Superelliptical Insert Gradient Coil with a Field Modifying Layer for Breast Imaging

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Abstract

Many Magnetic Resonance Imaging (MRI) applications such as Dynamic Contrast Enhanced MRI (DCE-MRI) of the breast require high spatial and temporal resolution, and can benefit from improved gradient performance, e.g. increased gradient strength, and reduced gradient rise time. The improved gradient performance required to achieve high spatial and temporal resolution for this application may be achieved by using local insert gradients specifically designed for a target anatomy. Current flat gradient systems cannot create an imaging volume large enough to accommodate both breasts, further, their gradient fields are not homogeneous, dropping off rapidly with distance from the gradient coil surface. To attain an imaging volume adequate for bilateral breast MRI, a planar local gradient system design has been modified into a superellipse shape, creating homogeneous gradient volumes (HGVs) that are 182% (Gₓ), 57% (Gᵧ), and 75% (Gᶻ) wider (left/right direction) than those of the corresponding standard planar gradient. Adding an additional field-modifying (FM) gradient winding results in an additional improvement of the homogeneous gradient field near the gradient coil surface over the already enlarged HGVs of the superelliptical gradients (67%, 89%, and 214% for Gₓ, Gᵧ, and Gᶻ respectively). A prototype y-gradient insert has been built to demonstrate imaging and implementation characteristics of the superellipse gradient in a 3T MRI system.

Keywords

Superellipse gradient; Breast imaging; Planar gradient; Insert gradient; Field modifying

INTRODUCTION

Many imaging tasks, such as dynamic contrast enhanced MRI (DCE-MRI) for breast lesion characterization, can benefit from high spatial and temporal resolution. The improved gradient performance required for high spatial and temporal resolution may be achieved by local gradient coils such as planar insert gradients. With higher gradient strength and slew rate, the local planar gradient can attain higher spatial and temporal resolution than the body gradients.

In general, local gradient coils are designed to produce a smaller homogeneous gradient volume (HGV) compared to that of the MRI system whole-body gradients. For local planar gradients, the x-gradient, which requires four fingerprint patterns per plane rather than two...
fingerprints as in the y-gradient, results in the most limited imaging volume. As a result, for certain applications such as breast imaging, the homogeneous gradient volume may be too small to cover the entire desired imaging region (i.e. both breasts). Because of the small imaging volume, there have been very few attempts to image both breasts simultaneously with a local gradient system (1).

MRI of the female breast is performed typically with the patient in a prone position to reduce breathing induced motion artifacts and increase reproducibility. This position places the breast near the planar insert gradient. However, the magnetic field as well as gradient strength of uni-planar gradient systems falls off nearly exponentially with distance from the coil surface. As a result, the gradient varies continuously and even an approximately linear region is narrow in the direction of each gradient. Image distortion caused by the non-linear gradient can be corrected to some extent by using a non-linear compensating transformation (2,3).

Non-planar surface geometries might be a good alternative for homogeneous breast gradient designs that can cover the volume of both breasts. The Boundary element method (BEM), which was introduced by Pissanetzky et al. (4), allows gradient coil design of irregular surfaces by using a finite element mesh with a great deal of geometrical freedom, and further research has followed (5,6). However, practical applications of unusually shaped gradient coil designs (7) remain unanswered due to manufacturing issues.

In this work, in order to create wider homogeneous imaging volumes, the planar insert gradient geometry was extended laterally and the edges were bent vertically using a superelliptical curvature on the left and right sides, creating a so-called box-in or superellipse shape to fit in the magnet bore. Bending the wire pattern on the vertical edges increases its effects on the imaging volume, increasing the gradient uniformity in the left/right and anterior/posterior directions. To increase the gradient strength and to further improve the gradient homogeneity in all directions, an additional Field-Modifying (FM) layer of current windings was added. Wire patterns are designed using stream functions (SF) (8,9).

We present design and optimization methods for a multi-layered superelliptical local insert gradient designed for breast imaging with a homogeneous imaging volume that is extended in all directions. This design overcomes previous planar gradient limitations and can accommodate both breasts. In addition, a prototype of the superelliptical y-axis gradient winding has been constructed and tested to assess imaging feasibility and address eddy current questions.

