Variability of Premature Ventricular Contraction Localization with Respect to Source and Forward Model Variation in Clinical Data

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Abstract

Objective: *Electrocardiographic imaging can provide preliminary locations of premature ventricular contraction (PVC) origins, which can lead to the shortening of the invasive radio-frequency ablation (RFA) procedure. However, PVC localization results vary significantly with respect to the equivalent cardiac source models and the level of complexity assumed in the forward model. This study aims to evaluate PVC localization based on different source and forward model assumptions.* ***Methods:*** *We use body surface potential (BSP) measurements from 5 patients with PVCs, indicated for RFA. Two equivalent source models (dipole-based and heart surface potential-based), and homogeneous and inhomogeneous torso geometries are used to evaluate the variability in the PVC origin.* ***Results:*** *Dipole-based PVC locations are in general more clustered than potential-based ones. Spatio-temporal AT estimates smooth the ATs improving the PVC localizations to be more clustered. Potential-based solutions were more sensitive to changes in the forward model than the dipole-based method.* ***Conclusion:*** *The AT and RRE maps provided similar information consistent with patient descriptions. Therefore, rather than specifying a single point PVC origin, these maps could guide the physicians for more accurate PVC localization.*

1. Introduction

Cardiovascular diseases are among the most common reasons of (sudden) death, even at the productive ages, in the developed countries. Therefore, non-invasive diagnosis of these diseases and developing effective treatment strategies have been the top priority for clinicians and researchers [1]. Abnormal electrical activity of the ventricles can result in various types of arrhythmias, ranging from single premature ventricular contractions (PVC) to sustained ventricular tachycardia.

Electrocardiographic imaging (ECGI) is a novel non-invasive approach that estimates the electrical activity of the heart from body surface potential measurements (BSPMs) and a mathematical model of the torso. ECGI can provide a preliminary location of the origin of the ectopic activity, which can lead to the shortening of the invasive radio-frequency ablation (RFA) procedure.

Previous studies show that the ECGI-based localization of the origin of PVCs varies significantly with respect to the equivalent source model for the cardiac sources (dipole, multipole, epicardial/endocardial potentials, etc.) and the level of complexity assumed in the forward model (homogeneous/inhomogeneous) [2].

In this study, we use BSPMs measured from 5 patients with PVCs, indicated for RFA. We assume two equivalent source models: dipole-based and heart surface potential-based. We obtain forward matrices using the boundary element method (BEM) for epicardial only/ epicardial-endocardial surfaces of the heart, and for homogeneous/ inhomogeneous torso models. We evaluate the variability in the PVC localization results based on these various source and forward models.

2. Methods

2.1. BSPM Measurments

BSPMs were measured on the patients' torso using 128 electrodes organised in 16 strips with 8 electrodes around the chest by ProCardio measuring system [3]. The measurements were performed on 5 patients with PVC, indicated for RFA. The measurement duration varied from 5 to 20 minutes. Annotations of these patient data by a cardiologist are summarised in Table 1.

The measured signals were processed using the high pass filter with a finite impulse response designed by the window method and the Blackman-Harris window with a cut-off frequency of 0.5 Hz. In the chosen signal R-peaks were found for all cardiac cycles and the cycles were clustered according to their morphology. Then the signals in clusters for PVC beats were averaged. Finally, the starting time instant was estimated for the PVC beat and all signals were corrected by a constant value in such a

way that at this instant the BSPM was zero. This signal-averaged PVC beat was used for ECGI [4]. Each patient also underwent a CT scan of the chest, and a patient-specific torso model was created by segmentation of these CT scans.

Table 1. Patient information.

Patient #	Description
P1(001)	The focus was in the septum near the atrioventricular node (no RFA). The earliest signal was detected in the left ventricle, at the septum, towards the back.
P2(004)	Ablation in RVOT, anterolateral on a free wall.
P3(006)	Focus in CS at LVOT level, laterally more backwards.
P4(008)	Focus in RVOT, upper part, laterally.
P5(029)	Patient with anterior infarction. Ablation in RVOT, under pulmonary valve, anterior free wall.

