Non-linear encoding gradient optimization for O-space imaging with a microstrip coil array

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Introduction: In parallel MRI, multiple receiver coils collect signal in locally specific regions providing some spatial encoding. Coil sensitivity profiles, are then used in methods such as SENSE, GRAPPA, and SMASH to reconstruct images from under sampled data[1]. Traditionally, magnetic gradient fields encode spatial information through manipulations of the phase and frequency of the MR signal, and signals are decoded using the Fourier transform. However, linear gradients may not be optimal for parallel imaging. This work is focused on designing non-linear encoding fields that provide complimentary encoding to the receiver coils. An optimal gradient shape creates signal frequency and phase separation where coil sensitivities poorly distinguish signal. Spherical harmonic field gradients, notable for their ability to form a complete basis set and current implementation in shim coils, may form a composite encoding function that increases resolution and decreases imaging time through improved signal localization. In the current work, a metric independent of reconstruction methodology factors coil sensitivity and encoding function shape to guide the optimization of linear combinations of spherical harmonic functions to maximally complement information from localized receive coil sensitivities.

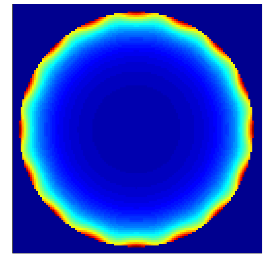

Figure 1. Sixteen element planar microstrip array coil sensitivity magnitude

Theory: In the vein of geometry factor g maps[2], but abstracted from a given reconstruction method, a mapping of inter-receive coil orthogonality relates patterns of strong and poor spatial localization. For instance in a given region, a receive coil with rapidly changing sensitivity would complement a coil that detects signal equally. To characterize coil inter-coil spatial localization, coils were considered pair-wise for the cosine of the angle between the directions of their maximum rate of change or the gradient vector. The pair-wise maps were summed to create a localization map L as follows:

$$L_{xy} = \sum_{i,j,i\neq j} \frac{(\nabla C_{i,xy}) \cdot (\nabla C_{j,xy})}{|\nabla C_{i,xy}| |\nabla C_{j,xy}|} \quad (1),$$

where C is the magnitude of the coil sensitivity profile, i and j are indices over the receive coils, and x and y are indices over the image matrix. In order to complement receive coil localization, magnetic gradient coils should vary rapidly in high L-valued regions. In particular, the objective function to be maximized is,

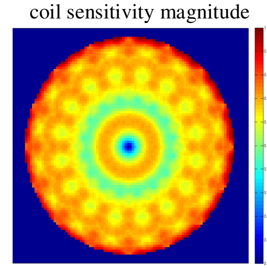

Figure 2. Receive coil spatial localization map

$$Obj = \sum_x \sum_y L_{xy} |\nabla G_{xy}| \quad (2),$$

where G is the gradient encoding field. Large values for the objective function indicate that regions of poor receive coil localization are compensated by encoding gradient localization.

Method: The sensitivity profiles of sixteen microarray strip coils circumferentially arranged around a field of view were simulated based on theoretical calculations by Lee[3]. In the calculation of Eq. (1), coil sensitivity magnitudes C, as depicted in fig. 1, were constructed from the in-plane components of the magnetic field. For the components of G, a linear combination of the nine non-degenerate single slice in-plane (z = 1) spherical harmonic functions through third order (fig. 2), were found using a line search optimization to maximize the objective function. The coefficients of the spherical harmonic functions represent the free parameters in the optimization. Values in L and ∇G were normalized to place spatial localizations capabilities on equal footing.

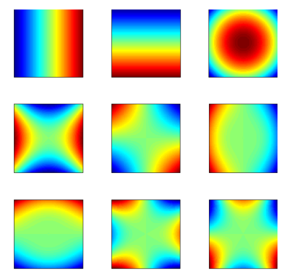

Figure 3. Spherical harmonics through third order, from upper left, common names: X, Y, Z^2, C^2, S^2, Z^2 X, Z^2 Y, X^3, Y^3

Results: The spatial localization map L, shown in fig. 2, reveal concentric localization patterns with maximum spatial localization achieved in the center of the field of view. In contrast, signal decay as visualized in fig. 1 decays rapidly towards the center. The optimized magnetic gradient fields, fig. 4, exhibit a rotational symmetry. A six-pole gradient shape is shown on the right in fig. 4, which was a local maximum, though not scoring as high as fig.4a. In fig. 4a, the greatest weighting is given to the z^2 shim, while in fig. 4b, the third order spherical harmonics receive greatest weighting, with the Y^3 function receiving over five times the weighting of the next heavily weighted spherical harmonic. Although not illustrated, the encoding gradient with reversed polarity has an equivalent value for the objective function.

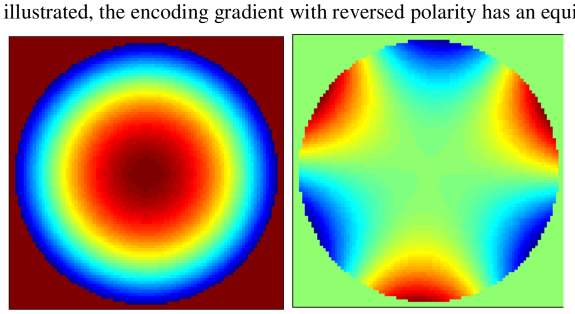

Figure 4 a, b. Magnetic gradient functions comprised of linear combinations of third order and lower spherical harmonics found by line search optimization of eqn. (2), with left gradient representing the largest maxima found.

Discussion: By considering the spatial localization abilities of a coil array independent of a reconstruction method, we have designed a gradient field that broadly complements spatial localization in the coil array. In addition, the components of the aggregate gradient shape are spherical harmonics, which are well-known, implemented solutions to Laplace's equation. Given a gradient shape better suited for spatial localization in parallel imaging, higher acceleration factors may be obtained without using conventional phase encoded acquisitions. The encoding function depicted in fig. 4b has been used for reduced slew rates and increased peripheral image resolution[4] but with phase encoding applied orthogonal to the read gradient. Using the encoding function of fig. 4a, a novel approach, termed O-space imaging, shifts the encoding function center placement between echoes or acquisitions to better utilize the full spatial localization capability available in parallel imaging and to do away with phase encoding completely.

References: [1]Sodickson, DK, McKenzie, CA. Med. Phys 2001; 28: 1629. [2]Preussman, KP, et. al. Magn. Reson. Med 1999; 42:952. [3]Lee, RF, et. al. Magn. Reson. Med 2004; 51:172. [4]Hennig, et. al. ISMRM 2007, 453.