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Low eddy current RF shielding enclosure designs for 3T MR applications

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Abstract

Purpose—MR-compatible medical devices operate within the MR environment while benefitting from the superior anatomic information of MRI. Avoiding electromagnetic interference between such instrumentation and the MR system is crucial. In this work, various shielding configurations for positron emission tomography (PET) detectors were studied and analyzed regarding RF shielding effectiveness and gradient-induced eddy current performances. However, the results of this work apply to shielding considerations for any MR-compatible devices.

Methods—Six shielding enclosure configurations with various thicknesses, patterns and materials were designed: solid and segmented copper, phosphor bronze mesh (PBM), and carbon fiber composite (CFC). A series of tests were performed on RF shielding effectiveness and the gradient-induced eddy current.

Results—For the shielding effectiveness, the solid copper with various thickness and PBM configurations yield significantly better shielding effectiveness (>15 dB) compared to CFC and segmented configurations. For the gradient-induced eddy current performance, the solid copper shielding configurations with different thicknesses showed significantly worse results, up to a factor of 3.89 dB, compared to the segmented copper, PBM and the CFC configurations.

Conclusion—We have evaluated the RF shielding effectiveness and the gradient-induced eddy current artifacts of several shielding designs and only the PBM showed positive outcomes for both aspects.

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Keywords

Faraday shielding enclosure; MR-compatible; gradient-induced eddy current; RF shielding effectiveness; PET/MR; PET detector

INTRODUCTION

MR systems provide exquisite high-resolution anatomic information ^{1–5}. To benefit from the anatomical information of MR images, PET ^{6–9} or single-photon emission computed tomography (SPECT) ^{10,11}, which provide sensitive visualization and quantification of bio-molecular pathways and signatures of disease in living subjects are combined with MRI for simultaneous hybrid imaging ^{6,7}. In addition, there are a range of other medical devices that have been brought into an MR system ^{12–16}.

Systems that simultaneously operate within an MRI system require special design considerations to avoid the electromagnetic (EM) interference between the systems ^{17–24}. MR images are acquired by fast-oscillating EM fields in a strong static magnetic field environment. To avoid EM interference, the inserted systems typically encapsulate electronic components inside a shielding Faraday enclosure to preserve the performances of both MRI and the inserted device ^{22–29}. Without sufficient shielding designs, intense radiofrequency (RF) fields in the MHz range could disturb the device electronics or, reciprocally, the noise generated from the device could corrupt the MRI image ^{21,22,26,30}.

The main function of a shielding enclosure is to shield the RF field, while also minimizing any secondary artifact, such as the eddy current induced from the MR gradient fields. The primary mechanisms of EM interference (EMI) shielding are reflection and absorption ^{30,31}. For reflection of the incident fields, the shield must have mobile charge carriers, which interact with the fields. Electrically conducting shields are preferred but not necessarily required. In addition, connectivity in the current path enhances the shielding effectiveness. The absorption loss is described by a function known as $\sigma_r \mu_r$, where σ_r is the relative electrical conductivity and μ_r is the relative permeability. With a few assumptions, including high conductivity ($\sigma \gg \omega \epsilon$), thick shield material ($t \gg \delta$), and the intrinsic impedance mismatch ($\eta_s \ll \eta_0$), the reflection loss is independent of the shield thickness, while the absorption loss is proportional to the thickness of the shield. The shielding “effectiveness” is equal to the sum of both loss mechanisms, and may be expressed in terms of material parameters as follows:

$$\text{Shielding Effectiveness (dB)} = \text{Reflection loss} + \text{Absorption loss} = 20 \log \left(\frac{\eta_0}{4\eta_s} \right) + 20 \log(e^{t/\delta})$$

