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A Permanent MRI Magnet for Magic Angle Imaging Having its Field Parallel to the Poles

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Abstract—A novel design of open permanent magnet is presented, in which the magnetic field is oriented parallel to the planes of its poles. The paper describes the methods whereby such a magnet can be designed with a field homogeneity suitable for Magnetic Resonance Imaging (MRI). Its primary purpose is to take advantage of the Magic Angle effect in MRI of human extremities, particularly the knee joint, by being capable of rotating the direction of the main magnetic field $B_0$ about two orthogonal axes around a stationary subject and achieve all possible angulations. The magnet comprises a parallel pair of identical profiled arrays of permanent magnets backed by a flat steel yoke such that access in lateral directions is practical. The paper describes the detailed optimization procedure from a target 150 mm DSV to the achievement of a measured uniform field over a 130 mm DSV. Actual performance data of the manufactured magnet, including shimming and a sample image, is presented. The overall magnet system mounting mechanism is presented, including two orthogonal axes of rotation of the magnet about its isocentre.

Keywords—Permanent magnet, Magnetic Resonance Imaging (MRI), Magic Angle imaging, Magnet design
1. **Introduction**

Magnetic Resonance Imaging (MRI) is the gold standard in many areas of medical diagnostics owing to its unrivalled ability to differentiate soft tissue structures, while 2D sections and 3D images of the anatomy can be obtained at arbitrary oblique angles. However, conventional MRI fails to detect the signals from tissues in which dipolar coupling dominates the relaxation mechanism and results in very short transverse relaxation time $T_2$, making signal acquisition very difficult. This is typical for tissues containing significant amounts of highly organized collagen, such as cortical bone, tendons ligaments and cartilage, in which their relaxation times can be changed by very large factors (typically $>10$) by changing the angle $\theta$ between the tissue and the main magnetic field. The orientation where the angular factor for dipolar coupling $(3 \cos^2 \theta - 1)$ vanishes (when $\theta = 54.7^\circ$ or, by symmetry, $125.3^\circ$) is what is termed the Magic Angle (MA) [1]. Thus MR Imaging at this ideal MA setting maximizes $T_2$, which makes MA imaging highly suitable for micro-structure investigations offering much better signal-to-noise ratio and image resolution [2]. Recent work has demonstrated how the MA effect can be used to analyse ligaments [3], meniscus [4] and articular cartilage [5] in terms of collagen fibre orientation, demonstrating promising solutions for future clinical diagnostics. Such investigation would be invaluable, for the preoperative assessment of patients being considered for knee replacement surgery, where in the US alone there are about 700,000 such procedures carried out annually. Another important area of application lies in the MA imaging of the peripheral nervous system in the body [6].

Unfortunately, in the range of magnet designs currently available for MRI, the required relative movement between the magnet field direction and the body is highly restricted or just impossible. For this reason in-vivo MA studies of ligaments and tendons reported to date have been restricted to the Achilles tendon [2] and the elbow [7].

The types of magnets suitable for use in MRI have been reviewed [8-10]. Essentially they fall into three main categories shown in concept form in Fig. 1.

1. **Solenoid magnet** (Fig. 1a), comprising a set of axially symmetric coils (usually superconducting) with the field along its main axis.
2. **Open magnet** (Fig. 1b), comprising permanent magnets of opposite polarity facing one another across a gap, shaped steel poles and a yoke forming a flux-return path. A very common variant of this magnet is driven by coils just behind both poles instead of the permanent magnets. The field orientation is perpendicular to the poles.
3. **Halbach cylinder** (Fig. 1c), comprises a ring of permanent magnets whose orientation of magnetization at azimuthal angle $\phi$ changes as $2\phi$. The orientation of field is across a diameter of the ring. Although not widely used in MRI, examples of this design are described in [11, 12].

