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## Surface EMG: the issue of electrode location

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## **Abstract**

This paper contributes to clarifying the conditions under which electrode position for surface EMG  
25 detection is critical and leads to estimates of EMG variables that are different from those obtained  
in other nearby locations. Whereas a number of previous works outline the need to avoid the  
innervation zone (or the muscle belly), many authors place electrodes in the central part or bulge of  
the muscle of interest where the innervation zone is likely to be. Computer simulations are  
presented to explain the effect of the innervation zone on amplitude, frequency and conduction  
30 velocity estimates from the signal and the need to avoid placing electrodes near it. Experimental  
signals recorded from some superficial muscles of the limbs and trunk (abductor pollicis brevis,  
flexor pollicis brevis, biceps, upper trapezius, vastus medialis, vastus lateralis) were processed  
providing support for the findings obtained from simulations. The use of multichannel techniques is  
recommended to estimate the location of the innervation zone and to properly choose the optimal  
35 position of the detection point(s) allowing meaningful estimates of EMG variable during movement  
analysis.

**Keywords:** surface electromyography, EMG, electrode location, electrode position, electrode  
arrays.

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**Acronyms:** CV: muscle fiber conduction velocity; DD: double differential; IED: interelectrode  
distance; IZ: innervation zone; MU: motor unit; MUAP: motor unit action potential; NMJ:  
neuromuscular junction; SD: single differential.

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## Introduction

During the last 30 years the effect of electrode location on estimates of conduction velocity (CV), amplitude and spectral variables of the surface EMG has been addressed in a number of methodological and clinical publications [4, 6, 7, 12, 13, 15, 16, 17, 19, 27, 31, among many others] considering muscles (or group of muscles) ranging from the masticatory muscles to the muscles of the shoulder, of the arm and leg. The most significant standardization effort took place in 1997-1999 within the European Project on “Surface EMG for Non Invasive Assessment of Muscles” (SENIAM) [www.seniam.org] where a detailed analysis of literature was presented for a number of muscles. Three strategies for placement of an electrode pair resulted from this analysis as the most used and are reported below (only data concerning the biceps brachii muscle are reported as an example):

- on the center or on the most prominent bulge of the muscle belly (in 10 out of 21 publications)
- somewhere between the innervation zone (IZ) and the distal tendon (in 6 out of 21 publications)
- on the motor point (in 1 out of 21 publications).

In 4 out of the 21 reviewed publications the electrode location was either not mentioned or was unclear, indicating the little attention paid by authors and reviewers to this issue.

The orientation of the detection system (usually constituted by an electrode pair) with respect to the direction of muscle fibers was rarely mentioned. The SENIAM recommendation of avoiding placing electrodes over the IZ was based on criteria that were only partially documented within the Project itself.

The objective of this paper is the discussion and updating of considerations and criteria for surface EMG electrode location on the basis of both computer simulations and experimental data.

## Methods

### *Simulations*

Computer simulation of the sources and of the volume conductor produces the potential distribution  
75 on the surface of the skin. The model described in [8] has been used to compute the single  
differential (SD) potential present under each electrode on the skin (skin thickness 1 mm,  
conductivity 0.022 S/m; fat thickness 3 mm, conductivity 0.04 S/m; semi-infinite muscle layer with  
longitudinal conductivity 0.4 S/m and transversal conductivity 0.09 S/m). Single fibers of  
semilength equal to 60 mm have been simulated and their contributions have been added to  
80 generate the motor unit action potential (MUAP). The contributions of three motor units (MU) have  
been added to simulate the interference EMG signal.

### *Experimental data*

Surface EMG signals from the muscles listed in Table 1 were considered. Some of the experimental  
85 data have been already discussed in previous works of our group (see references in Table 1), which  
the reader can refer to for a detailed description of the experimental set-up. Surface EMG signals  
were recorded in SD configuration during isometric contractions. Two contraction levels expressed  
in terms of percentage of maximal voluntary contraction (MVC) were considered (except for vastus  
lateralis and medialis), one level was low (lower than or equal to 30% MVC), the other high (equal  
90 to or higher than 50% MVC). In the case of the biceps brachii, the dominant arm was fixed in a  
brace with angle of the elbow 105°. Linear arrays aligned with muscle fibers were used to detect  
surface EMG except for upper trapezius and biceps brachii, in which case two dimensional arrays  
were used in the acquisition of data (for biceps brachii the same detection system described in [30]  
was used), but only the central column (out of five) was considered.

