# Monitoring of CRT by means of impedance multiple measurements – simulation studies



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**Abstract**. Cardiac resynchronization therapy (CRT) is a very promising treatment for patients with congestive heart failure. An Impedance Cardiography (ICG) is used for evaluation of heart mechanical activity. A simulation study with the use of a FEM model was performed. The developed realistic model of the thorax, based on CT data obtained from examination of a 68-year-old man, consisted of 212 000 tetrahedral elements. Different configurations of electrodes allowing extraction of ventricular components from impedance signal were examined. The associated sensitivity distributions were calculated for different phases of heart contraction and blood distribution inside the thorax. These sensitivity distributions were compared to that obtained for a uniform model of the thorax described by average conductivity. The performed simulation studies proved that it is possible to extract ventricular components of impedance signal with a reasonable accuracy.

## 1. Introduction

The synchronization of the ventricles can be compromised by a condition known as a bundle branch block. The bundle branch block occurs when the conduction of electrical impulse from the right atrium to the ventricles is blocked or slowed down. When this occurs, it may cause the ventricles to become out of synchronization, and when the heart loses its coordination, it cannot pump as well [1]. Cardiac resynchronization therapy (CRT) is a very promising treatment for patients with congestive heart failure. However, patients' classification for CRT as well as evaluation of the therapy effectiveness should be improved.

A magnitude of resynchronization can be achieved by means of measurement of the electromechanical delay. It is defined as the difference between the longest and the shortest activation time in the six basal segments of the left ventricle. Mechanical dyssynchrony can be assessed with the use of imaging modalities, e.g. MRI or ultrasound echo-Doppler [2]. In general, myocardial velocity timing can be evaluated by ultrasound tissue Doppler imaging and myocardial strain timing can be evaluated by MRI tagging [3].

An Impedance Cardiography (ICG) is used for evaluation of heart mechanical activity [4]. However, it reflects conductivity changes which occur in all regions of the thorax. The contribution of particular phenomenon to total impedance change depends on the value of conductivity change, its

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volume and associated sensitivity distribution [5, 6]. Thus, a method of extracting only ventricles', or in general selected, components from the total impedance change is needed.

## 2. Methods

Finite Element Models have been developed to examine a possibility of discriminating between contributions from different parts (ventricles) of the thorax for the assumed electrode configurations.

### 2.1. 2D model

A simplified 2D model was developed to study basic properties of the considered problem (figure 1). It consisted of four tissues: lung, myocardium, blood and effective one standing for tissue of thoracic wall. Each tissue was described by conductivity, respectively $\sigma_l$, $\sigma_m$, $\sigma_b$, and $\sigma_t$. The measured impedance change, due to conductivity changes undergoing in different regions, is described by equations (1-5), where $I_\psi$ – current flowing between current electrodes, hypothetical current $I_\varphi$ flowing between voltage electrodes (four-electrode method), $\Delta\sigma$ – conductivity change [6, 7].

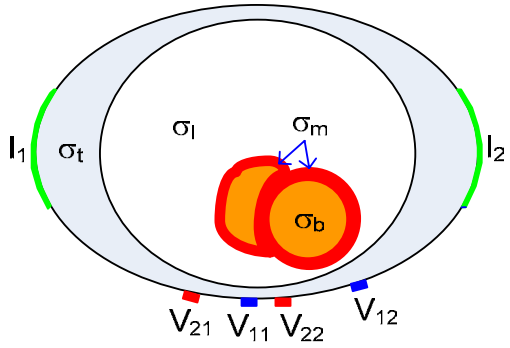


**Figure 1.** 2D model examined in the study, description in the text.

$$\Delta Z = \int_V -\Delta\sigma_s \frac{\nabla\varphi(\sigma+\Delta\sigma)}{I_\phi}\cdot\frac{\nabla\psi(\sigma)}{I_\psi}dv \qquad (1)$$

$$\Delta Z = \sum_i^I K_i\Delta\sigma_i \qquad (2)$$

$$K_i = \int_{V_i} -\frac{\nabla\varphi(\sigma+\Delta\sigma)}{I_\phi}\cdot\frac{\nabla\psi(\sigma)}{I_\psi}dv \qquad (3)$$

$$\Delta Z_1 = \int_{v1} K_{11}\Delta\sigma dv + \int_{v1} K_{12}\Delta\sigma dv \qquad (4)$$

$$\Delta Z_2 = \int_{v1} K_{21}\Delta\sigma dv + \int_{v1} K_{22}\Delta\sigma dv \qquad (5)$$