However, none of these types of magnet are suitable for the purpose of changing the general orientation of $B_0$ with respect to the patient. The space within the solenoid magnet (Fig. 1a) is extremely restricted, with little scope for moving patient’s limbs. In the conventional ‘open’ magnet (Fig. 1b) the scope for reorienting the main field relative to the subject is almost equally limited. The Halbach array offers the opportunity to reorient the field about the cylinder axis, but the rotation about a second axis is still severely restricted. An interesting magnet type based on combining Halbach and Aubert arrays was proposed in [13], producing a tilted field relative to the cylindrical axis, but the scaling up of this design for MRI applications appears to be difficult. Thus the primary motivation for the work in this paper was to provide a new MRI magnet configuration suitable for exploiting the Magic Angle effect, such that any required orientation of the main field relative to the subject may be achieved by rotating the magnet about two orthogonal axes.

We propose a novel MRI magnet configuration depicted in Fig.1d, as a solution to achieve unrestricted field orientation. The magnet essentially consists of two symmetrically opposed pole assemblies. Each pole assembly consists of a north and a south pole, and incorporates a flux return path. This results in a net magnetic field being parallel to the pole faces. The overall assembly may then be rotated about two orthogonal axes to provide a wide range of field orientations with respect to a stationary subject.

However, there are considerable technical challenges in providing a practical MRI magnet based on this concept. These include achieving high field uniformity in a sufficiently large imaging volume and keeping the overall size and weight manageably small. Our prime application is imaging of the extremities, particularly the study of knee joints. To address these challenges, we have developed bespoke magnet design and optimization software, which was shown to be well suited for the task of optimizing a large number of design parameters while allowing effective user control of the configuration at various stages of the optimization process. The ability to deal with strong non-linearities, 3D models and varying levels of magnet array...
resolution, were shown to be the key for successfully achieving the novel magnet design.

The next section describes the concept of the new magnet in more detail. Section III presents the details of the magnet design and optimization method, while the results are summarized in Section IV, together with an outline of a moveable MRI magnet system that is currently under construction. Discussion and conclusions are presented in Section V.

2. The Magnet concept

The new MRI magnet was primarily intended to meet the requirements of MA imaging studies of the extremities, particularly the knee, and also feet, elbows and hands. Being based on permanent magnets, the target field strength was set at 0.15 T, which is comparable to existing permanent magnet MRI systems for musculoskeletal imaging. The diameter spherical volume (DSV) was targeted at 150 mm centred on the origin, or the isocentre of the magnet, with a pole gap of 22 cm.

The basic concept of the new magnet design (Fig. 1d) involves two identical rectangular pole structures, comprising a backing yoke of low-carbon steel on which are mounted magnetized blocks of Neodymium-Boron-Iron (Nd-B-Fe) permanent magnets. The north and the south poles on the two halves of the assembly are arranged to oppose each other symmetrically, thus generating the net magnetic field $B_0$ in the central region between the poles parallel to the x-axis.

In practice however, such a simple arrangement will result in the desired field being a saddle point at the origin, comprising two maxima and one minimum, or two minima and one maximum with respect to the three Cartesian axes. This is largely similar to what has been reported in the proposed designs of one-sided magnets for nuclear magnetic resonance (NMR) [14, 15], for which the resulting small uniform field volume measuring only a few mm at 10-20 mm away from the pole surface is sufficient.

What is needed for MRI imaging is a significant extended central volume of uniform field with a homogeneity of the order of 10 μT. Thus we need to transform the saddle shaped field into a significant sized plateau of uniformity in all directions. Separating the north and south poles on each half of the assembly helps a little, but to make significant progress toward uniformity we need to replace the simple configuration of Fig.1d by a more general array of magnetized blocks indicated in Fig. 2, which modulates both the height and polarity of the blocks in the central region in both x and y directions. Note that for design purposes we have adopted a convention where the z-axis is perpendicular to the open magnet poles, but the field direction is now along x. Qualitatively, in performing this modulation we sacrifice some of the field strength for the sake of a much larger region of homogeneity.

3. Magnetic Design Method

The majority of design methods for MRI and NMR magnets deal with predominantly, or fully, axisymmetric configurations [16-18], using cylindrical polar 2D design analysis. However, for the proposed magnet the problem is inherently 3D, demanding development of new computational tools presented below.