2.2. Dipole-Based Inverse Solution

Considering only the first 20-30 ms of the PVC time interval the activated area is small enough to be represented by a single dipole. Therefore, we can use it for the computation of the PVC origin position. We assume that the PVC origin can be in every point of the epicardial or endo-epicardial triangulated surface of the ventricles [5].

For a dipole in each specific point position body surface potentials (BSP) are computed by the linear equation

$$\text{BSP} = \mathbf{T}\text{d} \tag{1}$$

where **T** is a transfer matrix computed by a boundary element method in a torso model, describing the relation between the dipole *d* in the given position and the potentials on the torso, that is assumed as a volume conductor. Then for the unknown dipole moments, measured BSP, and computed T the equation (1) leads to

$$\text{d} = \mathbf{T}^{-1}\text{BSP} \tag{2}$$

Such an equation has a unique solution fulfilling the criterion of the minimal least-squares method. In each time step for the beginning 20-30 ms of activation, we compute the result for each assumed position of the dipole on the ventricular surface. For each resulting dipole d_res the corresponding BSP dip_map and the relative residual error RRE to the measured BSP is calculated as:

$$\text{RRE} = \|\text{BSP-dip\_map}\|/\|\text{BSP}\| \tag{3}$$

where ||.|| means the Euclidean norm. Then the position of the dipole with minimal RRE value for each time step and all positions on the ventricles is considered as the location of the PVC origin.

2.3. Potential-Based Inverse Solution

In this work, two source models for heart potentials are used: Epi (i.e., epicardial electrograms) and EndoEpi (i.e., endocardial-epicardial electrograms). Given the BSPMs, Epi and EndoEpi potentials are estimated using the linear equation [1]:

$$\mathbf{Y} = \mathbf{AX} + \mathbf{N} \tag{4}$$

Such that $\mathbf{Y} \in R^{MxT}$ and $\mathbf{X} \in R^{NxT}$ are the BSPMs and Epi or EndoEpi potential matrices, respectively, where M is the number of ECG electrodes, N is the number of nodes on the heart, and T is the number of time samples. $\mathbf{A} \in R^{MxN}$ is the forward matrix and $\mathbf{N} \in R^{MxT}$ is considered as measurement noise.

Given the heart, torso, and lung geometries, the forward matrix **A** is computed with the Boundary Element Method based on Barr *et al.* [6] using two geometrical models: one with a homogeneous conductivity between epicardial surface and torso we call HT model, and an inhomogeneous HLT model with an additional lung conductivity layer.

Tikhonov regularization method [7] was used to deal with the ill-posed nature of the inverse electrocardiography problem of equation (1) and the heart potentials were found for each time instant separately such that the cost function below is minimized.

$$\hat{\boldsymbol{x}} = \boldsymbol{argmin}_{\boldsymbol{x}}(\|\boldsymbol{Ax} - \boldsymbol{y}\|_2^2 + \boldsymbol{\lambda}^2\|\boldsymbol{Rx}\|_2^2) \tag{5}$$

Here, $\boldsymbol{R}$ is a regularization matrix (identity matrix in zero order Tikhonov regularization employed in this study) and $\boldsymbol{\lambda}$ is the regularization parameter whose value was optimized with L-curve method [8] at each time instant then the median of those values was selected as the regularization parameter for all time instants in the inverse problem solution.

Since the accuracy of the activation times depends on the heart potential estimations, before calculating them the EGMs were passed through a low-pass Butterworth filter with a cut-off frequency of 30 Hz. Then two activation time sequences were computed from each of the EGMs of the four homogeneous and inhomogeneous Epi/EndoEpi models. First, the minimum derivative (dV/dt) of the filtered signals was calculated and a pacing location was marked at the earliest time detected. Second, a Spatio-Temporal method [9] was employed which starts with local activation time (LAT) estimation at each node of EGMs based on the min(dVdT) method. Sudden changes in LATs can occur due to signal inaccuracies, to avoid these fluctuations ST method regularizes AT sequences such that they are smoothed over space. Finally, the PVC location was marked as the node with the earliest activation time during the QRS region.

