High spatial and temporal resolution 4D FEM simulation of the thoracic bioimpedance using MRI scans

Mark Ulbrich\textsuperscript{1}, Bastian Marleaux\textsuperscript{1}, Jens M"uhlsteff\textsuperscript{2}, Felix Schoth\textsuperscript{3}, Ralf Koos\textsuperscript{3}, Daniel Teichmann\textsuperscript{1}, Steffen Leonhardt\textsuperscript{1}

\textsuperscript{1}RWTH Aachen University, Philips Chair for Medical Information Technology, Pauwelsstrasse 20, 52074 Aachen, Germany
\textsuperscript{2}Philips Research Eindhoven, High Tech Campus 34, 5656 AE Eindhoven, The Netherlands
\textsuperscript{3}University Hospital Aachen, Pauwelsstrasse 30, 52074 Aachen, Germany

E-mail: ulbrich@hia.rwth-aachen.de

Abstract. In this work, a finite element model was created using MRI scans of the main author to analyze sources of the dynamic thoracic bioimpedance. This model can be used to identify limitations of impedance cardiography (ICG) in practice. Heart beat (8.3 ms temporal resolution) and aortic wave propagation (2.6 ms temporal resolution) were implemented. The static volume contains all major organs of the thorax in high spatial resolution. Simulations were successfully conducted and a high correlation (r = 0.9) between the simulated aortic ICG signal and a measured signal of the same subject was obtained.

1. Introduction
ICG is a simple, cheap and non-invasive method to acquire hemodynamic parameters. Due to several limitations of this technology, it is not used in clinical practice today. The finite element model developed within this work shall provide remedy by analyzing multiple sources of the dynamic thoracic impedance and their influence on the signal. That way, the impedance signal of patients with certain pathologies can be simulated to identify problems occurring during measurements.

Others used MRI scans or volumetric representations of the thorax to investigate the thoracic impedance with a limited temporal resolution and no volumetric changes \cite{1, 2, 3, 4}. This work combines high spatial and temporal resolution to create a simulation model which is a novelty. Another advantage of this work is that the basis for the simulation model and the subject measured for the reference signal are the same person.

2. Methods
To build a geometric representation of the thorax of the main author, three different MRI scans were made: a static scan, a dynamic whole-heart scan and an angiographic scan. During the latter, a contrast agent was injected to visualize major blood vessels. The MRI scans were performed by using a Philips Gyroscan NT device with a magnetic field strength of 1.5 T. All three scans were used to build a dynamic representation of the thorax.
Major organs (12 different tissues) were identified from the static and angiographic scans using manual and semi-automatic segmentation. The volumetric geometry changes of both ventricles were extracted manually from the whole-heart scans, since they were ECG gated, resulting in 10 volumes per heartbeat. Unfortunately, diagnostic whole-heart scans do not contain information about atrial changes. Hence, static volumes were used to represent atria during heartbeat.

For the aortic wave propagation, the end-diastolic volume of the aorta obtained by the angiographic scan was semi-automatically segmented by Region Competition Snake Segmentation (RCSS) using the open source tool itk-SNAP [5] and subdivided into 13 parts. The volume of these parts was altered according to the measured pressure pulse propagating over the aorta. Here, 30 volumes per heartbeat were implemented.

Since the angiographic scans were not ECG gated, the dynamics of the aorta were implemented based on data derived from a project in which the arterial system was rebuilt by using silicone [6]. Within this work, pressure and flow data were acquired at various points on the aorta. Thus, the temporal development of pressure and flow for every point on the aorta was calculated and interpolated using MATLAB® (MathWorks, Inc. MA, USA). Using these results and the end-diastolic volume of the aorta, the radial change of the aortic segments was computed (figure 1).

**Figure 1.** Temporal development of the aortic radius along the aorta

The radius change in this figure was temporally normalized to facilitate the comparison of the curves, but the temporal delay of the propagating pulse was included into the model. All volumes were converted in stereolithography (STL) format and their surfaces were smoothed before being imported into the simulation software. Conductivity and permittivity values for 100 kHz obtained by Gabriel [7] were assigned to all tissues.

