(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
31 December 2008 (31.12.2008)

PCT

(10) International Publication Number
WO 2009/001084 A1

(51) International Patent Classification:
G01R 33/3815 (2006.01) *G01R 33/3875* (2006.01)

(21) International Application Number: PCT/GB2008/002182

(22) International Filing Date: 25 June 2008 (25.06.2008)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
0712421.7 26 June 2007 (26.06.2007) GB

(71) Applicant *(for all designated States except US)*: OXFORD INSTRUMENTS PLC [GB/GB]; Tubney Woods, Abingdon, Oxon OX13 5QX (GB).

(72) Inventors; and
(75) Inventors/Applicants *(for US only)*: McDOUGALL, Ian, Leitch [GB/GB]; 16 Lees Heights, Charlbury, Oxon OX8 3EL (GB). HANLEY, Peter [GB/GB]; Western Lodge, Parkend, Lydney, Gloucestershire GL15 4JQ (GB).

(74) Agent: GILL JENNINGS & EVERY LLP; Broadgate House, 7 Eldon Street, London EC2M 7LH (GB).

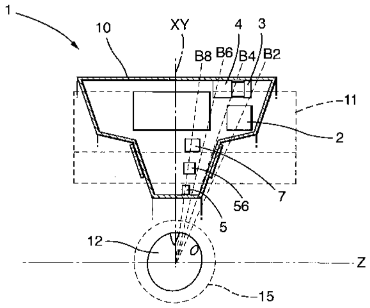
(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BR, BW, BY, BZ, CA, CH, CN, CO, CR, CU, CZ, DE, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IS, JP, KE, KG, KM, KN, KP, KR, KZ, LA, LC, LK, LR, LS, LT, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PG, PH, PL, PT, RO, RS, RU, SC, SD, SE, SG, SK, SL, SM, SV, SY, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LS, MW, MZ, NA, SD, SL, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European (AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MT, NL, NO, PL, PT, RO, SE, SI, SK, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, ML, MR, NE, SN, TD, TG).

Published:
— *with international search report*

(54) Title: MAGNET SYSTEM FOR USE IN MAGNETIC RESONANCE IMAGING

Fig.1.



(57) Abstract: A magnet system is described for use in magnetic resonance imaging. In use the system generates a resultant magnetic field, within a working region, of sufficient homogeneity to enable magnetic resonance imaging to be performed. The system comprises a primary magnet comprising a pair of coils positioned upon a common axis passing through the centre of the working region, the coils being spaced symmetrically about an origin upon the common axis and having dimensions such that each coil is at or adjacent a nominal position at which the second order derivative of a corresponding magnetic field in the working region is substantially zero for a coil pair located symmetrically about the origin at such a position. A shielding magnet comprises a pair of coils positioned upon the common axis, the coils being spaced symmetrically about the origin and having dimensions such that each coil is at or adjacent a nominal position at which the second order derivative of a corresponding magnetic field in the working region is substantially zero for a coil pair located symmetrically about the origin at such a position, the shielding magnet being operative in use to reduce the resultant magnetic field at locations distal from the magnet system. A compensation magnet comprises at least one pair of coils positioned upon the common axis, the coils being spaced symmetrically about the origin and having dimensions such that each coil is at or adjacent a nominal position at which at least one even order derivative of a corresponding magnetic field having a magnitude of 6 or above, in the working region, is substantially zero for a coil pair located symmetrically about the origin at such a position.

WO 2009/001084 A1

MAGNET SYSTEM FOR USE IN MAGNETIC RESONANCE IMAGING

Field of the Invention

The present invention relates to a magnet system for use in magnetic resonance imaging (MRI).

Background to the Invention

Magnets for MRI systems have evolved with a typically tubular examination space, derived from the use of “solenoid” magnet geometry. This is because the solenoid arrangement provides excellent uniformity of the magnetic field. For practical applications the “solenoid” arrangement has been effected using quasi-solenoids in which sets of circular coils are distributed along the axis of the solenoid.

In designing magnet systems it is known to use adapted Legendre polynomial functions to describe the way in which the magnetic field intensity due to a circular current loop behaves along a radius projected from the origin of a solenoid. The magnetic field vector can be described as rotating about a current carrying conductor, whereby the magnetic flux presents a closed path about the current source. At any point in the magnetic field, the rate of rotation of field can be described by the curl of the field vector. The designer of a magnet intended to produce a uniform field seeks to reduce to near zero the rate of field rotation at the origin of the solenoid. This is done conveniently by describing the field via a series of field derivatives and spatial coordinates and the dimensions and current value of each coil (expressed as a series of current loops). In general, values of the field derivatives describing rotation (uniformity) of the central magnetic field may be calculated and plotted against (for convenience) a given coil radius and current, at points along a uniform cylindrical bore tube. Such plots show that the value of field derivatives exhibit harmonic profiles with bore length, and that there are zero value positions, where the field derivative profile changes from a positive to negative value, or vice versa.

To develop central (origin) field uniformity, the design process seeks to balance a number of coils with respect to their strengths of field derivatives each side of such zero points, the result being that chosen field derivatives sum to zero for the coil set. Similar harmonic plots of field derivatives may be obtained against coil radius at a particular bore length. Again, sets of coils may be chosen to provide that the sum of a particular order field derivative is zero, and, consequently, this order does not contribute to field non-uniformity at the origin.

Standard tubular MRI magnets are therefore formed from a series of co-axial magnetic field coils, all of similar diameter. The coils, considered as pairs across the mid-plane, contribute magnetic field to the working total field for MRI at the origin, or centre of the magnet, along with characteristic non-uniformity in the central magnetic field. By judicious placing of the coil pairs, non-uniformity due to one coil pair may be cancelled out by the non-uniformity of another pair, when all coils carry current in the same sense. The non-uniformities can be described by nth orders of derivatives of the magnetic field, considered as an expansion of increasingly high orders.

We have realised that standard, high field MRI magnets cannot be made substantially shorter without reducing the uniformity of the magnetic field at the origin of the magnet (the centre of the working region), which would in turn cause reduction in the NMR signal quality. If a tubular magnet is made shorter, but the coils are retained on approximately the same diameter, it becomes increasingly difficult to cancel out higher order field derivatives. This is true even if an additional degree of freedom is added in allowing some coils to carry current in the reverse sense to that of the whole magnet.

There is therefore a need to produce shorter magnets without incurring the problems of non-uniformity that results when a solenoid design is shortened axially.

Summary of Invention

In accordance with the invention we provide a magnet system for use in magnetic resonance imaging, the magnet system being adapted when in use to generate a resultant magnetic field, within a working region, of sufficient homogeneity to enable magnetic resonance imaging to be performed, the system comprising:-

a primary magnet comprising a pair of coils positioned upon a common axis passing through the centre of the working region, the coils being spaced symmetrically about an origin upon the common axis and having dimensions such that each coil is at or adjacent a nominal position at which the second order derivative of a corresponding magnetic field in the working region is substantially zero for a coil pair located symmetrically about the origin at such a position;

a shielding magnet comprising a pair of coils positioned upon the common axis, the coils being spaced symmetrically about the origin and having dimensions such that each coil is at or adjacent a nominal position at which the second order derivative of a corresponding magnetic field in the working region is substantially zero for a coil pair located symmetrically about the origin at such a position, the shielding magnet being operative in use to reduce the resultant magnetic field at locations distal from the magnet system; and,

a compensation magnet comprising at least one pair of coils positioned upon the common axis, the coils being spaced symmetrically about the origin and having dimensions such that each coil is at or adjacent a nominal position at which at least one even order derivative of a corresponding magnetic field having a magnitude of 6 or above, in the working region, is substantially zero for a coil pair located symmetrically about the origin at such a position.

We have realised that it is possible to produce very short tubular magnets, that is, magnets powered by preferably, circular coils, by separating the coil pairs into two groups. The two groups are clustered, or "anchored" at or adjacent, respectively, the positions along the tube at which the lowest order field derivative is zero, and the positions along the tube at which the highest field derivative to be cancelled is zero. Preferably the coils are at or adjacent in the sense that each coil has a corner of its cross-section within a distance of the nominal position line in radial and axial co-ordinates, wherein the said distance is not greater than the maximum of the radial or axial extent of the coil section, whichever is the larger. More preferably, the coils are at or adjacent the nominal position in the sense that the coils have at least part of the coil cross-section intersecting the nominal position.

Typically this highest field derivative may be sixth order, or more typically, eighth, or even tenth order. The first group comprise the primary magnet and the shielding magnet, whereas the second group comprise the compensation magnet. The said "nominal positions" which cause the cancellation of nth order magnetic field gradients are nominal positions derivable from magnetic theory. The coils of the magnets according to the invention are positioned at or adjacent these positions, the differences between the actual positions and the nominal positions being due in particular to the finite size of the coils and the fact that, in order to produce a desired geometry, whilst minimizing the use of coil material, some non-zero magnitude in the particular field derivative may be advantageous, practically.

The respective nominal position is typically defined by a frusto-conical surface of rotation about the common axis, wherein the said derivative of the respective magnetic field is zero for a nominal set of ideal coils defining a circle, each part of which intersects with said surface. Such ideal coils have zero radial and axial thickness. The frusto-conical surface is preferably defined by a cone angle, the cone angle being the angle at the cone vertex between the common axis and a line lying parallel to said surface and passing through the vertex, and wherein the cone angle for each of the pairs of coils of the compensation magnet is in excess of that of the primary magnet.

The innovation of anchoring coil pairs in radial/axial space along the "tube" results in very short magnets with the ability to cancel high order field derivatives. To great advantage for access, the anchoring of coils to co-ordinates of radial/axial space with zero values of lowest and highest order field derivative produces a conical set of coils, in two separate groups. This is because the zero value of field derivative of order n has a characteristic ratio of radius (a) to axial length (b). All nth orders have at least one zero value for a ratio of a/b greater than one.

This means for a chosen order, as the diameter of the coil is increased, there is a proportionally smaller increase in coil length. However, for a given order n of field derivative, and considering the ratio a/b which provides zero value of the n order field derivative, all other orders of field derivative have finite values. Thus it is possible to groups coils on a chosen a/b "line" (cone in three dimensions) of order n, such that as pairs across the mid-plane of the tubular magnet (that having a normal parallel to the common axis and positioned passing through the centre of the working region), can be combined to cancel non-uniformities at the centre of the magnet for all orders other than the chosen nth order.

A magnet system with extremely short tubular bore has therefore been invented using "conical" arrangements of coils along the bore of the magnet. Typically the smallest diameter coils of the magnet are nearest to the mid-plane of the tubular bore, and the largest diameter coils furthest from the mid-plane of the tubular bore.

Preferably, each coil is defined by a circle centred upon the common axis and wherein each part of said circle is equally at or adjacent the nominal position.

Because the coils are typically circular in geometry, they have optimum (well known) mechanical properties. Thus high values of MRI field can be obtained when the coils will operate at high internal stress values.

A hitherto unobtainable combination of access to the subject being examined is achieved, with very high image quality, because of the high signal to noise provided by the high magnetic field strength obtainable. A high image quality and access to the "patient" subject is particularly valuable to the research community who wish to use MRI as a gauge of experiments in molecular medicine. For example it may be used to observe brain activity during investigation of the effect that drug compounds may have on cognitive ability.

In general the net current within at least one pair of the coils of the compensation magnet is caused when in use to flow about the common axis in a counter-running manner with respect to the net current in the primary magnet. The coil arrangement of the present invention provides for only a small amount of counter-running current which prevents the magnets unnecessarily working against one another. The efficiency of the system is therefore improved. Typically the compensation magnet comprises a number of pairs of coils, such as three pairs, each pair having a different radius. Advantageously, the pair of coils having the smallest radius is preferably counter-running.

Whilst the coils are provided at or adjacent the respective nominal position, a pair of coils for one or more of the primary, shielding or compensation magnets, typically has coil windings at the respective nominal position. This is particularly the case for the compensation magnet where it is preferred that each coil has windings at the nominal position. The windings may be distributed substantially uniformly within the coil cross-section and it is the centre of the said cross-section (at the average position of the windings) that may be used to define the position of the coil.

It will be appreciated that typically the shielding magnet has coils having the largest radius. Each of the shielding and primary magnets typically has coils having a radius in excess of those of the compensation magnet. The large radius coils of the shielding and primary magnets is operative to produce a relatively small high order (such as 8^{th} order) derivative of the magnetic field in the working region. The primary magnet typically provides the largest contribution to the resultant magnetic field, such as in excess of 75% of the resultant magnetic field strength.

It will be appreciated that the primary, shielding and compensation magnets are operative together to produce a sufficiently homogeneous magnetic field in the working region for performing magnetic resonance imaging. It should be noted however that the arrangement is such that, when taken alone, each of the primary, shielding and compensation magnets do not produce a sufficiently homogeneous magnetic field in the working region for imaging.

The shielding magnet is provided so as to limit the extent of the resultant magnetic field at locations distal from the magnet system. The shielding magnet is adapted when in use to have a net current flow in its coils which is counter-running with respect to the net current flowing within the coils of the primary magnet.

In some arrangements the coil pair of the shielding magnet is placed adjacent the nominal second order derivative position so as to cause a non-zero magnitude second order derivative having a first polarity and the coil pair of the primary magnet is placed adjacent the nominal second order derivative position so as to cause a non-zero magnitude second order derivative having a polarity opposite to the first polarity, such that the second order magnetic field derivatives of the primary and shielding magnets cancel.

In other arrangements the combined effect of the primary and shielding magnets is to cause a non-zero value of the second order magnetic field derivative. Thus in this case, the coil pair of the shielding magnet is placed adjacent the nominal second order derivative position so as to cause a non-zero magnitude second order derivative which is not fully cancelled by the primary magnet. The shielding magnet may comprise two sets of coils spaced apart along the common axis. These may each be upon one side of the nominal position. Each of the sets of coils of the shielding magnet may be positioned closer to the origin at the centre of the working region than the nominal second order derivative position.

As will be appreciated, although each of the magnets could be powered separately, typically they are each powered together by electrically connecting them in series. A suitable controller is preferably provided to control the electrical current within at least one and preferably each of the primary, shielding and compensation magnets.

In order to produce a magnetic field within the working region of high uniformity it is necessary to accurately position the respective coils of the magnets. We have further realised that it is advantageous to provide one or more of the pairs of coils of the compensation magnet with additional coils. These may be controlled by a second controller independently of the pair to which they relate in each case. It is advantageous to provide such additional coils for the compensation magnet because the compensation magnet has the highest engineering tolerance requirements. It is also possible to provide the additional coils for the primary or shielding magnets, although this typically requires greater use of conductor material. The additional coils may be used in one of two ways. They may be adapted to allow the positions of the magnetic centres of the said at least one pair of coils to be modified by controlling the current within the additional coils. They may alternatively, although preferably additionally, be adapted to allow the strength of the magnetic field of the compensation magnet to be modified by controlling the current within the additional coils. The additional coils may therefore modify the strength and/or position of the effective centre of the pairs of coils. At least three additional coils may be provided for each coil in question, these being preferably distributed symmetrically about each coil. Four additional coils are preferred, these additional coils being positioned upon either side of each of the said at least one pair both radially and axially with respect to the common axis. The powering of the coils independently of the coils to which they relate can be performed using the controller (or the controller and second controller) and suitable switching apparatus. Typically the second controller is adapted in use to provide electrical current to the said coils such that the corresponding strength of the magnetic field generated by the said additional coils alone is up to about 10% of the magnetic field strength of the compensation magnet.

In order to perform MRI procedures additional known apparatus is also required, this including a set of gradient coils. Thus, in order to maximise the benefit from the present invention, typically the system further comprises a set of gradient coils, each gradient coil being positioned at or adjacent the nominal second order derivative position.

Preferably, each magnet coil within the system is positioned substantially within a geometrical envelope defined by two symmetrical conical surfaces representing the second order position that extend away from the centre of the working region. Such an envelope may be further bounded by a cylindrical surface at the largest radius of the magnet coils. Typically the coils are enclosed within a common housing. The geometry of the apparatus may be such that the primary and shielding magnets are arranged in a first part of the housing which has a substantially annular geometry. The compensation coils may be arranged in a second part of the housing having substantially annular geometry of smaller axial and radial dimensions than the first part. The second part of the housing may be provided with walls which taper in axial dimension towards the common axis.

It is known in MRI apparatus to have a horizontal table upon which a subject to be imaged is located. This is then typically passed into the bore of a solenoid MRI magnet system. By contrast, the present system preferably further comprises a table upon which a subject is positioned for imaging when the system is in use, the table defining a table plane, and a system mounting, adapted to allow the magnets to be rotated with respect to the table plane, such that the common axis is rotated within a plane substantially normal to the table plane. Either alternatively or additionally the magnets may be arranged to be rotated in a plane substantially parallel to the table plane.

The coils of the magnets described above may be formed from high conductivity resistive wire. Preferably however, they are formed from superconducting wire, this being cooled when in use, using a coolant. We have further realised that the axial space between the respective coils of particularly the shielding and primary magnets can be used to accommodate a reservoir for coolant, such as a cryogen in the form of helium or nitrogen. The reservoir may be divided into two separate reservoirs, the reservoirs being positioned upon either side of a mid-plane of the system, the said plane passing through the centre of the working region and having a plane normal parallel to the common axis. This allows the system to operate in a wide range of orientations since the use of two dedicated reservoirs, one for each half of the magnet system, ensures a supply of coolant to the magnets during use.

