Modified Inverse Solution to One Dipole for Location of Lesions with Changed Repolarization

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Abstract

The aim of the simulation study was to modify the method for localization of one lesion with changed repolarization by inverse solution to one dipole to improve the localization error especially in the cases of large and transmural lesions.

The effect of one lesion was suggested to be represented by group of neighboring dipoles with the mean adjacent distance of 1 cm. The influence of number of dipoles in the group on the localization error was studied on simulated ECG data using analytically defined model of heart placed in inhomogeneous torso model. For comparison, analogical study was performed on ECG data simulated by the ECGSIM software.

The mean localization error for large/transmural lesions improved significantly from 2.3 cm to 1.0 cm or 1.3 cm for analytical or ECGSIM heart model when the group of five neighboring dipoles was used in the inverse solution instead of one dipole. The suggested method makes it possible to obtain comparable accuracy of the inverse solution regardless of the size of modeled lesion.

1. Introduction

Repolarization changes cause the changes in ECG signals during STT time interval, which reflect also in integral body surface potential maps (IBSPMs) [1]. The effect of changes of action potential duration can be observed in difference integral maps (DIMs) calculated from the difference between the IBSPM measured during manifestation of repolarization changes and IBSPM measured during normal heart activity as it was described in [2]. The method for non-invasive assessment of regions with changed repolarization (lesions) using DIMs was presented in [3]. The location of the lesion was represented by one inversely estimated dipole computed from DIM. In simulation study the localization error about 1cm was obtained for small and medium subendocardial or subepicardial lesions, but significantly larger error from 2 to 4 cm was obtained for large and transmural lesions.

In order to reduce the localization error for large and transmural lesions, in this study the former inverse procedure to one dipole was modified in such a manner, that the modeled lesion was assumed to be represented by a group of neighbouring dipoles.

2. Material and methods

2.1. Analytical heart model

The analytically defined model of the heart ventricles consisting of 1 mm^3 elements described in [4] was used in the simulation study. The ventricular walls were divided into five layers with different action potential duration, which differ from 240 to 282 ms. The activation of the myocardium started in selected points of ventricular endocardium according to the experimental results of Durrer [5]. The activation propagation velocity in specific parts of the endocardial layer was simulated three times higher than in the rest of the ventricular volume and represented the conduction system in the real heart. The spread of activation was simulated according the Huygens principle of wave propagation and computed using the principles of a cellular automaton.

Sixty six lesions with changed repolarization were modeled as caps of a sphere or caps of an ellipsoid located in myocardial areas typical for stenosis of three main coronary vessels: anterior - in the region supplied by LAD, posterior - in the region supplied by LCx and inferior - in the region supplied by RCA.

According to the size of the affected ventricular volume the lesions were denoted as small (0.5-1% of the volume), medium (2.5-6%) and large/transmural (8-14%). Mean radius of small lesions was 1.4 cm, of medium lesions 2.4 cm and of large/transmural lesions 3.3 cm. The repolarization changes in the lesions were modeled by shortening the action potential duration by 20%. The position of each lesion was defined by the gravity center of the affected region.

The equivalent heart generator was modeled in the form of a multiple-dipole. To compute the IBSPM and the DIM the heart model was inserted into a realistically shaped inhomogeneous torso derived from the Dalhousie

torso model [6]. Models of lungs with conductivity four times lower and ventricular cavities with conductivity three times higher than the average conductivity of the torso represented its main electrical inhomogeneities. Body surface potential map *bm*(t) in each time instant *t* was computed in 64 leads [7] on the torso surface using boundary element method (BEM) according to the equation

$$bm(t) = Ag(t) \quad (1)$$

where *A* is the time independent transfer matrix, which represents the properties of the inhomogeneous torso as the volume conductor and *g*(t) is the multiple dipole generator in particular time instant *t* of the heart activation. DIM *Δim* was computed by subtraction of STT integral map obtained during normal activation from STT integral map obtained during activation of the ventricular model with pathological lesion.

$$\Delta im = A\int_{STT} g_p(t) - A\int_{STT} g_n(t) = A(s_p - s_n) = A\Delta s \quad (2)$$

where g_n(t) and g_p(t) are multiple dipole generators representing the normal and pathological activation of the ventricular myocardium, respectively and *Δs* represents integral multiple dipole generators characterizing only the changes of the electrical activity in the modeled lesion. Computed DIM was used as the input data for the inverse localization of the lesion.