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**Table 1: list of muscle from which experimental data have been collected**

<b>Muscle</b>	<b>IED (mm)</b>	<b>Force levels (% MVC)</b>	<b>Number of subjects</b>	<b>Reference</b>
<i>Abductor pollicis brevis</i>	2.5	10 and 50	5	[25]
<i>Flexor pollicis brevis</i>	2.5	10 and 50	5	[25]
<i>Biceps brachii</i>	8	30 and 70	8	This work
<i>Upper trapezius</i>	8	20 and 80	6	[30]
<i>Vastus lateralis</i>	5	90	10	[26]
<i>Vastus medialis</i>	5	90	10	[26]

100 *Signal processing*

The position of the main IZ was identified by visual analysis of the EMG signals as the channel under which the phase inversion of the MUAPs detected in SD mode occurred, and where the MUAPs began to propagate in two opposite directions. Amplitude and spectral variables were estimated from SD channels, CV was estimated from pairs of double differential (DD) channels with a maximum likelihood algorithm. The EMG variables averaged rectified value (ARV), mean frequency (MNF) and CV were computed for the channel corresponding to IZ, the three distal channels and the three proximal channels (two distal and two proximal channels in the case of vastus medialis and lateralis), on eight 0.5 s long adjacent epochs (at the beginning of the contraction, to reduce the effect of fatigue), for each signal. The values of the variables obtained in eight different epochs were then averaged (considering only CV values between 1 and 10 m/s and discarding up to four outliers, i.e. values distant from the mean more than twice the standard deviation). Thus, for each signal array (i.e., fixing the muscle, the subject, and the force level), seven average values (or five, in the case of vastus medialis and lateralis) were obtained for each of the parameters under consideration (ARV, MNF and CV) corresponding to the seven (or five)

115 geometrical locations. The signals were processed only if three (or two) channels distal and proximal with respect to the IZ could be found, so that seven (or five) locations could be considered (only some of the subjects under study in the quoted papers were thus considered herein). Table 1 provides the number of the subjects considered for each muscle.

In order to compare signals from different subjects, the time averaged variables corresponding to  
120 each signal were linearly scaled so that the maximum value (across different channels for each subject) was assigned the value 1 and the minimum value was assigned to 0 and they were studied as a function of the distance from the IZ. The adopted scaling equation was

$$x_{norm} = \frac{x - x_m}{x_M - x_m}$$

where  $x$  is the current value of the considered variable,  $x_{norm}$  is its normalized value,  $x_m$  and  $x_M$   
125 are the minimum and maximum value observed along the array ( $\pm 3$  interelectrode distance (IED) or  $\pm 2$  IED for vastus medialis and lateralis).

One way non-parametric Friedman ANOVA for repeated measures was performed to assess if electrodes/channels (that is the positions over the muscles) play a role as factor in the distribution of variance. When the test indicated significant variations (significance level always set to  $p < 0.05$ ),  
130 pair-wise comparisons were performed with the Dunns post-hoc test looking for statistically significant differences among pairs of averaged variables from individual channels.

## Results.

### 135 *Simulations*

Let us first consider a single muscle fiber parallel to the skin. Two point electrodes placed on the skin symmetrically with respect to the Neuromuscular Junction (NMJ) and along the fiber direction detect a differential signal that is obviously zero, as shown in Fig. 1, due to specularity of the propagating signals.

140 Similar results are obtained in the case of a single MU with fibers parallel to the skin and NMJs uniformly distributed within the IZ. A sharp minimum of EMG amplitude is obtained when the electrodes are over or near the IZ. The signal amplitude, but not its global pattern, is affected by the averaging effect introduced by the scatter of the NMJs within the IZ and by the physical size of the electrodes. Fig. 2 shows the results of computer simulations [8] of a single MU (IZ 10 mm wide)  
145 whose MUAP is detected by a differential electrode pair with IED of 5 mm, 10 mm or 20 mm in different sets of simulations. MNF and CV (estimated from pairs of DD signals) are also shown. The detection point (center point between the two electrodes in the case of ARV and MNF; center point between the two DD detection systems used in the case of CV) is moved from the center of the IZ to the tendon junctions 60 mm away, in steps of 1 mm. Both point  
150 electrodes and rectangular electrodes are simulated.

The zone of stable EMG amplitude is relatively narrow, especially in the case of IED comparable with the fiber semilength, and 25-40 mm away from the IZ. Only in such a narrow region, estimation of MNF and CV is reliable.

