Analysis of the Spatial Resolution of Body-Surface Dominant-Frequency Mapping Systems

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Abstract

In this study we assess the ability of high-density body-surface lead systems to resolve local dominant-frequency (DF) values. In a detailed numerical model of the human thorax, we calculate the measurement sensitivity distribution (MSD) of the 117 unipolar leads of a Dalhousie lead system. Based on the MSD, we compute the lead equivalent volume (LEV) of each unipolar lead and use it to quantify the lead spatial resolution (SR). Anterior leads positioned in columns 4-6 and rows 3-5 of the Dalhousie system have the lowest LEV values ($2.4\% < LEV < 7\%$) and concentrate their sensitivity mostly in the right myocardium. Higher LEV values are achieved at posterior locations ($LEV > 11\%$). Optimal lead positioning can increase the resolution of local DF values. However, the ability to resolve local DF values in selected regions is limited by the intrinsic MSD of body-surface leads. Our results indicate that further signal processing stages are needed to improve the SR of body-surface DF maps.

1. Introduction

Cardiac fibrillation is a complex cardiac arrhythmia whose mechanisms are not well understood. In the past years, intracardiac dominant-frequency (DF) mapping has enhanced the understanding of the mechanisms of cardiac fibrillation and has been used as a guide for atrial fibrillation (AF) ablation therapies [1]. Body-surface DF techniques have also been developed to investigate the dynamics of cardiac fibrillation in a continuous and noninvasive manner. Richter et al. [2] showed that the DF of electrocardiogram (ECG) lead V_1 can be used to quantify AF organization. Also, Hsu et al. [3] observed that the DF of the ECG lead V_1 correlated with the DF of the anterolateral right atrium (RA) free wall, and Dibs et al. [4] reported a high correlation between the DF at the RA and at ECG lead V_2, and between the DF at the left atrium (LA) and at ECG leads V_4 and V_6.

The correlation between intracardiac and body-surface DF values during AF suggests that body-surface DF maps may be used as surrogates for intracardiac DF maps. Nevertheless, the detailed correspondence between intracardiac and body-surface DF values remains to date unclear. It is assumed that intracardiac and body-surface leads have different spatial resolution (SR) values. Due to their proximity to the myocardium, intracardiac leads mostly measure local activity. By contrast, body-surface leads measure global activity, since they are located further away from the myocardium. This discrepancy between intracardiac and body-surface SR values could potentially prevent the existence of a ono-te-one correspondence between intracardiac and body-surface DF values. The reasons are twofold. Firstly, global body-surface DF values would not reflect local DF values, but rather an average over the SR [5]. And secondly, even in the case of identical local DF values, the lead configuration could produce distortions in the spectrum of measured signals that depend on the lead SR [6, 7]. Consequently, in order to improve our understanding of the relationship between intracardiac and body-surface DF maps, and assess whether the latter can be used as surrogates of the former, it is of importance to quantify the SR of body-surface lead systems.

The measurement sensitivity distribution (MSD), also known as lead field, has been used to study the properties of different lead systems. In a classical work by Rush and Driscoll, the MSD of electroencephalographic (EEG) leads was analyzed [8]. Arzbaecher et al. investigated the sensitivity at the heart of unipolar precordial leads, and unipolar and bipolar esophageal leads [9]. The MSD has also been used to estimate the lead SR. In [10], the half sensitivity volume was proposed to compare the SR of EEG and magnetoencephalographic systems. In [11], the region of interest sensitivity ratio was defined as the ratio between the average sensitivities of two regions, and used to quantify the specificity of EEG measurements. Finally, the resolution volume was proposed for quantifying the SR of body-surface leads [12] and implantable defibrillator leads [13].

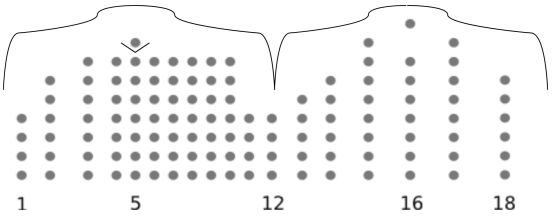

Figure 1. Electrode sites in the Dalhousie system.

In this study we investigate the ability of body-surface lead systems to resolve local DF values. Based on the MSD, we introduce a novel method for quantifying the lead SR called the lead equivalent volume (LEV), which corresponds to the fraction of the myocardium that contributes the most to the measurements. Then, we compute the LEV of a high-density body-surface lead system to quantify and analyze the SR of each lead.