3. Results

3.1. Dipole-Based Inverse Solution

The results obtained by a single dipole were quite robust to the changes of the cardiac surface model or to the complexity of the torso model. On the other hand, this method does not provide the information about the whole activation propagation like the potential based method.

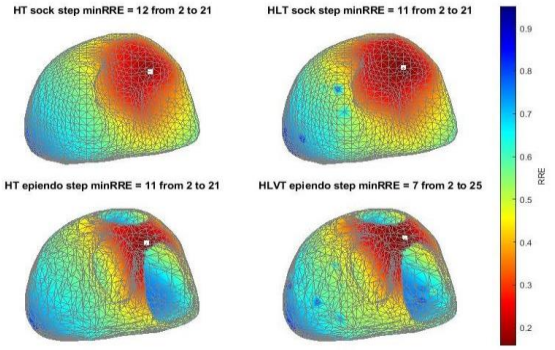

Figure 1. RRE value for each position of the inversely estimated dipole for the time step with minimal RRE value for P1. The white square indicates the estimated position of the PVC origin.

Fig. 1 shows the RRE value distributions for patient P1, for each position of the inversely estimated dipole for the time step with minimal RRE value. All models (Epi/EpiEndo and HT/HVLT) resulted in similar distributions for RRE. For this patient, minimum RRE location (PVC origin) is also consistent with all models.

3.2. Potential-Based Inverse Solution

The performance of this approach depends on the accuracy of the AT estimation method. Thus, we first evaluated the performances of dVdT and sptemp methods. Fig. 2 displays the AT maps and Fig. 3 shows a single lead electrogram with AT marks with both methods for P1, EndoEpi, HT model. Using the sptemp method decreased the abrupt changes seen in the minimum derivative (dVdT) activation time sequence, therefore sptemp results were more stable. However, sometimes this spatial smoothing produced LATs that are shifted compared to minimum derivative based AT estimates (Fig. 3).

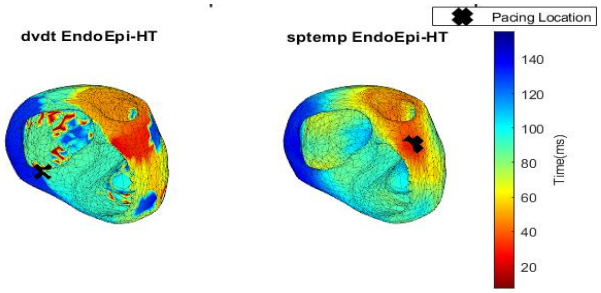

Figure 2. AT maps found by dVdT and sptemp methods for P1, EndoEpi, HT case.

Noise in the estimated electrograms resulted in inaccuracies in min(dVdT) AT calculations. Prefiltering the signals remedied this issue as illustrated in Fig. 4.

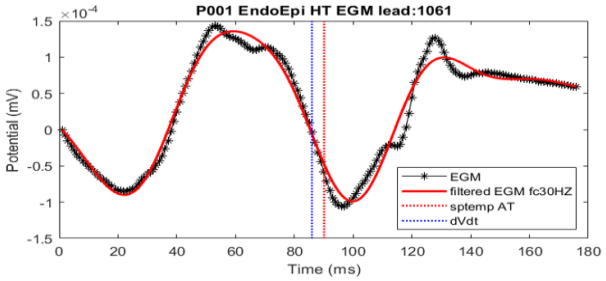

Figure 3. AT comparison of dVdT and sptemp methods for a single lead (P1, EndoEpi, HT).

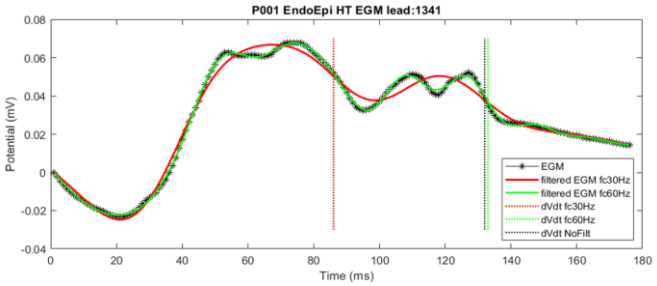

Figure 4. Effects of prefiltering of the electrogram on the AT estimates during the QRS region (P1, EndoEpi, HT).

