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Evaluation of Magnetic Field Tomography Inverse Problem Solutions Using Physical Phantoms

ABSTRACT

Computed tomography is the technique of estimating the interior of objects from measurements through the object, well-known from medical applications. In the case of magnetic field tomography (MFT) measurements of ionizing radiation is relieved by that of magnetic fields. The corresponding inverse field problems are usually solved applying minimum norm solutions. The required magnetic field data can be measured using MFT techniques. Then, the current density distribution producing the magnetic field can be reconstructed or certain current sources can be identified.
We describe inverse techniques usually applied in modern biomedical investigations of bioelectromagnetism and their aptitude for reconstruction of current density distributions. Furthermore, we consider the possibilities to verify the corresponding inverse solution methods. We demonstrate how physical phantoms can be used to verify computer simulation as well as measuring systems. Finally, we apply this strategy to the solution of an inverse problem in magnetic fluid dynamics (MFD) where the goal is the identification or reconstruction of the interface between two conductive fluids using MFT measuring data which have been recorded from phantom experiments.

INTRODUCTION

The word *´tomography´* is derived from Greek words *´tomos´* meaning *´to slice´* and *´graph´* meaning *´image´*. The Oxford English Dictionary defines ´Tomography´ as:

Radiography in which an image of a predetermined plane in the body or other object is obtained by rotating the detector and the source of radiation in such a way that points outside the plane give a blurred image. Also in extended use, any analogous technique using other forms of radiation.

As it can be concluded from this description, tomography is often perceived as an imaging tool for medical examination purposes. It has to be emphasized, however, that the concept of tomography and its non-invasive way of imaging are not restricted to the medical field. Tomography has been developed, over the last 15 years, into a reliable tool for imaging numerous industrial applications. This field of application is commonly known as *Industrial Process Tomography (IPT)* or simply *Process Tomography (PT)* [9].

Tomography is the study of the reconstruction of two- and three-dimensional objects from one-dimensional slices. Computed tomography is the technique of estimating the interior of objects from measurements through the object, well-known from medical applications where in the past usually X-rays have been used. In the case of the *Magnetic Field Tomography (MFT)* the dangerous, ionizing radiation is relieved by the harmless magnetic field measurements. In this paper is shown how this, in the framework of bioelectromagnetism well-established MFT concept can be applied, and which possibilities exist to verify the results of these computer simulation techniques.

TOMOGRAPHY TECHNIQUES

Tomography is a process that extracts a reconstruction of the imaged object from a set of input projections of the object. The basic problem in computerized tomography is the reconstruction of a function from its line or plane integrals. For example, X-ray and MRI tomography is used to image a cross-section through a patient, and taking a series of these cross-sections allows a complete 3-D density map of the body to be built up. A similar concept is used in the case of vector tomography, allowing vector maps of a field to be extracted from measurements by a suitable probe. The inverse Radon transform is a mathematical tool to reconstruct a cross section of an object if an infinite number of one-dimensional projections of an object taken at an infinite number of angles are available. However, the inverse Radon transform proves to be extremely unstable with respect to noisy data. In practice, a stabilized and discretized version of the inverse Radon transform is used, known as the filtered back projection algorithm.

In tomography, a variety of practical reconstruction algorithms have been developed to implement the process of reconstruction of a 3D object from its projections. These algorithms are designed largely based on the mathematics of the Radon transform, statistical knowledge of the data acquisition process and geometry of the data imaging system. In medical tomography, the filtered back projection algorithm and its variants are the most efficient algorithms currently in use. More recent kinds of tomography replace straight line models by an inverse problem for partial differential equations [22].

There are many quantities which can be utilized in tomography, e.g. measurements of voltage, current, field, magnetization, impedance, density, radiation, capacitance, inductance, etc. From those measurements the profiles or spatial distributions of diverse physical quantities, like conductivity, permeability, permittivity, magnetization, sound, light, radiation, etc. can be reconstructed. Consequently, we can find various applications of tomography techniques:

- X-Ray Tomography (CT, CAT)
- Infrared Tomography (IRT)
- Ultrasonic Tomography (UT)
- Magnetic Resonance Tomography (MRT, MRI, fMRI)
- Positron Emission Tomography (PET)
- Single Photon Emission Tomography (SPECT)
- Electrical Impedance Tomography (EIT)
- Electrical Capacitance Tomography (ECT)
- Electromagnetic Field Tomography (EFT)
- Magnetic Induction Tomography (MIT)
- Magnetic Field Tomography (MFT)
- . . .

