Published in final edited form as:

Magn Reson Med. 1994 August ; 32(2): 246–250.

A Modified Quadrupole Gradient Set for Use in High Resolution MRI Tagging

Walter G. O'Dell, Joseph S. Schoeniger, Stephen J. Blackband, and Elliot R. McVeigh

From the Department of Biomedical Engineering (W.G.O., E.R.M.) and Radiology (E.R.M.), Johns Hopkins University School of Medicine, Baltimore, Maryland; MRI Centre, Hull Royal Infirmary, Hull, England (S.J.B.); and the Center for Computational Engineering, Sandia National Laboratories, Livermore, California (J.S.S.).

Abstract

A special purpose gradient set with a $\cos(2\theta)$ current distribution has been constructed for high resolution MR imaging of small samples in a clinical Signa 1.5 T scanner using the existing gradient amplifiers. The X, Y, and Z gradient coils can attain gradient field strengths of 11.3, 4.7, and 15.2 G/cm at 100 amps current, respectively, with a slew rate of 20 G/cm/ms and usable ramp time of 150 μ s. Field distortions are less than 2% over the central 8 cm of the bore, suitable for high resolution tagging of isolated canine hearts.

Keywords

gradient sets; cardiac MRI; quadrupole; high resolution

INTRODUCTION

Cardiac mechanical dysfunction resulting from cardiovascular disease is the leading cause of mortality in the US. To understand the complex deformations of the heart, sophisticated mathematical models are employed. These models incorporate detailed finite element representations of wall geometry and fiber structure along with accurate estimates of local tissue material properties (1). However, the complexity of small-scale motions predicted by such models greatly exceeds our present ability to measure deformations in the living heart. Accurate, high density, global 3D strain data are crucial for reconstructing the deformations of the heart unambiguously. At high spatial resolution it is important to have very efficient data acquisition, that is, the fraction of time spent collecting signal over the time spent switching gradients should be as high as possible (2). For these reasons, rapidly switching gradients capable of high gradient amplitudes are preferable over the standard whole-body gradients.

The gold standard for high resolution myocardial strain analysis has been cine-radiography of implanted metal beads (3). This approach has significant limitations. At most two $\sim 1 \text{ cm}^3$ regions have been analyzed simultaneously because of the difficulty in bead implantation and identification. Additionally, scar tissue formation and edema from the implant surgery, and the physical presence of the beads all could contribute to deviations from natural motion.

To overcome these limitations while maintaining the 0.1 mm point tracking precision of the implanted marker method, we set out to construct a special purpose MR gradient set and radio-frequency (RF) coil assembly capable of generating high resolution tagged cardiac images.

Although the major constraint on the design was geometric, the gradient set was expected to at least match several performance goals. The working set was therefore designed with the following considerations: (a) be capable of at least 3 G/cm in each direction, (b) have switching times less than 500 μ s, (c) produce reasonably linear fields within a 8 cm³ volume, (d) interface with the GE Signa clinical instrument and pulse sequences with minimal disruption, (e) fit within the dimensions of the 55 cm diameter Signa magnet bore, (f) be compatible with a vertically suspended isolated heart preparation and a solenoidal RF coil system, and (g) be relatively inexpensive and easy to construct. All of these considerations are compatible with a *Quadrupole* gradient design which uses a $\cos(2\theta)$ current distribution on a cylinder oriented perpendicular to the B_0 field. An ideal current distribution of $\cos(2\theta)$ around an infinitely long cylindrical former generates a perfectly linear and homogeneous gradient field at all points inside and this field falls off as $1/R^3$ outside (4). Previous uses of quadrupole gradients in NMR imaging focussed on microscopy of relatively small samples (4) and in nonimaging applications for diffusion measurements (5,6) and in nuclear physics experiments (7).

THEORY

The $\cos(2\theta)$ gradient designs are derived from finite-pole approximations of the ideal current distribution $J \propto \cos(2\theta)$ on a cylinder oriented orthogonal to the static B_0 field, where J is directed parallel to the long axis of the cylinder and θ is the angle about the central axis of the gradient set, with $\theta = 0$ defined along the direction of the B_0 field. The finite pole approximation can be achieved by placing segments evenly spaced around the cylinder connected to a resistor network that allocates to each wire a current magnitude that varies as $\cos(2\theta)$. The same effect can be achieved by varying the number of conductors in series at each location. Alternatively each pole could carry the same current but with the density of wire segment spacing as $\cos(2\theta)$. For this latter approach the segment locations can be determined as follows. For an N pole approximation, each quadrant, located at the 0, 90, 180, or 270 degree lines, contains $N/4$ poles symmetrically placed about the quadrant line. The angle from the quadrant line to each pole, n , is given by

$$\theta_n = \frac{1}{2} \cdot \sin^{-1} \left(\frac{2n-1}{N/4} \right) \quad [1]$$

A self-shielded design is possible with the return paths parallel to the primary segments, at locations displaced radially outward. Shielding reduces the inductive coupling to conductors outside the gradient set thereby reducing the inductive load. However, shielding also reduces the gradient strength and requires a much larger space. For the dimensions of the isolated heart apparatus, the dimensions of the Signa magnet bore and the imaging goals, these drawbacks eliminated this design option.

