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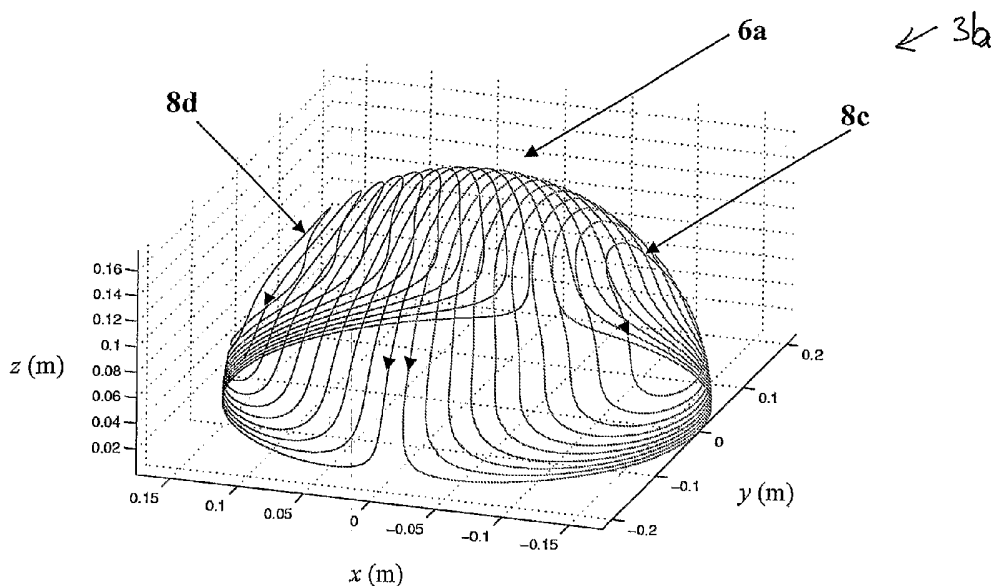
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- (71) Applicant (for all designated States except US): **THE UNIVERSITY OF NOTTINGHAM** [GB/GB]; University Park, Nottingham NG7 2RD (GB).
- (72) Inventors; and
- (75) Inventors/Applicants (for US only): **BOWTELL, Richard** [GB/GB]; School of Physics and Astronomy, The University of Nottingham, University Park, Nottingham NG7 2RD (GB). **GREEN, Dan** [GB/GB]; 1 Kedlestone Close, East Hunsbury, Northampton NN4 0WF (GB). **LEGGETT, James** [GB/GB]; "Greenacre", Tompkin Lane, Stanley, Stoke-on-Trent, Staffordshire ST9 9LX (GB).
- (74) Agents: **HIGGIN, Paul** et al.; Swindell & Pearson, 48 Friar Gate, Derby DE1 1GY (GB).
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(54) Title: AN APPARATUS FOR PROVIDING A MAGNETIC FIELD GRADIENT SUITABLE FOR MAGNETIC RESONANCE IMAGING



(57) Abstract: An arrangement of conductive loops for providing a linearly varying magnetic field in a first direction across an imaging region of a magnetic resonance imaging system, wherein the conductive loops are collectively arranged to define a dome.

TITLE

An apparatus for providing a magnetic field gradient suitable for magnetic resonance imaging.

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FIELD OF THE INVENTION

Embodiments of the present invention relate to an arrangement of conductors for providing a magnetic field that varies linearly in a first direction. The arrangement is suitable for, but not necessarily exclusively for, providing a magnetic field that varies linearly in a first direction across an imaging region of a magnetic resonance imaging system. It may for example, be used in other nuclear magnetic resonance applications.

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BACKGROUND TO THE INVENTION

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It is well known that in order to perform magnetic resonance imaging (MRI), it is necessary to provide magnetic field gradients in three, orthogonal spatial directions so as to encode the positions of the sources of magnetic resonance (MR) signals. The encoded MR signals are processed to form a magnetic resonance image.

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A magnetic field gradient is produced by passing electric currents through conducting wires that are arranged on a surface surrounding the subject to be imaged. Three such wire arrangements, known as the *gradient coils*, are needed to produce an image. Each of these three coils generates a linear variation of one vectorial component of the magnetic field, along one of the three orthogonal Cartesian axes, over a region of the subject to be imaged. This means that in this region, for example, the z-component of the magnetic field, B_z , is proportional to a constant, G , known as the gradient strength, multiplied by the positional co-ordinate, x , y or z . This is illustrated in Figure 1, which depicts the field variation produced by an x -gradient coil, such that $B_z = Gx$.

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In current practice, the magnetic field gradient coils are often wound on a plurality of cylinders that are arranged to be co-axial with the cylindrical bore of the magnet, as

shown in Figure 2, and which have an aperture that is large enough to accommodate the human body.

5 It is well known that, the speed of imaging and the quality of the MR images that are produced is improved by the use of gradient coils, in which the current in the coil can be changed rapidly in time (requiring gradient coils to have low inductance), and in which the strength of the gradient can be made large. It is also well known that it is advantageous for the electrical resistance of the gradient coil to be low, so that the Joule heating of the coil is limited. In addition, it is known that the oscillating Lorenz
10 forces experienced by the current carrying wires of the gradient coil lead to the production of acoustic noise that can be uncomfortable for the imaging subject and those operating the imaging system. It is thus beneficial to limit the average root mean square force acting on the coil. In addition, it is known that if the human body is subjected to large rates of change of magnetic field with time in areas of large cross
15 section, the resulting induced electric fields can cause uncomfortable nerve stimulation in the subject. It is thus beneficial to limit the exposure of the human torso to rapidly varying magnetic flux.

The above factors mean that it is advantageous to employ gradient coils in which the
20 current-carrying wires are brought close to the region of imaging. In particular for imaging the human head only, it has been shown that significant gains in relation to the above performance factors ((i) inductance; (ii) resistance; (iii) propensity for nerve stimulation; (iv) levels of acoustic noise generated) can be achieved by using gradient coils of reduced size, which are wound on a cylinder of diameter large enough to
25 accommodate the human head, but not the human body (Bowtell, R. et al, Analytic approach to the design of transverse gradient coils with co-axial return paths. Magnetic Resonance in Medicine, 1999. 41(3): p. 600-608)

30 However, the presence of the shoulders limits the length of such coils, which in turn reduces the gains in performance that can be achieved through reducing the coil diameter. To overcome this problem, asymmetric, cylindrical coil designs in which the axial location of the region where the magnetic field varies linearly with position

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is shifted towards one end of the coil, have been developed. In the case of coils which generate an x - or y -gradient it is necessary to modify such asymmetric coil designs to make the net torque on the coil zero, thus ensuring that such coils will not be driven to rotate in the magnet (Tomasi, D., et al., Asymmetrical gradient coil for head imaging. 5 Magnetic Resonance in Medicine, 2002. 48(4): p. 707-714; Alsop, D.C. et al, Optimization of torque-balanced asymmetric head gradient coils. Magnetic Resonance in Medicine, 1996. 35(6): p. 875-886.)

Even with the current availability of asymmetric, head-sized, cylindrical coils, it 10 would be desirable to improve gradient coil performance.

BRIEF DESCRIPTION OF THE INVENTION

According to one embodiment of the invention there is provided an arrangement of 15 conductive loops for providing a linearly varying magnetic field in a first direction across an imaging region of a magnetic resonance imaging system, wherein the conductive loops are collectively arranged to define a dome.

The dome shape allows the conductive loops to be brought close to the patient's head, 20 which improves performance.

According to another embodiment of the invention there is provided a magnetic resonance imaging system comprising an arrangement of conductive loops for providing a magnetic field that varies linearly in a direction across an imaging region 25 within a patient's head, wherein the conductive loops are collectively arranged to define a surface that overspreads at least a crown of a patient's head.

The overspreading surface allows the conductive loops to be brought close to the patient's head, which improves performance.

