A Monte Carlo method for overcoming the edge artifacts in MRI-based electrical conductivity mapping

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TARGET AUDIENCE – Researchers interested in algorithmic improvements of MRI-based electrical conductivity mapping.

PURPOSE – Quantitative conductivity mapping (QCM)1-2 is a non-invasive MRI-based technique to assess tissue electrical conductivity. Tissue conductivity has been shown to linearly depend on tissue sodium concentration, opening the door for clinical applications like tissue lesion delineation3. It provides important input for improving computations of the local specific RF energy absorption rate2. QCM is based on the spatial variation of the phase of the complex-valued B0 field, which may be acquired by using, e.g., 3D ultra-short echo time (UTE) sequences4,5. One major limitation of published algorithms for conductivity reconstruction that has hampered clinical application of QCM so far is the presence of partly severe artifacts near tissue interfaces (edge artifacts)6,7, which result from the assumption of locally piecewise constant conductivity in the Helmholtz equation8. In this contribution, we present a spatial-domain inversion algorithm for conductivity reconstruction that substantially reduces edge artifacts. The algorithm is demonstrated both in a dedicated phantom and in vivo experiment.

THEORY – The conductivity reconstruction formula under the phase-only assumption9 is \( \sigma = \frac{\Delta \phi}{\Delta B_0} \), which indicates that the basis of conductivity reconstruction is the calculation of the local Laplacian of the phase \( \phi \). One noise-robust implementation is to apply a Gaussian-filtered Laplacian to the phase maps1. Alternatively, local quadratic fitting is used, but the resulting conductivity maps are less smooth. In this contribution, the Gaussian-filtered Laplacian (quadratic fitting method) is used throughout the data. edge artifacts occur at tissue interfaces where the kernel affects tissues with different conductivity. One option is to reduce the size of the fitting kernel near tissue interfaces to reduce edge artifacts5. However, small fitting kernels generally result in excessive noise. To overcome this limitation, we propose to use individual fitting kernels for each voxel. We constructed the kernels as follows: First, identification of tissue interfaces is achieved based on edges in magnitude images using a 3D image edge detection algorithm9. Then, in each voxel a Monte-Carlo-based region growing is applied to determine the local region of homogeneous tissue. The resulting fitting kernels have arbitrary shapes and contain a sufficient number of voxels for noise suppression.

METHODS – (1) Phantom construction (Fig. 1a): Six cylindrical plastic bottles (diameter: 5 cm, wall thickness: 0.5 mm, height: 7 cm) containing different saline solutions (red label on bottle) were placed in a uniform concentration of Gd-DTPA (300 mg/mL), thus opening the door for quantitative conductivity mapping. The bottles were filled with water and saline solutions, with a concentration of 2.5% (mass fraction) to reduce vibration artifacts during scanning. To investigate the effect of magnetic susceptibility and T1 sensitivity on the UTE-based measurements of B0, the bottom row of bottles contained in addition 1 mol/L of Gd-DTPA (Magnevist, Schering-Plough, Germany), resulting in a susceptibility difference of 0.34 ppm (which corresponds to the order of magnitude of magnetic susceptibility contrast in human brain)9 (black labels in Fig. 1a). (2) Volunteer: A healthy 60-year-old male was examined. The institutional review board had approved the study before and written informed consent was obtained from the volunteer. (3) Data acquisition: An isotropic 3D UTE sequence with a radial “spiky ball” center-out acquisition10 was applied on a whole-body 3T MRI scanner (Tim Trio, Siemens Medical Solutions, Erlangen, Germany) with a single-channel T/R, double-tuned birdcage coil (1H/31P head coil, Rapid Biomedical, Germany). Parameters for the phantom study: TE 110 µs, TR 175 ms, FA 7°, voxel size 1.125 mm3, 80492 spokes, 2 averages, TA 584 s. Parameters for the volunteer study: TE 110 µs, TR 3.5 ms, FA 7°, voxel size 1.125 mm3, 80492 spokes, 2 averages, TA 584 s. (4) Computing server: A standard PC with a CPU of Intel® Xeon® E5520 (2.27 GHz, 8 cores) and 24 GB RAM was used. (5) Data processing: UTE phase images were reconstructed from raw MR data11, unwrapped12 and divided by two according to the transceiver phase assumption13. (6) Analysis: Edge artifacts in conductivity maps reconstructed by the Gaussian-filtered Laplacian algorithm1 and the novel Monte Carlo method were compared. Mean conductivities and standard deviations (STD) were calculated in the bottles and background of the phantom (central slices) and in white matter of the brain and compared to previously reported conductivity values (model of saline conductivity12 and study of tissue dielectric properties14).

RESULTS – (1) Phantom experiment: Compared to the conductivity maps reconstructed with the Gaussian-filtered Laplacian algorithm1 (duration: 1 min, Fig. 1b), the conductivity maps reconstructed with the novel Monte Carlo method (duration: 25 min, Fig. 1c) showed substantially decreased edge artifacts near the bottles’ walls without sacrificing homogeneity in the bottles. The conductivities of the bottles with and without Gd-DTPA are plotted in Fig. 1d (red and blue dots with error bars for Gaussian-filtered Laplacian calculus (b) and the novel Monte Carlo-based method (c)). Conductivity values versus saline concentration in the phantom (d).


Figure 1. Methods and results of the phantom experiment. Photo of phantom (a). Conductivity maps of the phantom calculated with Gaussian-filtered Laplacian calculus (b) and the novel Monte-Carlo-based method (c). Conductivity versus saline concentration in the phantom (d).

Figure 2. Conductivity maps of the brain calculated with Gaussian-filtered Laplacian calculus (a) and the novel proposed Monte Carlo method (b).