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Estimation of the Scope of Transvenous Lead Systems in Implantable Cardioverter Defibrillators

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Abstract— **The analysis of intracardiac Electrograms (EGM) recorded by transvenous lead systems in Implantable Cardioverter Defibrillators (ICD) often entails assumptions on the scope of the lead system. Based on bioelectric signal modeling and on numerical analysis, we studied quantitatively the scope of unipolar and bipolar lead configurations in ICD. We defined the scope in terms of the Mean Square Difference (MSD) between EGM generated by the whole myocardium, and EGM generated by different families of regions within the myocardium. For unipolar and bipolar lead systems, simulations showed that the smallest myocardial region involving a given value of MSD is characterized by the highest measurement sensitivity. Furthermore, the scope in the ventricles was found to be an order of magnitude smaller for bipolar leads than for unipolar leads. Bioelectric signal modeling combined with numerical analysis constitutes a powerful method to study quantitatively the scope of transvenous lead systems.**

I. INTRODUCTION

The Implantable Cardioverter Defibrillator (ICD) is the treatment of choice for patients at high risk of Sudden Cardiac Death (SCD). Modern ICD record intracardiac electrograms (EGM) by means of transvenous lead systems [1]. By implementing signal processing algorithms, modern ICD analyze intracardiac EGM in order to detect life-threatening arrhythmias and deliver an appropriate electric therapy to restore the normal rhythm [2]. It is often assumed that the quality of the information conveyed in intracardiac EGM relies on the scope of the lead configuration [3] [4] [5] [6] [7]. The assessment of the scope of transvenous lead systems remains largely qualitative, and it is based on the application of the principles of Lead Field Theory [8] to unipolar and bipolar lead configurations in an infinite and homogeneous conductor medium. As a consequence, it is widely accepted that for transvenous lead systems unipolar EGM capture global cardiac activity, while bipolar EGM capture local cardiac activity.

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In this paper we analyze quantitatively the scope of transvenous ICD lead systems based on bioelectric signal modeling and numerical methods. We study the scope of unipolar and bipolar lead configurations in terms of the Mean Square Difference (MSD) between EGM signals generated by the whole myocardium, and EGM signals generated by regions within the myocardium. For a given level α, the scope is defined as the region of minimum size involving a MSD smaller than α. Central to this study is the calculation of the lead field of unipolar and bipolar configurations in a realistic 3D human thorax. The lead field describes the ability of the lead system to measure electric sources distributed throughout the myocardium. By combining the lead field with a model of cardiac electrical activity, we are able to simulate numerically EGM signals generated by different myocardial regions, and estimate their MSD. By studying the MSD of different families of myocardial regions, we address the problem of determining the region of minimum size for every level α.

This paper is organized as follows. In Section II we introduce the model of intracardiac EGM signals and describe the numerical procedure to analyze the scope of lead systems based on MSD. In Section III we study the MSD of different families of myocardial regions for unipolar and bipolar lead configurations as a function of its size. Finally, in Section IV we discuss the limitations of this approach and we propose future applications.

II. METHODS

A. Modeling Intracardiac EGM

Lead Field Theory provides us with a convenient formulation of signals recorded by lead systems in terms of the measurement sensitivity distribution [8]. Let V be the whole myocardium. According to Lead Field Theory, the relationship between cardiac signal $z[n]$ recorded at electrodes sites and the distribution of cardiac currents $\overline{J}_i[n]$, generated in the myocardiom V, is:

$$z[n] = \sum_V \frac{1}{\sigma} \overline{J}_{LE} \cdot \overline{J}_i[n] \ , \qquad (1)$$

where σ is the conductivity of the medium and $\overline{J}_{LE}$ is the lead field associated to the lead system. Therefore, lead field $\overline{J}_{LE}$ corresponds to the measurement sensitivity distribution of the lead system.

