Modeling of Fluid Shifts in the Human Thorax for Electrical Impedance Tomography Using the Finite Element Method

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***Abstract*— Thoracic electrical impedance tomography (EIT) is a non-invasive technique for the estimation of the spatial conductivity distribution change in a thorax cross section. To reconstruct these changes from potential measurements at the boundary, the inverse problem has to be solved. The resulting images have a rather low spatial resolution. However, EIT seems to be particularly useful for the monitoring of patients suffering from Acute Respiratory Distress Syndrome (ARDS), since it is capable of detecting dynamic shifts of body fluids and lung aeration by the bedside. A Finite Element simulation of the human thorax is presented, and the results of the reconstruction from simulated values are being used to derive new indices to quantify the ventilation shifts.**

I. Introduction

Thoracic electrical impedance tomography (EIT) is a non-invasive technique for the estimation of the spatial conductivity distribution in a thorax cross section. N_E electrodes (here: $N_E = 16$) are equidistantly attached to the patient's skin around the thorax and small alternating currents injected (e.g. 5 mA_{eff}@50 kHz). The potentials at the remaining adjacent electrodes are then measured and used to reconstruct changes of the conductivity distribution [1].

The spatial resolution is rather low (about the distance of neighboring electrodes) if compared to other imaging modalities such as computed tomography (CT). However, EIT seems to particularly useful for the bedside monitoring of patients suffering from Acute Respiratory Distress Syndrome (ARDS) due to its non-invasiveness and high temporal resolution.

ARDS is a life-threatening state of the lung characterized by pulmonary inflammations, which lead to alveolar lung edema (water in the lung) and atelectasis (lung collapse) and, thus, changes of conductivity. ARDS patients have to be mechanically ventilated with higher oxygen concentrations and changing pressure levels resulting in dynamic shifts of body fluids and lung aeration.

EIT is capable of visualizing those shifts providing an instantaneous feedback to the medical staff, even though it still has to be examined what information exactly should be extracted and how it should be processed and quantified in order to achieve maximum benefit.

Therefore, we created a finite element model of the human thorax. Then, we divided the lung into four coronal layers and changed the fluid contents of the gravitationally dependent layers in order to simulate lung edema of differing severity. After reconstructing differential EIT images, we examined methods to quantify the detected changes by calculating the center of ventilation. Additionally, we determined the ratio of ventilation between both lobes of the lung which is influenced by the non-central position of the heart.

II. Methods

A. Modeling

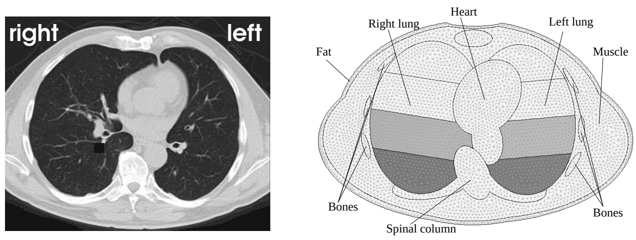

Fig. 1. Human thorax (left) and corresponding 2D-model (right). The two bottommost layers of the lung were filled with body fluid to simulate lung edema.

The geometric dimensioning of the human thorax model was based on the "Visual Human Project" [5], see Figure 1. In total, six different types of tissue were being modeled: muscle, heart, bones, fat, lung and body fluid, cf. Table I. The 2D-mesh consisted of 8,800 first order triangular elements and 4,600 nodes and was implemented in ANSYS®.

In order to simulate patient breathing, conductivity and permittivity of the lung was changed from deflated to inflated tissue. However, the size of the lung regions was kept unchanged which seems to be an acceptable simplification as the lung is mainly stretched along the longitudinal axis during breathing.

TABLE I

Electrical properties of the modeled tissues [6]

Tissue	Conductivity (S/m)	Rel. permittivity
Body Fluid	1.5	98.56
Bone (cortical)	0.02064	264.19
Fat	0.02424	172.42
Heart	0.19543	16982.00
Lung (deflated)	0.26197	8531.40
Lung (inflated)	0.10265	4272.50
Muscle	0.35182	10094.00

As mechanically ventilated patients are mostly treated in supine position and lung water mainly concentrates in the gravitationally dependent regions, we divided the lung into four coronal layers. Then, we modeled different amounts of lung water by first replacing the bottommost layer of the lung tissue with body fluid and, finally, the two bottommost layers (see Fig. 1). The remaining lung regions were still changed from deflated to inflated tissue to simulate breathing.