THEORY

In planar gradient coil geometry, the y-gradient has two fingerprint winding patterns along the z-axis with current flowing clockwise on one loop and counterclockwise on the other. Unlike cylindrical gradient coils where the x-gradient pattern can be obtained with a 90 degree rotation of the y-gradient, the x-gradient for planar gradient coils is achieved with windings that are laid out as a 2×2 fingerprint pattern on each plane with current flowing clockwise and counterclockwise alternatively (10–20).

To obtain the widest gradient for the limited space available, planar gradient coils are often designed with two dimensional stream functions (10,14,20,21), which, in turn, require more parameters to simulate and optimize as well as more calculation time when compared to optimizations using cylindrical stream functions.
In this work, wire patterns are first created on a cylindrical surface using a cylindrical stream function. These wire patterns are then transformed to the superelliptical geometry.

**Wire Pattern Design**

The stream function defines the current density flowing on the surface of a cylinder (9) as

$$j = \nabla \times [S(\phi, z) \cdot \hat{a}_r]$$

where, due to continuity, $\nabla \cdot j = 0$. $S(\phi, z)$ is an arbitrary stream function. With this stream function, we can derive an azimuthal component ($j_\phi$), and an axial component ($j_z$) of the cylindrical surface current density $j(z, \phi)$ as,

$$j_\phi = \frac{\partial}{\partial z} S(\phi, z) j_z = -\frac{1}{r} \frac{\partial}{\partial \phi} S(\phi, z).$$

To consider general current distributions, the stream function can be expressed as:

$$S(\phi, z) = h(z) \sum_{m=-\infty}^{\infty} v_me^{im\phi}$$

where $h(z)$ is an arbitrary function which describes the axial component of the stream function (9). For conventional cylindrical geometry, $m=1$ is used for the transverse gradient and $m=0$ for the axial gradient because of its axial symmetry. Finally, the resultant magnetic field can be calculated with this stream function by using the equations above (Eq [2] and Eq [3]), the Green’s function expansion (22), and the Fourier transform pair of the surface current density (9,14,20,23).

**Transformation to Superelliptical Geometry**

The wire patterns are defined on the cylinder surface based upon how they will be transformed to the superelliptic structure. In this paper, only the bottom half of the superelliptical cross section is fabricated, so only the wire pattern on the bottom half of the cylinder is transformed to the bottom half of the superellipse. For the y-gradient which needs two fingerprint wire patterns on the flat or superellipse surface, the stream function with $m=1$ is used. For the x-gradient, which needs four fingerprint winding patterns on the flat or superellipse surface, the stream function with $m=2$ is used:

$$S(\phi, z) = h(z) \cos(2\phi).$$

The top row of Figure 1 demonstrates y- and x-gradient winding patterns and their dipoles on the cylindrical surface before transformation to the superelliptical geometry. The figure is created with $S(\phi, z) = h(z) \cos(\phi)$ for the y-gradient, and with $S(\phi, z) = h(z) \cos(2\phi)$ for the x-gradient. The ‘x’ and ‘o’ mark indicate dipoles in and out of the surface alternatively.

Planar gradient systems have the inherent disadvantage of a relatively narrow linear homogeneous imaging volume (especially in the direction parallel to the gradient surface), despite the fact that they create such a strong local gradient field. To create a wider imaging volume, the width of the planar gradient has been increased and the edges have been bent up in a superelliptical shape which has geometry similar to a box with rounded corners. The entire winding pattern has been widened by moving the return paths of the planar gradient...
coil to the bent up edges as shown in Figure 1 (bottom row). Note that, in this design, the coil windings of the insert are on the bottom half of a superellipse-cylinder (former) placed under the patient as in a conventional planar geometry.

Once the stream function is set and the wire patterns on the cylindrical surface are determined, the cylinder-to-superellipse transformation is applied to convert wire patterns on the cylinder onto the superelliptical geometry. A typical superellipse is defined as,

$$|x/a|^m + |y/b|^n = 1,$$

where $a$ and $b$ are the semi-diameters, which define the aspect ratio of the superellipse and $m$ and $n$ are the parameters that determine the curvature of the superellipse. However, sampling points of the superellipse become coarsely spaced after the transformation for large $m$ and $n$.

