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Errors due to modeling mismatch between the simulated and experimental coil setup in magnetic induction tomography

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Abstract

Image reconstruction in magnetic induction tomography (MIT) is usually carried out by minimizing the residuals between the estimated and measured quantities assuming a structurally correct model. Thus, any mismatch between the simulated and the true experimental coil setup alters the data and may cause artifacts in the images. In this paper, we performed a simulation study to investigate the effect of modeling mismatches on measurements and corresponding reconstructed images. This is particularly important for adjustable MIT systems that have many degrees of freedom such as the Graz Mk2 MIT system and may also influence the research on wearable MIT systems, the sensors of which are not fixed rigidly in space.

1 Introduction

Magnetic induction tomography (MIT) is an emerging imaging modality which attempts to image the electrical conductivity of the human body [1, 2]. An eddy-current density is generated via magnetic induction using the transmitter coils and the changes in the magnetic field due to the conductivity perturbations within the body are recorded by an array of receiver coils. The noninvasive and contactless measurements makes the imaging modality attractive.

The corresponding inverse problem for the conductivity reconstruction is severely ill-posed and regularization methods are essential to overcome ill-posedness. The common approach for the inversion is minimizing the L_2-norm of the residuals between the estimated and measured data and the solutions are computed using the regularized Gauss-Newton method [3].

The image reconstruction is commonly formulated based on the norm of the residuals. Thus, modeling mismatches between the experimental system and the corresponding simulation model alter the data and may cause artifacts in the images. The errors originated from the body distortions, such as breathing and random movements, were investigated previously [4]. However, the errors originated from imperfect modeling of the experimental coils have not so far been investigated. To this end, a number of different coil distortions were defined and the corresponding errors in the measurement data and the corresponding imaging artifacts were presented.

This is particularly important for systems that have many degrees of freedom and require different adjustments for different imaging sessions as this is the case in the Graz Mk2 MIT system. This may also influence the research on wearable MIT systems the sensors of which are not fixed rigidly in space.

2 Methods

2.1 Simulation of data

To acquire the induced voltage data in the receiver coils, the computation of the electric field produced by an energized coil in the proximity of a volume conductor is needed [5]. According to the reciprocity theorem, the corresponding expression for data simulation is as follows:

$$v(\sigma) = \int_{\Omega_c} \sigma \mathbf{E}_1(\sigma) \cdot \mathbf{E}_2(\sigma) dV, \tag{1}$$

where $\mathbf{E}_1$ and $\mathbf{E}_2$, respectively, denote the direct and adjoint electric fields induced in the body, Ω_c and σ denote the conductive domain and electrical conductivity distribution.

Due to the nonlinear nature of the problem, absolute imaging fails to perform satisfactorily, and thus differential imaging is preferred. Thus, the change in measurements due to the change in conductivity for two different states can be expressed as follows,

$$\Delta v_{a,b} = v_a(\sigma_a) - v_b(\sigma_b), \tag{2}$$

where v_a and v_b represents two sets of voltages due to different conductivity distributions, σ_a and σ_b, respectively.

To image the temporal changes in the conductivity distribution (time diffential imaging), the measurements are taken at different times and time difference data is used to reconstruct tomograms. By this way, it is possible to image, for instance, fluid accumulation in the lungs, edema development or ventillation. Likewise, to image the spectral changes in the conductivity distribution (frequency diffential imaging), the measurements are recorded at different operating frequencies. However, in this case, due to the dependency of the amplitude of data on the frequency, the data must be scaled accordingly to form the difference dataset. This regime is more promising for clinical applications, since the measurements can be taken simultaneously by exciting the body using two different frequencies at the same time, and thus the systematic errors are considerably supressed. Imaging of motionless organs such as brain or pelvis seems more appropriate with this regime.

2.2 Simulation setup

Ideal coil geometry: For excitation, 16 identical solenoid transmit coils with 25 mm radius were used with their centres uniformly positioned on a circular ring of 140 mm radius. The arrangement is depicted in figure 1 (see, figure 2 for top-view). For measurement simulation, 16 receiver coils with 25 mm radius were placed with their centres on an inner ring of 130 mm radius.
Phantom geometry: A cylindrical volume conductor of 100 mm radius and 200 mm height with a local spherical inhomogeneity inside was used as a phantom. The inhomogeneity had a comparatively small radius of 10 mm and was located at [50,0,0] mm (see, figure 2). The electrical conductivity for the volume conductor and for the inhomogeneity were chosen as 0.1 Sm^{-1} and 0.2 Sm^{-1}, respectively.

