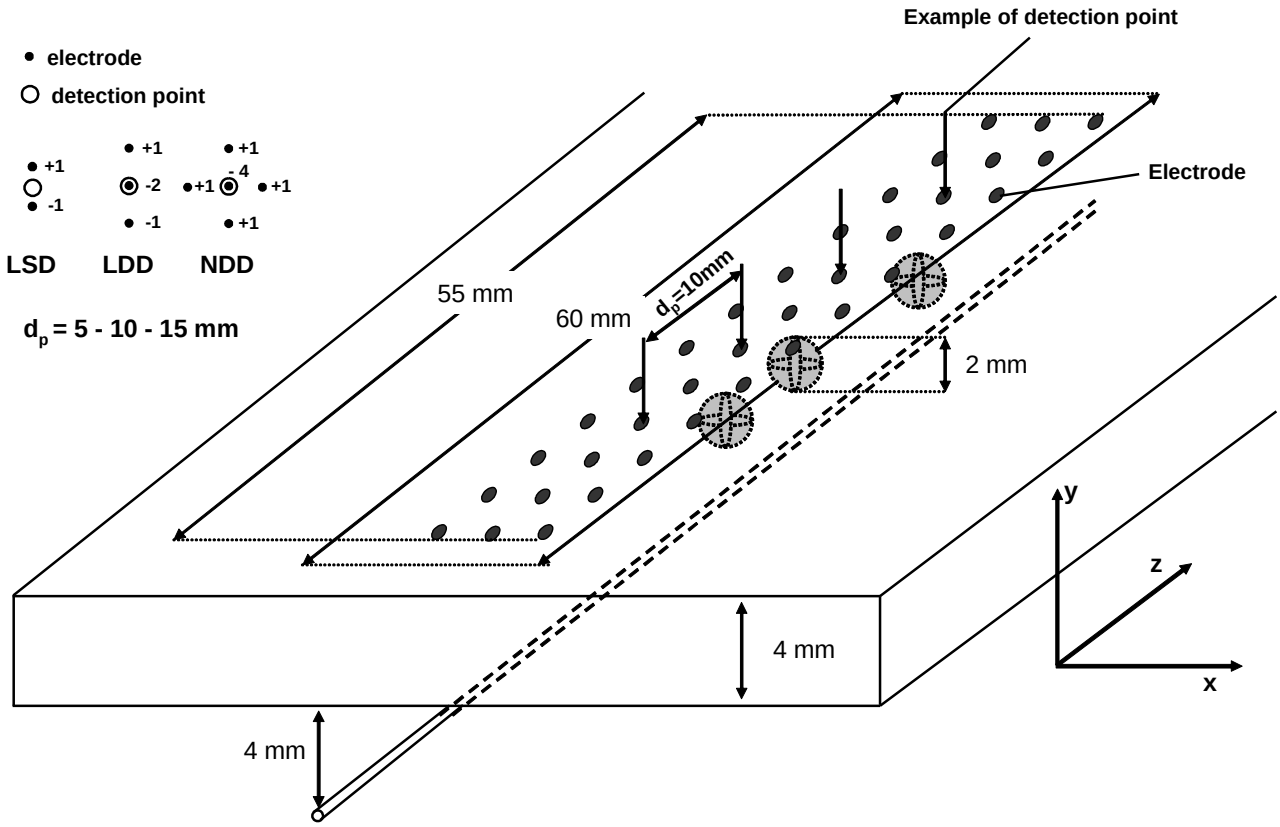

Example of detection point
electrode
detection point
+1
-1
+1
-2
-1
+1
- 4
+1
+1
+1
LSD
LDD
NDD
d_p = 5 - 10 - 15 mm
Electrode
55 mm
60 mm
d_p=10mm
2 mm
4 mm
4 mm
y
z
x

