The Equivalent Dipole Used to Characterize Atrial Fibrillation

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Abstract

This study aims at developing a measure for describing the spatial complexity of atrial signals during atrial fibrillation (AF). A biophysical model of the atria and a volume conductor model of the thorax were used to simulate the atrial electrical activity as expressed on the torso. Features were extracted from the equivalent dipole (VCG) derived from body surface potentials. The robustness of these features is documented by comparing their values with those derived from a limited set of electrodes. In addition, the effect of a possibly incomplete cancellation of ventricular activity is shown.

1. Introduction

The standard electrocardiogram (ECG) is the most common non-invasive tool for diagnosing atrial fibrillation (AF). So far, the assessment of the complexity or the hidden organization of AF dynamics from ECG signals has mainly been performed through frequency or time-frequency analysis [1]. In order to obtain a more complete picture of AF and therefore a more precise diagnosis, it would be desirable to extract features containing also information about the spatial dynamics of AF.

The complexity of the spatiotemporal behavior of the electrical sources during AF precludes in all likelihood the characterization of this complexity by means of an inverse procedure. This holds true in particular if the available ECG data are restricted to those of the standard 12-lead ECG, the data that are almost exclusively available in the clinical setting.

The vectorcardiogram (VCG) provides a global representation of the cardiac electric activity and the time course of the vector orientation in 3D space (that of the equivalent current dipole) may serve in extracting spatial features of AF. Moreover, this equivalent dipole can be estimated by a linear combination of the eight independent signals contained within the standard 12-lead ECG.

In this paper, we introduce some basic features for the spatio-temporal characterization derived from the atrial VCG. The robustness of these features is tested against the background of deriving them from a limited number of signals (the standard 12-lead ECG), or in the presence of a possibly imperfect quality of the required cancellation of the ventricular activity (QRST complexes).

The study is based on ECG data simulated by means of a computer model of the human atria embedded in a realistic inhomogeneous model of the volume conduction aspects within the thorax.

2. Methods

2.1. Simulated atrial signals

A three-dimensional, thick-walled, biophysical model of the atria was developed based on magnetic resonance (MR) images, which simulates the propagation of the electrical impulse [2]. Figure 1A shows the resulting atrial geometry as seen from an anterior view (on the left) and from a posterior view (on the right). The major anatomical details are indicated: the tricuspid valve (TV), the mitral valve (MV), the inferior vena cava (IVC), the superior vena cava (SVC), and the pulmonary veins (PV), the sino-atrial node (SAN), the Bachman's bundle (BB), and the left atrium appendage (LAA). The electrical propagation of the cardiac impulse was simulated using a reaction-diffusion system (monodomain formulation) based on a detailed ionic model of the cell membrane kinetics, the Courtemanche *et al.* model [3], comprising a total of 300,000 units.

In order to create a substrate for AF, patchy heterogeneities in action potential duration were introduced by modifying the local membrane properties [4]. Simulated AF was induced by rapid pacing in the left atrium appendage. Eighteen different episodes of AF were generated. The episodes differ by the arrhythmogenic substrate that was created in order to make the model vulnerable to AF. In addition, an episode of atrial flutter as well as an episode of normal rhythm were simulated, yielding a total of 20 data sets of atrial activity.

Body surface potential maps of atrial activity were computed at 300 points including as a subset the locations of the 9 ECG electrodes of the standard 12-lead system.

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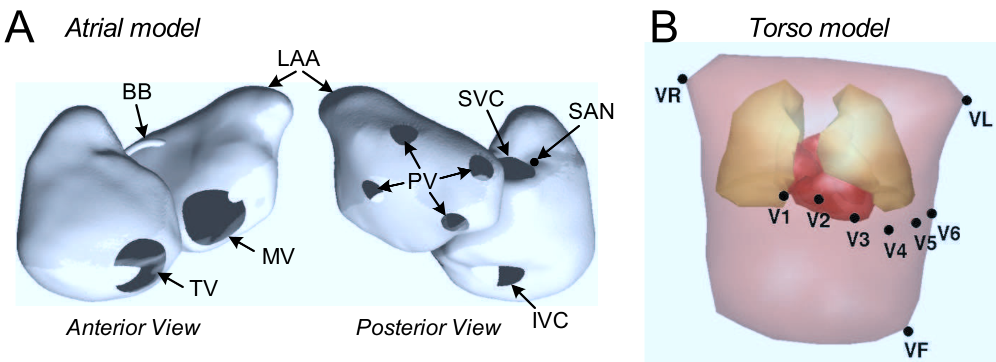

Figure 1. Geometry of the models representing the atria and the thorax.