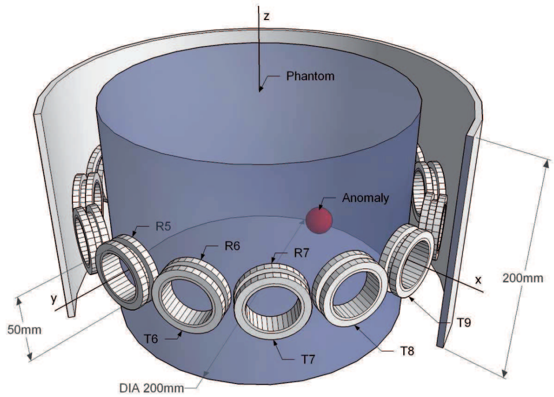

Figure 1: Sketch of an 16-channel MIT system.

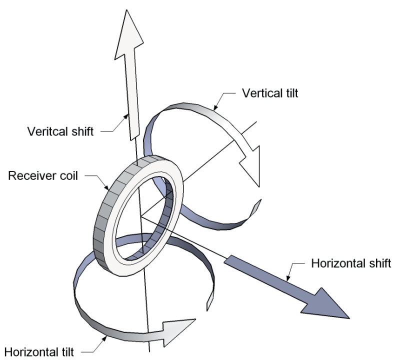

Figure 2: Sketch of the coil distorsions.

2.3 Coil distortions

We investigated 4 possible different types of distortions and the detailed sketch for each type is given in figure 3. "horizontal tilt" and "vertical tilt" denote the rotation around the sketched coil axes. Likewise, "horizontal shift" and "vertical shift" represent the displacements of the coil in z-axis and away from the phantom surface, respectively.

2.4 Model mismatch errors

The ideal synthetic data, Δv_{ideal}, is simulated using the given phantom and the ideal coil setup. In a second step, we assumed a slightly distorted geometry of the receiver coils, as defined previously in figure 3, to simulate the measurements, i.e. experimental data, Δv_{meas}. By doing so, the relative error between ideal and experimental data can be expressed as,

$$\eta = \frac{\Delta v_{meas} - \Delta v_{ideal}}{\Delta v_{ideal,RMS}}, \tag{3}$$

where $\Delta v_{ideal,RMS}$ is a constant which represents the root-mean-square of the ideal measurements and η denotes the relative error.

2.5 Imaging Artifacts

The artifacts in the images can be estimated by mapping the absolute errors onto the imaging domain as follows,

$$\delta = \mathbf{G}_{ideal}\left(\Delta v_{meas} - \Delta v_{ideal}\right), \tag{4}$$

where $\mathbf{G}_{ideal}$ denotes the "pseudo-inverse operator" for the ideal setup. In this study, the inversion was performed based on the least-squares data fitting by the Gauss-Newton method.

3 Results

The measurement data were simulated by changing the conductivity of the perturbations from 0.1 Sm^{-1} to 0.2 Sm^{-1} assuming a constant background conductivity of 0.1 Sm^{-1}. The relative errors, (i.e., η as defined in (3)) between the ideal and distorted receiver coil configurations were computed for each voltage data and presented in figure 3. The images represent the relative errors in the receiver channels (rows) for the corresponding active transmitter coil (columns). The relative error lies between 0 and 1. A value of 1 denotes that the errors due to distortion are as large as the ideal signals. Likewise, 0.1, 0.01 and 0.001 isolines represent the error levels. The topology of all relative errors showed similar "butterfly shaped" characteristics except in case of the horizontal tilt. For horizontal tilt, the most erroneous channel was R9, the data amplitude of which is consideably smaller than the adjacent channels' as contrary to other types of distortions. Vertical distortions up to 5 mm and 5^o yield relative errors less than 0.015, which is significantly less than the errors produced by horizontal distortions the errors of which reach up to 0.35.

