Noninvasive Location of Acute Ischemic Lesion in the Heart Ventricles Using a Few-lead System: Study on a Realistic Mathematical Model

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***Abstract**. With the use of an interactive computer model of the cardioelectric generator and body as conducting medium, a new method for locating the acute ischemic lesion in the heart is investigated. The method is based on recording the unipolar signals with the electrodes of the Frank vectorcardiographic lead system and calculating the displacement of the dipole equivalent cardiogenerator in the direction of its moment vector. It is shown that this method provides recognition of the major anatomical position of the injured region in the heart ventricles for various localizations of the lesion.*

1. Introduction

Determination of the anatomical position and stage of development (intensity) of the ischemic injury is one of the urgent problems of electrocardiographic diagnosis. There are well known efficient methods for solving this problem by body-surface electrocardiographic mapping with the use of rather complicated measuring procedure requiring several tens of leads [1]. In [2], we proposed a simplified method for estimating the characteristics of the ischemic zone on the basis of measurements using a modified vectorcardiographic Frank lead system including the same number of electrodes as the standard electrocardiographic system. This method was tested on the simplest mathematical models, in particular, a single point dipole model of the cardiogenerator and homogeneous parallelepipedal model of the chest.

Here, we describe an improved version of the aforementioned method and present some results obtained in modeling it on a sophisticated interactive computer model of the cardioelectric generator for the whole heart and body as volume conductor [3, 4].

According to electrophysiologic investigations, the electric generator set up by the local zone of acute ischemia in the heart can be represented with sufficient practical accuracy as an equivalent uniform electric double layer coinciding with the boundary between the regions of ischemia and healthy myocardium. The dipole moment density at each point of this double layer (double layer moment) is proportional to the difference of the action potential amplitudes in the cells of the aforementioned regions and is directed from the healthy to ischemic region. The double layer moment is maximal for the complete injury of the ischemic cells, when the cells are not able to polarize. The magnitude of the heart vector (total dipole moment of the cardioelectric generator) is proportional to the aforementioned difference of the action potential amplitudes and the area of the double layer projection onto the plane perpendicular to the ischemic heart vector. This vector is directed as the normal to the heart wall near the midpoint of the ischemic zone. The plane situated with minimum deviation from the double layer boundary, referred to as generator midplane, is perpendicular to the ischemic heart vector and is displaced by a distance dH from the geometric heart center, thus it characterizes the general position of the ischemic zone in the heart ventricles. In the case of transmural ischemia, the midplane is located between the endocardial and epicardial boundaries of the ischemic zone.

In electrocardiographic terms, the acute ischemia results in a stable shift of the systolic segment S-T of the electrocardiogram. So, for the estimation of the injured zone position, the generator midplane should be determined at the middle part of the S-T period.

2. Method and Model

It is assumed that the cardioelectric potentials are measured by the modified Frank lead system abbreviated as Frank-M system. It includes electrodes of the classical Frank vectorcardiographic lead system, in particular, the electrodes I, E, C, A, M in the transversal plane passing through the heart center and electrodes H and F on the head and left leg, respectively, with the additional electrodes R and L on the right and left arms, respectively (Fig. 1). The primary signals are the unipolar potentials of the individual Frank electrodes with respect to the Wilson's central terminal, while the signals proportional to the three components of the heart vector are also used. Two orthogonal coordinate systems with the origin at the geometric center of the chest are used, the main coordinate system *xyz* and auxiliary system *x'y'z'* turned by angle π/4 around the *z*-axis.

The displacement of the generator midplane with respect to the heart center is defined by the position of an equivalent dipole generator which is assumed to lie on a straight line passing through the origin of the *xyz* coordinate system and have a moment with the same orientation as the heart vector. The potential of this dipole should approximate the potentials of the measuring electrodes by the specified criterion.

For defining the position of the equivalent dipole, two orthogonal coordinates, z_D and u_D, are considered; these are the coordinate along the *z*-axis and coordinate along the *u*-axis, the latter coinciding with the projection of the heart vector onto the coordinate plane *x*0*y*. To determine the coordinates under consideration, the averaged values of the measured potentials in the coordinate half-spaces corresponding to the positive and negative directions of the dipole moment components *Dz* and *Du* are approximated by the potentials of the corresponding equivalent dipoles.

