Shimming with permanent magnets for the x-ray detector in a hybrid x-ray/MR system

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In this x-ray/MR hybrid system an x-ray flat panel detector is placed under the patient cradle, close to the MR volume of interest (VOI), where the magnetic field strength is ~ 0.5 T. Immersed in this strong field, several electronic components inside the detector become magnetized and create an additional magnetic field that is superimposed on the original field of the MR scanner. Even after linear shimming, the field homogeneity of the MR scanner remains disrupted by the detector. The authors characterize the field due to the detector with the field of two magnetic dipoles and further show that two sets of permanent magnets (NdFeB) can withstand the main magnetic field and compensate for the nonlinear components of the additional field. The ideal number of magnets and their locations are calculated based on a field map measured with the detector in place. Experimental results demonstrate great promise for this technique, which may be useful in many settings where devices with magnetic components need to be placed inside or close to an MR scanner. © 2008 American Association of Physicists in Medicine. [DOI: [10.1118/1.2963994](https://doi.org/10.1118/1.2963994)]

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I. INTRODUCTION

Our hybrid system provides the synergic image guidance of two powerful imaging modalities, x-ray fluoroscopy and magnetic resonance imaging (MRI).¹⁻³ Clinical trials have shown that the hybrid x-ray/MR system can greatly benefit a number of interventional procedures.^{4,5} In our system, an x-ray tube and a flat panel detector were placed inside the magnetic field of an open-bore MR system. Effects of the magnetic field on the x-ray tube have been studied and possible solutions proposed.⁶⁻⁸ The x-ray detector, being close to the MR imaging volume under the patient cradle, experiences a magnetic field approximately equal to the system field strength of 0.5 T. The effect of the magnetic field on the performance of the x-ray detector was evaluated and no appreciable image degradation was observed.⁹ However, the effect of the detector on the MR system has not heretofore been fully explored. The detector can affect the MR system if it has magnetic components. When placed in the main magnetic field, these components can be magnetized and degrade the homogeneity of the magnetic field of the MR scanner. In order to make the detector MR compatible, we have tried our best to replace magnetic components with nonmagnetic substitutes. Nonetheless, it is difficult to make the detector completely nonmagnetic and thus the residual field from the detector can still degrade the field homogeneity of the MR system.

Our MR scanner is able to produce a linear “shim” field to make the main field more homogenous, i.e., a gradient field to compensate for the linear spatial components of the inho-

mogeneous field in the imaging volume. Due to the proximity of the detector to the MR imaging volume, however, the detector produces a field that contains considerable nonlinear spatial components. As a result, even after linear shimming the MR image quality with the detector in position still suffers, especially with the steady-state free precession (SSFP) imaging that is frequently used in our interventional procedures.^{10,11} Our system does not have resistive higher order shim coils^{12,13} and the superconductive higher-order shims cannot be easily adjusted. Even if high-order shimming coils could be used, they are designed to produce fields with spatial variation comparable to the bore size and may not be an optimal choice for the effects of the x-ray detector. Passive shimming can also provide nonlinear field components to improve the system field homogeneity by strategically placing pieces of ferromagnetic material in the magnet.^{14,15} The sizes and locations of the ferromagnetic material need to be carefully calibrated. Since our x-ray detector is not permanently mounted in the MR system, it would be difficult to remove and replace the passive shimming according to the presence of the detector. Therefore, a more convenient and efficient method is very desirable.

Local shimming could be implemented with electric coils that can be tailored to specific situations. This active shimming scheme is especially well-suited in situations where the field inhomogeneity is slight and the imaging volume is localized, e.g., in fMRI studies.¹⁶ For interventional procedures, on the other hand, the imaging volume can be large and the deleterious field created by the magnetic material in

FIG. 1. Flowchart of the optimization process for shim magnets: B_{base} is the baseline field, $B_{base+det}$ is the field measured with detector in place. $B_{base+det}$ is fitted with field of dipoles, and the fitting result $dipoles_{tar}$ is the targeted dipoles for shim magnets. $B_{base+mags}$ is the field measured with shim magnets, and $dipoles_{mags}$ is the dipole fitting result of the net field produced by shim magnets $B_{base+mags} - B_{base}$. The final shimming efficacy is tested by measuring the field with detector and shim magnets $B_{base+det+mags}$.

the x-ray detector can be strong. Preliminary studies showed that for our applications the shimming coils would require very high power to be effective. Further, problems such as heat dissipation could also arise.