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According to another embodiment of the invention, a gradient coil is formed from wires arranged on a surface including a dome-shaped portion. Incorporation of a

dome-shaped surface, allows the wires of the gradient coil to be brought into closer proximity to the head, thus yielding substantial gains in gradient coil performance. In particular such coils allow larger gradients to be generated at fixed power dissipation and larger rates of change of gradient with time to be achieved at fixed amplifier power. In addition, gradient coils incorporating a dome-shaped surface produce lower levels of acoustic noise for a given gradient waveform, and weaker time-varying fields in the human torso.

The dome-shaped surface is preferably formed from a hemisphere, or other portion of a spherical surface, although other shapes such as a part-ellipsoid may also be employed. The dome-shaped surface may be attached to an extension, which reaches down to the shoulders. Wires forming the gradient coils also spread onto this surface. This surface extension may be cylindrical in form, with a circular cross-section, for use in conjunction with a part-spherical dome, or with an elliptical cross-section, for use with a part-ellipsoidal shaped dome. In other circumstances it may be flared in shape, so as to have a larger aperture than that formed by the opening to the dome.

According to another embodiment of the invention there is provided a method of manufacturing an arrangement of conductive loops for providing a linearly varying magnetic field in a first direction across an imaging region of a magnetic resonance imaging system, comprising: determining the coefficients of a stream function for a dome shape so that the inductance defined by Equation 6 and the resistance defined by Equation 7 are simultaneously minimized while the magnetic field is constrained to vary linearly in the first direction; and arranging conductive loops along the contours of the stream function.

According to another embodiment of the invention there is provided a method of manufacturing an arrangement of conductive loops for providing a linearly varying magnetic field in a first direction across an imaging region of a magnetic resonance imaging system, comprising: determining the coefficients of a stream function for a dome shape so that the applied torque in use is zero while the magnetic field is

constrained to vary linearly in the first direction; and arranging conductive loops along the contours of the stream function.

5 BRIEF DESCRIPTION OF THE DRAWINGS

For a better understanding of the present invention reference will now be made by way of example only to the accompanying drawings in which:

10 *Figure 1* shows the Cartesian x -, y - and z -axes and the linear variation of the z -component of the magnetic field, B_z , with x -position, as would be generated at the centre of an x -gradient coil;

Figure 2 shows the conventional coil arrangements used in a magnetic resonance imaging system;

15 *Figures 3A, 3B and 3C* show schematic diagrams (in perspective and plan views) of dome-shaped surfaces that may be used to support gradient coils in embodiments of the present invention;

Figures 4A, 4B and 4C show schematic diagrams (in perspective and plan views) of dome-shaped surfaces with cylindrical extensions that may be used to support gradient coils in embodiments of the present invention;

20 *Figure 5* shows a schematic diagram (in perspective and plan view) of dome-shaped surfaces, with a flared extension that may be used to support gradient coils in embodiments of the present invention;

Figure 6 shows the meaning of the conventional spherical polar co-ordinates, r , θ and ϕ , in relation to the Cartesian co-ordinates, x , y and z ;

25 *Figure 7* shows an axial gradient coil that is supported as the surface of a hemisphere and that generates a linear variation of the z -component of the magnetic field with z -co-ordinate;

Figure 8 shows the variation of the magnetic field, B_z , generated in the central x - z plane by an electric current in the axial gradient coil of *Figure 7*;

30 *Figure 9* shows a transverse gradient coil that is supported as the surface of a hemisphere and that generates a linear variation of the z -component of the magnetic field with x -co-ordinate;

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Figure 10 shows the variation of the magnetic field, B_z , generated in the central x - z plane by an electric current in the transverse gradient coil of *Figure 9*;

Figure 11 shows a transverse gradient coil supported as the surface of a hemisphere that generates a linear variation of the z -component of the magnetic field with x -coordinate and that is designed so that there is no net torque on the coil when it carries current in a uniform, z -directed, magnetic field;

Figure 12 shows the variation of the magnetic field, B_z , generated in the central x - z plane by an electric current in the transverse gradient coil of *Figure 11*;

Figure 13 shows an axial gradient coil that is supported on the surface of a hemisphere extended with a cylindrical extension and that generates a linear variation of the z -component of the magnetic field with z -co-ordinate; and

Figure 14 shows a transverse gradient coil that is supported on the surface of a hemisphere extended with a cylindrical extension and that generates a linear variation of the z -component of the magnetic field with x -co-ordinate.

DETAILED DESCRIPTION OF EMBODIMENTS OF THE INVENTION

Figure 1 – shows the Cartesian x -, y - and z -axes and the linear variation of the z -component of the magnetic field, B_z , with x -position, as would be generated at the centre of an x -gradient coil.

Figure 2 – shows the conventional coil arrangements used in a magnetic resonance imaging system 100. The system 100 comprise: a superconducting magnet 1; a set of shimming coils 2, gradient coils 3, a radio-frequency coil 4 and a patient bed 5. The gradient coils 3 are wound on the surface of a plurality of cylinders, located inside the cylindrical bore of the superconducting magnet 1. The diameter of the innermost cylinder forming the gradient coils is large enough to allow access for the human torso, which also fits inside the radio-frequency (RF) coil 4. The set of shimming coils 2 are often positioned on a cylinder surrounding the gradient coils. The subject of the imaging experiment is moved along the z -axis inside the aperture of the imaging system 100 on a patient bed 5.

- Figure 3A, 3B and 3C* show schematic diagrams (in perspective and plan views) of dome-shaped surfaces 6 that may be used to form gradient coils in the present invention. *Figure 3A* illustrates a hemispherical surface 6a in perspective and plan views. This hemispherical dome-shaped surface 6a appears as a circle in the plan view. *Figure 3B* illustrates a smaller portion 6b of a spherical surface in perspective and plan views. This dome-shaped surface 6b also appears as a circle in plan view. *Figure 3C* illustrates a portion 6c of an ellipsoidal surface in perspective and plan views. This dome-shaped surface appears as an ellipse in plan view.
- Figure 4A, 4B and 4C* show schematic diagrams (in perspective and plan views) of extended dome-shaped surfaces 6 that may be used to form gradient coils in the present invention. The dome-shaped surfaces are extended using cylindrical extensions 7.
- Figure 4A* illustrates a hemispherical surface 6a with a cylindrical extension 7a in perspective and plan views. The cylindrical extension has a circular cross-section that is matched to that of the hemispherical surface 6a. The extended dome-shaped surface appears as a circle in the plan view.
- Figure 4B* illustrates a smaller portion 6b of a spherical surface with a cylindrical extension 7a in perspective and plan views. The cylindrical extension has a circular cross-section that is matched to that of the smaller portion 6b. The extended dome-shaped surface 6b appears as a circle in plan view.
- Figure 4C* illustrates a portion 6c of an ellipsoidal surface with a cylindrical-like extension 7b in perspective and plan views. The cylindrical-like extension has an ellipse cross-section that is matched to that of the portion 6c. The extended dome-shaped surface appears as an ellipse in the plan view.
- Figure 5* shows a schematic diagram (in perspective and plan view) of an extended dome-shaped surface 6a that may be used to form gradient coils. The dome-shaped surface is extended using a flared extension 7c. The flared cylindrical extension 7c has

a circular cross-section that is matched to that of the hemispherical surface 6a where they meet. The flared cylindrical extension may have an increasing circular cross section, whose diameter is larger at the coil-aperture than at the junction with the hemisphere e.g. if the diameter of the circular cross section increases linearly with distance the flared extension has a frusto-conical shape.

Figure 6 shows the meaning of the conventional spherical polar co-ordinates, r , θ and ϕ , in relation to the Cartesian co-ordinates, x , y and z .

