A non-contact method for imaging the posterior chest using magnetic induction principles that allows to monitor pulmonary oedema

D Gürsoy and H Scharfetter
Institute of Medical Engineering, Graz University of Technology, Kronesgasse 5, A-8010 Graz, Austria
E-mail: guersoy@tugraz.at

Abstract.
Real time monitoring of lung function is of particular importance for the patients who are in the intensive care unit, and thus spend long durations of time in a supine position. This kind of recumbent positioning of the patients gives rise to a markedly increased fluid accumulation in the posterior lung regions associated with the gravity dependency. In order to monitor the temporal behavior of the accumulation, we proposed a non-contact semi-tomography method which uses magnetic induction principles. In the proposed method, an eddy current density is induced within the dorsal tissues including the posterior lungs via the transmitter coils which are embedded into the patient bed, and the magnetic field strength is measured similarly using an array of sensor coils in a non-contact manner. For the assessment of the method, we used a patient specific, MRI-guided realistic chest model and presented the reconstructed time-differential images.

1. Introduction
Real time monitoring of lung function is of particular importance for the patients who are in the intensive care unit, and thus spend long durations of time in a supine position. This kind of recumbent positioning of the patients gives rise to a markedly increased fluid accumulation in the posterior lung regions associated with the gravity dependency. In earlier studies, electrical impedance tomography was reported as a diagnostic tool in order to monitor the temporal behavior of the accumulation [1], [2]. However, using electrodes has some formidable drawbacks due to the poor skin-electrode interface. Therefore, we proposed an alternative method which uses magnetic induction principles. The non-contact measurement methodology has been known for years and at first proposed over 40 years ago to monitor the cardiac and respiratory cycles of the patients [3] and is gaining interest again [4] to be used for imaging purposes. In the proposed method, an eddy current density is induced within the dorsal tissues including the posterior lungs via the transmitter coils which are embedded into the patient bed, and the magnetic field strength is measured similarly using an array of sensor coils in a non-contact manner to be used in image reconstruction. For the assessment of the method, we used a patient specific, MRI-guided realistic chest model and presented the reconstructed time-differential images using simulated data.
2. Methods

2.1. Image reconstruction

The relationship between the conductivity parameters and the measurement data is nonlinear and requires the solution of a set of electromagnetic equations. The corresponding formulation is based on the quasi-stationary approximation of Maxwell’s equations and has been reported in earlier studies [5], [6]. Let us define this relation as, $v = \psi(\sigma)$, where $\sigma$ is a piecewise-constant function that defines the conductivity of the medium and $v$ represents the voltage measurements. The image reconstruction is known as estimating the conductivity $\sigma$ from the measurements $v$.

2.1.1. Linearization

Due to the nonlinear nature of the problem, absolute imaging fails to perform satisfactorily, and thus differential imaging is preferred. The corresponding reconstruction problem has commonly been formulated using the reciprocity theorem. According to the theorem, the sensitivities of the measurements to the conductivity perturbations at each voxel can be approximated as follows,

$$\left. \frac{\partial \psi}{\partial \sigma} \right|_{\sigma_0} \approx \Delta \sigma^{-1} \int_{\Omega_c} \Delta \sigma \mathbf{E}_1(\sigma_0) \cdot \mathbf{E}_2(\sigma_0) dV,$$

(1)

where $\mathbf{E}_1$ and $\mathbf{E}_2$ respectively denote the direct and reciprocal electric fields generated in the conducting domain $\Omega_c$ for a particular conductivity distribution $\sigma_0$ and $\Delta \sigma$ represents the conductivity deviation from $\sigma_0$. Assuming admissible conductivity deviations, by using (1), the following system of linear equations can be formed,

$$\mathbf{S}\Delta \sigma = \Delta v,$$

(2)

where $\mathbf{S}$ is referred to as the sensitivity matrix. Each row corresponds to a excitation-measurement coil pair and each column to a voxel.

2.1.2. Regularized Gauss-Newton Method

The regularized least-squares solution for $\Delta \sigma$ can be obtained by minimizing the $L_2$-norm of the residuals between the estimated and measured data as defined by means of a minimization problem and a single iteration of the Gauss-Newton algorithm gives a fairly convenient solution for $\Delta \sigma$ [7]. The corresponding solution can be obtained from,

$$\Delta \sigma = \left( \mathbf{S}^T \mathbf{S} + \lambda \mathbf{R}^T \mathbf{R} \right)^{-1} \mathbf{S}^T \Delta v,$$

(3)

where $\mathbf{S}^T \mathbf{S}$ is an approximation of the Hessian, $\mathbf{R}$ and $\lambda$ are the regularization matrix and regularization parameter, respectively [8]. A proper choice of $\lambda$ can be obtained, e.g. according to the Morozov’s discrepancy principle criterion [9].

