Individual vs. Adjusted Standard Torso Model in the Solution of the Inverse Problem in Electrocardiology

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Abstract. *To find a solution of the inverse problem of electrocardiography as precisely as possible, individual geometry of a patient's torso is required. The most precise method to obtain both the surface geometry and the position and shape of inner organs is an imaging method such as CT or magnetic resonance. In our work we tried to find a suitable surrogate torso model enabling to avoid the use of demanding imaging techniques and offering accurate enough results. We suggested doing that by adapting both the torso dimensions and positions of electrodes on an already existing parametrized standard torso model.*

Keywords: Individual Geometry, Standard Torso Model, Inverse Solution

1. Introduction

The topic of model geometry in the inverse problem of electrocardiography has been discussed for a long time. One of the overall accepted observations from numerous studies is that it is necessary to include inhomogeneities to the torso model – at least those most prominent, such as lungs and ventricular cavities, and that it is not possible to acquire an acceptably accurate result without individual geometry of a patient's body and the position of inner organs [1][2]. Other important observation is that the position of electrodes is to be adjusted to the heart position rather than to the reference point of the fourth intercostal space [3][4].

As opposed to imaging methods that have been used so far to acquire an individual human torso model (MRI, CT) and that require to construct a new torso model for each patient, different approach is employed in this study: a parametrized generic torso model that can be adjusted to fit the patient's individual torso shape. To do so, we pursued some procedures for torso- and electrodes position- adaptation. Our aim was to verify if a torso obtained by an imaging technique can be replaced with sufficient accuracy by a torso constructed in a much simpler way – using an adjusted standard realistically shaped torso model.

2. Subject and Methods

Simulation of body surface potentials

To simulate body surface ecg potential maps, we modeled 12 small ischemic lesions shaped as spherical caps, placed in the left ventricular myocardium, close to the positions of 3 main coronary arteries: Cx (circumflex artery), RCA (Right coronary artery) and LAD (left anterior descending artery). In each of these three positions, two subendocardial and two subepicardial lesions were modeled.

The model we used to simulate activation of ventricular myocardium used simplified analytical geometry of ventricles. The myocardium volume was divided into 1x1x1 mm cubic elements; each of them was assigned a realistically shaped action potential (AP) characteristic. In the elements within the ischemic lesion the AP was shortened to mimic the presence of ischemia. Activation sequence during ventricular depolarization and depolarization was simulated in normal ventricles as well as in ventricles with ischemic lesions. Boundary element method was used to sum the electrical activity of the elementary sources and to compute the cardiac electric field in the torso volume conductor.

From simulated torso surface potentials QRST integral maps were constructed. Then, for every lesion a difference QRST integral map (DIM) was computed, by subtracting a map computed for normal activation from the map computed in case of particular ischemic lesion incorporated in the ventricular model. This DIM characterizes topographical changes in surface cardiac electrical field due to the ischemia.

Three torso models were used in the forward computations. The torso shape models were obtained using MRI scanning from 3 patients, 1 woman and 2 men, while they had electrodes put on [2]. Realistically shaped lungs and analytical model of ventricular myocardium with cavities were inserted into the torso. The set of inhomogeneities (lungs and ventricles) was the same for all the torso models; the electrical conductivity assigned to the inhomogeneities was 0.25 and 3 times the mean electrical conductivity of the torso, respectively. The original positions of electrodes, as scanned by MRI, were used in the forward computations as well. The lead system used was the modified Amsterdam 62 system used in Warszaw [5].

Solution of the inverse problem

For the inverse computations, 4 torso models (see Fig 1.) were used for each of the patients: a) torso 1 – the original torso model with electrodes as scanned using MRI, b) torso 2 – a standard torso model (STM), actually a modified Dalhousie torso [6] with inhomogeneities and regularly placed electrodes, c) torso 3 – an adjusted STM ”stretched“ to fit the shape of the original torso model as good as possible, with regularly placed electrodes, d) torso 4 – an adjusted STM, similar to torso 3 but with electrodes positions adjusted according to the original torso by horizontally shifting the electrodes of the original torso set-up so they would lie on the surface of the adapted torso model.

