Effects of phase alternations in nonlinear inverse T2 reconstructions from undersampled data

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Introduction

Quantitative evaluation of the T2 relaxation time is of high importance for diagnostic MRI. Standard T2 mapping procedures rely on the time-demanding acquisition of several fully-sampled k-space data sets. We recently evaluated a new method that allows for the reconstruction of spin-density and T2 maps from highly undersampled Cartesian data by exploiting data redundancy in parameter space [1]. The approach works without parallel imaging and is therefore of particular interest for animal MRI studies with a limited number of coil-array elements available. However, unavoidable motion in in-vivo experiments can cause periodic phase alterations in the acquired spin echoes [2], particularly at high magnetic field strength. The effect appears most dominantly in the inferior cerebral veins (**Fig.1, green arrow**) and can cause artifacts in reconstructions from undersampled data. This work explains the origin of these artifacts and demonstrates the impact on reconstructions from undersampled high-field animal MRI data.

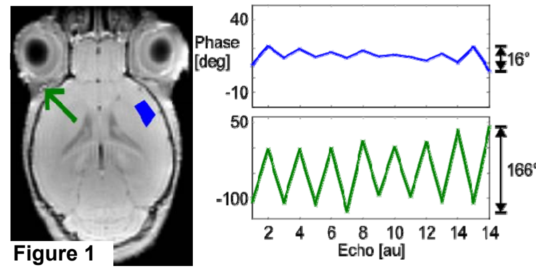

Figure 1

Materials and Methods

A healthy female adult C57BL/6 mouse was anesthetized by isoflurane (1.75% in ambient air) via an endotracheal tube. A 3D data set was obtained using a standard MSME sequence with 16 echoes (TR = 1.2 s, echo spacing ΔTE = 6.72 ms, matrix 128 × 86 × 64, resolution 160 × 160 × 160 μm^3) at 9.4 T (Bruker BioSpin, Germany). Signal detection was performed by a 4-channel mouse coil array. An artificial second data set with a constant phase throughout the echo train has been created from the DFT of magnitude images multiplied with the phase image of the first echo.

A numerical phantom was defined on a discrete pixel-grid in image-space, offering a compartment with M = 1, T2 = 100 ms and an alternating phase of φ = ± 50° throughout the subsequent echo images. Simulated noiseless single coil k-space samples were derived from the DFT of the pixels. The data comprised 16 echoes; echo spacing ΔTE = 6.72 ms; matrix 128 × 128.

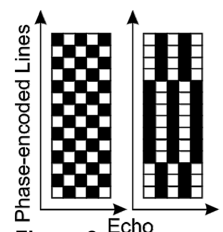

Figure 2

Spin-density and T2 map reconstructions of the different data were performed using the method described in [1] with the extended cost function:

$$\Phi(x) = \frac{1}{2}\sum_{c}\sum_{TE}\left\| P\,\mathrm{DFT}\,(M\, e^{-TE/T2+i\varphi}C_c) - s_{TE,c}\right\|_2^2 . \quad (1)$$

Here, M and T2 are the unknown parameter maps, φ the unknown image-phase map, P the sampling pattern and $s_{TE,c}$ the k-space samples. Interleaved and blocked undersampling schemes (**Fig.2 (left)** and **(right)**) were compared by selecting respective k-space lines from the fully-sampled data. The coil profiles C_c for the in-vivo data were calculated in a pre-processing step using the method [3] on the sum of the k-space samples of all echoes. For phantom studies, C was set to 1.

Figure 3

Results and Discussion

Fig.3 (left) shows the reconstructed spin-density map from fully-sampled simulated data. The results are similar to the reconstruction of simulated samples from the DFT of the real part of the original images (not shown). As the alternating phase cannot be reproduced by the model, the optimization terminates with residual energy in the cost function, i.e. $\Phi_{end} > 0$.

Fig.3 (center) shows a phantom reconstruction for interleaved undersampling. The even acceleration factor (AF) causes all odd k-space lines to be sampled from echoes with negative compartment phase and all even lines from echoes with positive compartment phase. As a consequence, all available samples can be perfectly modeled by the DFT of a mono-exponentially decaying object (M = 1, T = 100 ms) with the odd and even k-space lines multiplied by the respective phase values. The according image-space representation yields the original object convolved with the inverse FT of the multiplicative k-space phase pattern which causes the appearance of C' in **Fig.3 (center)**. As C' does not interfere with other image content, the result poses a numerically perfect solution with a final cost-function value of $\Phi_{end} \approx 0$. The ideal agreement of an artifact afflicted result with the samples can be precluded with an odd AF (not shown). However, for strong phase deviations (Δφ >> 20°), the optimization still tends to yield artifacts in the solutions that support the minimization of the residual energy from the first echoes. Similar observations can be made with the blocked sampling scheme, even though the impact of the artifacts is considerably reduced due to the sinc-shaped PSF of the pattern (**Fig.3 (right)**).

The simulations highlight the general effect of phase alternations on the reconstruction approach in [1] and explain how artifacts, shaped by the PSF of the sampling pattern, support the minimization of Eqn.1. Objects in true MR images are often embedded in a non-void surrounding and the multiplicative phase in k-space is neither scalar nor alternating with constant amplitude. Still, for undersampled data, regions with strong phase alternation as in **Fig.1 (green arrow)** provoke artifact in the reconstruction (**Fig.4 (center)**) that usually solve the cost function better than the desired visual optimum from fully sampled data (**Fig.4 (left)**). The artifacts disappear when reconstructing from artificial samples with constant image-space phase throughout the echo train (**Fig.4 (right)**).

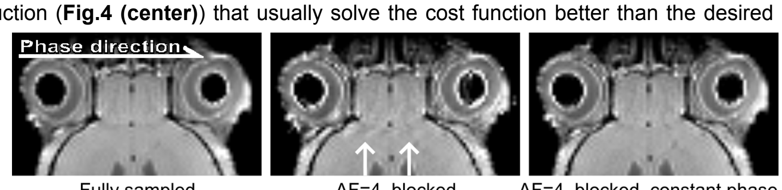

Figure 4

References [1] Sumpf TJ et al, JMRI 2011;34:420-428; [2] Wendt RE et al, MRM 1985;2:527-533; [3] Uecker M et al, MRM 2008;60:674-682