Figure 7 shows an axial gradient coil 3a wound on the surface of a 35 cm diameter hemispherical surface 6a. This coil 3a generates a linear variation of the z -component of the magnetic field with z -co-ordinate. The coil 3a is composed of loops of wire. A first set 8a of the loops are connected in series (not shown) and, in use, an electric current circulates through these adjacent loops in a clockwise sense. A second set 8b of the loops are connected in series (not shown) and, in use, an electric current circulates through the adjacent loops in a anti-clockwise sense. Each of the loops of wire in the first and second sets encircles an imaginary line that runs parallel to the z -axis.

Figure 8 shows the variation of the magnetic field, B_z , generated in the central x - z plane by a current flow of 1 A in the coil 3a shown in Figure 7. The surface of the hemisphere is denoted as 6a. Contours 9 of the magnetic field are shown at 5 μ T intervals. The thin-line, 10, denotes the 11-cm-diameter, spherical volume within which a linear magnetic field variation is required. The shading, 11, indicates the region within which the field deviates by more than 5 % from linearity.

Figure 9 shows a transverse gradient coil 3b wound on the surface of a 35 cm diameter hemispherical surface 6a. This coil 3b generates a linear variation of the z -component of the magnetic field with x -co-ordinate. The coil 3b is composed of a first set of distributed looped wire paths 8c at a first portion of the surface 6a and a second set of distributed looped wire paths 8d at a second portion of the surface 6a. The first set of distributed looped wire paths 8c are connected in series (not shown) and, in use,

an electric current circulates through the adjacent loops in an anti-clockwise sense. The second set of distributed looped wire paths 8d are connected in series (not shown) and, in use, an electric current circulates through the adjacent loops in a clockwise sense. Each of the loops of wire of the first and second sets surrounds an imaginary
5 line that runs parallel to the x-axis.

Figure 10 shows the variation of the magnetic field, B_z , generated in the central x-z plane by a current flow of 1 A in the coil 3b shown in *Figure 9*. The surface of the hemisphere is denoted as 6a. Contours 9 of the magnetic field are shown at 5 μ T
10 intervals. The thin-line, 10, denotes the 11-cm-diameter, spherical volume within which a linear magnetic field variation is required. The shading, 11, indicates the region within which the field deviates by more than 5 % from linearity.

Figure 11 shows a transverse gradient coil 3c wound on the surface of a 35 cm
15 diameter hemispherical surface 6a. This coil generates a linear variation of the z-component of the magnetic field with x-co-ordinate. It is designed so that there is no net torque on the coil when it carries current in a uniform, z-directed, magnetic field. The coil 3c is composed of separated first and second sets of distributed looped wire paths located upon the upper portion of the hemispherical surface and third and fourth
20 sets of distributed looped wire paths located upon the lower portion of the hemispherical surface. The first set of looped wire paths 8c are connected in series (not shown) and the electric current circulates in an anti-clockwise sense. The second set of looped wire paths 8d are connected in series (not shown) and the electric current circulates in a clockwise sense. The third set of looped wire paths 8e are
25 connected in series (not shown) and the electric current circulates in a clockwise sense. The fourth set of looped wire paths 8f are connected in series (not shown) and the electric current circulates in an anti-clockwise sense.

The first and second sets 8c, 8d of the distributed looped wire paths located upon the
30 upper portion of the hemispherical surface 6a surround a first imaginary line that runs parallel to the x-axis. The third and fourth 8e, 8f distributed looped wire paths located upon the lower portion of the hemispherical surface 6a surround a second imaginary

line that runs parallel to the x-axis. The first and second imaginary lines lie in the same y-z plane.

5 *Figure 12* – shows the variation of the magnetic field, B_z , generated in the central x-z plane by a current flow of 1 A in the coil 3c shown in *Figure 11*. The surface of the hemisphere is denoted as 6a. Contours (9) of the magnetic field are shown at 5 μ T intervals. The thin-line, 10, denotes the 11-cm-diameter, spherical volume within which a linear magnetic field variation is required. The shading, 11, indicates the region within which the field deviates by more than 5 % from linearity.

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Figure 13 – shows an axial gradient coil 3d wound on the surface of a 35 cm diameter hemispherical surface 6a, and a 35 cm diameter cylindrical extension 7a with a circular cross-section. This coil 3d generates a linear variation of the z-component of the magnetic field with z-co-ordinate. The coil 3d is composed of a first set 8a of circular loops of wire on the hemisphere in which the current circulates in a clockwise sense and a second set 8g of circular loops on the cylinder in which the current circulates in an anti-clockwise sense 8g. Each of the loops of wire encircles an imaginary line that runs parallel to the z-axis.

20 *Figure 14* – shows a transverse gradient coil 3e wound on the surface of a 35 cm diameter hemispherical surface 6a, and a 35 cm diameter cylindrical extension 7a with a circular cross-section. This coil 3e generates a linear variation of the z-component of the magnetic field with x-co-ordinate. The coil 3e is composed of first 8c and second 8d sets of looped wire paths on the hemispherical surface 6a in which the current circulates in respective anti-clockwise and clockwise senses and similar third and fourth sets of wirepaths (8h and 8i) on the cylindrical surface. The first and second sets 8c, 8d of distributed looped wire paths located upon the hemispherical surface 6a surround a first imaginary line that runs parallel to the x-axis. The third 8h and fourth 8i distributed looped wire paths located upon the cylindrical extension 7a surround a second imaginary line that runs parallel to the x-axis. The first and second imaginary lines lie in the same y-z plane.

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It is of course possible to use one or more domed gradient coils as described above in an MRI system 100. For example, a first dome-shape of conductive loops (wirepaths) may be used as an x-direction transverse gradient coil. A second dome-shape of conductive loops (wirepaths) may be used as a y-direction transverse gradient coil. A third dome-shape of conductive loops (wirepaths) may be used as a z-direction axial gradient coil. The first, second and third domes of conductive loops may be arranged as nested domes. For example, there may be three separate hemispherical layers of wires of decreasing radii. The hemisphere of greatest radius defines a recess that receives a hemisphere of smaller radius. The hemisphere of smaller radius defines a recess that receives the hemisphere of smallest radius.

It is of course possible to use more than one dome-shape of conductive loops (wirepaths) to create a magnetic field gradient in a particular direction. The multiple dome-shaped conductive loops may be arranged as a nest of domes.

The described gradient coil(s) may be actively-screened so that the field generated outside the coil is very small thus reducing interactions with external conductors.

The following description focuses upon an embodiment of the invention, in which the dome-shaped region is formed from a hemispherical surface. Other embodiments in which the dome-shaped surface is formed from an alternatively sized portion of a spherical surface or an alternative shape, such as a part ellipsoid are also possible, as shown in Figures 3, 4 and 5.

To produce a gradient coil, it is necessary to identify the current distribution, $\underline{J}$, needed on the dome-shaped surface to ensure that an appropriate field gradient is produced. This can be achieved using an expression, which describes the magnetic field, B_z , generated at all points inside a sphere, by a current distribution spread over a full spherical surface of radius, a . The current density can be defined via a stream function:

$$\psi(\theta, \phi) = \sum_{l=1}^{\infty} \sum_{m=-l}^l K_{lm} Y_{lm}(\theta, \phi) \quad [1]$$

such that

$$\underline{J}(\theta, \phi) = a^{-1} \left(\hat{\phi} \frac{\partial}{\partial \theta} - \hat{\theta} \frac{1}{\sin \theta} \frac{\partial}{\partial \phi} \right) \psi(\theta, \phi) . \quad [2]$$

