Design, Fabrication, and Testing of an Insertable Double-Imaging-Region Gradient Coil

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ABSTRACT: We have constructed a small-bore insertable gradient coil with two linear gradient imaging regions and interfaced it with an MRI scanner. We have also constructed an RF system capable of transmitting or receiving in both regions simultaneously. Designs for conductor placement for two-region X-, Y-, and Z-gradient coils were optimized by simulated annealing. Wire patterns for each axis were chosen that gave low inductance, reasonable homogeneity over a large imaging volume and high efficiency (gradient field per-unit-current). Imaging was performed on a Siemens 3T TIM Trio scanner equipped with three additional gradient amplifier channels and a second RF/gradient array controller. Phantoms were placed in the two imaging regions as well as the central non-imaging region to test gradient homogeneity and crosstalk between regions. Images acquired simultaneously in the two regions showed very little signal crosstalk between imaging regions and even less signal from the central, non-imaging region. When combined with an overlapping single-region gradient insert, extended field-of-view (FOV) imaging will be possible without moving the table or the subject and without increasing nerve stimulation. Construction and testing of a two-region gradient coil insert is a necessary intermediate step as a proof of concept for an extended field of view, contiguous, three-region human-sized gradient system.

INTRODUCTION

As part of our continuing work to develop high-performance gradient coils capable of imaging large fields of view without increasing the risk of nerve stimulation, we have designed and constructed a small-bore insertable gradient coil with two linear gradient regions of opposite polarity (1, 2) as shown in Fig. 1. Nerve stimulation can be caused by rapidly changing magnetic fields, specifically the rate of change and the magnitude of the field excursion (3–5). As demonstrated in Fig. 2, a two-region gradient can have the same maximum slope (in mT/m) as a single region gradient and yet much smaller maximum excursion. This is the redeeming feature to justify the extra hardware and difficulty of a multiple region gradient set.

We have now interfaced this gradient coil insert with a Siemens TIM Trio 3 Tesla MRI scanner (Fig. 3) equipped with two gradient amplifier systems.
Additionally we have constructed an RF Transmit/Receive (Tx/Rx) coil array capable of transmitting or receiving in both regions simultaneously as shown in Fig. 4. When combined with an overlapping single-region gradient insert capable of generating imaging gradients in the central region, extended FOV imaging will be possible without table motion (2). Construction and testing of a two-region gradient coil insert is a necessary intermediate step as a proof of concept for the three-region gradient system.

METHODS

Design

The dimensions of the two-region gradient insert were constrained by the diameter of the homogenous spherical volume (DSV) of the main B0 magnetic fields of the commercial human MRI scanners used to test the insert. The maximum DSV of the commercial MRI system available to us was approximately 50 cm. The Z-extent of the DSV was divided into three equal regions (compare Figs. 1 and 5). For this double-imaging-region prototype coil design, the two outer imaging regions were 17 cm in length along the magnet bore (Z-dimension) and were separated by a 17 cm non-imaging space.

Standard plastic piping for the gradient former were chosen to be of radii large enough to allow for reasonably large phantoms and RF coils yet small enough so that the gradient insert might be centered in the clinical system bore, as well as to keep the weight of the insert manageable. We chose 6, 8, and 10 inch inner-diameter (i.d.) schedule-40 polyvinyl chloride pipe (15.2, 20.3, and 25.4 cm) for formers of the X-, Y-, and Z-coil windings, respectively, and a 12 inch i.d. (30.5 cm) pipe to enclose the system.

Actual diameters of the conductor winding cylinders were 17.21, 22.29, and 27.69 cm for the X, Y, and Z-w windings, respectively.

After the radii budget was determined, designs for conductor placement for two-region X-, Y-, and Z-axis coils were individually generated with software we developed using the stream function approach (6) and optimized by simulated annealing (2, 7, 8). We explicitly computed coil inductance, efficiency, gradient homogeneity and magnitude of the \( B \)-field, which is important because it is the rapid change in the \( B \)-field that induces nerve stimulation. Figures of merits (FOM) for characterization of potential performance of the gradients were optimized by a simulated annealing algorithm. For the axial gradient, \( \text{FOM}_z = \sqrt{L/\sigma} / \eta \), where \( L \) is inductance, \( \sigma \) is root mean square of the deviation from desired field gradient, and \( \eta \) is gradient efficiency (gradient strength/unit current). For the transverse axes, \( \text{FOM}_{xy} = \text{FOM}_z \sqrt{B_{\text{0max}}} / \eta \) and \( B_{\text{0max}} \) is maximum field at \( 5/6 \) of the radius of the bore. Calculation of the field at \( 5/6 \) of the winding radius was chosen in order to get a qualitative comparison of the potential for nerve stimulation. The field at this radius is outside the imaging volume (assumed to be \( 3/4 \) of the radius) and yet still within the subject being imaged.

