# Dipole Modeling in Electrocardiographic Classification of Acute Ischemia

M Stenroos[1], M Lindholm[1], H Hänninen[2], I Tierala[2],
H Väänänen[1], T Katila[1]

[1]Laboratory of Biomedical Engineering, Helsinki University of Technology, Finland
[2]Division of Cardiology, Helsinki University Central Hospital, Finland

## Abstract

*Acute ischemia is commonly detected with electrocardiography. In diagnostic use 12–lead ECG is the standard setup, during monitoring reduced lead sets are often used. In this work we tested, whether equivalent dipole orientation can be used for ischemia classification.*

*120–channel Body Surface Potential Mapping (BSPM) was measured from 22 ischemic patients during PTCA operation. ST60 amplitude map was chosen as classification parameter. The equivalent dipoles between different coronary arteries differed in position and especially in orientation. Dipole moment orientation was used for classification of ischemic arteries. In addition to BSPM, 12–lead, and some other small lead sets were used for fitting.*

*The best classification results (not counting BSPM) were obtained with modified 12–lead layout (LAD 100%, LCX 86%, RCA 100%). The results show that dipole modeling is a potential tool for ischemia classification.*

## 1. Introduction

Acute ischemia is commonly detected with electrocardiography (ECG). Development of ischemia is also followed with ECG monitoring. In diagnostic use 12–lead ECG is the standard setup, during monitoring reduced lead sets are often used. The monitored ECG parameter is typically ST segment elevation and depression.

Studies on optimal lead selection [1–3] have shown that electrodes outside 12–lead setup improve diagnostic performance. In these studies, the results are usually derived straight (e.g. maximum ST–deviation) or statistically (t–test) from the data measured during percutaneous transluminary coronary angioplasty (PTCA). Body Surface Potential Mapping (BSPM), despite being too cumbersome for clinical use, is a valuable tool for straightforward comparison of different leads. It was used e.g. in studies [1, 2].

Although 12–lead ECG is strongly based on concept of a dipole as equivalent cardiac generator, dipole modeling has, to authors knowledge, not been used in classification of ischemia. However, ST segment BSPM patterns measured during severe ischemia show dipolar characteristics, and mean map orientation depends on culprit artery. In this work we tested, whether equivalent dipole orientation can be used for ischemia classification.

## 2. Methods

### 2.1. Data set and preprocessing

We measured 120–channel BSPM during PTCA operation. Our electrode layout is shown in Fig. 1. Total number of patients was 22 (19 M, 3 F). Nine of the patients had suffered myocardial infarction. The balloon was inflated either in left anterior descending (LAD, n = 8), left circumflex (LCX, n = 7), or right coronary artery (RCA, n = 7). Measurements were done before and during the inflation.

The data were pre-processed semi–automatically: 50 Hz adaptive filter was applied, baseline was removed with spline fitting, and data were averaged (moving average of 11 beats). Bad channels were interpolated from good ones by minimizing the surface laplacian [4]. Typical BSPM parameters, e.g. QRS integral, T-apex-end integral and ST60

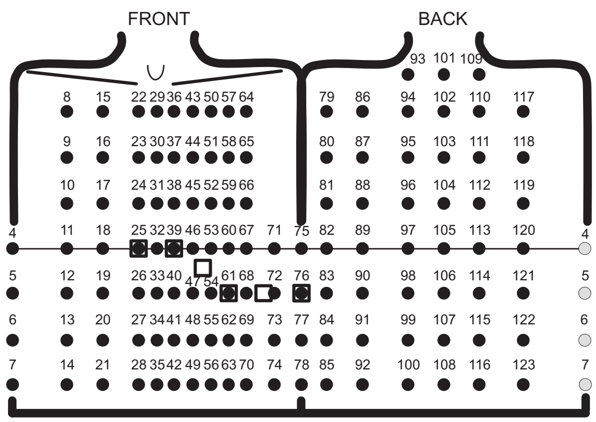


Figure 1. Helsinki BSPM electrode layout. Standard chest leads are marked with squares.

amplitude (J–point + 60 ms) maps were calculated, and further analysis was performed with delta maps (map before balloon inflation subtracted from map during ischemic state). In order to be able to reconstruct reduced lead sets correctly, the maps were interpolated to torso surface with the surface laplacian method.

## 2.2. Computational methods

Cardiac electric field was modelled with quasi–static Maxwell equations [5]. Poisson equation for volume conduction was converted to surface integral form by means of Green formulas [6, 7]. Surface integral equation was discretized with linear boundary element method (BEM): potential was modelled by linear combination of linearly varying nodal basis functions, and residual was weighed with point collocation technique [8]. Element integrals were calculated with DeMunck analytical formula [9]. Dalhousie standard torso was used as volume conductor model. Internal inhomogeneties were not taken into account, but anisotropic skeletal muscle layer was approximated with "torso extension" method [10]. The total number of nodes in the model was 704.

