Magnetic Sensitivity of MRI Systems to External Iron:
The Design Process

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Magnetic Sensitivity of MRI Systems to External Iron: The Design Process

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Magnetic Fields in Resonance Systems

High-performance Magnetic Resonance Imaging (MRI), Nuclear Magnetic Resonance (NMR), and Magnetic Resonance Spectroscopy (MRS) systems are extremely sensitive to any inhomogeneity in the magnetic field within a sample volume. One of the most common sources of such deleterious field perturbations is the magnetizable iron and steel in vehicles such as automobiles, trucks, buses, trolleys, subway trains, and ambulances. Additional sources of transient error fields include hospital carts and gurneys, portable X-ray equipment, and passenger and freight elevators, etc.

Computer software that employs the Boundary Element Method (BEM) is the fastest, most accurate, and easiest-to-use Computer-Aided Engineering (CAE) method, not merely to calculate the pertubrbing effects of such magnetized objects, but also to help design, analyze, and optimize passive and/or active magnetic shielding for the MRI system. The two competing techniques—Finite Element Method (FEM) and Finite Difference (FD) Method—both require the analyst to subdivide or "discretize" all regions that contain magnetic fields.

For the perturbed MRI system that is described in this article, this field volume includes not merely the huge volume of empty space that lies between the magnet and the magnetizable iron or steel, but also a substantial volume of space located beyond the steel.

To model such an MRI system using either FEM or FD software, one must either use an enormous number of finite elements, which increases the solution time interminably, or try a smaller number of larger finite elements, which often yields an unacceptable level of accuracy. Note that as the separation between the MRI magnet and the ferromagnetic object increases, the relative accuracy of any FEM or FD analysis degrades rapidly. In contrast, under these same geometric conditions, the accuracy of any BEM solution improves.

Magnetic Sensitivity of MRI Systems

Nuclear Magnetic Resonance (NMR) imaging—now renamed Magnetic Resonance Imaging (MRI)—and its allied procedure, Magnetic Resonance Spectroscopy (MRS), are both extremely sensitive to the inhomogeneity of the magnetic field. The better systems on the market today image at a field homogeneity of 10 parts per million (ppm), peak-to-peak, or better, in a 50-cm Diameter Spherical Volume (DSV). MRS is even more demanding, requiring a field homogeneity of 0.1 to 1.0 ppm or better, although over a considerably smaller volume.

Note that the magnetic-field homogeneity of 1 to 10 ppm generated by a typical MRI system that operates at a central field of 1.5 teslas (15,000 gauss) equals a field inhomogeneity of only 15 to 150 milligauss. This field increment is much smaller than the Earth's magnetic field of ~0.5 gauss! Fortunately, the Earth's field is constant, not only with respect to location within the DVS, but also with respect to imaging time. Thus, the MRI system can easily cancel such an inhomogeneity with its standard magnetic shim set.

As part of the installation process, whenever a new MRI system is installed in a hospital, the shim set is meticulously adjusted to cancel out not only the Earth's magnetic field, but also the ambient magnetic fields generated by any local stationary magnetized objects, such as the steel pilings, structural beams, pipes, HVAC ducts, and steel laminations of motors and transformers in the hospital building. However, the standard MRI shim set is incapable of canceling the transient magnetic fields generated by quasi-static, magnetizable objects, such as moving or parked vehicles and portable hospital equipment.

These quasi-static objects will be magnetized not only by the fringe field of the MRI magnet, but also by the Earth's magnetic field. If the object is large (~18 feet long), and massive (~5000 lbs.), then the deleterious effect on the MRI magnet from such an 18-foot dipole can be significant. Since most MRI systems are sited in the radiology area, they are often located near the emergency rooms (ER), the vehicle ramps to the emergency rooms, the parking lots for the staff, patients and visitors, and the freight or passenger elevators.

These nearby magnetized objects generate error fields that impair the MRI images. These transient, non-shimmable, error fields produce "phase ghosting" due to the misinterpretation of the actual source "voxel" of the stimulated RF signal emitted by the sample, and the consequent erroneous assignment of the corresponding "pixel" during the MRI image reconstruction. To yield MRI images of the highest quality and resolution, it is imperative that the ambient bias magnetic field throughout the DSV remain constant during the entire imaging cycle.