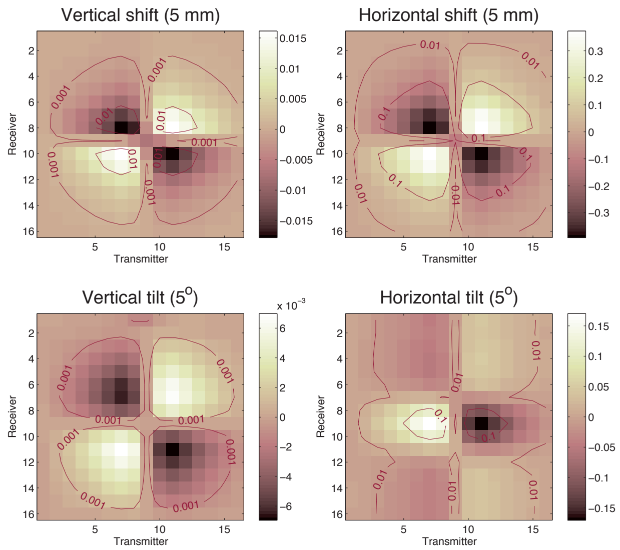

Figure 3: Relative error for each voltage data. The isolines represent the error levels.

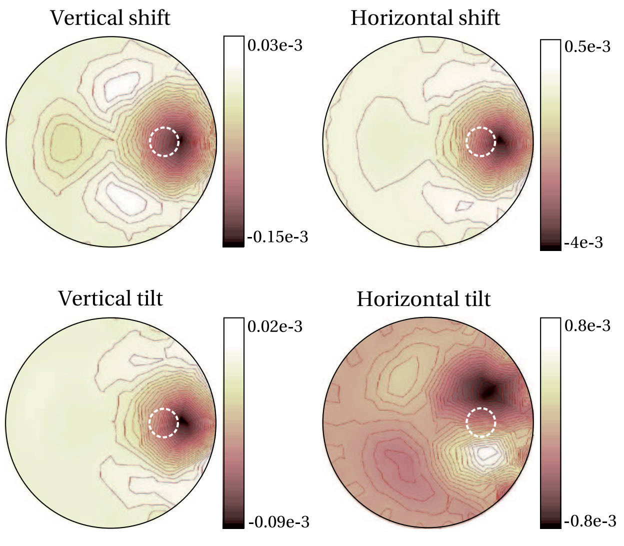

Figure 4: Artifacts in the tomograms for different coil distorsions.

Figure 4 presents the reconstructed artifacts of the central slice that are obtained with the erroneous data (δ in (4)). These images represent the imaging artifacts that one can encounter due the to wrong modeling of the system. The dotted circles delineate the position of the conductivity anomaly. For the vertical shifts, the artifacts appeared near the anomaly region but slightly shifted towards the boundary. For the horizontal tilt, two mirror artifacts with different signs were observed. The largest range in absolute magnitude was found to be for the horizontal shift, i.e. -4e-3. All the artifact images except the case of the horizontal tilt showed similar topologies that they tend to reduce the amplitude of the target anomaly.
Considering the horizontal tilt of the receiver coils, mirror artifacts with different signs can be expected near the sides of the anomaly. For other types of distortions, the amplitude of the reconstructed images decreases particularly at the location of the anomlaly.

4 Discussions

5 mm and 5^o distortions of the coils may cause up to 35% deviations in the data considering a small perturbation of 10 mm radius at 50 mm depth. When the exact model is not known accurately, this error may be interpretted as a noise term which corresponds to a level of up to -10 dB SNR and can be used as *a prior* information to regularize the inversions and improve the imaging performances. However, the uncertainties of the actual location and orientations must be estimated beforehand, e.g. using a Bayesian approach.
Due to the variablility of the body size and shape of the patients the coil system must be adjusted for each patient before data acquisition. Thus, possible solutions to accurately model the coil setup in a shorter time are essential to decrease the duration of the imaging sesison. If the system does not need adjustment, for instance in head applications, then a rigid mechanical support appears to be an important design issue to improve SNR.

5 Acknowledgements

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