Experimental verification of Electric Properties Tomography (EPT)

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Introduction: The electric properties of human tissue, i.e., the electric conductivity and permittivity, can be helpful for a more precise prediction of the local SAR distribution during MR measurements or can potentially be used as an additional diagnostic parameter [1]. In this study, a new approach "Electric Properties Tomography" (EPT) is presented, which derives the patient's electric properties from the spatial sensitivity distributions of the applied RF coils. Initial experiments underline the principle feasibility of EPT on a standard MR system. The use of special MR systems optimized for EPT might further enhance the potential of EPT. In contrast to previous methods to measure the patient's electric properties [2-4], EPT does not apply externally mounted electrodes, currents, or RF probes, thus enhancing the practicability of the approach. Moreover, in opposite to the previous methods, EPT circumvents the solution of an inverse problem, which might lead to significantly higher spatial image resolution.

Theory: From Maxwell's equations, we obtain

$$\nabla \times \vec{\underline{H}}(\vec{r}) = i\omega \underline{\varepsilon}(\vec{r}) \vec{\underline{E}}(\underline{\varepsilon}(\vec{r}), \vec{r}) \tag{1}$$

with $\underline{H}$ the magnetic field strength, $\underline{E}$ the electric field, ω the Larmor frequency, and $\underline{\varepsilon}$ the (supposed to be isotropic) permittivity. The underlines denote complex variables. Eq. (1) can be solved for the unknown $\underline{\varepsilon}$ by regarding only the z-component

$$\left(\partial_x \underline{H}_y(\vec{r}) - \partial_y \underline{H}_x(\vec{r})\right) \Big/ \underline{E}_z(\underline{\varepsilon}(\vec{r}), \vec{r}) = i\omega \underline{\varepsilon}(\vec{r}) \ . \tag{2}$$

The real and imaginary part of $\underline{\varepsilon}$ can be identified with the (non-complex) permittivity ε and the electric conductivity σ, respectively. $\underline{H}_x$ and $\underline{H}_y$ can be measured via MRI by utilizing the sensitivities $\underline{H}^+$ and $\underline{H}^-$ of an RF coil for the transmit and receive case, respectively [5]. These sensitivities are given by the $\underline{H}$ component circularly polarized in the positive and negative direction, respectively

$$\underline{H}^+ = \left(\underline{H}_x + i\underline{H}_y\right)/2 \ , \quad \underline{H}^- = \left(\underline{H}_x - i\underline{H}_y\right)^*/2 \ . \tag{3}$$

Thus, the wanted components $\underline{H}_x$ and $\underline{H}_y$ can be deduced from Eq. (3). Finally, $\underline{E}_z$ has to be estimated via simulations in order to solve Eq. (2). The corresponding simulation setup is given by the (known) RF coil geometry and the patient's geometry known from the measurement of $\underline{H}^+$ and $\underline{H}^-$. Furthermore, since E_z itself is a function of $\underline{\varepsilon}$, an iteration has to be applied

$$\left(\partial_x \underline{H}_y(\vec{r}) - \partial_y \underline{H}_x(\vec{r})\right) \Big/ \underline{E}_z(\underline{\varepsilon}^n(\vec{r}), \vec{r}) = i\omega \underline{\varepsilon}^{n+1}(\vec{r}) \ . \tag{4}$$

The iteration starts with an estimation $\underline{\varepsilon}_0$, e.g., literature values of healthy tissue. Simulations have been performed confirming the convergence of this iteration.

