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Asymmetric gradient coil design for use in a short, open bore magnetic resonance imaging scanner

Yaohui Wang, Feng Liu*, Yu Li, Fangfang Tang, Stuart Crozier

School of Information Technology and Electrical Engineering, The University of Queensland, St Lucia, Brisbane, QLD 4072, Australia

Corresponding author: Feng Liu Email: feng@itee.uq.edu.au

Abstract

A conventional cylindrical whole-body MRI scanner has a long bore that may cause claustrophobia for some patients in addition to being inconvenient for healthcare workers accessing the patient. A short-bore scanner usually offers a small sized imaging area, which is impractical for imaging some body parts, such as the torso. This work proposes a novel asymmetric gradient coil design that offers a full-sized imaging area close to one end of the coil. In the new design, the primary and shielding coils are connected at one end whilst separated at the other, allowing the installation of the cooling system and shim trays. The proposed coils have a larger wire gap, higher efficiency, lower inductance, less resistance and a higher figure of merit than the non-connected coils. This half-connected coil structure not only improves the coils’ electromagnetic performance, but also slightly attenuates acoustic radiation at most frequencies when compared to a non-connected gradient coil. It is also quieter in some frequency bands than a conventional symmetric gradient coil.

Keywords: MRI; gradient coil; asymmetric coil; magnetic field; noise

1. Introduction

A conventional long-bore cylindrical whole-body magnetic resonance imaging (MRI) scanner is claustrophobic for some patients, thus making them uncomfortable during scanning. One method to make the scanner more open is to enlarge the diameter of the bore. However, this will increase the cost of the magnet and influence the uniformity of the magnetic field. Another method is to move the region of interest (ROI) towards one end of the main magnet [1]. An asymmetric gradient coil design is focused on this to overcome the claustrophobia problem and is paired with an asymmetric magnet design concept [2, 3].

Gradient coils are a critical part of an MRI scanner, providing a means of frequency-encoding in the region of interest (ROI). With the development of MRI techniques, the gradient field must be strong and pulsed quickly to enable rapid imaging [4, 5]. However, fast, strong gradient switching can induce significant acoustic noise, making some patients uncomfortable [6, 7]. These issues can be attenuated to some extent by an appropriate coil design scheme, low inductance, low eddy current loss, low acoustic noise level, and so on.

Recent developments in gradient coil design methods allow for the design of arbitrary geometries [8-12] that provides many possibilities for improving the gradient coil performance. For example, the ultra-short gradient coil was designed with three-dimensional (3D) geometry for ultra-short
cylindrical MRI systems, where the length of the gradient coil can be controlled throughout the design process [13]. However, compared with standard long gradient coil sets, the short, layered gradient coils (both having the same ROIs, design methods, border conditions and similar coil patterns, and so on) tend to have a dense coil pattern. The gradient field generating arcs (x, y coils) are competing for space with the return wires, making some wire-wire distances too narrow to manufacture. In addition, thermal heating can be a problem and the inductance may be higher. Coils with two ends connected were proposed to allow the current flow from the primary surface to the shielding surface [8, 14]. Under the same design parameters, this method can relax the current distribution to some extent compared with the conventional non-connected primary and shielding coils, because some return path wires are laid on the connected surface. However, this design has a higher complexity in terms of the mechanical design.

Apart from considering the electromagnetic performance of the gradient coil design, the amount of noise generated by the coil also needs to be considered in the design process to improve patient comfort for seriously ill or anxious patients [7, 15-18]. With respect to the acoustic noise level reduction of the gradient coils, traditional methods for coping with this problem include wearing earplugs, earmuffs or even a helmet [19] or applying a damping treatment on the gradient assembly [20, 21]. Some scanners use a vacuum device to block the airborne noise propagation to the patients’ ears [22, 23]. Active actuators are also reported to be mounted on the ends of the gradient assembly to reduce the noise radiation [24]. In addition to these methods, a more straightforward method for acoustic noise reduction is to design the gradient coil by minimising the force/torque at the source [25].

In this work, we propose a novel asymmetric gradient coil design pattern matching an asymmetric magnet design concept [2]. The primary and shielding surfaces of the gradient coil were connected at one end, but separated at the other, to allow for the installation of the cooling device and shim tray, which also provided more space for the coil wire distribution. An equivalent magnetization current method was applied to the design of the gradient coil [8]. For the acoustic analysis, the finite element method (FEM) was used, where the gradient coil was inserted into an epoxy resin. The electromagnetic performance and acoustic radiation intensity of the designed asymmetric gradient coil were compared with a non-connected asymmetric gradient coil. Its acoustic characteristics were also compared with a conventional symmetric gradient coil.