Eq. 1

where σ is the conductivity of the shield, ω is the angular frequency of the RF field, ϵ is the relative permittivity of the shield, η_0 is the intrinsic impedance of air, η_s is the intrinsic

impedance of the shield, t is the thickness of the shield, and δ is the skin depth of the shield. From this expression it can be seen that metals with high conductivity are by far the most common materials for EM shielding. However, a changing magnetic flux from the time-varying gradient magnetic field induces eddy currents on any conductive surfaces based on Faraday's law of induction, which in turn may cause resistive heating or ghosting artifacts in MRI images^{32–36}. The power dissipation of eddy currents can be calculated using the following equation³⁷,

$$P = \frac{\pi^2 B_p^2 d^2 f^2}{6k\rho D} \quad \text{Eq.2}$$

where P is the power lost per unit mass (W/kg), B_p is the peak magnetic field (T), d is the thickness of the sheet (m), f is the frequency (Hz), k is a constant equal to 1 for a thin sheet and 2 for a thin wire, ρ is the resistivity of the material ($\Omega\cdot\text{m}$), and D is the density of the material (kg/m^3).

In turn, the eddy current generates magnetic fields that oppose the primary gradient fields due to Lenz's law^{38–40}; hence the imaging objects do not experience the gradient field magnitudes that were programmed. Accordingly, this results in a distorted MR image during gradient-intense sequences such as echo planar imaging (EPI). Eddy currents can generate resistive heat and vibration as well.

There have been several approaches to reduce the gradient-induced eddy currents while preserving the RF shielding effectiveness inside the MRI; one method is to reduce the shielding thickness, which directly reduces the induced surface current density. For good conductors ($\sigma \gg \omega\epsilon$) with thick shield material ($t \gg \delta$), the shielding effectiveness is dominantly dependent on reflection loss and slightly affected by thickness^{41–44}. Another way is to disturb the current loop path by segmenting or meshing the conductive surface so that only limited current loops are formed; the RF shielding effectiveness is still preserved by keeping the electrical connectivity throughout the paths^{45–48}. One example of the segmented design application is the RF shield placed between the RF coil and the gradient coil of commercial MR systems^{45,48–50}. Similarly, the wire mesh limits the induced eddy current loop based on the same principle; metal wire mesh has shown to be a good shield^{50,51}. Alternately, a lower conductivity shield material can be employed to reduce current induction. Although, metal is preferred for RF shields, carbon fiber composites (CFC) have been applied inside MRI systems for RF shielding and shown to reduce eddy current induction^{30,52}; in addition to the EM properties, its low density, high strength, high oxidation resistance and thermal stability are attractive as well.

In this paper we have designed various shielding enclosure configurations for detector modules in a PET insert for simultaneous PET/MR^{53–53} and evaluated the RF shielding effectiveness and gradient-induced eddy current performances of these enclosures in several direct and indirect measurements with the goal to understand their potential effects on MR performance. To date there has not been any comprehensive work on shielding designs to be

applied for PET detector modules inside an MR bore. However, the results of this work apply to shielding considerations for any MR-compatible devices.

METHODS

We have studied six shielding enclosures with various materials, thickness and patterns. The list of shielding enclosure properties is shown in Fig. 1.

All shielding enclosures under study had the same dimensions for consistency. Each cage was a trapezoidal cage (bases: 6 and 8 cm, height: 4 cm, length: 20cm) (Fig. 1, bottom); six side plates were built and then assembled into an enclosure with electrical contacts and mechanical support to form a full Faraday cage. For the bare copper designs (Fig. 1(A, B and C)), different thicknesses (36 μm , 18 μm , 9 μm) of copper were printed on a 31 mil (0.79 mm) FR4 sheets. Then the five plates, excluding the top plate of the enclosure, were soldered together at the inner perimeter edge (the dotted line in Fig. 1, bottom), and electrically connected to the outer copper plane through plated vias. The top plate was mechanically held in place with brass screws and electrically connected with copper tapes (0.25" width, 0.0014" thick foil with 0.0021" conductive acrylic adhesive) on the outer top edges. For the segmented copper (Fig. 1(D)), the segment pattern was based on the pattern from a double-sided RF shield⁵⁰. The 20 mil (0.51 mm) width slit was patterned on a copper plate with the maximum length of 20 cm and the smallest slit-to-slit pitch of 0.5 cm. The two segmented patterns were separated by a 31 mil (0.79 mm) FR4 sheet. The assembly method was the same as above. Then, CFC sheet (DragonPlate, NY, USA), with orthotropic laminates utilizing a twill weave at 0°/90° orientation, was computerized numerical control (CNC) machined to the reference shielding enclosure dimensions, mechanically assembled through brass screws, and electrically connected to the plates using copper tape on the outside edge between the plates (Fig. 1(E)). The copper tape was applied in short pieces to avoid creating a large copper loop for inducing eddy current. For the phosphor bronze mesh (PBM) shielding enclosure (Fig. 1(F)), the mesh sheet (36 μm wire diameter, 325 wires/in.) was laminated on to FR4 plates using epoxy with minimal heat and pressure. Then the assembly and electrical connection was achieved the same as for the CFC shielding enclosure.