To improve the uniformity of the sampling point spacing, a parametric form of the superellipse is used:

$$
\begin{align*}
  x(t) &= \cos t^{2/m} \cdot a \operatorname{sgn}(\cos t), \\
  y(t) &= \sin t^{2/n} \cdot b \operatorname{sgn}(\sin t),
\end{align*}
$$

(0 \leq t < 2\pi)

This parametric form can transform wire patterns on the cylindrical geometry to the superelliptical geometry without losing information on the cylinder (24,25). In this work, we use half (0 \leq \varphi < \pi) of the full cylindrical circumference to create uni-planar-like systems, and one can use a full (0 \leq \varphi < 2\pi) circumference to design bi-planar systems.

Methods

Field-Modification with an Extra Layer

The relatively small linear gradient region of the uni-planar gradient systems is caused by the limited width of the planar surface and the field drop-off along the $y$-axis as mentioned above. The magnetic field created by the uni-planar systems drops off approximately exponentially as a function of distance from the coil surface.

To increase the linear homogeneous region in all directions, an extra outer layer of current windings has been added to each of the primary superellipse gradient coil layers. The magnetic field of this extra layer is added to the magnetic field of the primary gradient coil to modify the profile of the resultant magnetic field. As a result, each axis is formed using two layers, a primary and a field modifying (FM) layer, resulting in a total of six layers for a complete (x-, y-, and z-axis) insert gradient coil.

Figure 2 shows the field modification process with the wire patterns and the magnetic field profiles of the $y$-gradient. Figure 2(a) and (b) show the primary and the field-modifying extra layers of the $y$-gradient coil where in this case the FM layer is formed as the 2nd harmonic ($\cos 2\varphi$) (with currents applied in the sense shown). By having four fingerprint windings (Figure 2(b)) rather than two windings as in typical $y$-gradient design (Figure 2(a)), and adding a separation, $\Delta_s$, between the wire patterns in the $z$-direction, a dip is created in the magnetic field profile (Figure 2(d)). After adding field profiles of the primary layer (Figure 2(c)) and the FM layer (Figure 2(d)), the resultant magnetic field yields a wider homogeneous imaging region than the primary winding alone.
The magnetic field from the FM layer strengthens the resulting magnetic field, as well as the gradient efficiency significantly. In a recent paper, Aksel et al (12) presented a six-layered local planar gradient coil with two layers with the identical wire patterns for each gradient axis to double the ampere-turns, which in turn, increased the gradient strength up to 170 mT/m, 230 mT/m, and 325 mT/m for the x-, y-, and z-coil (at 5 cm above coils, 320 A) respectively.

The winding patterns for the x-gradient layers are shown in Figure 3 (top row). Figure 3(a) shows the primary gradient layer and Figure 3(b) shows the FM layer in 3D view. The FM layer is obtained by replicating the primary layer and inserting a separation, $\Delta s$, along the z-axis. Non-zero $\Delta s$ creates a dip along the z-axis, which in turn, yields an extended homogeneous region along the z-axis after addition with the primary layer. For the z-gradient coil (Figure 3(c) and (d)), field modification is performed by an extra layer of windings that are similar to those of the primary y-gradient coil winding pattern, with the separation ($\Delta s$) in the middle along the z-axis. Since the wire pattern of the FM layer of the z-gradient (i.e. similar to the primary y-gradient) uses a different stream function than the primary z-gradient coil, two different stream functions were used to create these two layers for the simulation.

**Geometric Design**

In this study, the gradient set is designed to operate within a magnet bore diameter of 70 cm such as Siemens’ Magnetom Verio 3T (Siemens Medical Systems, Erlangen, Germany) or Toshiba’s Vantage Titan (Toshiba Medical Systems, Tokyo, Japan). The gradient system consists of one 6 cm thick superelliptical gradient plate below the patient. The proposed superelliptical system in this study has nine layers, six layers for gradient coil windings, and three cooling layers between the primary and FM layer of each axis.

To be practical for breast imaging, the target field-of-view (FOV) should be large enough to cover both breasts, and the space inside of the insert gradient has to be wide enough to accommodate a torso and both breasts for most female body habitus while being limited by the inner diameter of the magnet bore. As a reasonable design to meet these limitations, the proposed insert gradient design has 50 cm of inner width, 56 cm of outer width, and 80 cm in length with 20 cm of vertical sides on each end of the x-axis (Figure 4), and the minimum possible wire spacing is set to 7 mm. Note that the parameters in the equation [6] used for this particular design are $a=1$, $b=0.7143$, $m=6$, $n=6$.