2.2. Simulation Setup

2.2.1. Chest Model

The model for the simulation was obtained by segmenting the MRI images to construct a three dimensional volume (see, figure 1). Using diffused imaging modalities, reconstruction of small details is infeasible; therefore, the model was constructed roughly by considering only the extended tissue groups such as the lungs, heart and the surrounding muscle tissue. The conductivity values for the tissues were assigned based on a parameterized Cole equation. The conductivities of the muscle, lung and heart tissues were taken as 0.35 S m$^{-1}$, 0.18 S m$^{-1}$ and 0.21 S m$^{-1}$, respectively, at 200 kHz operating frequency.

2.2.2. Fluid accumulation

The fluid accumulation that causes the so-called gravitational dependent collapse was assumed to take place in the posterior lungs. The collapsed region is a mixture of oedematous fluid together with the deflated lung tissue causing the conductivity of that region to be higher than the healthy tissue. We assumed 0.6 S m$^{-1}$ for that region [1].
2.2.3. Coil Configuration  The $5 \times 5$ coil matrix was assumed to be embedded in the patient’s bed and the corresponding arrangement is depicted in figure 1. The coils are assumed to act both as a transmitter and a receiver and have a radius of 20 mm. The approximate distance from the coil center to the body surface is 1 cm.

3. Results
Assume that the patient is in supine position and the measurements are taken at different times, $t_1$ and $t_2$. Using time-differential voltage measurements, i.e., $\Delta v = v(t_2) - v(t_1)$, one can monitor the change of fluid accumulation in time, i.e., $\Delta \sigma = \sigma(t_2) - \sigma(t_1)$. The initial measurements were taken from the healthy lungs at $t_1$ and the measurements at $t_2$ are taken assuming that the fluid accumulation has reached a state in which 2 cm of the lower lungs have collapsed. 1% uncorrelated Gaussian noise was added to the voltage data in order to simulate the noise of the receiver channels.

The reconstructed images of the fluid accumulation which has happened in both lungs simultaneously is shown in figure 2-(a). From the images, both collapsed regions can be identified in the regions close to the surface and in central regions they overlap and cannot be distinguished clearly. The remaining images in figure 2 represent the cases that the accumulation develops faster in one lung than the other, i.e., (b) and (c) corresponding to the collapse in the left lung and right lung only. The images show similar blurry characteristics and tend to shift towards the central regions which explains the overlap of reconstructions in figure 2-(a).
4. Discussions and conclusions

In this work, the measurements were assumed to be insensitive to the cardiac and respiratory cycles. The heart is placed in the ventral portion of the chest and, knowing that the sensitivity decays rapidly with depth, it is reasonable to neglect the corresponding effects. Breathing, on the other hand, affects the measurements much more than the cardiac cycles. However, the conductivity change between the deflated and inflated lungs are way lower than the conductivity changes caused by the lung collapse and thus the effects are neglected.

Compensating the effects of the patient’s movement during data acquisition is critical for this system [10]. For the patient in a supine position, the chest motion due to respiration is assumed to be only on the ventral surface and the movement artefacts are expected to be minimal for this positioning. However, there may still be a need to determine the patient’s placement in the bed accurately.

This preliminary work points out the potential of the method for imaging of the fluidic changes in the dorsal parts of the lungs. Since the sensitivity of the measurements drops rapidly with depth below the skin, only the inhomogeneities that are very close to the skin can be monitored. The future work may include finding optimal coil matrices so that the spatial resolution of the images are increased for particular locations [11] which may provide better diagnostic information.

Acknowledgments

This work was supported by the SFB project F32-N18 granted by the Austrian Science Fund.

References

[1] Luepschen H, Riesen D, Beckmann L, Hameyer K and Leonhardt S 2008 Magnetics, IEEE Transactions on 44 1450–1453
[2] Preis C, Luepschen H, Leonhardt S and Gommers D 2009 Clinical Physiology and Functional Imaging 29 159–162
[3] Tarjan P P and McFee R 1968 Biomedical Engineering, IEEE Transactions on BME-15 266–278
[4] Steffen M, Heimann K, Bernstein N and Leonhardt S 2008 Physiological Measurement 29 291–306
[5] Gençer N G and Tek M N 1999 Medical Imaging, IEEE Transactions on 18 617–627
[6] Morris A, Griffiths H and Gough W 2001 Physiological Measurement 22 113–119
[7] Merwa R, Hollaus K, Brunner P and Scharfetter H 2005 Physiological Measurement 26 241–250
[8] Ziolkowski M, Gratkowski S and Palka R 2009 International Journal of Applied Electromagnetics and Mechanics 30 245–253
[9] Hansen P C 1998 Rank-deficient and discrete ill-posed problems (Philadelphia: SIAM)
[10] Gürsoy D and Scharfetter H 2009 Physiological Measurement 30 165–174
[11] Gürsoy D and Scharfetter H 2009 Biomedical Engineering, IEEE Transactions on 56 1435–1441