For each tested dipole position, orthogonal unit dipole potentials at selected lead positions – also called the lead fields – were calculated. Locally optimal dipole was fitted to measured data by pseudoinverting this lead field matrix. The field produced by the locally optimal dipole was compared to measured field with cost function $C$ and goodness of fit $G$:

$$C = \sum_i (\phi_{\mathrm{meas,i}} - \phi_{\mathrm{calc,i}})^2, \qquad (1)$$

$$G = \frac{\sum_i (\phi_{\mathrm{meas,i}} - \phi_{\mathrm{calc,i}})^2}{\sum_i \phi^2_{\mathrm{meas,i}}}. \qquad (2)$$

Optimal dipole position was searched by minimizing cost function with Nelder–Mead simplex method (as provided in Matlab Optimization Toolbox [11]).

Classification of test data was done against dipoles calculated from population average maps. Classification tests were done in leave-one-out manner: test patient was taken out of the set, and population average maps for each artery were calculated from rest of the set.

Dipole fitting and classification was done in three ways: In **Optimal method** the dipoles were allowed to move freely to cost function minimum, and in **Fixed method** the dipole position was fixed to midpoint of the Dalhousie model epicardium. In **Average method** optimal dipoles were fitted population average maps of each artery. Then dipole position was fixed to each of these three locations, and locally optimal dipole for test map or lead set (from now on "Test dipole") was fitted for each position. Moments of these three dipoles were compared to the dipole moments of population average maps ("Population average dipole"). Population average dipoles were be calculated both from BSPM data and from each lead set.

## 3. Results

QRS integral, T-apex-end integral and ST60 amplitude maps were used in initial dipole fitting tests. An equivalent dipole explained QRS and T-apex-end integral maps better in terms of goodness of fit than the ST60 amplitude map, but ST60 dipoles showed larger artery specific correlation and smaller variance than the dipoles calculated from other maps. Therefore ST60 amplitude was used as classification parameter. Population average ST60 maps are shown in Fig. 2. The maps resemble closely those published by Horáček et al [1]. Optimal dipoles for the average maps are shown on Dalhousie epicardium model in Fig. 3. LAD and LCX dipoles are located in the apical part of the left ventricle, not far away from each other. The RCA dipole is in basal part of the ventricles. The angle between LAD and LCX dipoles is 137 degrees, between LAD and RCA 121 degrees, and between LCX and RCA 88 degrees.

Basing on these results, cosine of the angle between dipole moment vectors – or normalized dot product – was chosen as comparison parameter between dipoles. Mean goodness of fit for each fitting method and artery is given in Table 1. As expected, the Optimal method yields best

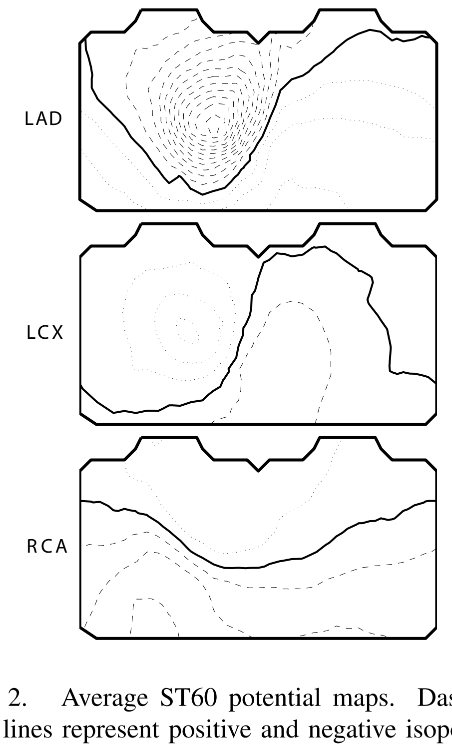


Figure 2. Average ST60 potential maps. Dashed and dotted lines represent positive and negative isopotentials, curve step is 20 $\mu V$.

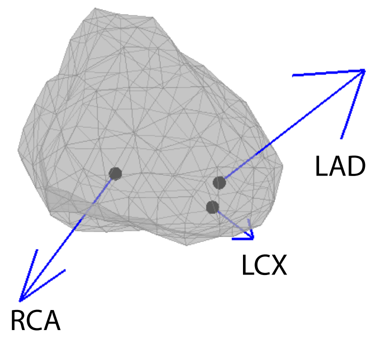


Figure 3. Optimal dipoles, frontal projection

goodnesses. Fixed method gives the goodnesses of the same order with Average method – this relates to the large variance of the dataset. LAD average and LCX average dipole positions are close to each others, so similar goodnesses for dipoles in these positions are not surprising.