The lead field can be obtained by measuring in the conducting volume the gradient of the potential field generated

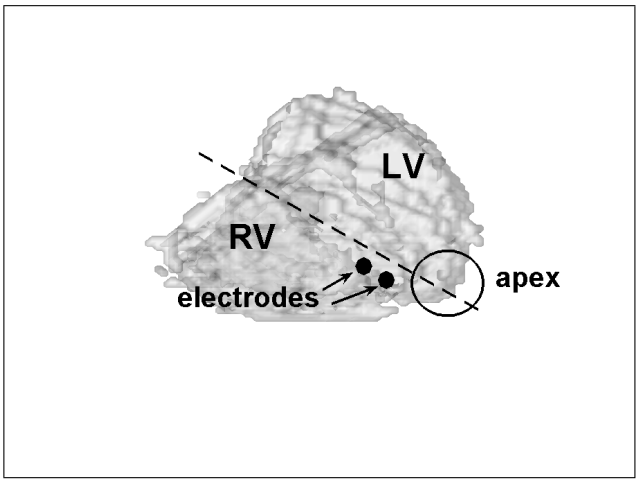

Fig. 1. Superior view of the ventricular myocardial mass as described by the VHM dataset. Right ventricle (RM) and left ventricle (LV) are, respectively, the regions below and above the dashed line.

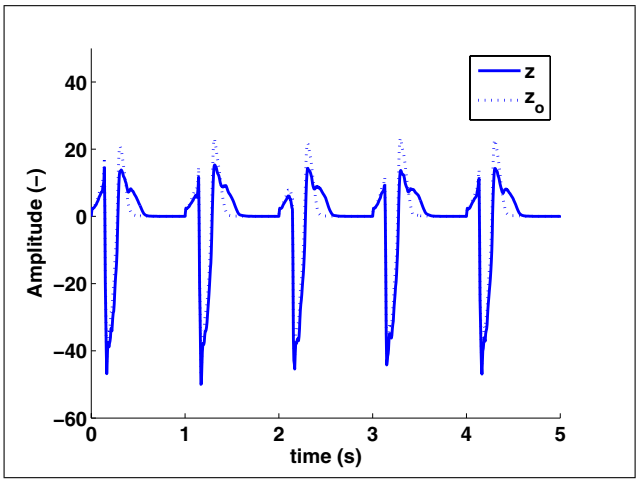

Fig. 2. Unipolar EGM generated by the whole myocardium V, $z[n]$, and by a region V_o within the myocardim, $z_o[n]$, when simulating the stimulation of the apex at 60 beats per minute during 5 seconds.

when the electrodes are energized reciprocally with one unit of current. We described the conducting properties of the human body based on Finite Difference Methods (FDM) featuring the realistic 3D male thorax of the Visible Human Man dataset (VHM) [9]. In the model implemented here, 20 different organs and tissue types were identified, and the whole volume was characterized by a resolution of $1.67mm \times 1.67mm \times 4mm$ in the heart and $1.67mm \times 1.67mm \times 8mm$ elsewhere. Based on this model of the human body, we calculated the lead fields for unipolar and bipolar configurations. The unipolar configuration was modeled as one point electrode located in the right ventricle (RV) and one can in left subpectoral position, while the bipolar configuration was modeled as two point electrodes in the RV (Fig 1).

B. Definition of Scope

The contribution of a myocardial region V_o, such that $V_o \subseteq V$, can be quantified by determining the similarity in mean power between signals generated by V and by V_o (Fig 2). From (1), if V_c is the complementary myocardial region of V_o with respect to V, i.e., if V_o and V_c are two disjoint myocardial regions and $V = V_o \cup V_c$, signal $z[n]$ recorded by the lead system can be expressed as:

$$\begin{aligned} z[n] &= \sum_{V} \frac{1}{\sigma}\overline{J}_{LE} \cdot \overline{J}_i[n] = \\ &= \sum_{V_o} \frac{1}{\sigma}\overline{J}_{LE} \cdot \overline{J}_i[n] + \sum_{V_c} \frac{1}{\sigma}\overline{J}_{LE} \cdot \overline{J}_i[n] = \\ &= z_o[n] + z_c[n] \ , \end{aligned} \tag{2}$$

