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A new approach for electrical properties estimation using a global integral equation and improvements using high permittivity materials

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ABSTRACT

Electrical Properties Tomography (EPT) using MRI is a technique that has been developed to provide a new contrast mechanism for in vivo imaging. Currently the most common method relies on the solution of the homogeneous Helmholtz equation, which has limitations in accurate estimation at tissue interfaces. A new method proposed in this work combines a Maxwell's integral equation representation of the problem, and the use of high permittivity materials (HPM) to control the RF field, in order to reconstruct the electrical properties image. The magnetic field is represented by an integral equation considering each point as a contrast source. This equation can be solved in an inverse method. In this study we use a reference simulation or scout scan of a uniform phantom to provide an initial estimate for the inverse solution, which allows the estimation of the complex permittivity within a single iteration. Incorporating two setups with and without the HPM improves the reconstructed result, especially with respect to the very low electric field in the center of the sample. Electromagnetic simulations of the brain were performed at 3 T to generate the B_1^+ field maps and reconstruct the electric properties images. The standard deviations of the relative permittivity and conductivity were within 14% and 18%, respectively for a volume consisting of white matter, gray matter and cerebellum.

1. Introduction

Over the last decade there has been a growing interest within the MR community in measuring the electrical properties of human tissue in vivo [1–3], in part due to the availability of ultra-high field (7 T and above) MR scanners and the associated high specific absorption rate (SAR) [2]. One particular motivation is the desire to estimate in real-time patient-specific local SAR [3]. In order to fulfill these needs accurate conductivity maps are required. Databases of tissue conductivities over different frequency ranges already exist, but such datasets are based on ex-vivo measurements [4,5], which are not necessary equivalent to in-vivo values, and cannot represent either potential changes in pathological tissue or differences between patients. In addition to SAR considerations, measurement of electrical properties, both conductivity and permittivity, can serve by itself as an important MRI contrast [6,7]. It has already been shown in a large range of ex-vivo studies that both conductivity and permittivity of malignant tissues vary significantly from the values in healthy tissue [6–9]. In-vivo breast imaging studies have shown also a correlation between post-contrast MR images and conductivity maps [9,10]. In addition, a study of the electrical properties measured using a radiofrequency (RF) probe showed improvements in the margin assessment in a surgical procedure, with the potential to reduce the need for repeated surgeries [11].

Currently the main MR-based imaging method for electrical properties estimation is termed Electrical Properties Tomography (EPT) and relies on a solution of the homogeneous Helmholtz equation for the electromagnetic field in space [1], involving a second order derivative of the estimated B_1^+ map [10]. This technique, however, has limitations at tissue interfaces as well as a high sensitivity to noise [3,9,10] due to the requirement for a second order derivative. An alternative approach generalizes the relevant Maxwell's equations and introduces a formulation that includes both transmit and receive sensitivity distributions, which allows to solve it for different coil designs [12]. Various approaches have been introduced to increase the sensitivity of the EPT method, including utilizing a multi-channel transmit coil for a more stable solution based on relative coil sensitivities [13] and gradient-based EPT [14]. Another method that relies on injected current from electrodes, called magnetic resonance electrical impedance

Abbreviations: EPT, Electrical Properties Tomography; HPM, high permittivity material; MREIT, magnetic resonance electrical impedance tomography; SAR, specific absorption rate; RF, radiofrequency; WM, white matter; GM, gray matter.

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tomography (MREIT) [15] was demonstrated even earlier than EPT, but it has lower sensitivity due to the limitations on the induced current that can be injected in an in-vivo setting.

Most of the methods for EPT rely on *local* field equations, either Maxwell's or Helmholtz's. A relatively new technique utilizes the *global* Maxwell integral equations. In this approach, each voxel can be described as a contrast source, representing the problem as an electromagnetic scattering problem. This approach is called Contrast Source Inversion EPT (CSI-EPT) and uses an iterative conjugate-gradient inverse solution and regularization methodology to reconstruct the complex permittivity [16,17]. Promising results using this approach have been shown in simulation studies performed at 3 T. A similar approach has been utilized in an inverse approach to design the optimal parameters of a high permittivity material (HPM) for a given transmit field distribution [18].

The aim of the current study is to design and evaluate a new method based on global integral equations which incorporates a priori information based on a simple reference EM simulation or scout MR scan. In order to improve the quality of the reconstruction we show how different EM fields can be generated within the body by scanning with and without high permittivity materials. Electromagnetic simulations of the brain were performed at 3 T to assess the feasibility of this approach, as well as investigating the effect of noise on solution stability.