In the case of interference signal generated by many MUs innervated in approximately the same  
155 location, the minimum is less sharp depending on the scatter of the IZs. Displacements of the electrodes of the order of 10 mm result in changes of amplitudes that depend on the width and spread of the IZs of the active MUs that are within the detection volume of the electrode system. Fig. 3 shows this concept for a pair of rectangular electrodes aligned with the fiber direction and depicts a more complex and realistic situation than Fig. 2, where three identical MUs, placed at the  
160 same depth next to each other, have the respective IZs shifted in space by 5 mm, 10 mm or 20 mm.

#### *Experimental data*

Fig. 4 shows an example of processing procedure applied to experimental signals from upper trapezius muscle (contraction level 80% MVC, IED = 8 mm, 6 subjects considered). IZ is



165 determined by visual analysis and the channels corresponding to IZ, the 3 distal channels and the 3  
proximal channels are selected (Fig 4A). ARV, MNF and CV are estimated for the 7 channels  
considered, for each of the 6 subjects (Fig. 4B on the left). These parameters are normalized (Fig.  
4B on the right). Thus, at this point, all data are aligned and are scaled in amplitude, so that data  
from different subjects can be compared. Statistical analysis is then performed on normalized and  
170 aligned data (Fig. 4C).

Non parametric ANOVA indicated statistically significant dependence of ARV and MNF from the  
detection point for all considered muscles (except for MNF of signals detected over the abductor  
pollicis brevis and ARV of signals detected over the abductor pollicis brevis, but only during high  
level of contractions). CV was statistically dependent on the detection point only for signals  
175 detected over the upper trapezius muscle. Dependence of parameters on channel position was  
statistically significant with  $p < 0.01$  in the case of all muscles except for abductor pollicis brevis and  
flexor pollicis brevis (for flexor pollicis brevis statistically significant dependence was found with  
 $p < 0.05$ ). Table 2 shows the results of the post-hoc test. The values of the variables ARV, MNF and  
CV estimated from signals detected in different channels were compared to the values estimated  
180 from the signals detected over the IZ. The minimum distance (in terms of number of channels) from  
the IZ for which the post-hoc test disclosed statistically significant difference is provided, for both  
proximal and distal direction. In general, it is less likely to obtain statistically significant differences  
when the values corresponding to the location of the IZ are compared to those obtained from signals  
recorded proximally (probably due to the presence of other proximal IZs or to the thicker fat layer  
185 usually found for proximal locations).

**Table 2: Electrode locations for which the differences, with respect to the values over the IZ, were found to be statistically significant (after the post hoc test) are reported (NS: not significant difference, N: number of subjects).**

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Muscle	Force levels (% MVC)	Minimum distance for statistical difference (p<.05)		
		ARV	MNF	CV
<i>Abductor pollicis brevis</i> (N=5)	10	3 channels (proximal)	NS	NS
	50	NS	NS	NS
<i>Flexor pollicis brevis</i> (N=5)	10	2 channels (proximal)	3 channels (distal)	NS
	50	2 channels (proximal)	3 channels	NS
<i>Biceps brachii</i> (N=8)	30	2 channels (distal)	3 channels (distal and proximal)	NS
	70	2 channels (distal)	3 channels (distal)	NS
<i>Upper trapezius</i> (N=6)	20	3 channels (distal or proximal)	2 channels (distal) 3 channels (proximal)	1 channel (distal) 2 channels (proximal)
	80	2 channels (distal or proximal)	3 channels (distal or proximal)	1 channel (distal) 2 channels (proximal)
<i>Vastus lateralis</i> (N=10)	90	2 channels (distal)	2 channels (distal)	NS
<i>Vastus medialis</i> (N=10)	90	2 channels (distal)	2 channels (distal)	NS

### Discussion and conclusions

A differential montage with both electrodes placed on one side of the IZ detects mostly unidirectionally propagating MUAPs. In this case, estimates of EMG amplitude, spectral variables

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and CV are less affected by minor electrode displacement as well as by the potentials propagating in the opposite direction. Multichannel montage (with  $N > 2$ ) must be used in this region if muscle fiber CV is to be estimated and if limited sensitivities of amplitude and spectral estimates with respect to electrode displacement are desired [9, 10, 13, 27].