Based on the origin of the detector's field, we hypothesized that shimming with permanent magnets would be able to counteract the detector induced field. If the magnetic components inside the detector are distributed in small clusters, the field from each cluster can be well approximated by the field of a magnetic dipole near the cluster pointing in the direction of the external field. In reality, the dipole field can be approximated with the field of a small permanent magnet. If the intrinsic coercivity of the magnet is sufficiently large, reversing the poles of the magnet in space will not change its magnetization or its magnetic field.¹⁷ As a result, the undesired magnetic field from the detector could be largely canceled by suitably selected and placed permanent magnets oriented in the direction opposite to the external magnetic field.

II. METHODS

The process to optimize the locations and strengths of the shim magnets is illustrated in the flowchart shown in Fig. 1. We first measured the baseline field (B_{base}) without the detector, and then we measured the field with the detector in place ($B_{base+det}$), which is the sum of the baseline field and the net field due to the detector (B_{det}). Considering the linear shimming of the scanner, we focused on the higher-order terms by numerically removing the offset and linear (x, y, z) components of the field. We hypothesized that the magnetic components in the detector can be grouped into a few clusters, and the field produced by each cluster can be approximated with a dipole field. In other words, the field due to the detector B_{det} can be modeled with the field of a few dipoles. As the ultimate goal is to reduce the inhomogeneity of

FIG. 2. Experimental setup for field map measurement: the x-ray detector sits underneath a phantom inside an open-bore MR scanner.

$B_{base+det}$, this measured field was fitted with the field of a number of dipoles. The fitting result $dipoles_{tar}$ was the targeted dipoles and they were used to guide the selection and placement of the shim magnets. The field with the shim magnets ($B_{base+mags}$) was measured, and the net field due to the magnets ($B_{mags} = B_{base+mags} - B_{base}$) was fitted with the field of the same number of dipoles. The fitting result $dipoles_{mags}$ was compared with $dipoles_{tar}$ in order to adjust the locations and quantities of the shim magnets. Iterations of adjustment of the magnets were performed until there was no significant difference in shimming performance between $dipoles_{tar}$ and $dipoles_{mags}$. Finally, the detector with shimming magnets was placed in the MR system, and the MR field homogeneity was evaluated qualitatively with MR images and quantitatively with measurement of $B_{base+det+mags}$.

All MR data were collected on a 0.5 T open magnet system (Signa SP, GE Healthcare, Waukesha, WI) using a transmit-receive birdcage body coil (Fig. 2). The impact of the detector on the field homogeneity was evaluated using a 28 cm diameter cylindrical phantom filled with doped water.

For qualitative assessment of image quality, axial 2D multislice gradient echo images were collected with TR/TE = 250/20 ms, FOV = 34×34 cm², 256×256 matrix, slice thickness = 7 mm, and 50° flip angle. Balanced steady state free precession (SSFP) images were also collected with TR/TE = 11.0/3.9 ms, slice thickness = 8.2 mm, 60° flip angle, and the same FOV and matrix size as the GRE images.

II.A. Field map measurement and data processing

II.A.1. Field map measurement

In order to obtain the field map, interleaved 2D multi-slice gradient echo (GRE) images of the phantom were collected with two consecutive single-echo scans at different echo times ($TE_1 = 4.8$ ms, $TE_2 = 6.8$ ms) for 66 3-mm contiguous axial slices (FOV: 34×34 cm², resolution: 64×64 , TR: 1500 ms). Both magnitude and phase images were reconstructed. The magnitude images were used to select only voxels within the cylindrical phantom for further processing.

