ACCURACY OF INVERSE LOCALIZATION OF PREEXCITATION SITES. A SIMULATION STUDY

Milan Tyšler, Marie Turzová, Jana Švehlíková, Mária Tiňová
Institute of Measurement Science SAS,
Dúbravská cesta 9, 842 19 Bratislava, Slovak Republic
E-mail: umertysl@savba.sk

ABSTRACT

In the paper, the accuracy which can be expected in inverse localization of preexcitation sites was studied on a computer model. Surface ecg potentials simulated by a forward model, realistic torso geometry and fixed-location multiple dipole model with 39 segmental dipoles were used in the inverse computations. Neglecting of torso inhomogeneities, use of limited number of leads, errors in heart position estimation and noise in the ecg potentials were analyzed as factors influencing the accuracy of the localization.

While the mean localization error computed from 198 leads in inhomogeneous torso was less than 0.6 cm, neglecting of torso inhomogeneities brought an error increase to 1.1 - 1.3 cm and the results were very similar for 198, 63 or 26 leads. When errors in heart position estimation were included, the mean errors for respective lead systems were 0.9, 1.3 and 1.0 cm in inhomogeneous torso and raised up to 1.6, 1.9 and 1.4 cm in homogeneous torso model. When only the noise was added to ecg signals, its standard deviation had to be less than 15 μV (about ±50 μV peak-to-peak) to hold the mean errors less than 1.5 cm. However, when all the factors were combined, only noise with $\sigma \leq 5$ μV was acceptable to get mean error 1.6 - 2.0 cm. For the noise with $\sigma \leq 5$ μV the 26-lead system performed better than the 63-lead system, while for higher noise the 63-lead system gave better results.

1. Introduction

The correlation between spontaneous and invoked body surface potential maps has been successfully used to locate early ventricular preexcitation sites during ablation surgery [1]. It is hoped that inverse techniques based on body surface potential and torso geometry knowledge could help in computer-supported localization of pathological initial activation sites. The achievable accuracy of the inverse localization based on the multiple dipole model was examined by computer simulation. In previous studies [2] the influence of a limited number of leads and simplification of torso structure was tested. In this study the synergetic effects of the mentioned factors combined with inaccurate knowledge of the heart position and presence of noise in ecg potentials were analyzed in cases with single preexcitation site on the atrio-ventricular ring.

2. Method

The multiple dipole (MD) model with predefined 39 dipoles representing heart segments shown in Fig.1 was used in the inverse computations. The m components of segmental dipole moments $\boldsymbol{M}$ were estimated from n surface potentials $\boldsymbol{\Phi}$ using a pseudo-inverse of the transfer matrix $\boldsymbol{T}$, which represents the used torso geometry [3]:

$$\boldsymbol{M} = \boldsymbol{T}^{+} \boldsymbol{\Phi}$$

where $\boldsymbol{T}^{+}$ was obtained using the truncated singular value decomposition of the transfer matrix $\boldsymbol{T}$:

$$\boldsymbol{T} = (\boldsymbol{ASB}^{t})$$

Then $\boldsymbol{T}^{+} = (\boldsymbol{A}_r \boldsymbol{S}_r \boldsymbol{B}_r^{\,t})^{+} = \boldsymbol{B}_r \boldsymbol{S}_r^{\,+} \boldsymbol{A}_r^{\,t}$

where $\boldsymbol{A}$ is eigenvector matrix of $\boldsymbol{TT}^{t}$

$\boldsymbol{B}$ is eigenvector matrix of $\boldsymbol{T}^{t}\boldsymbol{T}$

r is rank of the truncation, $r<m$

$\boldsymbol{S}$ is diagonal matrix,

$s_{ii}^{+} = 1/s_{ii}$ if $s_{ii} \neq 0$, else $s_{ii}^{+} = 0$.

Maximal segmental dipole moment during the initial period of activation was used to detect the segment on the atrio-ventricular (AV) ring with the earliest activation.

To get well defined ecg potentials needed as input for the inverse computations, a finite element model of the myocardium [4] was used to simulate activation sequences, each initiated in a single site on the AV ring. Corresponding ecg potentials were computed on the surface of a realistic inhomogeneous torso - the same as the one used in the inverse calculations. In the study, results from a set of 8 preexcitation sites along the AV ring were analyzed.

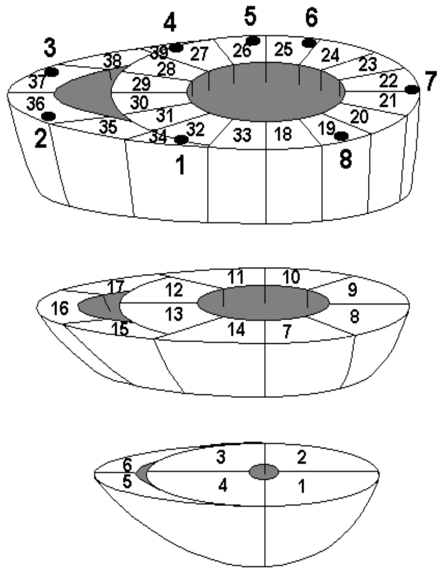


Fig.1. 39 heart segments represented by segmental dipoles and 8 preexcitation sites used in the study.

The influence of noise was studied using normally distributed deviations added to the surface potentials.