2.2. Inverse solution to a group of dipoles

The inverse solution to one dipole is based on the assumption that the local lesion with changed repolarization occupies such a small area that its effect can be represented by one dipole located at the centre of the lesion. To keep the inverse problem linear [8, 9], the inversely estimated integral dipole representing the equivalent integral generator (EIG) is computed in predefined positions in the modeled myocardium with mean distance of adjacent positions about 1cm. In this study we examined the possibility to represent the modeled lesion by up to 6 dipoles in adjoining positions. To each predefined position six nearest neighbors were found. Then all combinations of two to six positions of dipoles were examined as possible EIG.

Analogically to equation (2) the transfer matrix A_c was computed for each combination of positions of dipoles. Then the EIG was computed for each combination

$$EIG = A_c^+ .\Delta im \quad (3)$$

where A_c^+ is a pseudoinverse of the transfer matrix corresponding with particular combination.

The best EIG was chosen for each modeled lesion and for each number of neighbors according the criterion of the minimum of the relative residual error between the original input DIM and the map generated by the inversely estimated EIG.

The localization error of the inverse solution was computed as the distance between the defined position of the modeled lesion and the weighed center of the inversely estimated group of dipoles. The position of each member of the group of dipoles was weighed by its dipole moment, so that the center of the group was assigned closer to dipoles with bigger dipole moment.

The inhomogeneous torso model and the analytical model of ventricles used in the inverse solution are in the Figure 1.

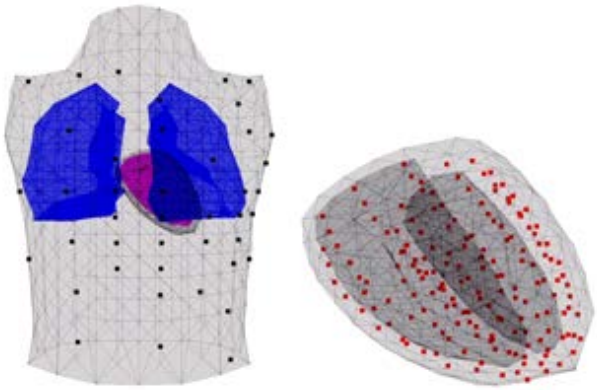

Figure 1. Left: Inhomogeneous torso model used with analytical heart model. Black dots indicate the anterior positions of measuring leads in which the DIM was computed. Right: Analytical heart model with predefined positions for inverse localization of dipoles representing the modeled lesion.

2.3. ECGSIM heart model

The suggested inverse method for identification of one lesion was then applied to another set of input data computed using the ECGSIM software [10]. In this model not only geometrical models of heart and torso are different from the model described in section 2.1., but also the equivalent heart generator is described by the transmembrane potential at the surface bounding the heart model and by the time instants when it occurs in particular part of the surface. The regions with shortened action potential duration by 20% were modeled in anterior, inferior and posterior parts of the left ventricle in similar positions as in the analytical heart model, 24 lesions in total. The position of each lesion was defined by selecting one epicardial or endocardial point on the triangulated surface of modeled ventricles that was assigned as the centre of lesion. The radius of the lesions varied from 1 cm for small lesions to 2 cm and 3 cm for medium superpicardial or subendocardial lesions. Transmural lesions were created also with a radius of 3 cm.

For each modeled lesion the DIM was computed in the same 64 measuring leads system as for the analytical heart model and used as the input data for the inverse

solution. Because the inverse solution was computed in the same way as for the analytical heart model, the predefined dipole positions within the modeled myocardium were similarly defined with the mean distance of adjacent positions of about 1cm.

The inhomogeneous torso and heart model used in ECGSIM program and predefined dipole positions for inverse solution are shown in the Figure 2.

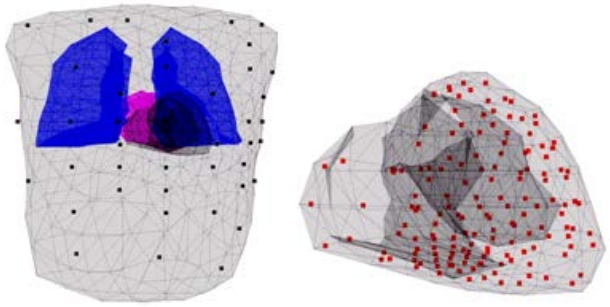

Figure 2. Left: Inhomogeneous torso model used with ECGSIM heart model. Black dots indicate the anterior positions of measuring leads in which the DIM was computed. Right: ECGSIM heart model with predefined positions for inverse localization of dipoles representing the modeled lesion.