Much research effort has been devoted to the development of tomography techniques for imaging the low-frequency (<2 MHz), passive electromagnetic properties of materials non-invasively. Most of the interest has been in the area of medical imaging, where cross-sectional images of the human body are sought, and industrial imaging for the visualization and control of processes in vessels, pipelines, etc. Electrical imaging has also been used in archaeology for imaging submerged remains or geoelectrical investigations. The oldest of the electrical imaging techniques is *Electrical Impedance Tomography (EIT)*, which normally involves attaching an array of surface electrodes around the region to be imaged. Currents are injected and electrical potentials measured via the electrodes.

Another technique, the *Electrical Capacitance Tomography (ECT)*, is very similar to EIT. It also uses an array of electrodes and applies an electric field to the material, but it differs in the way the measurements are made: instead of a measurement of transimpedance involving four electrodes at a time, capacitance is measured between different pairs of electrodes. ECT is designed for materials of low permittivity and negligible conductivity imaged through an insulating boundary.

The most recent and least developed technique is *Magnetic Induction*

Tomography (MIT), where the first reports appeared in the early nineties. MIT applies a magnetic field from an exciting coil to induce eddy currents in the material and the magnetic field from these is then detected by sensing coils. The technique has been variously named *Mutual Induction Tomography* (also MIT) and *Electromagnetic Tomography (EMT)*. MIT is sensitive to all three passive electromagnetic properties: conductivity, permittivity and permeability [9]. Currently, there are a number of tomography techniques available for studying complex multiphase phenomena. These include, for example, infrared, optical, X-ray and Gamma-ray tomography systems, *Positron Emission Tomography (PET)*, *Magnetic Resonance Tomography (MRT)*, and sonic or ultrasonic tomography systems. Each of these techniques has its advantages, disadvantages and limitations. The choice of a particular technique is usually dictated by many, very often contradictory, factors. These include physical properties of the constituents of multiphase flow, the desired spatial and temporal resolution of imaging, costs of the equipment, its physical dimensions, human resources needed to operate it, and potential hazards to the personnel involved (e.g. radiation).

MAGNETIC FIELD TOMOGRAPHY

In this paper we discuss possibilities to verify the results of numerical simulations for reconstruction of distributed current sources in 3D-space. Therefore, the current dipole localization and/or current density reconstruction in biomagnetism and the interface identification at a two-fluid-cell of a magnetic fluid dynamics problem have been considered. In both cases the MFT technique has been applied, i.e. multi-channel magnetic field measurements are used either to reconstruct the current density distribution in space (or on a certain surface) in biomagnetic studies or to identify the interface deformation between two conductive fluids in the case of a MFD application.

Application in Biomagnetism

Electric potential and magnetic fields associated with the intracellular currents that flow within the active cells of the cortex or heart muscles provide a means to monitor the spatial-temporal evolution of the cortical activity within the brain or the electrical activity of the heart. Recording electrical potential differences from the body surface (electroencephalography (EEG), electrocardiography (ECG)) is possible via surface electrodes. The weak magnetic fields outside the body can be obtained by superconducting quantum interference device (SQUID) magnetometers (magnetoencephalography (MEG), magnetocardiography (MCG)). Thus, by measuring potentials and magnetic fields, it is possible to demonstrate the continuous brain or heart activity. Moreover, one goal in electric and magnetic recordings is to form an image of the electrical sources distributed across the cortex or in the heart muscles. Such representations, obtained via event related measurements, provide a non-invasive means to obtain information about the centers (or spatial extension) of related brain or heart functions. For convenience, we call such a representation *electric source imaging (ESI)* when it is based on electric data, *magnetic source imaging (MSI)* when the images are obtained via magnetic data, and *electromagnetic source imaging (EMSI)* when it is based in bimodal data.

The electric potential patterns on the body surface and the magnetic field patterns measured near the surface are similar to the field patterns of a dipole source. As a result, the current dipole model is traditionally employed to represent the center of simultaneously active sources. Finding a representation for the source distribution corresponds to finding the locations and directions of the current dipoles from measurement patterns at consecutive instants. Several methods have been suggested to solve this problem which is referred to as the *inverse problem*. Multiple dipole fits are developed to find the directions and locations of a number of dipoles. Such methods are non-linear as the dipole locations are non-linearly related to the measurements. If the dipole locations are obtained by some means then it is possible to apply linear inversion methods, as the measurements are linearly dependent to the dipole strengths. The locations and directions of finite number of dipoles (typically ten thousands) are then known but their strengths are unknowns.

Linear and non-linear methods in numerical computations are based on differences between measurements and calculated fields obtained for an estimated source configuration. Thus, solution of potential and/or magnetic fields for a given dipole configuration (*forward problem*) is required.