FIG. 3. The field of a small magnet can be approximated with the field of a magnetic dipole ($\mathbf{m}$); the main MR field ($\mathbf{B}_0$) is in the z direction.

The phase difference between the two echo times was then calculated for voxels within the cylindrical volume of interest (VOI).

Phase unwrapping was performed starting from the center of the cylindrical VOI and proceeding along the central axis toward both ends in the slice direction (z). Within each slice, phase unwrapping was first performed from the central voxel toward both edges along the horizontal (y) direction and then in the vertical direction (x). At each voxel, a multiple of 2π was added or subtracted to make the difference compared to the previous voxel less than π .

The difference of the two echo times ($\Delta TE=2$ ms) was empirically chosen so that the phase evolution map could be unwrapped and the angular frequency of the spins at each voxel can be calculated. The angular frequency divided by the gyromagnetic ratio was used to generate the field map. Since all the processing was performed on images demodulated by the assumed Larmor frequency, the field map was defined as the field shift from the assumed main magnetic field of the MR system (B_0).

II.A.2. Field map processing

A least-squares (LS) fit of the field map to a three-dimensional linear spatial function (with a constant term) was performed. These components can be corrected using conventional methods, e.g., changing the linear shims and center frequency of the scanner. Thus, they were removed from all the field maps so that the resultant fields only contain the higher-order components for which the system is unable to compensate.

II.B. Fitting with the dipole field

II.B.1. Dipole field calculation

The static field of a small permanent magnet (or a small magnetized volume) at a distance much larger than the size of the magnet can be approximated by the field of a dipole:

$$\mathbf{B} = \frac{\mu_0}{4\pi} \left[\frac{3(\mathbf{m} \cdot \mathbf{r})\mathbf{r}}{r^5} - \frac{\mathbf{m}}{r^3} \right], \quad (1)$$

where $\mathbf{m}$ is the dipole moment, μ_0 is the permeability constant, and $\mathbf{r}$ is the displacement vector from the dipole to the point of field (Fig. 3).¹⁸ The mathematical model behind this approximation is to express the field in the local spherical coordinates (centered at the magnet) with Legendre functions, neglecting the terms of orders higher than the second order. In the presence of the main magnetic field $\mathbf{B}_0$, only the

component of the dipole field in the same direction as $\mathbf{B}_0$ (usually defined as the z direction) affects the resonance frequency. The perpendicular components of the dipole field $\mathbf{B}$ are negligible because $\mathbf{B}$ is orders of magnitude less than $\mathbf{B}_0$. For a dipole aligned with the z direction, the z component of its field is

$$B_z = \frac{\mu_0}{4\pi} \frac{m(3 \cos^2 \theta - 1)}{r^3}, \quad (2)$$

where θ is the angle between $\mathbf{m}$ and $\mathbf{r}$. This expression was used to calculate the dipole field with the constant and linear spatial components numerically removed.

II.B.2. Nonlinear LS fitting

In order to approximate a measured magnetic field with the field of one or more dipoles, the deviation between these two fields needs to be minimized. The selection of the number of dipoles is discussed below. The disagreement between the measured and modeled fields can be characterized by the span (max-min) of the difference field (peak-to-peak deviation), which is appropriate since MR image quality can often be limited by the maximum peak-to-peak deviation.¹⁹ However, fitting trials showed that in our case with a large number of field sampling points ($>10^5$), minimization of the peak-to-peak (p-p) deviation was less effective and less stable than minimization of the root-mean-square (rms) error. Therefore, LS minimization was adopted to find locations and strengths of the dipoles. Nonlinear LS fitting was performed with an optimization program (lsqcurvefit) by MATLAB (MathWorks, Natick, MA). If locations of the dipoles are known, the dipole strengths that minimize the rms error can be computed with linear fitting. Between iterations of nonlinear fitting, linear optimization of the dipole strengths was performed at then estimated positions. This was found to improve convergence. Iterations were terminated when changes in the dipole parameters were insignificant, and the total computation time was on the order of minutes.

