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Reconstruction of conductivity change in lung lobes utilizing electrical impedance tomography

Abstract: Electrical Impedance Tomography (EIT) is a novel medical imaging technology which is expected to give valuable information for the treatment of mechanically ventilated patients as well as for patients with obstructive lung diseases. In lung-EIT electrodes are attached around the thorax to inject small alternating currents and to measure resulting voltages. These voltages depend on the internal conductivity distribution and thus on the amount of air in the lungs. Based on the measured voltages, image reconstruction algorithms are employed to generate tomographic images reflecting the regional ventilation of the lungs. However, the ill-posedness of the reconstruction problem leads to reconstructed images that are severely blurred compared to morphological imaging technologies, such as X-ray computed tomography or Magnetic Resonance Imaging. Thus, a correct identification of the particular ventilation in anatomically assignable units, e.g. lung-lobes, is often hindered. In this study a 3D-FEM model of a human thorax has been used to simulate electrode voltages at different lung conditions. Two electrode planes with 16 electrodes at each layer have been used and different amount of emphysema and mucus plugging was simulated with different severity in the lung lobes. Patient specific morphological information about the lung lobes is used in the image reconstruction process. It is shown that this kind of prior information leads to better reconstructions of the conductivity change in particular lung lobes than in classical image reconstruction approaches, where the anatomy of the patients' lungs is not considered. Thus, the described approach has the potential to open new and promising applications for EIT. It might be used for diagnosis and disease monitoring for patients with obstructive lung diseases but also in other applications, e.g. during the placement of endobronchial valves in patients with severe emphysema.

Keywords: Electrical Impedance Tomography, obstructive lung diseases, medical imaging, simulation study.

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1 Introduction

Electrical Impedance Tomography (EIT) is an imaging technology which is mainly used to assess the regional ventilation distribution in the lungs. Compared to morphological imaging methods e.g. X-ray computed tomography, EIT images are less detailed and appear blurred. However, the advantages of EIT are its easy applicability, the lack of potentially harmful radiation, real-time image reconstruction and the high temporal resolution [1]. The benefit of real-time reconstruction has been used in several studies in the intensive care unit (ICU), where EIT was used to adapt ventilator settings in mechanically ventilated patients. Thereby clinicians can directly see how changes in ventilator settings, such as tidal volume or positive end-expiratory pressure (PEEP), affect the ventilation distribution and if parts of the lungs are collapsed or over-distended [2].

Recent studies also take advantage of the high temporal resolution of EIT. Thereby EIT has the potential to act as an additional technique during pulmonary function testing [3]. Established methods for pulmonary function testing (e.g. spirometry) are only able to provide global parameters to reflect the current lung condition. The high data acquisition rate enables its application during the determination of highly dynamic lung parameters, such as the 'forced expiratory volume in 1 second' (FEV_1). Consequently EIT might increase the information content of these global lung parameters by providing regional features of such established measures [4]. Thus, besides its use in the ICU, the most promising application of EIT is the usage as a supplemental tool during pulmonary function testing, especially in patients

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with obstructive lung diseases, such as chronic obstructive lung disease, asthma or cystic fibrosis.

However, a drawback of currently available and clinically approved EIT-systems is the limitation to 2D-image acquisition and reconstruction. In these systems a single electrode plane, consisting of 16-32 electrodes is attached around the thorax to inject small alternating currents and to measure changes in voltages due to the breathing of the patient. So far a single electrode plane has been considered as adequate enough to monitor the ventilation distribution in mechanically ventilated patients in supine or prone position, since the ventilation gradient is usually distributed along the gravitational axis. However, for sitting patients with obstructive lung diseases this assumption does no longer hold. Severe differences in ventilation distribution between cranial and caudal lung regions are likely, which has already been shown by sequential measurement at different thoracic layers [4].

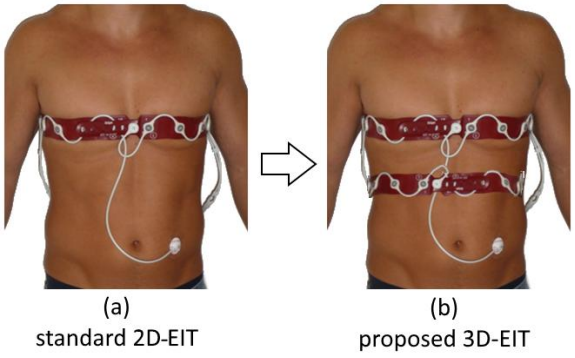

Figure 1: (a) standard 2D-EIT with electrodes attached at 5th intercostal space. (b) proposed 3D-EIT with two electrode belts attached at the thorax.

A possible procedure to overcome this hurdle is to use three-dimensional EIT (3D-EIT) [5]. Thereby, several electrode planes are used and current injection as well as voltage measurements are conducted within the planes but also across the planes (see **Figure 1**).

In this study it is shown how patient specific anatomical information about the patients' lung lobes can improve image reconstruction for 3D-EIT.