The source description used was the equivalent double layer specified at the closed surface bounding the myocardium [4]. The local double layer strength at the surface was taken to be the time course of the transmembrane potential computed in the atrial model. The effect of volume conduction heterogeneity on the atrial contribution to body surface potentials was computed by means of the boundary element method. This was applied to a compartmental torso model including the atria, the ventricles, blood cavities and the lungs [4]. Figure 1B displays this torso geometry (nearly frontal view), also derived from MR images, as well as the position of the 9 standard ECG electrodes.

2.2. Estimated dipoles

The time course of 3 different estimates of the equivalent dipole was computed for each of the 20 data sets:

• Estimate A was derived from the 300 ECG signals by integrating the body surface potential Φ over the body surface S:

$$\mathbf{D}_A(t) = \oint_S \mathrm{d}\mathbf{S}(\mathbf{x})\ \Phi(\mathbf{x},t)\ , \tag{1}$$

where $\mathrm{d}\mathbf{S}$ is the outward-oriented surface element. This expression is known as the Gabor-Nelson equations. Its results may be considered as the gold standard, as discussed in [5]. The integral in Eq. (1) was discretized by using a triangular mesh describing the torso surface.

• Estimate B was derived by means of a dedicated set of transfer coefficients applied to 12-lead ECG data, according to the formula:

$$\mathbf{D}_B(t) = \mathbf{T}\ \ \mathbf{\Phi}_{\mathrm{ECG}}(t)\ . \tag{2}$$

The $9{\times}1$-vector $\mathbf{\Phi}_{\mathrm{ECG}}(t)$ represents the instantaneous potentials generated by the atrial activity at the 9 standard electrode positions at time instant t. The $3{\times}9$ transfer matrix $\mathbf{T}$ was dedicated to the estimation of the atrial dipole as described in van Oosterom *et al.* [5].

• Estimate C used the same simulated 12-lead ECG atrial signals to which ventricular components were added [6] and subsequently cancelled by using an average beat subtraction approach [7]. This provided a means for evaluating the quality of the AF features in a practical situation involving an imperfect cancellation. The resulting estimate is:

$$\mathbf{D}_C(t) = \mathbf{T}\ \left(\mathbf{\Phi}_{\mathrm{ECG}}(t) + \Delta\mathbf{\Phi}_{\mathrm{ECG}}(t)\right)\ , \tag{3}$$

where $\Delta\mathbf{\Phi}_{\mathrm{ECG}}$ represents possible cancellation artifacts.

2.3. Spatial Features

For each of the 3 dipole estimates $\mathbf{D}_i$ with $i = A, B, C$, the eigenvalues of the covariance matrix of the 3 components of the dipole with normalized magnitude were taken as a measure of spatial complexity of the dipole signals. The $3{\times}3$ matrix $\mathbf{C}_i$ was defined as the second-order raw moment matrix of the normalized signals:

$$\mathbf{C}_i = \frac{1}{t_2 - t_1}\int_{t_1}^{t_2} \mathrm{d}t\ \frac{\mathbf{D}_i(t)\cdot\mathbf{D}_i(t)^\top}{\|\mathbf{D}_i(t)\|^2} \tag{4}$$

computed over the time interval $[t_1, t_2]$. The eigenvalues of $\mathbf{C}_i$, $\lambda_1 \geq \lambda_2 \geq \lambda_3 \geq 0$, were the features used in this analysis. Because the trace of the integrand in Eq. (4) is 1, we have $\lambda_1 + \lambda_2 + \lambda_3 = 1$.