Figure 1

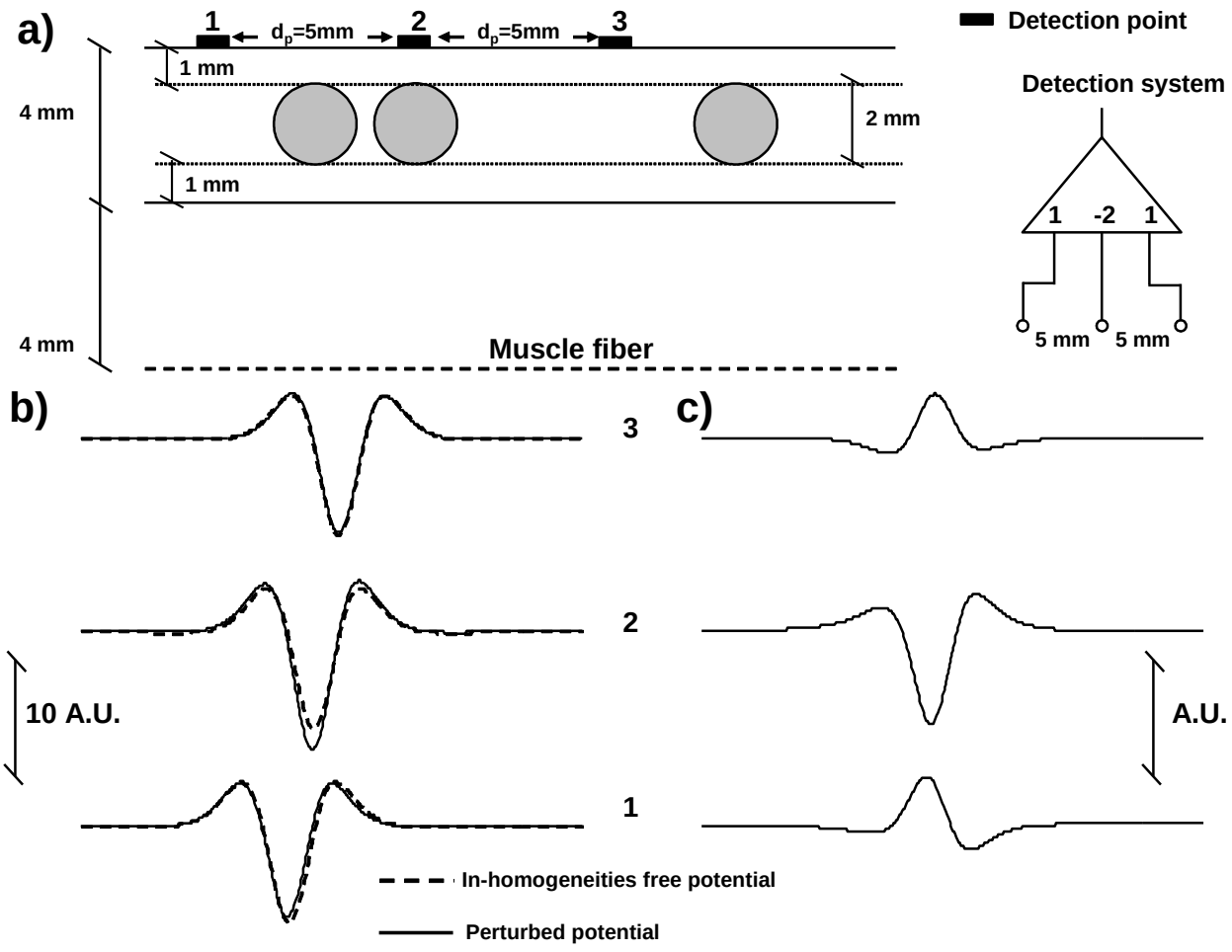