Standard electrode positions at neck and abdomen were used to apply a fixed voltage. The resulting complex current flowing through the thorax was calculated by integrating conduction and displacement current density over an area:

\[
L_{\text{cond}} = \int_A (\sigma \cdot \vec{E}) \ d\vec{A} = \int_A \vec{J} \ d\vec{A} = \int_A (Re\{\vec{J}\} + j Im\{\vec{J}\}) \ d\vec{A} \quad (1)
\]

\[
L_{\text{disp}} = \int_A (\epsilon \cdot \frac{\partial \vec{D}}{\partial t}) \ d\vec{A} = \int_A (\frac{\partial \vec{D}}{\partial t}) \ d\vec{A} = \int_A (j \omega Re\{\vec{D}\} - \omega Im\{\vec{D}\}) \ d\vec{A} \quad (2)
\]
\[
L_{\text{total}} = \frac{\text{Re}\{L_{\text{cond}}\} + \text{Re}\{L_{\text{disp}}\} + \text{Im}\{L_{\text{cond}}\} + \text{Im}\{L_{\text{disp}}\}}{\text{Re}\{L_{\text{total}}\} - \text{Im}\{L_{\text{total}}\}}
\]  

(3)

That way, the dynamic complex impedance was calculated. For every calculation volume, a new simulation was conducted. An electroquasistatic solver was used since the wavelength of alternating currents with a frequency of 100 kHz is much bigger than the calculation volume. For discretization, a hexahedral mesh was used with a maximum mesh density of 50 parts subdividing the simulation volume. A higher density could not be used due to the limitation of the main memory (48 GB RAM). The CST EM Studio® (Computer Simulation Technology, Darmstadt, Germany) was used to perform the simulations.

3. Results
Figure 2 shows the simulation model with voltage sources and areas for current density integration. Fat and muscle tissue are semi-transparent to show the inner organs. The extracted volumes for the dynamic volume changes in combination with static blood vessels are shown in figure 3.

![Figure 2. Simulation model.](image)

![Figure 3. Blood vessels.](image)

Impedance curves for both dynamic sources were assessed. Because discretization noise for a mesh density of 50 parts was unavoidable, a smoothing spline was calculated for further signal analysis.

Since it is known from previous simulations, that the aortic diameter change defines the characteristic morphology of the ICG signal [8], and since no other dynamic sources besides the mentioned ones are implemented so far in the model, the derivative of the aortic impedance change was compared to a measured ICG signal of the same subject using the Niccomo device (medis, Ilmenau, Germany) showing a high cross correlation factor of \( r = 0.9 \) (see figure 4).

Despite the discretization noise, the signal is good enough to extract characteristic points, such as closure (X-point) and opening (B-point) of the aortic valve, to calculate left ventricular ejection time (LVET). Table 1 shows the extracted times of characteristic points and LVET.

Although simulated and measured ICG curve show a rather large temporal offset, LVET is the same for both curves.
Figure 4. Simulated • and measured - - - -ICG.

Table 1. Extracted times from ICG curve.

|       | B-point | C-point | X-point | O-point | LVET |
|-------|---------|---------|---------|---------|------|
| Simulated | 93 ms   | 213 ms  | 373 ms  | 506 ms  | 280 ms |
| Measured  | 35 ms   | 105 ms  | 315 ms  | 430 ms  | 280 ms |

4. Conclusion
First, a finite element model of a human thorax with high temporal and spatial resolution was created containing important organs and major blood vessels. Besides these static volumes, heartbeat and aortic wave propagation were implemented to simulate the thoracic complex dynamic impedance change. Second, the impedance change for both dynamics was simulated showing a reasonable development. Third, the morphology of the ICG signal shows high correlation with the morphology of measured signals. Hence, a model was created that provides the possibility to identify changes of the ICG signal due to pathologies in the future.

Improvements of the model comprise the integration of dynamic atria by using additional MRI scans and the implementation of other important dynamics, such as erythrocyte orientation and lung perfusion, into the simulations. Since volumes of large venous vessels were extracted, the impact of volumetric changes of the vena cava on the impedance cardiogram can be analyzed.

References
[1] Patterson R 1985 Medical & Biological Engineering & Computing 23 411–417
[2] Sakamoto K, Muto K, Kanai H and Izuka M 1979 Medical & Biological Engineering & Computing 17 697–709
[3] Kim D, Baker L, Pearce J and Kim W K 1988 IEEE Transactions on Biomedical Engineering 35 993–1000
[4] Wang L and Patterson R 1995 IEEE Transactions on Biomedical Engineering 42 141–148
[5] Yushkevich P A, Piven J, Cody Hazlett H, Gimpel Smith R, Ho S, Gee J C and Gerig G 2006 Neuroimage 31 1116–1128
[6] Matthys K S, Alastruey J, Peir J, Khir A W, Segers P, Verdonck P R, Parker K H and Sherwin S J 2007 Journal of Biomechanics 40 3476–3486
[7] Gabriel C, Gabriel S and Corthout E 1996 Physics in Medicine and Biology 41 2231–2249
[8] Ulbrich M, Mühlsteff J, Walter M and Leonhardt S 2011 Computing in Cardiology vol 38 pp 149–152