2. Methods

2.1 Asymmetric MRI scanner configuration

Based on an asymmetric magnet design concept [2], an asymmetric gradient coil was designed for the short MRI magnet, where the ROI was located near one end of the gradient coil. Fig. 1 shows an asymmetric scanner whose ROI is near the end of the scanner and a symmetric counterpart with the ROI at the centre of the scanner, which is also plotted for comparison. When doing chest imaging for a normal adult patient in a symmetric scanner, the patient’s head will sit in the cylindrical tunnel of the scanner. In contrast, in an asymmetric scanner, the patient’s head will sit at the edge of the scanner, thus potentially reducing the patient’s discomfort and claustrophobia. In addition, as Fig. 1 shows, this design may have advantages for interventional imaging.
2.2 Gradient coil design

The gradient coil design was implemented using our recently-developed equivalent magnetization current method [8, 26]. An asymmetric x coil, whose primary current surface and shielding current surface were connected at one end, but separate at the other end (defined as a connected coil in this work), was designed to provide more dimensions for the spatial distribution of the current density. For the comparison of the coil performance, a corresponding layered asymmetric x coil (defined as a non-connected coil) was designed. The design strategy for both coils was a combination of conventional short (with high current density at both ends) and 3D short (with reduced current density at both ends) with a shift of ROI towards the patient end and an extended length at the other end. The target field and current density surfaces used in the coil design are plotted in Fig. 2 (a) and (b) and the ROI designed here was 500 ×500 ×400 mm (xxyxz). The coil and cryostat sizes are shown in Fig. 2 (c) and a diagram of the designed coil layers is illustrated in Fig. 2 (d). All the gradient coils designed here had a target gradient strength of 30 mT/m and the maximum field error in the ROI was constrained to ±5% when the coils were designed. The shielding ratios [27] were controlled to be 2% during the design process and the parameters of the cryostat used for the eddy current control are listed in Table I.
Fig. 2. Asymmetric gradient coil configurations, (a) current density surface of the connected coil, (b) current density surface of the non-connected coil, (c) dimensions of the designed x coils and the cryostat and, (d) diagram of the coil layers in a gradient assembly. The ROI shift size is 0.23 m

Table I. Parameters of the cryostat used for the eddy current control during the coil design process

<table>
<thead>
<tr>
<th></th>
<th>Length (m)</th>
<th>Radius (m)</th>
<th>Thickness (mm)</th>
<th>Conductivity (S/m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Warm bore</td>
<td>1.460</td>
<td>0.460</td>
<td>6</td>
<td>$1.1 \times 10^6$</td>
</tr>
<tr>
<td>First cold shield</td>
<td>1.460</td>
<td>0.472</td>
<td>3</td>
<td>$3.8 \times 10^7$</td>
</tr>
<tr>
<td>Secondary cold shield</td>
<td>1.460</td>
<td>0.482</td>
<td>3</td>
<td>$1.2 \times 10^9$</td>
</tr>
</tbody>
</table>

2.3 Acoustic analysis

With respect to the acoustic radiation intensity, a simple mechanical analysis can be used to explain the advantage of the designed coil. Eq. (1) is a Navier’s equation [28], where $f_v$ is volume force, $B$ is derivative matrix, $\sigma$ is stress and $\rho$ is density and $e$ is displacement. If a torque is applied at the ends of the gradient assembly, it is expected from the Navier’s equation that the radial displacement will be influenced. Fig. 3 (a) and (b) show a torque diagram at the connected end of the designed coil in a gradient assembly. Taking a cell of the gradient assembly for a spatial stress and strain analysis [29], as shown in Fig. 3 (c) and (d), the radial strain will be attenuated owing to a tangential stress. Eq. (2) is a derived expression based on the cell stress state analysis, where $\Delta$ is attenuated distance from the principal stress direction, $E$ is the Young’s modulus, $v$ is the Poisson’s ratio, $\tau_{xz}$ is the shear stress and $l$ is the edge length of the cell. From Eq. (2), increasing the shear stress $\tau_{xz}$ can better restrain the displacement of the principal direction. Based on the above analysis, it can be predicted that the torque at the end of the gradient coil can attenuate the displacement amplitude of the radial direction.
The dynamic vibration properties and possible acoustic advantage will be evaluated by an acoustic harmonic analysis.