**Simulation/Optimization Process**

Starting in the cylindrical geometry, wire patterns are first specified by a one dimensional stream function. Wire patterns for each axis in the cylindrical geometry are then mapped onto the superellipse surface. A functional form is chosen for the stream function so that it can be parameterized by a small number of coefficients. These coefficients can then be adjusted to find a desired optimal solution. In this study, the stream function is parameterized by a piecewise cubic hermite interpolating polynomial. The piecewise hermite interpolated stream function requires fewer control points and can create a wider range of useful stream function shapes than a simple power function (26). Further, the piecewise hermite interpolation creates smooth transitions leading to wire patterns without sharp wire bends for easier winding implementation in the actual gradient coil construction process. Figure 5 shows an example of a piecewise hermite polynomial interpolated stream function for a y-gradient coil design and its control points (circled points), which determine the shape of the stream function.
After the initial stream function is set, the cylindrical surface current density and exact wire pattern position for a regular cylindrical geometry are determined. Next, these cylindrical wire patterns are transformed to the desired superellipse geometry and the magnetic field map is calculated using the Biot-Savart law on the transformed superellipse wire pattern followed by our figure-of-merit (FOM) evaluation. Note that all the field maps from the wire patterns were simulated by using the Biot-Savart law (22):

$$B = \frac{\mu_0}{4\pi R^2} \int_\gamma Idl \times a_R$$

where $Idl$ is a small current segment of the wire pattern, $a_R$ is the unit vector directed from the source point to the field point (i.e. $a_R = \frac{\vec{R}}{R}$), and $R = |\vec{R}|$.

The cost function ($F$), which was to be minimized, was calculated based upon homogeneity ($\sigma_{rms}$), efficiency ($\eta$), inductance ($L$), and the maximum field near the edge of the coil surface for peripheral nerve stimulation considerations:

$$\text{cost function } (F) = \left( (\sigma_{rms})^{w_1} \cdot (L)^{w_2} \cdot (\max B_{0}\sigma_{5/6})^{w_3} \right) / (\eta)^{w_4},$$

$$\log(F) = w_1 \log(\sigma_{rms}) + w_2 \log(L) + w_3 \log(\max B_{0}\sigma_{5/6}) - w_4 \log(\eta).$$

Therefore,

$$F = \exp( w_1 \log(\sigma_{rms}) + w_2 \log(L) + w_3 \log(\max B_{0}\sigma_{5/6}) - w_4 \log(\eta)),$$

where $w_1, w_2, w_3, w_4$ are arbitrarily adjustable weighting factors for homogeneity, inductance, maximum field, and efficiency, respectively, that were chosen based upon the relative magnitude of each parameter and additional weighting by its importance. For the results in this paper, the values chosen were: $w_1=0.6$, $w_2=0.8$, $w_3=0.5$, $w_4=0.9$. The homogeneity and efficiency are defined as,

$$\sigma_{rms} = \sqrt{\frac{1}{V_{GFOV}} \int_{GFOV} (B_z(\vec{r}) - z \cdot \eta)^2 \, dx \, dy \, dz},$$

$$\text{and } \eta = \int_{GFOV} z B_z(\vec{r}) \, dx \, dy \, dz / \int_{GFOV} z^2 \, dx \, dy \, dz,$$

where $V_{GFOV}$ is the volume of the desired imaging FOV (GFOV), where GFOV is the desired imaging volume in 3D. FOVx, FOVy, and FOVz are defined as the width, height, and length of the GFOV respectively. Inductance values were calculated using Grover’s method (27).

For the optimization, an exhaustive search was performed, stepping through all combinations of all gradient design parameters, to find the optimized stream functions for both winding layers of each gradient axis in the given superelliptical geometry. The
exhaustive search method is superior in quality, even though it is time consuming, as long as the parameter sampling is not too coarse.

Since the parameters of the figure of merit are calculated in the superellipse geometry and the secondary FM layer has a separation in the middle of the z-axis, the separation factor ($\Delta s$) in the middle of the windings along the z-axis and parameters for the aspect ratio and curvature of the superellipse (i.e. $a, b, m, n$) are included in the optimization process.