In order to obtain accurate representations of physiological sources, the associated fields must be properly sampled. The recent advent of very large arrays of electrodes and superconducting sensors provide EEG/MEG or ECG/MCG data from over two hundred channels. In addition to the need for sufficient number of data, a methodology must be developed to accurately model the subject´s geometry. New generation magnetic resonance imaging (MRI) systems provide images with spatial resolution under 1 mm. The progress in computer technology continuously increases the speed and memory capabilities of even personal computers. Thus it is now an appropriate time to launch an effort on the development of accurate body models of individuals. This will incorporate the correct geometry of the body and electrical properties of the tissues, thus yielding a better interpretation of the measured data. On the other hand, there is a need to have possibilities to verify the computer simulations, i.e. to estimate the errors by means of comparing measured data with computed results both known from well-defined reference models. This could be done using physical phantoms.

Application in Magnetic Fluid Dynamics

There are a variety of problems in material processing where it would be useful to know the time-dependent distribution of the electrical conductivity of a single fluid or a multiphase flow. For instance, the knowledge of the position of the interface between highly conducting molten aluminium and poorly conducting electrolyte (cryolite) is important to prevent unwelcome instabilities in aluminium reduction cells. Recently, it has been demonstrated that the MFT concept can be successfully used for detection of interfaces between current carrying fluids of different electrical conductivity [8,6]. It was shown that the external magnetic field generated by the electrical current flowing in a highly simplified model of an aluminium reduction cell provides sufficient information to reconstruct the unknown interface characteristics. In the reconstruction process genetic algorithms have been applied. Evolutionary algorithms like *Genetic Algorithms (GA)* or *Evolution Strategies (ES)* are particularly effective when the goal is to find an approximate global minimum (or maximum) in a multimodal function domain [32,33]. We have already shown how these techniques can be applied to MFD problems [7,10,35].

SOURCE RECONSTRUCTION IN BIOELECTROMAGNETISM

Magnetic Inverse Problem

The goal of the (quasi-static) magnetic inverse problem is to estimate the source current density underlying the signals measured outside the object. Unfortunately, the primary current distribution cannot be recovered uniquely, even if the magnetic field (or the electric potential) were precisely known everywhere outside the body.

Source reconstruction for magnetic field tomography is an especially difficult inverse problem. The Maxwell's equations have to be applied and for the simpler cases these reduce to a second-order partial differential equation for which exist only very few analytic solutions. Numerical methods have to be applied to solve the equations, subject to boundary conditions which are specified on the electrodes or body surface.

Calculating the magnetic field distribution from predefined sources - the forward solution – is the easy part. The source identification/reconstruction using the magnetic field measurements - the inverse problem - usually progresses iteratively by guessing the answer and then calculating a new source configuration that will fit the measurements better. The found solution is then further refined because the inverse problem is strongly nonlinear. Care must be taken with the computations otherwise the iterations will not converge, or they will converge to local solutions. This situation is common to inverse problems and it is necessary to use a regularization technique to obtain stable solutions. Furthermore, it is often possible to use additional physical information to constrain the problem and to facilitate the solution. This is the reason why it is so important to have well-adapted source models, i.e. model descriptions which include those features that have to be estimated during the inverse solution.

Source Localization

There are two major considerations in interpreting biomedical signals. The first involves maximizing the signal-to-noise ratio of the measurement. The second involves *´the inverse problem´* i.e., the backwards inference needed to go from a distribution of measurements on the body to a set of sources in the body.

There are two technologies which have been mentioned so far for minimizing distant signals: positioning the MSI inside a magnetically shielded room, and using gradiometers to focus on local field differences. These aspects will not be considered in this paper.

The inverse problem applies quite similar to electrical case (EEG) as well as to MEG. One can solve the ´forward problem´ exactly, taking any number of sources in the brain and calculating what the detectors outside the body will measure. However, a given distribution on the scalp could be the result of any of an infinitely large number of different configurations of local current sources inside the volume of the brain. To solve this non-unique problem, it is necessary to make reasonable assumptions about the number of possible sources, and their approximate locations. Fortunately, there are many sources of information about specific brain areas and neural responses that permit this kind informational ´bootstrapping´. The situation is improving with the use of structural and functional imaging information from other modalities, especially MRI and functional MRI (fMRI).