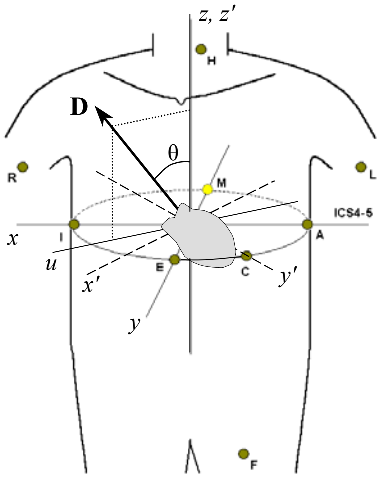

Fig. 1. Electrode positions of the Frank-M lead system and axes of the coordinate systems used.

The imaginary measuring points considered are situated on the positive and negative semiaxes of the coordinate systems *xyz* and *x'y'z'* at the distances r_x, r_y, r_z, $r_{x'}$, $r_{y'}$ from the coordinate origin (subscripts indicate the corresponding axes). The potential at the particular imaginary point is equal to the root-mean-square potential of the true measuring points situated in the same coordinate half-space and on its boundary coordinate plane. For example, the potential at the imaginary point on the positive *x*-semiaxis is

$$\varphi_{+x} = \sqrt{\frac{1}{5}\left(\varphi_{\mathrm{I}}^2 + \varphi_{\mathrm{M}}^2 + \varphi_{\mathrm{E}}^2 + \varphi_{\mathrm{H}}^2 + \varphi_{\mathrm{F}}^2\right)} \qquad (1)$$

(the expressions for the other half-spaces are composed in a similar way).

There are imaginary measuring points considered separately on the *z*-axis and on the coordinate axes lying in the *x*0*y* plane and deflected from the *Du* component of the equivalent dipole by an angle less than *π*/8. Thus the component *Du* is assumed to be collinear to a certain coordinate axis, if this component lies in the corresponding sectors of the *x*0*y* plane, particularly, the sectors along the *x*-axis, *y*-axis, *x'*-axis, or *y'*-axis.

Similar to the variation of the dipole potential in the infinite homogeneous conductor, it is assumed that the potentials at the imaginary measuring points on the *z*-axis or on the horizontal coordinate axis approximating the *u*-axis in the corresponding sectors is inversely proportional to the squared distance between this point and projection of the equivalent dipole onto the same axis. Then after some appropriate transformations the equivalent dipole coordinates (relative to the chest parameter *a*) $\hat{u}_{\mathrm{HD}} = u_{\mathrm{HD}}/a$ and $\hat{z}_{\mathrm{HD}} = z_{\mathrm{HD}}/a$ in the $u_H z_H$ coordinate system with the origin at the heart center and axes u_H and z_H coinciding in direction with the axes *u* and *z*, respectively, are expressed in terms of the square roots of the corresponding imaginary point potentials. Thus, for the direction of *u* inside the *x*-axis sectors

$$\hat{u}_{\mathrm{HD}} = \hat{r}_x \frac{\sqrt{\varphi_{+x}} - \sqrt{\varphi_{-x}}}{\sqrt{\varphi_{+x}} + \sqrt{\varphi_{-x}}} + \Delta_x \tag{2}$$

(Expressions for the other axis directions are similar).

Parameters $\hat{r}_x$, $\hat{r}_y$, $\hat{r}_{x'}$, $\hat{r}_{y'}$, $\hat{r}_z$, Δ_x, Δ_y, $\Delta_{x'}$, $\Delta_{y'}$, Δ_z depend upon the considered structures of the cardiogenerator and chest models. These parameters were initially chosen by means of an extremely simple mathematical model of the cardiogenerator in the form of a set of 9 point dipoles with various orientations situated in the main parts of the heart space and model of the chest in the form of a homogeneous conductive parallelepiped with the main dimensions equal to the assumed chest parameters (in particular, transversal diameter a, sagittal diameter b, and height h) and heart center coordinates x_c, y_c, z_c in the coordinate system xyz. For modeling the typical male torso, the following values were used: a = 30 cm, $b = 0.71a$, $h = 1.41a$, $x_c = -0.1a$, $y_c = 0.1a$, $z_c = 0$. For any spatial direction of the heart vector, the generator midplane crosses this vector at the point shifted with respect to the heart center by the distance d_{H} (which is positive, if this point is displaced in the direction of the heart vector, and negative for the displacement in the opposite direction). The relative value of this distance is expressed as

$$\hat{d}_{\mathrm{H}} = d_{\mathrm{H}}/a = \hat{u}_{\mathrm{HD}} \sin\theta + \hat{z}_{\mathrm{HD}} \cos\theta, \tag{3}$$

where θ is the angle between the z-axis and heart vector.