3. Results

3.1. Analytical heart model

For the analytical heart model 66 lesions with changed repolarization were modeled: 24 small, 28 medium and 5/9 large/transmural. The mean localization error for each size of lesion and for particular number of neighboring dipoles representing the lesion is summarized in Figure 3.

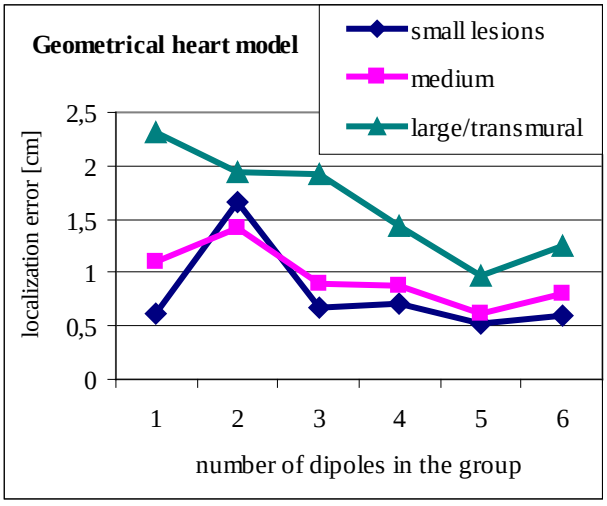

Figure 3. Mean localization error of lesions of three sizes as a function of the number of neighboring dipoles representing the lesion in the analytical heart model.

3.2. ECGSIM heart model

Using the ECGSIM software 24 lesions with changed repolarization were created: 6 small, 12 medium and 6 transmural. The mean localization error for each size of lesion and for particular number of neighboring dipoles representing the lesion is summarized in Figure 4.

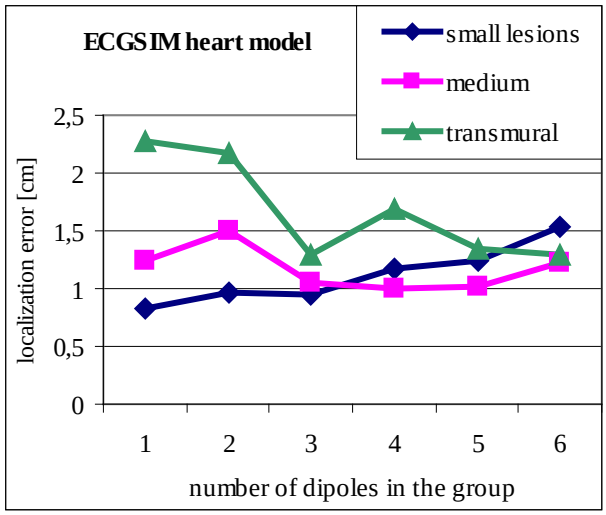

Figure 4. Mean localization error of lesions of three sizes as a function of the number of neighboring dipoles representing the lesion in the ECGSIM heart model.

4. Discussion

As can be seen from the results in Figure 3 and Figure 4, the use of more than three neighboring dipoles in the inverse solution decreased the localization error of large/transmural lesions. The best results for both heart models were obtained when the group consisted of five dipoles. When the group of five neighboring dipoles was used in inverse solution the localization error for large/transmural lesions decreased significantly from 2.3 cm to 1.0 cm for the analytical heart model (Figure 5) and from 2.3 cm to 1.3 cm for ECGSIM heart model. For lesions of medium size the localization error decreased only slightly and for small lesions the error increased for ECGSIM heart model.

When the inverse solution to one dipole was computed the mean localization errors varied for different sizes of lesions from 0.8 to 2.8 cm or from 0.8 to 2.3 cm for analytical or ECGSIM heart model respectively. For the group of five dipoles representing the modeled lesion the range of mean localization errors for different sizes of lesions decreased to 0.3 cm (0.6-0.9 cm or 1.0-1.3 cm resp.) for the both heart models. It means that the comparable values of localization error were achieved regardless of the size of the modeled lesion what is useful for practical application when the size of the affected area

is a priori not known.