$$f_i + B^T \sigma = \mu \ddot{e}$$

$$\mathbf{B} = \begin{bmatrix}
\frac{\partial}{\partial r} & 0 & 0 & 0 & \frac{1}{r} \frac{\partial}{\partial \theta} \\
0 & \frac{1}{r} \frac{\partial}{\partial \theta} & 0 & 0 & \frac{\partial}{\partial z} \\
0 & 0 & \frac{\partial}{\partial z} & 1 & \frac{\partial}{\partial r} \\
\sigma_{zz} & \sigma_{z \theta} & \sigma_{z \theta} & \sigma_{tt} \\
\sigma_{zt} & \sigma_{t \theta} & \sigma_{t \theta} & \sigma_{zz}
\end{bmatrix}$$

Fig. 3. Mechanical diagram of the connected coil, (a) connected coil in an assembly, (b) torque direction at the connected end of the designed coil, (c) two-dimensional (2D) stress state analysis without shear stress and, (d) 2D stress state analysis with shear stress, where $l$ is the edge length of the cell, $\sigma_i$ is the principal stress, $\tau_{xy}$ is the shear stress and $\theta$ is the torsion angle due to the shear stress. A coil example was illustrated to introduce the torque analysis.

$$\Delta = \left( 1 - \frac{E}{\sqrt{E^2 + 4(1+\nu)^2} \tau_{xy}^2} \right) l$$

For the acoustic noise analysis of the gradient coil, a three-dimensional (3D) model was established. The gradient assembly was simulated as an epoxy resin cylinder surrounded by air, the ends of which were fixed[30]. Fig. 4 (a) is a 3D finite element (FE) model, where the gradient assembly was placed in free space with an infinite boundary (no acoustic wave reflection). The infinite boundary was modelled using the Fluid 130 element type in the ANSYS element type library, which was applied on the outer surface of the air sphere. The size of the gradient assembly is displayed in Fig. 4 (b) and the mechanical and acoustic parameters used in the simulation are listed in Table I. The temperature-caused material property variation was not considered in the simulation. For the model establishment,
the element size was controlled to be less than 1/6 of the smallest wave acoustic wave length [31]. Harmonic analysis was used here from 100 Hz to 3000 Hz. The sinusoidal peak current used to energize the coil was the coil design current, producing a gradient strength 30 mT/m. Apart from the acoustic comparison between the connected asymmetric gradient coil and the non-connected gradient coil, a conventional symmetric gradient coil was also designed to compare the acoustic differences between the asymmetric coil pattern and symmetric coil pattern. As with the asymmetric coils, the symmetric coil also produces a 30-mT/m gradient strength. For the Lorentz force calculation, an asymmetric 3 T static magnetic field and a symmetric 3 T static magnetic field were designed to pair the asymmetric gradient coils and the symmetric gradient coil.

![Image](image-url)

**Fig. 4.** 3D FE model for the acoustic analysis of the gradient coil, (a) gradient assembly surrounded by air and (b) dimensions of the gradient assembly

<table>
<thead>
<tr>
<th>Item</th>
<th>$E$ (Gpa)</th>
<th>$\mu$</th>
<th>$\rho$ (kg/m$^3$)</th>
<th>$c$ (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gradient assembly (epoxy resin)</td>
<td>15.7</td>
<td>0.30</td>
<td>1835</td>
<td></td>
</tr>
<tr>
<td>Surrounding air</td>
<td>1.225</td>
<td></td>
<td>340</td>
<td></td>
</tr>
</tbody>
</table>

$E$, $\mu$ and $\rho$ are the Young’s modulus, Poisson’s ratio and density of the simplified gradient assembly. $c$ is the acoustic velocity.

3. Results

3.1 Coil performance evaluation

Fig. 5 shows the designed asymmetric gradient coil wire contours. The connected coil and non-connected coil are plotted in Fig. 5 (a) and (b) respectively. It can be seen from Fig. 5 that the wire distribution of the connected coil is slightly looser than the non-connected coil owing to the connected end — this can alleviate the heating problem to some extent [32].
Fig. 5. Asymmetric gradient coil wire contours, (a) connected coil and (b) non-connected coil