3.1 Magnet Modelling

In order to keep the number of optimisation variables reasonably small, we exploit the plane symmetries by dividing the Cartesian space into 8 octants with principal planes $x = 0$, $y = 0$, $z = 0$. We label the octants according to Table 1. By symmetry the magnetic field of the whole assembly can be determined wholly by the configuration of the magnet blocks located in the first octant and the imposition of mirror symmetry of the magnet blocks about the three principal planes, provided that the orientation of the magnetizations of the blocks are transformed in each octant according to Table 1. The symmetry boundary conditions imposed on the principal planes are that the magnetic field must be normal to the plane $x = 0$, and tangential to the planes $y = 0$ and $z = 0$.

For the magnet array, we divide the x-y plane area of the pole in the first octant into squares from which permanent magnets are considered to be independently extruded inwardly in the z direction into blocks with variable heights $t$. Since we know that the majority of the desired field is generated as in Figure 1 from magnetic dipoles oriented in the z direction we restrict ourselves in this magnet and for practicality to magnetizations in this direction. Each square element of the grid is conceptually allowed to vary in thickness between zero and a maximum $t_{\text{max}}$. By taking the sign of magnetization into account we can allow the magnets to vary smoothly in the range $-t_{\text{max}} < t < +t_{\text{max}}$.
, where the change in sign corresponds to a change in polarity. For maximum efficiency the side of blocks remote from the air gap is kept in contact with the steel yoke, so it is the inner z surface which varies.

If we were dealing with independent magnetic dipoles which did not interact with one another we might compute an analytic matrix of sensitivities of these variable dipoles at selected points in our volume of interest. We could then apply a simple least squares optimization method to seek the combination of dipoles producing the most uniform field. This would be the case if we did not have a steel yoke backing on the array of magnets. But as it is, each magnetized block will interact very strongly with the yoke, magnetizing it in a complex and hard to predict way. The yoke furthermore has nonlinear magnetization (BH) curve.

For these reasons, and in view of the required high precision of field computation, the system was modeled from the outset using finite element analysis (FEA). Specially written code was developed to manage the complexity of the design in a systematic fashion and to control the optimisation process (Fig. 3). All information about the magnet comprising geometry, system variables, field test points and material properties are held in a single design file. At any given stage we can read this file into a control program written in Python. We can then proceed to generate from this magnet data script files for the automatic generation of the geometry, meshing, solution and post-processing of the FEA program. For most of our work we have used Opera Simulation Software (Cobham Technical Services, UK). The models have also been independently verified using Comsol Multiphysics (Comsol Inc., USA).

### 3.2 Evaluation of Sensitivities

In the optimization software in Fig. 3, each calibration model can be requested, which perturbs the thickness of each magnetic block in turn. In Opera-3d, for efficiency, this was done by defining a specially prepared thin air layer, typically 1 mm thick, bounding the permanent magnet blocks in the inward axial direction. The thickness perturbation is then efficiently made by relabeling this region to be the magnet material, re-solving the model from that point, and then reversing the labeling back to air. This process is repeated for each magnet block \( i = 1 \ldots M \). Thus the same base FEA mesh was used for a model and all of its calibrations, resulting in a much faster convergence to the set of solutions than starting from scratch at each calibration would have done. In addition, by maintaining a common mesh for a set of base and calibrations model variants, systematic errors caused by changes in mesh for each calibration were eliminated.

The sensitivity of the magnetic field at test points \( \{ f \} \) to variation in parameters \( \{ x_i \} \) is expressed by the sensitivity matrix \( [S_{if}] \). Its elements are calculated by subtracting the base field \( B_f \) at each point from the calibration field value \( C_{if} \) at each point \( f \) and dividing by the amplitude of perturbation. In this way we automatically take into account any redistribution of flux that occurs in the yoke. Also by expressing \( S_{if} \) in the units of \( \mu T/mm \) we could simultaneously synthesise the main field from zero and shim it toward a configuration giving optimal homogeneity over our target volume.