The target FOV was specified, based upon a clinical breast imaging protocol, to be FOV$_x$ (left/right) = 38 cm, FOV$_y$ (anterior/posterior) = 19 cm, FOV$_z$ (superior/inferior) = 25 cm. For each FOM calculation, a line is fit to the generated true field through the middle of the axis and the slope of this line becomes the desired gradient. This gradient within the FOV is used during the optimization process to compare to the actual field generated. Each contour line of the gradient fieldmaps represented in this paper is a deviation from the desired magnetic field of the gradient. As an example of the x-gradient,

$$\text{Deviation}(\delta) = \frac{B_{x,\text{desired}} - B_{x,\text{true}}}{G_x - B_x(\alpha)} / G_{x,\text{FOV}}^x,$$

where $G_x$ is the gradient along the x-axis, and $B_{z,x}^x$ is the z component of the magnetic field from the x-gradient ($d B_z / dx$).

**Force and Torque Balance**

Due to the asymmetric characteristic of the uni-planar gradient system, force and torque balancing is important for realistic implementation and patient safety. In this study, we included net force and torque calculation as a limiting factor of the simulation results.

The net force calculation of the insert gradient is done by dividing the wire pattern into finite segments and using the Lorentz force equation (28):

$$\vec{F} = I \sum_{i=1}^{N} (\vec{d} \times \vec{B}_i)$$  \hspace{1cm} [14]

The torque on the insert gradient coil due to the external Lorentz force from the main magnetic field was calculated by (28):

$$\tau = I \sum_{i=1}^{N} (r_i \times (\vec{d} \times \vec{B}_i)).$$  \hspace{1cm} [15]

Both Eqs. [14] and [15] involve numerical integration over the entire wire pattern. Each wire segment of the insert gradient will contribute to the net force, and the force on each wire segment will contribute to the torque. The main magnetic field is simulated by a simplified design presented by Cheng (29). The simplified model has a diameter of spherical volume (DSV) of 30 cm, an inhomogeneity of 272 ppm, and a 3 T main magnetic field with a cold bore length and diameter of 1m. The current flowing through the insert gradient was scaled to 300 A to represent a practical setting. The mass of the insert coil was estimated by finding an approximate volume of the potted insert coil, and the density of a thermally conductive epoxy of 2 g/cm$^3$. The mass of the insert gradient coil was estimated to be 127 kg.
Construction

Local insert gradient systems can have eddy current problems when they are placed in close proximity to the metal of the magnet warm bore. To test for the evidence of eddy current effects on this gradient design, a prototype of the y-axis of the proposed gradient coil was constructed. It is expected that the y-axis will have the greatest interaction with the warm bore, and would therefore demonstrate the greatest magnitude of eddy current effects. The dimensions of this prototype y-axis were designed to sit as closely as possible to the warm bore of a 60cm bore Siemens TIM Trio, to provide the greatest chance to observe eddy current effects. Dimensions of the prototype are 48 cm (inner width), 51 cm (outer width), and 80 cm in length with 20 cm of vertical walls.

MRI Imaging

All MRI scans were performed on the Siemens 3Tesla TIM Trio scanner (Siemens Health Care AG, Erlangen Germany). The standard system was augmented with three additional gradient amplifiers and master/slave configured computers capable of controlling extra prototype y-axis gradient. The control hardware and software were developed and provided by Siemens. Pulse sequences were implemented to control both gradient coils synchronously. A separate pulse sequence was used to operate each gradient set. For our experiments, the master computer was used to 1) maintain all computer controlled shims, including the first order gradient shims obtained using the standard body gradients, 2) control RF excitation and reception, 3) control standard whole body imaging gradients to allow any combination of x, y, z to be on or off, and 4) synchronously trigger the slave computer. The slave computer executed a pulse sequence controlling the prototype y axis gradient to allow the y gradient to be on or off. In this work the Y- gradient was used as the readout or phase encoding gradient.

The water-filled sphere phantom was scanned with using the body RF coil and the prototype for the y-gradient and the system body gradient for the x- and z-gradients with a Gradient Echo (GRE) sequence with the following parameters; imaging matrix=256×256, TR=100ms, TE=5ms, BW=320 Hz/pixel, FOV=300mm, slice thickness=5.5mm, dist factor=200%.