We also provide a method of constructing such a magnet system, the method comprising the steps of:-

i) mounting one coil of each pair of coils of the primary, shielding and compensation magnets to a first support;

ii) mounting a first coolant reservoir to the first support, the first coolant reservoir being adapted to supply the said one coil of each of the primary, shielding and compensation magnets with coolant during use;

iii) mounting the other coil of each pair of coils of the primary, shielding and compensation magnets to a second support;

iv) mounting a second coolant reservoir to the second support, the second coolant reservoir being adapted to supply the said other coil of each of the primary, shielding and compensation magnets with coolant during use; and,

v) mounting the first and second supports together.

Thus the first support, coils and first reservoir may comprise a first half of the magnet system and the second support, coils and second reservoir may comprise a second half of the magnet system. The first and second halves are preferably arranged as substantially mirror images of one another.

Brief Description of the Drawings

Some examples of magnet systems according to the present invention are now described with reference to the accompanying drawings, in which:-

Figure 1 is a schematic section through a first example system;

Figure 2 is a table showing the field derivatives and required compensation magnet for a first coil pair;

Figure 3 is a table showing the field derivatives and required compensation magnet for a second coil pair;

Figure 4 is a table showing the field derivatives and required compensation magnet for a third coil pair;

Figure 5 is a table showing the field derivatives and required compensation magnet for a fourth coil pair;

Figure 6 is a table showing the field derivatives and required compensation magnet for a fifth coil pair;

Figure 7 is a table showing the field derivatives and required compensation magnet for a sixth coil pair arranged as a shielding magnet;

Figure 8 is a table showing the field derivatives and required compensation magnet for a seventh coil pair arranged as a shielding magnet;

Figure 9 is a table showing details of an example full magnet system;

Figure 10 is a table showing details of other example full magnet systems;

Figure 11 shows a coil with a set of additional coils;

Figure 12 shows the steps of a method of constructing the magnet systems;

Figure 13 shows a section through a magnet system;

Figure 14 shows relative movement between the system and a subject;

Figure 15 shows a first schematic perspective view of a magnet system;

Figure 16 shows a second schematic perspective view of the magnet system;

Figure 17 is a table showing calculated magnetic field values;

Figure 18 is a table comparing exact and calculated field values;

Figure 19 shows the effect of tolerances in coil fabrication; and,

Figure 20 shows the compensation of the tolerances using additional coils.

Detailed Description of Examples

We now describe a number of example magnet systems. We emphasize here that the magnets comprise three magnets. The first is a primary magnet (for generating the majority of the magnetic field) and the second is a shielding magnet to reduce the stray field. Each of these first and second magnets has coils which are positioned at or adjacent a theoretical position at which ideal coils would produce a zero second order derivative (order n=2 or “B2”), of the magnetic field B produced by such coils. The third magnet is a compensation magnet having coils at or adjacent an n=8 or B8 theoretical position.

This combination produces a very axially short magnet system where all the coil winding volumes lie outside a cone shaped region whose apex is the magnetic centre of the magnet system. In practice, the angles between the lines joining the geometric centres of the coil sections (the winding cross sections) and the magnetic centre of the system, and the magnetic axis, is greater than about 63 degrees. Taking into account the finite size of the coils, all the winding volumes typically lie outside a cone shaped region whose apex is the magnetic centre, whose axis is the magnetic axis and whose angle is greater than 45 degrees.

Each of the magnet systems described below produce a working region of sufficient uniformity to be suitable for MRI. Some of the coils of the magnet system are energised in opposition to others so that the combination produces the required magnetic field homogeneity and at the same time reduces the net magnetic moment of the system so as to effectively screen the system when viewed from a distance substantially greater than the radius of the largest coil (shielding).

The coils described may be resistive and continuously energised although they may be superconductive. Superconductive coils may be fitted with persistent mode switches, allowing them to be used in persistent mode (without being in constant connection to a power supply). Conventional superconductors or high temperature superconductors may be used to construct some or all of the magnet coils.

Practically, the systems described produce a short MRI magnet, with the length of the "patient" examination bore less than ½ the diameter of the examination bore (typically ¼ the diameter). This is achievable even though the coil sets of the primary, shielding and compensation magnets by themselves are not sufficiently uniform for MRI. The magnetic fields from each of the magnets need to be combined to produce a designed central field (at the origin of the short tubular bore) that is sufficiently uniform for MRI. When the coil sets are combined, the short nature of the tube allows a patient with their head in the central field to have line-of-sight to the examination room, external to the magnet system. This is a great advantage in the reduction of physiological stresses which otherwise may distort results from MRI observations.

Reference is now made to Figure 1 in which a first example magnet system 1 is illustrated. Figure 1 is schematic and representative of all examples described. It shows the system in section when viewed from the side. The system coordinates are defined by a common axis (Z) upon which the geometrical centre of each coil of the magnets are positioned. A mid-plane XY is also shown, this having a plane normal coaxial with the axis Z. The position of intersection between the plane XY and the axis Z defines an origin of the system which is also, by design, the centre of a working region of the magnet system 1.

Unless otherwise stated herein, the coils which comprise the respective magnets are of circular geometry having a centre upon the common axis and lying in a plane normal thereto (parallel with the mid-plane XY). Each coil is described by a radius “a” and a half-length “b”. The radius a defines the distance of the effective centre of the multiple windings of the coil from the common axis Z, whereas the half-length b describes the distance along the common axis Z from the origin at which the geometrical centre of the respective coil is located.

In designing the example systems according to the present invention, it will be understood that the higher the order n required for cancellation of that particular order of magnetic field derivative, the greater the ratio a/b needed. It will be further understood that the examples of magnet coils discussed herein are symmetrical about the magnetic centre, and thus in order to produce a B2 of zero for example, an ideal coil is placed at the respective position on each side of the origin. The ideal coil zero lines (for the respective order) are shown in Figure 1 as B2, B4, B6, B8.

It will be seen from Figure 1 that coils of the primary, shielding and compensation magnets are positioned at or adjacent the B2 and B8 lines. The primary magnet coils are illustrated at 2. Note that only one quadrant of the coil set is shown in detail. Thus the coil 2 encircles the axis Z and the second coil 2 (of the pair) is positioned at a similar mirrored position upon the other side of the origin. In this example the cone of rotational positions about the Z axis defining the line B2 passes through the middle of the primary magnet coil 2.

The coils of the shielding magnet are shown at 3,4. It will be noted that this comprises a set of two coils, with each having a similar radius a (in excess of the coils 2) but positioned in a split formation axially (different b). The coils 3 intersect the B2 line, whereas the coils 4 are positioned adjacent the line B2 (a short distance therefrom). Since the coils 3,4 form the shielding magnet, the current within them is counter-running with respect to that within the primary coils 2.

Turning now to the compensation magnet, this has three different coils 5,6,7, each of different a and b values. Notably, each of the coils of the compensation magnet is intersected by the B8 line. It will be understood that, with respect to the 8^{th} order of the magnetic field, very minor movements have a large effect for coils having small a and b values. Typically therefore, the B8 line intersects the coil windings of the compensation coils, although it should be noted that, due to their finite size, the B8 line may not pass through the geometrical centre of each coil section. The coils 6 and 7 each carry an electrical current having a flow sense similar to that of the primary magnet 2. The coil 5, is counter-running with respect to the coils 2,6,7.

In this example the coils are enclose in a support structure or housing 10, this having an approximate "T" shape in section (above the Z axis). The upper part or cross piece of the "T" accommodates at least the coils at or adjacent the B2 line, whereas the stem of the "T" accommodates the B8 coils. Thus the "T" has a distal part from the Z axis which has a greater axial dimension than the proximal part nearer the axis.

The purpose behind the design of the magnet systems of the present invention is to produce axially short magnets, which are economical to build and which are effectively open access in comparison with standard tubular magnets. This is shown in Figure 1 where a dotted envelope 11 illustrates the geometry required by known short solenoid magnet MRI systems.

A subject under examination (which is typically an animal, or more often a human patient) is illustrated in Figure 1. The head of a human subject 12 is shown positioned at the origin of the system 1. It will be understood from Figure 1 that the subject has line-of-sight out of the bore of the magnet system, in particular due to the axially short arrangement of the housing 10. Furthermore, the axially narrow part of the housing (the stem of the "T") enhances the open nature of the system. A person located as illustrated may therefore read a book held slightly above them (adjacent the axially short part of the housing). In addition, other measurements and interventions may be performed upon the subject located in this position because the short arrangement of the magnet now makes this possible.

The design can be arranged in this manner since the positioning of the coils on the B8 line allows very good access to the patient. The primary and shielding magnets are anchored about the B2 line on a larger diameter, and have a lower a/b ratio (which can also be thought of as a lower conic angle). This is less convenient for access, so the diameter is made large. In the present system we have arranged that the large diameter magnets comprise the coils that provide the main proportion of the MRI field intensity (more than 75% from the primary magnet), as well as the external shielding field coils (for the shielding magnet). It is known that the magnetic moments of the primary and shield magnets are similar for optimum shielding (Bm x Area m ~ Bs x Area s) and this approach is adopted here also.

An important benefit of the large diameter of the magnet anchored about B2 is that the intermediate even orders of n, B4 and B6, are well within the range of values that the compensation magnet will compensate. There are distinct benefits in designing the larger diameter magnets anchored to B2 arising from the necessity to run the shield coil with reverse current, in that the values of the field derivative B2 can be chosen to cover a wide range. In particular, the magnet anchored on B2 is physically much shorter in axial length than a standard MRI magnet, but it can produce values of B2 typical of a standard magnet. This can make the compensation of B4 and B6 errors by the smaller magnet, with larger conic angle, particularly efficient in terms of ampere turns. This is because the smaller diameter magnet does not have to "waste" its own ampere turns compensating for its own value of field derivative B2, as it compensates B4 and B6 of the larger diameter magnet.

The precision required to build the compensation magnet with the greatest conic angle (B8=0) is such that special techniques may be used. However, this precision means that a strategy for clustering coils about nth order field derivative zeros is needed to group coils in radial/axial space whilst being aware of the sensitivities of the magnetic field properties that affect the central field uniformity achieved.

It is not possible to provide very short tubular magnets using prior art techniques without reversing the current relative to the current direction for the main field to correct field derivatives up to eighth order. This can be costly in ampere turns and cause high magnetic stresses in the coils. Our innovation is to separate the coil system into groups of coils, based on two performance requirements of MRI magnets. These are that:-

i) a high field intensity is required, which for efficiency means the coils should be located close to the mid-plane of the tubular magnet. In our designs we put these on a large diameter to avoid high order field derivatives in the working region; and,

ii) a high field uniformity at the origin of the tubular magnet is also required.

The latter requirement usually means locating some coils remote from the mid-plane so as to provide possibilities for cancellation of particular field derivatives between sets of coils, without these coils producing their own high order field derivatives. However, we have shown that it is possible to place small diameter coils close to the origin of the tubular magnet, to correct those intermediate orders of field derivatives produced by the large diameter coils, without introducing high order field derivatives usually associated with small diameter coils near the mid-plane. Our basic approach to reconcile these requirements is to use both cancellation of field derivatives between, at least two coils, and to avoid values of certain field derivatives by placing some coils on or adjacent "zero locations". In particular, higher order field derivatives are also reduced to acceptable values by the use of large diameter coils.

Because it is possible to design sets of coils to control field derivatives by these means using predominantly radius as a position variable, with some use of axial position, a radially balanced magnet set can be achieved. This gives the desired short bore tube design. Adoption of coil positions placed at some naturally occurring zero points in the field derivative profile leads to an economic set of coils in which the magnitude of reverse ampere turns is at a minimum. This also provides the opportunity for effective zeroing of high order terms along conic projections from the origin of the bore tube. Considering a conic projection associated with a zero value of a high order field derivative, such as B6 or B8, the increase in the distance a coil is located from the mid-plane of the tube increases slowly for substantial increase in radius. This is useful in providing good access to the central region to provide an imaging volume with "line of sight" to the examination room.

It should be noted that the primary and active shield magnets on the smaller conic angle (with larger diameter coils) provide the main field for MRI and the external shielding. Each has only the second order field derivative controlled to be either zero, or a finite target value. Residual errors in field uniformity, described by 4^{th}, 6^{th} and 8^{th} order field derivatives are minimised as far as is practical by using large diameter coils, but achieving zero value for these orders only occurs when the fields of the first and second magnet are combined. In addition, the compensation magnet is chosen such that the radial and axial dimensions of are located on the zero of the highest order of field derivative required to be cancelled (eighth order for example).

There are three strategies for positioning the coils of the primary and shielding magnets so as to control of second order field derivative.

(i) Both the shielding magnet and primary magnet coils are located on B2=0. B4 and B6 are then cancelled by the compensation magnet that has no B8;

(ii) Both the shielding magnet coils and the primary magnet coils are axially just closer to the origin than the B2=0 line. For example the primary magnet may have a non-zero negative B2 value for a positive ampere turns ("NI) value, whereas the shielding magnet has a a positive B2 for negative NI. The sum of the coils B2 can be made to be zero. Thus the compensation magnet has only to compensate B4 and B6 while presenting no B8, which requires fewer coils in the compensation magnet; and,

(iii) Shielding magnet coils and primary magnet coils are on opposite sides of the B2=0 line in radial and axial co-ordinates. The primary coil, spaced further axially than the B2=0 line has positive B2 for a positive value of ampere turns. The shielding magnet coils (inside the line) have a positive B2 for negative NI. The sum of the coils B2 values is that normally provided by a tubular magnet of approximately twice the length, when using two coils with NI in the same sense. The conical magnet "looks" longer at its origin with respect to field uniformity described by B2 than it actually is. This reduces the ampere turn correction "load" required of the compensation magnet that compensates high order field derivatives.

The coils of the shielding magnet are placed in the vicinity of the conic angle projected from the origin of the bore tube that coincides with the lowest even order of field derivative (second order), so that adjustments of each coil in radial and axial space, along with adjustments in relative ampere turns provide controlled target values of the second order field derivative. As discussed above, it should be noted that because the largest diameter coils run in opposition to the primary magnet coils, the second order field derivative can be summed to zero by choice of current ratio and radial/axial space coordinates. It is also possible to create a sum value of second order field derivatives which is typical of a longer tubular magnet. This makes subsequent field uniformity at the origin easier to achieve when the compensation magnet is used to generate the full resultant field. In particular, the compensation magnet needs fewer ampere turns than otherwise needed, although the precision of the coil build is more demanding.

It is also a feature of the combination of magnets that the shield magnet is preferably divided into two coils, both with negative current (with respect to the primary magnet coils). The purpose of this is to achieve optimum values for the radial and axial shielding of magnetic field, while also achieving, where required, the option of a maximum value of B2. For the latter requirement, the inner shield coil is moved a controlled distance from the B2=0 projection from the magnet origin, towards the mid-plane XY of the magnet. This move also improves radial shielding efficiency.

It is preferable to arrange the compensation magnet to have at least one coil carrying current in a direction opposed to the current producing the main field. This is to compensate field values described by intermediate orders of field derivative such that the combined coil sets of the magnets have these intermediate values of field derivative compensated in sum to zero (i.e. no field non-uniformity at the origin of the tube due to these orders of field derivative). In addition, the compensation magnet has the coils arranged on the radius (from the origin of the bore tube) that coincides with a zero value for the field derivative of the highest order of field error that the magnet is intended to have near to or equal to zero. Typically this is eighth order (B8), although tenth order is feasible also, particularly using the current and geometry control technique described below.

For example, let the primary and shielding magnets be chosen at such a coil diameter that the 8th order field derivative tends to zero, and in any case the 8th order is chosen smaller than would produce unacceptable MRI image distortion. Then the compensation magnet is placed on the conic angle projected from the origin of the bore tube that coincides with the first zero value of the 8th order field derivative. Thus the compensation magnet coil set can be organised to compensate the intermediate orders of field derivative produced by the larger diameter coil set, without contributing unwanted high order field errors.

We now return to Figure 1. This provides a 30cm diameter working region 15 for imaging centred upon the origin of the bore tube. The primary magnet has a positive ampere turns values with the shielding magnet providing a negative ampere turns value. We note here that each of the primary and shielding magnet coils may be centred upon the conic projection of B2=0, this giving some B4, B6. However in the arrangement of Figure 1 the primary coils are positioned slightly closer to the origin. The shielding magnet coils are also closer to the origin. This produces an overall shorter magnet axially. The resultant B2 of the primary and shielding magnets together is cancelled by non-zero (positive and negative values of the ampere turns of the primary and shielding magnets). Again there is some residual B4,B6. In this arrangement and the arrangement where each (primary and shielding) magnet is on the B2 line the diameter is chosen such that B8 and above are close to zero. Typically therefore the primary and shielding coils have a radius of the order 80cm. The compensation magnet comprises, in this case, two positive NI coils and a small counter-running trimming coil on the smallest diameter. Each of the three coils of the compensation magnet are located on the closest to the centre conic projection for B8=0. The coils have, respectively, positive (coil 7), positive (coil 6) and negative (coil 5) ampere turns and sum B2=0. The B4 and B6 produced by the compensation magnet are equal and opposite to the combined primary and shielding magnet, B4 and B6. The compensation magnet coils are set in a radius range from 28cm to 50cm.

Thus the combined magnet system balances B2 to B6 to zero, and controls B8 to an acceptably low level. The compensation magnet may also be provided with additional coils (see below) to produce a net resultant B6 or B8 to "sharpen the imaging field of view".

Some tables of field profiles are shown in Figures 2 to 10. These are expressed in terms of field derivatives to show how a system of coils can be organised to give a uniform central field, and how grouping coils in particular regions of radial/axial space give emphasis to control of either low or high orders of field derivatives.