Fig. 6 is the z-component magnetic field distribution map of the connected coil and non-connected coil, whereby (a) and (b) are the magnetic field distributions of the connected coil and non-connected coil, respectively, on the cutting plane $y=0$. Comparing Fig. 6 (a) and (b), these two coils have nearly identical magnetic field distributions. However, small differences can be appreciated, especially at the peripheral magnetic fields outside the ROIs, where the $z$-component magnetic field of the non-connected coil is slightly higher than the connected coil. Fig. 7 (a) and (b) show the equipotential contours of the $z$-component magnetic field produced by the inputting current of the coils on the plane $y=0$, where the ROIs of the connected coil and non-connected coil were plotted. The lines in the ROIs of the two coils are approximately straight, which indicates uniform gradient magnetic fields. Outside the ROIs, the magnetic field uniformities of the two coils drop quickly and their stray fields show some differences. Fig. 7 (c) and (d) illustrate the equipotential contours of the $z$-component magnetic field produced by the eddy current on the cryostat. Also, uniform gradient magnetic fields were formed. The calculation of the magnetic field error on the ROI surface considering the eddy current induced field shows $4.49\%$ and $4.55\%$ deviations from the target field for the connected coil and non-connected coil respectively.
Fig. 6. Z-component magnetic field distributions of the asymmetric gradient coils on the cutting plane y=0, (a) connected coil, (b) non-connected coil

Fig. 7. Equipotential contours of the z-component magnetic field on the cutting plane y=0, (a) directly-produced magnetic field of the inputting current on the connected coil, (b) directly-produced magnetic field of the inputting current on the non-connected coil, (c) induced magnetic field of the connected coil on the cryostat and, (d) induced magnetic field of the non-connected coil on the cryostat
Minimum wire gaps are illustrated for both the connected coil and non-connected coil, as shown in Fig. 8. Because of the asymmetric properties of the gradient coils, the wires are concentrated towards one end together with the ROI. Minimum-gap wires occur at the ends of the primary coils for the two coils. This may become a problem for the fabrication and heating. However, the connected coil can alleviate the wire gap problem to some extent if both kinds of coils have a dense wire distribution, because the connected end provides another space for the wire laying. For the designed coils, the connected coil has a minimum wire gap 6.8 mm, while the non-connected coil has a minimum wire gap 4.7 mm. The comparative advantage of the connected coil may alleviate the heating problem and reduce the coil inductance.

For further comparison of the designed coils, some essential electromagnetic performances of the designed gradient coils and a symmetric coil are listed in Table III. The two asymmetric coils both have 168 wire turns. For a target gradient magnetic field strength of 30 mT/m, the connected coil is operated with a current of 323.5 A, while the non-connected coil has a current input of 328.1 A. Thus, the connected coil has a comparatively higher efficiency than the non-connected coil. In addition, the inductance and resistance of the connected coil are both smaller than the non-connected coil, where the smaller inductance can make a high slew rate and the smaller resistance can reduce a power loss. An integrated parameter figure of merit, which is used to evaluate the overall performance of the gradient coil, also indicates the advantage of the connected coil. When compared with a symmetric coil, the electromagnetic performances of the asymmetric coil do not always show an advantage, although the efficiencies are a little higher. The torques of the asymmetric coils and a symmetric coil were calculated under a 3 T homogeneous static magnetic field. As listed in Table IV, the asymmetric coils have larger torques than the symmetric coil on the y because of their asymmetric properties.
Fig. 8. Minimum wire gaps for the asymmetric gradient coil designs, (a) minimum-gap wires shown in the connected coil, (b) minimum-gap wires of the connected coil, (c) minimum-gap wires shown in the non-connected coil and, (d) minimum-gap wires of the non-connected coil

| Table III. Electromagnetic performances of the connected coil, non-connected coil and a symmetric coil |
|-------------------------------------------------|-------------------------------------------------|------------------|------------------|----------------------|
| Connected coil                                  | Non-connected coil                               | Symmetric coil   |
| Maximum current (A)                             | Efficiency (mT/m/A)                              | Inductance (μH)  | Resistance (mΩ) | Figure of merit (T²/m²/A²/H) |
| 323.5                                          | 0.093                                           | 1014.2           | 449.2           | 8.48×10⁻⁶             |
| 328.1                                          | 0.091                                           | 1073.5           | 459.7           | 7.79×10⁻⁶             |
| 387.7055                                       | 0.077                                           | 631.1            | 354.4           | 9.49×10⁻⁶             |

| Table IV. Torques of the connected coil, non-connected coil and a symmetric coil |
|-------------------------------------|---------------------|---------------------|
| Connected coil                      | Tx (N/m)            | Ty (N/m)            | Tz (N/m)         |
| 0.5172                              | 423.9               | 0                   |
| Non-connected coil                  | 0.6365              | 209.2               | 0                 |
| Symmetric coil                      | 0.2640              | 0.0399              | 0                 |
3.2 Acoustic radiation intensity evaluation