### 3.3 Optimization Procedure

The objective of magnet optimization is to minimize the deviation of the target uniform field from its target value by optimizing the magnetic dipole moments of individual magnetizing blocks, by modifying block height and polarities. At each iteration we define the following:

\[ f = \{ 1 \ldots N \} \] the test points that characterize the target field, selected to lie on the surface of a target spherical volume of radius \( r_0 \) (in this case \( r_0 = 75 \text{ mm} \)) and at the corners of all cubic cells lying within that volume at a typical spacing of 10 mm.

\[ T_f \] the target field values at points \( f \), (all set to 0.15 T)

\( B_f \) are the field values at points \( f \), evaluated from the latest iteration of the FE model

\[ x_i, i = \{ 1 \ldots M \} \] are the heights of the magnetized blocks to be optimized. These are more conveniently expressed in normalized form, such that \(-1 < \frac{x_i}{t_{\text{max}}} < +1\).

In the next iteration, let \( \delta B_f \) be the change in the field at a point \( f \) due to all changes to the magnets \( x_i \). These are determined by the sensitivities \( S_{if} \) of the field to parameter variations, such that:

\[ \delta B_f = \sum_i \delta x_i S_{if} \] (1)

The deviation of the new field from the target at each point \( f \) after the next iteration is:

\[ h_f = B_f + \delta B_f - T_f \] (2)

We seek to minimize a weighted root mean square value of \( h_f \) where \( w_f \) is an optional weighting factor for point \( f \) giving emphasis to selected regions of the target volume:

\[ F = \sqrt{\frac{1}{N} \sum_f w_f^2 h_f^2} \] (3)

The default unity value of \( w_f \) was suitable for the majority of optimisations, while its use to assign non-uniform weighting was found to be particularly useful during shimming of the manufactured magnet. We performed a constrained minimization of \( F \) using a modified Newton method [19] from the NAG C library.

<table>
<thead>
<tr>
<th>Octant</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coordinate sign</td>
<td>( X )</td>
<td>+</td>
<td>–</td>
<td>+</td>
<td>+</td>
<td>–</td>
<td>–</td>
<td>+</td>
</tr>
<tr>
<td></td>
<td>( Y )</td>
<td>+</td>
<td>+</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>+</td>
<td>+</td>
</tr>
<tr>
<td></td>
<td>( Z )</td>
<td>+</td>
<td>+</td>
<td>+</td>
<td>+</td>
<td>–</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Magnetization Component Direction</td>
<td>( M_x )</td>
<td>+1</td>
<td>+1</td>
<td>+1</td>
<td>+1</td>
<td>+1</td>
<td>+1</td>
<td>+1</td>
</tr>
<tr>
<td></td>
<td>( M_y )</td>
<td>+1</td>
<td>–1</td>
<td>–1</td>
<td>–1</td>
<td>+1</td>
<td>–1</td>
<td>–1</td>
</tr>
<tr>
<td></td>
<td>( M_z )</td>
<td>+1</td>
<td>–1</td>
<td>+1</td>
<td>–1</td>
<td>+1</td>
<td>+1</td>
<td>–1</td>
</tr>
</tbody>
</table>
nag_opt_bounds_2nd_deriv (Numerical Algorithms Group, UK). This method requires the first and second derivatives of the objective function. These can be derived by an analytic differentiation of (3) and its defining constituting expressions (1) and (2). From these it can be shown that $G_i$, the first derivatives of $F$ with respect to $x_i$, are given by

$$G_i = \frac{\partial F}{\partial x_i} = \frac{1}{NF} \sum_f \left( h_f S_{if} w_f^2 \right)$$

(4)

Also, the second derivatives of $F$, $H_{ij}$ are given by

$$H_{ij} = \frac{\partial^2 F}{\partial x_i \partial x_j} = \frac{1}{NF} \sum_f \left( h_f S_{if} w_f^2 \right) - \frac{1}{F^2} G_i G_j$$

(5)

The square matrix $H$ is known as the Hessian matrix and equation (5) is valid both for diagonal ($i = j$) and off diagonal ($i \neq j$) members. From previous experience optimisation, with access to these accurate derivatives, works very well where a large number of variables need to be optimised. For the optimisation routine, functions need to be supplied in software which implement these formulae for computation of the objective function and its derivatives dynamically, on demand whenever the optimiser requires them.