The tables each comprise the error term orders (calculated field derivatives) B0 to either B6, or in some tables to B10 or B12, for finite coil cross-sections (radial build and length), where the Bn (with "n" being the derivative order) are tabulated against axial position of the coil. Each table represents one coil of one mean radius. Figures 2 to 6 illustrate tables for coils configured as a primary magnet, with Figures 7 and 8 being for a shielding magnet. Note the values of B0 to Bn are for half of the magnet, these being the sets of coils along the axis of the magnet to one side of the mid-plane. All Bo values are given in Gauss, with dimensions in centimetres. The Bn values given are those taken on a 15cm radius position from the origin, that is, at the edge of the working region 15.

The coil finite dimensions are typically those of a whole body, 2 Tesla, magnet, for main field and a shield field. Cross-section (inner radius a1; outer radius a2; inner half-length b1; outer half-length b2), ampere turns (NI), turns per coil ("T coil-1"), turns per square centimetre ("T cm-2") and current (I) are appropriate examples for this type of magnet.

Each table also shows the necessary strength in terms of ampere turns (NI) required by a three-finite-coil compensation magnet. The dimensions of the compensation coils are given on the left hand side of each table (these being "L" for large radius, "M" for middle radius and "S" for small radius). Note that the same compensation coil dimensions are used for all of the included tables. The centroids of the cross-section of each coil are located close to the B8=0 line, defined in a/b terms. The compensation coil dimensions are chosen to be practical in engineering terms with convenient spacing between coils along the B8=0 line.

For each tabulated value of error term orders of a "main 2T type coil of mean radius a", against length b, there are given the necessary ampere turns (NI) of each coil of the compensation magnet required to cancel B2, B4 and B6 of a “main” coil. Tables are given for a “main” coil with a mean radius (“a mean”) of 65cm, 70, 75, 80, 90, 100 and 110 cm respectively (Figures 2 to 8). The tables show the compensation NI for the large, middle and small radius compensation coils for axial positions of a main coil. Note that the “main coil” discussed here may be a primary magnet or a shielding magnet as appropriate, although for a shielding magnet the signs of the values would be reversed since the current is reversed.

As an overview, firstly we consider the table for a chosen main coil with a given mean radius a. We then look down the table of main coil values of B2, B4, B6 against coil axial position (mean b) (fourth to sixth column). The compensation coil NI required to provide equal magnitude B2, B4, B6 but of opposite sign are shown against the b position of the main coil (see columns 11 to 13 of Figure 1 for example). Thus, it is possible to select a desired position and radius of the main coil to create a given bore field Bo, and find the "ampere turns" cost of the compensation magnet. The outcome is the engineering parameters of a combined magnet system, corrected to B8. B8 is zero in the compensation arrangement, and small for the set of main coil radii in the tables by virtue of the absolute magnitude of main coil radius, a mean. Note also that small (sub-millimetre) b changes of the inner radius compensation coil can cancel small B8 arising from the main coil, without significant changes in the NI values of the compensation coils required to compensate B2, B4, B6 of the main coil. Note further that the compensation magnet produces its own Bo value when the NI set provides the required B2, B4, B6 compensation. The combined main and compensation magnet have B2, B4, B6, B8 cancelled to zero, and the combined Bo is the signed sum of the respective value in the Bo column and that of the “Comp Bo” column. Whilst the procedure may be used to compensate a primary magnet, similarly it is used to compensate a shielding magnet (whilst taking the polarity reversal into account).

We now look at an example of using the tables in more detail.

As a first step a general decision is made about the overall magnet dimensions. In this example we select “a mean” for the main coil (primary magnet) = 75cm and “a mean” for the shield coil (shielding magnet) = 100cm. Note that the compensation three coil set for all tables has a minimum radius “a“ mean of 41.48cm. This sets the internal bore as about 40cm radius. Of course other compensation magnet dimensions could be calculated to use in look-up tables.

In a second step the main coil (primary magnet) is selected using the table with the mean radius = 75cm (Figure 4). We select the mean half-length b towards the top of the table where compensation NI are smaller and Bo from the compensation magnet is positive, for efficiency, but not the shortest possible length. For example, we take "b mean" = 54cm.

We then select the active shield coil. Using the table with "a mean" = 100cm (Figure 7), we select "b mean" towards the middle of the table where compensation NI are of similar magnitude to those selected for the main coil, and a negative value of Bo is about an average value in the tabulated list Bo. For example, we take "b mean" = 33cm. Note that the shielding magnet coils will have its NI run in opposition to the main coil, so all NI and Bn have a sign change when used as a shield.

We then combine the tabulated "Bo" and "Comp NI" properties arithmetically, thus:-

a mean	b mean	NI	Bo	CompNI L	CompNI M	Comp NI S	Bo comp
75cm	54cm	3E6	12745	-810123	1165685	-251435	2254
100cm	33cm	-1.4E6	-8092	1034356	-688391	103020	3595
Half-System total:			4653	224233	477294	-148415	5849

This tells us that the available Bo from the half-system is the sum of Main, Shield, and Compensation magnets is:-

4653 + 5849 = 10502 (Both halves = 21004)

The actual ampere turns in coils of the Compensation magnet are:

L= 224233 M= 477294 S= -148415

Figure 9 gives the details of the primary, shielding and compensation magnet coils of this example, together with all Bn values up to n=12. Note that there are slight differences in NI because the exact dimensions were recalculated. In the table of Figure 9, the axial re-adjustment of "Comp inner" to cancel B8 exactly has not been performed, but is a straight forward extra step.

In this case therefore it will be seen that the only counter-running coils are those of the shielding magnet and the innermost coil of the compensation magnet. In the lower right part of Figure 9 the homogeneity of the resultant field is shown to be about 5 parts per million at a working region radius of 15cm, and about 100 parts per million at 20cm radius.

It should be noted that the compensation magnet coils L and M both have positive NI values, and because these coils are closest to the main coil, peak fields should be low, meaning high field systems are economically practical. Note further that, by interpolation of the tables it would be possible to exactly cancel the NI of the smallest compensation coil, thereby achieving a system magnet with increased bore radius. Figure 10 shows some details of other example systems derived using a similar process.

In order to construct physical coils having the dimensions given in Figures 2 to 10, precision engineering is needed to ensure the correct size and position of each coil cross-section. This is particularly the case for the coils of the compensation magnet which are located close to the XY plane (small axial displacement). Figure 11 shows the cross-section of a coil 20 which is representative of a generic coil of the compensation magnet (although similar principles can be applied to the primary and shielding magnets). The coil 20 has a rectangular cross-section with dimensions defined by maximum and minimum radial (a2 and a1 respectively) and axial dimensions (b2 and b1 respectively). The Z direction is also indicated in Figure 11. Additional coils 21,22,23,24 are positioned upon each face of the coil 20. These are intended to have an ampere turns (NI) value which is a fraction of that of the coil 20, each typically having an NI value of 10 percent or less that of the coil 20. The coils 21,22,23,24 can be used in two possible modes. The first is to provide a net change to the effective ampere turns of the coil 20 in total, without changing the centre of the coils section as a whole. It will be appreciated that all main coils of the magnets are preferably connected in series and therefore adding currents to specific coils is not possible in such an arrangement. It is also undesirable to have separate main coil power supplies. The arrangement in Figure 11 solves this problem by allowing a current flow within the additional coils 21,22,23,24, and this may be in the same sense as that flowing in coil 20 so as to increase the strength of the magnetic field from coil assembly as a whole. The net current may also be reduced by providing a counter-running current in each of coils 21,22,23,24.

The second mode is the powering one or more individual coils of the additional coils 21,22,23,24. This may also include providing normal running current in one or more coils and none or counter-running current in others. The effect of this is to change the effective geometrical centre of the coil cross-section which provides the ability to produce very small effective movements of the coil without any physical movement being required. Thus the precision needed for the compensation coils in terms of location accuracy may be conveniently achieved. The additional coils are distributed symmetrically with respect to the cross section of each coil. Specifically, additional windings controlled by switches are placed on the inner and outer radial faces of each coil cross section in equal amount, and on the axial inner most and axial outer most longitudinal faces of each coil in equal amount. The radial and longitudinal additional coil faces may be separately adjusted. An efficient compensation of build tolerances is thereby created. In Figure 11 an additional (second) coil controller 25 is illustrated to control the current flowing in each of the additional coils. This may form part of a general controller for the other coils of the magnets, or it may be a separate device. Note that this includes suitable switching apparatus, particularly to effect the use of superconductors in persistent mode (if required).

Each coil may therefore have its net ampere turns increased by switches across the auxiliary turns of the additional coils, independently from the other coils of the set. To prevent stresses from the use of counter-running current, each of the coils of the compensation magnet is preferably organised such that with no current through its own additional coils, the coil in question provides a lower value of the field derivatives (that it contributes) than is required for full compensation. Thus in that condition the compensation magnet "under-compensates" the intermediate field derivatives of the combined primary and shielding magnets. The network of switches is therefore preferably used to increase the effective ampere turns in each coil of the compensation magnet. It should be noted that the changes in effective ampere turns do not produce pure changes in field derivatives for compensation of the field derivatives of the primary and shielding magnets. The ampere turn changes for each coil are made to adjust the sum of the field derivatives for the compensation magnet to "meet and compensate" as a whole the field derivatives of the primary and shielding magnets in combination.

In addition, the compensation of the field derivatives of the shielding and primary coils is improved as the relative strength of intermediate order field derivative for each coil of the compensation magnet may be changed. Additional field derivatives may be included in the correction list by moving the electric centres. This allows for change to be made in the volume of the working region at the origin of the tubular bore of the system. For example, if the compensation magnet coil set is placed to coincide with the first zero value of the 8th order field derivative, the volume of the uniform zone suitable for MRI is at a maximum (all other lower even orders being also zero in sum between the magnets). In some experimental situations it may be desirable to reduce the field of view derived by reducing the volume of uniformity. This might be relevant to getting "focussed" high quality images more quickly by reducing the size of the data set gathered. Allowing the electric centres of the compensation coil set to be displaced using the additional coils could provide some 8th order field derivative that would reduce the diameter of the uniform field region suitable for MRI.

We now discuss a beneficial assembly method for systems according to the invention in association with Figure 12. There are six basic steps to the assembly of the magnet system, such that the combined fields of the three magnets produce a central region suitable for high quality MRI. It is given that the magnet coils discussed below are previously wound on structural formers and encapsulated for mechanical integrity using well known vacuum impregnation techniques.

Figure 6 shows the major steps in the construction of a generic magnet system of the types described above, each of the coils being superconducting. The principle is to construct the magnet in two halves, these being two halves upon either side of the mid-plane XY.

At step 200, one set of primary coils 2 of the primary magnet are positioned horizontally upon a main support structure such that the their common axis points upwards. The coils 3,4 of the shielding magnet are likewise positioned co-axially upon the main support.

At step 201 a compensation support is added, to which coils 5,6,7 of the compensation magnet are attached. The main and compensation supports are mounted together such that all coils have a common axis orientated upwards.

At step 202, the first part of the cooling system is added, this being the helium system which provides helium coolant for each of the coils so that they may operate in superconducting mode. The helium system is shown schematically as a bold outline in Figure 12, and is illustrated at 30. This is achieved by placing the magnet coils into a bottom half of a horizontal "helium can"; followed by placing a top half of the helium can on top of the bottom half. The two halves of the helium can are then closed by welding.

At step 203 a liquid nitrogen cooling system 31 is added, this having an annular geometry analogous to that of the coils. The liquid nitrogen system comprises a liquid nitrogen coolant reservoir which is positioned at a similar axial location, although radially outwardly of the compensation magnet coils (with respect to the common axis). In this case a "can" for liquid nitrogen (for absorbing environmental heat at intermediate cryogenic temperatures) is placed horizontally on the top of the helium can, using appropriate insulating fixtures. It is also understood that appropriate nitrogen temperature (and other lower intermediate temperature) shields are assembled to surround the helium can, to further improve thermal insulation.

By adopting the storage of the cryogen within the space provided by the coils the outside diameter of the magnet system may be kept to a value similar to that established for standard MRI magnet of "long bore" tube design. Inserting the cryogen storage in this way makes good use of the space and allows easily fabricated cylindrical storage tanks to be used, along with conventional top filling service turrets.

At step 204, a second instance of the system produced by step 203 is generated, this forming the second of the two halves of the system. The two halves are then brought together such that they represent mirror images of one another across the mid-plane XY (requiring the inversion of the second). The halves are appropriately located by a fiduciary plate on the mid-plane of the magnet system.

At the final step 205, the two halves are encapsulated within a cryostat 32. The two halves are mounted horizontally in the bottom half of the vacuum cryostat structure, and appropriately fixed by a support insulation structure. The top half of the vacuum cryostat is subsequently mounted horizontally. The two halves of the vacuum cryostat are joined by welding at the mid-plane. Thereafter the system is ready for additional support apparatus to be added and it will be understood that detailed electrical and cryogenic liquid handling conduits are installed as appropriate during steps 200 to 205.

Figure 13 shows a further diagram of the magnet system, similar to that of Figure 1, although in this case with all four quadrants illustrated. It will be noted that the system is indeed significantly shorter than a more conventional tubular magnet illustrated at 11. In addition to the magnets shown in Figure 1, in this case the location of gradient coils 60 is also illustrated. The gradient coils are situated on the face of the magnet where the conic angle is largest. The gradient coils arcs contributing the main gradient field components are located on a similar diameter to standard systems, but do not intrude into the bore of the short magnet. Thus similar gradient field strength to a standard system may be achieved without adverse impact on the view to the exam room afforded the patient.

The gradient coils arcs are provided with additional turns under separate control (using suitable apparatus) so as to provide odd order field derivatives, both in an axial direction and a transverse direction. Normally only first order gradient fields are provided in scanners to reference the source of NMR signals in the imaging volume. In the system of the present invention, there are desirable uses that require the field of view to be limited to a small central volume to speed up the overall imaging process, and focus on a particular volume of interest. This is facilitated by adjustment of the values of high order field derivatives using the compensate magnet coil set on the larger conic angle (small diameter coils). This produces a focussing effect on the image field of view by reducing the volume of the uniform zone suitable for MRI. However, surrounding the uniform zone there is, for example, a spherical hollow zone with a uniformity of field insufficient for good signal to noise, containing local volumes with alias values of magnetic field. A radio frequency (rf) receive coil may be able to receive from these alias regions. It is an intention of the present invention to apply imaging pulse gradients that are non-linear in the hollow sphere surrounding the imaging field of view such that alias regions are overcome and unwanted NMR resonances avoided.

It should be noted that non-linear pulse gradients (or nmr spatial reference gradients) may be used in this invention either alone, or in conjunction with rf "preparation pulses" which can augment the avoidance of alias effects when using a focused field of imaging view inside a large diameter rf coil. For example, the outer hollow spherical volume, where the nmr resonance frequency generally has a value different to that of the central uniform zone (or "sweet spot"), may be prepared by a saturation rf pulse, before application of the spatial reference gradient pulses. Then, when the reference gradients are applied for localisation of the nmr signal, there is no magnetisation located outside the sweet spot which can be brought back into the rf band width detected for constructing image data.

It is important to note that the method of assembly shown in Figure 12 allows for the high field, open access, magnet system to be operated either vertically, or horizontally, or at any intermediate angle, without cryogenic handing problems. Thus is provided a very flexible MRI magnet, which can, for example, facilitate scanning patients either standing or lying down.

In this regard reference is made to Figure 14 which shows an example use of the system in a medical procedure. A human subject 50 is positioned upon a horizontal table and undergoes a head scan whilst the head of the subject is located within the working region of the magnet system 1. Conveniently, due to the short axial length of the magnet, it is possible for a medical practitioner 51 or technician to interact physically with the subject. In the case shown in Figure 14, the common axis of the magnet system is not horizontal, but angled with respect to the horizontal, such that the upper part of the system moves away from the feet of the subject. This, coupled with a recess in the floor where the medical practitioner or technician is standing (or alternatively by raising the scanner and subject), allows good access to the head of the subject. Conceivably, this would allow a surgeon to perform simultaneous brain surgery whilst performing an MRI scan of the brain without moving the scanner or subject. It is this sort of breakthrough in scanning practice that is now achievable as a result of the short axis of the magnet combined with the imaging capability of a conventional tubular scanner.

If, as discussed above, the magnet is tilted such that the upper half moves away from the feet of the patient, then during head scanning the patient subject has a further improvement in view to the examination room, as well as there being more space around the head of the subject for experimental equipment.

Furthermore, if the magnet system is tilted such that the upper half moves towards the feet of the patient subject 50, it becomes more convenient for a medical attendant to reach into the "imaging zone", namely the working region. This is particularly the case if the floor standing place of the medical practitioner is lowered, so the patient table and magnet are raised relative to the medical attendant.

The extremely short parallel bore tube also allows the patient table to be skewed to right or left about the origin of the magnet and the working region. The patient table is skewed to the right or left about the vertical axis of the magnet, or the magnet may be skewed with the table remaining stationary. In either case, this feature improves access for medical practitioner or technician when the MRI field of view concerns the torso or lower limbs. Figures 15 and 16 are schematic perspective views of example systems which illustrate the extremely short axial nature of the magnets, allowing the much improved access to the subject whilst providing line-of-sight outside the magnet system for the subject.

It will be further recognised that the invention provides benefits in that the working region produced by the magnet system is similar in form to that produced by a conventional tubular magnet. This means that the established imaging technology used with tubular magnet systems may be also used with magnet systems of the present invention.

In the attached Appendix further discussion is provided relating to the mathematical theory underlying the invention, together with a consideration of the accuracy of the calculations used in designing the disclosed short tubular magnets.