Regularization

The inverse problems of bioelectromagnetism are ill-posed meaning that small measurement errors can lead to large artefacts in the reconstructed images. A regularization technique is needed to come to a stable solution. Sometimes the moving dipole model, the multiple dipole approach and the multipole expansion are also called "regularization techniques" since they stabilize a solution by using specific model assumptions. From a more general point of view, we can describe the problem of source identification and source reconstruction in the following way. The basic equation for the forward problem is

$$\underline{L} \cdot \underline{x} = \underline{b}$$

with lead field matrix $\underline{L}$, the unknown source term $\underline{x}$ and measured signals $\underline{b}$. The minimum-norm-solution (Tikhonov zero-order [27])

$$\underline{\hat{x}} \ = \ \min \left\{ \left\| \underline{L} \cdot \underline{\hat{x}} - \underline{b} \right\|^2 + \gamma \left\| \underline{\hat{x}} \right\|^2 \right\}$$

is easily given by:

$$\underline{\hat{x}} \ = \underline{L}^T \left(\underline{L}\underline{L}^T + \gamma \underline{I} \right)^{-1} \cdot \underline{b}$$

A generalization of this approach is:

$$\underline{\hat{x}} \ = \ \min \left\{ \left\| \underline{W} \cdot \left(\underline{L} \cdot \underline{\hat{x}} - \underline{b} \right) \right\|_2^2 + \gamma \cdot \left\| \underline{B} \cdot \underline{\hat{x}} - \underline{d} \right\|_p^p \right\}$$

There is possible to choose

- a weighting matrix W to compensate for different SNR of various sensors
- a weighting matrix B for depth normalization and/or a mollifier matrix B, e.g. the Laplacian Δ ([23,24])
- other norms for the regularization term (e.g., p = 1, p = 2, p = ∞)

The following approaches can be found in the literature [11]:

- spatial techniques (Tikhonov 0. and 2. Order, adaptive local regularization)
- maximum entropy/expectation/likelihood
- temporal techniques
- spatio-temporal techniques
- multiple constraints
- admissible solution approach

If Tikhonov regularization methods are used with different assumptions of the norm (L_1, L_2, L_∞), every choice demands for a specific depth-normalization - otherwise the solution will have a bias in the depth direction. Every regularization leads to a spatial filtering of the solution, i.e. the imaging system behaves like a spatial low-pass filter. This is typical for any imaging system. Unfortunately, in our case the smallest wavelength (that is the highest spatial frequency) that can be detected is in the range of 10 mm. Many authors try hard to push the lowest detectable wavelength down e. g. to 8 mm by taking the risk that large artefacts might show up in the images. The right choice of the regularization parameter also is a critical issue. The L-curve method is used quite often [15].

Magnetic Source Imaging

Magnetic Source Imaging (MSI) is a non-invasive technique that measures signals from the magnetic field generated by the brains (or other electrically active sources, like heart, etc.) electrical activity. These signals, one billion times smaller than those of an ordinary light bulb, reveal how information travels through and thus are processed by the brain. MSI is a relatively new clinical tool that provides neuroradiologists and neurosurgeons with dynamic spatial maps of neural events, tracking the electrical activity of neural units by detecting the tiny magnetic fields that they generate and which propagate outside the head. MSI can thereby monitor the inner communications in the brain which may include spontaneous rhythmic events, localized sensory and cognitive functions, and abnormal signals that may indicate either brain injury or sources of epileptic seizures. Very similar investigations can be done in the case of *Magnetocardiography (MCG)*, where the currents in the heart muscles generate the magnetic field.

A technology that can map the relevant brain processes which cause certain disorders would have to be able to distinguish and locate changes in brain activity that may occur over just tens of milliseconds, and which may reflect communications between small millimetre-sized functional units of cortex. The MSI technology has the properties needed to provide such information.

MSI is actually the product of a combination of *Magnetic Resonance Imaging (MRI)* and *Magnetoencephalography (MEG)*. A subject undergoes a MEG measurement and a MRI scan. Where MRI reveals the brain structure and any macroscopic structural abnormalities, MEG detects ongoing electrical currents generated by the neural activity, and then localizes significant sources of activity in the volume of the brain. By using a common frame of reference with landmarks on the head, MEG and MRI are then combined into MSI images, which consist of pictures showing neural activity localized to specific structures in the brain. These images can be viewed as MR slices and as 3D reconstructions of the brain´s outer surface, with the sources projected onto the surface, providing a surgeon´s-eye view of the brain as it would be seen with part of the skull removed.

Being a completely passive and non-invasive technology, MSI provides a relatively rapid and comfortable experience for patients being measured. A clinical measurement can be taken with the subject in either a seated or a supine position. MSI measurements, which may take 30 minutes to two hours (depending on the involved clinical application) are much quieter and less claustrophobic than typical MR examinations.

Source Modelling

Current Dipole Models

The simplest model for the current distribution consists of one or more point sources, current dipoles. In the simplest case the field distribution measured at one time instant is modelled by that produced by one current dipole. The best-fitting current dipole, commonly called the equivalent current dipole (ECD), can be found reliably by using standard non-linear least-squares optimization methods [18].

In the time-varying dipole model, first introduced to the analysis of biomedical data, an epoch of data is modelled with a set of current dipoles whose orientations and locations are fixed but the amplitudes are allowed to vary with time. This approach corresponds to the idea of small patches of the cerebral cortex or other structures activated simultaneously or in a sequence. The precise details of the current distribution within each patch cannot be revealed by the measurements, performed at a distance of 3 cm (or more) from the sources.