This approach is related to the analysis of directional data by parametric statistical modeling [8]. Bingham's distribution provides a model to represent spherical data. Its probability density function is a trivariate Gaussian restricted to the unit sphere [9]:

$$p(\mathbf{x}) = K(k_1,k_2,k_3)\cdot \exp\left(k_1(\mathbf{x}^\top\mathbf{u}_1)^2 + k_2(\mathbf{x}^\top\mathbf{u}_2)^2 + k_3(\mathbf{x}^\top\mathbf{u}_3)^2\right)\ , \tag{5}$$

where k_1, k_2 and k_3 are parameters, and $\mathbf{u}_1$, $\mathbf{u}_2$ and $\mathbf{u}_3$ form an orthonormal basis. The normalization constant K is such that the integral of $p(\mathbf{x})$ over the unit sphere is 1. It can be shown that the maximum likelihood estimate of

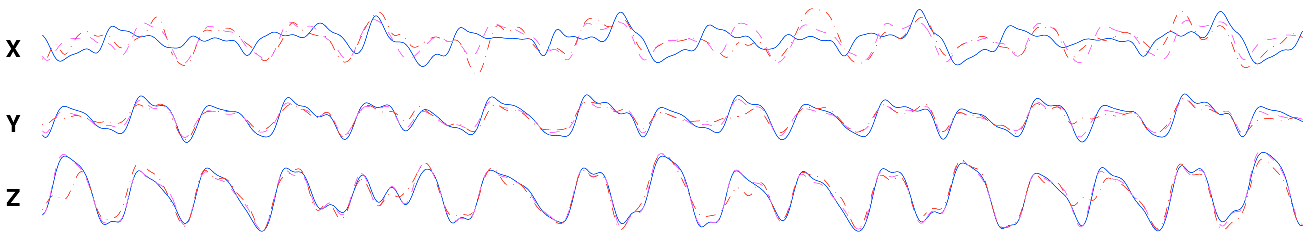

Figure 2. Example showing the time course (during AF) of each component of the estimated dipoles A (blue solid line), B (magenta dashed line) and C (red dash-dotted line). The signal segment is 2-s long.

the vectors $\mathbf{u}_1$, $\mathbf{u}_2$ and $\mathbf{u}_3$ is given by the eigenvectors of the matrix $\mathbf{C}_i$. The maximum likelihood estimate of k_1, k_2 and k_3 is a (nontrivial) function of the eigenvalues λ_1, λ_2 and λ_3 [8, 9].

To evaluate the robustness of these eigenvalues with respect to the perturbing elements involved, the Pearson correlation coefficients between the estimates A, B and C of λ_1, λ_2 and λ_3 were computed over the 20 simulations, as well as their mean relative difference.

3. Results

Figure 2 shows the time course of the dipole components (X,Y and Z) as estimated using the methods A, B and C during an episode of simulated AF. The X-axis is oriented back-to-front, the Y-axis right-to-left and the Z-axis feet-to-head from the patient's viewpoint. The X component is clearly more difficult to estimate from the 12-lead ECG because the position of the electrodes involved has a X-coordinate within a narrow range. Although some residual ventricular activity is still visible, the quality of the cancellation algorithm seems to be sufficient for determining the dipolar activity of the atria.

In order to illustrate the spatial information provided by the equivalent dipole, Fig. 3 displays the dipole estimate A projected on a sphere for selected episodes of simulated sinus rhythm, atrial flutter and AF. During the P-wave, the equivalent dipole follows a well-documented loop [10]. The parameters λ for this simulation were $\lambda_1 = 0.86$, $\lambda_2 = 0.13$ and $\lambda_3 = 0.01$. The distribution of equivalent dipole orientation during atrial flutter is concentrated around a great circle ($\lambda_1 = 0.72$, $\lambda_2 = 0.26$, $\lambda_3 = 0.02$). The AF episodes presented in Fig. 3 differ by the complexity of their dynamics as expressed by the VCG, the first one ($\lambda_1 = 0.65$, $\lambda_2 = 0.23$, $\lambda_3 = 0.12$) being less disorganized than the second one ($\lambda_1 = 0.42$, $\lambda_2 = 0.31$, $\lambda_3 = 0.27$), in which the values of the features are closer to the uniform case ($\lambda_1 = \lambda_2 = \lambda_3 = 1/3$). This reflects the fact that the average wavelength of the depolarization waves is shorter in the second AF episode (7.3 cm vs 9.4 cm), leaving enough space for more wavelets and reentries.