Here, K_{lm} are constants and $Y_{lm}(\theta, \phi)$ are the spherical harmonics, while θ and ϕ are the usual spherical polar co-ordinates (Fig. 6). The magnetic field inside the sphere is then given using Maxwell's Equations by:

$$B_z(r, \theta, \phi) = a^{-1} \sum_{l=1}^{\infty} \sum_{m=0}^l \left(\frac{r}{a} \right)^{l-1} \left\{ \sqrt{\frac{(2-\delta_{m,0})(l-m)!}{(l+m)!}} \right\} (l+m) P_{l-1,m}(\cos \theta) \times [g_l^m \cos \phi + h_l^m \sin \phi] \quad [3]$$

where $P_{l,m}(x)$ represents an associated Legendre function and δ_m the Kronecker delta, and the constants g_l^m and h_l^m are given by

$$g_l^m = \frac{\mu_0(l+1)}{2\sqrt{2\pi(2l+1)}} (K_{lm}(-1)^m + K_{l,-m} \left[1 - \delta_{m,0} \times \left[1 - \frac{1}{\sqrt{2}} \right] \right]) \quad [4]$$

and

$$h_l^m = \frac{\mu_0(l+1)}{2\sqrt{2\pi(2l+1)}} (iK_{lm}(-1)^m - iK_{l,-m}) \quad [5]$$

In addition expressions for the inductance, L , and resistance, R , of a coil design based on the current distribution, $\underline{J}(\theta, \phi)$, have been originally derived:

$$L = \frac{\mu_0 a}{I^2} \sum_{l=1}^{\infty} \sum_{m=0}^l \frac{l(l+1)}{(2l+1)} (-1)^m K_{l,-m} \times K_{lm} \quad [6]$$

and

$$R = \frac{\rho}{tI^2} \sum_{l=1}^{\infty} \sum_{m=0}^l l(l+1) (-1)^m K_{l,-m} \times K_{lm} \quad [7]$$

where I is the current carried by the final coil design, ρ is the resistivity and t is the conductor thickness. These parameters can be used in the design process.

The process of designing and manufacturing a coil incorporating wires wound on a hemispherical region, involves representing the stream function that describes the

current on the hemisphere in terms of an expansion of the form described in Equation [1] which relates to a full-spherical stream function.

- 5 For an **axial gradient coil**, which generates a linear variation with the z -co-ordinate of the z -component of the magnetic field, the appropriate form of the hemispherical stream function is:

$$\begin{aligned} S(\theta, \phi) &= \sum_{l=1}^{\infty} c_l (P_l(\cos \theta) - P_l(0)P_0(\cos \theta)) & 0 \leq \theta \leq \pi/2 \\ &= 0 & \theta > \pi/2 \end{aligned} \quad [8]$$

- 10 (with $P_l(x)$ being the Legendre function) that can also be represented using Equation (1) with

$$K_{pm} = \delta_{m,0} \left(\sqrt{(2p+1)\pi} \sum_{l=1}^N c_l \left[\int_0^1 (P_l(x) - P_l(0)P_0(x)) P_p(x) dx \right] \right). \quad [9]$$

- 15 It is necessary to include the 2nd term in the brackets of Equation [8] to ensure that the stream function goes to zero at the equatorial edge of the hemisphere. These expressions can be used in conjunction with known expressions for cylindrical coils, with either circular or elliptical cross-section, to design coils with a dome-shaped end and cylindrical extension. On their own they may be used to design purely hemispherical coils.

- 20 In either case, a gradient coil design may be arrived at by choosing a set of N coefficients, c_l , that minimise a functional, U , of the form:

$$U = \sum_{q=1}^Q \left(\frac{\partial B_z(\mathbf{r}_q)}{\partial z} - G \right)^2 + \alpha L + \beta R \quad [10]$$

- 25 in which the first term represents the deviation of the field from a perfect gradient over a set of Q points at positions $\mathbf{r}_q$, while the second and third terms represent the inductance and resistance multiplied by adjustable weighting coefficients, α and β . Once optimal values of c_l have been identified, the hemispherical stream function can be generated using Equation [8]. Paths of wires which will produce a pattern of

current flow that mimics the required current distribution are given by the contours of the stream function.

Considering a purely hemispherical coil arrangement, Figure 7 shows the positions of wires on a hemispherical surface of 35 cm diameter which when energised with current generate a z -gradient which deviates from linearity by less than 5 % within an 11 cm diameter spherical volume (dsv) that is centred 7.5 cm in from the edge of the hemisphere. Figure 8 shows the variation of the field generated by this coil within the hemisphere – indicating that a linear variation of the field with z -position is present and showing that the required homogeneity has been achieved over the 11 cm dsv. This coil has an efficiency, η , (gradient strength per unit current) of $522 \mu\text{Tm}^{-1}\text{A}^{-1}$ and has an inductance of $102 \mu\text{H}$, giving a value of η^2/L of $2 \times 10^{-3} \text{T}^2\text{m}^{-2}\text{A}^{-2}\text{H}^{-1}$. The value of η^2/L , which is a good indicator of coil performance, is more than 2.5 times higher than that of a cylindrical coil of similar diameter. In addition, the length of wire needed to make this coil is much less than that required in a comparable cylindrical coil, thus leading to a significant reduction in resistance when the same wire diameter is employed in both coils.

In the case of a **transverse gradient** coil that generates a linear variation of the z -component of the magnetic field with the x -co-ordinate, the appropriate form of the hemispherical stream function is:

$$S(\theta, \phi) = \sum_{l=1}^{\infty} c_l (P_l(\cos\theta) - P_l(0)P_l(\cos\theta)) \cos\phi \quad 0 \leq \theta \leq \pi/2 \quad [11]$$

$$= 0 \quad \theta > \pi/2$$

This can again be represented using Equation [1] with

$$K_{pm} = (\delta_{m,1} - \delta_{m,-1}) \left(\sqrt{\frac{(2p+1)}{p(p+1)}} \pi \sum_{l=1}^N c_l \left[\int_0^1 (P_l(x) - P_l(0)P_l(x)) P_{p1}(x) dx \right] \right) \quad [12]$$

In this case, values of c_l are chosen so as to minimise a functional,

$$U' = \sum_{q=1}^Q (B_z(\mathbf{r}_q) - Gx_q)^2 + \alpha L + \beta R \quad [13]$$

To design a transverse, y -gradient coil the “ $\cos\phi$ ” term in Equation [11] is replaced by
 5 “ $\sin\phi$ ” and slightly modified versions of Equations [12] and [13] can then be formed.

Figure 9 shows the positions of wires on a hemispherical surface of 35 cm diameter which when energised with current generate an x -gradient which deviates from linearity by less than 5 % within an 11 cm diameter spherical volume (dsv) 10 that is centred 7.5 cm in from the edge of the hemisphere. Figure 10 shows the variation of the field generated by this coil within the hemisphere – indicating that a linear variation of the field with x -position is present and showing that the required homogeneity has been achieved over the 11 cm dsv. This coil has an efficiency of 345 $\mu\text{Tm}^{-1}\text{A}^{-1}$ and has an inductance of 67 μH , giving a value of the ratio, η^2/L of $1.8 \times$
 15 $10^{-3} \text{ T}^2\text{m}^{-2}\text{A}^{-2}\text{H}^{-1}$. This value is more than 3 times higher than that of a cylindrical coil of similar diameter. In addition, the length of wire needed to make this coil is much less than that required in a comparable cylindrical coil, thus leading to a significant reduction in resistance when the same wire diameter is employed in both coils.

20 The coil shown in Fig. 9 experiences no net force when carrying current in a uniform magnetic field, but will experience a net torque which acts to rotate the coil about a y -directed axis. This torque can however be eliminated by appropriately choosing the coefficients, c_l , in Equation [11]. The torque, $\underline{\mathbf{T}}$, experienced by a current distribution of the form described in Equation [11] exposed to a magnetic field $B_0 \hat{\mathbf{z}}$ is given by:

$$25 \quad \underline{\mathbf{T}} = -\hat{\mathbf{y}} \sum_{l=1}^N \pi B_0 a^2 c_l \left(\int_0^1 (1-x^2)^{\frac{1}{2}} [P_{ll}(x) - P_{ll}(x)P_{ll}(0)] dx \right) \quad [14]$$

By defining the N^{th} coefficient, c_N , in terms of the other $N - 1$ coefficients such that $\underline{\mathbf{T}} = 0$, and then choosing the values of these coefficients so as to minimise a functional

16

of the form shown in Equation [13] it is possible to design a torque-balanced hemispherical, transverse gradient coil.

