

Evidence of tissue conductivity as a source of signal inhomogeneities in Ultrashort Echo Time (UTE) imaging

Ferdinand Schweser^{1,2}, Li Huang^{1,3}, Karl-Heinz Herrmann¹, Martin Krämer¹, Andreas Deistung¹, and Jürgen Rainer Reichenbach¹

¹Medical Physics Group, Institute of Diagnostic and Interventional Radiology I, Jena University Hospital - Friedrich Schiller University Jena, Jena, Germany, ²School of Medicine, Friedrich Schiller University Jena, Jena, Germany, ³Abbe School of Photonics, Faculty for Physics and Astronomy, Friedrich Schiller University Jena, Jena, Germany

TARGET AUDIENCE – Researchers interested in understanding the different biophysical contributions to the *Ultrashort echo time* (UTE) signal, in particular the relative contribution of B_1 -effects in magnitude imaging.

PURPOSE – UTE magnitude imaging allows direct visualization of tissues with short T_2 , such as tendons and cortical bone.¹ Despite the applied short echo times, a surprisingly high contrast was recently observed in UTE phase images of the meniscus.² Carl and Chiang² investigated the physical origin of UTE phase in simulations and phantom experiments. While the authors noted that UTE phase is affected by both B_1 - and B_0 -inhomogeneities their investigations, however, focused on B_0 -related effects concluding that UTE phase is strongly affected by these contributions. A thorough understanding of the different biophysical contributions to the UTE signal is important not only for interpretation of pathological signal alterations, but also for developing techniques to increase the signal homogeneity of the UTE images or for quantitative analyses of the images.²

In this contribution we investigated B_1^+ -contributions to UTE signal inhomogeneities with a dedicated phantom experiment to demonstrate that B_1 -related effects substantially affect the UTE signal. By applying *Electric Property Tomography* (EPT) to the complex-valued UTE signal we demonstrate that B_1 -effects due to the underlying tissue electrical conductivity are a major source of both magnitude and phase UTE signal inhomogeneity.

MATERIALS AND METHODS

Phantom: A phantom was created by placing three cylindrical tubes (diameter: 5 cm, approx. wall thickness: 100 μm) in a glass bowl (diameter: 24 cm) with tap water (Fig. 1). The tubes were filled with 1.5%, 3%, and 4.5% NaCl solutions (made from tap water), respectively, resulting in four different conductivity values in the physiological range but similar susceptibility values (negligible B_0 -effects).

Data Acquisition and Processing: UTE data were acquired on a 3T whole-body MRI scanner (Tim Trio, Siemens Medical Solutions, Erlangen, Germany) using a single-channel T_x/R_x birdcage coil and a radial 3D “spikey ball” center-out acquisition (TE = 100 μs , TR = 3.8 ms, RF pulse duration 10 μs , FA = 12°, 0.94 mm isotropic voxels, TA approx. 2 min). Magnitude and phase images were reconstructed by Fourier transforming the UTE data after state-of-the-art 3D gridding with iterative grid weights estimation⁶. Phase images were unwrapped with a spatial-domain best-path algorithm³. Since in this setup the UTE signal is supposed to contain similar contributions from B_1^+ and B_1^- , the square-root of the magnitude was calculated and the phase was divided by two, resulting in complex-valued images S . The complex-valued admittivity $\gamma = \sigma + i\omega\epsilon$ (permittivity ϵ and conductivity σ) was calculated according to Katscher et al.⁴: $\gamma(\vec{r}) = \nabla^2 S(\vec{r}) / i\omega\mu_0 S(\vec{r})$, where ω is the Larmor frequency and μ_0 is the permeability constant. This formula allows reconstructing from the complex signal absolute admittivity images within locally homogeneous regions.⁴ The Laplace was evaluated by parabolic fitting⁵ using the same fitting kernel for the whole dataset. To this end, the optimal kernel size was determined by successively reconstructing conductivity maps with different kernel sizes between 7 and 23 voxels and measuring the noise level (standard deviation) in the surrounding water.

Evaluation: The conductivity map was quantitatively analyzed by manually drawing ROIs in the center of the tubes and in the surrounding water, calculating mean and standard deviation within the ROIs, and correlating mean values with the known NaCl concentrations in the tubes. All data processing was carried out in MATLAB (2011b, The MathWorks, Natick, MA).