For most of the widely-used pulse sequences, the total imaging time can average twenty to thirty minutes or more. While Echo-Planar Imaging (EPI), Gradient-Recalled Echo (GRE), and "RODEO" imaging require less time, these newer imaging techniques are significantly more sensitive to minor field perturbations. In the absence of time-dependent, adaptive shimming, a nearby car, truck, bus, ambulance, elevator, or portable X-ray machine can easily render an MRI scan worthless.

Figure 1. Side Profile of the Shielded Room with the Magnetizable Steel Bar Located 20 Feet from the Isocenter of the Magnet. This places it outside the wall of the shielded room.

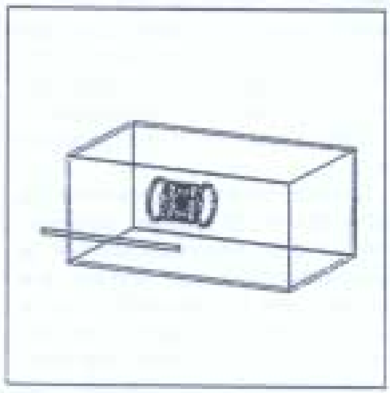

Figure 2. View of the Magnet Coils and the Surrounding Passive, Ferromagnetic Shield.

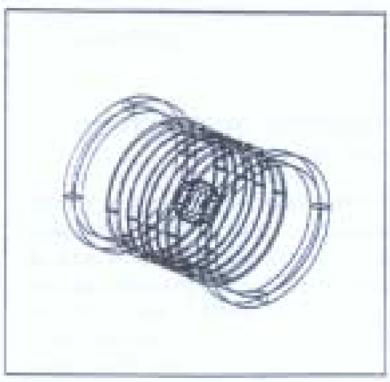

Figure 3. Close-up of the Magnet Coils and the Diameter Spherical Volume.

Error Field Due to Nearby Steel

SYSTEM GEOMETRY

A shielded, 1.5-Tesla, MRI system can easily cost $1,000,000 to $2,000,000 or more. Any "build-and-measure" evaluations of the effectiveness of the magnetic shielding of such a system would obviously be cost-prohibitive. Computer-Aided Engineering (CAE) modeling of the MRI system is the only cost-effective method to design, analyze, and optimize the magnetic shield for an MRI system that will be immune to the magnetic-field inhomogeneities generated by nearby magnetized steel.

This article describes a process using CAE software that employs a sophisticated, commercial BEM software program to model an calculate the magnetic-field inhomogeneity generated by an 18-foot-long bar of steel weighing 5,000 pounds which simulated a full-sized car or small truck. This steel bar was located at various different positions (Figure 1) with respect to the isocenter of an ultra-high homogeneity, 1.5-tesla MRI magnet which was designed expressly for this study (Figures 2 and 3).

This proprietary MRI magnet uses four pairs of superconducting coils. The inside diameter of all eight coils is 40 inches; the maximum length of this magnet is 66.45 inches. The magnet generates a field homogeneity of 0.4664 ppm, peak-to-peak, over a 50-cm DSV. The field profile of the magnet is "16th order;" i.e.,

$B(z) = 1.500\ (1\text{-}2.672(10^{-7})(z\ /z_0)^{16} + \ldots)$

where

$B(z)$ = the magnetic field along the axis of the magnet

z = the distance along the axis of the magnet

z_0 = the radius of the DSV

= 25 centimeters

The test positions of the center of the steel bar are located at the horizontal midplane, 10 feet above the midplane, and 10 feet below this midplane. This set of three positions is repeated 20 feet to the rear, 20 feet to the side, and 27.75 feet to the front of the isocenter of the magnet. As expected, our analysis confirmed that, due to the horizontal mirror symmetry of the geometry, the field homogeneity at any point located below the plane of mirror symmetry is identical to that at the symmetric point located above the plane of mirror symmetry. The first set of CAE test cases analyzes the effects of the magnetized steel bar on an unshielded MRI system. A second set of CAE test cases analyzes the effects of the steel bar on a similar system that includes a passive ferromagnetic shield.