From a mathematical point of view, finding the best-fitting parameters for the time-varying multi-dipole model is a challenging task. Since the measured fields depend nonlinearly on the dipole position parameters, standard least-squares minimization routines may not yield the globally optimal estimates for these parameters. Therefore, more complex optimization algorithms have been suggested to take into account the physiological characteristics related to particular experiments [28].

Current Density Distributions

An alternative approach in source modelling is to assume that the sources are distributed within a volume or surface, the so-called source space, and then to use various estimation techniques to find out the most plausible source distribution. In MCG, the source space may be a volume defined by the heart muscle or restricted to the epicardial surface, determined from MR images. Distributed source modelling techniques may provide reasonable estimates of complex source configurations without having to resort to complicated dipole fitting strategies. However, the size of an activated region in the source images does not necessarily relate to the actual dimensions of the source but rather reflects an intrinsic limitation of the imaging method. In fact, without an extremely high signal-to-noise ratio it is unrealistic to claim that it would be possible to determine the real extent of a source giving rise to the magnetic signals.

The first current distribution model applied in MEG analysis was the (unweighted) minimum-norm estimate, one in a group of linear approaches which can be described in a common framework [16]. Here linearity means that the amplitudes of the currents are obtained by multiplying the data with a (time-independent) matrix. This kind of estimates has been employed by several authors [14, 18].

It is also possible to enter into the source imaging method the assumption that the activated areas have a small spatial extent. For example, a special algorithm, also called *´Magnetic Field Tomography (MFT)´*, obtains the solution as a result of an iteration in which the probability weighting is based on previous current estimate [17].

The L_1-norm approach employs the sum of the absolute values of the current over the source space as the criterion to select the best current distribution among those compatible with the measurement [14, 11]. The resulting *Minimum Current Estimation* is focally and may resemble the time-varying dipole model solution.

However, an important difference is that the source constellation is allowed to change as a function of time. Consequently, sequentially activated sources can be identified without the cross-talk problems inherent to the current dipole model [29-31].

VERIFICATION OF BIOMAGNETIC SOURCE RECONSTRUCTIONS

Evaluation Techniques

A lot of work has been devoted to the evaluation of the localization power of ECG and/or MCG concerning single current dipoles or other (extended) source configurations. Therefore, the following types of studies can be used:

- Computer simulation
- Phantom experiment
- Animal experiment
- Patient study

Whereas there is a high standard of animal experiments and patient studies in bioelectromagnetism, and an increasing number of works related to additional computer simulations, only a very few reports of phantom experiments can be found in the literature. Because of the high potential of such investigation including the chance to verify computer simulations (where usually no real error estimations are available) several joint studies with physical MCG/ECG phantoms have been started about ten years ago combining the theoretical expertises and technical experiences of scientists from the Biomedical Center at the Friedrich-Schiller-University Jena [19] and from the Department of Electromagnetic Fields/Theoretical Electrical Engineering at the Ilmenau University of Technology.

Torso Phantom and Modelling

Human body is often modelled as being homogeneous and isotropic in its electrical properties. However, in some tissue types the conductivity is strongly direction-dependent. Myocardium, brain and ordinary muscle tissue are examples of such anisotropic materials. In the myocardium, as well as in ordinary muscles, the anisotropy is due to the cell structure of the muscles, which favours the electric conduction into the direction of the cells and hinders the conduction in the perpendicular directions. In our computer simulations anisotropic behaviour could not be considered, a homogeneous realistically shaped boundary-element model of the phantom was utilized. The torso surface was tessellated with 1252 nodes and 2500 triangular elements, and a constant conductivity of 0.16 S/m was assigned inside of it. A single moving equivalent current dipole (ECD) was used as source model.

A fiberglass phantom with the shape of a truncated adult male torso was employed (for details, see http://www.biomag.uni-jena.de/romeo.htm) [1, 26, 2, 25, 3, 4, 5, 13]. An artificial current dipole was placed at several positions, where the depth ranged from 43mm to 133mm with 15mm spacing. The dipole and electrode locations were digitized with a 3-D digitization system (3SPACE ISOTRAK II, Polhemus Inc., Colchester, VT, USA) (Fig. 1, 2).