APPENDIX

1 Calculation accuracy

The outline magnet designs discussed herein are dependent on the calculation of the field derivatives. If we consider the variation of the field in the axial (Z) direction only, we can write down the value of B_z at any point on the axis using a Taylor expansion (see section 4):

$$B_z(z) = B_0 + B_1 z + B_2 z^2 + \ldots = \sum_{n=0}^{\infty} B_n z^n$$

where B_0 is the field at the origin and $B_n = \frac{1}{n!}\frac{\partial^n B_z}{\partial z^n}\Big|_{z=0}$

The object is then to achieve field uniformity by arranging for as many of the derivatives as possible to be zero. The design calculations assume uniform current density within a cylindrical region of conductor characterised by inner and outer radii, a_1, a_2, and the positions of the ends b_1, b_2.

Hitherto, the derivatives have been calculated using a general-purpose program which allows for non-coaxial elements of arbitrary orientation. The method is to calculate $2n+1$ values of B over a length $2r_0$ and to fit a Taylor series of nth order to these values. The accuracy of the results can be influenced by the choice of n and r_0 and also by the precision of the calculation of B. For example, r_0 must not extend into the conductor, where there would be a discontinuity in the derivatives, nor should it be so small that the variation of the nth derivative is lost in the error in B.

The field values are calculated by integrating the contributions from elementary hoops over the conductor cross-sectional area. When the field point is on axis, only one (radial) numerical integration is required, but the result is not exact. Also, the calculation depends on using library functions such as "sqrt()" and "log()" which are performed numerically. The precision of these depends on the number of bits used in the arithmetic. Using "double" floating point arithmetic (64 bit) and comparing it with "long double" (96 bit) suggests that the "double" functions used in the magnetic field calculations are accurate to 1 x 10^{-16}.

If we confine ourselves to the axial derivatives of coaxial, cylindrical systems, it is possible to arrive at exact analytical expressions for the derivatives. The method is as follows:

We can write down the on-axis field due to a hoop:

$$B_h(a,b) = \frac{\mu_0 I}{2} \frac{a^2}{(a^2+b^2)^{3/2}}$$

and integrate this over a conductor cross section to find the field of a solenoid:

$$B_s(a_1,a_2,b_1,b_2) = \int_{a_1}^{a_2} \int_{b_1}^{b_2} B_{h(a,b)dbda}$$

We now make the substitution $b_1 = b_1 - z$, $b_2 = b_2 - z$ and differentiate:

$$B_n = \frac{\partial^n}{\partial z^n} B_s(a_1,a_2,b_1 - z,b_2 - z)\Big|_{z=0}$$

This can be done with the help of computer algebra.

These expressions can now be used to compare the results of the Taylor fit to those of the analytical expressions.

As an example, take a near-Helmholtz coil with a_1 = 0.99, a_2 = 1.01, b_1 = 0.49, b_2 = 0.51 and J = 1 x 10^6 (all units amperes and metres). The values are shown in Figure 17.

This is generally fairly good, except that the 8th order is a bit unreliable, and a long way out for r_0 = 0.1, n_{max} = 16. Notably, this is for a coil of small section which is almost a hoop.

A comparison between exact and calculated “Taylor” values is shown in Figure 18.

Given that we are only interested in the order of magnitude of the 8th order, this is satisfactory, and we can have confidence in the method of calculation.

2 Practical tolerances

The engineering tolerances are concerned with the positions of the coils, the dimensions of the coils and spatial variations of the winding densities. Having evaluated the possible errors arising from these illustrate how a strategy for correcting them is devised.

2.1 Position accuracy

The effect of an axial movement on the derivations may be calculated as follows:-

The derivatives are defined as:

$$B_n = \frac{1}{n!}\frac{\partial^n B(z)}{\partial z^n}\bigg|_{z=0}$$

We can therefore recover $\frac{\partial^n H(z)}{\partial z^n}$ by replacing b with $(b-z)$ in the expression for B_n:

$$\frac{\partial^n B(z)}{\partial z^n} = f(b-z) \qquad \text{where } f(b) = n!B_n$$

$$\frac{\partial f(b-z)}{\partial b} = -\frac{\partial f(b-z)}{\partial z}$$

Therefore:

$$\frac{\partial B_n}{\partial b} = -\frac{1}{n!}\frac{\partial f}{\partial z}\bigg|_{z=0} = -\frac{1}{n!}\frac{\partial^{n+1} B(z)}{\partial z^{n+1}}\bigg|_{z=0} = -(n+1)B_{n+1}$$

Hence:

$$\delta B_n = -(n+1)B_{n+1}\delta b$$

From this, the relative field excursion at r_0 due to the error in the nth order from a shift of Δb is:

$$\Phi = -\frac{n+1}{n!} r_0^n \frac{B_{n+1}}{B_0}\delta b$$

For the case of Helmholtz coils, $b = a/2$, and writing $\rho = r_0/a$ and $\delta\beta = \delta b/a$ these are simply:

n	Φ_n
1	0
2	$11.52\rho^2\delta\beta$
3	$18.43\rho^3\delta\beta$
4	$-5.38\rho^4\delta\beta$
5	$-45.42\rho^5\delta\beta$
6	$-43.35\rho^6\delta\beta$
7	$24.54\rho^7\delta\beta$

Similarly, for an error in radius of δa and putting $\delta\alpha = \delta a / a$

$$\Phi_n = \frac{r_0^n}{n!}\frac{\partial B_n}{\partial a}\delta a$$

n	Φ_n
1	$-2.4\ \rho\delta\alpha$
2	$-1.92\rho^2\delta\alpha$
3	$2.816\rho^3\delta\alpha$
4	$6.298\rho^4\delta\alpha$
5	$2.494\rho^5\delta\alpha$
6	$-7.734\rho^6\delta\alpha$
7	$-8.611\rho^7\delta\alpha$

If we want a homogeneity of 1ppm over a radius of $a/3$ or $a/2$, the maximum error in position is then:

n	$\delta b / a$	
	$\rho = a/3$	$\rho = a/2$
1		
2	9.65×10^{-7}	3.47×10^{-7}
3	2.01×10^{-6}	4.34×10^{-7}
4	-2.29×10^{-5}	-2.97×10^{-6}
5	-9.06×10^{-6}	-7.05×10^{-7}
6	3.16×10^{-5}	1.48×10^{-6}
7	1.86×10^{-4}	5.22×10^{-6}

This implies that micron precision is needed, which is clearly very difficult.

2.2 Finite coils

The above discussion is for thin hoops. For finite or "thick" coils, errors arise from dimensional errors and also from irregularities in turns density within the cross-section of the windings. These can be represented by a superposition of the effects of a multiplicity of thin hoops, so that the above arguments hold.

Figure 19 shows a misplaced thick coil (solid line) superimposed on the correct size and position (broken line). The regions contributing to the errors are confined to the hatched areas.

3 Correction schemes

In general, the lower order gradients are relatively easy to correct using "shim" coils. First, second and third order correction coils can be made sufficiently powerful to correct those gradients in short magnets. The size of the useful working volume is determined by the higher order gradients, such as the fourth, fifth, sixth and seventh, and these are much more difficult to correct. In the very short magnets under consideration these can be significant due to the practical working tolerances.

The proposed scheme has additional correction coils associated with each main coil. Consider a main coil whose effective centre is positioned at $r = a$, $z = b$. We position a set of four correction coils around this at $(a, b - \Delta b)$, $(a, b + \Delta b)$, $(a - \Delta a, b)$, $(a + \Delta a, b)$ denoted by 1, 2, 3 and 4.

Alternatively, the four coils could be at

$(a - \Delta a, b - \Delta b)$, $(a - \Delta a, b + \Delta b)$, $(a + \Delta a, b - \Delta b)$, $(a + \Delta a, b + \Delta b)$.

The first of these arrangements is illustrated in Figure 20.

It can be seen that the correction coils are close to, or even coincident with, the regions responsible for errors. Correction is achieved by choosing the currents in the correction coils, I_1, I_2, I_3, I_4 so as to cancel the errors from the misplacement of the main coil. There are four degrees of freedom so that a linear combination can always be found to cancel at least four orders of gradients. The correct strategy is to correct the most significant high orders, for example 4th, 5th, 6th and 7th order gradients. Because the lower orders, although strong, are relatively less sensitive to the position of conductors, the discrepancy between the positions of the correction coils and the regions responsible for the errors is less significant than the higher orders. Consequently, fairly good correction is achieved for the lower orders, with imperfections being manageable by conventional shim coils.

4. Mathematical Basis of Calculations

Finally, we provide the mathematical basis for the calculations underlying the systems of the present invention.

Laplace's equation – Solution in terms of spherical harmonics

The magnetic field vector, H, can be defined in terms of a scalar potential V:

$$\vec{H} = -\nabla V$$

In the absence of an electric current, or magnetised materials

$\nabla.\vec{H} = 0$ hence $\nabla^2 = 0$ which is Laplace's equation.

In the spherical coordinate system, the complete solution to Laplace's equation is

$$V = \sum_{l,m} \left[A_{l,m}\rho^l + \frac{B_{lm}}{\rho^{l+1}} Y_l^m(\theta,\phi) \right]$$

where the functions Y_l^m are spherical harmonics.

The field and its derivatives contain no discontinuity at $\rho = 0$ so that all the coefficients $B_{lm} = 0$:

$$V = \sum_{l,m} A_{lm} \rho^l Y_l^m(\theta, \phi)$$

Associated Legendre functions and Legendre polynomials

The functions Y_l^m can be written as:

$$Y_l^m(\theta,\phi) = (-1)^m \left\{ \frac{2l+1}{4\pi} \frac{(l-m)!}{(l+m)!} \right\} \exp(im\phi) P_l^m(\cos\theta)$$

The functions P_l^m are the associated Legendre functions:

$$P_l^m(\mu) = (1-\mu^2)^{\frac{m}{2}} \frac{d^m}{d\mu^m} P_l(\mu)$$

Where the *Legendre polynomials* are:

$$P_l(\mu) = P_l^0(\mu) - \frac{1}{2^l l!} \frac{d^l}{d\mu^l} (\mu^2 - 1)^l$$

These spherical harmonics form a complete, orthonormal set, so that:

$$Y_l^m . Y_j^k = \delta_{lj} . \delta_{mk} \qquad \delta_{lj} = \begin{cases} 1 & l = j \\ 0 & l \neq j \end{cases}$$

This means that any continuous function can be expressed as a series of spherical harmonics:

$$f(\theta,\phi) = \sum_{l=0}^{\infty} \sum_{m=-l}^{l} f_{lm} Y_l^m(\theta,\phi)$$

In particular, an arbitrary magnetic potential, or its magnetic field can be expressed as a series of spherical harmonics. The first few terms of a series are generally all that is required. These are:

$Y_0^0 = \frac{1}{\sqrt{4\pi}}$	
$Y_1^0 = \sqrt{\frac{3}{4\pi}}\cos\theta$	$Y_1^1 = -\sqrt{\frac{3}{8\pi}}\exp(i\phi)\sin\theta$
$Y_2^0 = \sqrt{\frac{5}{15\pi}}(3\cos^2\theta - 1)$	$Y_2^1 = -\sqrt{\frac{15}{8\pi}}\exp(i\phi)\sin\theta\cos\theta$ $Y_2^2 = \sqrt{\frac{15}{32\pi}}\exp(2i\phi)\sin^2\theta$

The Legendre polynomials are:

$$P_0(\cos\theta) = 1$$
$$P_1(\cos\theta) = \cos\theta$$
$$P_2(\cos\theta) = \tfrac{1}{2}(3\cos^2\theta - 1)$$
$$P_3(\cos\theta) = \tfrac{1}{2}(5\cos^3\theta - 3\cos\theta)$$
$$P_4(\cos\theta) = \tfrac{1}{8}(35\cos^4\theta - 30\cos^2\theta + 3)$$
$$P_5(\cos\theta) = \tfrac{1}{8}(63\cos^5\theta - 70\cos^3\theta + 15\cos\theta)$$
$$P_6(\cos\theta) = \tfrac{1}{16}(231\cos^6\theta - 315\cos^4\theta + 105\cos^2\theta - 5)$$

The Legendre polynomials (and their derivatives) have zeros at the values of $\cos\theta$:

n	$P_n(\cos\theta) = 0$	$P'_n(\cos\theta) = 0$
1	0	
2	0.57735027	0
3	0 0.77459667	0.44721360
4	0.339981044 0.861136312	0 0.65465367
5	0 0.538469310 0.906179846	0.28523152 0.76505532
6	0.238619186 0.661209387 0.932469514	0 0.186156788 0.935113127
7	0 0.405845151 0.741531186 0.949107912	0.20929921 0.59170018 0.87174015

Application to cylindrically symmetric systems

Writing the magnetic potential as:

$$V = \sum_{l,m} A_{lm} \rho^l Y_l^m(\theta, \phi)$$

there is no ϕ variation so that $m = 0$.

Therefore:

$$V = \sum_l \tfrac{2l+1}{4\pi} A_l P_l(\cos\theta)$$

Now $\vec{H} = -\nabla V$ hence:

$$H_\rho = \sum\nolimits_l -\tfrac{2l+l}{4\pi} A_l l \rho^{l-1} P_l(\cos\theta)$$
$$H_\theta = \sum\nolimits_l -\tfrac{2l+l}{4\pi} A_l l \rho^{l-1} \sin\theta \tfrac{\theta P_l(\cos\theta)}{\theta(\cos\theta)}$$

Using the relationship:

$$\tfrac{\theta P_l(\mu)}{\theta\mu} = \tfrac{l}{1-\mu^2} P_{l-1}(\mu) - \tfrac{l\mu}{1-\mu^2} P_l(\mu)$$

$$H_\theta = \sum \tfrac{2l+1}{4\pi} A_l \rho^{l-1} \tfrac{l}{\sin\theta} [P_{l-1}(\cos\theta) - \cos\theta P_l(\cos\theta)]$$

Transform from spherical to cylindrical polar coordinates:-

$$H_\rho \cos\theta + H_\theta \sin\theta$$

$$H_z = \sum_{l=1}^{\infty} \tfrac{2l+1}{4\pi} A_l \rho^{l-1} l P_{l-2}(\cos\theta)$$

$$H_\rho \sin\theta + H_\theta \cos\theta$$

$$H_r = \sum_{l=1}^{\infty} \tfrac{2l+1}{4\pi} A_l \rho^{l-1} l [(\cos^2\theta - \sin\theta) P_l(\cos\theta) - \tfrac{\cos\theta}{\sin\theta} P_{l-1}(\cos\theta)]$$

Evaluating the coefficients – on-axis

If we consider the variation of the field in the axial (Z) direction only, we can write down the value of H_z at any point on the axis using a Taylor expansion:

$$H_z(z) = H_0 + H_1 z + H_2 z^2 + \ldots = \sum_{n=0}^{\infty} H_n z^n$$

where H_0 is the field at the origin and $H_n = \frac{1}{n!}\frac{\partial H_z}{\partial z^n}\Big|_{z=0}$

This must be identical to the previous equation using spherical harmonics, if we put $\theta = 0$ and $\rho = z$.

Hence:

$$H_z(z) = \sum_{1}^{\infty} \frac{l(2l+1)}{4\pi} A_l z^{l-1} = \sum_{0}^{\infty} H_n z^n$$

so that $A_l = \frac{4\pi}{l(2l+1)} H_{l-1}$ and:

$$H_z = \sum_{0}^{\infty} H_n \rho^n P_n(\cos\theta)$$

CLAIMS

1. A magnet system for use in magnetic resonance imaging, the magnet system being adapted when in use to generate a resultant magnetic field, within a working region, of sufficient homogeneity to enable magnetic resonance imaging to be performed, the system comprising:-

a primary magnet comprising a pair of coils positioned upon a common axis passing through the centre of the working region, the coils being spaced symmetrically about an origin upon the common axis and having dimensions such that each coil is at or adjacent a nominal position at which the second order derivative of a corresponding magnetic field in the working region is substantially zero for a coil pair located symmetrically about the origin at such a position;

a shielding magnet comprising a pair of coils positioned upon the common axis, the coils being spaced symmetrically about the origin and having dimensions such that each coil is at or adjacent a nominal position at which the second order derivative of a corresponding magnetic field in the working region is substantially zero for a coil pair located symmetrically about the origin at such a position, the shielding magnet being operative in use to reduce the resultant magnetic field at locations distal from the magnet system; and,

a compensation magnet comprising at least one pair of coils positioned upon the common axis, the coils being spaced symmetrically about the origin and having dimensions such that each coil is at or adjacent a nominal position at which at least one even order derivative of a corresponding magnetic field having a magnitude of 6 or above, in the working region, is substantially zero for a coil pair located symmetrically about the origin at such a position.

2. A system according to claim 1, wherein each nominal position is located such that the respective magnetic field derivative is zero at the origin.

3. A system according to claim 1 or claim 2, wherein each coil is defined by a circle centred upon the common axis and wherein each part of said circle is equally at or adjacent the nominal position.

4. A system according to any of claims 1 to 3, wherein the nominal position is defined by a frusto-conical surface of rotation about the common axis, wherein the said derivative of the respective magnetic field is zero for a nominal set of coils defining a circle, each part of which intersects with said surface.

5. A system according to claim 4, wherein the frusto-conical surface is defined by a cone angle, the cone angle being the angle at the cone vertex between the common axis and a line lying parallel to said surface and passing through the vertex, and wherein the cone angle for each of the pairs of coils of the compensation magnet is in excess of that of the primary magnet.