Alternatively, a nonshielded design is constructed by connecting each pole in one quadrant to a pole in an adjacent quadrant, as shown in Fig. 3. If the connecting paths are far enough away from the sample, they will not introduce significant adverse field effects.

Theoretical field profiles for both an idealized 40 pole $\cos(2\theta)$ and common Maxwell pair configurations of similar dimensions and gradient field strengths are shown in Fig. 1. The $\cos(2\theta)$ configuration has a length equal to the diameter, but is idealized by having no current path end connections. Contour plots of the gradient of the field are shown in Fig. 2. The contour lines represent gradient field deviations from the central field value in 1 percent increments. Linearity and homogeneity can be further improved by fine tuning the exact conductor locations (8).

Since the $\text{Cos}(2\theta)$ coil provides a fairly uniform field gradient over a substantially greater percentage of space than the Maxwell pair, the actual gradients can be made to closely fit the sample and RF coil.

The scaling of gradient field strength with former radius is given by

$$\frac{dB_z}{dz} \propto \frac{nI}{a^2} \quad [2]$$

where n is the number of turns, I is the current and a is the cylindrical radius of the coil. The total inductance scales with radius as

$$L \propto n^2 \cdot a + L_{\text{ext}} \quad [3]$$

where L_{ext} is due to the coupling of the gradient field to the external constant field magnet and other surrounding conductors. The field from a working $\text{Cos}(2\theta)$ gradient set is expected to fall off by $1/R^3$ to $1/R^2$ outside the coil (4), thus the external inductive coupling to surrounding conductors decreases significantly when the gradient size is reduced. As the expressions for the gradient field and inductance indicate, employing a $\text{Cos}(2\theta)$ gradient set of minimum size for the sample (possible because of the excellent linearity of the field profile) maximizes the magnitude of the magnetic field gradient produced, and minimizes the inductance.

As mentioned earlier, the axis of the cylindrical quadrupole gradient set must be aligned perpendicular to the B_0 field. This arrangement works well when imaging isolated heart preparations since the heart may be suspended vertically in a perfusate bath, reducing gravitational alterations to the mechanics of the heart. The orthogonal configuration of the gradient set is also compatible with an inductively coupled solenoidal RF coil of a minimum diameter to contain the isolated dog heart; providing high efficiency and high sensitivity (9).

GRADIENT SET CONSTRUCTION

The X and Z gradients were made on concentric, acrylic, cylindrical formers with 10 poles per quadrant. The placement of the 40 poles around the cylindrical former is shown in Fig. 3. The inner cylinder is rotated 45 degrees with respect to the outer to generate the orthogonal gradient field. For the Y gradient, two poles per quadrant were used to allow for the insertion of the sample. Solid rectangular, acrylic formers were used for this gradient.

Mechanical fatigue due to the forces generated by the large currents in the presence of the static magnetic field is also an important consideration. For a conducting element oriented perpendicular to a static magnetic field the force per length acting on it is given by

$$\frac{F}{x} = I \cdot B \quad [4]$$

A 10 cm strand of conductor in a 1.5 Tesla field with 50 amps of current feels a force of 7.5 N and a group of 15 wires experiences a net force of 112.5 N (25.2 pounds). The method of fastening the wires to the former must have the mechanical integrity to withstand repeated pulsing of this force.

The completed gradient set is shown in Fig. 4. The conducting wires (Belden Heavy Armored Poly-Thermaleze Magnet Wire; Cooper Industries, Beldon Division, Richmond IN) are set into grooves machined into the formers. Layers of epoxy and fiberglass wrap fix the wires to the formers. A technique of etching copper film onto a plastic sheet which is then fixed to a rigid cylindrical former has also been demonstrated (8).

For the isolated dog heart experiment, the individual coil dimensions and wiring specifications are given in Table 1. Since the return paths should be placed as far away from the imaging

volume as possible, and these paths are located at the ends of the formers, the gradient set length reflects the largest dimensions possible within the scanner bore that is compatible with the isolated heart apparatus. Smaller gauge wire was used for the Y gradient because a decreased wire size allowed for a greater number of windings. The total cost of materials for the finished gradient set was approximately \$200 including wire, acrylic formers, and epoxy resin. This is a considerable reduction in cost over commercially available probes.

