Compressive B1$^+$ mapping: Towards faster transmit coil sensitivity mapping

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Introduction: B1$^+$ mapping is an important prerequisite for multi-element transmit applications. Several B1$^+$ mapping methods have been proposed in the past [1-3]. Although typically low resolution data are acquired, B1$^+$ mapping is still challenging with respect to scan time, because the measurements have to be performed for each individual coil [4]. In this work, we investigate the potential to accelerate the B1$^+$ mapping measurement by means of compressed sensing (CS) [5-6]. The basic feasibility of the proposed method is evaluated on simulations and in vivo data from a 3T 8-channel parallel transmit system.

Theory: Compressed sensing allows for accurate image reconstruction from incomplete k-space data [7]. A successful CS reconstruction requires appropriate sparsity representation, incoherent sampling and a sparsity promoting non-linear reconstruction. Here we consider these requirements specifically in the context of B1$^+$ mapping.

To perform B1$^+$ mapping a couple of images are measured for each TX coil element modifying the encoding of the actual local transmit B1$^+$ field in the individual images. It is important to note that the images obtained for the different TX coils contain the same magnetization distribution, weighted by the corresponding transmit sensitivity. Thus, they share a common sparse support. The joint sparsity functional [8] accounts for these correlations. The images are reconstructed by solving the unconstrained optimization problem: $\min. \sum_c \left\| |\mathcal{F}_{uc} x_c - y_c| \right\|_2^2 + \lambda \sum_r \sqrt{\sum_c |\Psi\, x_c(r)|^2}$ (1), where x_c is the image for TX coil c, $\mathcal{F}_{uc}$ is the undersampled Fourier transform, y_c is the corresponding k-space data vector, Ψ is the sparsifying transform, and λ is a regularization parameter. A nonlinear conjugate gradients algorithm was used to solve (1).

Variable density quasi random undersampling provides high incoherence and preserves a large part of the signal energy. Since the data for each transmit coil are acquired separately, a different set of phase encoding lines could be acquired for each measurement, providing random sampling also in the coil dimension.

Methods: Simulations were performed to assess the quantitative error of the reconstruction method. The actual flip angle imaging sequence (AFI) [2] was used for B1$^+$ mapping. However, the reconstruction is independent of the actual B1$^+$ mapping method. AFI data were simulated using the Shepp-Logan phantom as magnetization distribution for an 8-element RF transmit array, TR1 = 20 ms, TR2 = 100 ms, matrix 128x128. Gaussian noise with $\sigma = 0.001$ was added, and the data were undersampled in the phase encoding direction. The images for each TR were reconstructed according to Eq. 1 using finite differences as a sparsifying transform, and B1 maps were obtained from the reconstructed images as described in [2].

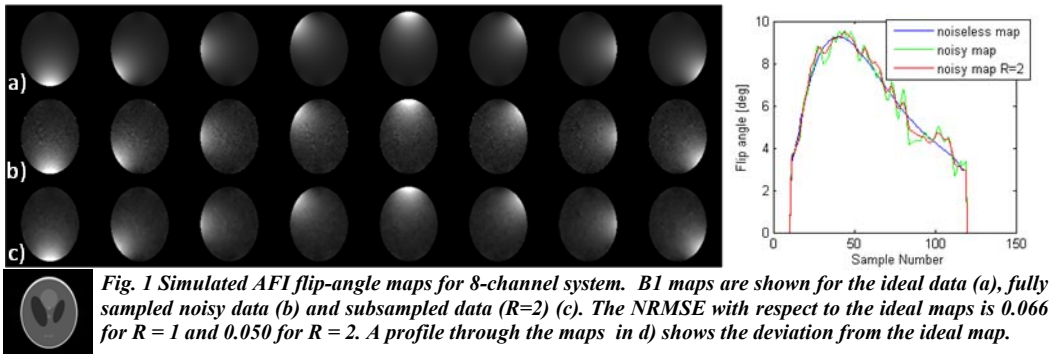

Fig. 1 Simulated AFI flip-angle maps for 8-channel system. B1 maps are shown for the ideal data (a), fully sampled noisy data (b) and subsampled data (R=2) (c). The NRMSE with respect to the ideal maps is 0.066 for R = 1 and 0.050 for R = 2. A profile through the maps in d) shows the deviation from the ideal map.

In vivo experiments were performed to acquire abdomen data on a 3T 8-channel body transmit system (Philips Medical Systems, Best, The Netherlands) with TR1 = 20 ms, TR2 = 100 ms, scan matrix 40x128. The data were undersampled with reduction factor of 2 (leaving only 20 phase encodings in each image).

Results: Fig.1 shows the ideal B1$^+$ maps simulated for the 8 channel coil array (a), maps reconstructed from noisy data with Nyquist sampling (b) and from undersampled data with acceleration factor of 2 (c). The accelerated maps show good correspondence to the true maps and slight denoising, typical for CS reconstruction. This could be seen better by looking at a single profile of the resulting maps (Fig.1 (d)). Fig. 2 shows the results of the in vivo experiments. The B1$^+$ maps obtained from the CS reconstructed images do not show significant differences from the fully sampled maps.

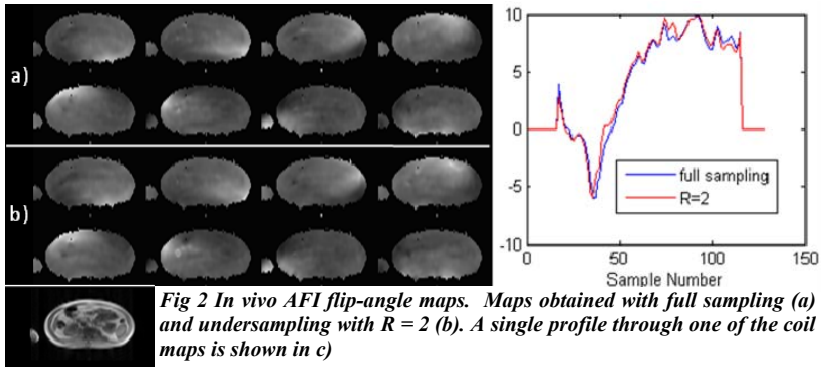

Fig 2 In vivo AFI flip-angle maps. Maps obtained with full sampling (a) and undersampling with R = 2 (b). A single profile through one of the coil maps is shown in c)

Conclusion: The proposed reconstruction method allows accurate B1$^+$ mapping with reduced acquisition time. In this setting, undersampling can be performed in several different dimensions: the phase encoding dimension, across different coils, and across the data with different TR, which improves the reconstruction robustness. Together with the joint sparsity constraint, this allowed a very good reconstruction from very small amount of data. In a further refinement, the reconstruction can be extended applying prior knowledge about the ratio between the images at different TR obtained from previous scans of the same patient or from different patients with similar size.

References:

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