To verify the obtained relations, we used an interactive computer model of propagated ventricular activation in conjunction with a realistic torso model [3, 4]. The geometry of ventricles was defined analytically and was based on several ellipsoids. The properties of myocardial cells, such as conduction velocity, action potential amplitude and duration were defined for all elements. The activation spread was governed by the Huygens principle, the isotropic myocardial tissue was supposed. The cardiac generator was placed into a realistic torso model with basic inhomogeneities. The boundary element method was used for computation of the potential distribution. The potentials corresponding to particular depolarization-repolarization sequence were computed at 198 points on the model torso surface.

3. Results and Conclusion

In this study, two different positions of the acute ischemic lesion in the heart ventricles were considered, as shown in Fig. 2. There is assumed the maximum degree of injury, so that the cells in the ischemic zone are persistently depolarized, while the intracellular spaces of the ischemic and normal tissues are electrically connected. The electric generator on the ischemic zone boundary produces the external electric field and corresponding shift of the S-T segment in the body surface electrocardiograms.

The unipolar potentials at the aforementioned Frank electrode points were obtained by the model for the considered cases of acute ischemia and used to determine the heart vector components and the position of the generator midplane during the middle part of the S-T period of the cardio-cycle. Calculations were carried out by means of equations (1) and (2) for corresponding orientations of the heart vector, then the intersection of the generator midplane with this vector was determined by equation (3).

The resulting positions of the generator midplane are shown in Fig. 2 by the dashed straight line. As well, these results are illustrated in the decarto-map format (see [2]) in Fig. 3, where the recognized injury regions are indicated on the surface of the imaging sphere with the radius of 6 cm enclosing the heart. The regions under consideration are bounded by the circular line of intersection of the generator midplane and imaging sphere. This sphere is cut along the right meridian, unrolled and projected onto a plane in the isoareal form. It is seen that the determined zones rather well correspond to the true localizations of the injury regions preassigned in the model. In particular, good sensitivity of the midplane position to the displacement of the injury zone in the heart vector direction was obtained (such information cannot be delivered by corrected orthogonal vectorcardiographic systems).

Fig. 1. Localization of the simulated acute ischemic lesion (crossfilled region) for the cases of injury in the subendocardium of the ventricular septum (left) and subepicardium of the left ventricular free wall (right). The heart cross-section close to the transversal plane of the chest is sketched. The heart vector D (arrow) lying in the same plane, and the generator midplane trace (dashed straight line) obtained by the described method are indicated.

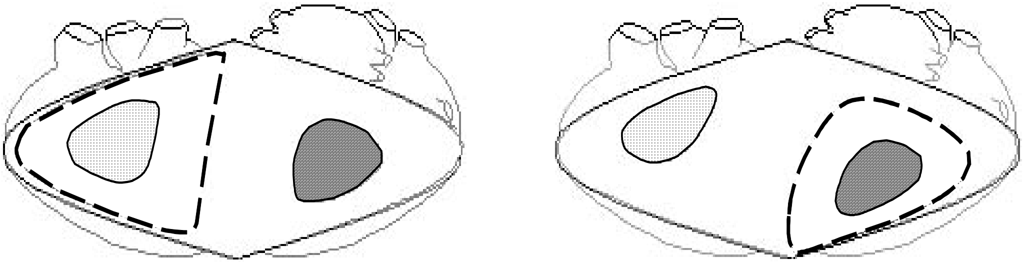

Fig. 2. Decartograms for a time instant at the middle of the S-T period for the same simulated cases presented in Fig. 2. The dashed line indicates the intersection of the generator midplane with the imaging sphere. The projections of positive and negative sides of the injured zone are shown by dark and light filled regions, respectively.

The accuracy of the method presented can be further improved by means of additional adjustment of the parameters in equations of the type (2) with taking into account in more detail the anthropometric parameters and individual position of the heart.

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