RESULTS

Figure 6 compares the gradient field maps (contours are at 5% deviation from the desired field of the gradient) for each gradient axis between the conventional planar and the superelliptical geometries, demonstrating that the superellipse geometry allows for a wider homogeneous gradient region than the conventional planar geometry. An identical wire pattern on the full cylinder (0 ≤ φ < 2π) is transformed to the conventional planar geometry and to the superelliptical geometry separately, and then fieldmaps are acquired from the transformed wire patterns for each case. The left column of Figure 6 (i.e. Figure 6 (a), (c), and (e)) are the results from the planar gradients and the right column of Figure 6 (i.e. Figure 6 (b), (d), and (f)) displays the results from the superelliptical transformation.

The cylinder-to-conventional-planar transformation is performed with the transform published previously (26), and the cylinder-to-superellipse transformation is performed with the transformation described by Equation [6] with a=1, b=0.7143, m=6, n=6.

It is well known that the linear gradient regions of the planar gradient system are very small as shown in the left column of Figure 6. However, this result demonstrates that the superellipse design can create a wider (left/right) homogeneous imaging region than the conventional planar design. The 10% homogeneous gradient region in the transverse plane...
of the superellipse geometry is increased by 182% for the x-gradient design, by 57% for the y-gradient design, and by 75% for the z-gradient design (See Table 1).

Gradient field maps with and without the field modification layer are shown in Figure 7. The linear gradient region after the field modification is extended notably compared to the linear gradient region of the primary layer alone. The extra field-modifying layers increase the HGV by 67% (in the transverse plane) for the x-gradient coil, by 89% (in the transverse plane) for the y-gradient coil, and by 214% (in the coronal plane) for the z-gradient coil when combine with the already extended superelliptical gradient field.

Figure 8 shows contour plots of the z-component of the magnetic field for each axis of the gradient coil within the target FOV. The figure indicates that the x-gradient coil design reaches up to 300 mT/m at a distance of 7 cm from the surface of the coil, the y-gradient coil reaches up to 217 mT/m, and the strength of the z-gradient coil reaches 173 mT/m at 7 cm off the coil surface, assuming a current of 300 A. For the unit current, the efficiency values are 1.0 mT/m/A for the x-gradient coil, 0.723 mT/m/A for the y-gradient coil, and 0.577 mT/m/A for the z-gradient coil. These results clearly show extended HGVs compared to the results presented by Aksel et al. for an optimized flat gradient insert (see Figure 3 in (12)). The gradient efficiencies for the superelliptical design show similar efficiencies, but lower efficiency for the z-gradient coil.

Gradient fieldmaps from the superellipse gradient including both the primary and FM windings, with contour lines at 10% deviation from the desired field of the gradient, are shown in Figure 9. Superimposed on each figure are dashed lines representing contours of a female breast. The extent of the HGV for each gradient for 5%, 10% and 20% deviations are given in Table 2. From these numbers, the minimum imaging extent in x, y, and z is 29.6 cm, 19.0 cm, and 26.2 cm, respectively. These are close to the design parameters: 38 cm, 19 cm, and 25 cm.

Force and torque were calculated using Equations [14] and [15]. The force-safe region was defined as those locations where the net force on the insert gradient coil is below 10% of the force of gravity. Similarly, the torque-safe region represents the locations where the net torque on the insert gradient coil is less than 10% of the force of gravity applied at a distance of 1 m to the centre of mass of the insert gradient system. The intersection of these two regions gives a final safe volume of approximately 46 cm in x-, 46 cm in y-, and 65 cm in z-axis respectively at the iso-center.

With the threshold being a fraction of the force of gravity, the insert gradient coil should not lift, or rotate off the bed of the main magnet as long as the center of the insert gradient coil is positioned inside the safe region. This calculation includes consideration of failure modes including partial and complete failure of a fingerprint winding section, and miss-alignments such as a 5 degree rotation along the x- and/or y-axis.

Figure 10(a) shows the manufactured y-axis prototype and its loading harness. The axial images of the water-filled sphere phantom using the prototype are shown in Figure 10(b).