The reason of the large localization errors in the inverse solution to one dipole for the large/transmural lesions inhered in the fact that their size was not small enough to fulfill the assumption of being represented by a single dipole, mentioned in section 2.2. The group of five dipoles seems to be better representative for most of the modeled lesions in this study. Further increase of the number of dipoles in the group need not yield to additional decrease of the localization error as indicate the values in the charts in Figure 3 and Figure 4 for six neighboring dipoles.

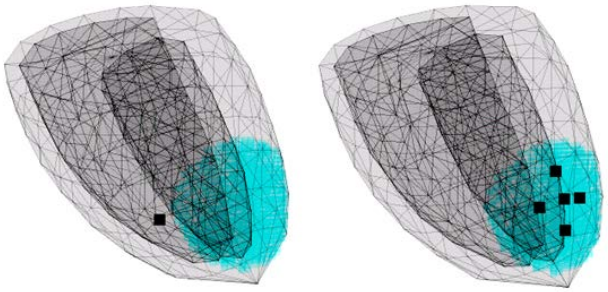

Figure 5. Inversely estimated position of modeled transmural anterior lesion in the analytical heart model when the effect of the lesion is taken to be represented by one dipole (left) and by a group of five neighboring dipoles (right). Black dot/dots indicate the inversely estimated dipoles; cyan color marks the modeled lesion.

5. Conclusion

Modification of the inverse solution for estimation of the position of single local area with changed repolarization was suggested to reduce the unacceptable large localization errors obtained from the inverse solution in case of large/transmural lesions. In the modified method, instead of one dipole a group of several neighboring dipoles is used to represent the lesion.

The method was applied on analytical heart model with simplified shape of ventricles and multiple-dipole equivalent heart generator and then tested on realistically shaped model of ventricular cavities and equivalent heart generator based on the model of transmembrane potential spreading over the heart surface.

The use of a group of five neighboring dipoles in the inverse solution using two different heart models resulted in significant decrease of the mean localization error of large and transmural lesion positions to 1.0 or 1.3 cm, which is comparable with the localization error for small lesions.

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References

[1] Hanninen H, Takala P, Rantonen J, Makijarvi M, Virtanen K, Nenonen J, Katila T, Toivonen L. ST-T integral and T-wave amplitude in detection of exercise-induced myocardial ischemia evaluated with body surface potential mapping. Journal of Electrocardiology 2003;36:89-98.

[2] Trudel MC, Dube B, Potse M, Gulrajani RM, Leon LJ. Simulation of QRST integral maps with a membrane-based computer heart model employing parallel processing. IEEE Transactions on Biomedical Engineering 2004;51:1319-1329.

[3] Tysler M, Kneppo P, Turzova M, Svehlikova J, Karas S, Heblakova E, Hana K, Filipova S. Noninvasive assessment of local myocardium repolarization changes using high resolution surface ECG mapping. Physiological Research 2007;56:S133-S141.

[4] Szathmary V, Osvald R. An Interactive Computer-Model of Propagated Activation with Analytically Defined Geometry of Ventricles. Computers and Biomedical Research 1994;27:27-38.

[5] Durrer D, van Dam RT, Freud GE, Janse MS, Meijler FL, Arzbaecher RC. Total Excitation of the Isolated Human Heart. Circulation 1970;41:899-912.

[6] Horacek BM. Numerical Model of an Inhomogeneous Human Torso. Adv. Cardiol. (Body Surface Mapping of Cardiac Fields) 1974;10:51 - 57.

[7] Fereniec M, Kacprzak M, Karpinski G, Maniewski R, Zbiec A, Ircha D. The 64 channel system for high resolution ECG mapping. Computers in Cardiology 2001;28:513-515.

[8] Svehlikova J, Lenkova J, Turzova M, Tysler M, Kania M, Maniewski R. Influence of individual torso geometry on inverse solution to 2 dipoles. Journal of Electrocardiology 2012;45:7-12.

[9] Oostendorp T, Van Oosterom A. Source Parameter Estimation in Inhomogeneous Volume Conductors of Arbitrary Shape. IEEE Transactions on Biomedical Engineering 1989;36:382 - 391.

[10] van Oosterom A, Oostendorp TF. ECGSIM: an interactive tool for studying the genesis of QRST waveforms. Heart 2004;90:165-168.

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