This steel-walled room has a rectangular floor plan, 216 inches wide, 345 inches long, and 162 inches high, and has two planes of mirror symmetry. The cylindrical axis of the MRI magnet coincides with the axis of symmetry of the shielded room. The isocenter of the magnet is located 126 inches in front of the rear wall of the room. The rear wall is 3 inches thick; the front wall is 1 inch thick. The remaining four walls of the room are each 2inches thick. For quicker and easier modeling during the preliminary analysis, this simple shielded room is modeled without doors, windows, or other complexities. Of course, these features would be included in any subsequent detailed analysis. The total mass of steel in this shield is 93.75 tons.

The CAE Analysis Process

The analysis of the effectiveness of passive, ferromagnetic shielding in reducing the perturbing effects of nearby magnetized steel on an MRI system is divided into three sequential stages. First, we use CAE software to calculate the fringe field generated by the unshielded MRI magnet, with no perturbing magnetized steel anywhere in the vicinity of the magnet. Next, we recalculate the fringe field of the same unshielded MRI magnet with the 5000-pound, magnetized steel bar located in one of its nine test positions. Finally, we once again calculate the fringe field generated by the combination of the MRI magnet and the nearby magnetized dipole, but this time with the intervening steel walls of the 93.75-ton, passively-shielded room. As an illustrative example, Figure 4 is a contour plot of the fringe field generated by the passively-shielded MRI system with the magnetizable steel bar located 20 feet from the isocenter of the magnet. This places it outside the wall of the shielded room.

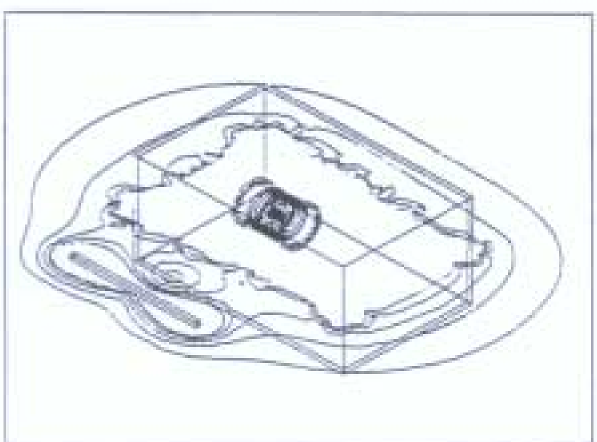

Figure 4. Contour Plot of the Fringe Field Generated by the Passively-shielded MRI System with a Magnetizable Steel Bar Located 20 Feet from the Isocenter of the Magnet. This places it outside the wall of the shielded room.

The BEM solution for all 19 test cases took less than five hours on a basic 200-MHz Pentium with 32-MB of RAM memory. For this study, we assumed that the permeability of the steel was constant, with a relative permeability, μ_r = 4000. However, we could easily have used the tabulated nonlinear B-H curve for any actual steel, with only a modest increase in solution time. Furthermore, if we had used a version of commercial BEM software that is capable of solving time-dependent (eddy-current) problems (such as FARADAY), our analysis could easily have also included any transient effects such as eddy currents due to moving objects, e.g., car, trucks, and trolleys.

Conclusions

CAE software is a powerful, cost-effective tool with which to design, analyze, and optimize magnetic shielding of MRI and NMR systems. Using such CAE software, one can demonstrate that a passive ferromagnetic shield greatly reduces the perturbing effects of external magnetizable iron and steel both outside of the passive shield and inside the magnet DSV.

This magnetic shielding yields two major benefits: First, shielding protects the environment from the magnet. Thus, nearby sensitive equipment such as cardiac pacemakers, surgical implants, electronic equipment, wristwatches, and credit cards will function properly and safely. Second, shielding also protects the MRI system (specifically, its field homogeneity) from the environment. As imaging techniques become ever more sophisticated and imaging speed, resolution, and/or discrimination is improved, magnetic shielding is likely to become progressively more crucial. For example, MRS and high-speed, gradient-recalled echo-imaging (without the advantage of 180-degree time-reversal pulses) both require such superb field homogeneity that magnetic shielding is almost imperative. Very large vehicles—such as a light-rail transit system, or perhaps a major subway system—may even generate such large field perturbations that they require a combination of both active shimming and passive shielding.

Of these three competing methods for the analysis for electromagnetics (Finite Element Method, Finite Difference.

Method, and Boundary Element Method), the BEM is ideally suited to design, analyze, and optimize any magnetic system in which magnetic field fills substantial volumes of otherwise inactive space.

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