RESULTS

Orthogonal images of a plastic grid phantom in Fig. 5 show high grid uniformity throughout, indicative of the spatial uniformity of the gradients. In the central 8 cm diameter of the probe the spatial distortion was measured to be less than 2% in the long axis view (Y gradient) and less than 1% in the short axis view (X and Z gradients). From grid spacing measurements the gradient field strengths, scaled to 100 amps, were calculated. Each of the X , Y , and Z gradient inductances were measured within the 3D gradient set using a Hewlett-Packard Impedance Analyzer (model 4192A LF). These values are presented in Table 2.

The waveform of the amplifier current for a step from 0 to 1 G/cm for the Y gradient (the slowest of the three gradients) shows a very quick ramp phase lasting only $\sim 75 \mu\text{s}$, overshooting the 1 G/cm line by less than 1/15 G/cm, and a recrossing of the 1 G/cm line three times before stabilizing at 1 G/cm less than 75 μs after the initial peak. The working rise time for this gradient-amplifier circuit is thus 150 μs .

Using a rapid gradient echo tagging pulse sequence (10), high resolution tagged images of an isolated heart were obtained. The tag spacing was 3.5 mm, tag thickness ~ 0.6 mm; 256×128 image matrix; FOV 11–13 cm, corresponding to a pixel dimension across the tags of 0.43–0.5 mm; 1 NEX; slice thickness 5 mm; TE 2.8 ms; and TR 8.9 ms. Examples are shown in Fig. 6. All software corrections in the image reconstruction program were turned off.

Due to the low resistance of the wire used in the gradients, heat dissipation was not found to be a problem. At the inaximum current (40 amps peak) and maximum continuous imaging time for our protocol (on the order of 10 to 15 min) an immediate inspection did not find the wire temperature noticeably above room temperature.

DISCUSSION

In general, when choosing the optimal number of windings to include at each pole, there is a tradeoff between maximum gradient field strength and fast switching time. If the inductive load of a gradient set is considerably less than the output impedance of the given gradient amplifiers, either the output impedance of the amplifiers must be decreased, or a series inductance must be used to stabilize the circuit. Thus, there is no rise-time cost of increasing the number of windings when the inductance is very small if the amplifiers are unchanged. The built-in self-shielded GE Signa gradients have an inductance of 1 milli-henry, and they are matched to the GE Signa gradient amplifiers output impedance of 1 mH, 1 ohm. This gives a slew rate of 600 $\mu\text{s}/\text{G}/\text{cm}$. The $\text{Cos}(2\theta)$ gradients with ~ 0.1 mH inductances could be expected to switch 10 times faster than the conventional gradient set with the output impedance of the amplifiers properly matched to the gradients.

Since the heart is suspended vertically and most of the complicated deformations are expected to occur within the short axis view of the heart, the lower amplitude Y gradient is primarily used as the slice-selection gradient. Thus the strongest performing gradients, X and Z , will be used for phase, frequency and tagging where higher spatial resolutions are required.

Theoretical estimates on the precision of tag location measurements for these tagging-imaging parameters are ≤ 0.05 mm (11). In conjunction with the tagging/imaging sequence given in (10), and a suitable reconstruction scheme, 3D deformations throughout the canine left ventricle can be evaluated at a 0.1 mm precision (12). These measures are being used to validate complex finite element models of the deforming left ventricle. With an improved understanding of high resolution 3D strains in the living heart, we hope to prescribe imaging and tagging schemes in a more efficient fashion and evaluate strain measures at lower resolution in the clinical setting.

CONCLUSION

A practical quadrupole gradient set suitable for high resolution imaging on a clinical 1.5 T GE Signa scanner has been demonstrated. Although designed for use with isolated canine hearts, these gradients are suitable for other perfused organs and vertically oriented MRI samples. Clinical scanner compatibility allows these gradients to be used with conventional pulse sequences and imaging protocols, such as cardiac tagging. The gradients have a simple and inexpensive design which enables them to be made for specific applications. High resolution MRI tagging using these gradients allows the measurement of displacements to within 0.05 mm (11) making this a viable research tool for cardiac wall motion analysis.

ACKNOWLEDGMENTS

The authors thank Gary Glover for useful discussions about the Signa hardware and Rolf Lipowitz for machining of the acrylic formers.

This research was supported by the National Institutes of Health grant HL45683. W.G. O'Dell is funded through a Whitaker Foundation Biomedical Engineering Development Award.