B. Field Simulation

The field simulation is performed using the Finite Element Method (FEM). The unknown field variable is the electric potential V. The used formulation is given in Galerkin form:

$$\int_{\Omega} (\sigma + j\omega\varepsilon)\, \nabla\cdot \alpha_i\, \nabla\cdot \underline{V} \mathrm{d}\Omega = \int_{\Gamma} \alpha_i \underline{J}_n \mathrm{d}\Gamma \quad \forall i = 1, \ldots, n_n \tag{1}$$

Here, σ and ε are the conductivity and permittivity of the materials, ω the angular frequency and J_n the normal current flowing into the model, which is introduced using a Neumann boundary condition.

C. Image Reconstruction and Processing

Differential images $F(x, y)$ $(x, y = 1..32)$ were reconstructed using a modified Sheffield back-projection algorithm [7]. The reconstructed images describe lung ventilation based on the changing lung conductivities due to breathing.
In order to quantify shifts of the ventilation distribution resulting from different amounts of water in the lung, we calculated a center of ventilation index y_s and an ventilation ratio between the left and the right lobe of the lung:

$$y_s = \frac{\sum\limits_x \sum\limits_y y \cdot F(x,y)}{\sum\limits_x \sum\limits_y F(x,y)} \qquad LR_{ratio} = \frac{\sum\limits_{x=17}^{32} \sum\limits_y F(x,y)}{\sum\limits_{x=1}^{16} \sum_y F(x,y)} \tag{2}$$

Finally, the differential images were bilinearly interpolated to 156×156 pixels for a smoother representation.

III. RESULTS

The reconstructed images for different levels of lung edema can be seen in Figure 2. If the gravitationally dependent part

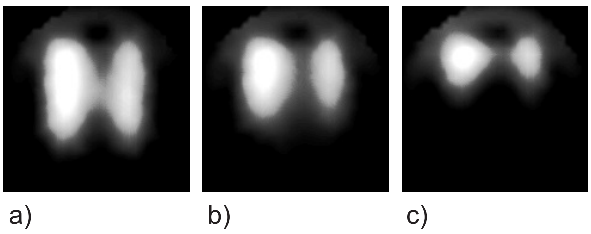

Fig. 2. Reconstructed differential images. White areas describe a great change in conductivity, i.e. a well-ventilated part of the lung. a) Ventilation of the whole lung. b) The bottommost layer is not ventilated due to lung edema. c) The two bottommost layers are filled with water.

of the lung is filled with lung water, the center of ventilation is shifted upwards which can clearly be seen in the images. The left lobe of the lung (i.e. the right lobe in the image) is less ventilated than the right one due to the asymmetrical position of the heart, even though the conductivity changes were identical.
The center of ventilation index and the ventilation ratio is listed in Table II.

TABLE II
CENTER OF VENTILATION INDEX y_s AND VENTILATION RATIO LR_{ratio} FOR DIFFERING SEVERITY OF LUNG EDEMA.

# of water-filled layers	**Center of Ventilation**	**Ventilation Ratio**	**Lung volume (relative)**
0 (no water)	14.32	0.79	1
1	12.45	0.75	0.73
2	9.00	0.71	0.42

The current density images of the FEM simulation can be seen in Figure 3.

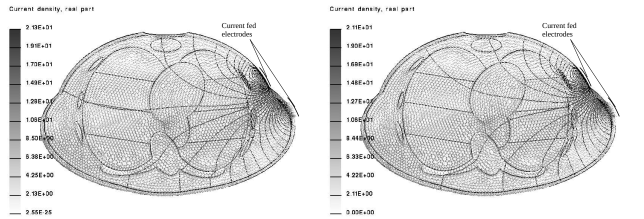

Fig. 3. Current density distribution and potential isolines from the FEM simulation for the deflated lung (left) and the deflated lung with the bottom half filled with body fluid (right).

IV. DISCUSSION AND CONCLUSION

An EIT application was simulated using the FEM. The resulting potentials were fed into a reconstruction algorithm. The reconstructed images reflect the simulated changes in conductivity distribution quite well, although the spatial resolution is low. The changes of the ventilation can be quantified by means of a ventilation index. Some effects were not regarded (like the 3rd dimension) or simplified (like the constant lung size). The influence of these effects has to be studied in future work. The EIT proved to be a method well suited for detecting changes in the ventilation distribution of the lung.

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