Eddy currents induced by this gradient design should not be a problem when using GRE DCE (30). GRE images acquired with the Y gradient superellipse insert have demonstrated that the B0 shift is less than a pixel when using 320 Hz/pixel bandwidths when phase and frequency directions are swapped at all possible echo times. We anticipate that pulse sequences more prone to eddy current artifacts such as DWI EPI will be possible with ECC since EPI ghosting artifacts we see when using the Y superellipse insert are no worse than those of the systems Y gradient when ECC is disabled.
DISCUSSION

The goal of this study was to design a bilateral breast insert gradient for faster imaging capability, high gradient efficiency with a homogeneous imaging volume large enough for clinical imaging of both breasts. Two new concepts applied in this design included the transformation to a superellipse for improved wire space, and the addition of a field modifying layer to allow variations in wire patterns to improve the extent of the gradient HGV for all axes. Transformation from planar to the superelliptical geometry for the local insert gradient resulted in a substantial increase (50% to 100%) in the dimensions of the HGV. The simulation results also demonstrate that a second field-modifying (FM) winding layer can be used to further extend the homogeneous gradient volume.

This paper used the transformation of winding patterns from a cylinder to the superellipse to allow optimization of a one dimensional stream function rather than the two dimensional stream function required for a two dimensional current distribution. Although this gives up some generality in definition of the current distribution, it is much faster and provided current distributions that work well for our purpose. Further, the field modifying layer was specified and optimized using a few additional parameters, and the addition of this layer further improved the imaging HGV. It remains to be shown how much improvement would be obtained by specifying current distributions more generally.

Although it was not possible to construct the full 6 layer coil for testing, we were able to construct a prototype superelliptical y-gradient coil that was positioned on the magnet rails, as close as possible to the magnet bore. Experiments using this insert gradient demonstrated no measureable effects due to eddy currents. Although this does not totally rule out the possibility that eddy currents may ultimately need to be addressed, it is likely that the y-axis will have the greatest interaction with the metal of the main magnet and therefore this does provide strong evidence that eddy currents can be dealt with adequately using standard techniques.

In conclusion, the small homogeneous gradient volume limitations of conventional planar gradient coil designs have been improved with the use of a superellipse insert gradient design. The superelliptical gradient characteristics have been further improved by implementing an additional field modifying current winding layer for each gradient axis. As a result, the homogeneous imaging volumes of the field-modified superelliptical insert gradient can be increased to a volume that should work well for bilateral breast MRI.

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REFERENCES


Figure 1.
Gradient coil wire patterns before and after the cylinder-to-superellipse transformation. (**Top row**) Wire patterns on the cylindrical surface acquired with $S(\phi, z) = h(z) \cos(\phi)$, for the $y$-gradient (a), and with $S(\phi, z) = h(z) \cos(2\phi)$ for the $x$-gradient (b). The ‘x’ and ‘o’ mark indicate dipole direction for each winding. (**Bottom row**) The transformed superellipse wire patterns of the $y$-gradient (c) and $x$-gradient (d).
Figure 2.
Y-gradient coil field-modification (FM). (a) primary y-gradient wire pattern, (b) wire pattern for the FM layer of the y-gradient, (c) field profile of the primary y-gradient coil, (a), along the dotted line in the middle, (d) field profile of the FM layer, (b), along the dotted line in the middle, (e) the resultant field profile after combining (c) and (d).
Figure 3.
X- and z-gradient coil field-modification (FM). (Top row) primary x-gradient coil wire pattern (a), and FM layer winding patterns (b) of the x-gradient coil. (Bottom row) primary z-gradient coil winding pattern (c) and FM layer (d) of the z-gradient coil.
Figure 4.
Superellipse insert gradient system geometry. (a) 3D view, (b) axial view, and (c) superellipse geometry specifications.
Figure 5.
A piecewise cubic hermite interpolating polynomial stream function for the y-gradient design. Circles represent control points. (Units are nominal)
Figure 6.
Gradient map (5% deviation contours) comparison between the planar and superelliptical geometry transformed from the same cylindrical wire pattern demonstrating that the superellipse geometry creates a wider HGV along the x-axis than the conventional planar geometry. (Top row) x/y view of the x-gradient: (a) flat and (b) superelliptical geometry, (Middle row) x/y view of the y-gradient: (c) flat and (d) superelliptical geometry, (Bottom row) x/z view of the z-gradient at 7 cm off the coil surface: (e) flat and (f) superelliptical geometry. All units are expressed in cm.
Figure 7.
Gradient fieldmap (5% deviation contours) comparisons of before and after the field-modification for each axis. (Top row) x-gradient comparison (x/y view): (a) before and (b) after the field-modification, (Middle row) y-gradient comparison (x/y view): (c) before and (d) after the field-modification, (Bottom row) z-gradient comparison (x/z view at 7 cm off the coil surface): (e) before and (f) after the field-modification. All units are expressed in cm.
Figure 8.
Magnetic fieldmaps of each gradient coil (in mT at 300A) within the target FOV. (Top row) $B_x^1$, from the x-gradient coil (a) x/z view at 7 cm off the coil surface (b) x/y view at z=0 cm (i.e., center), (Middle row) $B_y^2$, from the y-gradient coil (c) x/y view at z=0 cm, (d) y/z view at x=0 cm, (Bottom row) $B_z^3$, from the z-gradient coil (e) x/z view at 7 cm off the coil surface (f) y/z view at x=0 cm.
Figure 9.
The resulting gradient fieldmaps including field modification. (Top row) x-gradient results: (a) x/y-axis view (b) x/z-axis view, (Middle row) y-gradient results: (c) x/y-axis view (d) y/z-axis view, (Bottom row) z-gradient results: (e) x/z-axis view (f) y/z-axis view. Each contour line represents 10% deviation from the desired gradient. Superimposed on each figure are dashed lines representing contours of a female breast. All units are expressed in cm.
Figure 10.
(a) Y-axis only superelliptical gradient insert prototype and (b) phantom images (axial).
Table 1