where $z_o[n]$ and $z_c[n]$ are the signals generated by V_o and V_c, respectively. The similarity between $z[n]$ and $z_o[n]$ is defined in terms of their Mean Square Difference (MSD):

$$\begin{aligned} MSD\{V_o\} &= E\left[(z[n] - z_o[n])^2\right] = \\ &= E\left[(z_c[n])^2\right] = \\ &= P_{z_c z_c} \ , \end{aligned} \tag{3}$$

where E denotes statistical mean. Thus, $MSD\{V_o\}$ is the error of approximating the signal $z[n]$ to $z_o[n]$. If we divide $MSD\{V_o\}$ by the mean power of $z[n]$, $P_{zz} = E\left[(z[n])^2\right]$, we can obtain a more convenient measure of similarity between $z[n]$ and $z_o[n]$ that is relative to the mean power of $z[n]$:

$$\begin{aligned} MSD_n\{V_o\} &= \frac{MSD\{V_o\}}{P_{zz}} = \\ &= \frac{P_{z_c z_c}}{P_{zz}} \ . \end{aligned} \tag{4}$$

We used (4) as a basis for investigating the scope of transvenous ICD lead systems in the ventricles. For a family of N regions within the ventricular myocardium, $\{V_o\}_N$, we studied numerically MSD_n for every region V_o. Subsequently, we calculated numerically its size. Fixing a level α of acceptable MSD, we defined the scope as the region of minimum size that involves an MSD_n smaller than α.

C. Numerical Analysis of the Scope

In order to estimate the contribution of a myocardial region V_o based on (4), a statistical description of cardiac current sources $\overline{J}_i[n]$ is needed. We implemented a model of electric activity of cardiac tissue previously defined [10] to generate cardiac current sources by means of numerical simulations. This model is based on the restitution properties of cardiac tissue, and is able to reproduce complex behavior, such as the restitution of Action Potential Duration (APD) and Conduction Velocity (CV), and curvature effects.

We simulated the stimulation of the apex at 60 beats per minute during 2 seconds. For every myocardial region V_o,

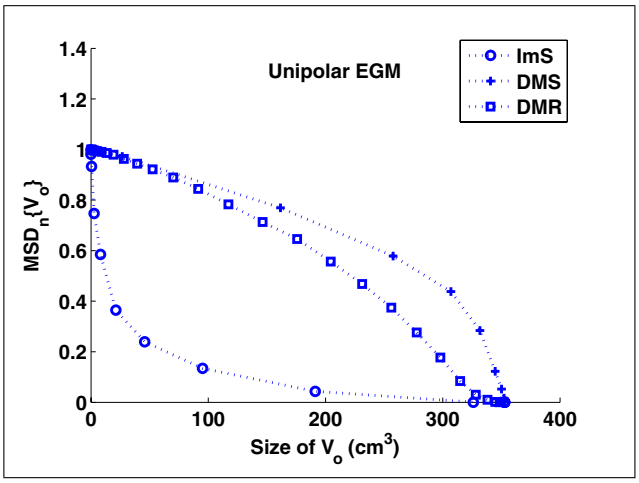

Fig. 3. $MSD_n\{V_o\}$ as a function of the size of V_o for families of myocardial regions ImS, DMS and DMR for unipolar EGM.

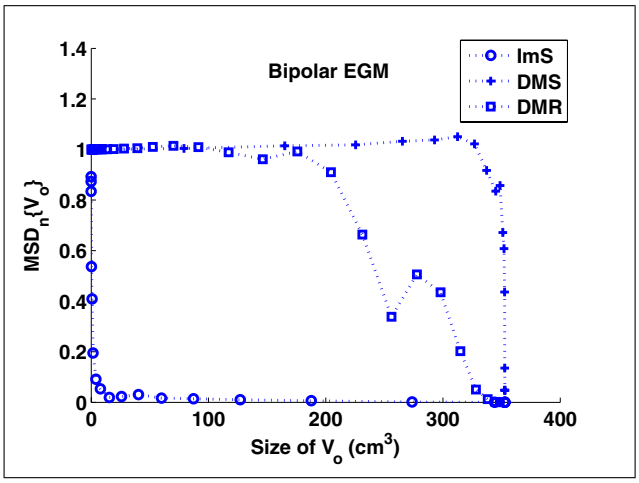

Fig. 4. $MSD_n\{V_o\}$ as a function of the size of V_o for families of myocardial regions ImS, DMS and DMR for bipolar EGM.