To test the shielding effectiveness and the gradient-induced eddy current artifact of various shielding designs, bench tests and MRI experiments were performed.

For all of the bench (outside MRI) tests, a network analyzer (HP 4195A, USA) (Fig. 2(A)) along with custom-built instruments (Figs. 2(C, D)) were used. Then, a 3T MRI (GE MR750, GE, USA) system equipped with the built-in body coil consisting of a liquid-cooled 35 kW solid-state RF power amplifier and a gradient system driven by an amplitude system (amplitude 50 mT/m, slew rate 200 T/m/s) were used for assessing the performance in MRI (Fig 2(B)) along with custom-built instruments (Figs. 2(E)–(G)). Each shielding enclosure was individually placed radially 16 cm away from the MRI isocenter to mimic its realistic position for a 32cm diameter head PET inserted placed inside the MRI bore.

RF shielding effectiveness

For testing the shielding effectiveness at 127.76 MHz on the bench, two identical magnetic field loops (Fig. 2(C)) were built; one of the loops was placed inside the shielding enclosure and the other loop was placed outside the enclosure. Each loop was a shielded magnetic field loop, which was only sensitive to RF magnetic fields and insensitive to electric fields. The coil pair was used with a “S21 measurement” (transmission loss)⁵⁵ calibrated at 127.76 MHz with bandwidth of 20 MHz with no shield. Each shield is then placed between the 2 coils.

For the MR experiment, two experiments were performed. A 127.68 MHz RF field source was built using a miniature 42.56 MHz clock chip (FOX electronics, FL, USA) powered by a non-magnetic battery (Powerstream, UT, USA). The 42.56 MHz square wave was driven into a series of resonant LC circuits tuned to the 3rd harmonic, which was 127.68 MHz. This RF source was then measured with the 3T MRI system’s body coil receiver tuned with a center frequency at 127.7 MHz with a bandwidth of ± 31.25 kHz; the experiment began with the 127.68 MHz source unshielded as a reference and then again inside each of the shielded enclosure configurations (Fig. 2(E)). When placing the RF source inside the shielding enclosure, to avoid a frequency peak shift, the source was not placed too close to the shielding planes inside the enclosure. Then, a single magnetic field detection loop (Fig. 2(F)), which has the same design as in Fig 2(C), was placed inside each shielding enclosure and was connected to an oscilloscope for detecting the induced voltage from the RF-intense MRI fast spin echo (FSE) pulse inside the enclosure (Fig 2(D)).

Gradient-induced eddy current investigation

At the bench, the standard eddy current evaluation tests⁵⁶ were performed on different shielding enclosure configurations. By using a high inductance magnetic surface probe (25mH) and an impedance analyzer (Fig. 2(D)), we can perform a S11 measurement⁵⁵ to analyze the magnitude of eddy currents induced at different frequencies. To test a range of frequencies appropriate for the gradients used in an actual MRI study, the S11 measurement was acquired at a range of frequencies from 0.5 kHz to 10 kHz in steps of 500 Hz. Then, the resolution bandwidth and the source amplitude were set to 100Hz and 15 dBm, respectively, to increase the receive sensitivity and lower the noise. Three sets of S11 data were measured and the mean and standard deviation (as error bars) were plotted respect to the frequency. Shields with less eddy current induce smaller change in coil impedance. This change indirectly indicates the magnitude of the opposing magnetic field due to the induced eddy current.

