Partial Fourier accelerated spiral SENSE imaging using magnetic field monitoring

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Introduction: Partial Fourier imaging is a common method to shorten the acquisition duration in Cartesian MRI. It is based on the assumption that the image phase is smooth in space and thus is contained in the central part of k-space. After subtraction of the image phase estimate, the k-space representation of the image is considered to be conjugate symmetric, thus allowing for asymmetric partial k-space acquisition, if the image phase is accurately known [1-6].

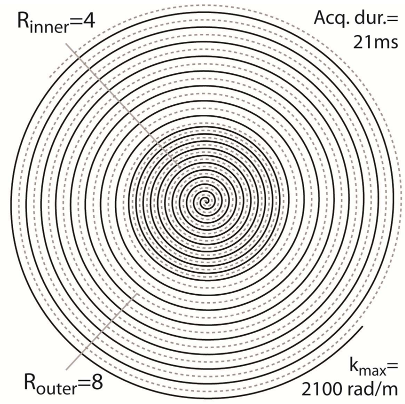

Fig.1: Monitored variable density spiral trajectory (solid); mirrored trajectory (dashed).

Spiral MRI is among the most efficient ways of covering k-space, which can be further increased using parallel imaging. A combination with partial Fourier phase constrained imaging could allow to further decrease the acquisition time and thus increase the achievable temporal resolution in dynamic MRI studies as well as robustness against motion and B_0 off-resonance effects. Since spiral imaging is very sensitive to encoding deficiencies such as gradient delays, eddy currents, concomitant fields and static B_0 inhomogeneity, a combination with partial Fourier phase constraints (where precise phase information is required) is hard to achieve in vivo.

Here we present a partial Fourier spiral MRI approach where an accurate encoding model of the dynamic field encoding is obtained by using dynamic field monitoring [7-9]. The performance and the sensitivity against inaccuracy of the phase estimate are evaluated in vivo.

Methods: Imaging was performed on a Philips 3T Achieva system using an 8-element head receive array. For all scans the dynamic field encoding information was acquired using a 2^{nd} order concurrent field monitoring setup [9] based on ^{19}F-NMR probes. Variable-density spiral gradient echo scans with 1 and 4 interleaves were acquired (TE=1.3ms, readout=21ms, k_{max}=2100rad/m, FOV=22cm). Each interleave (Fig. 1) being undersampled by a factor of R=4 and R=8 in the inner and outer k-space region respectively. The same slice was acquired with two Cartesian gradient echo scans (TE=3.0ms/3.6ms, resolution=1.8mm^2) from which B_1-coil-sensitivity maps and B_0-maps were determined.

The incorporation of phase constraints to parallel imaging is based on previously proposed iterative algebraic reconstruction approaches [10, 5, 6]. The image $\boldsymbol{\rho}$ is obtained by minimizing the objective function

$$\|\boldsymbol{\sigma} - \boldsymbol{E\rho}\|_2^2 + \lambda^2 \|\mathrm{Im}\{\boldsymbol{\rho}\}\|_2^2 \qquad (1)$$

where $\boldsymbol{\rho}$ is a vector of image pixel values, $\boldsymbol{\sigma}$ is a vector holding the signal samples and $\boldsymbol{E}$ is the encoding matrix defined as

$$\boldsymbol{E}_{(\kappa,\gamma),\nu} = s_\gamma(\boldsymbol{r}_\nu) e^{i\phi(\boldsymbol{r}_\nu, \boldsymbol{t}_\kappa)}$$

with the spatial coordinate $\boldsymbol{r}$, time t and coil sensitivity s counted by ν, κ and γ respectively. The phase evolution is described by $\phi(\boldsymbol{r}, t) = \phi_{B_0}(t) + k(t)\boldsymbol{r} + \Delta\omega(\boldsymbol{r})t + p(\boldsymbol{r})$,

where the B_0-phase ϕ_{B_0} and k-space coordinates k are obtained from magnetic field monitoring [8,9], $\Delta\omega$ is the static off-resonance obtained from the B_0-map and $p(\boldsymbol{r})$ is the phase estimate map. The penalization of the imaginary part in $\boldsymbol{\rho}$ (and thereby the residual phase) is controlled by the regularization factor λ (Eq. 1) [6] which was heuristically set to 0.5 for all phase constrained reconstructions.

