Note

A non-axial superconducting magnet design for optimized patient access and minimal SAD for use in a Linac-MR hybrid: proof of concept

Shima Yaghoobpour Tari\textsuperscript{1,2}, Keith Wachowicz\textsuperscript{1,2} and B Gino Fallone\textsuperscript{1,2,3}

\textsuperscript{1} Department of Medical Physics, Cross Cancer Institute, 11560, University Avenue, Edmonton, Alberta, T6G 1Z2, Canada
\textsuperscript{2} Department of Oncology, University of Alberta, 11560, University Avenue, Edmonton, Alberta, T6G 1Z2, Canada
\textsuperscript{3} MagnetTx Oncology Solutions, PO Bx 52112, Edmonton, Alberta, T6G 2T5, Canada

E-mail: yaghoobp@ualberta.ca

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Abstract
A prototype rotating hybrid magnetic resonance imaging system and linac has been developed to allow for simultaneous imaging and radiation delivery parallel to $B_0$. However, the design of a compact magnet capable of rotation in a small vault with sufficient patient access and a typical clinical source-to-axis distance (SAD) is challenging. This work presents a novel superconducting magnet design as a proof of concept that allows for a reduced SAD and ample patient access by moving the superconducting coils to the side of the yoke. The yoke and pole-plate structures are shaped to direct the magnetic flux appropriately. The outer surface of the pole plate is optimized subject to the minimization of a cost function, which evaluates the uniformity of the magnetic field over an ellipsoid. The magnetic field calculations required in this work are performed with the 3D finite element method software package Opera-3D. Each tentative design strategy is virtually modeled in this software package, which is externally controlled by MATLAB, with its key geometries defined as variables. The optimization variables are the thickness of the pole plate at control points distributed over the pole plate surface. A novel design concept as a superconducting non-axial magnet is introduced, which could create a large uniform $B_0$ magnetic field with fewer geometric restriction. This non-axial 0.5 T superconducting magnet has a moderately reduced SAD of 123 cm and a vertical patient opening of 68 cm. This work is presented as a
proof of principle to investigate the feasibility of a non-axial magnet with the coils located around the yoke, and the results encourage future design optimizations to maximize the benefits of this non-axial design.

Keywords: magnet design, optimization, longitudinal linac-MR

(Some figures may appear in colour only in the online journal)

1. Introduction

There has been an increasing interest in using magnetic resonance (MR) imaging in radiotherapy in the recent years. MR can produce images with exquisite soft-tissue contrast helping to improve the accuracy of the delineation of both tumor and surrounding tissues, and uses no ionizing radiation. Given these imaging abilities, a hybrid system combining a medical linear accelerator (Linac) and an MR would permit real-time image-guided radiotherapy, where a continuous stream of images throughout the radiation treatment can be acquired. As a result, any tumor motion can be identified, and the radiation beam can be adapted to accommodate the change. Having the beam precisely on the target during the treatment minimizes the harm to normal tissue near the tumor. At the present time, there are a number of prototypes of hybrid Linac and MR systems both in production and development (Fallone et al 2009, Raaymakers et al 2009, Pechenaya Green et al 2012, Fallone 2014, Keall et al 2014, Keyvanloo et al 2016).

The Linac-MR system as developed by Fallone et al (Fallone 2014, Keyvanloo et al 2016) is designed as a linac mounted onto a bi-planar rotating MR scanner to allow for simultaneous imaging and radiation delivery. One of the unique features of this rotating design is the ability to transmit radiation parallel to the $B_0$ field, which has been shown to have dosimetric advantages compared to the transverse configuration (Keyvanloo et al 2016). These advantages take the form of a reduced penumbra width (Litzenberg et al 2001), reduction of tissue-air interface effects, and increased dose in the PTV, resulting in reduced dose to the surrounding normal tissue (Kirkby et al 2010).