RESULTS – Signal inhomogeneities in the tubes and in the surrounding water were observed in both the magnitude (Fig. 2-left) and the phase (Fig. 2-right) images. Optimization of the fitting kernel yielded an optimal size of 15 voxels (see Fig. 3-left, arrow). The calculated conductivity map (Fig. 3-middle) demonstrated homogeneous contrast in the surrounding tap water and increased values in the tubes, though with some remaining inhomogeneities. Quantitative analysis revealed a linear relationship (0.932 S/m · c - 0.093 S/m; R=0.996, p=0.004) between the conductivity values and the NaCl concentrations (Fig. 3-right).

DISCUSSION – The linear relation between NaCl concentration and conductivity (Fig. 3-right) as well as the reasonable qualitative appearance of the reconstructed conductivity map (Fig. 3-middle) represent clear evidence of conductivity as a substantial source of contrast in both UTE magnitude images and UTE phase images (Fig. 2). Inhomogeneities in the reconstructed conductivity map are likely caused by vibrations of the liquid during signal acquisition; the use of agarose gel in future experiments may reduce these effects. More research is required to elucidate the relative contributions of admittivity- (B_1) and susceptibility-related (B_0) effects on the UTE signal. This may be achieved in future experiments by adding a paramagnetic substance such as Gd to the NaCl solutions.

CONCLUSION – Signal inhomogeneities in UTE images are substantially affected by B_1^+ phase contributions due to the underlying tissue conductivity distribution. Disentangling the different contributions to the UTE signal, similar as in a recent work on GRE phase contrast by Kim et al.⁷, may in future allow high-resolution EPT with acceptable measurement time.

REFERENCES – [1] Gatehouse PD and Bydder GM, 2003. *Clin Radiol.* 58:1-19. [2] Carl M and Chiang J-TA, 2012. *Magn Reson Med.* 67(4):991-1003. [3] Abdul-Rahman H et al., 2007. *Appl Opt.* 46(26):6623-35. [4] Katscher U et al., 2009. *IEEE Trans Med Imaging.* 28(9):1365-74. [5] Katscher U et al., 2012. *ISMRM.* #3482. [6] Zwart NR et al., 2012. *Magn Reson Med.* 67(3):701-10. [7] Kim et al. 2012. *ISMRM.* #3464.

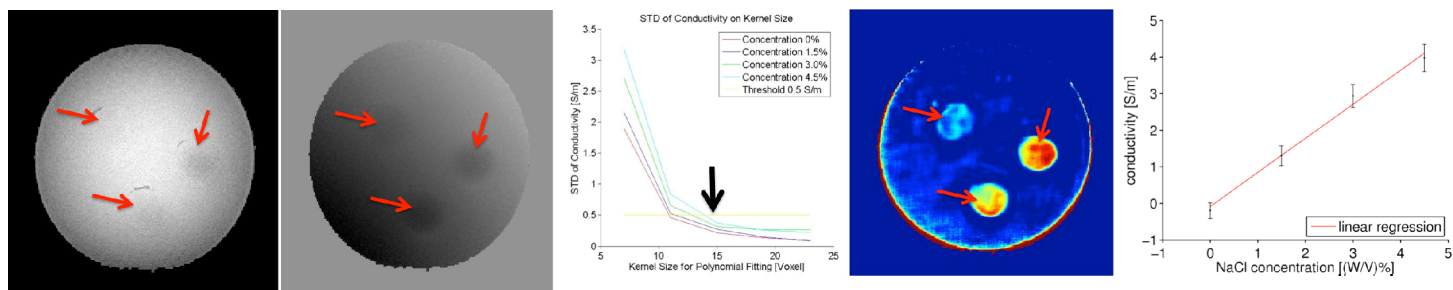


FIGURE 2. Coronal slice of the UTE magnitude (left) and **FIGURE 3.** Left: Optimization of the fitting kernel size. The threshold standard deviation was set to 0.5 unwrapped UTE phase (right). Note the signal inhomogeneities S/m (yellow line). **Middle:** Conductivity map calculated from the complex-valued UTE images. The in both magnitude and phase at the locations of the tubes color coding ranges from blue (0 S/m) to red (5 S/m). **Right:** Quantitative analysis of the conductivity map.