5 Figure 11 shows such a coil wound on a 35 cm diameter hemispherical surface. This coil has an efficiency of $182 \mu\text{Tm}^{-1}\text{A}^{-1}$ and an inductance of $76 \mu\text{H}$, giving a value of η^2/L of $4.4 \times 10^{-4} \text{ T}^2\text{m}^{-2}\text{A}^{-2}\text{H}^{-1}$, which is still higher than that achievable in comparably sized cylindrical coils.

10 Gains in performance over coils comprising purely hemispherical shapes can be achieved by combining the hemisphere with a cylindrical section as shown in Fig. 3. Figures 13 and 14 show exemplary coils of this form that generate x - and z -gradients respectively.

15 Although embodiments of the present invention have been described in the preceding paragraphs with reference to various examples, it should be appreciated that modifications to the examples given can be made without departing from the scope of the invention as claimed.

20 Whilst endeavoring in the foregoing specification to draw attention to those features of the invention believed to be of particular importance it should be understood that the Applicant claims protection in respect of any patentable feature or combination of features hereinbefore referred to and/or shown in the drawings whether or not particular emphasis has been placed thereon.

25 I/we claim:

CLAIMS

1. An arrangement of conductive loops for providing a magnetic field that varies linearly in a first direction across an imaging region of a magnetic resonance imaging system, wherein the conductive loops are collectively arranged to define a dome.
2. An arrangement as claimed in claim 1, wherein the first direction is parallel to a longitudinal axis of the magnetic resonance imaging system and the conductive loops surround an imaginary line that extends in the first direction.
3. An arrangement as claimed in claim 1 or 2, comprising:
a first plurality of serially interconnected adjacent conductive loops at a first portion of the dome and a separated, second plurality of serially interconnected adjacent conductive loops at a second portion of the dome, wherein in use the electric current in the first plurality of serially interconnected adjacent conductive loops circulates in the opposite sense to the electric current in the second plurality of serially interconnected adjacent conductive loops.
4. An arrangement as claimed in claim 4, wherein all of the first plurality of serially interconnected adjacent loops surround a first imaginary line, that extends in the first direction, at the first portion of the dome and all of the second plurality of serially interconnected adjacent loops surround the first imaginary line at the second portion of the dome.
5. An arrangement as claimed in claim 1 or 2, comprising:
a first plurality of serially interconnected adjacent conductive loops at a first portion of the dome
a second plurality of serially interconnected adjacent conductive loops at a second portion of the dome
a third plurality of serially interconnected adjacent conductive loops at a third portion of the dome

a fourth plurality of serially interconnected adjacent conductive loops at a fourth portion of the dome, wherein in use the electric currents in each of the first plurality of serially interconnected adjacent conductive loops and the third plurality of serially interconnected adjacent conductive loops circulate in the opposite sense to the electric currents in each of the second plurality of serially interconnected adjacent conductive loops and the fourth plurality of serially interconnected adjacent conductive loops.

6. An arrangement as claimed in claim 5, wherein all of the first plurality of serially interconnected adjacent loops surround a first imaginary line, that extends in the first direction, at the first portion of the dome and all of the second plurality of serially interconnected adjacent loops surround the first imaginary line at the second portion of the dome and wherein all of the third plurality of serially interconnected adjacent loops surround a second imaginary line, that extends in the first direction, at the third portion of the dome and all of the second plurality of serially interconnected adjacent loops surround the second imaginary line at the fourth portion of the dome.

7. An arrangement as claimed in any preceding claim, wherein the paths of the conductive loops are defined by the contours of a stream function bounded by the shape of the dome.

8. An arrangement as claimed in claim 7, wherein the stream function is such that the magnetic field produced by the conductive loops varies linearly in the first direction.

9. An arrangement as claimed in claim 7 or 8, wherein the stream function is such that the total inductance and the total resistance of the conductive loops are minimal

10. An arrangement as claimed in claim 7, wherein the stream function is defined as a series of Legendre functions wherein the coefficients of the series minimize the inductance and the resistance of the conductive loops while the magnetic field produced by the conductive loops is constrained to vary linearly in the first direction.

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11. An arrangement as claimed in any preceding claim wherein, in use, the net torque on the conductive loops is zero.

5 12. An arrangement as claimed in any preceding claim, wherein the dome is a section of an ellipsoid.

13. An arrangement as claimed in any claim 12, wherein the dome is a section of a sphere.

10 14. An arrangement as claimed in any preceding claim, wherein the conductive loops are supported by a dome shaped surface.

15 15. A method of manufacturing an arrangement of conductive loops for providing a linearly varying magnetic field in a first direction across an imaging region of a magnetic resonance imaging system, comprising:
determining the coefficients of a stream function for a dome shape so that the inductance defined by Equation 6 and the resistance defined by Equation 7 are simultaneously minimized while the magnetic field is constrained to vary linearly in the first direction; and
20 arranging conductive loops along the contours of the stream function.

16. A method of manufacturing an arrangement of conductive loops for providing a linearly varying magnetic field in a first direction across an imaging region of a magnetic resonance imaging system, comprising:
25 determining the coefficients of a stream function for a dome shape so that the applied torque in use is zero while the magnetic field is constrained to vary linearly in the first direction; and
arranging conductive loops along the contours of the stream function.

30

17. A gradient coil for magnetic resonance imaging, incorporating a dome-shaped surface upon which the current carrying conductors are arranged.

5 18. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a hemisphere.

10 19. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a hemisphere and is attached to a cylinder of circular cross-section upon which current carrying conductors are also arranged.

15 20. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a hemisphere and is attached to a flared cylinder of circular cross-section whose diameter increases upon moving away from the circle of attachment, upon which current carrying conductors are also arranged.

21. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a portion of a spherical surface which is not a hemisphere.

20 22. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a portion of a spherical surface which is not a hemisphere and is attached to a cylinder of circular cross-section upon which current carrying conductors are also arranged.

25 23. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a portion of a spherical surface which is not a hemisphere and is attached to a flared cylinder of circular cross-section whose diameter increases upon moving away from the circle of attachment, upon which current carrying conductors are also arranged.

30 24. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a portion of an ellipsoidal surface.

25. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a portion of an ellipsoidal surface and is attached to a cylinder of elliptical cross-section upon which current carrying conductors are also arranged.
- 5 26. The gradient coil of Claim 17 wherein the dome-shaped region is formed from a portion of an ellipsoidal surface and is attached to a flared cylinder of elliptical cross-section whose major and/or minor axes increase in size upon moving away from the ellipse of attachment, upon which current carrying conductors are also arranged.
- 10 27. The gradient coils of Claims 18 to 20 in which the conductor positions are established by expressing the current distribution on the hemisphere in terms of a weighted series of spherical harmonics on a complete spherical surface.
28. The gradient coils of Claims 18-20 in which the torque on the gradient coil is
15 eliminated by expressing the current distribution on the hemisphere in terms of a stream function, $S(\theta, \phi)$, where θ and ϕ are the usual spherical polar co-ordinates, which is represented as a series of associated Legendre functions of the form

$$S(\theta, \phi) = \sum_{l=1}^{\infty} c_l (P_l(\cos \theta) - P_l(0)P_{l1}(\cos \theta)) \cos \phi \quad 0 \leq \theta \leq \pi/2$$

$$= 0 \quad \theta > \pi/2$$

- 20 and then choosing the coefficients, c_l , so that

$$\sum_{l=1}^N c_l \left(\int_0^1 (1-x^2)^{\frac{1}{2}} [P_l(x) - P_{l1}(x)P_{l1}(0)] dx \right) = 0.$$

- 25 29. A magnetic resonance imaging system comprising an arrangement as claimed in any one of claims 1 to 14 or a gradient coils as claimed in any one of claims 17 to 28.

