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Amplitude indicators and spatial aliasing in high density surface electromyography recordings

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ABSTRACT

Average Rectified Value (ARV) and Root Mean Square (RMS) are amplitude indicators commonly used in the field of EMG either in time or space. These two indicators are compared (a) analytically for a one dimensional sinusoid, sum of sinusoids, two dimensional sinusoids, and (b) numerically by simulating a high density detection system, sampling in space the distribution of propagating surface action potentials generated by a muscle motor unit (MU). For any signal sampled above the Nyquist frequency the estimated RMS does not depend on the sampling rate while the estimated ARV does. The surface potential is often sampled in space below the Nyquist frequency, by high density surface EMG detection systems (HDsEMG), generating aliasing in space. For point-like electrodes, the lowest spatial sampling frequency corresponding to the largest inter-electrode distance (IED), which avoids spatial aliasing for a simulated MU action potential, is 100 samples/m (IED = 10 mm). Therefore, IEDs below this value are recommended for measurements of EMG image features. From the theoretical point of view, the spatial RMS of sEMG images is more robust than the ARV with respect to the IED and should be preferred.

1. Introduction

Surface EMG (sEMG) signals are the algebraic summation of the motor unit action potentials (MUAP), occurring within the "detection volume" of the electrodes. The amplitude of a sEMG signal is usually estimated by the average rectified value (ARV) or the root mean square (RMS) of the signal. When a 2D electrode array of MxN channels is used, the spatial map of MxN ARV or RMS values, each estimated over the K time samples of a given time window (epoch), gives the distribution of amplitude in space during that time window and can be considered as one 2D sample (frame) in time. Maps obtained from subsequent epochs provide a movie describing the time evolution of the map (see Fig. 1). A single ARV or RMS value over MxN spatial samples and K time samples may be obtained by summation first in time and then in space or vice versa.

Since EMG signals are band-limited in both space and time, reconstruction becomes possible if the sampling process satisfies the Nyquist criteria. However, this may not always be true [1] and, in such cases, signal reconstruction becomes a non-trivial problem. The estimated ARV and RMS equal the true ARV and RMS when the signal is sampled above the Nyquist rate and the proper reconstruction filter (a sinc function) is used. These concepts are also relevant for the estimation of geometrical features of the distribution, such as centroid, moments of the spatial distribution and regions of interest (segmentation).

EMG amplitude has been studied for several years [2], mostly by analog rectification and low-pass filtering, to estimate the EMG linear envelope. Modern systems sample the signals and use ARV or RMS indicators. For a Gaussian distributed amplitude, these indicators are given by RMS = σ and ARV = $\sigma\sqrt{2/\pi} = \mathrm{RMS}\sqrt{2/\pi}$, where σ is the standard deviation of the distribution. ARV and RMS have been used in EMG-based force estimation [3–6], study of fatigue [7,8], muscle activity distribution [8,9], crosstalk studies [10,11], and many other applications.

Some researchers have applied techniques such as whitening [12,13] or high pass filtering [14] to reduce the variance of the EMG amplitude estimates in force estimation applications.

In this work we address the following questions: (1) Are there reasons to prefer the ARV or the RMS value considering non-Gaussian distribution of EMG? (2) What should the spatial sampling frequency be (samples/m or pixels/m) in order to avoid spatial aliasing and what would be the consequences of under sampling on estimated EMG features?

The issue is addressed from a general point of view and not for the special case of signals propagating along the fiber direction (z), where $z = z_0 \pm vt$ and high sampling rate in t implies high sampling

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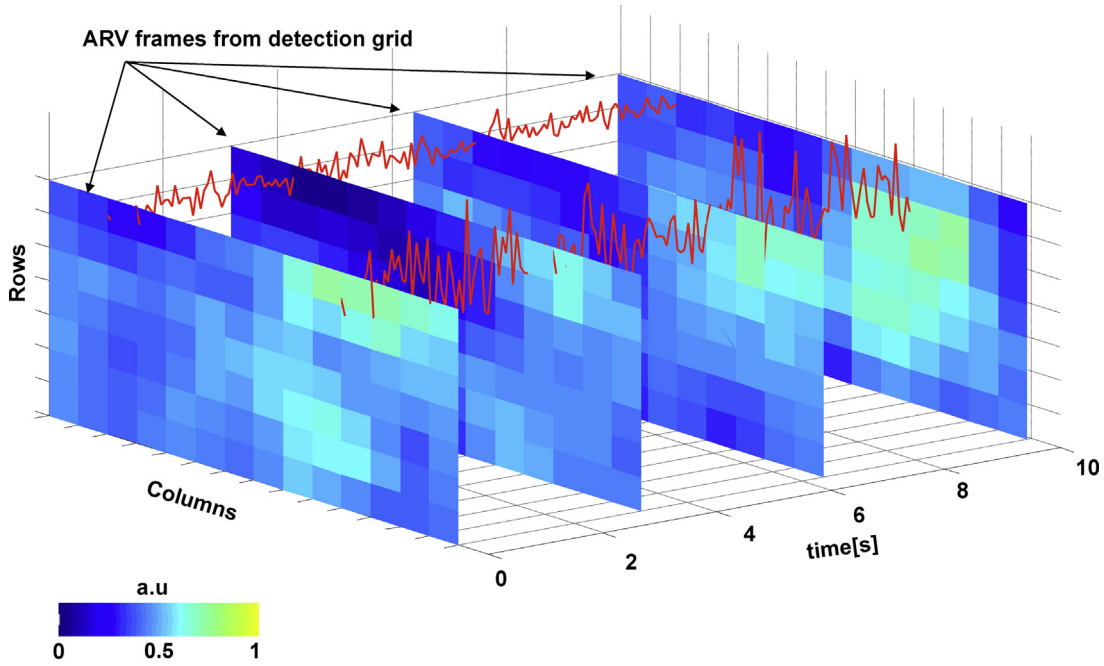