The robustness of the features λ_1, λ_2 and λ_3 against the use of a limited set of electrodes for estimating the VCG, and incomplete cancellation of ventricular activity is documented in Tables 1 and 2. Table 1 presents the mean relative difference of the features as estimated by method B and C, the method A being considered as the reference. Similarly, Table 2 shows the Pearson correlation coeffi-

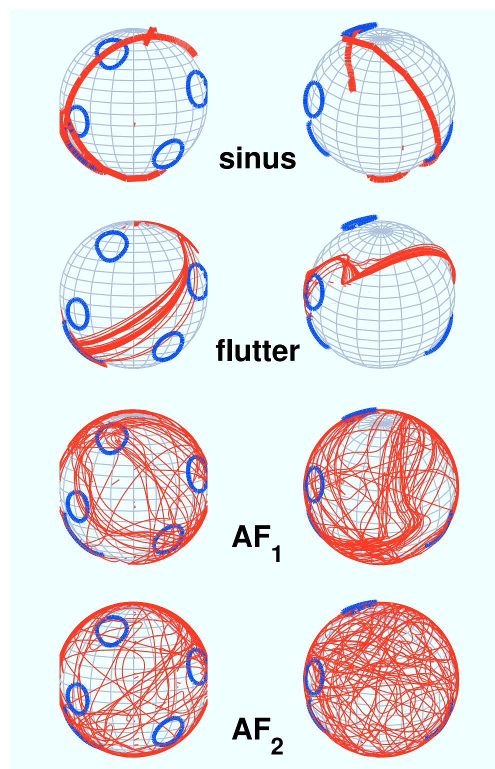

Figure 3. Equivalent dipole (estimate A) projected on a sphere for four different types of atrial activity: posterior view (left panels) and anterior view (right panels). The approximate orientation of the projection of the four pulmonary veins are shown, as well as of the mitral valve.

cient between the estimates A, B and C of the features λ_1, λ_2 and λ_3.

Table 1. Mean relative differences of the estimates B and C of the eigenvalues (A is the reference).

Eigenvalue	B vs A	C vs A
λ_1	9%	9%
λ_2	13%	17%
λ_3	39%	33%

Table 2. Correlation between the eigenvalues of the different dipole estimates.

Eigenvalue	A vs B	B vs C	A vs C
λ_1	0.91	0.84	0.81
λ_2	0.85	0.94	0.83
λ_3	0.75	0.69	0.48

4. Discussion and conclusions

The results indicate that the first eigenvalue λ_1 yields a robust measure of spatial complexity of the VCG during AF, even when only a standard 12-lead ECG is available, as is usually the case in a clinical environment. The estimate of the second eigenvalue λ_2 still provides reliable information about the distribution of the dipole orientations. The third eigenvalue $\lambda_3 = 1 - \lambda_1 - \lambda_2$ has a lower amplitude and accumulates the errors of the two previous ones, notably due to the difficulty in estimating the X component of the VCG from the 12-lead ECG.

The use of an optimized lead system dedicated to the extraction of an atrial VCG would significantly improve the estimation of the equivalent dipole provided that a dedicated transfer matrix is computed [5]. The accuracy of the QRST cancellation algorithm also plays a role, but should major artifacts remain, the features λ_i may be extracted based on the TQ segments only.

A combination of the features presented in this paper and the frequency-domain features developed previously may be used to identify and characterize subforms of AF that differ in their spatial organization such as multiple wavelets-based chronic AF, ectopic foci in the pulmonary veins, or fibrillo-flutter.

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