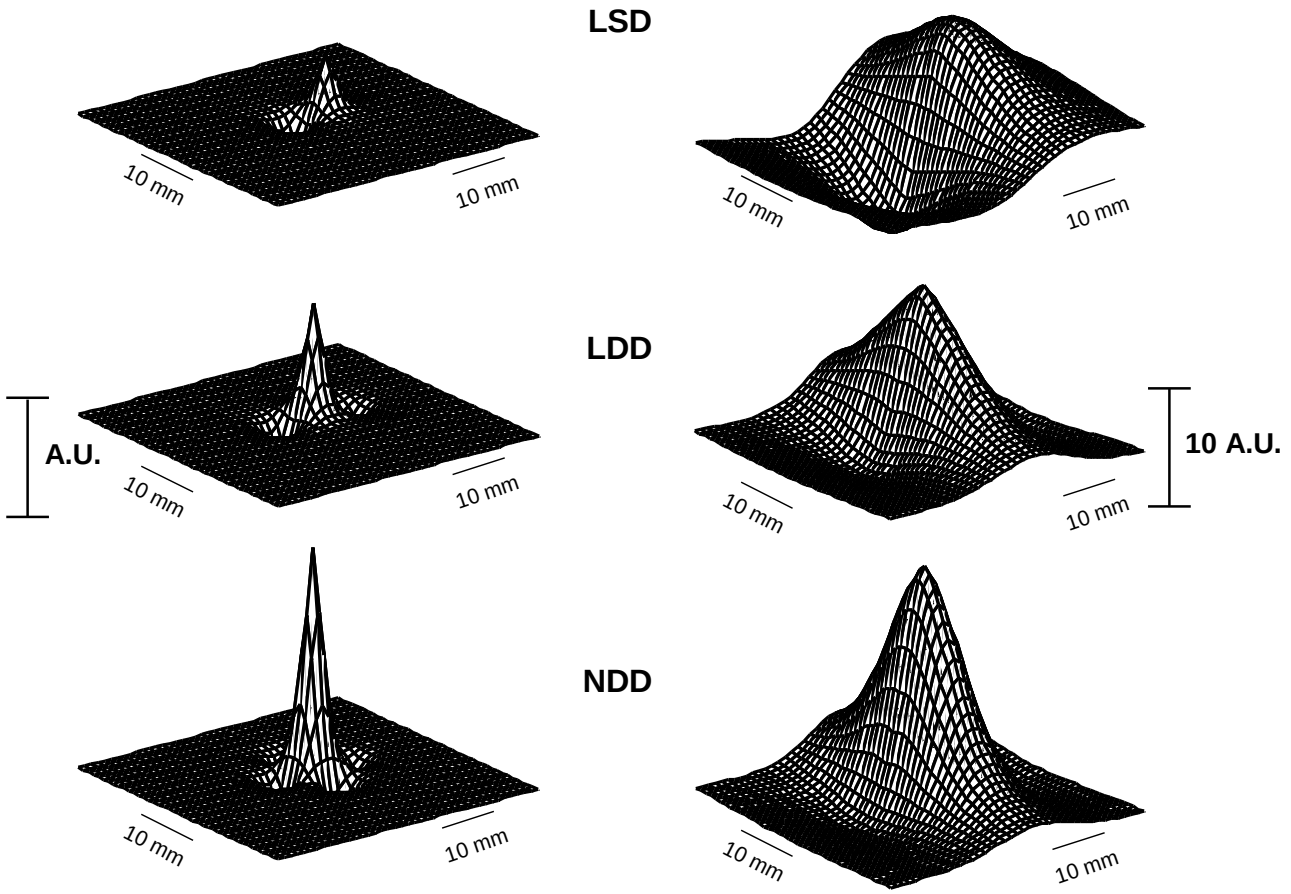

