Electric Properties Tomography (EPT) via MRI

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Introduction: MR is able to image a large variety of physiologic parameters. However, MR is not yet able to image electric properties, i.e., the electric conductivity and permittivity of human tissue. Up to now, these parameters are measured, e.g., with the help of externally applied electrodes [1,2] or RF probes [3]. All such imaging approaches define a (typically ill-posed) inverse problem, leading to very limited spatial resolution. In the current 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. Minor modifications of standard MR systems seem to be sufficient to perform EPT. However, also without these modifications, a principle proof of EPT might be feasible. EPT is not based on an inverse problem, and thus, might be able to achieve images of higher quality than achievable with previous approaches. The knowledge of the electric properties might be valuable, e.g., for tumour diagnostics (see, e.g., [4]) or the determination of the local SAR distribution during MR measurements.

Theory: From Maxwell's equations, we obtain

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

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

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

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

$$\underline{H}^+ = \left(\underline{H}_x + i\underline{H}_y\right)/2 \;\cdot\; \underline{H}^- = \left(\underline{H}_x - i\underline{H}_y\right)^*/2 \;\cdot \qquad (3)$$

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

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

The iteration starts with an estimation $\underline{\varepsilon}_o$, e.g., literature values of healthy tissue.

Methods: To test the feasibility of the described approach, a spherical phantom (diameter = 10 cm) with a saline solution representing healthy tissue ($\varepsilon_r = 60$, σ=0.5 S/m) was assumed. The phantom contained a small spherical compartment ($\varepsilon_r = 90$, σ=0.75 S/m, diameter = 2 cm). The fields $\underline{H}^+$, $\underline{H}^-$, and $\underline{E}$ were simulated for a birdcage head coil at 128 MHz using the software package CONCEPT [6]. The derivation of $\underline{H}^+$ and $\underline{H}^-$ was performed via Savitzky-Golay filtering [7]. The nature of the birdcage coil leads to $\underline{H}^+ >> \underline{H}^-$, and thus, also the assumption $\underline{H}^- = 0$ were tested. Furthermore, the convergence of the iteration Eq. (4) was tested, starting with $\underline{\varepsilon}_o$ homogeneous in the whole phantom. Artificial noise was added to the simulated $\underline{H}$ fields measurements.

Results: Figure 1 shows the reconstructed conductivity $\underline{\sigma}$ and permittivity $\underline{\varepsilon}_r$ for simulated SNR = 100. The mean error in the small compartment is below 1% for both $\underline{\sigma}$ and $\underline{\varepsilon}_r$. Figure 2 shows the same reconstruction, but assuming $\underline{H}^- = 0$. Here, the mean error in the small compartment is roughly 10% for $\underline{\sigma}$ and $\underline{\varepsilon}_r$. The enhanced noise in the center of Figs. 1,2 corresponds to a denominator in Eq. (2) $\underline{E}_z$ close to zero. Figure 3 visualizes the reconstruction iteration leading to Fig. 1 and Fig. 2.

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. Simulations confirmed the basic feasibility of the approach. For an assumed SNR = 100, a ratio of $\underline{\sigma}$ and $\underline{\varepsilon}_r$ of 50% between inner and outer compartment is clearly visible in the reconstruction. The results are still sufficient, if one polarization direction of the magnetic field is neglected. Thus, it should be possible to perform the technique on a standard MR scanner. For the investigated scenario, the reconstruction iteration (necessary to take the influence of $\underline{\sigma}$ and $\underline{\varepsilon}_r$ on the electric field into account) converges after 5-10 steps. Further studies should test the possible resolution and accuracy of the approach experimentally. For clinical cases, a realistic patient model has to be included in the approach. This model might be simplified by segmenting patient compartments of constant electric properties, as performed, e.g., in [3].

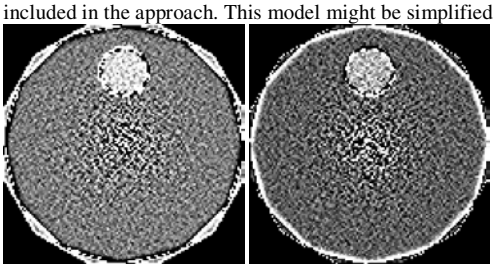

Fig. 1: Simulation results for reconstructing the electric conductivity (left) and permittivity (right). Both components $\underline{H}^+$ and $\underline{H}^-$ have been taken into account. The chosen noise corresponds to SNR = 100. The mean error in the small compartment is below 1% for both $\underline{\sigma}$ and $\underline{\varepsilon}_r$. The enhanced noise in the center corresponds to E_z close to zero.

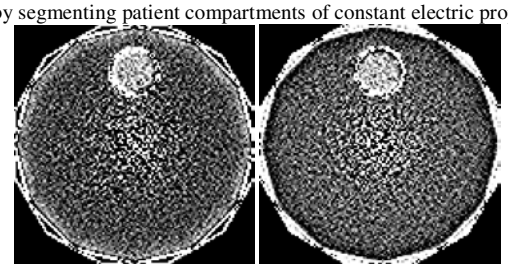

Fig. 2: Simulation results for reconstructing electric conductivity (left) and permittivity (right) as in Fig. 1. However, only the component $\underline{H}^+$ has been taken into account, thus enabling the approach on a standard MR system. The mean error in the small compartment is roughly 10% for both $\underline{\sigma}$ and $\underline{\varepsilon}_r$.

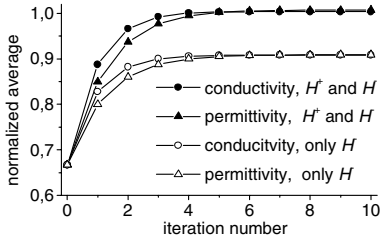

Fig. 3: Iteration of normalized average electric conductivity and permittivity in the small compartment of Fig. 1 (solid symbols) and Fig. 2 (open symbols). Neglecting $\underline{H}^-$ leads to an error of roughly 10%. The iteration started assuming homogeneous electric properties for the whole phantom.

References: [1] Saulnier GJ et al. *IEEE Sig. Proc. Mag.* 18 (2001) 31, [2] Gencer NG et al. *IEEE Trans on Biomed. Eng.* 43 (1996) 139, [3] Huang F et al. Proc. ISMRM 13 (2005) 683, [4] Joines WT et al. *Med. Phys.* 21 (1994) 54, [5] Collins CM et al. *MRM* 47 (2002) 1026, [6] CONCEPT II, Technical University Hamburg–Harburg, Department of Theoretical Electrical Engineering, [7] Press WH et al. *Numerical Recipies in C*. Cambridge University Press, 1995