Fig. 1. Concept of the evolution of sEMG images in time and space. The intensity of each pixel in the grid is the average rectified value (ARV) over a 250 ms epoch window. The first frame shows the ARV map obtained from the first epoch (from 0 to 250 ms). The amplitude maps (four ARV frames out of 40) are shown for 10 s.

rate in *z*. ARV and RMS indicators are compared for one dimensional (1D) single sinusoid, 1D sum of sinusoids, 2D sinusoids, simulated action potential signals in space (1D and 2D), and recorded sEMG signals using a custom made electrode array with 1 mm inter electrode distance. The issue of sampling in space and time, a little above or below the Nyquist frequency, spatial aliasing and inter electrode distance are also discussed for a simulated single fiber, single motor unit and real sEMG signals.

2. Materials and methods

Any signal satisfying Dirichlet conditions [15] can be described as a finite summation of sine and cosine functions in the Fourier series plus residual error. We start from the case of a sinusoid in space or time and evolve toward simulated motor unit action potentials sampled over the skin by a two dimensional detection grid.

Aliasing is introduced by sampling, when the signal reconstructed by convolution of the sample train with a sinc function [16] is different from the original continuous signal. To study spatial aliasing, the monopolar potential distribution generated by the propagating single muscle fiber or motor unit action potential was simulated using a previously developed model [17]. The fiber(s) was (were) parallel to the skin and placed at different depths in the muscle. Table 1 shows the general muscle model parameters. Tables 2 and 3 show specific setting parameters for simulating either single fiber or the motor unit. In the simulated signals a non-homogeneous, layered, anisotropic volume conductor, constituted by muscle (anisotropic), fat (isotropic), and skin (isotropic) layers was considered.

In order to study the effect of inter electrode distance (IED) on ARV and RMS on experimental signals, we recorded sEMG signals with an electrode array with IED = 1 mm. By down sampling in space, higher IEDs were obtained. A linear electrode array including 63 electrodes with 1 mm IED was applied to the Biceps Brachii of a young healthy subject. Signals were collected in Monopolar configuration from the upper portion (proximal with respect to the innervation zone) of the long head of the Biceps Brachii when the

Table 1
General EMG model parameters for simulating both the single fiber and the motor-unit using model developed by Farina and Merletti [17]; σ is the conductivity; the rows and columns of the detection grid are considered in "*z*" and "*x*" direction respectively; the fiber depth changes along the "*y*" direction.

Parameters	Value
Skin layer thickness	1 mm
Fat layer thickness	3 mm
Conductivity ratio between skin and fat layers	20
Conductivity ratio between fat and muscle layers ($\sigma_{fat}/\sigma_{muscle-y}$)	0.5
Muscle anisotropy ratio ($\sigma_{muscle-z}/\sigma_{muscle-x}$)	5
Electrode dimension	Point like
Spatial filter	Monopolar
Inter electrode distance (IED)	1–20 mm
Area of the skin covered by the detection grid	128×128 mm^2
Center of the detection system (x, y, z)	(64, 0, 64)

Table 2
EMG model parameters for simulating the single fiber (see Table 1).

Parameters	Value (mm)
Depth in the muscle	0–10
Fiber length	125
Upper semi fiber length	65
Neuromuscular junction location along fiber direction (z axis)	0
Lower semi fiber length	60

Table 3
EMG model parameters setting for simulating the motor unit (see Table 1).

Parameters	Value
Mean depth of the fibers in the muscle	15.5 mm
Radius of the motor unit territory (MUT)	15 mm
Number of fibers (uniformly distributed in the MUT)	150
Fiber length	125 mm
Lower semi fiber length	60 mm
Upper semi fiber length	65 mm
Spread of innervation zone (IZ) (uniformly distributed in the MUT)	10 mm
Spread of lower tendon region (uniformly distributed in the MUT)	8 mm
Spread of upper tendon region (uniformly distributed in the MUT)	10 mm