Critical to the success of this biplanar Linac-MR system is the design of a magnetic yoke that allows for a homogeneous magnetic field over a large field of view. Magnetic field homogeneity allows for artifact-free imaging and accurate geometrical information of the target volume during the treatment. In addition to the stringent requirements on the magnetic field uniformity for the image quality, there are other criteria that are imposed on the system due to the unusual circumstances in which the MR must operate. Specifically, the yoke should be designed to have ample patient access, and to be sufficiently compact to rotate in a typical clinical vault. In addition, it must allow room for the linac assembly to be positioned within a limited distance from the isocentre of the device.

The biplanar MR radiotherapy hybrid as being developed by Fallone (Fallone 2014), has an increased source-to-axis distance (SAD) compared to the 100 cm standard, mainly due to the position of the linac, which must be outside the yoke and in a low magnetic field zone. The latest system has an SAD of 130 cm (Keyvanloo et al 2016), where the radiation beam is parallel to the direction of magnetic field. The 3D schematic of the magnet assembly is shown in figure 1(a). This increase in SAD will cause the reduction of the dose rate at the isocentre by 41% due to the inverse square law, with potential consequences for patient treatment times. Consequently, the reduction of this SAD will be a valuable objective of any new design.

Conventionally, bi-planar superconducting magnets have the coils placed axially adjacent to the pole plates to produce the desired magnetic field (Tadic and Fallone 2011, 2012). The
Coils are enveloped by a bulky cryostat, which extends on the order of 30 cm or more in the vertical dimension. For the new design, the superconducting current loops are moved to surround the yoke supports that connect the pole plates. The intention in this new design is to reposition as many structures that are currently between the patient and the linac as possible. Moving the coils to the side allows for reduced SAD while improving the patient access, which is important to reduce patient claustrophobia and clinician access.

This work presents the development of a unique superconducting biplanar magnet design as a proof of concept, shown in figure 1(b). Using a finite-element analysis technique, the shape of the pole plate for the magnet assembly is optimized to achieve the desired field uniformity over a field-of-view suitable for MR imaging. The magnetic field calculations required in this work are performed with the 3D finite element method (FEM) software package Opera-3D (Opera-3D 2007). Each tentative design strategy is virtually modeled in this software package, which is externally controlled by MATLAB, with its key geometries defined as variables. The optimization variables are the thickness of the pole plate at control points distributed over the pole plate surface. An optimized magnet assembly that generates a homogenous 0.5T magnetic field over an ellipsoid with large axis of 60 cm and small axes of 40 cm is obtained. For the final optimized 0.5 T magnet, the inhomogeneities of the magnetic field are studied.

2. Methods

2.1. Magnet assembly

The open superconducting magnet assembly is depicted in figure 1(b). The superconducting coils have been moved from their conventional position (being centred on the magnet axis, with one adjacent to each pole plate) to a non-axial position surrounding the yoke supports. The yoke supports are composed of 1020 steel and serve both a mechanical and magnetic function. Mechanically they resist the compressive force between the two magnetic poles, and magnetically they serve as a flux pathway, guiding magnetic flux generated by the coils to the pole plates, which are sculpted to distribute the field uniformly over a region-of-interest. The pole plates are composed of grain-oriented (GO) silicon steel. GO silicon steel is an electrical/transformer steel developed under controlled condition to produce grains with preferred crystal orientation (Dorner et al 2006), which improves magnetic properties in the rolling direction. Silicon increases the electrical resistivity of the steel, which decrease the induced
eddy currents (Tumanski 2016). In addition to being a steel with relatively high resistivity, the sheets of GO-SiSteel are often insulated in some manner to resist the flow of current through a laminar stack. A wide array of industrial methods are available for improving inter-sheet resistance (ASTM A976-13 2013). The sheets themselves are on the order of 0.1 mm, with thinner sheets having higher resistivity (Haiji et al 1996). The sculpted nature of the pole plates, as described above, is achieved by considering this structure to be defined by a continuous array of vertical projections. Vertical projections are part of the pole plate assembly. Projections are attached to the pole plate to guide magnetic flux toward the target region, an ellipsoid with its center located at isocentre. In this work, a homogeneous field over the target region is achieved by optimizing the height of all these projections. This procedure will be described in more detail below.

