Non-linear TRASE

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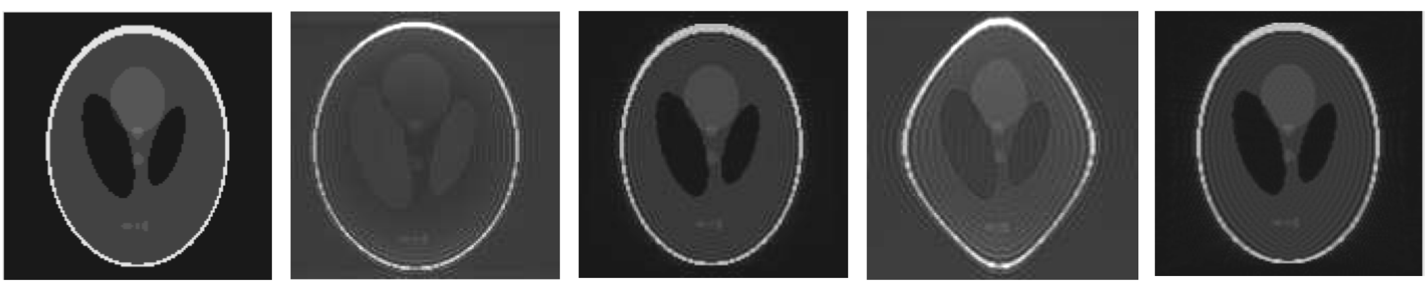
Target Audience: Those interested in developing "gradient-free" MRI for portability and accessibility.

Purpose: The TRansmit Array Spatial Encoding (TRASE) method encodes spatial information through the use of spatially varying B_1 transmit phase fields instead of using traditional B_0 gradient coil systems [1]. An ideal TRASE coil array creates B_1 fields with linearly-varying spatial phases. The imaging data generated with an ideal array may be reconstructed via a Fourier transform similar to the reconstruction of data acquired using B_0 gradients. Existing TRASE array coils create linear B_1 phase variations only over a limited spatial region after some optimization of coil geometry [2]. The design and manufacture of TRASE RF coils would be simplified if it were not necessary for the phase of the transmitted B_1 fields to vary linearly. Here we explore the feasibility of the reconstruction of non-linear TRASE signals from a mathematical perspective.

Theory/Methods: TRASE signals were simulated for a discrete Shepp and Logan mathematical phantom inside two different RF coil configurations: (a) a parallel conductor coil set that follows an optimized TRASE coil as designed by Deng et al. [2] and (b) a circular coil set composed of a combination of circular Maxwell and Helmholtz coils. The simulated signals were reconstructed using a Fourier transform and using a least squares reconstruction [3]. The detected signal data for 2D TRASE imaging was modelled as $S_L(\rho) = \sum_i \sum_j \rho(x_i, y_j) e^{i\phi_L(x_i,y_j)}$, where $\phi_L(x_i, y_j)$ is the accumulated encoding phase associated with the L^{th} data point. When the phase gradient is linear, the encoding phase is related to k-space by $\phi_{L(p,q)}(x, y) = p\Delta k_x x + q\Delta k_y y$. The phantom signal may be decomposed using the signal of image basis functions as $S_L(\rho) = \sum_m \sum_n \rho(x_m, y_n) S_L(E_{(x_m,y_n)})$ where $S_L(E_{(x_m,y_n)})$ is simulated signal of image basis function which explicitly is $S_L(E_{(x_m,y_n)}) = \sum_i \sum_j E_{(x_m,y_n)}(x_i, y_j) e^{i\phi_L(x_i,y_j)}$ and $E_{(x_m,y_n)}$ is the image basis function defined as: $E_{(x_m,y_n)}(x,y) = 1$ if (x,y) is in the pixel centered on (x_m,y_n) and 0 otherwise. The signal from the phantom ρ may be written in matrix form, with appropriate re-indexing of i and j, as $S_L = [T]\,\rho$ where $[T]$ is the matrix of simulated signals of the image basis functions. The image can be reconstructed using least squares Explicitly, if $P\rho$ denotes the reconstructed image then $P\rho = [[T]^T[T]]^{-1}[T]S_L$ gives the least squares reconstruction.



(a) (b) (c) (d) (e)

Fig. 1: 128 × 128 pixel phantom and reconstructions. (a) Phantom. (b), (c) Reconstructed images from a parallel conductor coil set using DFT and least squares respectively. (d), (e) Reconstructed images from a circular coil set using DFT and least squares respectively.

Results/Discussion: Fourier reconstruction of simulated TRASE signals from the optimized TRASE coil set showed minor distortions over the reconstruction of signals from the ideal coil set but the Fourier reconstruction of signals from the circular coil set produced severe geometric distortion. Least squares inversion of both cases produced reconstruction errors comparable to Fourier reconstruction from the ideal coil set simulated signal.

Conclusion: Least squares reconstruction of data from a non-ideal circular TRASE coil set is able to produce results comparable to the DFT reconstruction of data from an optimized TRASE coil set. Therefore TRASE coil design complexity or imperfect construction may be traded off against more complex image reconstruction to reduce hardware cost.

References: [1] J.C. Sharp, S.B King. Magn Reson Med 63:151-161(2010). [2] Q. Deng et al. Magn Reson Imaging 31:891-899(2013). [3] R. Penrose. Math Proc Cambridge Phil Soc 51:406-413(1955).