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30. A magnetic resonance imaging system comprising an arrangement of conductive loops for providing a magnetic field that varies linearly in a direction across an imaging region within a patient's head, wherein the conductive loops are collectively arranged to define a surface that overspreads at least a crown of a patient's head.

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31. A magnetic resonance imaging system as claimed in claim 30, further comprising a primary magnet for producing in the imaging region a primary magnetic field that travels through the patient's head and the surface at least where it overspreads the crown of the patient's head.

10

32. A magnetic resonance imaging system comprising:

a first dome of conductive loops for providing a first magnetic field that varies linearly in a first direction across an imaging region; and

a second dome of conductive loops for providing a second magnetic field that varies linearly in a second direction, orthogonal to the first direction, across the imaging region,

15

wherein the first dome of conductive loops and the second dome of conductive loops are arranged as nested domes.

20

25

Figure 1

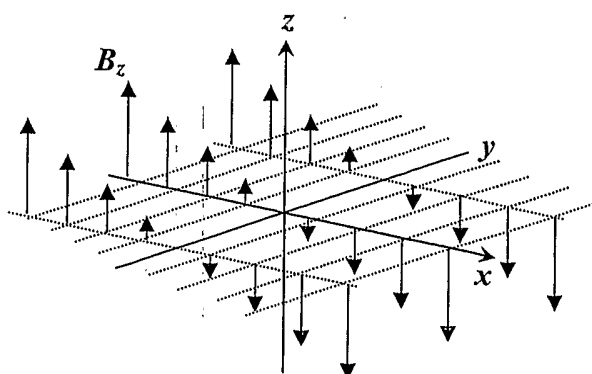


Figure 2

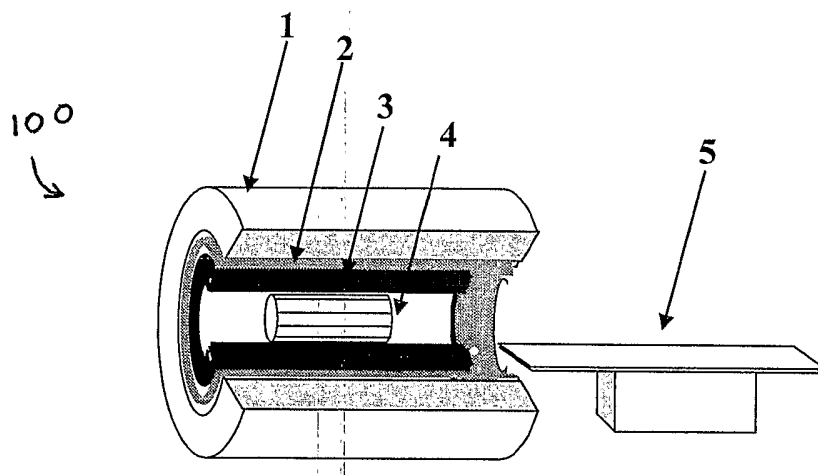
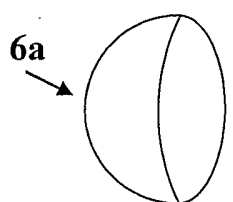
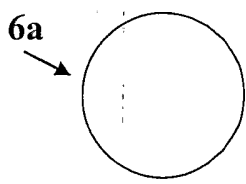


Figure 3A

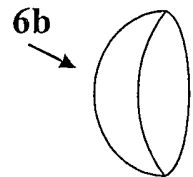


Perspective

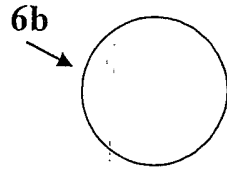


Plan

Figure 3B

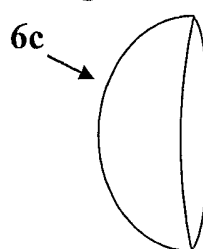


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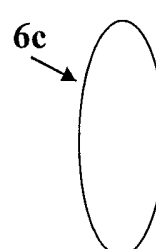


