Overcoming Coil Phase Effects in Highly Accelerated Imaging with a Dedicated Fourth Gradient Channel

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Introduction:

Magnetic Resonance Microscopy (MRM) has become an important tool for in-vivo microscopy and in-vitro/ex-vivo histology [1,2]. One challenge is that very small RF coils are used to obtain high signal-to-noise ratio in small voxel sizes. These small coils limit the field-of-view, necessitating the use of arrays or multiple studies. In this abstract, the extension of our previous work in Single-Echo-Acquisition (SEA) imaging to enable wide-field-of-view MRM is discussed.

SEA MRI uses arrays of small coils along with multiple receiver channels to obtain entire images in a single echo [3,4]. In Wide FOV Microscopy, multiple phase encode lines are obtained, but the narrow dimension of the coils remains similar to the voxel dimensions, leading to a phase dispersion across the voxel due to the coil sensitivity pattern. This phase dispersion in turn shifts the center of k-space (peak signal) away from the origin, impacting which k-space lines one would obtain in a highly accelerated parallel acquisition. This phase shift is dependent on the relative position of the voxel to the coil, complicating 3D imaging, commonly used in MR Microscopy. Additionally, as the phase can be shown to reverse from one side of the coil to the other, it effectively prevents imaging on both sides of the array or using two arrays in a "sandwich" configuration as might be desirable for histological applications.

Here we describe the addition of a fourth gradient channel to a small animal MRI system in order to provide a spatially dependent, non-linear gradient field that helps offset the spatially dependent phase dispersion of the array elements. This gradient amplifier is controlled by the host MRI system, and one or two gradient coils, placed underneath or around the RF coil array, minimize the signal dropout due to the coil phase pattern, improving 3D MR Microscopy and enabling the use of "sandwich" coils.

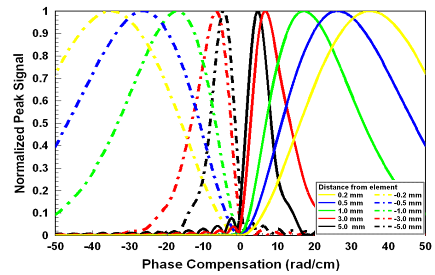

Figure 1. Without proposed fourth planar gradient coil, each "slice" of spins refocuses at a different phase encoding line (expressed as phase compensation rad/cm).

Methods:

In SEA imaging, the phase encode table is replaced by a fixed y-gradient value (along the narrow axis of the coils), and an image is acquired from the array with each echo acquisition. The fixed gradient pulse is referred to as a phase compensation gradient [4,5]. This method has enabled imaging of flow and motion at frame rates of up to 250 images per second.

A potentially interesting application of the array coils used for SEA imaging is to use them for microscopy applications. Instead of forming an image from a single echo, partially parallel imaging can be used to provide high resolution and reduced imaging times. There are two primary difficulties with the application of the array coils to microscopy. First, as shown in Figure 1, the optimal value of the phase compensation gradient varies with the distance of the imaging plane from the coil. This creates a problem for 3D imaging, as there is no "optimal" value of the phase compensation gradient to use over a thick slice. Second, and of more importance when considering the development of a dedicated probe for wide-field MR microscopy, is that the optimal phase compensation gradient reverses sign depending on whether the sample is "above" or "below" the array [5].

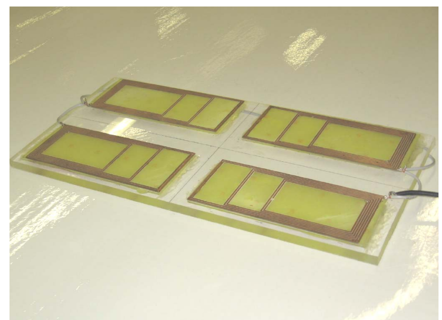

Figure 2. Prototype gradient coil constructed using an in-house milling machine and two-sided 4 oz copper clad board, o.d. 20 x 40 cm.

Results and Discussion:

The value of 3D-SEA as a wide-FOV microscopy tool is diminished because the SNR drops off rapidly with distance from the array and due to the position dependent array phase shift. We are implementing two modifications to help overcome these problems. First, we are adding a second planar array, on top of the sample. This will improve the sensitivity to slices far from the array, originally on the bottom of the sample. Second, we have added a fourth gradient channel and gradient coil to our MRI scanner. By adding a fourth gradient coil, in this case a conventional surface gradient coil as proposed by Cho [6], it becomes possible to vary the amplitude of the y-directed phase compensation gradient in the x direction. Figure 2 shows a prototype phase compensation gradient coil to be used as the fourth gradient coil.

The planar gradient, combined with a fixed offset from the standard cylindrical x-gradient coil, provides a y-dependent phase compensation gradient. The combination of the two gradient coils enables the phase compensation gradient to be reasonably well-optimized for all slices, including the sign reversal at the midpoint of the volume where the opposite array becomes dominant. Figure 3 shows the resulting signal-vs-k-line curves for all slices of both arrays, and shows convergence of the optimal k-space line. By combining arrays of microcoils with high-speed digital receivers it becomes feasible to provide high-resolution imaging over large imaging regions in reasonable time. The addition of the dedicated planar gradient coil generates a 'height-tailored' phase compensation gradient, allowing the use of 3D imaging and dual or "sandwich" arrays to greatly enhance the sensitivity of this technique.

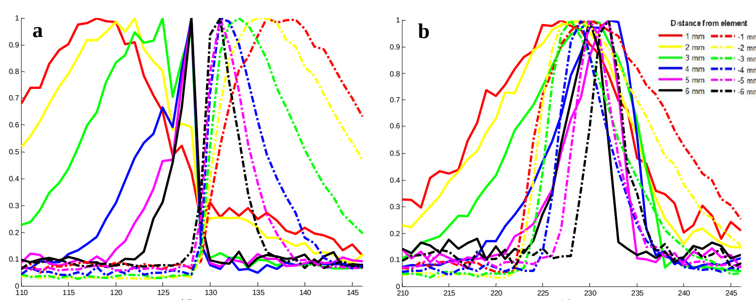

Figure 3. Measured results corresponding to Figure 1 showing signal strength in the echo peak vs. k-space line (phase encoding line). (a) Without the planar gradient, the peaks are dispersed, preventing 3D imaging. (b) With the planar gradient the signal peaks cluster near one value of the phase compensation gradient (k-space line).

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