Therefore at each iteration the optimization algorithm in Fig. 3 could be summarized as follows:

1. Prepare the base FEA model from the parameters in current design file.
2. Prepare $M$ calibration FEA models for each magnet block $1 \leq i \leq M$ by perturbing the thickness of each block.
3. Run the FEA solver for the base model and re-solve for each of the calibration models.
4. Run postprocessor to extract evaluated field $B_f$ at the prescribed points from the FEA solution.
5. Run postprocessor to extract N calibration fields $C_i$ from each calibration model $i$.
6. Subtract base field from calibration fields to generate $S_{if} = B_f - C_{if}$ sensitivity matrix for all $i, f$.
7. Set upper and lower limits on allowed changes in block thicknesses $x_i$ for the next iteration.
8. Run constrained solver to minimize $F$ for $x_i$ with known first and second derivatives.
9. Update design file based on the latest block changes $x_i$.

When starting an optimization study the variables could start out at the equivalent of zero thickness of the dipole moments in all positions and thus zero field so that no pre-conceived bias would influence how the poles would grow and evolve. Instead the magnet patterns emerged organically.

For each iteration a constrained optimization was performed where the objective function is the sum of the squares of the deviation in field from the target field of 0.15 Tesla, as in equation (3). Constraints were applied at each iteration so as to ensure that the thickness of each of the magnet blocks must not grow thicker than the allowed $t_{max} = 50$ mm.

The field optimization points, $f$, were chosen to sample the computed fields at points at the surface of the 150 mm sphere, corresponding to the Gaussian angles corresponding to a 13 plane plot and sampled at 36 points azimuthally at 10° intervals, and on a regular 10 mm grid of points lying inside the sphere.

3.4 Evolution of the Magnet Design

Initial trial optimizations were conducted on models based on a 50 mm square grid of magnets with variable height. These were used to establish the basic dimensional parameters of the magnet that can realistically achieve the required field level and homogeneity over the target DSV of 150 mm, which would accommodate the intended imaging of the human knee. As expected, the field homogeneity rapidly improved as the pole gap in Z decreased, so it was necessary to fix the pole gap at a suitable compromise value. Similarly, the X and Y pole dimensions needed to be established to allow maximum patient access while achieving the field targets. Table 2 summarizes the established overall magnet parameters, which were then fixed in all subsequent optimizations.

Subsequent magnet optimizations involved increasing resolution of the base magnet grid. Initially, the coarsest pattern, shown in Fig. 4a was used, involving 50 mm square magnet blocks as the largest available magnet cross section. A pattern emerged, whereby all dipoles in the outer regions of the pole piece evolued to be at their allowed maximum value of the same sign and, for efficiency, they could be excluded from further optimization. These magnets provide the majority of the flux for the main field, while those in the central region of the pole show a heavily modulated distribution of dipole strength including reversed signs (as indicated in Fig. 2).
It was found that in practice it was also beneficial to impose a further constraint on the magnitude of change of each variable for a single iteration. Failure to do so would result in the field overshooting its target, due to the nonlinearity of the steel and the complex interaction between magnets and yoke. Therefore patience was needed in gradually reducing this constraint on maximum amplitude of change on each block from an initial 5 mm to as little as 0.1 mm per iteration.

In view of this, attempts were made to save the computing time and reuse the calibrations more than once on the basis that the parameter changes were very small. However this approach was generally unsuccessful and often resulted in the field quality deteriorating, highlighting the fact that with iron present the process is highly nonlinear. Therefore the magnet model and all its calibrations had to be refreshed after each iteration.

After optimizing with the 50 mm square grid it was found that the progress of field improvement had stalled, impeded by the coarseness of this grid. Therefore the grid was refined to 25 mm squares. With increased resolution, optimization runs again established the dipoles in the outer regions of the pole pieces that evolved to their maximum value, which were then excluded from further iterations. In addition, it was found that the corners of the poles were not contributing significantly to the field and to keep the pole areas more streamlined it was decided to fillet them along with the yoke to a radius of one magnet block width or 50 mm. All this has led to the final magnet pattern shown in Fig. 4b where 32 variables were modulated in the final optimization stages. A typical model mesh comprised 1 million nodes and 6 million elements. The computation for one complete iteration including all calibrations was typically 3 hours on a Xeon 2.5 GHz, 2-processor PC with 96 Gb memory.