Plan

Figure 3C



Perspective



Plan

Figure 4 A

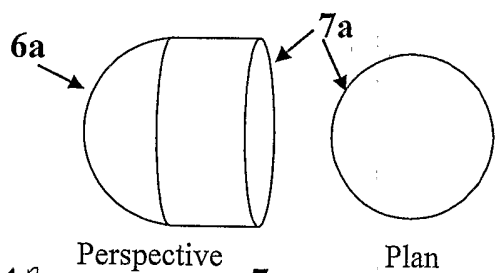


Figure 4C

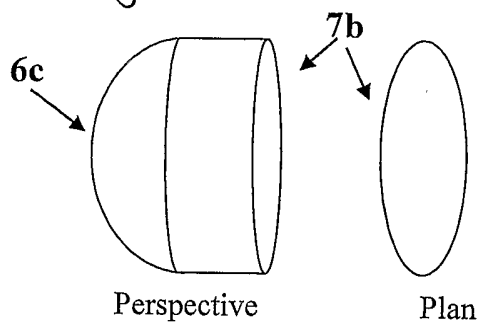


Figure 4B

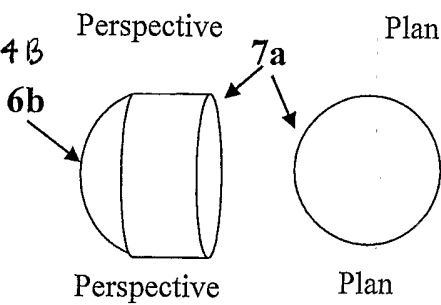


Figure 5

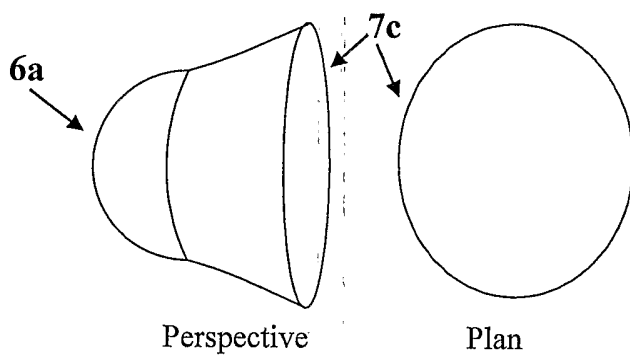


Figure 6

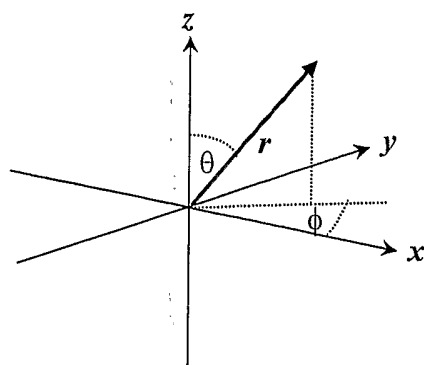


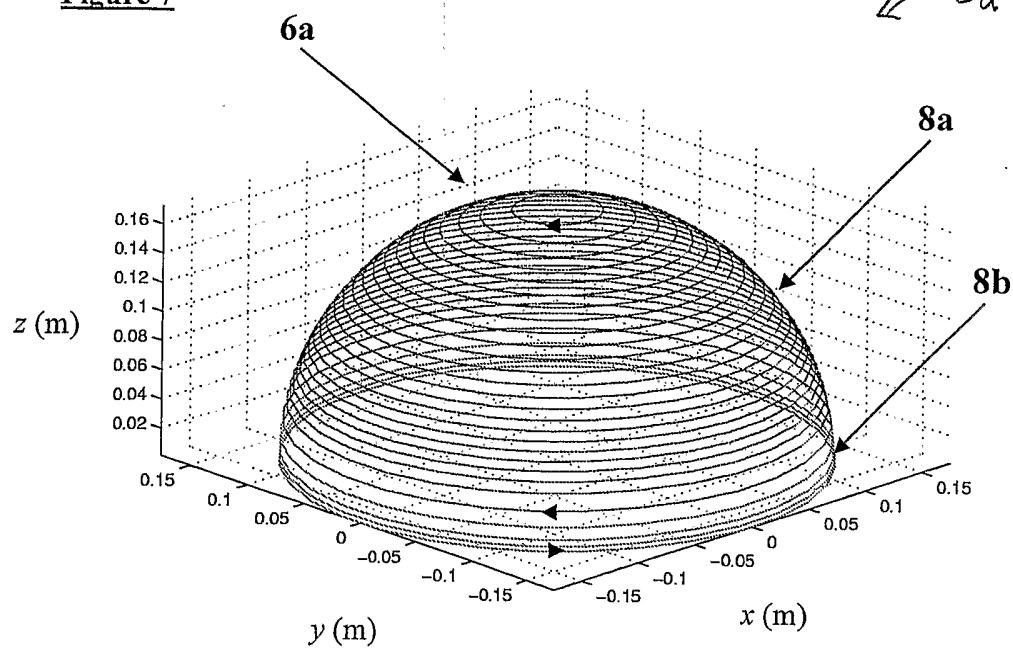
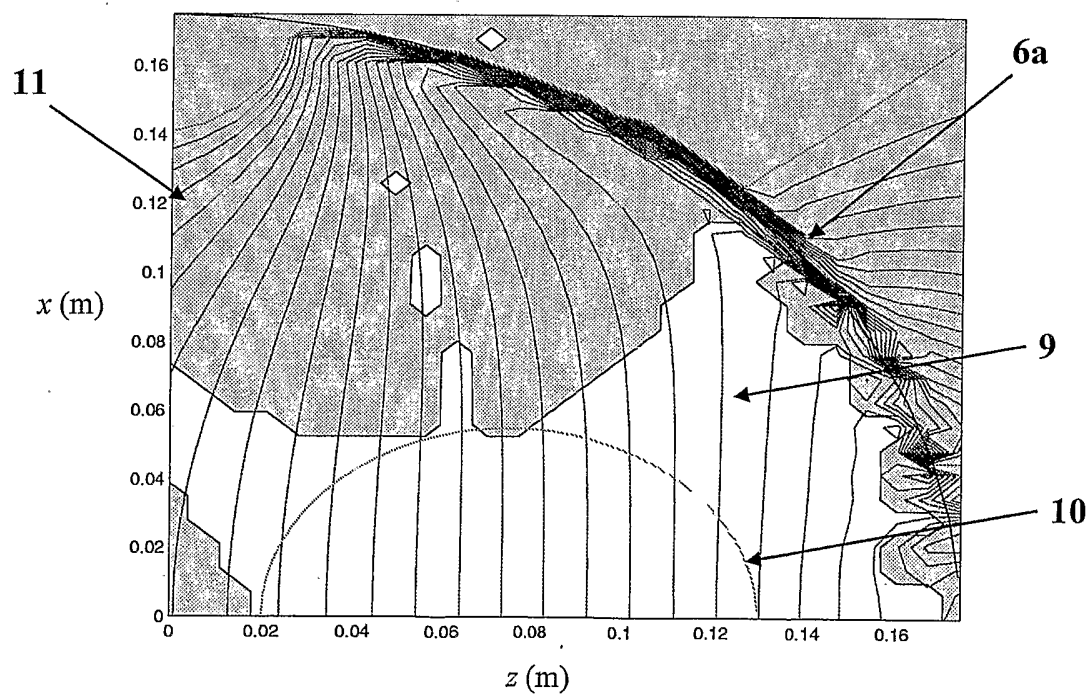
Figure 7**Figure 8**

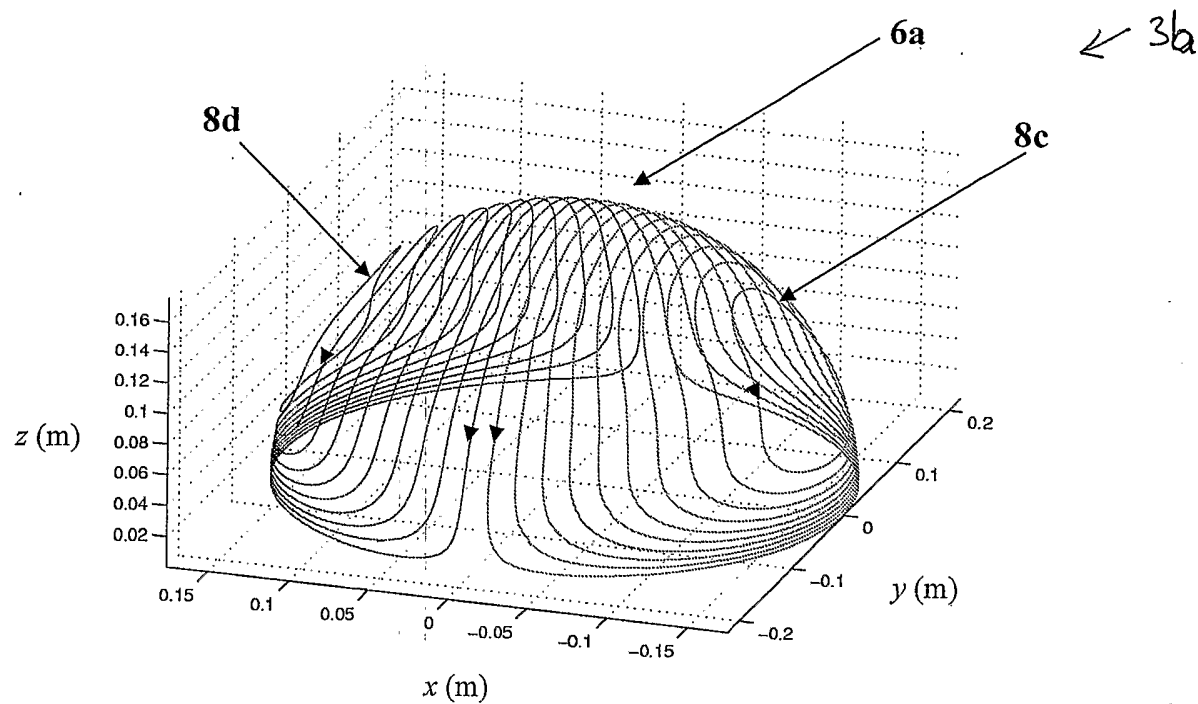
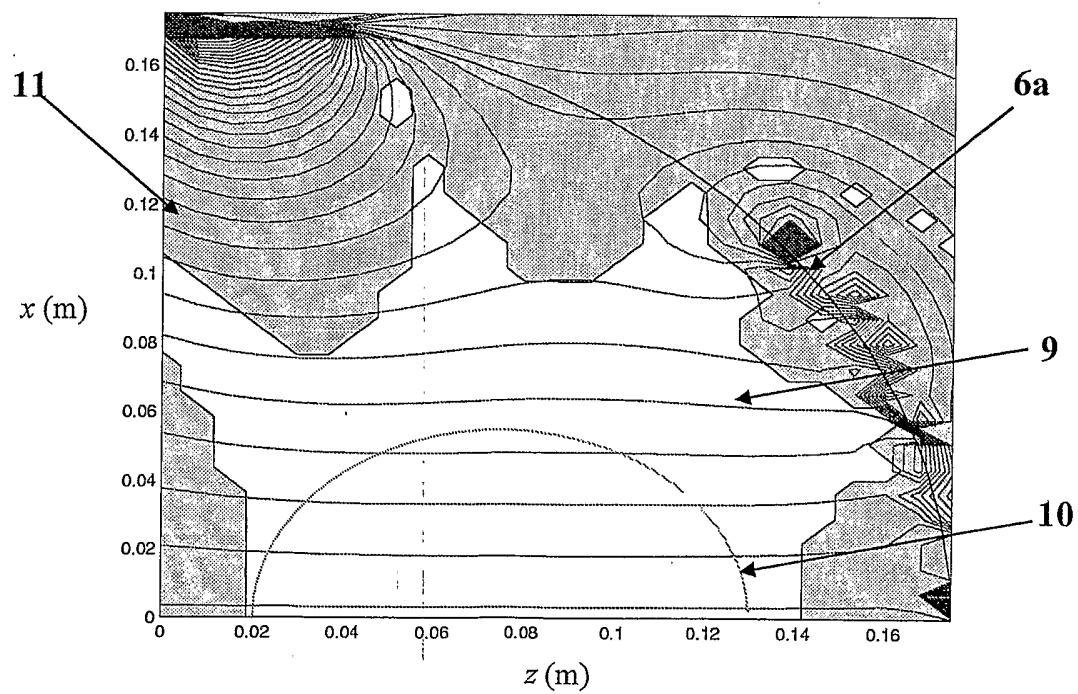
Figure 9**Figure 10**