According to the relationship (1) the impedance change for two different electrode's configurations and for object consisting of two regions can be described by the following relationships (4) and (5). Thus, the impedance change measured by a certain electrode probe was a weighted sum of conductivity changes occurring in selected regions. Without loss of generality it could be assumed that the changes of conductivity were sinusoidal. Thus, assuming that both impedance signals are sinusoidal, however, of different amplitude and phase shift, the resulting signal is also sinusoidal i.e. $A_1\sin(\omega t)+A_2\sin(\omega t+\Delta\varphi_{12})=A\sin(\omega t+\varphi)$. The resulting amplitude and phase can be easily calculated with the use of simple trigonometric relations. Additionally, assuming that relation between amplitudes is a simple ratio, i.e. $A_2 = kA_1$ the resulting amplitude and phase is calculated.

### 2.2. 3D model

A realistic model of the thorax, based on CT data obtained from examination of 68 years old man was developed [8, 9]. A FEM model included of lungs, heart, thorax wall, liver, skeleton and other anatomical organs was developed to mimic impedance measurements. It consisted of 212 000 tetrahedral elements (figure 2). Three different matrix configurations were examined (figure 3).

The associated sensitivity distributions were calculated for different phases of heart contraction and blood distribution inside the thorax. These sensitivity distributions were compared to that obtained for a uniform model of the thorax described by average conductivity. The value of average conductivity was calculated on the assumption equal impedance for uniform and non-uniform models. It was assumed that in both cases the measured impedance is identical. Similar to equation (4) relations were calculated, however, using a FEM approximation. It allowed estimating sensitivity to conductivity changes for regions occupied by organs included in the model [5, 6, 10]. Three different models were

developed: 1. non-uniform with conductivity of heart muscle assumed to be equal to conductivity of blood, 2. non-uniform with conductivity of heart muscle different from that of blood, and 3. uniform.

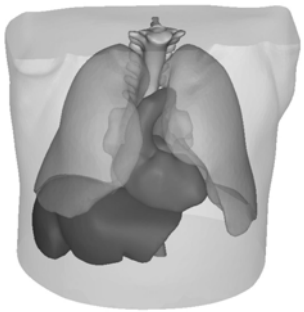

**Figure 2.** A developed geometric model of the thorax.

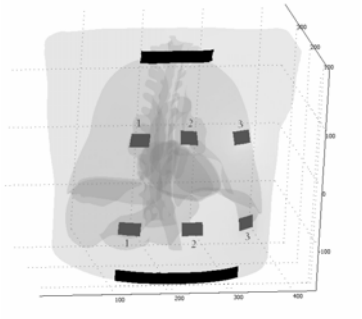

**Figure 3.** Examined electrode configurations in FEM model.

Conductivity values of different organs are gathered in table 1. Aside conductivity a relative value of electric permittivity used in study is also given [11].