As can be appreciated from Eq. (2), at large distances from a dipole, a change in the dipole distance can be mostly compensated by a change in the dipole strength. A more distant dipole of higher strength produces a similar field within the volume of interest as a weaker but closer dipole. Thus, it was possible to constrain the dipoles to be located on a plane below the imaging volume (i.e., dipole locations with only two degrees of freedom), and the residual error after the LS fit was comparable to that of a fit that allowed full freedom in the locations of the dipoles.

II.B.3. Detector field measurement

The field measured with the detector inside the MR scanner ($B_{\text{base+det}}$) had two parts, the intrinsic baseline field of the scanner (B_{base}) and the field perturbation due to the detector (B_{det}). The measured total field ($B_{\text{base+det}}$) was fitted with the field of dipoles since MR image quality is dependent on the total field. Initially the field with the detector was fitted with the field of a single dipole. Results showed that the calcu-

FIG. 4. Two sets of magnets are mounted at the bottom of the detector with two pairs of slotted bars allowing separate 2D location adjustment for each of the two sets of magnets.

lated location of the model dipole was considerably below the bottom of the detector and the predicted residual field was significantly less homogeneous than the baseline field. When two dipoles were used, their locations were close to the detector and the predicted residual field had equal or better homogeneity than the baseline field. Therefore, a model containing two magnetic dipoles was used to fit the field, and two sets of magnets were used to correct the field inhomogeneity.

II.C. Magnet selection, mounting device, and calibration

The requirements for the magnetic material of the shim magnets are high strength and coercivity, which allow the magnets to be small and unaffected by the main MR field. We selected neodymium-iron-boron (NdFeB), a high-powered magnetic material that can withstand an external field of more than 1 T. A number of NdFeB magnets, each approximately $0.8 \times 0.9 \times 1.5 \text{ cm}^3$ in size, were obtained since their small size allowed fine adjustment to their total magnetic dipole strength by changing the number of magnets used.

In order to cancel the field of the magnetized components in the detector, we expect the ideal locations of the shim magnets to be near these magnetic components. Due to space limitation, it is difficult to place the magnets inside the detector. At best, they could be placed under the detector where the magnets and their mounting device would not obstruct the x-ray beam. Importantly, fitting results showed that if two dipoles were on the same horizontal plane, the specific horizontal plane could be placed at a range of heights while maintaining satisfactory shimming effects. The farther the plane was from the detector, the stronger the needed dipoles, while the residual field was not strongly dependent on the location of the plane. A plane close to the detector is preferred so as to reduce the size and strength of the magnets needed.

A device that allowed two sets of magnets to be installed under the detector was custom built (Fig. 4). For MR com-

TABLE I. Results of field measurement and dipole fitting.

Field	Description		Dipole field fitting results (dipole strength @ y, z) unit: (A m ² @ mm, mm) $x=365$ mm for fitting with two dipoles (approximate vertical location for the magnet holders)	Residual field (measured field— dipole field)	
				p-p ($\times 10^{-7}$ T)	rms ($\times 10^{-7}$ T)
Baseline (B_{base})	No detector or magnets		None	53.8	5.51
With detector ($B_{\text{base+det}}$)	Detector in place		None	125.8	16.8
			Single dipole	71.4	6.49
			(A: 2.7 @ -13, -78; B: 1.7 @ -21, 97)	55.4	5.20
Magnets (net field: $B_{\text{mags}} = B_{\text{base+mags}} - B_{\text{base}}$)	Two groups of magnets	Position 1	(A: -2.2 @ -15, -85; B: -1.4 @ -18, 99)	18.1	1.22
		Position 2	(A: -2.2 @ -16, -75; B: -1.4 @ -17, 100)	16.3	1.27
With detector and magnets ($B_{\text{base+det+mags}}$)	Detector in place with magnets in Position 2		None	55.6	5.66
			(A: 0.5 @ -13, -87; B: 0.5 @ -25, 103)	50.5	4.80
	Detector shifted	$\Delta y = 20$ mm	None	54.7	5.51
		$\Delta z = 20$ mm	None	59.5	5.84