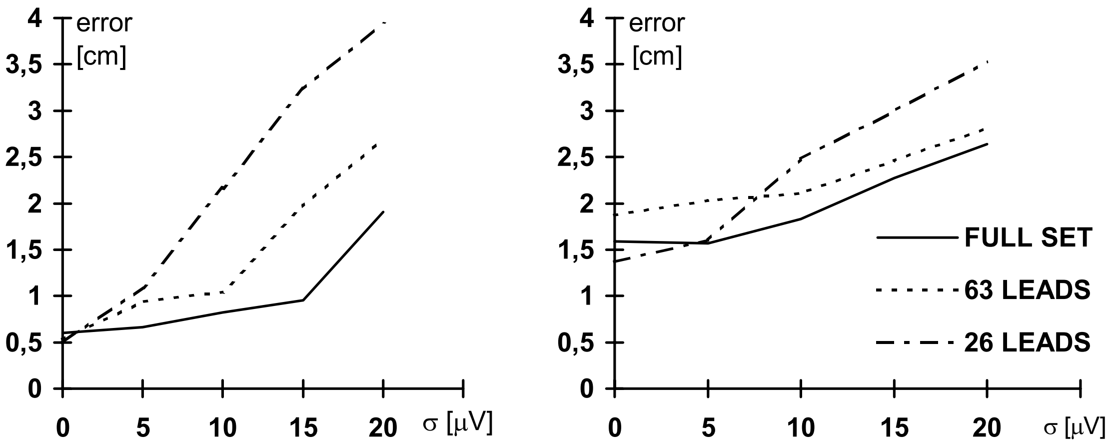
2. Results

The mean localization error of the method was 0.5 - 0.6 cm for a full set of torso potentials and for both tested lead systems, when the inhomogeneous torso was used. It is in agreement with the previously reported results.

Examples of the inverse localization results obtained supposing some errors in heart position are shown in Table 1. When shifts or rotations of the heart model by approximately 1 cm were included, the mean errors for tested lead systems were 0.9 - 1.3 cm and increased to 1.4 - 1.9 cm in homogeneous torso model. The 63 lead system proved to be the most sensitive to the incorrect heart model position. In several individual cases errors greater than 3 cm were obtained.

Table 1. Sample mean localization errors [cm] for incorrect heart positions

heart position	inhomogeneous torso			homogeneous torso		
	198 leads	63 leads	26 leads	198 leads	63 leads	26 leads
shift z-1cm	0.9	2.5	1.0	1.7	3.3	1.1
rotation y 20 deg.	1.0	0.7	1.5	1.8	1.2	1.6
rotation z 20 deg.	0.9	0.8	0.5	1.3	1.0	1.5



a) b)

Fig. 2. Mean localization error in dependence on σ of the noise present in ecg signals:
a) computed in inhomogeneous torso with the precise heart position,
b) computed in homogeneous torso with inaccurate heart position

When only noise was added to the ecg signals, its standard deviation had to be less than 15 μV (about ±50 μV peak-to-peak) to hold the mean localization errors less than 1.5 cm (Fig.2a). However, when all the examined error factors were combined, only the noise with $\sigma \leq 5$ μV (less then ±20 μV peak-to-peak) was acceptable to get mean errors 1.6 - 2.0 cm (Fig.2b). With an increasing amplitude of the noise, in a few cases distant segments were identified, what yielded an unacceptable localization error. When all the examined factors were present, for a small noise with $\sigma \leq 5$ μV the 26 lead system performed better than 63 leads, while for noise with $\sigma > 10$ μV the 63 lead system gave better results.

The results for all combinations of the studied error factors are shown in Table 2, the mean localization errors for the optimal and the worst acceptable conditions are highlighted.

3. Discussion

In this study, unlike previous reports, the surface potentials were averaged over 10 or 20 ms interval to minimize the influence of noise. When noise with uniform or Gaussian distribution was added to the surface potentials, the use of averaged body

Table 2. Mean localization errors [cm] for combinations of error factors
(simplified torso structure, limited number of leads, changed heart position, noise)

		inhomogeneous torso			homogeneous torso		
heart position	noise σ [μV]	198 leads	63 leads	26 leads	198 leads	63 leads	26 leads
exact	0	0.6	0.5	0.5	1.3	1.2	1.1
	5	0.7	0.9	1.1	1.3	1.1	1.4
	10	0.8	1.0	2.1	1.3	1.32	2.2
incorrect	0	0.9	1.3	1.0	1.6	1.9	1.4
	5	1.0	1.5	1.5	1.6	2.0	1.6
	10	1.1	2.1	2.6	1.8	2.1	2.5

surface potentials significantly improved the accuracy and stability of the inverse solution. The averaging interval 15-35 ms was selected, however, earlier intervals performed better for small noise ($\sigma < 5$ μV). The 20 ms interval length was used to filter the 50 Hz interference.

4. Conclusions

For the above described method, the knowledge of torso geometry, heart position and preferably also geometry of the main torso inhomogeneities with accuracy better than 1.0 cm and the noise in ecg signals with $\sigma \leq 5$ μV were necessary to obtain mean localization errors less than 2 cm. Precise knowledge of the heart position proved to be crucial for the correct detection of the initial activated segment. On the other hand, the usage of limited number of 63 or even 26 leads did not deteriorate the results significantly and seemed to be reasonable.

Acknowledgments

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