Phase Encoding without Gradients at High Field: TRASE RF MRI at 3T

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Introduction

The development of fast MRI methods has focused on new efficient single-shot k-space trajectories or parallel acquisition of undersampled k-space trajectories. Recently, a new RF B1-field method of spatial encoding was introduced whereby k-space is traversed in the phase encoding direction without using B_0 magnetic field gradients, but by applying different B_1-phase gradient fields produced by a Tx-array, TRansmit Array Spatial Encoding (TRASE) [1,2]. This simple method has been shown to work very well at 0.2T for both gradient-free 2D MRI [3] as well as gradient-free slice selection [4]. Here we demonstrate our first step at TRASE MR imaging in more inhomogeneous B_1 fields at 3T using both a switched 1- and 8-channel transmitter system, gradient-free encoding in one dimension with a two-channel spiral birdcage array B1-fields.

Theory

If an array of Tx-elements are driven to produce a B1-phase gradient along a particular direction of the form, $\mathbf{B_1} = |\mathbf{B_1}| e^{i\phi_1(\mathbf{r})}$ where $\phi_1(\mathbf{r}) = 2\pi \mathbf{k_1} \cdot \mathbf{r} = \mathbf{G_1} \cdot \mathbf{r}$, then the magnetization after applying a refocusing 180° RF pulse with this B1 phase gradient is $\mathbf{M}_+ = \mathbf{M}_-^* e^{i2\phi(\mathbf{r})} = \mathbf{M}_-^* e^{i2\pi[2\mathbf{k_1}.\mathbf{r}]}$, so that $\mathbf{k}_+ = -\mathbf{k}_- + 2\mathbf{k_1}$. The key is to then apply a pulse sequence consisting of excitation followed by an alternating train of refocusing pulses: $90_A - 180_A - 180_B - 180_A - 180_B - \ldots$, using a Tx-array that can produce two different phase gradients $G_{1A}(r)$ and $G_{1B}(r)$, with respective k-space origins k_{1A} and k_{1B}, then the k-space coordinate (after each 180° pulse) as observed by field B_{1A} accumulates as: $(0, -\Delta k_{BA}, +\Delta k_{BA}, -2\Delta k_{BA}, +2\Delta k_{BA}, \ldots)$, where $\Delta k_{BA} = 2(k_{1B} - k_{1A})$. (Note that in reception, a phase gradient coil receives the signal with an additional k-space shift of $-k_1$). Therefore the key to high resolution TRASE MR imaging is to repeatedly refocus using alternating phase gradients, allowing the signal phase or k-space trajectory to accumulate to higher k-space states. Then the NMR signal becomes spatially encoded by the transmit B1-field. The field-of-view which can be encoded without aliasing, in this single-shot, single receive field experiment, is given by $FOV_{shot} = 1/\Delta k_{BA}$, and is equivalent to the distance over which $\phi_A(r) - \phi_B(r)$ is equal to π. Spatial resolution is then FOV_{shot}/N.

Methods/Results

A +π and -π spiral Birdcage array was constructed at 123.2 MHz, 29 cm long and 28 cm in diameter, for a FOV_{shot} (where $\Delta\phi = \pi$) of about 20cm (*Fig.1a*) mapped on the bench using a small search probe and a network analyzer (Agilent). The two channel Tx-array was configured to operate using a single transmitter system with scanner TTL signals used to control a two-way power director, or configured to operate without this extra Tx-switching, using two separate Tx-channels of the Siemens prototype 8-channel transmitter on a 3T TRIO Tim MRI system. Images were collected using a single shot Turbo Spin Echo (TSE) sequence of 64 echoes (4 mm Coronal slice, 1 Average, 20cm FOV, TE=11ms, S-I Phase Encoding), with a 90° SINC slice selective excitation and Rectangular 180° refocusing, first with normal phase encoding using a single spiral birdcage coil: $90_A - 180_A - 180_A - 180_A - 180_A \ldots$, and secondly for TRASE a 64 echo acquisition was acquired without phase encoding gradients, by applying Tx-power sequentially to the two spiral birdcage arrays: $90_A - 180_A - 180_B - 180_B - 180_B \ldots$, followed by standard unfiltered, uncorrected 2DFT reconstruction. Images were collected of a three-compartment saline phantom (11cm S-I x 8 cm) and a formalin-fixed brain (12cm S-I x 14cm).