FIG. 5. With no detector inside the MR scanner, axial images of the phantom (256×256 , FOV=34 cm) were obtained with (a) GRE (TE/TR=20/250 ms, slice thickness: 7 mm, flip angle: 50°) and (b) SSFP (TE/TR=3.9/11.0 ms, slice thickness: 8.2 mm, flip angle: 60°). (c) Measured field map. (d) Histogram of the field sampled at voxels in the phantom (offset and linear spatial terms removed).

patibility, the mounting device was made of nonmagnetic material. The main body was a plastic plate of the same size as the detector that could be attached to the bottom of the detector with stainless steel screws. A pair of slotted bars made of aluminum was fixed on the vertical edges of the plate, and another pair of bars was placed horizontally over the first pair with screw-stud fasteners. Two magnet holders made of plastic (Delrin, DuPont, Wilmington, DE), each with a cavity capable of holding multiple magnets, were situated on the horizontal bars, held with screw-stud fasteners. The in-plane positions of the magnet holders could be adjusted by sliding the horizontal bars vertically and sliding the holders horizontally.

The strengths of the magnets were calibrated using the MR scanner. Two sets of magnets were placed in the holder without the detector and then into the MR scanner, approximately where the detector would be. The baseline field (B_{base}) and the field with the magnets in place ($B_{\text{base+mags}}$) were measured. The field due to the magnets (B_{mags}) was obtained by subtracting the baseline field from the field measured with the magnets ($B_{\text{mags}} = B_{\text{base+mags}} - B_{\text{base}}$). The resulting field B_{mags} was fitted with the field of two dipoles in order to obtain the dipole strengths and locations that could be used to characterize the field of the magnets.

III. RESULTS

Field measurement and fitting with dipole field were performed in order to optimize the shim magnets. Results are tabulated in Table I and described in detail below.

III.A. Baseline field map (B_{base})

Figure 5 shows MR images of an axial slice near the center of the phantom with GRE and SSFP sequences when the detector was not inside the MR field. With linear shimming, the GRE sequence could produce images with high SNR and no artifacts [Fig. 5(a)]. Slight banding artifacts can be seen in images with the SSFP sequence that is more sensitive to field inhomogeneity [Fig. 5(b)]. The signal nonuniformity at the bottom of the phantom in both images is due to proximity to the RF coil. Figure 5(c) shows a slice of the measured baseline field (B_{base}) and Fig. 5(d) is a histogram of B_{base} at voxels in the cylindrical VOI after removing the offset and linear components. The peak-to-peak (p-p) and rms deviations were 53.8×10^{-7} T and 5.51×10^{-7} T, respectively.

III.B. Field map with the detector ($B_{\text{base+det}}$)

The detector significantly perturbed the main magnetic field and the effects were obvious in the MR images, even after linear shimming. The GRE images contained banding artifacts throughout the FOV [Fig. 6(a)]. The SSFP images had strong banding artifacts typical of the behavior of SSFP with significant field inhomogeneity [Fig. 6(b)]. The measured field map ($B_{\text{base+det}}$) [Fig. 6(c)] within the cylindrical VOI after linear shimming had a p-p deviation of 125.8×10^{-7} T (more than two times of the baseline) and an rms deviation of 16.8×10^{-7} T (about three times of the baseline) [Fig. 6(d)].

FIG. 6. With the detector inside the scanner, images of the phantom were obtained with (a) GRE and (b) SSFP sequences. (c) Measured field map. (d) Histogram of the field sampled in the phantom. Imaging parameters were the same as in Fig. 5.

Fig. 7. With the detector and shimming magnets inside the scanner, images of the phantom were obtained with (a) GRE and (b) SSFP sequences. (c) Measured field map. (d) Histogram of the field sampled in the phantom. Imaging parameters were the same as in Fig. 5.