6. A system according to any of the preceding claims, wherein for the compensation magnet, the nominal position is with respect to the 8^{th} order derivative of the magnetic field of the coils at the nominal position.

7. A system according to any of the preceding claims, wherein the net current within at least one pair of coils of the compensation magnet is caused when in use to flow about the common axis in a counter-running manner with respect to the net current in the primary magnet.

8. A system according to claim 6, wherein two or more pairs of coils are provided within the compensation magnet, each pair having a different radius, and wherein the pair of coils having the smallest radius is operated when in use in the said counter-running manner.

9. A system according to any of the preceding claims, wherein the coils are at or adjacent the nominal position in that each coil has a corner of its cross-section within a distance of the nominal position not greater than the maximum radial or axial extent of the coil section, whichever is the larger.

10. A system according to any of the preceding claims, wherein at least part of the coil cross-section intersects the nominal position.

11. A system according to any of the preceding claims, wherein a pair of coils for one or more of the primary, shielding or compensation magnets, has coil windings at the respective nominal position.

12. A system according to any of the preceding claims, wherein, when in use, the primary magnet provides in excess of 75% of the resultant magnetic field strength.

13. A system according to any of the preceding claims, wherein the individual respective magnetic field from each of the primary, shielding and compensation magnets is of insufficient homogeneity within the working volume for performing magnetic resonance imaging.

14. A system according to any of the preceding claims, wherein the shielding magnet is adapted when in use to have a net current flow in the coils which is counter-running with respect to that flowing within the coils of the primary magnet.

15. A system according to claim 14, wherein the coil pair of the shielding magnet is placed adjacent the nominal second order derivative position so as to cause a non-zero magnitude second order derivative having a first polarity and wherein the coil pair of the primary magnet is placed adjacent the nominal second order derivative position so as to cause a non-zero magnitude second order derivative having a polarity opposite to the first polarity, and wherein the second order derivatives of the primary and shielding magnets are arranged in use to cancel.

16. A system according to claim 14, wherein the coil pair of the shielding magnet is placed adjacent the nominal second order derivative position so as to cause a non-zero magnitude second order derivative.

17. A system according to claim 16, wherein the shielding magnet comprises two sets of coils spaced apart along the common axis.

18. A system according to claim 17, wherein each of the sets of coils of the shielding magnet is positioned closer to the origin than the nominal second order derivative position.

19. A system according to any of the preceding claims, wherein the primary, shielding and compensation magnets are electrically connected in series, when in use.

20. A system according to any of the preceding claims, further comprising a controller adapted to control the electrical current within at least one of the primary, shielding and compensation magnets.

21. A system according to claim 20, wherein at least one of the pairs of coils of the compensation magnet comprises a set of additional coils and a second controller, wherein the second controller is adapted to control the electrical current within the additional coils independently of the said at least one pair

22. A system according to claim 20, wherein the set of additional coils are adapted to allow the positions of the magnetic centres of the said at least one pair of coils to be modified by controlling the current within the additional coils.

23. A system according to claim 21 or claim 22, wherein the set of additional coils are adapted to allow the strength of the magnetic field of the compensation magnet to be modified by controlling the current within the additional coils

24. A system according to claim 22 or claim 23, wherein at least three additional coils are provided for each coil and are distributed symmetrically about each coil of the said at least one pair.

25. A system according to claim 24, wherein four additional coils are provided for each coil of each at least one pair, the additional coils being positioned upon either side of the said at least one pair radially and axially with respect to the common axis.

26. A system according to any of claims 22 to 25, wherein the second controller is adapted in use to provide electrical current with the said coils such that the corresponding strength of the magnetic field generated by the said additional coils alone is up to about 10% of the magnetic field strength of the compensation magnet.

27. A system according to any of the preceding claims, further comprising a set of gradient coils, each gradient coil being positioned at or adjacent the nominal second order derivative position.

28. A system according to any of the preceding claims, wherein the magnetic centre of each magnet coil section defines the location of each magnet.

29. A system according to any of the preceding claims, wherein each magnet coil within the system is positioned substantially within a geometrical envelope defined by two symmetrical conical surfaces representing the second order position and extending away from the centre of the working region.

30. A system according to any of the preceding claims, wherein the primary and shielding magnets are arranged in a first part of a housing having a substantially annular geometry and wherein the compensation coils are arranged in a second part of the housing having substantially annular geometry of smaller axial and radial dimensions than the first part.

31. A system according to claim 30, wherein the second part of the housing has walls which taper in axial dimension towards the common axis.

32. A system according to any of the preceding claims, wherein the system further comprises a table upon which a subject is positioned for imaging when the system is in use, the table defining a table plane, and a system mounting adapted to allow the magnets to be rotated with respect to the table plane, such that the common axis is rotated within a plane substantially normal to the table plane.

33. A system according to any of the preceding claims, wherein the system further comprises a table upon which a subject is positioned for imaging when the system is in use, the table defining a table plane, and a system mounting adapted to allow the magnets to be rotated with respect to the table plane, such that the common axis is rotated within a plane substantially parallel to the table plane.

34. A system according to any of the preceding claims, wherein the magnet coils are superconducting coils.

35. A system according to claim 34, wherein the superconducting coils are cooled using a coolant and wherein the coolant is held within a reservoir positioned at an axial location between that of the coils of the shielding magnet.

36. A system according to claim 35, wherein the reservoir is divided into two separate reservoirs, the reservoirs being positioned upon either side of a mid-plane of the system, the said plane passing through the centre of the working region and having a plane normal parallel to the common axis.

37. A method of constructing a magnet system according to claim 33, the method comprising:-

i) mounting one coil of each pair of coils of the primary, shielding and compensation magnets to a first support;

ii) mounting a first coolant reservoir to the first support, the first coolant reservoir being adapted to supply the said one coil of each of the primary, shielding and compensation magnets with coolant during use;

iii) mounting the other coil of each pair of coils of the primary, shielding and compensation magnets to a second support;

iv) mounting a second coolant reservoir to the second support, the second coolant reservoir being adapted to supply the said other coil of each of the primary, shielding and compensation magnets with coolant during use; and,

v) mounting the first and second supports together.

38. A method according to claim 34, wherein the first support, coils and first reservoir comprise a first half of the magnet system and wherein the second support, coils and second reservoir comprise a second half of the magnet system and wherein the first and second halves are arranged as substantially mirror images of one another.

Fig.1.

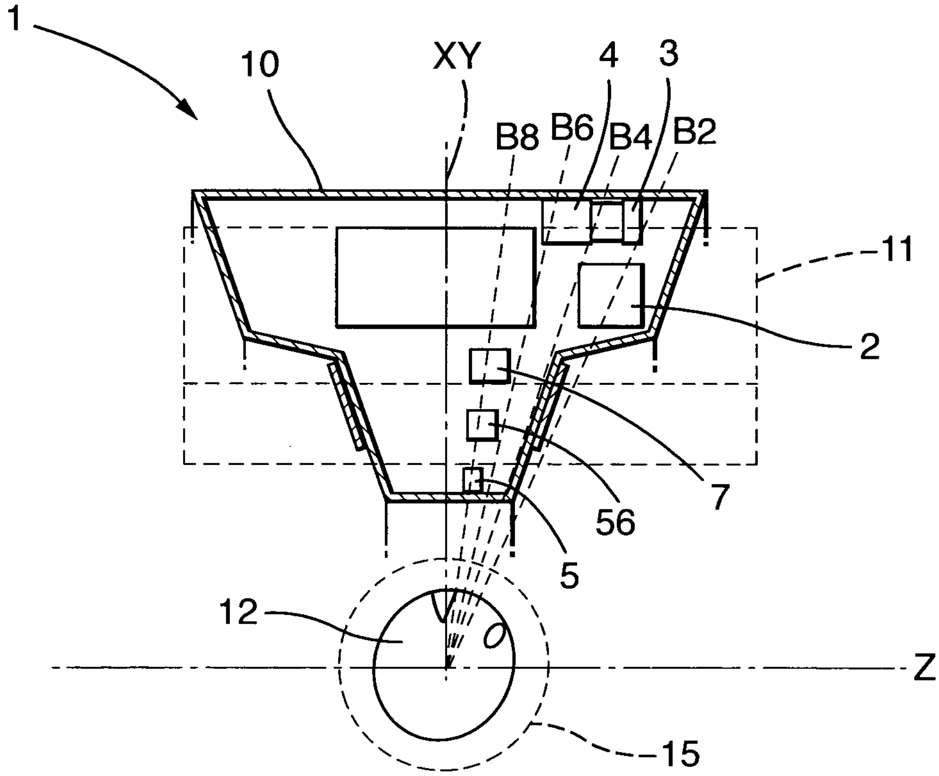


Fig.2.

a mean = 65			b mean 12 to 54		Comp -B2	+B4 -B6												
a1	a mean	a2	b1	b mean	b2	Bo	B2	B4	B6	B8	B10	B12, r=15	NI L	NI M	NI S	B0 comp	B8	B10
54	65	76	43	54	65	12536	606.5018	-28.4345	-0.5191	0.05	0	0	-568374	1403860	-360231	5974	-0.007	-0.08
Coils B2,4,6 Opposite sign to main			41	52	63	13112	596.1622	-32.831	-0.3951	0.05	-0.0003	0						
L a	60		39	50	61	13710	578.8091	-37.4358	-0.2182	0.06	-0.001	0	-1919744	2469715	-533063	2728	-0.0061	-0.11
L a1	58.78		37	48	59	14328	553.4677	-42.1543	0.0191	0.07	-0.002	0						
L a2	61.203		35	46	57	14965	519.1344	-46.8596	0.3227	0.07	-0.003	0	-3297354	3477612	-677480	-1067	-0.004	-0.12
L b	9.995		33	44	55	15621	474.8063	-51.3898	0.6959	0.07	-0.004	0						
L b1	8.796		31	42	53	16294	419.5225	-55.5458	1.1379	0.06	-0.005	0						
L b2	11.194		29	40	51	16982	352.3995	-59.0921	1.6416	0.05	-0.005	0.002						
M a	50		27	38	49	17682	272.6854	-61.7604	2.1946	0.027	-0.005	0.002	-5419856	4660727	-753407	-9242	0.006	-0.11
M a1	47.19		25	36	47	18392	179.8174	-63.2572	2.7736	-0.004	-0.006	0.0003						
M a2	52.81		23	34	45	19108	73.4801	-63.2749	3.3480	-0.044	-0.005	0.0004						
M b	8.328		21	32	43	19827	-46.3239	-61.5089	3.8779	-0.09	-0.003	0.0004						
M b1	5.552		19	30	41	20545	-176.2131	-57.6789	4.3159	-0.14	-0.0003	0.0003	-5669698	3932636	-442480	-16011	0.019	-0.02
M b2	11.104		17	28	39	21255	-324.3661	-51.5530	4.6102	-0.19	0.003	0.0002						
S a	41.48		15	26	37	21954	-480.4736	-42.9779	4.7086	-0.24	0.008	0						
S a1	40		13	24	35	22637	-645.7048	-31.9018	4.5641	-0.28	0.01	-0.0003						
S a2	42.96		11	22	33	23296	-817.6991	-18.4040	4.1416	-0.29	0.01	-0.0006	-4431914	2014350	-24674	-20622	0.02	0.07
S b	6.9115		9	20	31	23926	-993.5798	-2.7105	3.4236	-0.29	0.02	-0.0008						
S b1	5.43		7	18	29	24520	-1170.0070	14.7960	2.4156	-0.26	0.01	-0.0009						
S b2	8.393		5	16	27	25074	-1343.2539	33.5793	1.1495	-0.2	0.01	-0.0009						
			3	14	25	25579	-1509.3276	52.9669	-0.3163	-0.12	0.01	-0.0007	-3569631	985060	-10874	-24162	0.007	0.02
			1	12	23	26031	-1664.1043	72.1801	-1.8995	-0.02	0.006	-0.0004						
			-1	10	21	26432	-1803.5006	90.3794	-3.5015	0.09	-0.0005	-0.0007						
			-3	8	19	26752	-1923.6482	106.7153	-5.0122	0.2	-0.007	0.0003						
			-5	6	17	27012	-2021.0789	120.3869	-6.3237	0.3	-0.013	0.0005	-3867699	1423842	-376059	-27177	-0.02	-0.11
			-7	4	15	27200	-2092.8872	130.6990	-7.3389	0.3772	-0.01762	0.0007						
			-9	2	13	27314.996	-2136.8804	137.1138	-7.981	0.427	-0.0206	0.0008						
			-11	0	11	27353	-2151.6992	139.2908	-8.2007	0.4442	-0.021	0.0009	-4088757	1710463	-520957	-28050	-0.026	-0.15

Fig.3.

Finite coil B0 B4 B6 along axis of model															
NI = 3E6 per coil, I = 588 turns = 4840 @ 10Tcm-2 delta b ~22cm 20pnt b/a=1 per coil															
NOTE B0 = x 2 for full magnet						M+AS +B2 -B4 +B6									
a mean = 70			b mean 10 to 50			Comp -B2 +B4 -B6									
a1	a mean	a2	b1	b mean	b2	Bo	B2	B4	B6 15cm r	NI L	NI M	NI S	B0 comp	B8	B10
			41	52	63	13220	447.79	-28.6119	-0.0173	-1237570	1677470	-361221	2532	-0.004	-0.07
59	70	81	39	50	61	13767	423.63	-31.6457	0.142						
Coils B2,4,6 Opposite sign to main			37	48	59	14329	392.73	-34.6021	0.3368	-2237011	2398052	-464830	-332	-0.002	-0.08
L a	60		35	46	57	14905	354.43	-37.3774	0.5673						
L a1	58.78		33	44	55	15494	308.18	-39.8486	0.8316	-3171620	3001509	-535341	-3477	0	-0.09
L a2	61.203		31	42	53	16094	253.43	-41.8749	1.125						
L b	9.995		29	40	51	16703	189.76	-43.3004	1.4392	-3935302	3395687	-557083	-6717	0.003	-0.08
L b1	8.796		27	38	49	17318	116.86	-43.9578	1.7623						
L b2	11.194		25	36	47	17938	34.6108	-43.6757	2.0782	-4437979	3512159	-521990	-9855	0.007	-0.07
M a	50		23	34	45	18559	-56.9436	-42.2863	2.3668						
M a1	47.19		21	32	43	19177	-157.496	-39.637	2.6051	-4636775	3335558	-435571	-12726	0.01	-0.04
M a2	52.81		19	30	41	19790	-266.478	-35.6016	2.768						
M b	8.328		17	28	39	20394	-383.0327	-30.0944	2.8302						
M b1	5.552		15	26	37	20984	-505.9788	-23.0833	2.7685						
M b2	11.104		13	24	35	21556	-633.8229	-14.6021	2.5643	-4297766	2391266	-207451	-17387	0.01	0.009
S a	41.48		11	22	33	22105	-764.7578	-4.76	2.2066						
S a1	40		10	21	32	22375	-830.7398	0.6149	1.9694						
S a2	42.96		9	20	31	22628	-896.6878	6.2519	1.6938						
S b	6.9115		7	18	29	23118	-1027.267	18.1601	1.0364	-3860725	1700761	-122012	-20076	0.009	0.01
S b1	5.43		5	16	27	23572	-1153.964	30.617	0.2518						
S b2	8.393		3	14	25	23985	-1274.125	43.2063	-0.6103						

Fig.4.