2. Methods

2.1. Body-surface lead system

The high-density body-surface lead system included in this study is based on the Dalhousie lead system [14]. The Dalhousie lead system consists of 117 electrodes sites arranged in 18 vertical strips which are located according to precisely defined landmarks (Figure 1). In the resulting system, strip 1 is located in the right mid-axillary line, strip 12 in the left mid-axillary line, strips 2-11 are located on the anterior torso, and finally strips 13-18 are located posteriorly. Wilson's central terminal, obtained by including three additional limb electrode sites, is used as the reference for obtaining unipolar signals from each electrode.

2.2. The measurement sensitivity distribution

The MSD describes the ability of a lead system to measure bioelectric sources that are distributed throughout a volume conductor [15]. Let $\mathbf{J}(t, v)$ denote a distribution of dipoles in a volume V and $\mathbf{L}(v)$ denote the MSD of a lead system in V, where $v \in V$ and t denotes time. According to the lead field theory, the signal $x(t)$ that is measured by the lead system is expressed as follows:

$$x(t) = \int_V \mathbf{L}(v) \cdot \mathbf{J}(t, v) dv \tag{1}$$

where $(\cdot)$ denotes the dot product. Consequently, dipoles originated in a region where the MSD is high will have a larger effect on the measured signal than dipoles originated in a region where the MSD is low.

2.3. The lead equivalent volume

The LEV is used to quantify the lead SR. Given the MSD of a lead in a volume source V_M, the LEV is defined as follows. Let $L_n(v)$ be the normalized magnitude of the MSD,

$$L_n(v) = \frac{\|\mathbf{L}(v)\|}{\max \|\mathbf{L}(v)\|} \tag{2}$$

The normalized magnitude $L_n(v)$ is a scalar field whose maximum is 1. This value is reached where the lead sensitivity magnitude $\|\mathbf{L}(v)\|$ reaches its global maximum. Also it is worth noting that when the MSD is uniform across V_M, $L_n(v) \approx 1$ everywhere; on the other hand, if the MSD is concentrated within a small region of V_M, $L_n(v) \approx 0$ everywhere except for the region where the MSD is concentrated, in which $L_n(v) \approx 1$.

Based on $L_n(v)$, the volume V_E is the defined as the integral

$$V_E = \int_V L_n(v) dv \tag{3}$$

The volume V_E can be interpreted as the size of the myocardial region that contributes the most to the measured signals. Since $0 \leq L_n(v) \leq 1$, it follows that $0 \leq V_E \leq V_M$. Moreover, in the case of leads with a uniform sensitivity $V_E \approx V_M$, whereas for leads with a concentrated sensitivity $V_E \approx 0$. Therefore, the volume V_E describes the ability of a lead system to concentrate its MSD within a small region of V_M.

Since volume sources can have different sizes, it is convenient to normalize V_E to V_M. The resulting quantity is defined as the LEV

$$LEV = \frac{V_E}{V_M} = \frac{\int_V L_n(v) dv}{\int_V dv} \tag{4}$$

The LEV is a dimensionless quantity and by definition $0 \leq LEV \leq 1$. Furthermore, when the MSD is uniform across V_M, $LVE \approx 1$. On the other hand, when the MSD is highly localized, $LVE \approx 0$.

Finally, based on the LEV we define the SR of a lead system as

$$SR = \frac{1}{LEV} \tag{5}$$

It follows easily that $1 \leq SR \leq \infty$, with $SR = 1$ for uniform MSD, and $SR = \infty$ for a highly localized MSD.

2.4. Human thorax model

The MSD of each body-surface lead defined in Section 2.1 was calculated in the ventricular myocardium of a realistic 3D model of the human thorax. This model was constructed based on the Visible Human Man (VHM) dataset [16] [17] and consisted of 95 slices of the human thorax with a resolution of 1.67mm×1.67mm×1.67mm in the

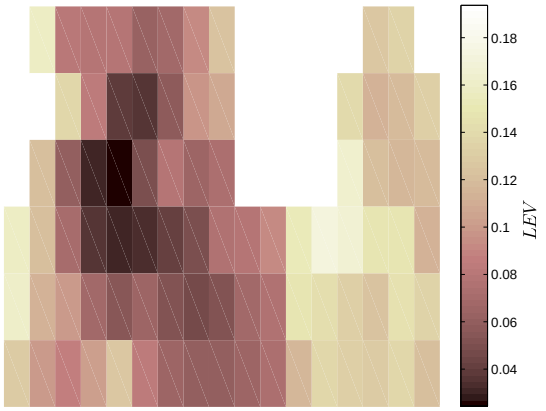

Figure 2. Map of the LEV values in the numerical model of the Dalhousie lead system.