a)
1
d_p=5mm
2
d_p=5mm
3
Detection point
1 mm
4 mm
2 mm
1 mm
Detection system
1 -2 1
5 mm
5 mm
4 mm
Muscle fiber
b)
c)
3
2
1
10 A.U.
A.U.
In-homogeneities free potential
Perturbed potential

Figure 2


LSD
LDD
NDD
A.U.
10 A.U.
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10 mm
10 mm

Figure 3

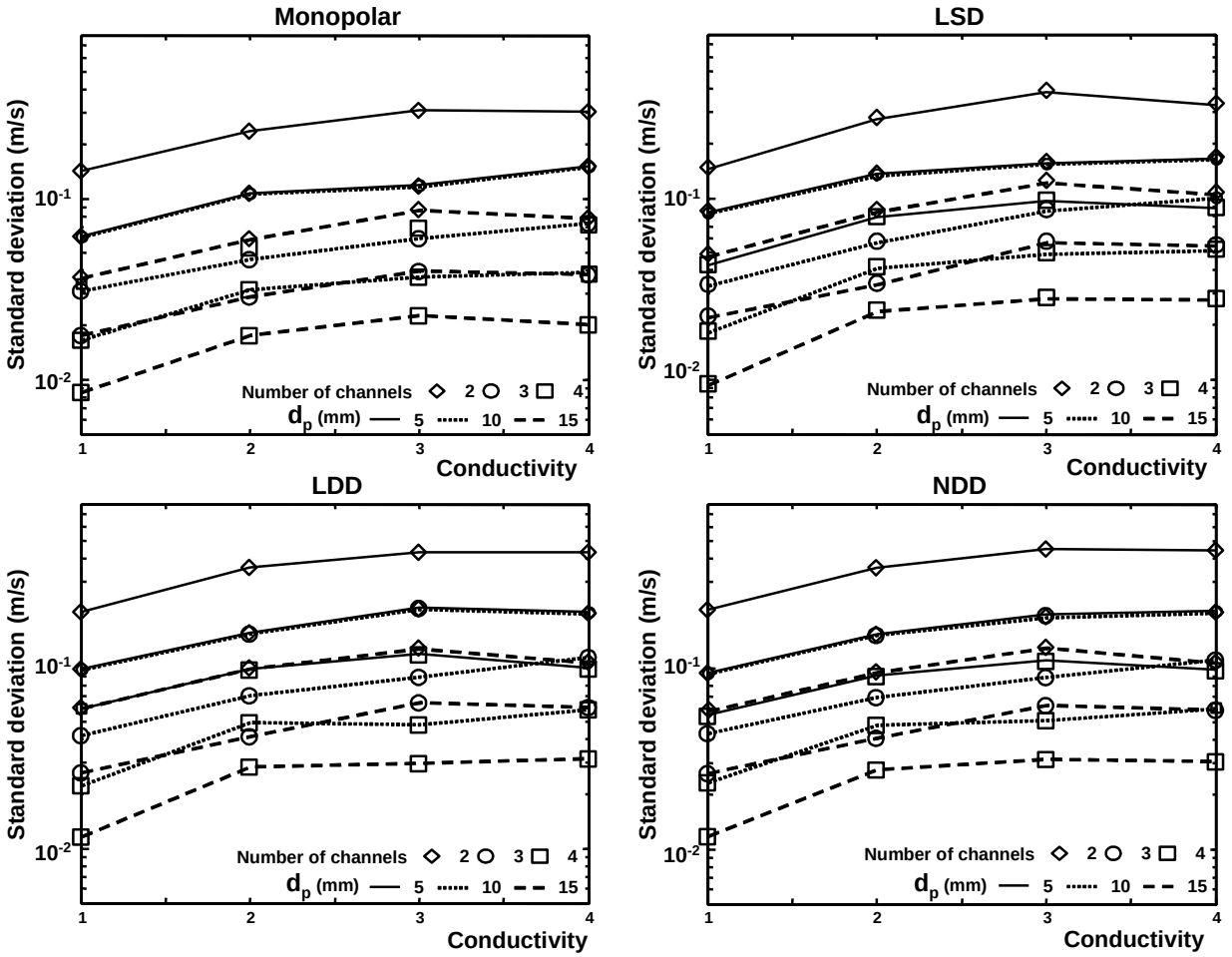
Monopolar
LSD
LDD
NDD
Standard deviation (m/s)
10-1
10-2
Number of channels 2 3 4
dp (mm) 5 10 15
1
2
3
4
Conductivity