Methods: To verify the described method experimentally, a cylindrical phantom (diameter = height = 18 cm) with a saline solution representing healthy tissue (ε=81, σ=0.6 S/m) was examined. The phantom contained two smaller cylindrical compartments (diameter = 7 cm, height = 18 cm) representing damaged tissue (left compartment: ethylenglycol solution [6], $\varepsilon_r = 55$, $\sigma = 1.2$ S/m, right compartment: saline solution, $\varepsilon_r = 81$, $\sigma = 0.3$ S/m). Measurements have been performed on a Philips Achieva 3T system using a birdcage head coil. TSE images have been acquired (TE = 12ms, TR = 4000ms, 256×256 pixels, FOV=20×20cm) using eight different flip angles α=10°,30°,50°,…,130°,150°. From this series, the coil sensitivity $\underline{H}^+$ has been derived by fitting [5]

$$\underline{S}(\alpha, \vec{r}) \sim \sin(const \cdot \alpha \cdot \underline{H}^+(\vec{r})) \tag{5}$$

in each pixel using a Levenberg-Marquardt algorithm [7]. Here, $\underline{S}$ denotes the images for the different flip angles, and *const* denotes a system-dependent constant. The nature of the birdcage coil leads to $\underline{H}^+ >> \underline{H}^-$, and thus, $\underline{H}^- = 0$ is assumed. A stack of five coronal images has been acquired to enable a through-plane numerical derivation of $\underline{H}^+$ via Savitzky-Golay filtering [7]. The electric fields are simulated using the software package CONCEPT [8].

Results: The raw MR image for α=90° is shown on Fig. 1. The image shows slight variations of the intensity, which, however, do not coincide intuitively with the underlying electric properties of the phantom. Figure 2 shows the reconstructed electric conductivity σ and Fig. 3 the reconstructed permittivity ε_r. The mean values in the corresponding areas are $\sigma = 1.03 \pm 0.13$ S/m, $\varepsilon_r = 40{,}9 \pm 17.0$ for the left compartment and $\sigma = 0.37 \pm 0.08$ S/m, $\varepsilon_r = 80.7 \pm 6.9$ for the right compartment. The values are normalized with respect to the reconstructed σ and ε_r in the outer area. Eight iterations via Eq. (4) have been performed.

Discussion/Conclusion: A method is presented determining the electric conductivity and permittivity of human tissue, based on the determination of the spatial sensitivity distributions of the involved RF coils. Furthermore, the corresponding electric fields of the RF coils have to be determined numerically. First experiments have been performed with a standard MR system. The satisfying results might serve as an initial proof of principal for the approach. Several simplifications have been applied, in particular the neglect of $\underline{H}^-$. Further studies should investigate the technique without these simplifications and test the possible resolution and accuracy. For clinical cases, a realistic patient model has to be included. This model might be simplified by segmenting patient compartments of constant electric properties, as performed in [4].

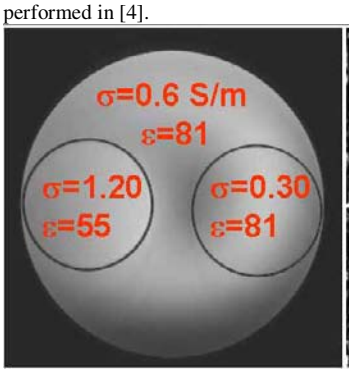

Fig. 1

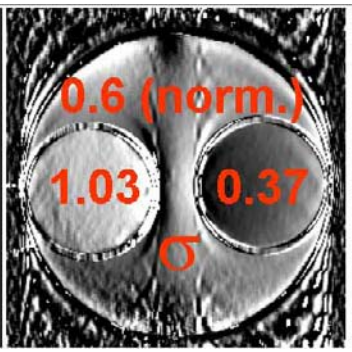

Fig. 2

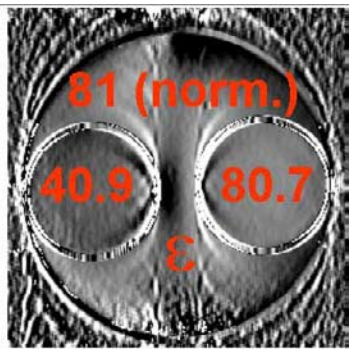

Fig. 3

Fig. 1: Raw MR image for α=90. The left compartment contains an ethylenglycol solution (modified σ and ε_r), the right compartment a saline solution (modified σ). Fig. 2: Reconstructed electric conductivity. The average values in the two inner compartments are normalized to the outer region. Fig. 3: Reconstructed permittivity. The average values in the two inner compartments are normalized to the outer region.

References:

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