k-Space trajectory mapping for ultra-short, single-shot, non-Cartesian imaging

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INTRODUCTION

In-vivo MR image encoding speed is physiologically limited by gradient-induced peripheral nerve stimulations [1]. Due to the linear nature of the magnetic gradient field, the acceptable gradient slew rate increases with decreasing subject dimensions. This makes high performance gradient insert coils an attractive choice for performing small-animal MR imaging and microscopy studies in a whole-body, clinical MR scanner [2]. Such enhanced performance also amplifies gradient imperfections due to Eddy currents, coupling effects, mechanical vibrations, etc. On the other hand, the smaller dimensions and higher gradient amplitude often result in only suboptimal gradient calibration results using standard service tools [3].

Recently, magnetic field sensors in the form of small NMR probes have been described as a highly accurate tool for spatiotemporal magnetic field mapping [4, 5]. In this work such magnetic field sensors were used in combination with a high-performance gradient insert coil for ultra-fast, single-shot, high-resolution, non-Cartesian imaging.

MATERIALS and METHODS

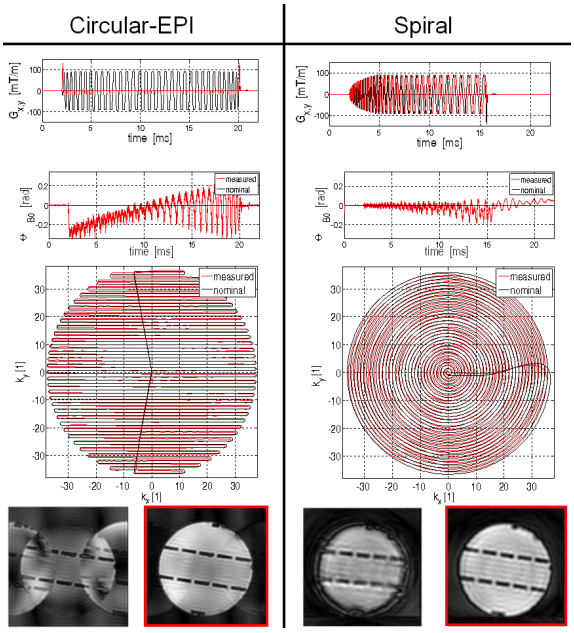
Spatiotemporal magnetic field mapping was achieved using susceptibility-matched, transmit-receive NMR probes as described in Ref. [5]. The probes consist of a ~1µl small H_2O signal droplet, closely surrounded by a solenoid coil for high SNR signal detection. Signal lifetimes on the order of 100ms could be achieved by applying susceptibility-matching methods in order to minimize static ΔB_0 variations across the water droplet. Using a distributed array of such NMR probes, the spatially-constant, B_0 phase variation $\Phi_{B0}(t)$, and the k-space trajectory $\mathbf{k}(t)$ can be extracted based on the Larmor relation. The measured k-space encoding information is then used in the subsequent image reconstruction, according to:

$$\varphi_{n^{th}\,\mathrm{Probe}}(t) - \Delta\omega_{n^{th}\,\mathrm{Probe}}\, t = \Phi_0(t) + \mathbf{k}(t)\,\mathbf{r}_{n^{th}\,\mathrm{Probe}}, \quad \text{with: } \Phi_0(t) = \gamma\int \Delta B_0(t')dt',\ k(t) = \gamma\int \mathbf{G}(t')dt' \quad \Rightarrow \quad \text{image}(\mathbf{r}) = \int \text{data}(\mathbf{k})\,\text{dcf}(\mathbf{k})\,e^{-i\Phi_0(t)}\,e^{-i\mathbf{k}(t)\mathbf{r}}\,d\mathbf{k}$$

with $\mathbf{r}_n$ and $\Delta\omega_n$ denoting the n^{th} probe spatial position and off-resonance, respectively. Both can be obtained from a single FID-type calibration measurement. The NMR probes were analyzed and found to be sensitive enough ($\Delta B_0 * \Delta t \sim 10^{-11}$ Ts) to capture magnetic field alterations, which when uncorrected would become apparent as noticeable image artifacts ($\Delta B_0 * \Delta t \sim 10^{-10}$ Ts). The above image encoding equation was solved using a gridding-based, conjugate-gradient (CG) algorithm with optional multi-frequency interpolation (MFI) off-resonance correction [6, 7].