![Figure 1](image1.png) Schematic for gradient insert with two separate imaging regions of opposite polarity. The dashed line indicates the Z-gradient field, which is linear in the front and back regions. Transverse gradient fields generated by the insert for the front and back region are also opposite polarity.

![Figure 2](image2.png) Comparison of gradient slope and excursion of a single region configuration (top) and double region configuration (bottom). In the top diagram, coils A and C have current in opposite directions and result in a somewhat linear gradient field in the volume between the coils (dotted line). By having the current in A and C in the same direction and another coil (B) with current in the opposite direction a double region gradient field is created. The slope of the dotted lines is an indication of the change of B field. Notice that the slope of the gradients are equal (this makes the resolution of both systems equal) but the excursion of the gradient B-field is much less for the double region gradient thus reducing the potential of nerve stimulation! Arrows indicate relative current direction.
During the simulated annealing process all solutions were saved and used to generate a feature space with three axes: 1) \( \eta_h = \eta / \sqrt{L} \), 2) \( \sigma \), 3) \( B_{\text{max}} \). After the three-dimensional feature space was generated, two-dimensional projections were used to help visually compare homogeneity, efficiency/inductance and maximum \( B \)-field tradeoffs for different operating points. After operating points were chosen for each axis the software generated wire patterns (Fig. 6). These conductor patterns were chosen from operating points of feature space with homogeneity and normalized efficiency \( \left( \eta_h = \eta / \sqrt{L} \right) \) as the most heavily weighted considerations. Many of the solutions in this feature space had reasonable maximum \( B \)-field so we were able to choose a solution with excellent homogeneity, efficiency, and low inductance without being limited by maximum \( B \)-field considerations.

**Construction**

The chosen wire patterns were milled with a ball-nose cutter into flat sheets of polytetrafluoroethylene (PTFE) that were treated on one side for improved adhesion with glues. These sheets were bonded with an adhesive to the pvc pipe forms and 12 AWG copper wire was inserted into the milled grooves. Thin-walled PTFE tubing for water-cooling was placed between layers. MRI compatible (Type-E) thermocouples were fastened to the X, Y, and Z conductors at regions of highest current density. The voids between axes were filled with epoxy that was poured into the insert and left to harden for mechanical stability. The epoxy potting compound (Durapot 865, Cotronics Corporation, NY) was specially chosen to have a high thermal conductivity (2.9 W/m/K) allowing efficient heat transfer from the winding layers to the cooling layers.

**Bench Tests**

Before imaging, the coil’s inductance, capacitance, and resistance were characterized over a frequency range of 12 Hz – 10 kHz using a GW Instek LCR meter model number LCR-817.

To quantify the thermal performance of the coil, two methods of temperature measurement were employed. The first method measured temperature directly from thermocouples placed inside the coil during construction. Every attempt was made to position the thermocouples at the point of greatest wire density and hence highest temperature region for...
each gradient axis. A second temperature method was employed that measures the average coil temperature via the change in resistance of the copper wire. To facilitate these measurements, a LabView 7.1 Virtual Instrument (VI) was created to interface with the thermocouples and Agilent 34420A Ohm Meter. This VI directly recorded thermocouple temperatures and calculated the average coil temperature from resistance measurements recorded every 5 sec.

An exponential model proved suitable for both the heating and cooling characteristics of this coil. For the heating portion of the experiment, the data were suitably fit by the function:

\[ T(t) = T_0 + \Delta T(1 - e^{t/t_{\text{heat}}}) \]

For each gradient axis, the heating time constant, \( t_{\text{heat}} \), and asymptotic temperature rise, \( \Delta T \), was determined by driving the axis with 75A DC for 30 min with forced-water-cooling. A fit of a simple exponentially decaying function with cooling time constant, \( t_{\text{cool}} \), was fit to the temperature data recorded as the system was left to cool for 30 min in the idle state with the water cooling flowing. The water cooling system was run with 15°C tap water with a flow rate of \( \sim 6 \text{ L/min} \).

**Imaging**

Imaging was performed on the Siemens 3T TIM Trio scanner (Siemens Medical, Erlangen Germany). The standard system was augmented with three additional gradient amplifiers and master/slave configured computers capable of controlling extra RF and gradient channels. The control hardware and software were developed and provided by Siemens. Two-region RF excitation and detection were accomplished by feeding a single transmit pulse to both 17 cm long Tx/Rx surface coils via a splitter and using a separate Rx channel for each coil as illustrated in Fig. 4. The RF coils and phantoms were placed inside the bore of the insert. RF coils were placed on top of the outer 2 phantoms. (see Fig. 5). A separate pulse sequence was used to control each gradient set. For these experiments, the master computer served three purposes: 1) to maintain the first-order gradient shims using the standard system gradients, 2) to control RF excitation and reception, and 3) trigger the slave computer. The slave computer executed a pulse sequence controlling the amplifier currents in the double-region gradient insert.