Additional optimization trials were made to assess the effect of refining the squares by a further factor of 2 (to 12.5 mm squares). However, little tangible improvement in the field quality seemed to accrue from this higher resolution, while the number of variables quadrupled and the time taken to re-run the FEA to obtain the calibrations became very long (>10 hours). In addition the large number of such small squares would make manufacture significantly more difficult.

It was noticed during optimization that there was a tendency for the magnitude of the modulated dipoles to be mostly less than 0.75 (sometimes much less) on our normalized scale, where \( t_{\text{max}} = 50 \text{ mm} \). Therefore, to take advantage of this effect, a constraint for all variable dipoles was applied that they were only now allowed heights in the range \( \pm 0.75 \). This created a recess in the front face of the pole of 0.25 (12.5 mm) and the optimizer was still able to find as good a solution with this constraint in place. We note that this emergent effect bears a resemblance to the Rose ring feature on conventional magnet poles where the thicker pole rim compensates for the truncation effect of the edge of the pole and maintains a larger volume of uniform field. The intention was to use the space that this recess provided to house a light, low-profile passive shim set to compensate for deviations in the field caused by manufacturing and magnetisation tolerances without impinging further on the net accessible gap of the magnet. It was also discovered that the optimization led to two squares in each quadrant to be very close to zero thickness, which were then both constrained to zero and the remainder re-optimized with no ill-effects on the field. These vacated locations in the pole pieces later proved to be useful for mechanically anchoring the shimset.

### 3.5 Permanent Magnet Grades

<table>
<thead>
<tr>
<th>Table 2</th>
<th>Magnet dimensions fixed at the initial design stage</th>
</tr>
</thead>
<tbody>
<tr>
<td>Parameter</td>
<td>Dimension (mm)</td>
</tr>
<tr>
<td>Length in X</td>
<td>600</td>
</tr>
<tr>
<td>Width in Y</td>
<td>350</td>
</tr>
<tr>
<td>Clear gap between poles Z</td>
<td>220</td>
</tr>
<tr>
<td>Thickness of yoke</td>
<td>50</td>
</tr>
<tr>
<td>Thickness of magnet blocks (maximum)</td>
<td>50</td>
</tr>
</tbody>
</table>
Neodymium-Iron-Boron is available in many grades with remanence ranging from around 1.2 to 1.425 Tesla, but with the lower energy grades being available in harder versions – that is, having a higher coercivity. In general the policy in our magnet design was to use the highest grade N50 where possible for maximum field strength but subject to the magnetic intensity in the model not getting too close to the coercivity of the material which might cause demagnetization. By monitoring in the FEA models this condition was found to be satisfied for the majority of the magnets at their maximum values of thickness. However where approached the specified value of coercivity a harder grade was chosen (N48M or N48H) to give a larger margin of safety against demagnetization. The trade-off is that the harder grades are a little less strong and more expensive through having more of the exotic rare earth elements (e.g. dysprosium) in their composition. We note that magnetic remanences in data sheets are not normally specified with a precision better than 1% which we would expect to lead to appreciable errors in fields measured in ppm.

4. Results

4.1 Magnet Optimization

The resulting design fields attained after an evolutionary period of over 200 models is shown in Fig. 5a which plots the field in Tesla along the Cartesian axes between ±100mm. The orientation of the field is along the long axis parallel to the poles designated x. Clearly a significant plateau of uniform field can be achieved, formed by maxima with respect to x and y and a minimum with respect to z.

As an indicator of the field off-axis, Fig. 5b shows the field plotted over the surfaces of the Cartesian planes in terms of the deviation in field in the range ±50 µT relative to the centre, which gives a more complete picture than the axial field plots in Fig. 5a. Notice that in the y-z and z-x planes we see alternating lobes of positive and negative variation in field separated by cusps which can trace their origins back to the alternating signs of dipole moments in the permanent magnet array. On the other hand, in the equatorial x-y plane (halfway between the poles) the field falls negatively off the central plateau in all radial directions.