The experiments were performed on a clinical 3T GE Signa HDx MR scanner (GE Healthcare, Milwaukee, WI, USA) equipped with a high-strength gradient insert coil (peak strength = 600 mT/m, peak slew rate = 3200 T/m/s) [2]. This gradient system was operated at a maximum amplitude of 500 mT/m with a slew rate of 1800 T/m/s to minimize vibration and acoustic problems. The insert gradient coil was designed to be driven by the body coil gradient amplifiers. A hardware switch enables the switching of amplifier output between the two gradient systems.

Non-Cartesian, single-shot, circular-EPI [8] and constant radial density spiral k-space trajectories were designed for the following imaging parameters: FOV = 6.4 cm, BW = ±125kHz, ACQ mtx = 64x64, Recon mtx = 256x256. The resulting readout time (t_{ACQ}) and the maximum gradient strength and slew rate were found to be t_{ACQ} = 18.2 ms, G_{max} = 92 mT/m, S_{max} = 1750 T/m/s for circular EPI, and t_{ACQ} = 13.7 ms, G_{max} = 92 mT/m, S_{max} = 1800 T/m/s for the spiral, respectively (cf. figure).



RESULTS

The left column of the figure shows k-space trajectory mapping and imaging results for the single-shot, circular EPI acquisition. Performing image reconstruction based on the calibrated k-space trajectory resulted in significantly reduced ghosting artifacts compared to the reference image obtained using the nominal prescribed gradient waveforms. The right column shows the same comparison for the single-shot spiral acquisition. MFI deblurring was applied based on separately acquired off-resonance maps. Also in this case the image quality was significantly improved using the measured k-space trajectory. The reproducibility of the gradient insert system was tested by multiple repetitions of the k-space trajectory measurement, with the gradient insert removed in between. The obtained k-space trajectories resulted in no significant difference in the reconstructed images, verifying the assumption of time stability for the gradient insert system. An initial attempt to apply an advanced frequency-domain, linear-system gradient calibration approach failed, however, calling for further investigations.

DISCUSSION and CONCLUSIONS

In this work, ultra-fast, high-resolution, single-shot, non-Cartesian imaging was demonstrated using a high performance gradient insert coil. This was enabled using an NMR-probe based k-space trajectory calibration method. In comparison to alternative k-space trajectory methods [9, 10], susceptibility-matched, transmit-receive NMR probes provide important advantage like high SNR, and fast single-shot calibration of two gradient axes at once. The method is expected to be particularly valuable for applications, which either perform advanced encoding (single-shot acquisitions, spectral-spatial Non-Cartesian imaging, microscopy, etc.) or are particularly susceptible to gradient imperfections (Diffusion, Phase Contrast, etc.).

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REFERENCES

[1] Reilly et al, *Med Biol Eng Comput*, vol. 27, 101-10, 1989. [2] Chronik et al, *Magma*, vol. 10, 131-46, 2000. [3] Glover et al, *U.S. Patent 4,591,789*, 1986. [4] De Zanche et al, *Magn Reson Med*, vol. 60, 176-86, 2008. [5] Sipilä et al, 16th ISMRM Toronto, p680, 2008. [6] Barmet et al, 13th ISMRM Miami, p682, 2005. [7] Schulte et al, 15th ISMRM, Berlin, p 1741, 2007. [8] Kerr et al, *Magn Reson Med*, vol. 38, 355-67, 1997. [9] Takahashi et al, *Magn Reson Med*, vol. 34, 446-56, 1995. [10] Duyn et al, *J Magn Reson*, vol. 132, pp. 150-3, 1998.