Finite coil B0 B4 B6 along axis of model							,+Comp coil balancing NI											
NI = 3E6 per coil, I = 588 turns = 4840 @ 10Tcm-2 delta b ~22cm 20pnt b/a=1 per coil										M+AS +B2 -B4 +B6								
NOTE B0 = x 2 for full magnet										Comp -B2 +B4 -B6								
a mean = 75			b mean 12 to 54															
a1	a mean	a2	b1	b mean	b2	Bo	B2	B4	B6	B8	B10	B12, r=15	NI L	NI M	NI S	Bo comp	B8	B10
			51	62	73	10926.0	395.68710	-14.14570	-0.18820	0.01246	0.00001	-0.00001	554251	106691	-79200	5705	-0.0033	-0.0230
			49	60	71	11360.0	389.74060	-16.01760	-0.14870	0.01442	-0.00007	-0.00001						
			47	58	69	11809.0	380.37480	-17.96820	-0.09491	0.01628	-0.00016	-0.00001	-91359	620573	-165308	4159	-0.0031	-0.0370
			45	56	67	12270.0	367.17570	-19.96860	-0.02490	0.01787	-0.00028	-0.00001						
64	75	86	43	54	65	12745.0	349.71650	-21.98160	0.06281	0.01901	-0.00043	-0.00001	-810123	1165685	-251435	2254	-0.0025	-0.0500
Set B2,4,6 equal B opposite sign			41	52	63	13232.0	327.56820	-23.96010	0.16940	0.01946	-0.00059	0.00000						
Coil			39	50	61	13731.0	300.30960	-25.84760	0.29530	0.01897	-0.00076	0.00000	-1561493	1701344	-329209	29	-0.0002	-0.0500
L a	60		37	48	59	14241.0	267.54020	-27.57660	0.44000	0.01726	-0.00093	0.00001						
L a1	58.78		35	46	57	14760.0	228.89160	-29.07020	0.60180	0.01407	-0.00108	0.00002	-2258899	2175756	-388904	-2437	-0.0001	-0.0600
L a2	61.203		33	44	55	15287.0	184.04730	-30.24230	0.77740	0.00915	-0.00118	0.00003						
L b	9.995		31	42	53	15820.0	132.75859	-30.99930	0.96195	0.00236	-0.00121	0.00004	-2927894	2531298	-420755	-5030	0.0020	0.0200
L b1	8.796		29	40	51	16359.0	74.86590	-31.24320	1.14870	-0.00635	-0.00113	0.00005						
L b2	11.194		27	38	49	16899.0	10.31880	-30.87470	1.32890	-0.01685	-0.00093	0.00006	-3422859	2726770	-419671	-7633	0.0040	-0.0500
M a	50		25	36	47	17441.0	-60.80220	-29.79825	1.49230	-0.02882	-0.00058	0.00006						
M a1	47.19		23	34	45	17980.0	-138.26610	-27.92710	1.62698	-0.04170	-0.00009	0.00006	-3759374	2765005	-392039	-10153	0.0060	-0.0400
M a2	52.81		21	32	43	18514.0	-221.67230	-25.19000	1.72000	-0.05470	0.00055	0.00004						
M b	8.328		19	30	41	19040.0	-310.43620	-21.53800	1.75870	-0.06681	0.00128	0.00002	-3873089	2600252	-332134	-12411	0.0080	-0.0300
M b1	5.552		17	28	39	19556.0	-403.77820	-16.95100	1.73110	-0.07687	0.00205	-0.00002						
M b2	11.104		15	26	37	20057.0	-500.71994	-11.44410	1.62725	-0.08365	0.00278	-0.00006						
S a	41.48		13	24	35	20540.0	-600.08783	-5.07302	1.44044	-0.08601	0.00338	-0.00010	-3824535	2203633	-242784	-15406	0.0090	-0.0100
S a1	40		11	22	33	21001.0	-700.52454	2.06252	1.16845	-0.08300	0.00375	-0.00013						
S a2	42.96		9	20	31	21438.0	-800.50916	9.81755	0.81433	-0.07402	0.00382	-0.00015						
S b	6.9115		7	18	29	21846.0	-898.39240	18.00363	0.03870	-0.05898	0.00354	-0.00016						
S b1	5.43		5	16	27	22223.0	-992.43152	26.39354	-0.01930	-0.03833	0.00289	-0.00014	-3506653	1604849	-169673	-18366	0.0050	-0.0100
S b2	8.393		3	14	25	22564.0	-1080.84009	34.72920	-0.62188	-0.01311	0.00190	-0.00011						
			1	12	23	22867.0	-1161.84680	42.73259	-1.15803	0.01512	0.00066	-0.00006						
			-1	10	21	23128.0	-1233.74756	50.11907	-1.67904	0.04439	-0.00073	-0.00001						
			-3	8	19	23346.0	-1294.97241	56.61201	-2.15595	0.07249	-0.00213	0.00006						
			-5	6	17	23518.0	-1344.13977	61.95837	-2.56098	0.09719	-0.00340	0.00011						
			-7	4	15	23641.0	-1380.11511	65.94305	-2.86971	0.11648	-0.00442	0.00016	-3711813	1666891	-299000	-21560	-0.0060	-0.0600
			-9	2	13	23716.0	-1402.04773	68.40247	-3.06305	0.12875	-0.00507	0.00018						
			-11	0	11	23741.0	-1409.41809	69.23387	-3.12889	0.13296	-0.00530	0.00019						

Fig.5.

Finite coil B0 B4 B6 B8 along axis of model																
NI = 3E6 per coil, I = 588 turns = 4840 @ 10Tcm-2 delta b ~22cm 20pnt b/a=1 per coil																
NOTE B0 = x 2 for full magnet					M+AS +B2 -B4 +B6											
a mean = 80			b mean 10 to 50		Comp -B2 +B4 -B6											
a1	a mean	a2	b1	b mean	b2	Bo	B2	B4	B6	B8 r=15cm	NI L	NI M	NI S	B0 comp	B8	B10
			53	64	75	10649	321.6468	-11.5938	-0.09548	0.008	506202	30832	-50893	4719	-0.02	-0.16
			51	62	73	11043	314.5378	-12.9182	-0.06389	0.009						
			49	60	71	11448	304.6723	-14.2761	-0.02334	0.009						
			47	58	69	11864	291.761	-15.6461	0.027	0.01	-257287	620313	-147230	2751	-0.02	-0.03
			45	56	67	12290	275.5121	-17.00126	0.08785	0.01						
			43	54	65	12726	255.6371	-18.30903	0.15945	0.0108						
69	80	91	41	52	63	13171	231.8600	-19.5313	0.24170	0.0102	-1117106	1236232	-239309	195	-0.001	-0.04
Coils B2,4,6 Opposite sign to main			39	50	61	13624	203.9100	-20.6241	0.33410	0.0089						
L a	60		37	48	59	14084	171.5700	-21.5384	0.43530	0.0068						
L a1	58.78		35	46	57	14551	134.6427	-22.2202	0.54310	0.0038	-1966171	1778197	-308304	-2807	0.0006	-0.05
L a2	61.203		33	44	55	15022	92.9764	-22.6122	0.65470	-0.0001						
L b	9.995		31	42	53	15496	46.4924	-22.6551	0.76600	-0.005						
L b1	8.796		29	40	51	15973	-4.8174	-22.2896	0.87240	-0.0107	-2679033	2141449	-337978	-6025	0.003	-0.04
L b2	11.194		27	38	49	16449	-60.8741	-21.4593	0.96800	-0.0171						
M a	50		25	36	47	16923	-121.4995	-20.1135	1.04670	-0.0239						
M a1	47.19		23	34	45	17392	-186.4059	-18.2106	1.10150	-0.031	-3158836	2262897	-324313	-9185	0.0048	-0.04
M a2	52.81		21	32	43	17855	-255.1868	-15.7216	1.12570	-0.037						
M b	8.328		19	30	41	18309	-327.3109	-12.6338	1.11290	-0.043						
M b1	5.552		17	28	39	18751	-402.1199	-8.9544	1.05760	-0.0467						
M b2	11.104		15	26	37	19179	-478.8289	-4.7131	0.95390	-0.049						
S a	41.48		13	24	35	19590	-556.5346	0.0359	0.80600	-0.048	-3426524	2020191	-251613	-13745	0.006	-0.023
S a1	40		11	22	33	19981	-634.2247	5.2127	0.60860	-0.044						
S a2	42.96		10	21	32											
S b	6.9115		9	20	31	20349	-710.7973	10.7120	0.36720	-0.038						
S b1	5.43		7	18	29	20692	-785.0805	16.4056	0.08840	-0.028						
S b2	8.393		5	16	27	21008	-855.8640	22.1449	-0.21810	-0.015						
			3	14	25	21293	-921.9249	27.7674	-0.54010	-0.001	-3380552	1674129	-221692	-17019	0.002	-0.028

Fig.6.

Finite coil B0 B4 B6 along axis of model							,+Comp coil balancing NI											
NI = 3E6 per coil, I = 588 turns = 4840 @ 10Tcm-2 delta b ~22cm						20pnt b/a=1 per coil				M+AS +B2	-B4 +B6							
NOTE B0 = x 2 for full magnet										Comp -B2	+B4 -B6							
a mean = 90			b mean 12 to 54															
a1	a mean	a2	b1	b mean	b2	Bo	B2	B4	B6	B8	B10	B12, r=15	NI L	NI M	NI S	Bo comp	B8	B10
			57	68	79	10094	216.31874	-7.75327	-0.01532	0.00335	-0.00003	0.00000	408810	-44708	-19495	3286	-0.0013	-0.0076
			55	66	77	10420	208.83997	-8.43066	0.00309	0.00358	-0.00004	0.00000	258806	69718	-38299	2885	-0.0013	-0.011
			53	64	75	10754	199.56253	-9.10625	0.02502	0.00373	-0.00006	0.00000						
			51	62	73	11095.0	188.34288	-9.76832	0.05070	0.00379	-0.00008	0.00000	-69049	312944	-77209	1956	-0.0010	-0.0160
			49	60	71	11443.0	175.03976	-10.40318	0.08012	0.00371	-0.00010	0.00000						
			47	58	69	11797.0	159.51811	-10.99515	0.11322	0.00348	-0.00012	0.00000	-426169	567256	-116240	865	-0.0008	-0.0200
			45	56	67	12156.0	141.65187	-11.52657	0.14978	0.00306	-0.00014	0.00000						
79	90	101	43	54	65	12520.0	121.32799	-11.97800	0.18936	0.00243	-0.00015	0.00000	-800160	820865	-153195	-373	-0.0004	-0.0260
Set B2,4,6 equal B opposite sign			41	52	63	12889.0	98.45067	-12.32830	0.23135	0.00155	-0.00016	0.00000						
Coil			39	50	61	13262.0	72.94547	-12.55516	0.27485	0.00042	-0.00016	0.00000	-1175891	1060392	-185788	-1736	0.0002	-0.0300
L a	60		37	48	59	13636.0	44.76447	-12.63541	0.31875	-0.00098	-0.00016	0.00000						
L a1	58.78		35	46	57	14013.0	13.89091	-12.54570	0.36165	-0.00263	-0.00014	0.00001						
L a2	61.203		33	44	55	14390.0	-19.65562	-12.26323	0.40192	-0.00450	-0.00011	0.00001						
L b	9.995		31	42	53	14766.0	-55.81450	-11.76659	0.43770	-0.00654	-0.00006	0.00001						
L b1	8.796		29	40	51	15140.0	-94.47944	-11.03680	0.46696	-0.00868	0.00000	0.00001	-2017325	1514594	-236499	-5456	0.0017	-0.0330
L b2	11.194		27	38	49	15511.0	-135.49454	-10.05836	0.48758	-0.01083	0.00007	0.00000						
M a	50		25	36	47	15877.0	178.65073	-8.82035	0.49743	-0.01287	0.00015	0.00000						
M a1	47.19		23	34	45	16237.0	-223.68391	-7.31761	0.49449	-0.01466	0.00024	0.00000						
M a2	52.81		21	32	43	16589.0	-270.27347	-5.55179	0.47698	-0.01608	0.00330	0.00000						
M b	8.328		19	30	41	16931.0	-318.04285	-3.53228	0.44352	-0.01696	0.00041	0.00000	-2573901	1680683	-240419	-9105	0.0028	-0.0310
M b1	5.552		17	28	39	17262.0	-366.56171	-1.27699	0.39321	-0.01719	0.00048	-0.00001						
M b2	11.104		15	26	37	17580.0	-415.34915	1.18726	0.32586	-0.01664	0.00053	-0.00001						
S a	41.48		13	24	35	17883.0	-463.88049	3.82457	0.24201	-0.01525	0.00055	-0.00001						
S a1	40		11	22	33	18169.0	-511.59308	6.59027	0.14305	-0.01990	0.00053	-0.00002						
S a2	42.96		9	20	31	18438.0	-557.89862	9.43170	0.03122	-0.00988	0.00048	-0.00002	-2832249	1628540	-225232	-12109	0.0023	-0.0280
S b	6.9115		7	18	29	18686.0	-602.19177	12.28935	-0.09037	-0.00600	0.00039	-0.00002						
S b1	5.43		5	16	27	18913.0	-643.86438	15.09853	-0.21789	-0.00152	0.00026	-0.00001						
S b2	8.393		3	14	25	19117.0	-682.31726	17.79142	-0.34689	0.00338	0.00010	-0.00001						
			1	12	23	19297.0	-716.97894	20.29948	-0.47251	0.00840	-0.00007	0.00000						
			-1	10	21	19452.0	-747.31390	22.55589	-0.58974	0.01339	-0.00025	0.00000	-2921923	1534204	-220720	-14082	0.0007	-0.0320
			-3	8	19	19580.0	-772.84247	24.49837	-0.69367	0.01794	-0.00042	0.00001						
			-5	6	17	19680.0	-773.15198	26.07169	-0.77980	0.02181	-0.00057	0.00001						
			-7	4	15	19753.0	-807.90637	27.23005	-0.84429	0.02477	-0.00069	0.00002						
			-9	2	13	19796.0	-816.85986	27.93922	-0.88421	0.02663	-0.00077	0.00002						
			-11	0	11	19811.0	-819.86163	28.17803	-0.89772	0.02726	-0.00079	0.00002	-2944185	1498266	-223099	-14774	-0.0003	-0.0350

Fig.7.

AS/main 45%															
NI = 1.4E6 per coil, I = 588 turns = 2560 @ 10Tcm-2 delta b ~16cm 20pnt b/a=1 per coil															
NOTE B0 = x 2 for full magnet						*NOTE sign reverse of Bn symmetrically reverses sign NI*									
a mean = 100			b mean 0 to 60		*NOTE*	*-5883*	*-56.0719*	*3.6993*	*-0.04281*	*118351*	*-172200*	*35742*	*-366*	*0.0002*	*0.007*
a1	a mean	a2	b1	b mean	b2	Bo	B2	B4	B6 15cm r	NI L	NI M	NI S	B0 comp	B8	B10
92	100	108	53	61	69	5883	56.0719	-3.6993	0.04281	-118351	172200	-35742	366	-0.0002	-0.007
Coils B2,4,6 Opposite sign to main			51	59	67	6040	43.5854	-3.8159	0.05216						
L a	60		49	57	65	6200	35.2854	-3.9049	0.0619	-294687	290420	-53551	-239	-0.00001	-0.0009
L a1	58.78		47	55	63	6360	26.1528	-3.9607	0.072						
L a2	61.203		45	53	61	6521	16.176	-3.9781	0.0821	-434123	379426	-66194	-751	0	-0.01
L b	9.995		43	51	59	6683	5.3514	-3.9517	0.0921						
L b1	8.796		41	49	57	6845	-6.3255	-3.8763	0.1015	-572331	462791	-77439	-1299	0.00002	-0.01
L b2	11.194		39	47	55	7007	-18.8083	-3.7467	0.11						
M a	50		37	45	53	7168	-32.0994	-3.5585	0.1174						
M a1	47.19		35	43	51	7328	-46.1485	-3.3079	0.1231						
M a2	52.81		33	41	49	7486	-60.9022	-2.9916	0.1267	-828111	600696	-94296	-2452	0.00006	-0.01
M b	8.328		31	39	47	7642	-76.2928	-2.6079	0.1279						
M b1	5.552		29	37	45	7796	-92.2386	-2.1569	0.1264						
M b2	11.104		27	35	43	7946	-108.6432	-1.637	0.1217						
S a	41.48		25	33	41	8092	-125.3958	-1.0532	0.1137	-1034356	688391	-103020	-3595	0.00008	-0.01
S a1	40		23	31	39	8234	-142.3721	-0.4088	0.1024						
S a2	42.96		22	30	38	8303	-150.9016	-0.0659	0.0949						
S b	6.9115		21	29	37	8371	-159.4345	0.2897	0.0868						
S b1	5.43		19	27	35	8502	-176.4345	1.0345	0.068	-1149126	721875	-105450	-4380	0.00009	-0.01
S b2	8.393		17	25	33	8627	-193.2129	1.8153	0.0459						
			15	23	31	8745	-209.6039	2.6205	0.0209						
			13	21	29	8855	-225.4354	3.4369	-0.0065						
			11	19	27	8958	-240.5338	4.2496	-0.0357	-1251803	735578	-106039	-5256	0.00008	-0.01
			9	17	25	9052	-254.7262	5.0433	-0.0659						
			7	15	23	9137	-267.843	5.8019	-0.0961						
			5	13	21	9212	-279.7231	6.5092	-0.1255						
			3	11	19	9278	-290.2158	7.1496	-0.153	-1309196	732839	-106210	-5868	0.00005	-0.015
			1	9	17	9333	-299.1846	7.7086	-0.1778						
			-1	7	15	9377	-306.5107	8.1729	-0.1986						
			-3	5	13	9411	-312.0988	8.5316	-0.2151						
			-5	3	11	9433	-315.8615	8.7757	-0.2265	-1333380	726730	-106559	-6166	0.00004	-0.015
			-7	1	9	9445	-317.7578	8.8993	-0.2322						
			-8	0	8	9446	-317.9951	8.9149	-0.233	-1335213	728293	-106594	-6190	0.00003	-0.015
			-9	-1	7	9445	-317.7578	8.8993	-0.2322						

Fig.8.