200 Simulations of simple situations are provided in Fig. 2 and 3. Experimental data were investigated during isometric non fatiguing contractions, confirming the results obtained from the simple simulations considered. The superficial muscles of the hand, the limbs and the trunk listed in Table 1 and 2 were studied for which fibers can be considered rectilinear, parallel to each other and to the skin surface within a first approximation. Potentials propagating from IZ to tendons (with small  
205 distortion in shape) could be detected.

Fiber inclination with respect to the skin surface and other geometrical factors which vary from muscle to muscle, person to person [10, 23, 24], and for the same muscle during shortening [29] or fatigue conditions can strongly influence the detected signal. Furthermore, there are (relatively infrequent) cases of muscles with multiple IZs lacking an area of unidirectional propagation. In  
210 these cases (not shown) estimation of global CV and EMG amplitude may indeed be affected by large fluctuations in space and it may be necessary to resort to electrode arrays from which the individual MUAPs may be extracted, classified and analyzed [11, 14] (some fluctuation in CV estimation is also shown for the simple simulations shown in Fig. 3, where local maximum of CV estimation are found over each of the three simulated IZs). The increasing use of mathematical  
215 models [8, 13] and of electrode arrays and multichannel amplifiers [20, 21, 22, 28] at the clinical research level [2, 3, 5] is providing insight in these situations.

The basic principle depicted in Fig. 1 and 2 explains and confirms the clinical observation that the amplitude of the EMG signal detected with a pair of electrodes has a minimum over the IZ [2, 24]. Furthermore, Fig. 2 shows that reliable estimates of MNF and CV can be obtained only in regions  
220 where EMG amplitude is stable to small displacements along fiber direction. This result is also

supported by the experimental signals considered in this study. To compare data from different subjects, a normalization procedure was used which could possibly affect the specific results from experimental signals. In particular, the considered normalization forces the data of each subject to cover the full range 0-1, losing information about relative changes of the variables across channels.

225 Nevertheless, statistically significant dependence of EMG variables from the detection point was found for all the muscles under study.

When the MUs active in the detection volume have IZs scattered in space or when bipolar detection is obtained using large electrodes with large inter-electrode distance, the minimum of ARV of the signals detected over the IZ is less marked and may become a ripple when the MUs have widely  
230 scattered IZs making estimates of MNF very sensitive to electrode position and estimates of CV impossible. The scatter of IZs may be different in different muscles and, for the same muscle, in different subjects. This is the reason why such a minimum has been detected by some authors and not by others. In any case, the general rule of avoiding the IZ for placement of an electrode pair should be followed. Of course, as indicated by Campanini et al [3], the criticality of the issue  
235 depends on the application but, nevertheless, it is a good practice to follow the general rule. In dynamic conditions the IZ zone moves under the skin and if it reaches the electrodes it may cause changes of EMG amplitude that are easily misinterpreted as changes of muscle activation level [15, 18, 29]. In dynamic conditions it is therefore important to verify that the IZ does not approach the detection electrodes during the entire range of movement.

240 The growing use of electrode arrays creates a need for both continuing education of clinical users and for the definition of standards. The need for a standardization effort, continuing the endeavors of Project SENIAM, is perceived worldwide. A recent meeting held at the Worcester Polytechnic Institute (Massachusetts, USA) concerning multichannel detection, decomposition and interpretation of EMG constitutes a first effort in this direction [[www.emglab.stanford.edu](http://www.emglab.stanford.edu)] that  
245 should be supported and expanded.

Some authors did not recognize this problem [1] since large electrodes were adopted in their work. Such electrodes introduce low pass filtering and spatial averaging resulting in limitations in the quality of the estimated EMG variables. For these reasons we strongly suggest to avoid electrode placing over the IZ and locating them in between the IZ and tendon zones.

250 The individual variations of limb and muscle anatomy require efforts to minimize the resulting variability of EMG features. It is foreseen that surface EMG computer assisted systems may soon detect EMG from two dimensional arrays (High Density EMG) and automatically identify the electrode pairs or groups that are most suitable as a source of information concerning the location of IZs, the anatomy of individual active MUs, the fiber direction and the features of surface EMG that  
255 are relevant for meaningful physio-pathological observations and conclusions.

In conclusion, our results show that when a single electrode pair is used the IED must be small with respect to the distance between the IZ and the tendon and neither electrode of the pair should be over the IZ for the entire range of the movement. If this is not the case, EMG variables may be highly affected by small geometrical changes.