Imageable area in the transverse plane (x/y) comparison data, based on 10% contours of deviation from the desired gradient field for both conventional uni-planar and superellipse gradient coils.

<table>
<thead>
<tr>
<th>Gradient</th>
<th>Linear region (10%) (x- / y-axis in cm)</th>
<th>Imageable region (cm$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>x-gradient</td>
<td>Planar 20.0 / 8.5</td>
<td>170</td>
</tr>
<tr>
<td></td>
<td>Superellipse 24.0 / 17.0</td>
<td>408</td>
</tr>
<tr>
<td>y-gradient</td>
<td>Planar 19.2 / 12.0</td>
<td>230.4</td>
</tr>
<tr>
<td></td>
<td>Superellipse 30.5 / 11.9</td>
<td>363</td>
</tr>
<tr>
<td>z-gradient</td>
<td>Planar 11.0 / 7.5</td>
<td>82.5</td>
</tr>
<tr>
<td></td>
<td>Superellipse 18.0 / 8.0</td>
<td>144</td>
</tr>
</tbody>
</table>
Table 2

Field-modified superelliptical insert gradient performance measures. The inductance values include mutual and self inductance for both primary and field-modifying layers, and the efficiencies for the x- and z-gradients are measured at 7 cm off from the surface of the coil.

<table>
<thead>
<tr>
<th></th>
<th>X-gradient</th>
<th>Y-gradient</th>
<th>Z-gradient</th>
</tr>
</thead>
<tbody>
<tr>
<td>Efficiency (mT/m/A)</td>
<td>1.0</td>
<td>0.723</td>
<td>0.577</td>
</tr>
<tr>
<td>Inductance (µH)</td>
<td>2979</td>
<td>2050</td>
<td>1810</td>
</tr>
<tr>
<td>HGV (5%, x-/y-/z-axis in cm)</td>
<td>20.3 / 11.0 / 34.2</td>
<td>26.3 / 13.8 / 12.6</td>
<td>9.0 / 8.0 / 27.0</td>
</tr>
<tr>
<td>HGV (10%, x-/y-/z-axis in cm)</td>
<td>25.0 / 15.8 / 46.0</td>
<td>32.0 / 15.0 / 15.2</td>
<td>23.0 / 13.0 / 30.0</td>
</tr>
<tr>
<td>HGV (20%, x-/y-/z-axis in cm)</td>
<td>29.6 / 23.8 / 54.0</td>
<td>43.0 / 22.0 / 26.2</td>
<td>37.0 / 19.0 / 32.0</td>
</tr>
</tbody>
</table>