III.C. Field correction with shim magnets

III.C.1. Dipole fitting of the field with the detector

When the field with the detector in place ($B_{\text{base+det}}$) is fitted to the field of a single dipole (B_{dipole}), the result called for exceedingly large dipole strength (11.5 A m^2) at a location about 10 cm below the detector. Still, the predicted residual field ($B_{\text{base+det}} - B_{\text{dipole}}$) had p-p and rms deviations of $71.4 \times 10^{-7} \text{ T}$, and $6.49 \times 10^{-7} \text{ T}$, which are somewhat worse than the baseline field. This indicated that the field produced by the detector could not be well represented by the field of a single dipole. Consequently, the field of two dipoles was used to fit the measured field. When the locations of the two dipoles were constrained to be on a horizontal plane near the bottom of the detector ($x=365 \text{ mm}$, measured from the center of the VOI), the calculation called for a dipole (A) of 2.7 A m^2 placed at $y=-13 \text{ mm}$, $z=-78 \text{ mm}$ (see Figs. 2 and 4 for the y - z coordinate system) and another dipole (B) of 1.7 A m^2 placed at $y=-21 \text{ mm}$, $z=97 \text{ mm}$. For simplicity, the two dipoles are noted as (A: $2.7 @ -13, -78$; B: $1.7 @ -21, 97$) and the same notation format for dipoles will be used from now on. The predicted residual field (p-p: $55.4 \times 10^{-7} \text{ T}$, rms: $5.20 \times 10^{-7} \text{ T}$) was comparable to the baseline field. The locations of the dipoles corresponded to the locations of components in the detector that are believed to be magnetic.

III.C.2. Shimming magnet deployment

It is desirable to use as few shim magnets as possible for better dipole approximation. Fitting results showed that when the maximum allowed dipole strength was limited to 2.2 A m^2 , satisfactory shimming could still be obtained. For example, with two smaller dipoles (A: $2.2 @ -10, -73$; B: $1.4 @ -20, 95$) the predicted residual field (p-p: $60.0 \times 10^{-7} \text{ T}$, rms: $5.75 \times 10^{-7} \text{ T}$) was close to that with stronger dipoles. Two sets of magnets were selected with their strengths close to 2.2 and 1.4 A m^2 respectively. Then they were placed in the magnet holders at the appropriate locations (Position 1 in Table I) on the shimming plate. In order to cancel the field due to the detector, the direction of magnetization of the magnets was oriented opposite to the MR field. The shimming plate was placed in the MR scanner without the detector, and the field ($B_{\text{base+mags},1}$) was measured. The field due to the magnets ($B_{\text{mags},1}$) was estimated as

the difference field $B_{\text{base+mags},1} - B_{\text{base}}$, which was fitted with the field of two dipoles. Fitting results showed that the two sets of shimming magnets could be represented by two magnetic dipoles (A: $-2.2 @ -15, -85$; B: $-1.4 @ -18, 99$) with small residual field (p-p: $18.1 \times 10^{-7} \text{ T}$, rms: $1.22 \times 10^{-7} \text{ T}$), confirming the similarity between the field of magnets and that of two dipoles. The negative sign of the dipole strength indicates its direction opposite to the external field. By comparing the equivalent dipole locations of the magnets (A: $y=-15, z=-85$; B: $y=-18, z=99$) and the predicted dipole locations of the field measured with the detector (A: $y=-10, z=-73$; B: $y=-20, z=95$), we adjusted the location of the stronger set of magnets (A) by shifting them in the $-z$ direction by 12 mm (Position 2 in Table I). Other location differences were ignored because they were small ($\leq 5 \text{ mm}$) and difficult to achieve at sufficient accuracy with our device. The field with magnets after location adjustment ($B_{\text{base+mags},2}$) was again measured and $B_{\text{base+mags},2} - B_{\text{base}}$ was fitted with field of two dipoles (A: $-2.2 @ -16, -75$; B: $-1.4 @ -17, 100$). The dipole locations showed that the two sets of magnets were well placed at the targeted positions. The field with the detector and shimming magnets could be predicted with all experimentally measured fields as $B_{\text{base+det}} + B_{\text{base+mags},2} - B_{\text{base}}$ (p-p: $59.2 \times 10^{-7} \text{ T}$, rms: $5.58 \times 10^{-7} \text{ T}$).