AS/main 45%																
NI = 1.4E6 per coil, I = 588 turns = 2560 @ 10Tcm-2 delta b ~16cm 20pnt b/a=1 per coil																
NOTE B0 = x 2 for full magnet																
a mean = 110			b mean 0 to 60													
a1	a mean	a2	b1	b mean	b2	Bo	B2	B4	B6	B8 r=15cm	NI L	NI M	NI S	B0 comp	B8	B10
			59	67	75	5355	38.1886	-2.3022	0.02216	0.00025	-52976	94826	-20667	310	-0.00019	-0.004
			57	65	73	5485	33.07529	-2.36833	0.02653	0.0002	-99366	126518	-25542	155	-0.0001	-0.0047
			55	63	71	5616	27.45694	-2.42029	0.03109	0.00013						
102	110	118	53	61	69	5749	21.3225	-2.4557	0.0358	0.00050	-196144	191103	-35290	-180	0.0000	-0.006
Coils B2,4,6 Opposite sign to main			51	59	67	5882	14.6647	-2.4719	0.0405	-0.00005						
L a	60		49	57	65	6016	7.4797	-2.4666	0.0452	-0.00016	-296633	255919	-44814	-547	0.0000	-0.007
L a1	58.78		47	55	63	6149	-0.2309	-2.4372	0.0497	-0.00029						
L a2	61.203		45	53	61	6283	-8.4610	-2.3812	0.0539	-0.00040	-398615	319193	-53839	-941	0.0000	-0.009
L b	9.995		43	51	59	6417	-17.1986	-2.2966	0.0577	-0.00060						
L b1	8.796		41	49	57	6550	-26.4257	-2.1811	0.0610	-0.00073	-499544	378997	-62076	-1354	0.0002	-0.009
L b2	11.194		39	47	55	6682	-36.1175	-2.0331	0.0636	-0.00090						
M a	50		37	45	53	6613	-46.2423	-1.8513	0.0651	-0.00100	-597286	433964	-69380	-1780	0.0003	-0.010
M a1	47.19		35	43	51	6942	-56.7615	-1.6347	0.0658	-0.00120						
M a2	52.81		33	41	49	7069	-67.6288	-1.3831	0.0652	-0.00130						
M b	8.328		31	39	47	7194	-78.7906	-1.0960	0.0633	-0.00140						
M b1	5.552		29	37	45	7316	-90.1856	-0.7767	0.0599	-0.00150						
M b2	11.104		27	35	43	7434	-101.7461	-0.4248	0.0552	-0.00150	-813613	542267	-82779	-2844	0.0005	-0.010
S a	41.48		25	33	41	7549	-113.3965	-0.0435	0.0488	-0.00150						
S a1	40		23	31	39	7661	-125.0556	0.3638	0.0410	-0.00150						
S a2	42.96		22	30	38											
S b	6.9115		21	29	37	7767	-136.6366	0.7929	0.0317	-0.00130						
S b1	5.43		19	27	35	7869	-148.0486	1.2388	0.0209	-0.00120	-947340	596754	-88951	-3621	0.0006	-0.010
S b2	8.393		17	25	33	7965	-159.1958	1.6958	0.0090	-0.00100						
			15	23	31	8056	-169.9813	2.1576	-0.0039	-0.00075						
			13	21	29	8141	-180.3066	2.6173	-0.0176	-0.00040						
			11	19	27	8219	-190.0734	3.0675	-0.0317	-0.00009						
			9	17	25	8291	-199.1865	3.5009	-0.0458	0.00030	-1059927	631378	-92796	-4391	0.0006	-0.013
			7	15	23	8355	-207.5528	3.9098	-0.0598	0.00070						
			5	13	21	8413	-215.0858	4.2869	-0.0730	0.00100						
			3	11	19	8462	-221.7047	4.6252	-0.0852	0.00140						
			1	9	17	8504	-227.3380	4.9181	-0.0960	0.00170						
			-1	7	15	8537	-231.9229	5.1598	-0.1051	0.00200	-1117449	643645	-94422	-4845	0.0005	-0.013
			-3	5	13	8563	-235.4079	5.3455	-0.1122	0.00230						
			-5	3	11	8580	-237.7538	5.4714	-0.1171	0.00240						
			-7	1	9	8588	-238.9334	5.5351	-0.1195	0.00250						
			-8	0	8	8589	-239.0812	5.5431	-0.1199	0.00250	-1128875	645622	-94756	-4941	0.0004	-0.014
			-9	-1	7	8588	-238.9333	5.5351	-0.1195	0.00250						

Fig.9.

Finite 8th order CYCLOPS Main outside B2=0, Shield inside B2=0									
			pinched OD/ID						
Look up Tables		**80cm clear bore (coils) max OD "200cm"**							
Step 1 use tables to get Net B0 for Compensation NI investment B2,4,6=0 0n B8=0									
Main + its required comp, AS + its required comp = Net B0 and Net NI comps									
Step 2 fine tune B8 between AS+M and net Comp coils									
a mean= 75	b index	Bo Main	NI compL	NI compM	NI compS	Bo comp			
75	43/54/65	12745	-810123	1165685	-251435	2254			
		NI=3E6							
		Bo AS							
100	25/33/41	8092	-1034356	688391	-103020	-3595			
		NI=1.4E6							
change sign all NI		-8092	1034356	-688391	103020	3595			
Sum 1/2 magnet		4653	224233	477294	-148415	5849			
Sum Bo=10502									
Main outside B2=0, AS inside B2=0				20pnts calc					
20pnt calc a/b=1					Compensation 8th order magnet				
Coil	Main	AS		Outer	Middle	Inner			
a	75	100		60	50	41.48			
a1	64	92		58.78	47.19	40			
a2	86	108		61.203	52.81	42.96			
b	54	33		9.995	8.328	6.9115			
b1	43	25		8.796	5.552	5.43			
b2	65	41		11.194	11.104	8.393	ex look up tables		
Tcoil-1	4840	2560		381.029	811.959	252.4	*381.35*	*811.72*	*252.41*
Tcm-2	10	10		65.578	26.022	28.778	*66.207*	*26.834*	*28.045*
NI	3000000	1400000		224045.1	477432.3	-148435.2	*224233*	*477294*	*-148415*
I	588	-588		588	588	-588			
Coil sec	22x22	16x16		2.4x2.4	5.5x5.5	3x3			
			Target	Comp NI recalculated against target				**Comp Tot**	**Total**
B0 1/2mag	12745.0000	-8092	4652.0000	2251.0000	5755.0000	-2157.0000		**5850.0000**	**10502.0000**
B2 r=15	349.7165	125.3968	**475.1122**	-177.7114	-653.3988	355.9465		**-475.1637**	**0.0515**
B4	-21.9816	1.05316	**-20.9284**	9.9538	52.6717	-41.6906		**20.9348**	**0.0064**
B6	0.0628	-0.11366	**-0.0508**	-0.3784	-2.8862	3.3148		**0.0503**	**-0.0005**
B8	0.0190	0.00358	0.0226	0.0007	0.0092	-0.0133		**-0.0034**	**0.0192**
B10	-0.0004	-0.00007	-0.0005	0.0016	0.0255	-0.0627		**-0.0355**	**-0.0360**
B12	0.0000	0	0.0000	-0.0002	-0.0047	0.0166		**0.0117**	**0.0117**
				No axial shuffle to reduce B8					5ppm r=15
									100ppm r=20

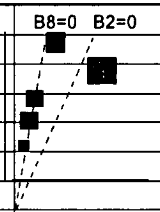


Fig.10.

Finite 8th order CYCLOPS Main outside B2=0, Shield inside B2=0								
Expanded all coil dimension by 1.48 x to get min coil am a mean 40cm								
This is intended to remove B10 B12 B14 by diameter effect								
FOV 50cm dsv								
After step 4		**Main outside B2=0, AS inside B2=0**						
Coil	Main	AS		Outer	Middle	Inner 8th nudge		
a	70	110		50	36	28		
a1	59	102		49	34	27		
a2	81	118		51	38	29		
b	42	25		8.33	6	4.665		
b1	31	17		7.33	4	3.665		
b2	53	33		9.33	8	5.665		
Tcoil-1	4840	2560		14.029	276.174	42.282		
Tcm-2	10	10		3.507	17.261	10.571		
NI 1/2 mag	3000000	1400000		8249.27	162390.2	24861.95		**4595500**
I	588	-588		588	588	-588		
Coil sec	22x22	16x16	Target	2x2	4x4	2x2	Comp sum	**Combined**
B0 1/2mag	16094.0000	-7965	Float	99.0000	2718.0000	-535.0000	2282.0000	**10411.0000**
B2 r=15	253.4339	159.1958	-412.6297	-11.3009	-595.2066	193.8823	-412.6251	**0.0046**
B4	-41.8749	-1.6958	43.5707	0.9112	92.4991	-49.8410	43.5693	**-0.0014**
B6	1.1250	-0.009	-1.116	-0.0499	-9.7646	8.6988	-1.1157	**0.0003**
B8	0.0213	0.001	Float	0.0001	0.0571	-0.0777	-0.0205	**0.0018**
B10	-0.0025	-0.0003		0.0005	0.3223	-0.7916	-0.4688	**-0.4716**
B12	0.0007	0		0.0000	-0.1130	0.4595	0.3465	**0.3465**
								78ppm erro
Coil	Main	AS		Outer	Middle	Inner 8th nudge		
a	103.6000	163		74.0000	53.2800	41.4400		
a1	87.32	151		72.52	50.32	39.96		
a2	119.9	175		75.48	56.24	42.92		
b	66.65	37.02		12.33	8.88	6.904		
b1	45.9	25.2		10.85	5.92	5.424		
b2	87.4	48.84		13.81	11.84	8.384		
Tcoil-1	4840	2560		14.029	276.174	42.282		
Tcm-2	4.565	4.565		1.601	7.88	4.826		
NI 1/2 mag	3000000	1400000		8249.27	162390.2	24861.95		
I	588	-588		588	588	-588		
Coil sec	33x33	24x24		3x3	6x6	3x3	Totals	
B0 1/2mag	10861	-5439		67	1836	-361	**6964**	
B2 r=15	78.0273	49.5371		-3.4856	-183.5972	59.8072	**0.2889**	
B4	-5.89	-0.2406		0.1283	13.026	-7.0192	**0.0045**	
B6	0.07224	-0.0006		-0.0032	-0.6278	0.5593	**0.00001**	
B8	0.0006	0.00003		0	0.00168	-0.0023	**0.00004**	
B10	-0.00003	0		0.00001	0.0043	-0.0106	**-0.0063**	
B12	0	0		0	-0.0007	0.0028	**0.0021**	
B14	0	0		0	0	0	**0**	
				error 30cm dsv B2reshimmed =0			1ppm	
				error 50cm dsv B2reshimmed =0			300p m	

Fig.11.

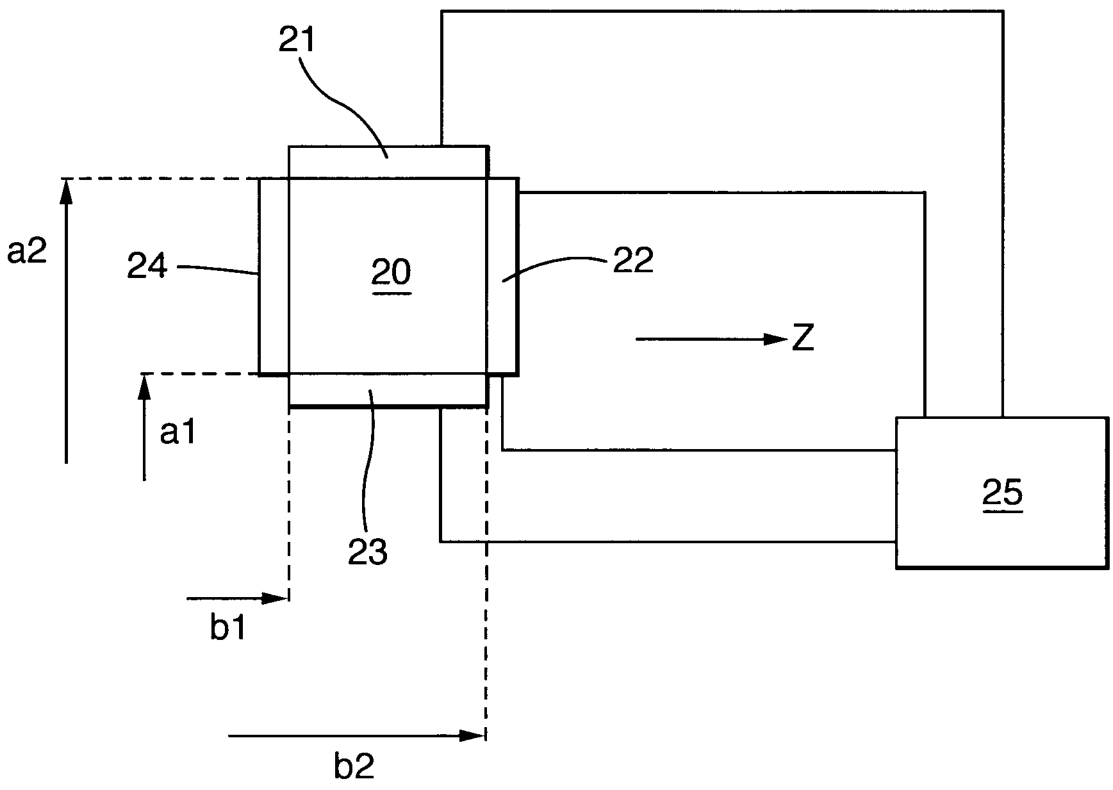


Fig.12.

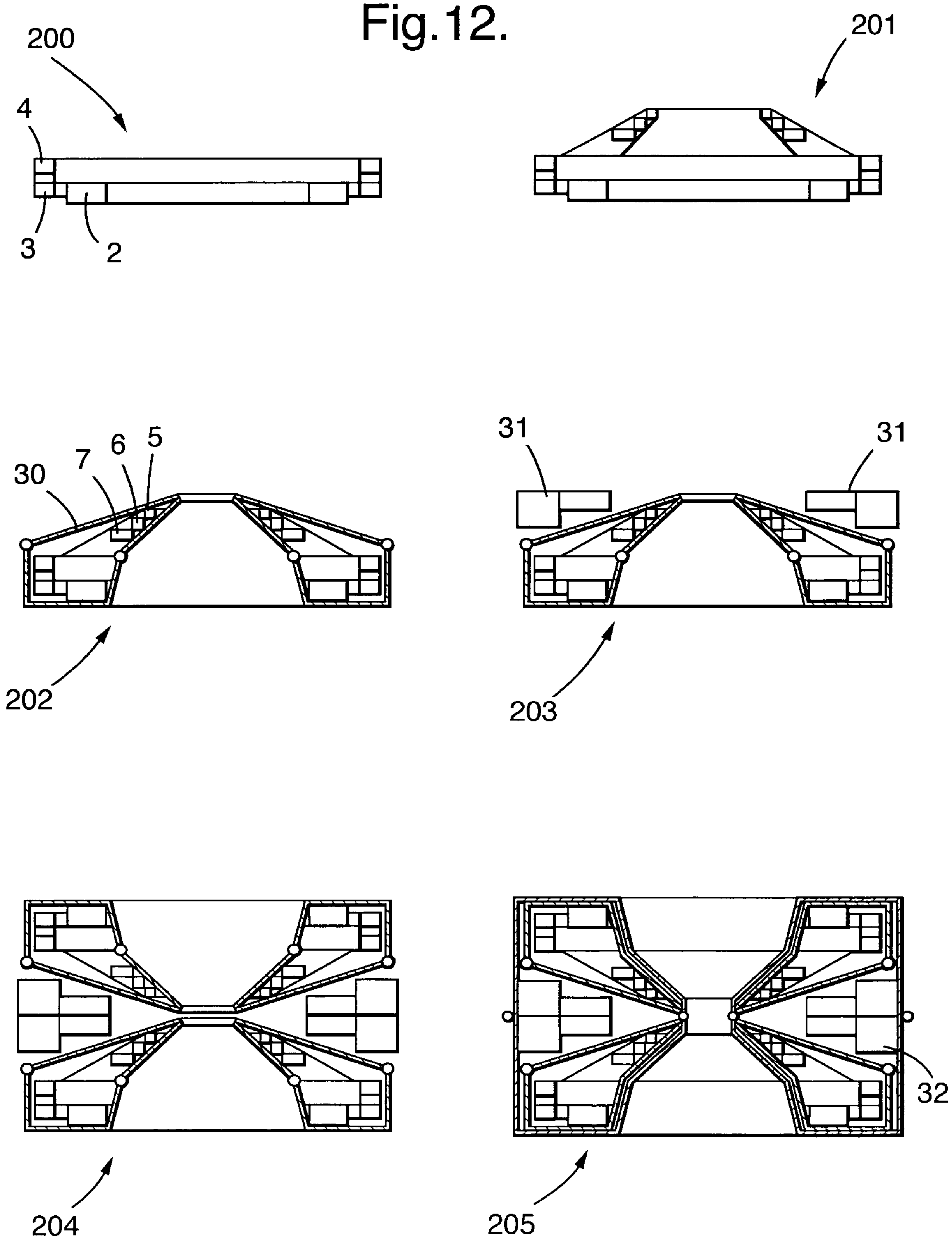


Fig.13.

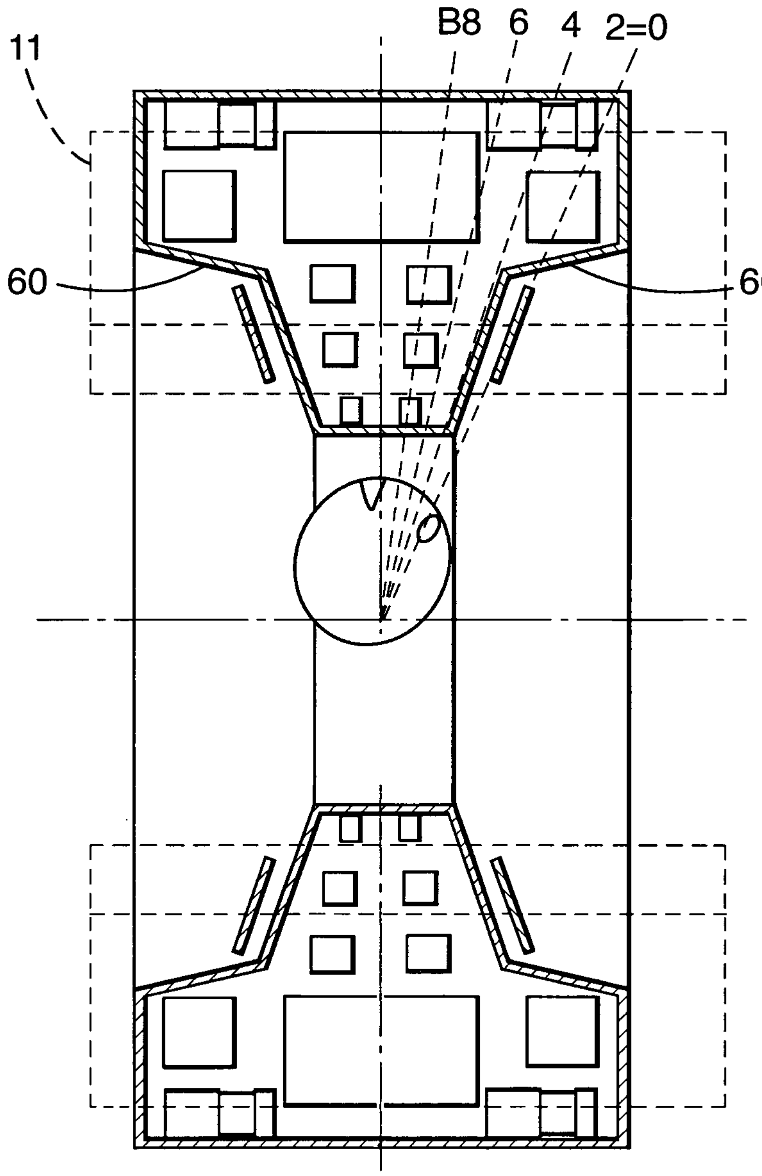


Fig.14.