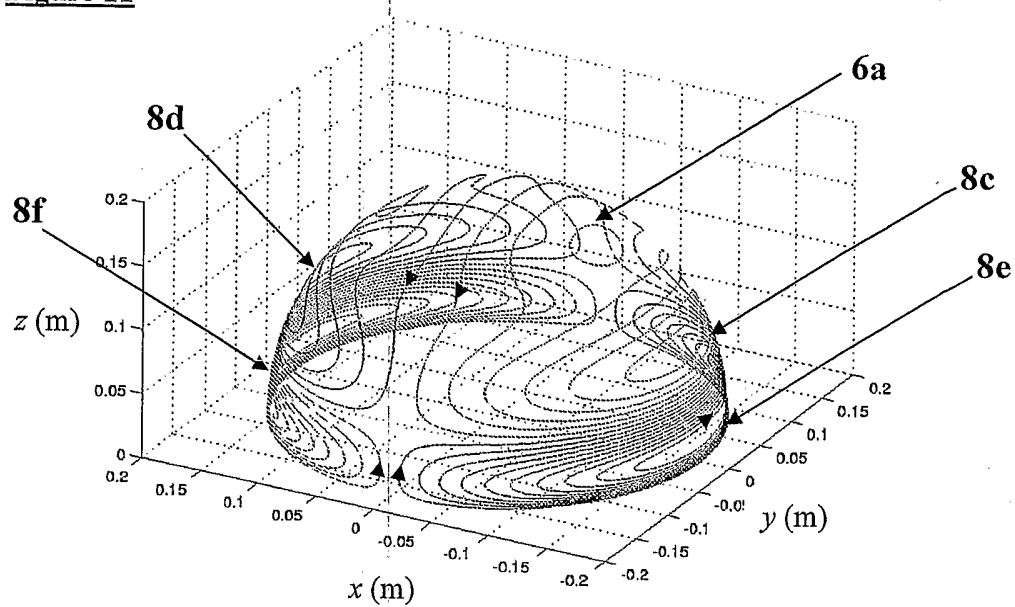
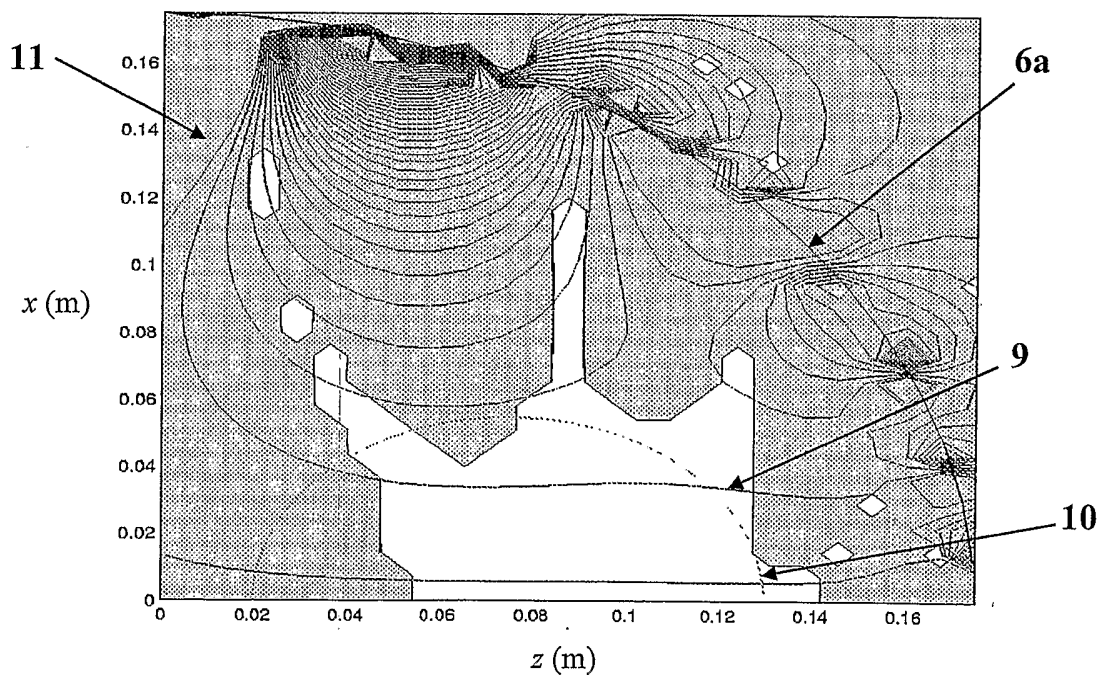
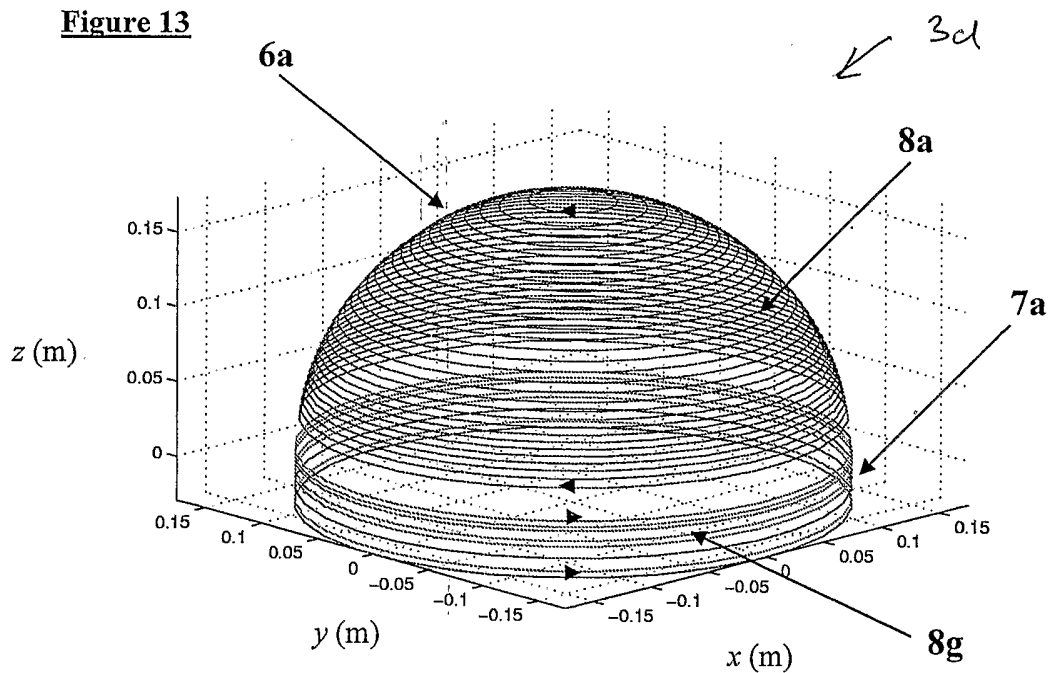
Figure 11**Figure 12**

Figure 13**Figure 14**