The designed static magnetic fields used for the Lorentz force calculation are shown in Fig. 9, where Fig. 9 (a) and (b) display the longitudinal (z) component and radial (r) component of the asymmetric static magnetic field, and Fig. 9 (c) and (d) display the longitudinal (z) component and radial (r) component of the symmetric static magnetic field. Note that the designed static magnetic fields are axi-symmetric with respect to the z axis. The illustrations of the static magnetic fields around the gradient coils are plotted in Fig. 9 (e) and (f), where the connected asymmetric gradient coil under the asymmetric 3 T static magnetic field was taken as an example. The calculated Lorentz force was mapped on the gradient assembly.

![Fig. 9. Static magnetic fields for the Lorentz force calculation (target field in the ROI is 3 T), (a) longitudinal z component of the asymmetric static magnetic field, (b) radial r component of the asymmetric static magnetic field, (c) longitudinal z component of the symmetric static magnetic field, (d) radial r component of the symmetric static magnetic field, (e) connected coil under longitudinal z component of the asymmetric static magnetic field and, (f) connected coil under radial r component of the asymmetric static magnetic field. The designed static magnetic fields are axisymmetric with respect to the z axis](image)

A modal analysis was conducted on the gradient assembly before the acoustic investigation. The mode frequencies and their corresponding mode participation factors between 100 Hz to 3000 Hz were
indentified. The main bending vibration modes and their participation factors on six degrees of
freedom are listed in Table V. These modes have a participation factor larger than five in the X or Y
displacement direction. Obviously, a gradient pulse containing much energy at these frequencies may
cause large bending vibration in the gradient assembly.

Table V. The main bending vibration modes of the gradient assembly and their corresponding participation factors

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>X</th>
<th>Y</th>
<th>Z</th>
<th>Rot-X</th>
<th>Rot-Y</th>
<th>Rot-Z</th>
</tr>
</thead>
<tbody>
<tr>
<td>463.9</td>
<td>-0.2285</td>
<td>24.0180</td>
<td>-0.1043e-3</td>
<td>-0.5282e-2</td>
<td>0.1614e8</td>
<td>0.1256e-4</td>
</tr>
<tr>
<td>464.4</td>
<td>24.0190</td>
<td>0.2286</td>
<td>-0.4381e-3</td>
<td>0.1033e-3</td>
<td>0.2176e2</td>
<td>-0.5339e-4</td>
</tr>
<tr>
<td>1296.6</td>
<td>-0.5003e-1</td>
<td>7.8332</td>
<td>0.9168e-2</td>
<td>-0.1932</td>
<td>-0.9095e3</td>
<td>-0.1366e-4</td>
</tr>
<tr>
<td>1318.5</td>
<td>-8.0528</td>
<td>-0.5728e-1</td>
<td>-0.1260e-1</td>
<td>0.4295e-2</td>
<td>-0.2178e1</td>
<td>-0.1414e-3</td>
</tr>
<tr>
<td>2066.6</td>
<td>-0.4030e-1</td>
<td>-5.0534</td>
<td>0.1334e-2</td>
<td>0.1658</td>
<td>0.1942e-2</td>
<td>0.1157e-3</td>
</tr>
</tbody>
</table>

Note: the mode frequencies with participation factors larger than 5 in the X or Y displacement direction are listed.

Fig. 10 illustrates the displacements of the connected coil and non-connected coil in the gradient assemblies by a sinusoidal maximum current input (see Table I) with a frequency of 464 Hz around a
main bending mode of the gradient assembly. Obviously, the large-displacement regions of the non-connected coil are restrained by using a connected coil design that will attenuate the radiated acoustic field from the assembly surface. Fig. 11 shows the sound pressure level (SPL) comparison between the connected and non-connected coils. Also, the SPL of the conventional symmetric coil was plotted to characterize the acoustic differences of the new asymmetric coil and conventional symmetric coil designs. The SPL for a frequency was the averaged result in the cylindrical tunnel [33]. Fig. 11 (a) is
the SPL comparison from 100 Hz to 3000 Hz. Fig. 11 (b), (c) and (d) are partial plots of Fig. 10 (a)
that show only the frequencies from 1500 Hz to 2500 Hz, 500 Hz to 1000 Hz, and 1000 Hz to 1500
Hz, to clearly reveal the SPL differences between the connected and non-connected coils. Fig. 11 (a)
shows that the connected coil has an SPL reduction compared to the non-connected coil for nearly the
whole frequency band; this is further partially illustrated in Fig. 11 (b) and (c).