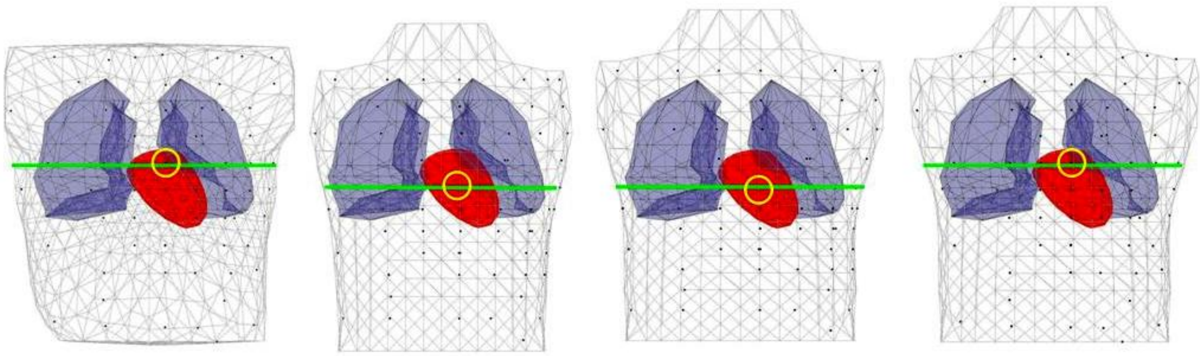

Fig 1. Four torso models created for patient 3 and used in the inverse solution. Left to right: torso 1, torso 2, torso 3 and torso 4 (described in text). The analytical model of ventricular cavities is surrounded by the epicardium, forming thus the ventricular myocardium model with lungs model on both sides. The square marks depict the electrodes of the used lead system, the line marks the level of the V2 lead (circled).

The model of ventricular myocardium consisting of 168 evenly lain segments as well as the models of lungs and ventricular cavities was the same for all torso models (torso 1 to 4) used in the inverse computations. In our inverse computations, difference integral maps were used as input data. Therefore, we searched the equivalent electrical source in form of an equivalent integral dipole (EID) that represented the difference over the whole QRST period between ischemic and normal activation. The position of EID was restricted to the centres of defined segments of ventricular model. The dipole moment of EID was computed for each of 168 possible positions using the equation

$$d_i = T_i^+ p \tag{1}$$

where d_i is the dipole moment of the EID placed in the *i*-th heart segment (i =1,2,...,168), T_i^+ is the pseudoinverse of the transfer matrix T_i describing how the EID positioned in the *i*-th segment and integral body surface potentials in measured points are related, *p* is the vector of integral potentials in measured points (corresponding DIM). To compute the equation (1), singular value decomposition was applied on the transfer matrix.

To assess the precision of the results for all 4 torso models used in the inverse computations for each of the patients, we evaluated a localization error parameter defined as the distance between the inversely localized EID and the gravity center of the simulated lesion that was looked for. The average localization error for all 12 lesions was then evaluated [5].

3. Results

As seen in Table 1, the best result for each patient was obtained when the torso 1 model was used in the inverse computation. In all three patients, the second best result was for torso 4, in case of patient 3, the result for torso 4 and torso 1 were the same. For patient 1, the third best result was achieved using torso 3, with the worst result being for the torso 2. In patients 2 and 3, the order of the torso models with two worst results was changed – the results for torso 2 were better than for torso 3.

Table 1. The results of the inverse computations. For each patient, four torso models were used in the inverse solution. For every torso set-up 12 DIMs (one for each lesion) were used as the input and the resulting localization error was averaged.

Mean localization error [cm]	Torso 1	Torso 2	Torso 3	Torso 4
Patient 1	0.59	2.88	2.62	0.97
Patient 2	0.49	1.04	1.31	0.76
Patient 3	0.49	1,70	1.80	0.49

4. Discussion and Conclusion

In this work we tried to suggest a surrogate torso model that could be used in the inverse solution with one dipole instead of a torso obtained from an imaging technique with sufficiently accurate results. We investigated the situation in 3 MR – scanned torso surface models with electrodes placed on them. For each original torso we suggested 3 types of simplified torso models to compare the accuracy of lesion location.

Our hypothesis was that the results of the inverse solution would show the least localization error parameter using the original torso+electrode set-up. From the other three torso models, the best results were expected for the torso with adapted electrode positions (torso 4). This proved to be true in all the three patients. The mean localization error was only 1.53 and 1.65 times higher for patient 1 and 2, respectively, and the same for patient 3, when compared to the mean localization error obtained by using the original torso model. The mean localization error values using the original torso model were not zero due to other factors such as the error coming from rough segmentation of the heart.

We supposed, that the third best results will be achieved using the torso model with adapted dimensions and with regularly distributed electrodes (torso 3) and that the worst results will come with using the STM with regular electrodes placement (torso 2). This, however, was only true in patient 1 (a woman) while the other two men patients showed the localization results in reversed order. Determining the reason for this behaviour requires further observations.

It is worth noticing that the span between the value of the mean localization error for the adjusted torso with adjusted electrode positions and the value for the third best result (whichever torso model produced it) was rather large. Apparently, adjusting only the outer torso shape does not influence the results of the inverse solution as much as adjusting both, the torso shape and the electrode positions. However, it is not possible to omit the procedure of adjusting the torso model shape, since the standard torso model may not be large enough for the range of electrodes of the original lead system positioning to be kept.

Based on the obtained results we can conclude that using some standard torso model instead of realistic patient geometry obtained by imaging techniques can give sufficiently accurate inverse solution only if the dimensions of the standard torso are properly adjusted in accordance with patient torso dimensions and electrodes are positioned according to their original placement on the patient's body as well. It can be expected that the results could be even better if more precise position and orientation of the heart is available. That will be the subject of our future study.

Acknowledgements

The work has been supported by research grant No. 2/0210/10 of the Vega Grant Agency in Slovakia. We also thank to prof. van Oosterom and R. Hoekema for providing the data of three torso models obtained from MRI and used in the forward computations of surface potentials.

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