The phantom was first filled with a saline solution with the conductivity of 1.6 mS/cm. The surface potentials were recorded when activating each dipole with a sinusoidal current of 25 Hz (amplitude 1 µA). A layer of different conductivity was then inserted into the phantom. An artificial PVC-sternum was first mounted, with grooves on the left and right side. The thin inhomogeneity layer was fixed in these grooves (Fig. 3, 4). It was constructed of six 10x18 cm^2 acrylic frames. Ionic exchange membranes (NEOSEPTA type CM-2, NISSHO IWAI Deutschland GmbH, Duesseldorf, Germany) were fixed between these frames and PVC-covers. The membranes kept a steady concentration of NaCl in the compartments while the ionic current could flow freely. The maximal spacing between the inside wall of the phantom and the frames was 15 mm. The conductivity inside of the membrane was 16 mS/cm (i.e. 10 times the conductivity elsewhere in the phantom).

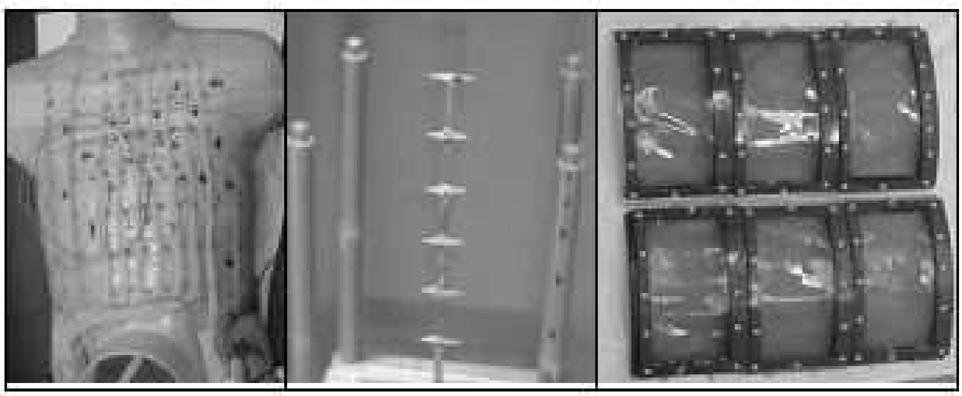

Fig. 1: Torso phantom, dipole sources and artificial inhomogeneity layers

In this special study, a fully anisotropic layer was not available. Still, the sheet of differing conductivity and the artificial sternum structure showed similar behaviour for the dipole localizations than in the above mentioned FEM study: the dipole localizations were systematically too deep [21, 20, 12]. More detailed FEM computations are still needed to assess more accurately how much error is introduced in the inverse solutions when the skeletal muscle layer and rib cage are not taken into account. Further investigations with fully anisotropic layers are under construction.

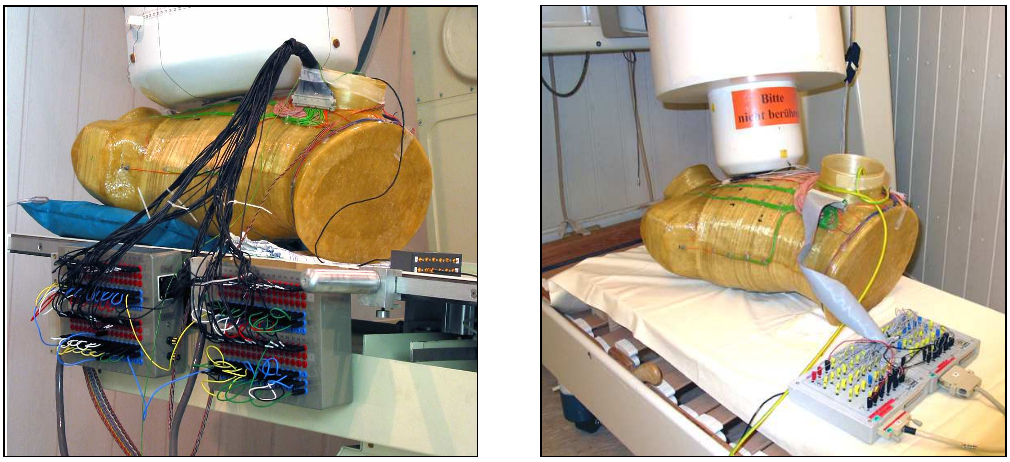

Fig. 2: Phantom measurements with vector magnetometer system (left) and one-component magnetometer (right) [Biomagnetic Center Jena]