simulated cardiac current sources were combined with the lead field according to (2) for synthesizing signals $z[n]$ and $z_o[n]$ for both unipolar and bipolar leads. Then, we replaced statistical means by time averages to estimate $MSD_n\{V_o\}$. We studied $MSD_n\{V_o\}$ as a function of the size of V_o. Three families of myocardial regions were investigated. They consisted of the whole ventricular myocardium mass V and progressively smaller nested myocardial regions, $V_{n+1} \subseteq V_n$, defined according to the following rules:

- *Increasing minimum sensitivity* (ImS). The surface of V_n of minimum sensitivity was excluded for consecutive region V_{n+1}. Therefore, minimum sensitivity was progressively higher for higher values of n.
- *Decreasing maximum sensitivity* (DMS). The surface of V_n of maximum sensitivity was excluded for consecutive region V_{n+1}. Therefore, maximum sensitivity was progressively smaller for higher values of n.
- *Decreasing maximum radius* (DMR). The surface of V_n containing the most distant points to the center of mass of V was excluded for consecutive region V_{n+1}. Therefore, maximum distance to the center of mass of V was progressively smaller for higher values of n.

III. Results

Simulations showed that, in general, for both unipolar and bipolar lead systems, and for families of myocardial regions ImS, DMS and DMR, an increase in the size of V_o involves a decrease of $MSD_n\{V_o\}$ (Fig 3 and Fig 4). Indeed, according to (2), the larger V_o, the closer it is to V and, therefore, the more similar z_o is to z. Further, the slope of $MSD_n\{V_o\}$ curves can be interpreted as the rate of convergence of z_o to z, when V_o tends to V. The rate of convergence differed from one family to another. For family ImS the convergence was found to be the fastest, while for family DMS the convergence was found to be the slowest. This observation can be explained by recalling that myocardial regions belonging to ImS always include the regions of highest sensitivity and, therefore, consecutive regions exclude surfaces of minimum sensitivity. As a consequence, myocardial regions of minimum size for a given level α were always found to belong to the family ImS. We can, thus, state that, given the conditions of the simulations, the scope of unipolar and bipolar lead system corresponds to the myocardial region of maximum sensitivity.

Comparing unipolar to bipolar lead system, it is worth noting that the curves of $MSD_n\{V_o\}$ as a function of the size of V_o are steeper in the latter than the former. This observation is related to the fact that, in an infinite and homogeneous medium, the lead field for unipolar systems decreases inversely to r^2, while for bipolar systems it decreases inversely to r^3, where r is the distance to the lead system. Thus, the rate of decrease of $MSD_n\{V_o\}$ will be expected to be higher for bipolar than for unipolar lead systems. As a consequence, the scope of bipolar systems will always be smaller for bipolar systems than for unipolar leads. Fig 5 and Fig 6 represent, respectively, the scope of unipolar and bipolar lead systems at a level of 10%. In both cases, the scope is located close to the electrode placement, where the sensitivity is highest. Nevertheless, while for unipolar lead systems the size of the myocardial region that constitutes the scope is roughly $100cm^3$, for bipolar lead systems is approximately $10cm^3$, an order of magnitud smaller. Thus, our analysis confirm that the scope of unipolar lead systems is global, and the scope of bipolar lead system is local.