Fig. 5 shows the sptemp-AT distributions for P1 for all models (Epi/EpiEndo and HT/HLT). AT distributions for Epi and EndoEpi source models were in agreement with each other for both the HT and the HLT forward models. However, HT and HLT results had some differences on the RV side of the heart. Depending on where the minimum AT is achieved, HT and HLT model PVC origins were also different.

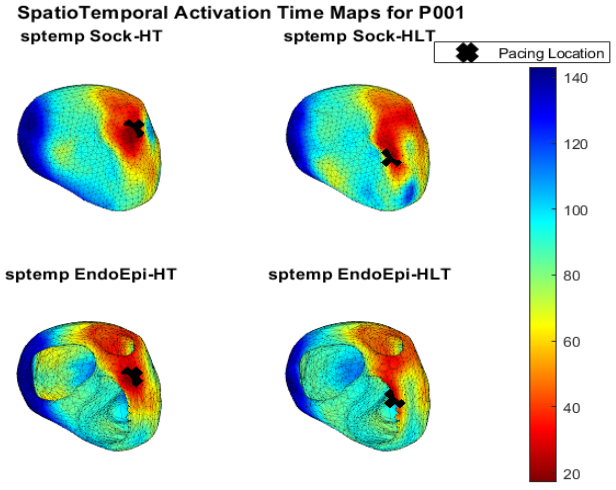

Figure 5. Pacing location on activation time maps during QRS for P1.

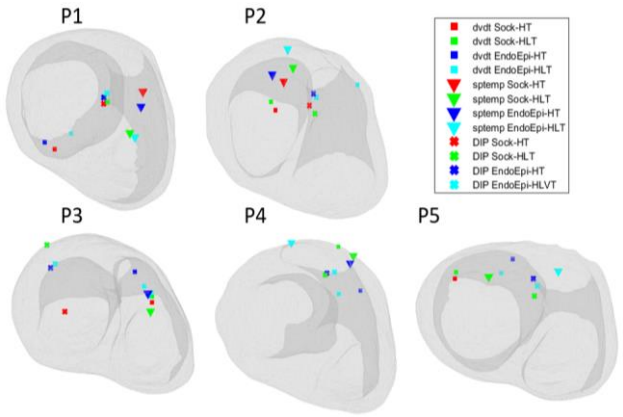

Figure 6. Pacing locations found by all methods for all patients.

Fig. 6 shows the PVC origins for all 5 patients estimated by all source and forward models, and both AT estimation methods. Dipole-based PVC locations (DIP) are in general more clustered compared to potential-based PVC locations. Only in P3, dipole-based PVC locations are also spread around the LV. Among the potential-based solutions, min(dVdT)-based AT's, due to aforementioned artifacts, result in PVC origin estimates (dVdT) spread around the heart and usually with conflicting results. Spatio-temporal AT estimates smooth the ATs, therefore result in more clustered PVC origin estimates (sptemp).

4. Discussion and Conclusion

In this paper, the variation of PVC localization based on various cardiac source and forward models were examined on clinical data.

Even though Tikhonov regularization generally resulted in smooth EGMs, the measurement artifacts and geometrical errors can cause the activation time calculation to be inaccurate for some nodes. Prefiltering the signals temporally improved the minimum derivative (dVdT) results and hence the spatiotemporal activation times to some extent, however, the final localization of PVC is still prone to AT performance. The spatiotemporal AT method produced stable results for both homogeneous and inhomogeneous models in general. Still, there is a need for a more robust AT estimation method to improve PVC localization performance.

Although there was too much variation in the PVC origin estimates for each case, the AT maps and the RRE maps were usually quite consistent. Therefore, rather than relying on a single point PVC origin estimate, these maps could guide the physicians for more accurate PVC localizations.

Acknowledgments

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