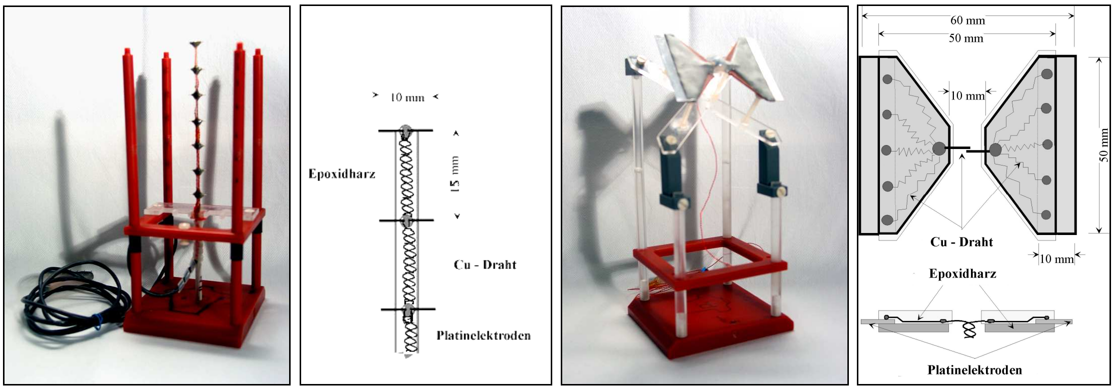

Fig. 3: Multiple dipole model (left) and extended source model (right), used for magnetic phantom studies at Biomagnetic Center Jena

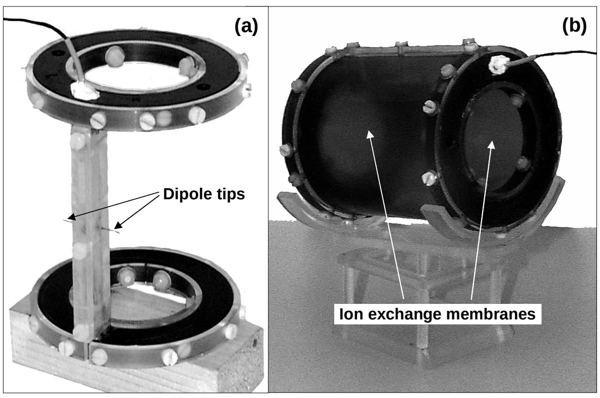

Fig. 4: Plastic frame that holds the compartment membranes, with the dipole tips integrated in the bar (a) and compartment with membranes on the mounting rack in measurement position (b).

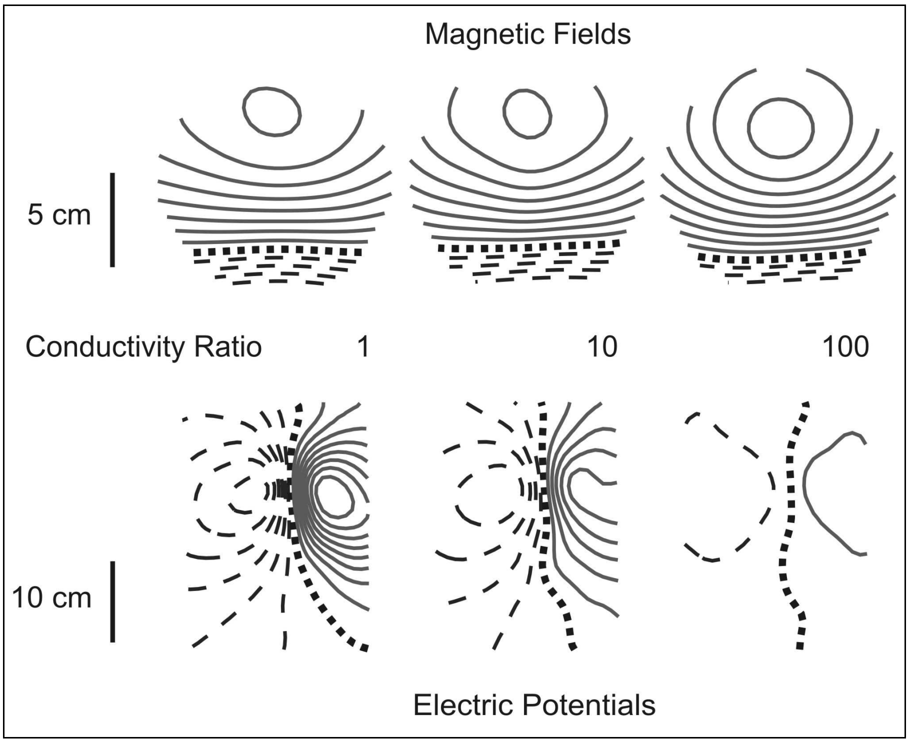

Fig. 5: Iso-contour lines of the magnetic fields (top row) and the electric potentials (bottom row) for different conductivity ratios. The magnetic field line increment is 5 pT and the electric potential line increment is 500 µV (solid lines indicate positive, dashed lines negative values, zero line is dotted).

An accuracy of 2 mm to 10 mm was achieved and MCG localizations have proven to be a little bit better than ECG localizations. Some results are promising others are a little bit disappointing. Nevertheless, this is an obviously extremely interesting "work in progress" [12].