In this proof of principle design, the minimum gap between two pole plates is set to 68 cm and the distance from the isocentre to the yoke’s top is 63 cm, which results in a SAD equal to 123 cm. This assumes a 60 cm distance between the target and the top of the yoke, which is distance from target to end of the MLC for the standard, unmodified Varian 600C linac head (Varian Linac Document; Drawing number: 1106021). The distance between pole plates is 70 cm, shown in figure 7. A few centimeters of gradient coils will be added to the final magnet, which is not modeled in the current design due to great range of design possibility. There is an opening with a diameter of 24 cm bored through the magnet to open a path for the radiation. The linac head would be placed on the top of the outer surface of the magnet into which the portal is drilled, and it is the height of this surface relative to isocentre that would ideally be minimized. Further, this yoke design is constructed to maintain the 110 cm lateral patient access that the current model allows, being important for patient comfort and treatment flexibility. The bi-planar nature of the magnet allows for a wide-range of motion in a direction perpendicular to the radiation beam. Given that, a peripheral tumor treatment positioning (PTTP) system and method, explained in detail in Wachowicz et al (2016), can be employed to make sure that the tumor is always kept in the axis of the beam. This allows for the treatment of peripheral tumors such as breast or lung tumors.

2.2. Distribution of coils

In this design, the material used for the superconducting coils is MgB$_2$, which is a high-temperature superconducting (HTS) material (Nagamatsu et al 2001). This superconductor (with $T_c \sim 39$ K (Razeti et al 2008) compared to $T_c \sim 9.5$ K (Kuperman 2000) for niobium-titanium (NbTi), which is material commonly used for superconducting magnet) allows for greater flexibility, especially for a rotating magnet, as no cryogens need to be stored at the coils, given that the coils can be conduction cooled. In this work, the current density for the MgB$_2$ coil is assumed to be $2.1 \times 10^8$ Am$^{-2}$, as was done in previous publication (Tadic and Fallone 2012). This current density is theoretically achievable at temperature of 10 K and a magnetic field of 6.5 T (Li et al 2015), even assuming a superconducting fill factor of only 20%.

For this proof-of-principle implementation, different scenarios were explored to find a suitable current distribution. Two examples of these distributions are shown in figure 2. These examples reveal the non-linear nature of the current distribution with respect to the field produced at isocentre. For both configurations, the magnetic fields generated at the isocentre are about 0.5 T. Therefore, roughly the same field with considerably less current cross section can be produced using distributed coil elements. This can be explained by the fact that the 1020 steel yoke supports saturate magnetically above roughly 2T, after which additional current
surrounding a similar location will provide little additional flux through isocentre. As shown in figure 2, both distributions generate the same level of magnetization, which means that they are fully saturated.

Any implementation will be very dependent on the exact manufacturing process of the wire, and the exact conductor distribution will need to be considered on a case-by-case basis, given that each conductor yields different specifications in terms of critical field, temperature and current density. Further optimizations of current distribution may be considered to limit superconducting material and reduce intra-conductor $B$ fields. However, given that our primary interest was the ability to sculpt a homogeneous field in this non-axial design by manipulation of the pole plates, the three-coil distribution as shown in figure 2(b) was used for the remainder of the work.

2.3. Magnetic field simulation

The magnetic field calculations required in this work is performed with the 3D FEM software package Opera-3D (Opera-3D 2007), using a magnetostatic model. The electromagnetic analysis is based on the following Maxwell’s equations:

$$\nabla \times \mathbf{H} = \mathbf{J},$$

$$\nabla \cdot \mathbf{B} = 0,$$

where $\mathbf{H}$ is the magnetic field, $\mathbf{B}$ is the magnetic flux density, and $\mathbf{J}$ is the current density. The total magnetic field $\mathbf{H}$ can be defined as (Simkin and Trowbridge 1979, 1980):

$$\mathbf{H} = \mathbf{H}_c + \mathbf{H}_m.$$  

$\mathbf{H}_c$ is the conductor magnetic field, which can be calculated using the Biot–Savart law:

$$\mathbf{H}_c = \frac{1}{4\pi} \int_\mathcal{V} \frac{\mathbf{J} \times \mathbf{r}}{|\mathbf{r}|^3} \, dV.$$
\( \mathbf{H}_m \) is the rest of the field where \( \nabla \times \mathbf{H}_m = 0 \). Therefore, the magnetic field \( \mathbf{H}_m \) can be determined by:

\[
\mathbf{H}_m = -\nabla \phi,
\]

where \( \phi \) is the magnetic scalar potential.