Table 1. Goodnesses of fit

| | Optimal | Fixed | |
|---|---|---|---|
| LAD | 0.91 ± 0.09 | 0.86 ± 0.09 | |
| LCX | 0.85 ± 0.12 | 0.80 ± 0.11 | |
| RCA | 0.88 ± 0.12 | 0.79 ± 0.16 | |
| | Ave. (LAD) | Ave. (LCX) | Ave. (RCA) |
| LAD | 0.84 ± 0.07 | 0.86 ± 0.07 | 0.82 ± 0.08 |
| LCX | 0.80 ± 0.13 | 0.80 ± 0.12 | 0.76 ± 0.10 |
| RCA | 0.76 ± 0.16 | 0.77 ± 0.16 | 0.81 ± 0.15 |

Table 2. Cosines of the angles between dipole moment vectors

Optimal method

| | LAD mean | LCX mean | RCA mean |
|---|---|---|---|
| LAD | **0.80 ± 0.20** | -0.63 ± 0.37 | -0.33 ± 0.27 |
| LCX | -0.19 ± 0.61 | **0.46 ± 0.34** | -0.23 ± 0.40 |
| RCA | -0.21 ± 0.49 | -0.01 ± 0.48 | **0.75 ± 0.10** |

Fixed method

| | LAD mean | LCX mean | RCA mean |
|---|---|---|---|
| LAD | **0.75 ± 0.18** | -0.60 ± 0.36 | -0.37 ± 0.32 |
| LCX | -0.14 ± 0.61 | **0.46 ± 0.31** | -0.23 ± 0.44 |
| RCA | -0.30 ± 0.37 | 0.05 ± 0.44 | **0.79 ± 0.11** |

Average method

| | LAD mean | LCX mean | RCA mean |
|---|---|---|---|
| LAD | **0.77 ± 0.18** | -0.64 ± 0.34 | -0.39 ± 0.30 |
| LCX | -0.15 ± 0.61 | **0.44 ± 0.34** | -0.22 ± 0.45 |
| RCA | -0.28 ± 0.39 | 0.04 ± 0.45 | **0.78 ± 0.10** |

Variation of the dipole moment was studied by comparing the Test dipole against Population average dipoles of each artery. The results are displayed in Table 2. It is seen in the results that if Test dipole is compared to Population average dipole of some other artery, the cosine is negative or almost zero, but in comparison to Population average dipole of the same artery the cosine is clearly positive. In LCX maps the dipole orientation varies considerably more than in other arteries.

Classification of the dipole moment orientations was done by comparing Test dipoles to Population average dipoles. BSPM layout, derived 12–lead and 18–lead ECG (12 + $V_{4-6\mathrm{R}} + V_{7-9}$, [3, 12]), modified 12–lead, and lead setups proposed by Horáček [13] and Kornreich [2] were tested. In 12– and 18–lead setups the limb potentials were approximated by potentials at right and left shoulder and left hip. In our modified 12–lead setup, the left leg and $V_5$ electrodes were replaced by electrodes 21 and 98 ($V_9$) from Helsinki BSPM setup. Classification percentages are shown in Table 3. With BSPM data all the fitting methods perform equally well in terms of classification performance. In all the other setups both Fixed and Average methods perform better than Optimal method. When the Population average dipoles are fitted to BSPM data (a priori database), the classification results are better than with lead set based Population average dipoles. In general, the best results are obtained with Average method and modified 12–lead setup or Horáček setup.

Table 3. Classification percentages from all the patients. Table cells contain result for all the arteries in order LAD / LCX / RCA.

With BSPM database

| | Optimal | Fixed | Average |
|---|---|---|---|
| BSPM | 100 / 86 / 100 | 100 / 86 / 100 | 100 / 86 / 100 |
| 12-Lead | 88 / 100 / 57 | 88 / 71 / 86 | 88 / 71 / 86 |
| Mod. 12 | 100 / 57 / 57 | 100 / 86 / 100 | 100 / 86 / 100 |
| Horacek | 100 / 57 / 57 | 75 / 57 / 100 | 100 / 71 / 100 |
| 18-Lead | 100 / 57 / 86 | 100 / 71 / 86 | 100 / 86 / 86 |
| Kornreich | 100 / 43 / 0 | 100 / 29 / 14 | 100 / 43 / 86 |