The 17 cm long, 12.7 cm diameter (5” o.d.) hollow cylindrical phantoms contained a plastic grid and were filled with CuSO4-doped water. \( T_1 \) and \( T_2 \) relaxation times were approximately 300 ms measured at 3T. Phantoms were placed in the two imaging regions as well as the central non-imaging region to test gradient homogeneity and crosstalk between regions.

All images presented here were acquired with a fast low-angle shot (FLASH) pulse sequence. Ranges for acquisition parameters were as follows: repetition

<table>
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<th>Axis</th>
<th>12 Hz</th>
<th>100 Hz</th>
<th>1 kHz</th>
<th>5 kHz</th>
<th>10 kHz</th>
<th>12 Hz</th>
<th>100 Hz</th>
<th>1 kHz</th>
<th>5 kHz</th>
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<td>235</td>
<td>235</td>
<td>232</td>
<td>233</td>
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<td>0.178</td>
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<tr>
<td>Y</td>
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<td>316</td>
<td>0.216</td>
<td>0.219</td>
<td>0.214</td>
<td>0.221</td>
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time (TR) 8.6–25 ms, echo time (TE) 2.8–6 ms, bandwidth (BW) 200–320 Hz/pixel, slice thickness (SL) 2–7 mm, field of view (FOV) 180 × 180 mm–480 × 480 mm, and image matrix 256 × 256 pixels to 460 × 512 pixels.

Signal Measurements

All measurements were made using magnitude images. For coronal localizer images obtained using standard gradients, four-sided polygon ROIs were chosen to include all signal from each of the three phantoms. For each axial and coronal image, one ROI was chosen that contained coherent signal only from the intended region and a second ROI was chosen to contain signal from the opposite region. A third ROI was chosen in the noise.

RESULTS

Bench Tests

Inductance and resistance measurements for each axis as a function of frequency from 12 Hz to 10 kHz are displayed in Table 1.

Mutual inductance and capacitance between axes as a function of frequency are displayed in Table 2.

Table 3 illustrates the temperature rise for each gradient axis under the conditions described in the Methods section. The columns labeled “Temp Rise” are the asymptotic temperature rises for each axis for a 75A DC current extrapolated for infinite duration from the fit of the temperature model to the acquired data. Of the three gradient axes, the Z-axis experienced the largest temperature rise. This is most likely because this wire pattern contains the most wire and thus largest resistance. The average temperature of the Z-axis rose 37°C while the thermocouple showed a temperature rise at the point of greatest wire density of 51°C. Both figures represent safe operating temperatures.

The two temperature measurement methods are consistent and show that the Z-axis runs the hottest followed by the X-axis, and the Y-axis. While the thermocouple measurements from the X- and Y-axes report smaller temperature rises than their respective average temperature rise, this is likely the result of the thermocouples having been displaced during coil potting. This illustrates the usefulness of the average temperature method which is not susceptible to this problem and does not require thermocouples to be included in the early designs.

The thermal performance testing shows that the cooling system effectively removes deposited power from the coil, and that the coil can be operated continuously in a nondestructive manner.

Imaging Results

Three-plane localizers were acquired using the standard Siemens gradients and our custom-built Tx/Rx RF coils. Coronal localizer images are displayed in Fig. 7 with and without gradient field inhomogeneity compensation image reconstruction to demonstrate the need for field-homogeneity compensation using the standard gradients across a large FOV.

After localizer acquisition we simultaneously acquired axial images in the front and back regions. For this acquisition, the slave computer was used to control the insert gradients as imaging gradients.

### Table 2 Mutual Inductance and Capacitance Between Axes as a Function of Frequency

<table>
<thead>
<tr>
<th>Axis</th>
<th>Capacitance (pF)</th>
<th>Mutual Inductance (μH)</th>
</tr>
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<tr>
<td></td>
<td>12 Hz 100 Hz 1 kHz 5 kHz 10 kHz</td>
<td>100 Hz 1 kHz 5 kHz 10 kHz</td>
</tr>
<tr>
<td>XY</td>
<td>1,190 869 755 725 718</td>
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<td>XZ</td>
<td>927 545 428 402 396</td>
<td>0.86 0.85 0.70 0.81</td>
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<tr>
<td>YZ</td>
<td>1,348 1,015 848 808 799</td>
<td>0.42 0.40 0.39 0.35</td>
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### Table 3 Temperature Rise for Each Gradient