Using the magnetic permeability, \( \mu \), the magnetic field and the magnetic flux density can be related by the following equation:

\[
\mathbf{B} = \mu \mathbf{H}.
\]

Given that divergence of the flux density is always zero. The magnetic field can be derived using the following equation:

\[
\nabla \cdot \mu \nabla \phi - \nabla \cdot \mu \left( \frac{1}{4\pi} \int_{V} \frac{\mathbf{J} \times \mathbf{r}}{|\mathbf{r}|^3} \, dV \right) = 0.
\]

When there is a structure made of a material with high permeability in the problem, the difference between the first and second terms of (7) is quite small inside the high permeability region. Therefore, the error in the field calculation dominates the difference between the first and second terms of (7). To avoid this problem, the volume of the problem is divided into two regions (Simkin and Trowbridge 1979, 1980). The first region is the one with all the conductors and no material with high \( \mu \), where (7) is applied to calculate the scalar potential. The second region is the one with no currents and the material with the high permeability. Since \( \mathbf{J} = 0 \) in the second region, the scalar potential, \( \psi \), is calculated for this region using \( \nabla \cdot \mu \nabla \psi = 0 \), derived from (1) and (2). Then the solutions for two regions must be matched at the interface of two regions with proper boundary conditions valid at the interface (Simkin and Trowbridge 1979, 1980).

The nonlinear magnetization curves for the materials used in magnet are shown in figure 3. The 1020 steel data are assembled based on data from Brauer (2013) and Gloria et al (2009) and the GO silicon steel are obtained from product documentation from AK steel (M-2MILL-ANNEAL grain-oriented electrical steel). The crystal alignments for GO silicon steel are such that they have a greater \( \mu \) in one direction versus the other. This design is modeled based on the magnetization curve for the rolling direction (preferred direction), which is placed along \( B_z \) direction for this design. This is the optimal direction for maximal permeability constant. GO silicon steel responds better at low field than the 1020 steel used for the yoke structure. As a result, small amounts of flux that reach the pole plate will still result in large magnetization. The relatively high resistivity of GO silicon steel (Haiji et al 1996) is also an asset, allowing for a reduction in eddy currents that can be induced by MR imaging gradient coils.
A large cube with a dimension of 12 m is set to be the outer boundary. The tangential magnetic field condition, $\mathbf{H} \cdot \mathbf{n} = 0$, is applied for the outer boundary, which is equivalent to $\frac{\partial \psi}{\partial n} = 0$. $\mathbf{n}$ is the outward normal unit vector at the surface of the cube. However, the model is symmetric so one can solve only one-eighth of the model using three cutting planes, which are the internal boundary of one-eighth of the model. The internal $XZ$ and $YZ$ planes should have the tangential magnetic field condition. But, for the internal $XY$ plane, the normal magnetic field condition is applied, $\mathbf{H} \times \mathbf{n} = 0$, which means having a constant scalar potential across this boundary plane. Given a well-defined boundary-value problem, the FEM technique can be applied to derive the scalar potentials $\phi$ and $\psi$ numerically using Opera-3D, allowing for a magnetic field solution.

The FEM model is divided into approximately $1.3 \times 10^5$ tetrahedral elements. The sizes of the elements vary for the various compartments of the model. The minimum element size is about 1.5 cm for the surface of the target ellipsoid and the maximum size is about 240 cm for the outermost boundary.