This is more significant in a higher frequency band than a low frequency band. Although this SPL
difference is not considerable — around 2.2 dB at some frequencies — this SPL reduction indicates
that the connected coil will radiate a lower sound intensity compared with the non-connected coil with
the same pulse pattern. However, there is a small frequency band around 1350 Hz where the SPL of
the connected coil is higher than that of the non-connected coil. That may be because the Lorentz
force distribution of the connected coil was near the peaks of the mode shape (or most forces were
applied near the peak of the mode shape) around that frequency; this could easily cause resonance.
However, if most forces were applied near the zero displacement points of the mode shape around that
frequency, the resonance would be difficult to induce. As with the SPL comparison between the
asymmetric coils and the symmetric coil, the SPL difference has a close relationship with the
frequency. In a frequency band lower than 300 Hz, the SPLs of the asymmetric coils are robustly
lower than that of the symmetric coil. In the frequency band 2200-2550 Hz, the asymmetric coils are
much quieter than the symmetric coil, where the SPL difference is considerable. In other frequency
bands, the SPLs of the asymmetric coil and symmetric coils fluctuate up and down with small
differences.
Fig. 10. The displacement illustrations of the designed coils in the gradient assemblies under a sinusoidal current input at 464 Hz, (a) connected coil and (b) non-connected coil.

Fig. 11. Acoustic radiation intensity comparisons of the connected and the non-connected coils, (a) SPL comparison from 100 Hz to 3000 Hz, and also a comparison with a symmetric coil, (b) SPL comparison from 1500 Hz to 2500 Hz, (c) SPL comparison from 500 Hz to 1000 Hz and, (d) SPL comparison from 1000 Hz to 1500 Hz.
Fig. 12. Acoustic field distribution on the cutting plane y=0 of the cylindrical tunnel, (a) acoustic field distribution of the connected coil at 500 Hz, (b) acoustic field distribution of the connected coil at 1000 Hz, (c) acoustic field distribution of the non-connected coil at 500 Hz, (d) acoustic field distribution of the non-connected coil at 1000 Hz, (e) acoustic field distribution of the conventional symmetric coil at 500 Hz and, (f) acoustic field distribution of the conventional symmetric coil at 1000 Hz.

Fig. 12 shows the acoustic field distributions on the cutting plane y=0 of the cylindrical tunnel in the gradient assembly. Fig. 12 (a), (c) and (e) are the acoustic field comparisons of the connected coil, non-connected coil and conventional symmetric coil respectively, at 500 Hz; and Fig. 12 (b), (d) and (f) are the acoustic field comparisons of the connected, non-connected and conventional symmetric coils, respectively, at 1000 Hz. Similar acoustic field distributions can be observed from the comparison between the connected coil and non-connected coil. However, the connected coil has smaller loud areas than the non-connected coil, thus resulting in a lower average SPL. For the acoustic field comparison between the asymmetric coils and the symmetric coil, the loud areas of symmetric coil tend to concentrate in the central part of the cylindrical tunnel, while the central part of the asymmetric coil is relatively quieter.

Conclusions

A novel asymmetric gradient coil pattern was proposed in this work for a cylindrical short-bore asymmetric MRI scanner. The coil was designed with one end connected, which gave more space for the coil wire placement and also made it possible to mount shim trays and to access them from the other end. This design increased the wire gap and improved the electromagnetic performances of the
coil compared to a non-connected coil and also had a lower acoustic radiation. The new coil design pattern had a higher efficiency, lower inductance and resistance than the corresponding non-connected coil. Using an overall parameter figure of merit to evaluate the proposed coil pattern, it also behaved better than the non-connected coil. According to the acoustic analysis, the proposed coil pattern had an SPL reduction at most frequencies compared with the non-connected coil pattern. It is noted that the finding is based on our FEM model with simplifying approximations, such as the lack of discrete wires or detailed structural components. Future work will be conducted to include the fabrication of the coil and experimental measurement of its performance.

References


Graphical abstract

Highlights

a) A novel asymmetric gradient coil design with ROI shifted was proposed for use in short, open bore MRI scanner.

b) The designed connected coil loosened the wire layout compared to the corresponding layered coil pattern.

c) The designed connected coil has better electromagnetic performance than the corresponding layered coil.

d) The designed connected coil has smaller vibration amplitude and lower SPL than the corresponding layered coil at most frequencies.