IV. Discussion

In this study we have addressed the problem of determining quantitatively the scope of unipolar and bipolar ICD transvenous lead systems. A quantitative framework for studying the scope of transvenous lead systems can provide us with a sound basis (1) to develop new methods of EGM analysis, (2) to determine differences in the scope of existing leads, and (3) to investigate sources of variability, such as lead positioning, type of heart disease and underlying dynamics.

We have defined the scope of lead systems based on

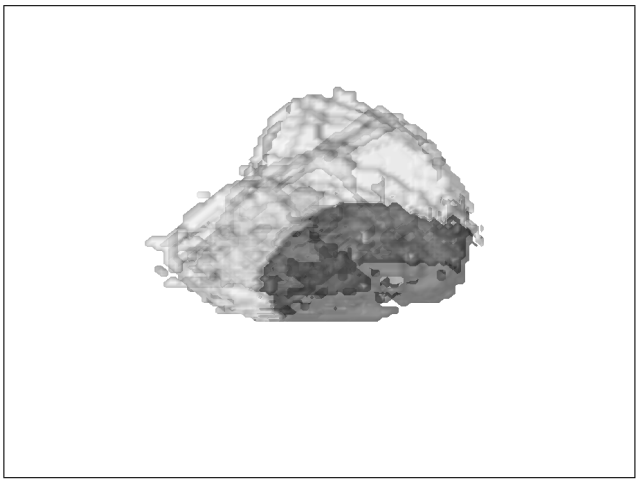

Fig. 5. Myocardial region of minimum volume involving an error MSD_n of 10% for unipolar leads.

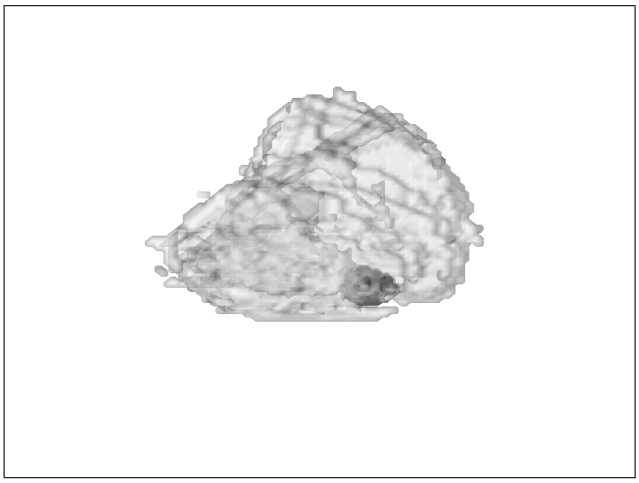

Fig. 6. Myocardial region of minimum volume involving an error MSD_n of 10% for bipolar leads.

the MSD between EGM signals generated by the whole myocardium and EGM signals generated by regions within the myocardium. In order to estimate the MSD, we combined bioelectric signal modeling and numerical methods. By means of this procedure, we can estimate quantitatively the scope of trasvenous lead systems, defined as the myocardial region of minimum size that involves an MSD_n smaller than α. We studied $MSD_n\{V_o\}$ as a function of the size of V_o for three families of myocardial masses. We found in our simulations that the scope corresponds to the myocardial region of maximum sensitivity. Moreover, for bipolar EGM the scope was found to be an order of magnitude smaller than for unipolar EGM, confirming that bipolar leads capture local activity, while unipolar EGM capture global activity.

This numerical approach has, however, certain limitations. Since an underlying model is assumed, the significance and generality of the results depend on the ability of the model to describe the physical phenomena under investigation. In this study we have described the conducting properties of the body based on the VHM dataset, and the electrical activity of the heart based on a state machine model. Besides, the dynamics of the sources elements have been generated by simulating the stimulation of the apex at 60 beats per minute. In order to investigate the impact of the choice of the signal model on the scope of the results, in the future we will implement different thorax models and simulate cardiac electrical activity based on other descriptions and stimulation protocols. Furthermore, this procedure will be extended to study other factors involved in the determination of the scope of a lead system, such as lead design (integrated and true bipolar), lead positioning, type of heart disease and underlying cardiac dynamics.

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