2 Methods

2.1 Model generation

For the simulation of electrode voltages several 3D-FEM models (comprising 153795 elements) were generated to reflect different lung conditions. The shape of the FEM models was generated based on a CT-dataset. Additionally, the lung lobes were determined in the CT-data and corresponding FEM elements were assigned to the respective lobe. **Figure 2 (a)** shows the FEM model used for voltage simulation, whereas different colours were assigned to the lung lobes.

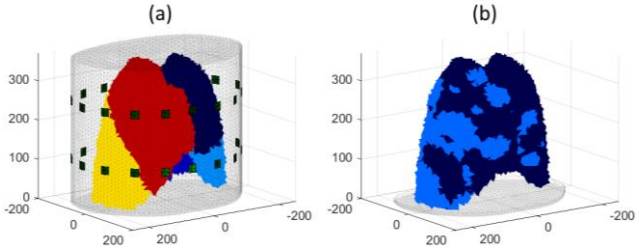

Figure 2: (a) FEM model used for simulation of electrode voltages. Different lung lobes are represented by different colours. (b) simulation of different severities of obstruction in the lobes. Dark blue represents healthy lung tissue with conductivity set to $\sigma_{healthy} = 60\ mS/m$, light blue represents obstructed regions with $\sigma_{obst} = 120\ mS/m$.

In all models FEM elements, which are not assigned to a lung lobe, are set to a conductivity of $\sigma_{bkg} = 480\ mS/m$. The FEM model representing the reference conductivity was modelled with all 'lung FEM-elements' set to a conductivity of $\sigma_{ref} = 120\ mS/m$. The voltages simulated with this model are contained in the vector $\mathbf{v}_{ref} \in \mathbb{R}^{928\times1}$ and represent the lung conditions at end-expiration.

Similarly the voltages $\mathbf{v} \in \mathbb{R}^{928\times1}$ were simulated with heterogeneously distributed values for the lobe conductivity. These voltages $\mathbf{v}$ represent the conditions after end-inspiration. The 'lung FEM elements' were assigned either to a conductivity of $\sigma_{healthy} = 60\ mS/m$, which represents healthy lung tissue, or to a conductivity of $\sigma_{obst} = 120\ mS/m$, which represents obstruction or air-trapping. The amount of 'simulated obstruction' was varied by distributing an arbitrary amount of spheres with different sizes over the lung. This results in different mean conductivity changes for each of the lobes. The mean ground truth conductivity change in the lobes is contained in $\mathbf{x}_{lobeTruth} \in \mathbb{R}^{5\times1}$. **Figure 2 (b)** shows an exemplary condition with arbitrarily sprinkled regions of reduced ventilation (e.g. due to mucus plugging or air trapping). Overall 100 arbitrary patterns of obstructions were generated and from each of them the voltages $\mathbf{v}$ were simulated.

2.2 Standard image reconstruction

In this study normalized difference EIT is applied, which uses the normalized difference voltages $\mathbf{z}$, where the i-th element is defined as $z_i = (v_i - v_{ref,i})/v_{ref,i}$, with v_i and $v_{ref,i}$ being the i-th components of the vectors $\mathbf{v}$ and $\mathbf{v}_{ref}$, respectively. For image reconstruction, a 3D-FEM model (comprising 78290 elements) with the same outer contour as the model for voltage simulation was utilized. The relative conductivity change $\hat{\mathbf{x}} \in \mathbb{R}^{78290\times1}$ is reconstructed employing a reconstruction matrix $\mathbf{B}$, such that

$$\hat{\mathbf{x}} = \mathbf{B}\mathbf{z} \tag{1}$$

where

$$\mathbf{B} = (\mathbf{J}^{\mathbf{T}}\mathbf{J} + \lambda^2\mathbf{I})^{-1}\mathbf{J}^{\mathbf{T}} \tag{2}$$

The matrix $\mathbf{J} \in \mathbb{R}^{928\times78290}$ is the Jacobian, which relates changes in conductivity to voltages changes and matrix $\mathbf{I} \in \mathbb{R}^{78290\times78290}$ represents the identity matrix to regularize the solution $\hat{\mathbf{x}}$. The amount of regularization is controlled with the hyperparameter λ, which was defined heuristically in this study. A more detailed description of this standard approach for image reconstruction can be found e.g. in [6]

Subsequently the mean conductivity change in the particular lung lobes $\hat{\mathbf{x}}_{lobeStd} \in \mathbb{R}^{5\times1}$ is calculated from the reconstructed distribution $\hat{\mathbf{x}}$.

2.3 Lobe image reconstruction

The lobe based image reconstruction utilizes a modified Jacobian $\mathbf{J}_{lobe} \in \mathbb{R}^{928\times5}$, which is generated by

$$\mathbf{J}_{lobe} = \mathbf{J}\mathbf{L} \tag{3}$$

with $\mathbf{L} \in \mathbb{R}^{78290\times5}$. Each element $L_{i,j}$=1, if FEM element i belongs to lobe j and $L_{i,j} = 0$ otherwise.