2.4. Pole plate optimization

The outer surface of the pole plate, surface of the pole plate in $XY$-plane (shown in figure 4(a)), is optimized subject to the minimization of a cost function, which evaluates the uniformity of the magnetic field over an ellipsoid. The cost function, $f$ with units of $T^2$, is calculated as:

$$f = \sum_{i=1}^{n} \left[ B(\mathbf{r}_i, \mathbf{P}) - B_0 \right]^2. \quad (8)$$

The target surface defined in this work is an ellipsoid with the major axis $(Y)$ of 0.60 m, the minor axes $(X, Z)$ of 0.40 m, and its center located at the isocenter. Due to the symmetry, magnetic flux is calculated over only one-eighth of the ellipsoid surface, at $n = 692$ uniformly distributed points over the surface. $B(\mathbf{r}_i, \mathbf{P})$ is the magnetic flux magnitude at point $i$ on the surface patch for $\mathbf{P}(Z_1, Z_2, ..., Z_N)$ is the design vector, where $Z_i$ is the thickness of the pole plate for the defined point on the surface of the pole plate in the $XY$ plane. $B_0$ is the magnetic flux at the isocenter. The cost function is therefore minimum when the field distribution is uniform relative to that at isocenter.
Considering the symmetry of the pole plate, only one fourth of the surface needs to be optimized. The pole plate quadrant is divided to four sections using five spokes, \( N_{\text{spk}} = 5 \), located at \( 0, \pi/8, 3\pi/8, \pi/2 \). Further, each spoke is represented by \( N_r \) control points as is displayed in figure 4(b). The inner radius of the pole plate is located at \( r = 0.12 \) m and its outer radius extends to \( r = 0.87 \) m. \( N_r = 31 \) evenly-spaced parameters were chosen for the radial direction. Therefore, the distance between two control points is 2.5 cm. This sampling density resulted in many optimization parameters, but given the unknown nature of the solution, it was felt that it was better to err on the side of excess samples than to impose a potentially limiting assumption to the sampling density.

The total number of control points on the surface of the pole plate in \( XY \) plane are \( N_{\text{spk}} \times N_r \). A value \( Z_i \) is assigned to each control point. Using four neighboring points located in two adjacent spokes and arcs, a structure called a projection was constructed, depicted in figure 4(b). The surface of the pole plate is constructed piecewise using these projections. In order to have a surface that remains faithful to the thickness at the four neighbouring points, defining the boundaries of the projection, the projection is divided to four triangular prisms. The middle point for the surface of the projection in the \( XY \) plane is the centre of the trapezoid, which formed using four neighboring points, and the \( Z \) value of the middle point is the average \( Z \) values of those points.

The expectation of a smoothly varying pole plate in the azimuthal direction allows the variation of these control points to be modeled as a Fourier series. However, the model is symmetric about \( \theta = 0 \) so there are no antisymmetric terms (i.e. Sine function). In addition, given that by design the model must be periodic in \( \pi \), only even frequencies are permitted. Finally, the thickness, \( Z \), for each point is defined by a truncated Fourier series where cyclic changes to four cycles per revolution have been included as:

\[
Z = A_0(r_k) + A_2(r_k) \cos(2\theta_j) + A_4(r_k) \cos(4\theta_j). \tag{9}
\]

\( r_k \) is the radial coordinate of point \( k \) on \( y = 0 \) axis and \( \theta_j \) is the angle of each spoke. Using (9), the total number of parameters are reduced from \( N_{\text{spk}} \times N_r \) to \( 3N_r \) since the points on one arc but different spokes are not independent. Finally, the thickness of the pole plate can be defined by (9) using low-resolution sinusoids with even coefficients of \( \theta_j \).

The surface of the pole plate is optimized in 3 steps by considering one term of (9) at each step. \( A_0(r_k), A_2(r_k), \) and \( A_4(r_k) \) are the variables at each step. This further reduces the number of parameters at each stage of the optimization to \( N_r \). The optimization was started with five spokes in angular direction. After all three terms in (9) were optimized, the resolution was increased in angular direction to nine spokes using the best result for five spokes to avoid the sharp edges in angular direction.

Because of having many variables to be optimized, the particle swarm optimization (PSO) (Kennedy and Eberhart 1995), which is a stochastic algorithm searching for the global best solution, was found to be the most suitable choice to do the optimization. The pattern search algorithm, on the other hand (also known as the Nelder-Mead simplex method (Lagarias et al 1998)), can be used as a complementary algorithm to locate the local best solution after the PSO search for the global minimum. The pattern search algorithm is a derivative-free algorithm that converges quickly to a local minimum and it requires a good initial solution. Therefore, in this work, the PSO was used as the prime algorithm used for the optimization and the pattern search algorithm was used for the fine-tuning of the solution.