Figure 4

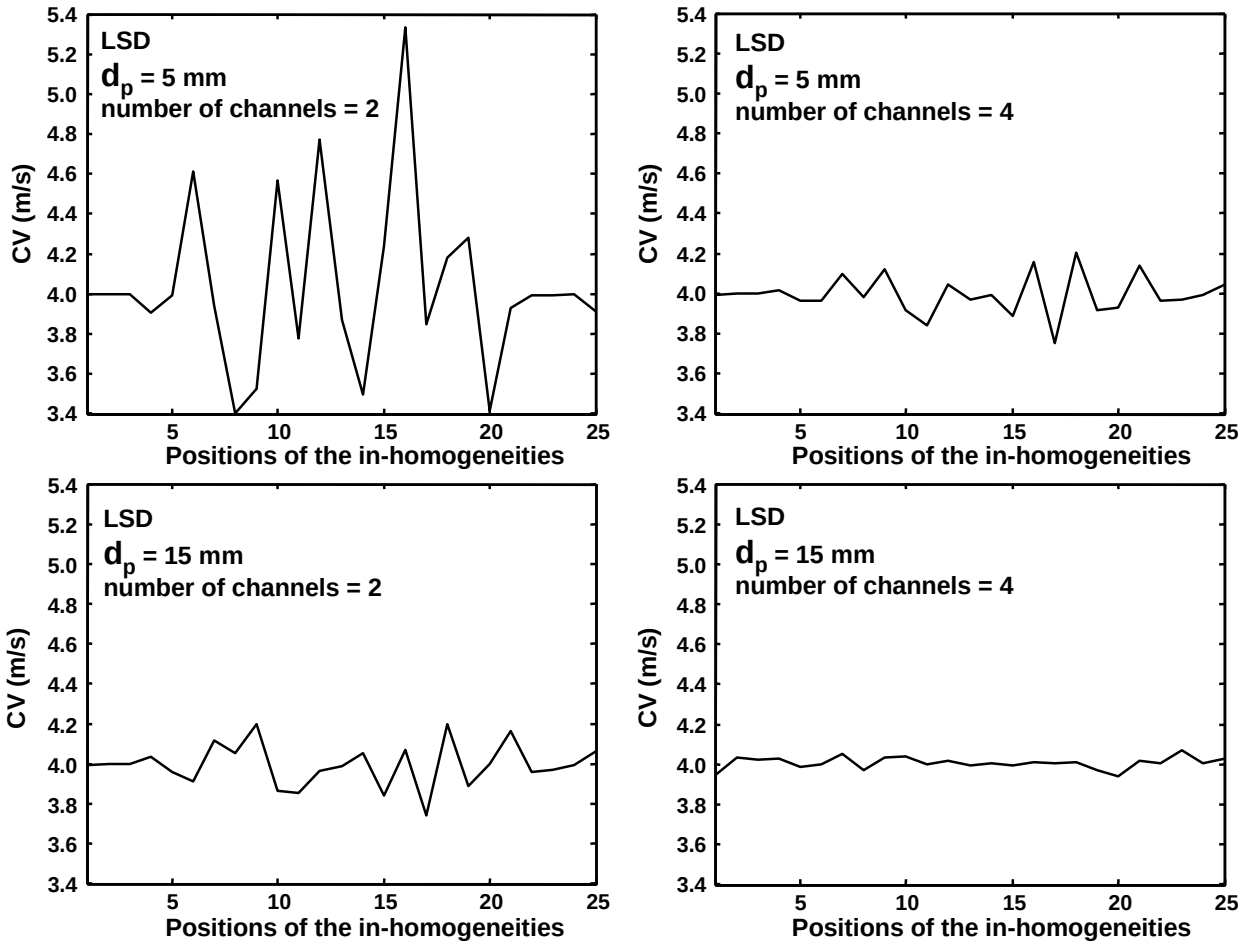
LSD
d_p = 5 mm
number of channels = 2
LSD
d_p = 5 mm
number of channels = 4
LSD
d_p = 15 mm
number of channels = 2
LSD
d_p = 15 mm
number of channels = 4
CV (m/s)
Positions of the in-homogeneities
5.4
5.2
5.0
4.8
4.6
4.4
4.2
4.0
3.8
3.6
3.4
5
10
15
20
25

Figure 5