<table>
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<tr>
<th>Axis</th>
<th>Average Temperature</th>
<th>Thermocouple Temperature</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Temp rise [°C] t_{Heat} [s] t_{Cool} [s]</td>
<td>Temp rise [°C] t_{Heat} [s] t_{Cool} [s]</td>
</tr>
<tr>
<td>X</td>
<td>38.32 533 573</td>
<td>38.32 533 677</td>
</tr>
<tr>
<td>Y</td>
<td>31.76 299 357</td>
<td>31.76 299 474</td>
</tr>
<tr>
<td>Z</td>
<td>37.03 339 348</td>
<td>37.03 339 317</td>
</tr>
</tbody>
</table>
while the master computer controlled the RF transmission and reception. Example axial images are displayed in Fig. 8. These images are not processed to compensate for gradient field inhomogeneity. We also acquired sagittal and coronal images using the same computer control described above for axial acquisition. Example coronal images are displayed in Fig. 9.

We observed out-of-region signal (stray signal) from the opposite region as well as the central region. Stray signal was always less than 5% of the maximum signal. When care was taken to isolate RF coil connecting wires from conducting surfaces within the insert bore even less stray signal was evident. Mean signal from the central phantom from the localizer shown in Fig. 7 was the worst with stray signal from the central region at 4.7% of the mean from the adjacent phantom. Stray signal from axial images in Fig. 8 was entirely from the opposite region. Mean crosstalk signal from an ROI sampling the signal that is obviously from the opposite imaging region measured in magnitude reconstructed images was 4.3 times mean background noise and 2.0% of maximum signal from the back region, and in the front region inappropriate signal was 2.9 times mean background noise and 1.4% of maximum signal.

Stray signal observed in the coronal images in Fig. 9 was also entirely from the opposite region. The slices labeled 6 were the most straightforward to analyze since the overlapping slices were easy to distinguish. Slice 6 of the back region had mean signal from the opposite region (stray signal) measured at 4.0 times the mean level of noise. Slice 6 of the front region was similar with stray signal 5.1 times the mean noise. The mean percent stray signal relative to

Figure 7  Coronal localizer images showing the phantoms using the “standard” Siemens gradients and custom-built dual RF surface coils. Uncorrected images are shown in the left panel and distortion-corrected images in the right. Imaging parameters: TR = 8.6 ms, TE = 2.8 ms, BW = 320 Hz, SL = 7 mm, FOV = 480 x 480 mm, image matrix = 460 x 512 pixels. Faint signals from the phantom located in the central region can be seen with appropriate display parameters. The signal from the central region is less than 5% of that in either the front or rear regions.

Figure 8  Axial GRE images of the grid phantoms acquired simultaneously (a) in the back region of gradient insert (correct gradient polarity) and (b) front region of gradient insert (reversed gradient polarity, notice air bubbles) Imaging parameters: TR = 9.1 ms, TE = 4.8 ms, BW = 390 Hz, SL = 5 mm, FOV = 180 x 180 mm, image matrix = 256 x 256 pixels.
the maximum for slice 6 front and back was 3.2 and 4.4%.

It should be noted that none of the images acquired using the double-region insert gradients were corrected for gradient-field inhomogeneities. Due to the difficulty of sorting out gradient linearity from field shimming, phantom and insert placement we have not attempted to quantify gradient linearity here. However, it can be seen from comparison of coronal slices of Figs. 7(a) and 9 that qualitatively our double region insert has better linearity over the imaged volume than the manufacturers’ gradients and from the uncorrected axial and coronal images of Figs. 8 and 9 the overall linearity is quite good.

**DISCUSSION AND CONCLUSIONS**

We have shown that imaging can be performed simultaneously in two separate imaging regions of the gradient insert without significant crosstalk between the two regions. Additionally, stray signal detected is most likely due to the proximity of RF system components to the RF shield and RF traps rather than shortcomings of the gradient insert. Separate RF coils in each region of the insert are necessary to isolate received signal coupling from the two different regions.

It was fairly obvious when stray signal from the opposite region was contaminating images. Stray signal could easily be determined when it did not overlap signal from the appropriate region, however, ROIs chosen in appropriate areas may have some small signal from the central or opposite regions that cannot be sorted out. Also, since surface coils were used for transmit and receive, the image signal is heavily determined by the sensitivity of the surface coils, so comparisons of maximum or even mean signal is more qualitative than we would like.

Notice the polarity is reversed in the front region as expected. Very little unwanted signal from the central region was evident (<5% of signal of the desired region) as well as stray signal from the opposite region (<5% of signal of the desired region). Eddy current compensation does not seem to be a problem, probably due to the distance between our insert and conducting surfaces within the magnet. Field shimming issues remain to be addressed.

In future work, this double region gradient coil can be combined with another gradient set to image the central region. Images acquired in each region could be “stitched” together for a combined extended FOV image.

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**REFERENCES**