In the MRI system, two experiments were performed. To test the gradient induced eddy current performance, the gradient-intense EPI sequence (Table 1) images of a 17.5 cm diameter spherical agar phantom were acquired. To assess the eddy current induced ghosting artifact from the EPI MR images, 2.5 cm² region-of-interest (ROI) in ghosting and background noise and 15 cm² ROI in signal were measured. Then, the ghost level intensity (GLI) in the images was calculated as following⁵⁷:

$$\text{Ghosting Level Intensity (\%)} = \left(\frac{S_{ghost} - S_{noise}}{S_{signal}} \right) \times 100$$

Then, the temperature of the shielding enclosure was steadily measured for 8 minutes using a fluoroptic thermometer (Luxtron 790, LumaSense, CA) (Fig. 2(G)) with the EPI sequence running (Table 1). Four temperature sensors of the fluoroptic thermometer were placed on four side plates of each of the shielding enclosures and the average of the four temperatures were analyzed.

RESULTS

RF shielding effectiveness

The S21 measurements at the bench are shown in Fig. 3. As expected, the solid copper shielding enclosures with 36, 18 and 9 μm thickness showed the best shielding effectiveness of -43.10 ± 1.72 , -42.47 ± 3.36 , and -42.27 ± 2.62 dB, respectively. The thickness of solid copper did not affect the shielding effectiveness due to the skin depth effect⁴⁴. PBM also showed a comparable shielding effectiveness performance of -37.55 ± 3.55 dB. However, the CFC result of -29.15 ± 1.51 dB was not as effective and the segmented copper result of -10.98 ± 0.66 dB showed the worst shielding performance.

The field attenuation results of the RF pulse voltage readout at 90° and 180° flip angle MR FSE sequence are shown in circles and triangles, respectively (Fig. 3). For both bench and MRI results, the 'No shield' case was set as the reference and the relative attenuation levels of bench and MRI results were closely matched.

The RF signal from the 127.7 MHz source inside the various enclosure was detected by the body coil (Fig. 4). When no shield was present, the raw peak value was the maximum as expected. For solid copper shielding enclosures, regardless of the copper thicknesses (36, 18 and 9 μm), almost no RF signals (-26.11 ± 1.18 , -26.02 ± 1.17 , and -26.18 ± 1.22 dB, respectively) were detected by the body coil. However, when a segmented copper or a CFC was present, some RF fields leaked (-10.89 ± 1.91 and -12.30 ± 2.02 dB, respectively) from inside the shielding enclosures and were detected by the body coil, indicated by the peaks at the 127.7 MHz. The PBM showed -26.08 dB shielding effectiveness, which was comparable to that of the solid copper enclosures at the Larmor frequency.

Gradient-induced eddy-current artifact investigation

The S11 attenuation magnitude result was significantly higher with the solid copper configurations compared to those of the other configurations (Fig. 5), which were close to the values observed from the "No shield" configuration. As the frequency decreased, the S11 magnitude decreased as well. The results below 1.5 kHz were noisy.

The ghosting level intensities of EPI images were acquired for various shielding enclosures (Fig 6). As shown in the plot, EPI images with solid copper enclosures showed high ghosting artifact intensity, and the intensity is higher for thicker copper, while other

enclosures showed negligible artifact. The worst ghosting artifact was shown when imaging the axial slice with frequency encode direction of left-right and phase encode direction of anterior-posterior. This is the case where the maximum gradient field is applied on the largest conductive area (top and bottom planes of the shielding enclosures, see Fig. 1). In other cases (sagittal and coronal orientations) where the gradient field was applied to the side plates, the ghosting artifact was not as high as that of the axial ghosting artifacts.

Temperature rose when the EPI sequence was running due to the Joule heating from the induced eddy current (Fig. 7). The temperature drift with the 36 μm solid copper resulted in the highest temperature rise rate of 0.29 $^{\circ}\text{C}/\text{min}$, and 18 and 9 μm solid copper enclosures showed minor temperature rise of 0.08 and 0.09 $^{\circ}\text{C}/\text{min}$ rate, respectively. Other shielding enclosures showed negligible temperature drift.

