

Exploration of TRASE MRI at Low Magnetic Field: Potential Performance and Limitations

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TRansmit Array Spatial Encoding (TRASE) MRI uses trains of radiofrequency (RF) π -pulses produced by "phase-gradient" coils which generate transverse B1 fields of uniform magnitude and spatially varying directions.¹⁻³ If coils A and B are two such coils – with phase-gradients characterized by vectors k_A and k_B – the building block of TRASE MRI sequences is the pair of π -pulses $[P_A P_B]$ which shifts transverse magnetisation by an amount $2(k_A - k_B)$ in k-space and has the same action as a gradient pulse of the static field B_0 . We are currently exploring the benefits and limitations of TRASE MRI at low field. In the mT field range, hyperpolarisation can be employed to obtain high-SNR images but the adverse effect of concomitant B_0 gradients puts a limit to the resolution achievable in standard MRI: this issue is automatically solved in TRASE. However, to accommodate for changes in B1-field orientation, concomitant RF field components necessarily exist. Moreover, oscillating RF fields are used, consisting of equal rotating and counter-rotating (CR) components. In contrast with high-field NMR for which the concomitant and CR components simply contribute to RF power deposition, both components may influence time evolution in low field since B1 magnitudes are not much smaller than B_0 . We perform experiments in a 15-cm bore resistive magnet (B_0 : 1–5 mT), equipped with a usual (uniform-field) RF coil, standard B_0 -gradient coils, and additional phase-gradient coils (Fig. 1a).⁴ The efficiency of RF phase-encoding is compared to that of encoding with B_0 -gradient pulses both in 1D and 2D imaging of water samples and sealed cells of laser-polarised ^3He gas. The potentially adverse effects of the RF concomitant and CR fields are systematically explored theoretically, numerically, and experimentally.^{5,6} In particular, their joint effect can lead to significant image artefacts when short π -pulses are used (Fig. 1b). We have designed robust pulse sequences which are fairly immune to such effects. Generalisation of these studies to 2D and 3D gradient-free MRI is under way.

[1;2] JC Sharp et al., Magn Reson Med 63:151 (2010); NMR Biomed, 26: 1602 (2013)

[3] Q Deng et al., Magn Reson Imaging 31:891 (2013)

[4] P-J Nacher et al., Proc ESMRMB15, Magn. Reson. Mater. Phys. 28 Suppl. 1 (2015) p. S64

[5;6] CP Bidinosti et al., Proc. ISMRM 2018 abstract 4309; abstract 8814

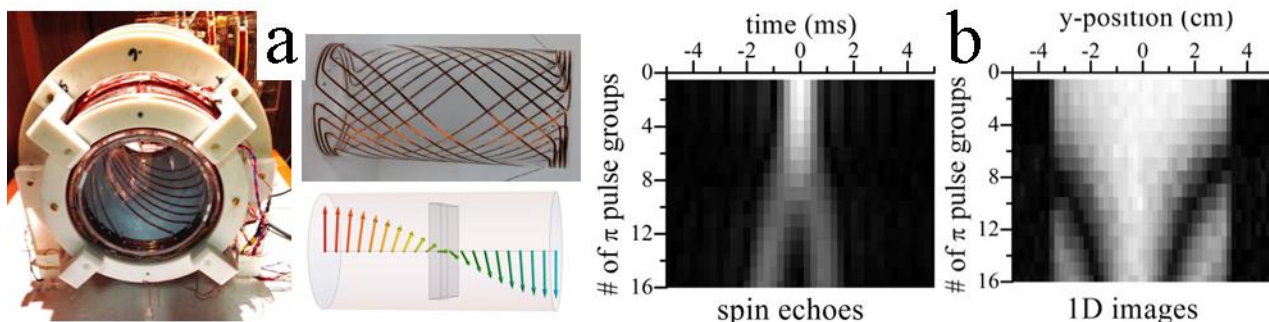


Fig1 a: low-field MRI system equipped with a spiral B1 coil^{4,6} (coil S, B1 pitch $k_s=0.12 \text{ rad cm}^{-1}$ along z). A standard B1 coil, U, (not shown) surrounded coil S and generated a uniform field ($k_U=0$). b: Results of TRASE phase-encoding steps, chained groups of π -pulses $[P_S P_U]_x [P_S P_U]_{-x}$, followed by refocussing B_0 -gradient pulses G_z of nominally opposite effect. Experiments were performed at 83.682 kHz ($N_R=300$) on doped water (a $7 \times 7 \times 1 \text{ cm}^3$ slab in the $z=0$ plane) using rectangular P_s pulses lasting 6 RF periods. Standard 1D spin-echo projection imaging was performed using B_0 -gradient pulses in the y -direction of the expected concomitant RF field gradient. In the time-domain, the echoes (the horizontal lines in the composite image) were strongly distorted and attenuated for more than 6-8 pulse groups. The corresponding 1D composite projection image displayed severe artefacts.