With lead set database

| | Optimal | Fixed | Average |
|---|---|---|---|
| 12-Lead | 88 / 100 / 57 | 88 / 86 / 71 | 100 / 71 / 43 |
| Mod. 12 | 100 / 71 / 43 | 88 / 86 / 100 | 100 / 71 / 86 |
| Horacek | 100 / 86 / 57 | 88 / 100 / 86 | 100 / 71 / 57 |
| 18-Lead | 100 / 43 / 72 | 100 / 71 / 71 | 100 / 43 / 71 |
| Kornreich | 100 / 14 / 43 | 100 / 29 / 71 | 100 / 14 / 86 |

## 4. Discussion and conclusions

This work shows that dipole modeling can be used for ischemia classification. Of the three tested dipole fitting methods, Average method gave the best results. Optimal method worked very well for LAD ischemia, but quite poorly for LCX. This performance can be explained by the variance of the data and the lack of use of a priori information – the optimal dipole for a small lead set may be far away from the ventricles, pointing in totally wrong direction. On the contrary, in Fixed method and Average method the dipole is at least in the correct region, thus giving a possibility for somewhat realistic field reconstruction even with less than ten leads. Use of BSPM a priori knowledge with Population average dipoles improved the classification results obtained with reduced lead sets.

Best classification results are obtained with lead sets that contain electrodes outside the 12–lead setup; moving one electrode close to RCA map maximum (see Fig. 2 and one electrode to dorsal side, e.g. $V_9$, improves detection of LCX and RCA ischemia. Horáček layout performed very well, taking into account that it uses only five electrodes. Kornreich setup seems to be not suitable for use with dipole approach: LCX recognition accuracy was poor with all the fitting methods.

In order to evaluate the performance of the dipole modeling method better, and to find an optimal electrode layout for this approach, a larger dataset is needed – especially the large variance of the LCX data degrades the results. To sum this paper up, dipole modeling is a potential tool for ischemia classification, especially when combined with a BSPM database providing a priori information on Population average dipole positions for each coronary artery.

## Acknowledgements

The torso model is courtesy of Milan Horáček, Dalhousie University, Nova Scotia, Canada.

## References

[1] Horáček BM, Wagner GS. Electrocardiographic ST–segment changes during acute myocardial ischemia. Cardiac Electrophysiology Review 2002;6:196–203.

[2] Kornreich F. Identification of best electrocardiographic leads for diagnosing acute myocardial ischemia. Journal of Electrocardiology 1998;31.

[3] Wung SF, Drew B. Comparison of 18–lead ECG and selected body surface potential mapping leads in determining maximally deviated ST lead and efficacy in detecting acute myocardial ischemia during coronary occlusion. Journal of Electrocardiology 1999;32 Supplement:30–36.

[4] Oostendorp T, van Oosterom A, Huiskamp G. Interpolation on a triangulated 3D surface. J Comp Physics 1989;80:331–343.

[5] Plonsey R, Heppner D. Considerations of quasistationarity in electrophysiological systems. Bull Math Biophys 1967; 29:657–664.

[6] Geselowitz D. On bioelectric potentials in an inhomogeneous volume conductor. Biophys J 1967;7:1–11.

[7] Barnard A, Duck I, Lynn M. The application of electromagnetic theory to electrocardiology. I. Derivation of the integral equations. Biophysical Journal 1967;7:443–462.

[8] Brebbia C, J.Dominguez. Boundary Elemens: an Introductory Course. McGraw–Hill, 1989.

[9] de Munck J. A linear discretization of the volume conductor boundary integral equation using analytically integrated elements. IEEE Transactions on Biomedical Engineering 1992;986–989.

[10] Mailloux G, Gulrajani R. Theoretical evaluation of the mcfee and frank vectorcardiographic lead systems usnig a numerical inhomogeneous torso model. IEEE Transactions on Biomedical Engineering 1982;BME-29:322–332.

[11] MathWorks. MATLAB – the language of technical computing. http://www.mathworks.com/products/matlab/, April 2005.

[12] Zalenski R, Rydman R, Sloan E, Hahn K, Cooke D, Fagan J, Fligner D, Hessions W, Justis D, Kampe L, Shah S, Tucker J, Zwicke D. Value of posterior and right ventricular leads in comparison to the standard 12-lead electrocardiogram in evaluation of ST-segment elevation in suspected acute myocardial infarction. American Journal of Cardiology 1997; 79:1579–1585.

[13] Horáček BM, Warren JW, Penney CJ, MacLeod RS, Title LM, Gardner MJ, Feldman CL. Optimal electrocardiographic lead for detecting acute myocardial ischemia. Journal of Electrocardiology 2001;34.

Address for correspondence:

Matti Stenroos
Laboratory of Biomedical Engineering, PO BOX 2200, FIN - 02015 TKK, Finland
matti.stenroos@tkk.fi