REFERENCES

1. Guccione, JM.; McCulloch, AD. *Theory of Heart: Biomechanics, Biophysics, and Nonlinear Dynamics of Cardiac Function*. Springer-Verlag; New York: 1991. Finite element modeling of ventricular mechanics; p. 121-144.
2. McVeigh, ER.; Atalar, E. Balancing contrast, resolution and signal-to-noise ratio, *on AIP*. New York, NY: 1993. *The Physics of MRI*.
3. Douglas AS, Rodriguez EK, O'Dell WG, Hunter WC. Unique strain history during ejection in canine left ventricle. *Am. J. Phys* 1991;260:H1596-H1611.
4. Schoeniger, JS. *Theory and Equipment Design for NMR Microscopy and Its Application to the Study of Water Properties in Isolated Neurons from the Gastropod Aplysia California*. Johns Hopkins School of Medicine; 1992. Ph.D. thesis
5. Odberg G, Odberg L. On the use of a quadrupole coil for NMR spin-echo diffusion studies. *J. Magn. Reson* 1974;16:342-347.
6. Webster DS, Marsden KH. Improved apparatus for the NMR measurement of self-diffusion coefficients using pulsed field gradients. *Rev. Sci. Instrum* 1974;45(10):1232-1234.
7. Lobb DE. Some useful two-dimensional magnetic fields. *Nuclear Instrum. Methods* 1968;64(3):251-267.
8. Chu, KC.; Barberi, EA.; Rutt, BK. Optimized quadrupole gradient coil construction, for high resolution MR imaging; *Proc. SMRM, 12th Annual Meeting*; 1993. p. 311
9. Hoult DI, Richards RE. The signal-to-noise ratio of the nuclear magnetic resonance experiment. *J. Magn. Reson* 1976;24:71-85.
10. McVeigh ER, Atalar E. Cardiac tagging with breath hold CINE MRI. *Magn. Reson. Med* 1992;28:318-327. [PubMed: 1461130]
11. Atalar E, McVeigh ER. Optimization of tag thickness for measuring of position with magnetic resonance imaging. *IEEE Trans. Med. Imaging* 1994;13(1):152-160. [PubMed: 18218493]

12. O'Dell WG, Moore CC, McVeigh ER. Displacement field fitting approach to calculate 3D deformations from parallel-tagged MR images. *J. Magn. Reson. Imag* 1993;3(P):P208.

FIG. 1. Magnetic field profiles of B_z for the 2 loop Maxwell (top) and an idealized 40 pole $\text{Cos}(2\theta)$ (bottom) gradients coils. The dotted trapezoids cutting through the small gradient diagrams depict the plane over which the field is computed. The $\text{Cos}(2\theta)$ profile has been clipped at $0.9 \times$ radius due to large field values near the conducting elements.

FIG. 2.

Contour plots of the theoretical B_z field gradient for an idealized 40 pole $\cos(2\theta)$ Z gradient coil normalized with radius of 1 and length of 2 with no end current connections. (Left) Cross-sectional cutting plane as in Fig. 1. (Right) The orthogonal, longitudinal cutting plane with the long axis oriented left to right. The contour lines represent gradient field deviations from the central field value in 1% increments.

FIG. 3.
(Top) 3D diagram of the 40 pole $\cos(2\theta)$ Z gradient. (Bot) Schematic wiring diagram with current path connections.

FIG. 4. Constructed gradient sets. The X and Z gradient coil wires are set into concentric vertically oriented cylindrical formers. The Y gradient coil windings are on four rectangular solid acrylic formers that enclose the other gradient coils.

FIG. 5.

Images of a plastic phantom grid at immersed in water contained in a vertically oriented cylindrical chamber. The “long axis” view of the x/y plane (top) has a 16-cm FOV. The “short axis” view of the x/z plane (bottom) has a 12-cm FOV.

FIG. 6. Images of an isolated canine heart at a 13-cm FOV in the long and short axis views. A balloon filled with D₂O provides the signal void in the left ventricular cavity.

Table 1

Specifications for the Constructed Gradient Coils

Gradient direction	Number of poles	Number of wires per pole	Diameter (cm)	Length (cm)	Wire gauge
X	40	2	20.3	30.5	16
Y	8	15	34.3	35.5	18
Z	40	2	17.8	30.5	16

Table 2

Properties of the Constructed Gradient Coils

Gradient direction	Gradient field strength (G/cm)	Inductance (mH)	Resistance (ohms)
X	11.3	0.132	1.8
Y	4.7	0.604	7.7
Z	15.2	0.128	1.8

Gradient field strengths were measured at the gradient centers and are scaled to 100 amps current. Inductances were measured at 1 kHz.