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## Figure captions

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### Fig. 1

Surface EMG signal generated by a single muscle fiber. Two spatial distributions of potentials, each associated to a depolarized zone, propagate from the neuromuscular junction to the tendon endings. Two electrodes placed symmetrically over the neuromuscular junction detect identical signals whose difference is zero, while two electrodes placed on one side of the junction detect a non zero differential signal.  $V(x)$  is the transmembrane potential. IED is the interelectrode distance, the detection point is the center point of the electrode pair.  $V_1(t)$  and  $V_2(t)$  are the outputs of the differential amplifiers.

360 **Fig. 2**

A) Simulation of a motor unit action potential detected by point or rectangular (20x5 mm) electrodes for different interelectrode distances (IED) and detection point locations. B) Average rectified value (ARV) and C) mean frequency (MNF) of the single differential (SD) EMG signal detected between two electrodes (5 mm, 10 mm or 20 mm apart) moved over the surface of the skin above a single 120 mm long motor unit in steps of 1 mm. D) Muscle fiber conduction velocity (CV) estimated from a pair of double differential (DD) signals (with IED 5 mm, 10 mm or 20 mm) moved over the surface of the skin above the same motor unit in steps of 1 mm. The neuromuscular junctions of the individual fibers are uniformly distributed over the innervation zone (IZ) which is 10 mm wide. For IED = 20 mm, the length of the four electrode array required to generate the two DD signals for CV estimation equals the distance between the IZ and the tendon.

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### Fig. 3

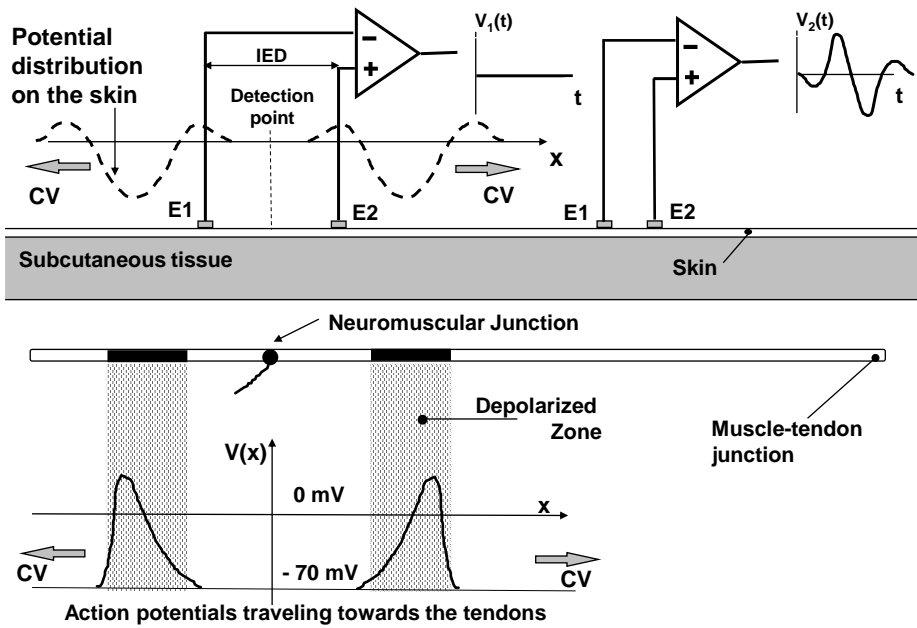
A) Geometry of three identical simulated motor units with different IZs. B) Geometry of the electrode pair (rectangular electrodes, 20x5 mm, IED = 20 mm) and of the IZ. C) Examples of

375 simulated raw signals for different locations of the detection point. D) Plots of ARV, MNF and CV  
of the EMG for different spread of the three IZ (each is 10 mm wide) and different locations of the  
detection point from one tendon to the other.

**Fig. 4**

380 A) Example of experimental signals recorded during the contraction at 80% MVC of the upper  
trapezius muscle. IZ determined by visual analysis and the channels selected for subsequent  
processing are indicated. B) ARV, MNF and CV (dimensional and normalized values) estimated for  
the 7 channels considered, for each of the 6 subjects. C) Statistical analysis performed on the data  
normalized with respect to the minimal and maximal values and scaled in the range 0-1. Similar  
385 results were obtained from other muscles (abductor pollicis brevis, flexor pollicis brevis, biceps  
brachii, vastus medialis, vastus lateralis).