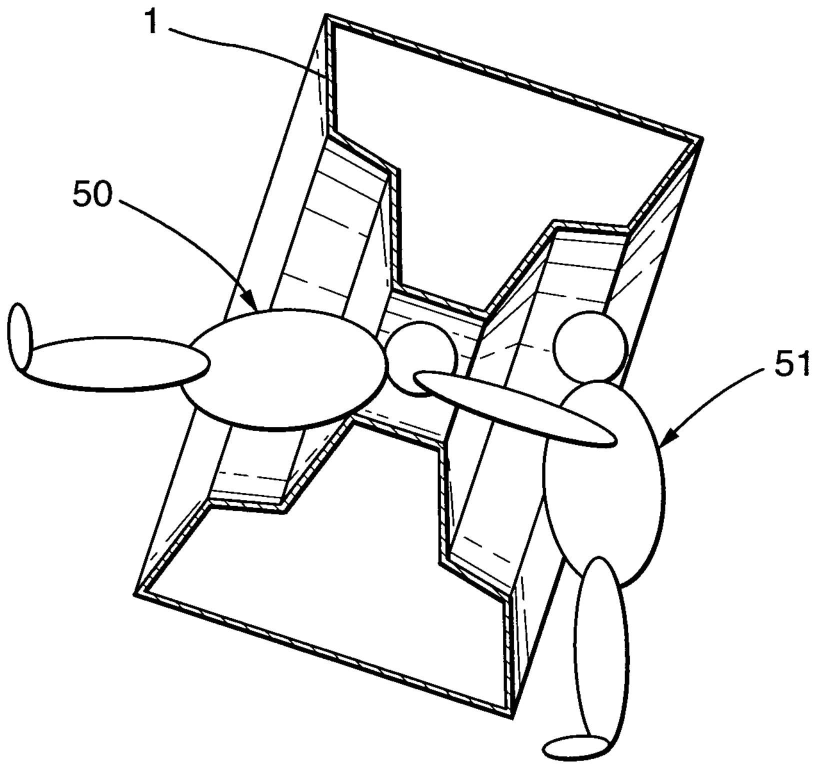


Fig.15.

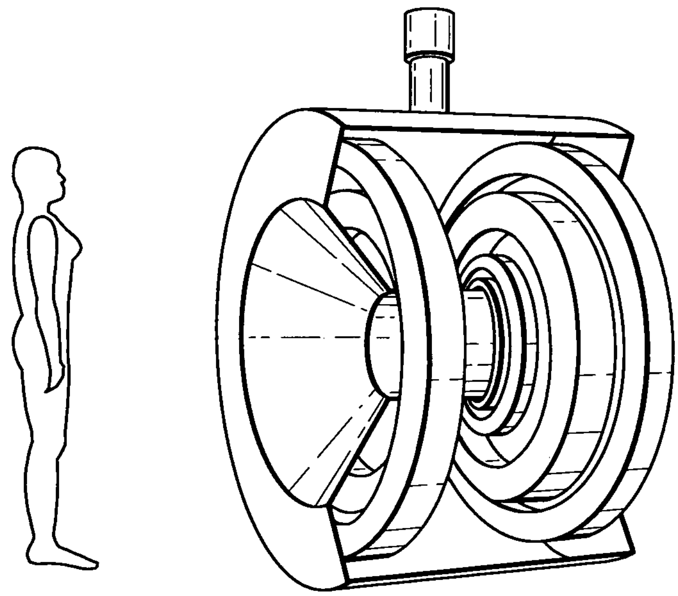


Fig.16.

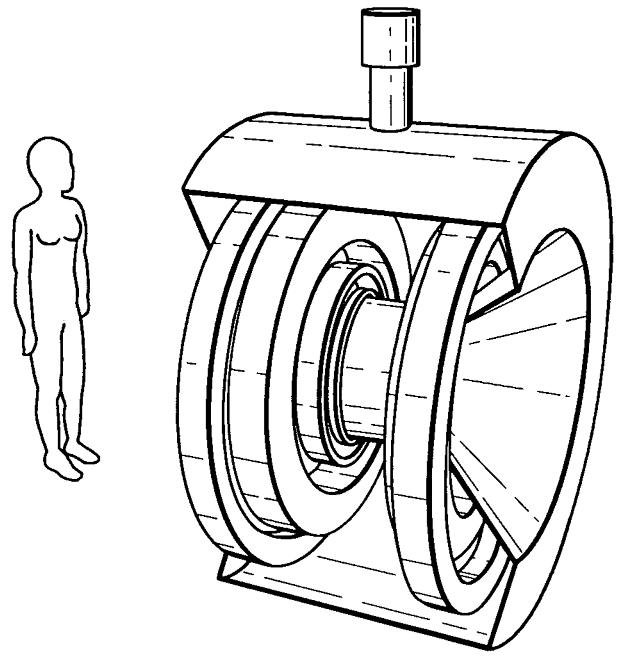


Fig.17.

		0	2	4	6	8
'exact'		1.79834E-04	-1.15250E-08	-4.97163E-03	1.63337E-01	-2.78011E+00
$r_0 = 0.1$	n_{max} = 8	1.7983E-04	-1.1525E-08	-4.9716E-03	1.6334E-01	-2.7786E+00
$r_0 = 0.1$	n_{max} = 12	1.7983E-04	-1.1525E-08	-4.9716E-03	1.6334E-01	-2.9758E+00
$r_0 = 0.1$	n_{max} = 16	1.7983E-04	-1.1525E-08	-4.9716E-03	1.6327E-01	2.0965E+00
$r_0 = 0.3$	n_{max} = 8	1.7983E-04	-1.1524E-08	-4.9716E-03	1.6338E-01	-2.9541E+00
$r_0 = 0.3$	n_{max} = 12	1.7983E-04	-1.1525E-08	-4.9716E-03	1.6334E-01	-2.7794E+0
$r_0 = 0.3$	n_{max} = 16	1.7983E-04	-1.1525E-08	-4.9716E-03	1.6334E-01	-2.7699E+00
$r_0 = 0.9$	n_{max} = 8	1.7983E-04	-9.9250E-09	-4.9738E-03	1.6520E-01	-3.8418E+00
$r_0 = 0.9$	n_{max} = 12	1.7983E-04	-1.1522E-08	-4.9716E-03	1.6336E-01	-2.8091E+00
$r_0 = 0.9$	n_{max} = 16	1.7983E-04	-1.1525E-08	-4.9716E-03	1.6334E-01	-2.7800E+00

Fig.18

	0		2		4		8	
coil	exact	Taylor	exact	Taylor	exact	Taylor	exact	Taylor
1	-2.71908E-03	-2.7207E-03	-4.81445E+02	-4.8142E+02	1.75852E+05	1.7591E+05	-3.34364E+07	-3.5961E+07
2	3.47847E-02	3.4783E-02	9.34566E+02	9.3459E+02	-2.19335E+05	-2.1937E+05	2.76224E+07	2.8893E+07
3	-1.23235E-01	-1.2326E-01	-5.01304E+02	-5.0129E+02	4.65925E+04	4.6592E+04	-2.55786E+06	-2.5945E+06
4	5.13778E-03	5.1412E-03	5.15380E+01	5.1536E+01	-2.49252E+03	-2.4924E+03	6.34046E+04	6.4069E+04

Fig.19.

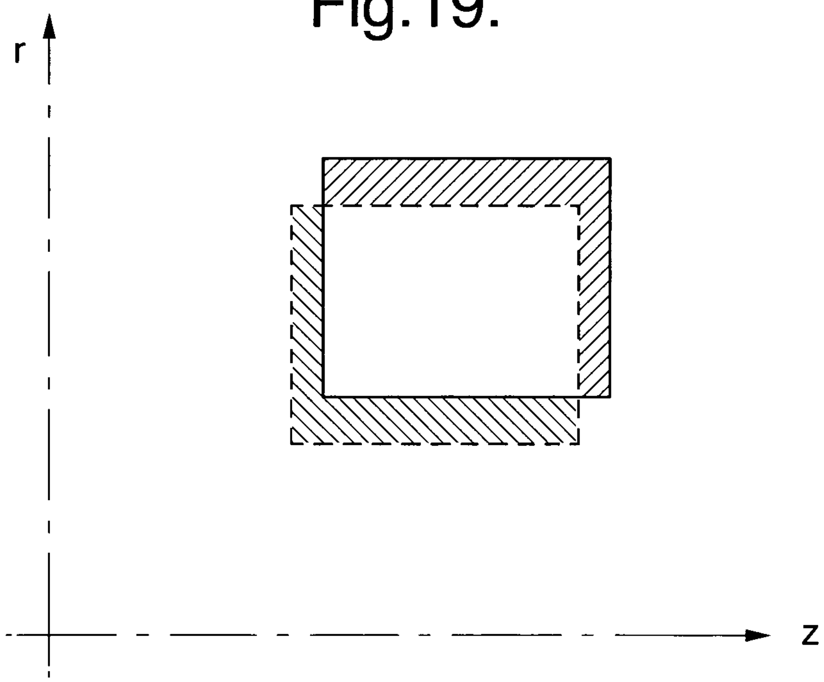
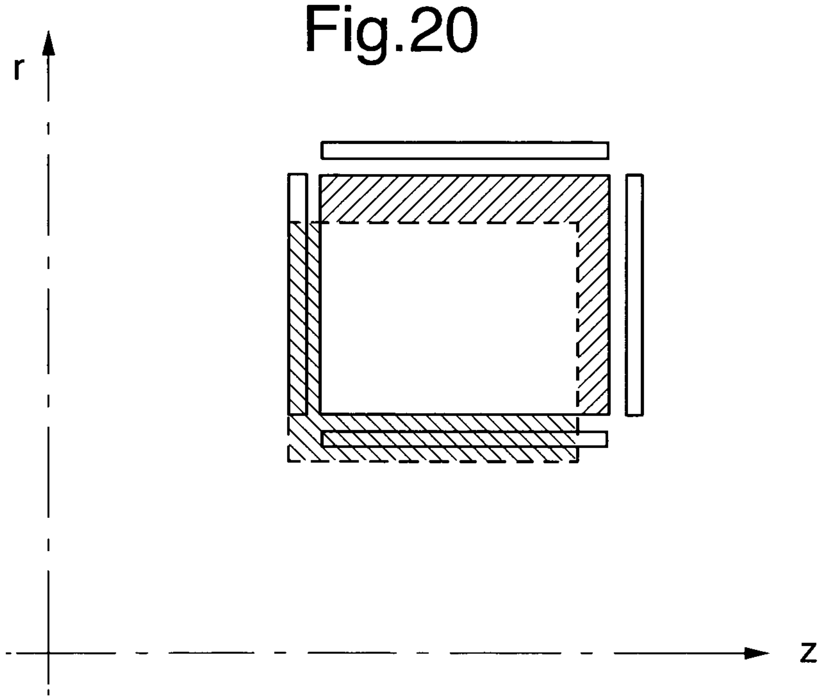


Fig.20



INTERNATIONAL SEARCH REPORT

International application No
PCT/GB2008/002182

A. CLASSIFICATION OF SUBJECT MATTER
INV. G01R33/3815
ADD. G01R33/3875

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
G01R

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

EPO-Internal, WPI Data, COMPENDEX, INSPEC, BIOSIS, MEDLINE, EMBASE, IBM-TDB

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	WO 2004/095046 A (HITACHI LTD [JP]; KIKUTA TOMOMI [JP]; WAKUDA TSUYOSHI [JP]; MAKI KOJI) 4 November 2004 (2004-11-04) the whole document	1-20,27, 28,34-38
Y		32,33

X	EP 0 496 368 A1 (TOKYO SHIBAURA ELECTRIC CO [JP]) 29 July 1992 (1992-07-29) column 6, line 31 - column 9, line 52 figures 3-4C,9	1-20,27, 28,34-38
Y		32,33

	-/--	

[X] Further documents are listed in the continuation of Box C.

[X] See patent family annex.

* Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier document but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
30 September 2008	13/10/2008

Name and mailing address of the ISA/	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31-70) 340-2040, Tx. 31 651 epo nl, Fax: (+31-70) 340-3016	Streif, Jörg Ulrich

3

Form PCT/ISA/210 (second sheet) (April 2005)

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	EP 1 036 541 A1 (HITACHI MEDICAL CORP [JP]; HITACHI LTD [JP]) 20 September 2000 (2000-09-20) paragraph [0017] - paragraph [0037] paragraph [0041] - paragraph [0051] figures 1,2,13-23	1-20,27, 28,34-38
Y		32,33
Y	EP 0 943 929 A2 (PICKER INT INC [US] KONINKL PHILIPS ELECTRONICS NV [NL]) 22 September 1999 (1999-09-22) figure 2	32,33
A	EP 0 244 843 A2 (TOKYO SHIBAURA ELECTRIC CO [JP]) 11 November 1987 (1987-11-11) the whole document	1-38
A	EP 0 460 762 A1 (KONINKL PHILIPS ELECTRONICS NV [NL]) 11 December 1991 (1991-12-11) the whole document	1-38
A	DJERDIR A ET AL: "3D modelling and design of current conical magnets" SIMULATION '98. INTERNATIONAL CONFERENCE ON (CONF. PUBL. NO. 457) YORK, UK 30 SEPT.-2 OCT. 1998, LONDON, UK,IEE, UK, 30 September 1998 (1998-09-30), pages 420-425, XP006502706 ISBN: 0-85296-709-8 the whole document	1-38
A	WO 94/06034 A (GEN ELECTRIC [US]) 17 March 1994 (1994-03-17) the whole document	1-38
A	JP 58 027306 A (TOKYO SHIBAURA ELECTRIC CO) 18 February 1983 (1983-02-18) the whole document	1-38
A	WO 00/33100 A (KONINKL PHILIPS ELECTRONICS NV [NL]) 8 June 2000 (2000-06-08) the whole document	1-38
A	JP 63 272335 A (TOKYO SHIBAURA ELECTRIC CO) 9 November 1988 (1988-11-09) the whole document	1-38

-/--

INTERNATIONAL SEARCH REPORT

International application No
PCT/GB2008/002182

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	ROMEO F ET AL: "MAGNET FIELD PROFILING: ANALYSIS AND CORRECTING COIL DESIGN" MAGNETIC RESONANCE IN MEDICINE, ACADEMIC PRESS, DULUTH, MN, US, vol. 1, no. 1, 1984, pages 44-65, XP009021071 ISSN: 0740-3194 the whole document -----	1-38
A	MCKEEHAN LW: "Combinations of circular currents for producing uniform magnetic fields" REVIEW OF SCIENTIFIC INSTRUMENTS, vol. 7, March 1936 (1936-03), pages 150-153, XP002449124 the whole document -----	1-38

INTERNATIONAL SEARCH REPORT

Information on patent family members

International application No
PCT/GB2008/002182

Patent document cited in search report		Publication date	Patent family member(s)		Publication date
WO 2004095046	A	04-11-2004	JP	4034224 B2	16-01-2008
			JP	2004325252 A	18-11-2004
EP 0496368	A1	29-07-1992	DE	69215114 D1	19-12-1996
			DE	69215114 T2	12-06-1997
			JP	4240440 A	27-08-1992
			US	5343182 A	30-08-1994
EP 1036541	A1	20-09-2000	WO	9927851 A1	10-06-1999
			US	6853281 B1	08-02-2005
EP 0943929	A2	22-09-1999	DE	69931833 T2	11-01-2007
			US	6029081 A	22-02-2000
EP 0244843	A2	11-11-1987	DE	3772616 D1	10-10-1991
			JP	1808622 C	10-12-1993
			JP	5016648 B	05-03-1993
			JP	62264606 A	17-11-1987
			US	4768009 A	30-08-1988
EP 0460762	A1	11-12-1991	DE	69130590 D1	21-01-1999
			DE	69130590 T2	24-06-1999
			JP	3238425 B2	17-12-2001
			JP	4269942 A	25-09-1992
			NL	9001300 A	02-01-1992
WO 9406034	A	17-03-1994	GB	2275538 A	31-08-1994
			JP	3556948 B2	25-08-2004
			JP	7501737 T	23-02-1995
			US	5307039 A	26-04-1994
JP 58027306	A	18-02-1983	NONE		
WO 0033100	A	08-06-2000	JP	2002531196 T	24-09-2002
JP 63272335	A	09-11-1988	JP	1783497 C	31-08-1993
			JP	4072540 B	18-11-1992