Frequency-Differential Reconstruction Algorithm for Thoracic EIT Using Absolute Values

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Abstract: Frequency-differential EIT reconstructs complex regional conductivity differences. We propose to extend the traditional Gauss-Newton approach with a weighted voltage minimization similar to absolute-EIT. The reconstruction results show a smoother background area and an improved contrast between organs and background.

1 Introduction

Most EIT-systems monitor time-differential conductivity distributions. Measuring at multiple frequencies simultaneously gives information about passive electrical properties and the dispersion of biological tissue. In the long term, the underlying tissue should be identified [1]. One application and the dispersion of biological tissue. In the long term, the underlying tissue should be identified [1]. One application of frequency-differential Electrical Impedance Tomography (fdEIT) is to monitor lung edema, due to the reduced extracellular resistance [2].

2 Methods

2.1 The Inverse Problem of EIT

In a three-dimensional domain $\Omega$ with a given conductivity $\gamma^*$ the resulting voltages $\bar{U}$ of an injected current can be easily calculated by $A : \gamma^* \mapsto \bar{U}$, where $A$ is the forward map of $\Omega$. EIT tries to reconstruct $\gamma^*$ from the measured surface potentials. The inverse of $A$ cannot directly be calculated, as the inverse problem is ill-posed. Therefore, a least-squares minimization $\gamma_{opt} = \min \left\{ \Psi \right\}$ is often used to estimate $\gamma^*$, where $\Psi$ is an objective function of the minimization. $\gamma_{opt}$ is optimized in a way that the difference of $\bar{U}$ and the calculated forward solution is minimized. Different approaches exist to solve this problem, e.g. GREIT and Gauss-Newton algorithm [3, 4].

2.2 Frequency-Differential Reconstruction

The objective function of the traditional Gauss-Newton approach is adjusted to meet the requirements for fdEIT:

$$\Psi_{absGN}(\gamma_{tot}^k) = \frac{1}{2} \left( \left\| \Delta A(\gamma_{tot}^k) \right\| - \left\| \Delta \bar{U} \right\| \right)^2 + \frac{\lambda}{2} \left\| L^tot, \gamma_{tot}^k \right\|^2 + \frac{\beta}{2} \left( \left\| A(\gamma_{high}^k) - \bar{U}_{high} \right\|^2 + \left\| A(\gamma_{low}^k) - \bar{U}_{low} \right\|^2 \right).$$

The differential-term includes the voltage difference at two frequencies $\Delta \bar{U} = \bar{U}^h - \alpha \bar{U}^l$, where $\Delta A(\gamma_{tot}^k) = A(\gamma_{tot}^k) - \alpha A(\gamma^k)$ is the difference of the forward solution of the current conductivity solution. In addition, $\alpha$ describes a weighted difference coefficient as described by Jun [4] to cancel out common errors. The regularization-term performs a Tikhonov-Regularization with a NOSER prior [5]. The hyperparameter $\lambda$ is chosen with the L-curve criterion [6]. We introduced an absolute-term, which minimizes the voltage difference at each single frequency similar to absolute-EIT. The control coefficient $\beta \in [0, 1]$ was introduced to define a ratio between the absolute- and differential-minimization.

3 Results and Discussion

The reference model in fig. 1 mimics an edema in the left lung. Typical tissue-conductivities were derived from the Gabriel-database [7].

The model used for the reconstruction has a different shape as the reference model to avoid over-fitting. The background area of the image is smooth and the contrast between organs and background is improved. Especially the phase contour of the edema has a similar shape as the reference lung and has similar values.

4 Conclusions and Outlook

We developed an improved version of a weighted difference algorithm for fdEIT. The newly introduced absolute-minimization term leads to a smoother background and an improved contrast between organs. In the future, the algorithm will be evaluated on different models in simulations as well as in real fdEIT measurements.

References