The dipole localization results in the homogeneous phantom showed an excellent agreement with the digitized dipole locations. When the inhomogeneity layer was inserted the surface potential patterns did not exhibit significant changes (Fig. 5). The inhomogeneous maps had slightly lower maximum amplitudes than corresponding homogeneous maps. Average 3D error increased considerably (from 6.7 mm to 11.3 mm) for the inhomogeneity layer. The most significant impact found in several studies was on the localization accuracy of the dipole depths. Dipoles close to the inhomogeneity layer seemed to be much deeper than the corresponding ones in the homogeneous phantom. The effect weakened for dipoles lying deeper.

PHANTOM STUDY FOR A MFT PROBLEM IN MFD

The application of the MFT principles is well-established in bioelectromagnetism but it is not the only one. There is a variety of problems in material processing where it would be useful to know the time-dependent distribution of the electrical conductivity of a single fluid or a multiphase flow. For instance, the knowledge of the position of the interface between highly conducting molten aluminium and poorly conducting electrolyte (cryolite) is important to prevent unwelcome instabilities in aluminium reduction cells. Recently, it has been demonstrated that the MFT concept can be successfully used for detection of interfaces between current carrying fluids of different electrical conductivity [33, 10, 35]. It was shown that the external magnetic field generated by the electrical current flowing in a highly simplified model of an aluminium reduction cell provides sufficient information to reconstruct the unknown interface characteristics.

Genetic algorithms have been applied to solve this inverse problem, because they are particularly effective when the goal is to find an approximate global minimum (or maximum) in a multimodal function domain. In this application numerical simulations of the interface shape function in a cylindrical two-compartment system have been compared with magnetic field measurements recorded from a simple test configuration (Fig. 6, 7).

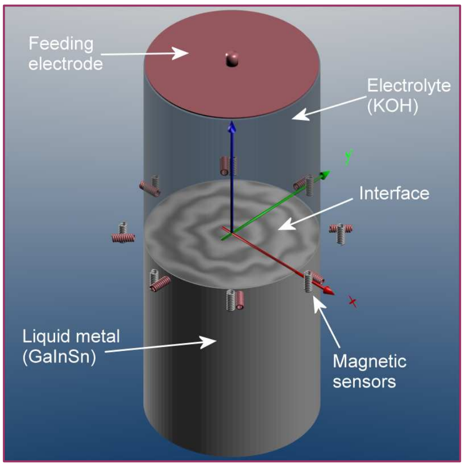

Fig. 6: Cylindrical two-fluid-cell phantom used for studies of MFT systems applied to a MFD problem

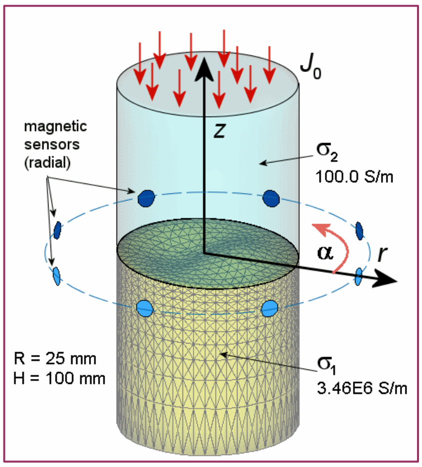

Fig. 7: Numerical model of the cylindrical two-fluid-cell used for FEM simulations

This model system can be considered as a physical phantom, where we have been able to study computer simulations as well as the measuring system which is a MFT system. Due to a certain controlling of the interface deformation modes we could test different approaches of the forward solution and of the inverse solution, i.e. the interface mode identification, as well. And thus, we could also optimize the measuring system consisting of eight 2D fluxgate magnetic field sensors mounted on a ring for measuring the radial and vertical components of the magnetic flux density. Because the sensor ring could be shifted into different heights (z-levels) the measuring data sets are very similar to those which are recorded with biomagnetic measuring systems.

CONCLUSIONS

In both applications we could proof that studies of physical phantoms are valuable tools for the evaluation of inverse problems solutions. Although not every feature of the real system can be modelled by the phantom with all details, the studies at least enable the verification of computer simulations where usually reference solutions are not available. Furthermore, in some cases the numerical solutions can even be validated by means of phantom studies. This can be of particular importance if stochastic optimization strategies are applied. Usually there are no reference solutions available. As an example for such an approach, i.e. use of stochastic optimization techniques, there has been applied recently to the solution of the inverse MEG problems where dipolar sources have been localized in the brain by means of the Genetic Algorithm and Simulated Annealing [28].

Another aspect strongly supporting further phantom studies is the optimization and evaluation as well as the comparison of different measuring systems. Thus, they can also be utilized for benchmarking of inverse solution strategies and magnetic field tomography systems, respectively.

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