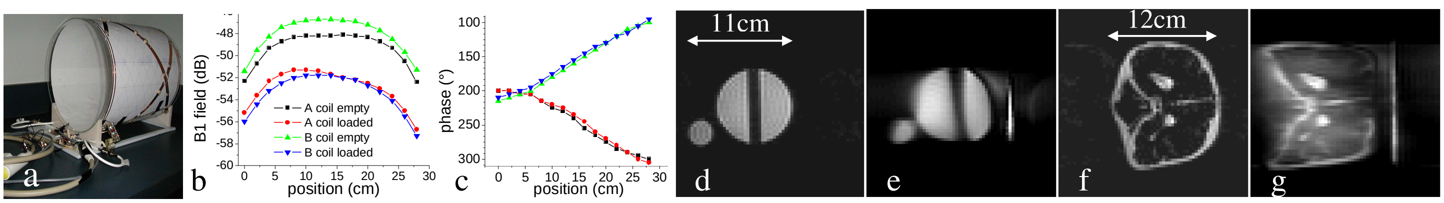

Fig. 1: (a) Constructed +π and -π two channel array. (b) B_1 magnitude maps of two coils. (c) B_1 phase maps show a FOV_{shot} (where $\Delta\phi = \pi$) about 20cm. Coronal images of the phantom (d,e) and of a formalin fixed brain (f,g) acquired using: (d,f) Standard 1-shot 2D Turbo SE, TE=11ms, 64 echo, 20cm x 20cm FOV using +π coil only; (e,g) same but without phase encode gradients with single shot, 64 echo, S-I (left-to-right in image) TRASE encoding $90_A - 180_A - 180_B - 180_A - 180_B \ldots$ by alternating between the two spiral birdcage coils during transmission.

Discussion/Conclusions

Using only two B_1 phase gradients, TRASE produced images (*Fig.1e*) comparable to those acquired using standard gradient encoding (*Fig. 1d*). The brain phantom results (*Fig. 1g*) are also good, but due to the larger phantom size, TRASE MRI is susceptible to B_1 inhomogeneity artifact. The on-axis phase gradients were very linear (*Fig. 1c*), but the magnitude of the phase gradient off-axis is stronger, leading to some gradient non-linearity image distortion seen as a smaller effective FOV away from the isocenter axis of the coil in the brain image (*Fig. 1g*). Although the FOV_{shot} for the 2-channel phase gradient array was about 20cm, the less than optimal magnitude uniformity (*Fig.1b*) results in a loss of spatial encoding at spatial locations where the refocusing pulse flip angle is far from 180°, and hence the corresponding encoding artifact [5] is seen as a bright line in the image. At this high field strength, B_1-shimming could improve TRASE image quality, although this can be done using a single Tx system, as we move to 3D TRASE MRI, all the Tx-switching will also become complicated, and the multi-Tx method may prove simpler. Future research will include optimizing for B_1 homogeneity by B_1 Shimming and composite RF pulses and a full 3D TRASE 3T implementation including 2D encoding and slice selection using a multi-Transmitter. Once perfected, this new silent k-space traversal method will allow the exploration of new data acquisition schemes that may reduce gradient related artifacts/deficiencies and improve imaging speed.

References: [1] S.B. King, D. Yin, S. Thingvold, J.C. Sharp, B. Tomanek, Proc. ISMRM, p.2628, 2006, [2] S.B. King, P. Latta, V. Volotovskyy, J.C. Sharp, B. Tomanek, Proc. ISMRM, p.680, 2007, [3] J.C. Sharp, S.B. King, D. Yin, V. Volotovskyy, B. Tomanek, Proc. ISMRM, p.829, 2008, [4] J.C. Sharp, S.B. King, D. Yin, V. Volotovskyy, B. Tomanek,, Proc. ISMRM, p.1083, 2008, [5] J.C. Sharp, S.B. King, D. Yin, P. Latta, B. Tomanek, Proc. ISMRM, 2009 (submitted) .