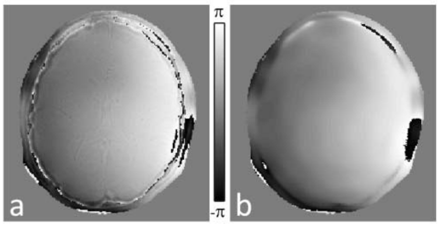

Fig.2: Phase estimate maps $\boldsymbol{p}$: (a) High resolution phase map from the reference scan. b: Low resolution map reconstructed from the inner k-space portion (single-shot acquisition)

As a reference, the 4-interleave scan (a) was reconstructed without the phase constraint ($\lambda = 0$). The single-shot images were reconstructed using a high resolution phase map (Fig.2a) taken from the reference scan (b), without PF constraint (c), using the low resolution phase map (Fig.2b) reconstructed from the central k-space portion of the variable density spiral (d), and the same while applying (masking) the phase constraint only in the central part of the image where a smooth behavior of the phase is expected (e).

Results and Discussion: As expected from the drastic undersampling (R_{outer}=8), the non-phase constrained image (Fig.3c) showed spiral fold-over artifacts, while the anatomy is faithfully reflected when incorporating the high resolution phase estimates (Fig.3b). Fig.3d,e show the phase constrained reconstructions based on the smooth phase estimate (Fig.2b). Artifacts due to imposing incorrect phase at the object border are visible in Fig.3d. The artifact is greatly reduced when applying the phase estimate only in the central regions (Fig.3e). Residual effects of the low resolution phase approximation are apparent as slight noise-like structures (Fig.3e).

Conclusion: The successful application of partial Fourier imaging to single-shot-spiral MRI was demonstrated in vivo for the first time. Using only 8 receive channels, 8-fold undersampled in-vivo images were reconstructed without notable aliasing. The apparent artifacts in the reconstructed images seem relatively sensitive to inaccurate phase estimates. However, given that the image phase is sufficiently smooth or can be obtained with sufficient resolution respectively, the method enables a significant decrease in readout time and thereby can increase the achievable temporal resolution in MRI.

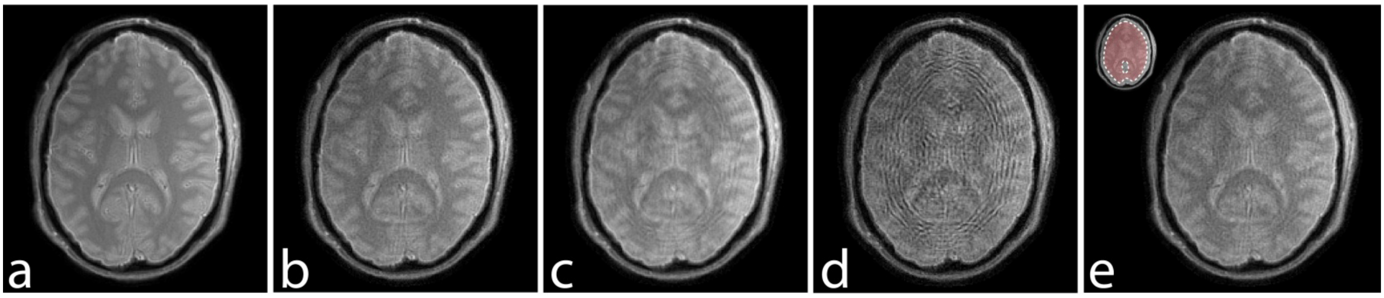

Fig.3: a: Non phase constrained reference image from 4-interleave scan (2-fold SNR). b-e: Single-short reconstruction: using the high resolution phase estimate(b), non phase constrained (c), phase constrained using the low resolution phase map in the entire image (d) and in the masked region (e).

References: [1] Margosian et al. Health Care Instrum. 1 (1986) 195–197; [2] Noll et al. IEEE TMI (1991) 154–163; [3] Cuppen et al. Topical Conf. Fast MRI Techniques, Cleveland, Ohio (1987); [4] Youla and Stark et al. IEEE TMI (1982) 81–101. [5] Willig-Onwuachi et al. JMR (2005) 176:187–198; [6] Bydder et al. MRM (2005) 53:1393–1401 [7] De Zanche et al., MRM 60:167:186. [8] Barmet et al., MRM 60:187–197. [9] Barmet et al. ISMRM2010,216, [10] Pruessmann et al. MRM 46:638–651 (2001)