From the MRI experiments, the segmented copper, CFC and PBM configurations showed negligible eddy current artifacts.

DISCUSSION

The aim of this work was to evaluate the shielding effectiveness and eddy current performance of several shielding configurations inside the MRI. Several shielding designs with various thicknesses, patterns and materials were developed and studies were performed outside and inside the MRI.

Before performing experiments, simple theoretical calculations with ideal assumptions were performed using equations 1 and 2 in order to understand the eddy current induction mechanisms in the shielding. However, as these equations are only valid with these ideal assumptions, the theoretical calculation and experimental measurements can have mismatches. For example, the multilayer design of the PCB with conductors and inductors and the different shielding patterns including the segment, mesh and weave are difficult to model in the theoretical calculation, but these can affect the shielding and eddy current induction performance^{43,51,58}.

The shielding thicknesses (36, 18 and 9 μm) for solid copper designs were thicker than the skin depth (5.8 μm) of copper at 127.76 MHz. Thus, all three were sufficient for shielding the RF field at 127.76 MHz (Figs. 3 and 4). Despite the slightly higher absorption in the thicker copper; the shielding effectiveness is mostly dominated by reflection rather than absorption, so the three RF shielding effectiveness values were not significantly different.

On the other hand, as the shielding thickness and the EM frequency are proportional to the eddy current power loss (Eq. 2), the ghosting artifact and the temperature increase during the gradient-intense EPI sequence were significantly higher for the thicker solid copper compared to the thinner copper (Figs. 6 and 7). On the bench using a network analyzer (Fig. 5), the eddy current experiment was performed in a range of frequencies from 0.5 kHz to 10 kHz expected for actual gradients in an MR study. The eddy current performance was worse (higher S11 magnitude) at higher frequencies due to increased eddy current induction with high EM field frequencies. Below 1.5 kHz, the data was noisy and S11 measurement was at the noise level. This is due to the limitation of the network analyzer with narrow bandwidth

and also the result of low eddy current induction at low frequencies. However, most importantly, the segmented copper, CFC and PBM configurations showed significantly lower S11 magnitudes compared to the solid copper configurations.

Therefore, reducing the shielding thickness diminishes the induced eddy current; however, this does not completely remove the artifact, as we have observed slight ghosting artifacts (Fig. 6) and temperature increment (Fig. 7) for thin solid copper enclosures. Instead, as mentioned earlier, there are other methods to remove the induced eddy current. By adding a precise design of segmented or meshed patterns, the current loop area can be reduced while conserving the electrical connectivity for sufficient shielding. The RF shield between the RF coil and the gradient coil inside a typical MRI system is an example of this. Although, the custom-built segmented copper was motivated by these considerations, the shielding effectiveness result was poor in our experiment (Figs. 3 and 4). While the segmentation definitely mitigates the induced eddy current issues, depending on the segmentation pattern, undesirable resonances or RF leak into the slots may result. If the slots in the segment pattern are long, they behave like a “slot aperture”, which affects the shielding effectiveness of the enclosure. The longest slot aperture in our design was ~20 cm, which is ~1/12 of the wavelength. To improve the shielding performance, the longest aperture length should be reduced down to at least 1/20 of the wavelength (~12 cm) for sufficient RF shielding efficiency^{58,59}.

For an alternative method, CFC was tested as its continuous fibers have been found to be an excellent EM shielding material due to the low surface impedance and high reflectivity in the frequency range from 0.3 MHz to 1.5 GHz³⁰. There also has been successful attempts in other groups of using the CFC as a shielding material for their PET Inserts⁵². Conversely, the result in our experiment showed poor RF shielding performance, which could be due to the structural parameters of the design, such as, the carbon fiber orientation of individual layers, fiber distribution and the number of layers, which can affect the shielding and eddy current performance⁶⁰. The current CFC prototype comprises orthotropic laminates utilizing a 0°/90° twill weave; the shielding performance could be enhanced by a quasi-isotropic layup having a -45°/45° weave laminations⁶¹.