During optimization, it was found that the homogeneity can be improved by placing a bulk metal beside the coil. Considering a circular field map (0.40 m diameter) centered at isocentre in the \( XY \) plane, the regions close to the superconducting coils exhibit hot spots, such as that seen in the magnetic field map shown in figure 5(a). This makes the optimization process
difficult and this problem cannot be addressed by the pole plate optimization alone, since there is not enough room with the pole plate for the adjustment to accommodate these hot spots. Therefore, two metal bulk shims consisting of GO silicon steel were added in the vicinity of the coils to compensate. The presence of these metal bulk shims pulls some of the excess magnetic flux away from the field of interest and partially corrects for the hot spots seen on the lateral edges. Figure 5(b) shows the magnetic field map for the same FOV as in figure 5(a) after adding the bulk metal shims, using trial and error to determine a thickness that generates a map with a symmetric appearance.

3. Results and discussion

The optimization was started using the model with the pole plate as a bored cylinder. In this initial state, the $Z$ value equal to $0.45 \text{ m}$ was assigned to all control points on the pole plate. The objective function value defined by (8) is $2.25 \times 10^{-2}$ for the bored cylinder as the pole plate.

While beginning the optimization with a flat pole plate is feasible and has been performed successfully in-house for previous models, in order to reduce the number of iterations required to converge (each iteration takes on the order of 90 min) the solution from a different model was imported to be the initial solution. This other model was a 0.2 T unit with a similar yoke design, and while the pole plate will clearly not be exactly appropriate for this new 0.5 T design, it was felt that the solution would be closer to convergence than a flat pole plate. When this solution (displayed in the appendix for completeness) was imported to the new design, the cost function value was reduced to $2.84 \times 10^{-4}$. From this starting point, the three terms of (9) were sequentially optimized using the PSO, eventually reducing our cost function to $8.59 \times 10^{-5}$.

At this stage we elected to increase the azimuthal pole-plate resolution by defining (9) over 9 spokes instead of 5 to better approximate the higher order terms. Unfortunately this process increased the cost function to $1.37 \times 10^{-4}$. To restore some of the deteriorated cost function before turning the problem over to the pattern search algorithm for refinement, slight trial-and-error modifications were made to the solution by looking at the field solutions in the XY plane, and observing what orders of fluctuating field were present. As seen in table 1, these adjustments brought the cost function back down to $7.1 \times 10^{-5}$, after which the pattern search algorithm was used to refine the solution down to a final cost function value of $6.09 \times 10^{-5}$. 

Figure 5. (a) Magnetic field map of the XY plane of an sphere with no bulk metal shim. (b) Magnetic field map of the XY plane of a sphere with added bulk metal shim. The maximal extent of the bulk metal shim in the Y-direction is defined between 55 and 76 cm from isocentre, maintaining a total patient access of 110 cm. The color bars have the same span for both designs.
Convergence rate of \( f_{val} \) values from PSO optimization (table 1) shows that the first term is the most dominant one. The higher order terms respond to cyclic changes in magnetic field induced by asymmetries in the pole plate. However, cyclic changes up to four cycles per revolution have been included in equation (9), and based on the progression of \( f_{val} \) it is expected that higher order terms (6 and above) would result in only incremental improvement.

The inhomogeneity of the field was measured over the defined surfaces for the final solution using:

\[
\Delta B = \max \left( \left| \frac{B(\vec{r}) - B_0}{B_0} \times 10^6 \right| \right)
\]

\( B(\vec{r}) \) is the magnetic flux value at point \( \vec{r} \) located on a defined surface and \( B_0 \) is the magnetic flux value at the iso-centre. The \( \Delta B \) value for the flat pole plate is equal to 21 750 ppm for the surface of the ellipsoid 40-60-40. The \( \Delta B \) values for the final solution for different surfaces are reported in table 2.

