Local Planar Gradients with Order-Of-Magnitude Strength and Speed Advantage

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Introduction: We report on the design and fabrication of local gradients that can reach 500 mT/m strength at 320 A while keeping the loads and temperatures within safe limits [1]. The local gradient coils can reach gradient strengths and slew rates an order-of-magnitude higher than conventional gradient coils for a given input current, over a limited but useful imaging volume. Since the gradients are applied over a small imaging volume, local coils have the additional advantage of keeping the maximum dB/dt within safe limits. Because of these features, we believe that local gradient coils could play an important role in high speed cardiac MRI, diffusion-weighted MRI and applications requiring the highest resolution.

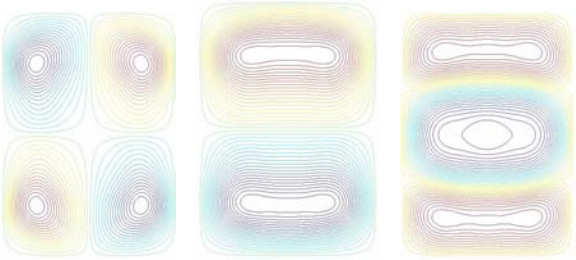

Figure 1. Current patterns for (left-to-right) *x*, *y*, and *z* gradient coils. Red and blue are opposite current directions.

Methods: The flat gradient coil design first reported in [1] has now been constructed, mounted, and used in imaging experiments. Figure 1 shows the current patterns in the x, y, and z gradient boards. Figure 2 shows the completed assembly of the coils. Each axis was composed of two sets of boards connected in series, assembled with cooling lines between them. The completed assembly was then impregnated with high-temperature epoxy under vacuum. The assembly was instrumented with 12 thermocouples to monitor internal and surface temperatures. All these features enable the coils to be operated at high performance while keeping the internal and surface temperatures within safe limits. The coil was mounted on a modified bridge inside a GE Signa scanner as shown in Fig. 2 (right). Threaded rods were cast into the epoxy as an integral part of the coil assembly. The bridge was reinforced at the coil mounting locations to ensure structural integrity. Resistance values for the final assembly were 65 mΩ, 91 mΩ, and 129 mΩ, for the X, Y and Z axis, respectively. Inductances were 472 μH for the X axis, 858 μH for the Y axis, and 922 μH for the Z axis. GE HFD gradient amplifiers were tuned to match the flat gradient characteristics.

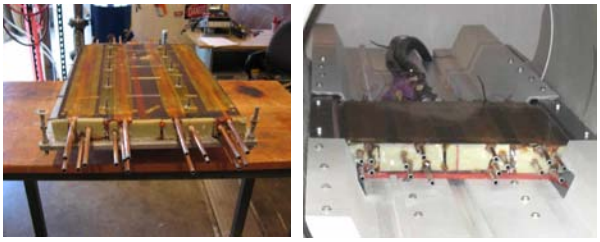

Figure 2. Assembled gradient coils (left) and installed in the bore of the scanner (right).

Axial, sagittal, and coronal imaging experiments were performed on phantoms and normal volunteers using spin-echo and fast gradient-echo pulse sequences

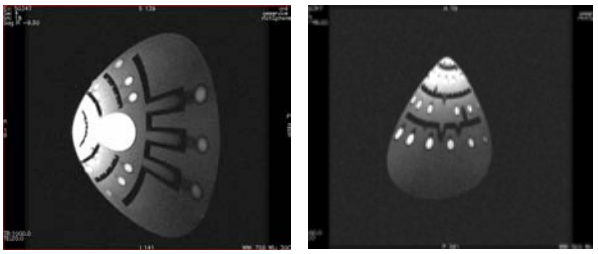

Figure 3. Sagittal (left) and axial (right) images of spherical quality control phantom.

Results and Discussion: Initial operation of the coil was found to be satisfactory without either excessive movement of the coil during MRI, or significant temperature increases. Figure 3 shows uncorrected sagittal (left) and axial (right) spin-echo images of a standard spherical phantom designed as a quality control phantom for structural MRI studies of neurological disorders [2]. The conventional images of the sphere are distorted by non-linearity in the gradient fields. Figure 4 shows a sagittal spine image acquired using a breath-held multi-NEX fast gradient-echo pulse sequence before (left) and after (right) application of an unwarping algorithm based on the field maps.

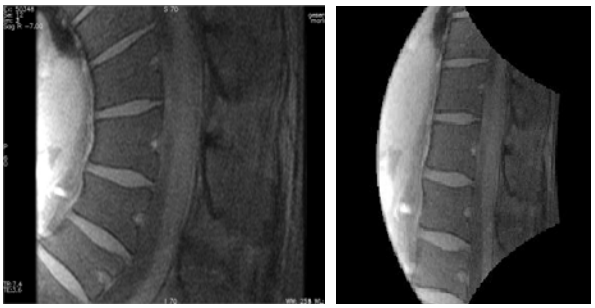

Figure 4. Spine images, with 14 cm FOV. Original image is on the left, unwarped image on the right.

Smaller flat gradient coil systems provide gradient field strengths and slew rates an order-of-magnitude higher than that are achievable with existing clinical MRI systems at modest power levels and net dB/dt rates. Gradient field aliasing and non-linearities can be dealt with by restricting the field-of-view and with non-linear reconstruction

References:

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2. R. Mallozzi, et al. Proc ISMRM 13 (2005), 1246.

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