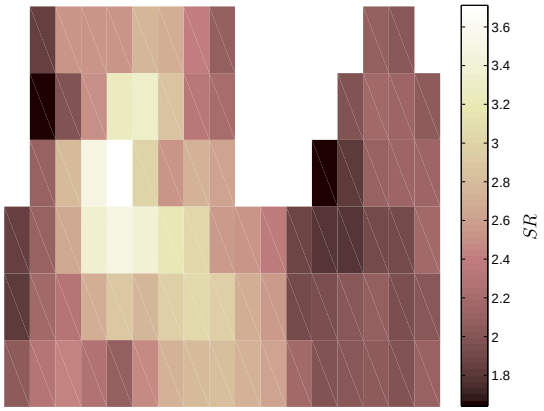

Figure 3. Map of the SR values in the numerical model of the Dalhousie lead system.

heart and 1.67mm$\times$1.67mm$\times$4mm in the rest of the thorax. During segmentation, 20 different organs and tissues were identified and assigned resistivity values according to [18]. This process resulted in a cubic resistive grid which was used to calculate the MSD by invoking the principle of reciprocity [15] and by implementing numerical methods previously discussed in [19] and [20].

3. Results

Our results show that anterior leads have lower LEV values than posterior leads; in other words, the fraction of the myocardium that contributes the most to body-surface measurements is lower when leads are located in the anterior torso than when they are located in the posterior torso (Figure 3). Accordingly, the SR is higher in the anterior torso than in the posterior torso (Figure 3). This observation agrees with the usual assumption that the closer the electrodes to the volume source, the more concentrated the MSD.

As expected, the lowest LEV values are achieved in the proximity of the myocardium, specifically in columns 4-6 and rows 3-5 of the Dalhousie system. In this region, LEV values of $2.4\% < \text{LEV} < 7\%$ are observed. By contrast, larger LEV values are observed at the back, where LEV $> 11\%$. The largest LEV values are achieved in the left posterior axillary line, where LEV $> 15\%$. As a consequence, leads located in the posterior torso can have LEV values up to 8 times larger than leads located in the anterior torso. Finally, the MSD of leads located anteriorly is concentrated mostly in the right myocardium, whereas the MSD of leads located posteriorly is concentrated mostly in the left myocardium.

4. Discussion and conclusions

Noninvasive diagnostic methods for cardiac fibrillation can benefit from understanding the correspondence between intracardiac and body-surface DF maps. Experimentally, a correlation between DF values at precordial leads and intracardiac DF values during AF have been previously reported in the literature [3, 4]. However, the detailed relationship between intracardiac and body-surface DF values remains unclear.

One potential obstacle preventing a one-to-one correspondence between intracardiac and body-surface DF values is the theoretical difference between the SR of intracardiac and body-surface leads. Intracardiac leas are assumed to measure local bioelectric phenomena, whereas body-surface leads are assumed to measure global bioelectric phenomena. It is believed that body-surface DF values correspond to either the mean or the most common firing rate within the myocardial region that contributes the most to the measurement [5]. However, this view could be inaccurate for two reasons. Firstly, DF values are extracted from the power spectrum of cardiac signals and hence, since the power spectrum is a non-linear operation, DF values from global measurements will not in general be equivalent to the average DF values from the corresponding local measurements. Secondly, it has been shown that the power spectrum of cardiac signals can be distorted by the intrinsic SR of the lead system [6, 7]. In other words, the power spectrum of signals from different lead systems will in general be different and hence, the corresponding DF values may be different too.

In this study we have quantified the SR of a high-density body-surface lead system based on the notion of the LEV. Our results show that body-surface unipolar leads can have large LEV values. Therefore, the possibility of obtaining a one-to-one correspondence between intracardiac and

body-surface DF values could be prevented by the practical impossibility of measuring local values by means of body-surface leads. We conclude that further signal processing stages are needed in order to estimate local DF values by using body-surface lead systems.

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