The final design of the magnet poles is as indicated in the 3D view of Fig. 2, which shows the variable height s of the arrays of permanent magnets and the solid backing yokes. The repulsive magnetic force between the poles was estimated from FE models to be about 7000 N.

4.2 Magnet as manufactured

The magnet (Fig. 6) was manufactured by Arnold Magnetics (Sheffield, UK) and tested by mapping the field, using a NMR probe (Metrolab, Geneva, Switzerland) mounted on a 3 degree-of-freedom robotic manipulator. Initial tests confirmed that the centre field was within 0.6% of the target and the homogeneity was largely symmetrical about the principal planes and in line with the specified magnet material tolerances, but it did require shimming. It was found that the probe could achieve a locked NMR...
signal at 75-80 mm from isocentre along the x and y axes but no more than 65 mm along z. Therefore it was decided to plot and optimise on a sphere of radius 65mm using 15 plane Gaussian angle plots, each sampled azimuthally at 24 points (15°).

A shim set was designed in the form of 4 plastic plates per pole, designed to fit in the central recess of the magnet face and providing an array of 1024 locations for mounting 5mm diameter button magnets in thicknesses between 1 and 5mm of either polarity. The shimming process involved five iterations of field mapping and shim optimization using the software described in Sec. 3.3 but slightly modified so the objective function was the least-squares deviation from the isocentre rather than the fixed target 0.15 Tesla thus allowing the main field to float. Fig. 7 shows the raw measured axial field plots in units of μT before and after shimming thus showing both field level achieved and variation. The shimming process reduced the exacting peak-peak range of the 15 plane, 24 angle, 130 mm DSV plots from 979μT to 133μT in four iterations. Factor of 6.7 converts variation in field from μT to ppm at 0.15 Tesla. The shimmmed plots indicated that the residual linear field errors could be corrected by small gradient offsets driven by less than 0.5Amp from our gradient set.

Preliminary integration of the complete MRI system was carried out, including the transverse field magnet, gradient system and a prototype spectrometer developed by our group. Fig. 8 shows an early sample image of a test phantom (seen in Fig. 6) consisting of a cylindrical jar (67mm internal diameter), filled with CuSO₄ doped water, containing 10 plastic tubes (15mm diameter, wall thickness 1.4mm) 3 of which were inserted inside a plastic tube (32mm diameter, wall thickness 1.8mm). The image was obtained using single channel 100mm diameter Helmholtz pair receiver coil, using a Gradient Echo sequence; Field of View (FOV) was 70 x 70mm, pixel size 0.5mm, slice thickness 5mm, Flip Angle 30°, TE=8ms, TR=200ms, Number of Excitations =4.

Having a large thermal mass, it was found that temperature variations in the room would typically change the operating NMR frequency of the system by less than 20 kHz during an operating day which is well within the capability of a B₀ coil built into the gradient set designed to compensate for such drift.

### 4.3 The Moveable MRI Magnet System

The design of moveable MRI magnet for extremity imaging, based on the results of this study, is shown in Fig.9. The assembly involves a 2 degree-of-freedom (DOF) mechanism, the outer part of which comprises an aluminium ring, which is supported by a set of four rollers to provide the ‘roll’ motion. The inner part comprises the magnet and an aluminium frame, which are supported in bearings and

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**Fig. 7.** Field (in μT) of the constructed magnet measured along the three principal axes (a) before shimming and (b) after shimming.

**Fig. 8.** Sample image of a test phantom. FOV 70 x 70 mm, pixel size 0.5mm. Gradient Echo, Flip Angle 30°, TE 8 ms, TR 200 ms, slice thickness 5 mm, NEX 4. The black meniscus shaped area at the top corresponds to an air bubble in the phantom.