With this dimension reduction, the conductivity change in each lobe can be reconstructed according to

$$\hat{\mathbf{x}}_{lobe} = (\mathbf{J}_{lobe}{}^{\mathbf{T}}\mathbf{J}_{lobe})^{-1}\mathbf{J}_{lobe}^{\mathbf{T}}\mathbf{z} \tag{4}$$

where $\hat{\mathbf{x}}_{lobe} \in \mathbb{R}^{5\times1}$.

3 Results

Exemplary reconstructions of conductivity change in the lobes are shown in **Figure 3**. The right upper lobe (RU) has the lowest change in conductivity, whereas all others lobes have equally values of conductivity change. **Figure 3 (a)** shows normalized values of ground truth mean conductivity change in the lobes $\mathbf{x}_{lobeTruth}$ (red) and the reconstructed mean conductivity change $\hat{\mathbf{x}}_{lobeStd}$ in the lobes (blue), which were reconstructed using a standard 3D reconstruction method, according to eq. (1) and eq. (2). The $recError$ visualized the difference between 'ground truth' and reconstructed values. Similarly, in **Figure 3 (b)** the reconstructed conductivity change $\hat{\mathbf{x}}_{lobe}$ (green) is depicted, which was calculated by aggregating respective columns of the Jacobian $\mathbf{J}$, as described in eq. (3).

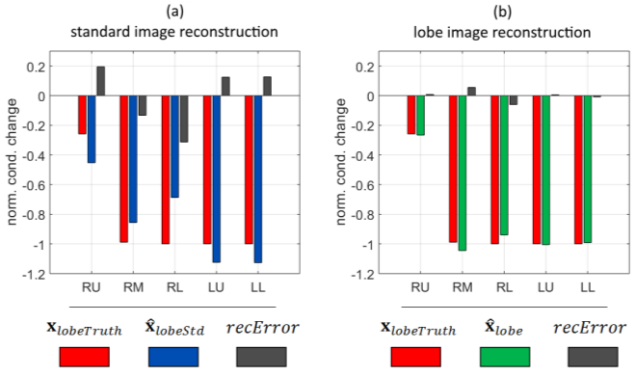

Figure 3: Comparison of different methods to reconstruct conductivity change in lung lobes (RU: right upper, RM: right middle, RL: right lower, LU: left upper, LL: left lower). (a) normalized lobe conductivity change (red: ground truth $\mathbf{x}_{lobeTruth}$, blue: standard 3D-reconstruction $\hat{\mathbf{x}}_{lobeStd}$, gray: reconstruction error). (b) normalized lobe conductivity change (red: ground truth $\mathbf{x}_{lobeTruth}$, green: lobe image reconstruction $\hat{\mathbf{x}}_{lobe}$, gray: reconstruction error).

The example in **Figure 3** already indicates that the reconstruction error is higher if the standard approach for reconstruction is used. **Figure 4** shows a box plot for comparison of reconstruction errors of both methods. The Anderson-Darling test ($p<0.001$) was used for normality testing, and the Wilcoxon signed rank test ($p<0.001$) was used to evaluate significant statistical differences between the median values of the reconstruction errors.

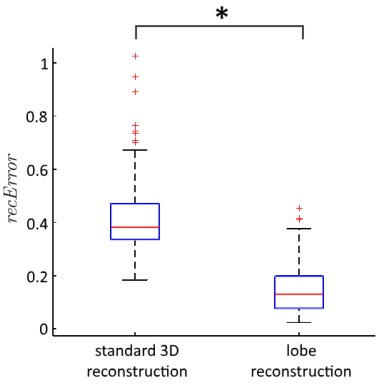

Figure 4: Reconstruction error for the standard reconstruction method and lobe reconstruction. The box indicates the 25% percentile, the median (red line) and the 75% percentile. The whiskers represent the minimum and maximum values. Outliers are plotted as plus signs. Median values of reconstruction error between both methods show significant differences ($p<0.001$).

4 Discussion

In this paper we have shown in a simulation study that 3D-EIT might be used to assess the conductivity change in particular lung lobes. More particularly, it is demonstrated that the reconstruction error of mean conductivity change in the lobes is minimized, if anatomical information about the lobes is used to aggregate respective columns of the Jacobian.

If patient specific morphological information is available (e.g. CT-data) such a methodology might be useful to monitor disease progression in patients with obstructive lung diseases. Another possible application could be during the placement of endobronchial valves [7]. Thereby one-way valves are places in the bronchus of a lung lobe with severer emphysema. The valve allows the air to slowly exit from the lobe, while during inspiration no more air is flowing into the treated lobe. In such an application EIT might be used to directly asses the correct functioning of the valves during surgery, but as well in follow-up examinations.

To fully reveal the potential of this approach additional studies on patients are necessary. However, this study should encourage researchers and manufacturers of EIT-devices to consider 3D-EIT as a valuable technology to assess the lung conditions.

Author's Statement

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