Most of the anatomy will fit in an ellipsoid with major axis of 50 cm and minor axes of 30 cm. The total \( \Delta B \) value for this surface is 400 ppm. Magnetic field maps for this ellipsoid at different planes are shown in figure 6.

The \( \Delta B \) can be translated into spatial distortion using:

\[
\text{Spatial distortion (SD)} = \frac{\Delta B(\text{mT})}{G(\text{mT m}^{-1})},
\]

where \( G \) is the gradient strength measured in mT m\(^{-1}\). The spatial distortions associated with the different gradient strengths, using \( \Delta B \) values from table 2, are reported in table 3.

The spatial distortion is not sufficient for the final design implementation. However, it is standard MR practice to incorporate a system of passive and sometimes active shims to accommodate any remaining inhomogeneity for the magnetic field upon installation. These established techniques allow for reduction of the inhomogeneity significantly using passive shims. This involves the installation of magnetic pieces at the specific locations to correct for the inhomogeneity. For example, in Zhu et al (2015), the homogeneity is reduced by the order 10 using passive shimming and Jin et al (2009) showed the inhomogeneity reduction of the order 60 for DSV36 by applying passive shimming. Following this standard procedure, the residual spatial distortion is expected to be sub-millimetre for small volumes and on the order of a few millimeters on the outer regions of larger volumes.

While for diagnostic purposes, spatial distortions on the order of several millimeters will not affect the utility of the imaging, an accuracy of a millimetre or better is required for

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**Table 1. Optimization results.**

<table>
<thead>
<tr>
<th>Pole plate structure</th>
<th>Cost function using (8)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flat pole plate</td>
<td>( f = 2.25 \times 10^{-2} )</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Projections</th>
<th>Algorithm</th>
<th>( f_{val} )</th>
<th>( f_{val} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>30 projections—5 spokes</td>
<td>PSO-first term</td>
<td>2.84 \times 10^{-4}</td>
<td>2.16 \times 10^{-4}</td>
</tr>
<tr>
<td></td>
<td>PSO-second term</td>
<td>2.16 \times 10^{-4}</td>
<td>1.69 \times 10^{-4}</td>
</tr>
<tr>
<td></td>
<td>PSO-third term</td>
<td>1.69 \times 10^{-4}</td>
<td>8.59 \times 10^{-5}</td>
</tr>
<tr>
<td>30 projections—9 spokes</td>
<td>Manual solution</td>
<td>1.37 \times 10^{-4}</td>
<td>7.10 \times 10^{-5}</td>
</tr>
<tr>
<td></td>
<td>Pattern search algorithm</td>
<td>7.10 \times 10^{-5}</td>
<td>6.09 \times 10^{-5}</td>
</tr>
</tbody>
</table>
the radiotherapy (Breeuwer et al 2001, Baldwin et al 2007). When using the images from the larger field-of view, the residual distortion can be reduced to the sub-millimetre level by applying geometric correction algorithms to correct for the spatial distortion (Breeuwer et al 2001, Wang et al 2004, Doran et al 2005, Baldwin et al 2007). Using these algorithms to do further correction on the larger FOV regions, the magnet can also be used for the patient set-up.

Table 2. ΔB for the best solution of 0.5 T model.

<table>
<thead>
<tr>
<th>FOV</th>
<th>ΔB (ppm)</th>
<th>ΔB_{XY} (ppm)</th>
<th>ΔB_{XZ} (ppm)</th>
<th>ΔB_{YZ} (ppm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ellipsoid 40-60-40</td>
<td>1370</td>
<td>1300</td>
<td>1150</td>
<td>980</td>
</tr>
<tr>
<td>Ellipsoid 30-50-30</td>
<td>400</td>
<td>320</td>
<td>300</td>
<td>140</td>
</tr>
<tr>
<td>DSV40b</td>
<td>1150</td>
<td>240</td>
<td>1150</td>
<td>700</td>
</tr>
<tr>
<td>DSV30</td>
<td>300</td>
<td>65</td>
<td>300</td>
<td>90</td>
</tr>
<tr>
<td>DSV20</td>
<td>70</td>
<td>40</td>
<td>70</td>
<td>50</td>
</tr>
</tbody>
</table>

*a* Part per million.