Even though significant ghosting artifacts and temperature rise were observed for the solid copper shielding enclosures, one thing to note here is that these are only caused by the gradient-intense high-duty-cycle EPI sequences, but not in other sequences, such as, gradient echo or FSE sequences.

The bench and MRI results of shielding effectiveness were comparable (Fig. 3); nonetheless, the slight mismatches could be due to the dynamic range difference between the network analyzer and the MRI RF receiver; the MRI body coil has a much higher dynamic range compared to that of the network analyzer.

To fully satisfy the MR-compatible requirements of MR-inserted medical devices, all of the electronic circuit parts of the device are required to be carefully examined for MR susceptibility. As the parts inside the enclosure satisfy the MR susceptibility requirements,

the appropriate shielding configuration will allow mutual EM interference-free operation inside the MRI.

For potential applications, if more than one shielding enclosure is placed inside the bore, the eddy current performance would proportionally depend on the amount of the integrated amount of conductive materials inside the bore, and also on the orientation of the shielding enclosures.

As the number of medical devices that are inserted into MR systems increases, considerations of proper shielding is a critical factor for simultaneous measurements with an MR system with low mutual interference. We believe some of the studies in this paper will be highly relevant to these goals.

CONCLUSION

The RF shielding effectiveness and the gradient-induced eddy current performance of several shielding configurations inside a 3T MR system were evaluated. Several shielding designs with various thicknesses, patterns and materials were developed and tested inside and outside the MRI. Solid copper shielding enclosures and PBM showed outstanding RF shielding performance, while significant RF leaks were observed in the segmented copper and the CFC configurations. On the other hand, the solid copper shielding designs showed poor eddy current performance and the others showed negligible eddy current artifacts. Therefore, PBM, which shields the RF field effectively while blocking the eddy current induced by the gradient fields, showed the best overall performance. Although we plan to incorporate these findings into enclosure design for a RF-penetrable PET insert system for simultaneous PET/MR^{53–53}, these results can be applied to any other electronic devices inserted into an MR system.

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	(A)	(B)	(C)	(D)	(E)	(F)

Material	Solid copper			Segmented copper	Carbon fiber composite	Phosphor bronze mesh
Thickness	36 μm	18 μm	9 μm	36 μm	0.8mm	36 μm wire diameter, 325 wires/in.
Dimension

Figure 1. Overview of the custom-built shielding enclosure configuration under study. Fig 1a, 1b and 1c are Solid copper shielding cages with thicknesses of 36, 18 and 9 μm. Fig 1d, 1e and 1f are 36 μm segmented copper, 0.8 mm carbon fiber composite and 36 μm phosphor bronze mesh shielding cages, respectively.

Figure 2.

Experimental materials for bench tests (Fig. 2a: network analyzer, 2c: magnetic field loops for shielding effectiveness measurements, and 2d: magnetic surface probe for eddy current measurement) and MRI experiments (Fig. 2b: 3T MRI, 2e: 127.68-MHz RF source, 2f: magnetic field detection loop, 2g: fluoroptic thermometer) of the different shielding configurations.

Figure 3.

S21 measurement at 127.8 MHz at the bench and FSE pulse detection measurement inside the MRI system

Figure 4.
Shielding measurements using the 127.7 MHz RF source

Figure 5.
S11 measurements of the shields from 0.5 kHz to 10 kHz.

Figure 6. (Left) Schematics of the shielding enclosure at different orientations with phase encoding and frequency encoding directions. (Right) Ghosting level intensity (%) on EPI images for axial, sagittal and coronal orientations.

Figure 7.

Temperature measurement of the shields while running the EPI pulse sequence.

Table 1

MR pulse sequence parameters

Sequence	TR (ms)	TE (ms)	Voxel (mm)	Matrix	Flip angle (°)	Number of average (NEX)	Echo train length (ETL)	Pixel bandwidth (Hz/Pixel)
FSE	3000	71	$0.86 \times 0.86 \times 4.00$	256×256	90	1	1	122.1
EPI	2000	30	$3.44 \times 3.44 \times 4.00$	64×64	90	1	1	7812.5