**Table 1.** Values of conductivity and relative permittivity of tissues considered in the study.

| Organ/tissue | σ[S/m] | | | $\varepsilon_r$ |
|---|---|---|---|---|
| | $\sigma_1$ | $\sigma_2$ | $\sigma_{average}$ | |
| **Lungs** | 0.1012 | 0.1012 | 0.16 | 5090.8 |
| **Aorta** | 0.7 | 0.7 | 0.16 | 5211.3 |
| **Left Atrium** | 0.7005 | 0.7005 | 0.16 | 5211.3 |
| **Right Atrium** | 0.7005 | 0.7005 | 0.16 | 5211.3 |
| **Left Ventricle** | 0.7005 | 0.56 | 0.16 | 5211.3 |
| **Right Ventricle** | 0.7005 | 0.56 | 0.16 | 5211.3 |
| **Liver** | 0.0689 | 0.0689 | 0.16 | 12100 |
| **Bone** | 0.0206 | 0.0206 | 0.16 | 281.06 |
| **Thoracic wall** | 0.15 | 0.15 | 0.16 | 5000 |

## 3. Results

Estimated values of amplitude and phase of the measured impedance change due to conductivity changes undergoing in regions of ventricles are presented in figure 4. These results have been obtained with the use of a simple 2D model.

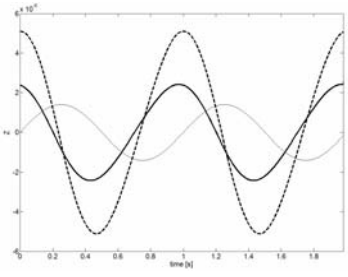

**Figure 4.** Impedance changes calculated for electrodes $V_{11}$-$V_{12}$ (black line), $V_{21}$-$V_{22}$ (dashed line) and conductivity change in region of left ventricle (grey line).

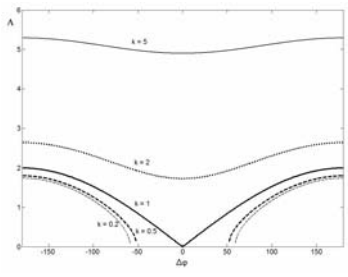

**Figure 5.** Amplitude, A, of the resulting impedance change due to phase shift between conductivity changes localized in left and right ventricle.

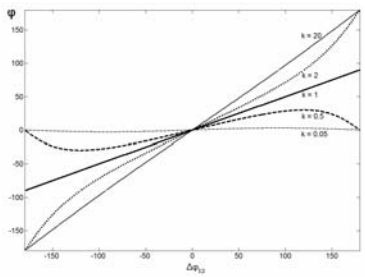

**Figure 6.** Phase of the resulting impedance change due to phase shift between conductivity changes localized in left and right ventricle.

Conductivity changes in the left and right ventricles had the same frequency, however, they were shifted in phase. Extension of that result, for different phase shift and amplitudes of conductivity changes are presented in figure 5 and figure 6. Sensitivity values of the chosen electrode configuration to conductivity changes located in different organs are gathered in table 2.

**Table 2.** Sensitivity values for different conductivity distribution, 2nd pair of voltage electrodes.

| Organ/tissue | Sensitivity [%] | | |
|---|---|---|---|
| | $\sigma_1$ | $\sigma_2$ | $\sigma_{average}$ |
| **Left Lung** | 1.601 | 1.64 | 7.82 |
| **Right Lung** | 0.91 | 0.94 | 6.17 |
| **Aorta** | 3.224 | 3.229 | 1.943 |
| **Left Atrium** | 0.059 | 0.064 | 0.291 |
| **Right Atrium** | 0.251 | 0.268 | 0.917 |
| **Left Ventricle** | 2.655 | 2.300 | 2.724 |
| **Right Ventricle** | 5.953 | 4.892 | 2.724 |
| **Liver** | 0.765 | 0.769 | 9.396 |
| **Bone** | 0.007 | 0.007 | 0.662 |
| **Thorax** | 84.566 | 85.889 | 68.697 |

## 4. Discussion and conclusions

As expected, in case of one dominant component the resulting signal of impedance change almost follows it. Nevertheless, an extraction of the component, which arises from a certain source, is not trivial when amplitudes of all components are comparable. However, the relation between contributions from different organs (regions) strongly depends on conductivity distribution inside the thorax. Multiple simultaneous measurements which use optimally designed electrode configurations allow, in some cases, enhancement of the signal's component from the region of interest.

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