**Fig. 9.** Magnet system for MA imaging, comprising 2DOF mechatronic system for controllable positioning of the magnet at various orientations. Shimset (in white) can be seen embedded in the pole face.
motorized to provide the ‘yaw’ motion. As stated previously, the dimensions of the pole pieces are 600mm × 350mm × 100 mm, the outer ring diameter is 1300 mm. The total weight of the magnet poles is about 350kg, of which 155kg are the permanent magnets. The mechatronic system was carefully designed to minimise any out-of-balance loads due to gravity, which enabled us to employ modestly sized (200W) electric motors, each combined with a harmonic gearbox (Harmonic Drive, Limburg, Germany). All actuation components and bearings are located behind the pole iron and at the base, such that they minimally influence the magnetic field.

In order to achieve all the angulations needed to exploit the magic angle effect, it is in principle sufficient that the range of movements for each axis is ±55°. Nevertheless, the system was designed to allow a full 360° rotation about the roll axis, which meant that all angulations can be achieved with a necessary yaw axis range of little more than ±35°. This solution was considered preferable for ergonomic reasons.

4.4 The Gradients and Spectrometer System

An unsheilded gradient coil system was designed using a target field method the details of which will be the subject of a further paper. Effective eddy current-inhibiting slits were incorporated into the first steel structure that the gradients encountered. Gradient field strengths of 263, 201 and 487 μT/m/A were measured for the X, Y and Z axes respectively. The coils were water-jet cut from 1.2 mm thick copper. Each axis had a resistances of ~0.35Ω and inductance of ~ 130 μH and were driven by Performance Controls Inc (USA) GA-300 amplifiers.

The spectrometer is based on National Instruments (USA) PCI-bus D/A, A/D and frequency synthesiser cards hosted in a Windows 7 PC running the low-level NI-DAQmx library hosted in a Python environment and with additional external Mini-Circuits (USA) mixers and analogue electronics boards and scan controller software designed by the group.

5. Conclusion

The designed field of 0.15 Tesla at an accessible gap of 220mm was chosen as a realistically achievable target using Nd-B-Fe permanent magnets while having a proven record of providing clinically useful MRI images. However a space budget of 10mm was allowed per pole for the provision of the gradient set and RF transmit coil, making the net usable gap 200 mm. Having chosen a target DSV of 150mm suitable for imaging of the knee the dimensions of the magnet in the direction parallel to the main field x and the width y in Table 2 emerged as the minimum values at which the target field level was achievable. In practice, owing to material and engineering tolerances and the single-shot nature of setting up such a complex assembly for the first time a measured uniform field was actually achieved over 130mm DSV of which we estimate the central 120 mm DSV should be imageable at a uniformity of at least 10µT or 67 ppm - improving very rapidly with decreasing radius. It is expected that subsequent magnet builds will improve on this as the magnet formula is adjusted slightly. Extending the magnet in x and y could certainly increase the size of the homogeneous volume but at the price of more permanent magnet material and patient accessibility. Other applications may have different priorities and constraints which could be readily accommodated using the design and software tools developed in this paper.

There may be other applications of such a homogeneous magnet both inside MRI in susceptibility imaging where angulation can result in new data [20] and outside MRI such as in physics experiments where a design optimised to suit, for example, at higher field and narrower gap could similarly rotate a magnetic field about two orthogonal axes.

This magnet could be scaled up for whole body imaging if the gap were increased to around 400 mm. However it may require an unfeasibly large amount of permanent magnet material particularly if the field was to be maintained at 0.15T. An alternative approach may be to use the same fundamental principle of opposing like poles but to replace the array of permanent magnets with equivalent ‘islands’ of positive and negative dipole moment generated by superconducting coil arrays. Such a magnet should have the capability of generating fields of at least 0.5T.

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References

Graphical Abstract

[Diagram showing a mechanical system with labels for Yaw Motor, Magnet, Roll Motor, and magnetic field标注：B₀.]
A Permanent MRI Magnet for Magic Angle Imaging
Having its Field Parallel to the Poles

Highlights

- Novel open magnet design having field parallel to the poles.
- Magnet suitable for human in-vivo MRI studies of the magic angle effect.
- 3D modelling and optimization method using specially developed software for configuration control, dealing with large number of parameters.
- Magnet performance verified by field measurements made on the manufactured magnet.
- High quality MR imaging demonstrated following integration of the magnet with MRI system.