Fig 1



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Fig 2

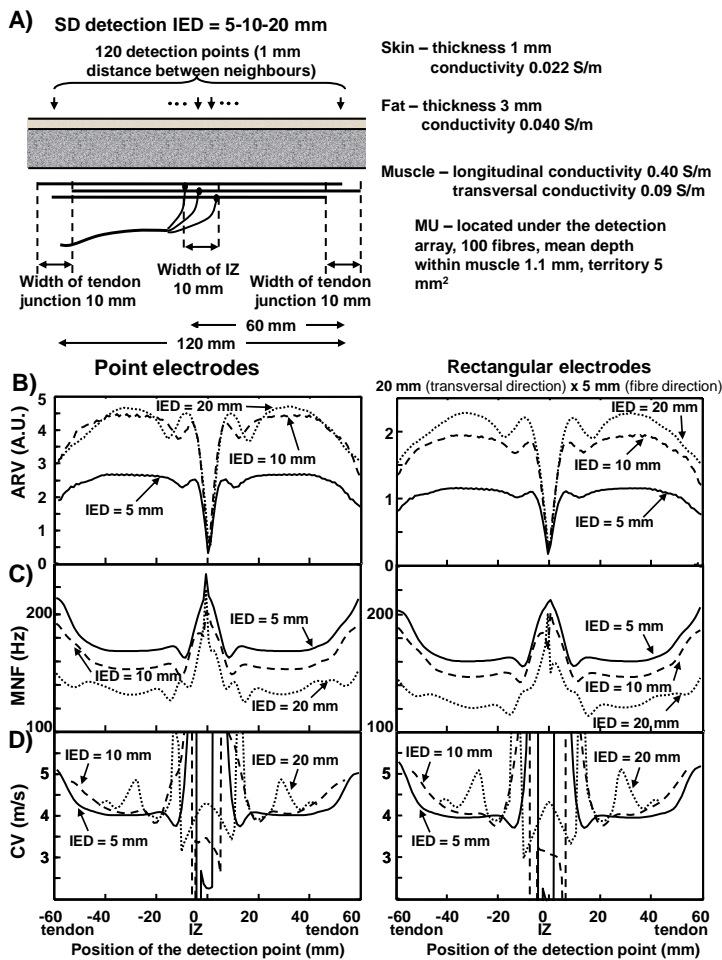
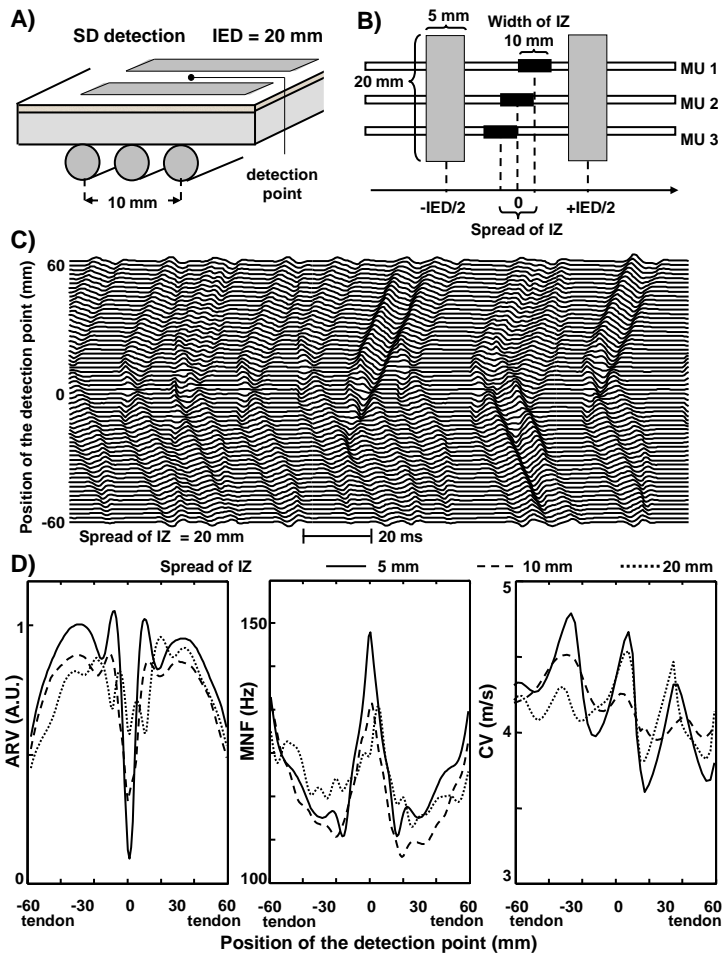


Fig 3



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Fig 4