2. Theory

2.1. Integral equation for the electromagnetic scattering problem

The transmit and receive RF fields are a function of the sample's electrical properties. The spatially-dependent complex permittivity can be expressed as

$$\varepsilon(\vec{r}) = \varepsilon_r(\vec{r})\varepsilon_b + i\frac{\sigma(\vec{r})}{\omega}, \tag{1}$$

where $\varepsilon_r(\vec{r})$ is the relative permittivity, ε_b the free space permittivity, $\sigma(\vec{r})$ the conductivity and ω the angular frequency. A contrast function, χ, is defined as

$$\chi(\vec{r}) = \frac{\varepsilon(\vec{r}) - \varepsilon_b}{\varepsilon_b} = \frac{\varepsilon(\vec{r})}{\varepsilon_b} - 1, \tag{2}$$

such that it normalizes the complex permittivity with respect to the free space permittivity at the frequency of interest. The global integral equation represents both electrical and magnetic fields as a background field plus a term proportional to the vector potential A, that can be expressed as a convolution of the induced currents $(\chi(\vec{r})E(\vec{r}))$ with the Green's tensor function [16,19]. B_1^+ is the circularly-polarized representation of the magnetic flux $(B_x + iB_y)/2$ and the relevant global integral equation is

$$B_1^+(\vec{r}) = B_1^+{}_{\text{unloaded}}(\vec{r}) + c\int_{\vec{r}} G^+(\vec{r} - \vec{r}')\chi(\vec{r}')E(r')dV \tag{3}$$

In this case the background field is the B_1^+ of an unloaded coil. The Green's tensor function G^+ represents the relationship between the displacement current source and the magnetic field [16,19]. Since a convolution is involved in this expression, we can also represent Eq. (3) by

$$B_1^+ \approx B_1^+{}_{\text{unloaded}} + \Delta x_1 \Delta x_2 DFT^{-1}\{DFT(G^+)\cdot DFT(\chi E)\} \tag{4}$$

where all the variables in this representation are matrix forms of the relevant parameters.

In this work a 2D in-plane solution is used, assuming components $E_x = E_y = B_z = 0$.

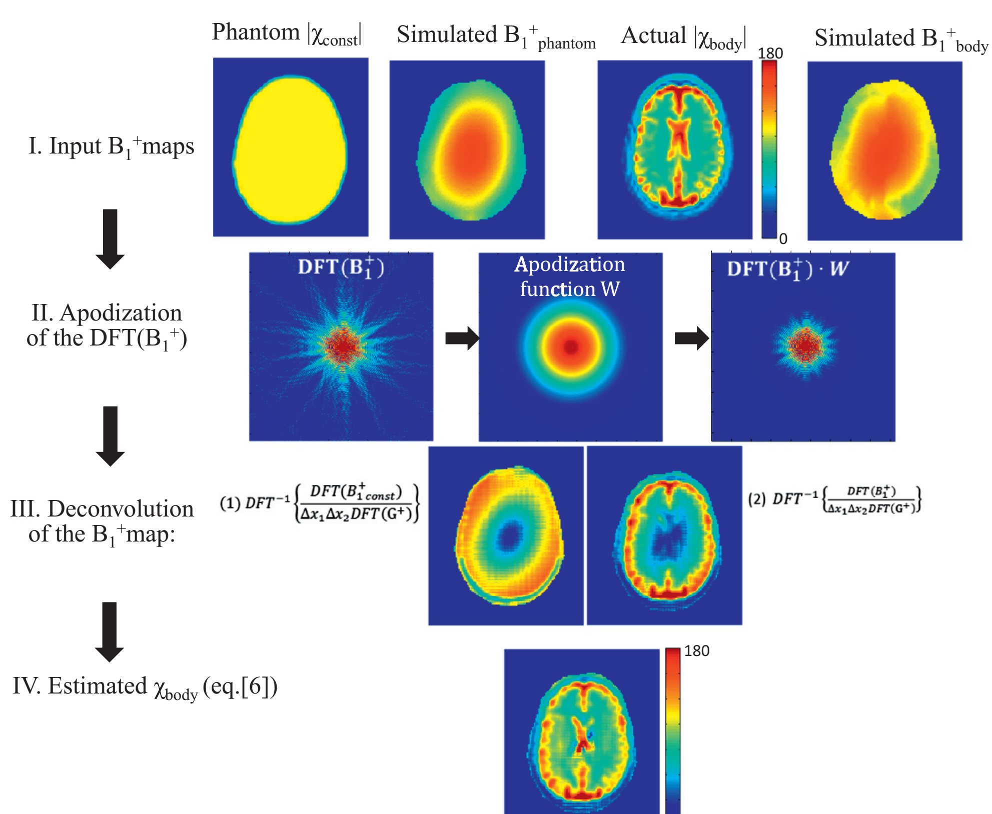

Fig. 1. Complex permittivity estimation procedure steps using a deconvolution of the B_1^+. (I) The input B_1^+ maps for two setups and the respective $|\chi|$, with χ_{const} and with χ_{body} (from left to right, respectively). (II) Example of apodization of the $DFT(B_1^+)$ using a Gaussian function W. $DFT(B_1^+)$ maps are normalized to 1 and the windowing is chosen to emphasize the apodization effect. (III) The deconvolved maps for both B_1^+ setups are shown. (IV) Final calculation using Eq. (6) is shown. The scaling for all χ maps is the same.