*b* Diameter of spherical volume.

Figure 6. (a) Magnetic field map for the XY plane of the ellipsoid 30-50-30. (b) Magnetic field map for the XZ plane of the ellipsoid 30-50-30. (c) Magnetic field map for the YZ plane of the ellipsoid 30-50-30.

Table 3. The spatial distortions associated with the different gradient strengths.

<table>
<thead>
<tr>
<th>G (mT m^{-1})</th>
<th>FOV</th>
<th>SD (mm)</th>
<th>SD_{XY} (mm)</th>
<th>SD_{XZ} (mm)</th>
<th>SD_{YZ} (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>Ellipsoid 40-60-40</td>
<td>69.5</td>
<td>65.9</td>
<td>58.3</td>
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It was shown that the presence of the field at the electron gun results in a target current loss in the parallel configuration (St Aubin et al 2010). In past designs, passive shielding has been implemented to reduce the field at the electron gun (Santos et al 2012). Figure 7 shows a cross-section of the magnet and optimized pole plate, together with a field map in the region of the MLC and linac target. As can be seen in figure 7, the field at the target is on the order of 400 G, meaning shielding at the linac would be essential for this model.

4. Conclusions

A superconducting magnet was designed with moderately reduced SAD of 123 cm and the gap of 68 cm with the magnetic field of 0.5 T at the isocentre. This novel design has several advantages over the current one (Fallone 2014). The separation distance between two magnetic poles is larger so better accessibility and more room to accommodate the patient can be achieved. The magnet assembly is designed to be more compact in certain dimensions so the linac head can be moved closer to the isocentre compared to the current magnet. This model is optimized to address the magnetic field homogeneity over a larger region; establishing the potential for a wider field-of-view. For the current magnet in the linac-MR system, the modelled $\Delta B$ value for DSV40 is 6441 ppm before any secondary shimming. However, it is 1150 ppm for DSV40 for this magnet design. The final magnetic field is not sufficiently homogenous for raw implementation, but further passive shimming can be implemented to create a field capable of the use with radiotherapy.

This non-axial design is in the early stages of development. While the model presented in this work achieves a magnet design with somewhat better dimensions than the conventional approach, the lack of a cryostat directly above and below the patient permits further optimizations (in the form of a sculpted pole plate with reduced profile) that either enhance patient access or further reduce the SAD. In addition, the 5 G field currently extends out to the walls of the vault, where shielding would prevent public exposure. Further yoke optimization with the goal of reducing magnetic saturation could allow this 5 G field to be improved. Lastly, the pole plate parameters for this non-axial design are sampled uniformly in the radial direction. Exploring the non-linear sampling such as logarithmic sampling for the pole plate in the radial direction to minimize sharp projections that may not have a large effect on homogeneity is the subject of the future work. It is possible that reducing the sharp projections will allow for a reduced pole-plate profile.
Acknowledgments

The authors would like to acknowledge Alberta Innovations-Health Solutions (AIHS) for funding.

Conflict of Interest

B Gino Fallone is a co-founder and CEO of MagnetTx Oncology Solutions.

Appendix. Vector solution for $f_{val} = 2.84 \times 10^{-4}$ presented in table 1 ($A_2(r_k)$ and $A_4(r_k)$ are zero vectors for PSO-first term.)

$$A_0(r_k) = \begin{bmatrix} 0.3789 & 0.3998 & 0.4207 & 0.4952 & 0.5697 & 0.4599 & 0.3501 \\ 0.4583 & 0.5664 & 0.4980 & 0.4295 & 0.4839 & 0.5382 & 0.4882 \\ 0.4382 & 0.4478 & 0.4575 & 0.5078 & 0.5581 & 0.4781 & 0.3982 \\ 0.3790 & 0.3599 & 0.4270 & 0.4942 & 0.4792 & 0.4641 & 0.4485 \\ 0.4329 & 0.4235 & 0.4142 \\ \end{bmatrix}. \quad (A.1)$$

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