The shimming plate was then attached to the detector and MR images were acquired to confirm the shimming efficacy. The GRE images [Fig. 7(a)] were visually indistinguishable from the baseline images [Fig. 5(a)]. The SSFP images [Fig. 7(b)] had only two bands at the edge of the phantom. Improved field map can be appreciated in Fig. 7(c). Quantitatively, the field inhomogeneity with the detector and shimming magnets ($B_{\text{base+det+mags}}$) was reduced to $55.6 \times 10^{-7} \text{ T}$ (p-p) and $5.66 \times 10^{-7} \text{ T}$ (rms), less than 4% above the baseline level of $53.8 \times 10^{-7} \text{ T}$ (p-p) and $5.51 \times 10^{-7} \text{ T}$ (rms) [Fig. 7(d)].

III.C.3. Prediction of additional shimming with electric coils

Further improvement of the magnetic field can be achieved by using both permanent magnets and electric shim coils placed around or near the detector. The active shim coils could be used to fine-tune the residual field after shimming with the permanent magnets. Since the permanent mag-

nets are able to handle the major portion of the field inhomogeneity, the power requirements for the active coils should be modest. In order to verify this, the measured field with detector and shimming magnets in place ($B_{\text{base+det+mags}}$) was fitted with two magnetic dipoles. With two small dipoles (A: 0.5 @ -13, -87; B: 0.5 @ -25, 103), the residual field became more homogeneous than the baseline, with p-p deviation at 50.5×10^{-7} T and rms deviation at 4.80×10^{-7} T.

III.C.4. Shift tolerance test of shimmed detector

The x-ray field of view is limited by the finite size of the detector. During a procedure, the physician may want to see anatomy outside the FOV of the detector, requiring either the patient or the detector to be moved. Moving a patient can be difficult and undesirable. Thus, moving the detector may be the better option. When the detector is moved to a new location, it may experience a different polarizing field. As a result, the magnetic components may be magnetized differently, producing a different field. The distance between the detector and the MR imaging volume may change as well, which is another factor that can change the detector-induced field in the MR imaging volume. Since the shimming magnets are optimized for the total field that includes the baseline field and the detector-induced field, a shift in detector location may degrade the shimming effect of the permanent magnets because they move with the detector. If the MR magnetic field at the detector is fairly uniform, this effect should be small. In our case, if the detector was shifted by 2 cm horizontally (z) and perpendicular (y) to the sliding direction of the patient table, little change of the overall field in the MR imaging volume was observed [horizontal: 59.5×10^{-7} T (p-p), 5.84×10^{-7} T (rms); perpendicular: 54.7×10^{-7} T (p-p) 5.51×10^{-7} T (rms), Table I]. Larger detector shifts may require change in number and location of the shimming magnets in order to maintain the required field homogeneity. If the detector is shifted frequently during an interventional procedure, the shimming magnets can be optimized for only the detector-induced field so that the magnets may not need to be repositioned.

IV. DISCUSSION

Using permanent magnets near magnetized components in an external magnetic field can be a simple and efficient shimming method. However, it requires the magnets to be strong enough not to be affected by the external magnetic field. Otherwise, the shim magnets can be weakened or even repolarized. To prevent this from happening, the material of the magnets should be selected so that its intrinsic coercivity is much larger than the intensity of the external magnetic field. For our MR system with a field of 0.5 T, NdFeB appears to be a good choice. However, for 1.5 or 3 T systems, other magnetic materials with higher coercivity may need to be considered, e.g., SmCo (samarium cobalt).

When a magnetic dipole is placed in a strong magnetic field ($\mathbf{B}_0$), it experiences a torque ($\boldsymbol{\tau} = \mathbf{m} \times \mathbf{B}_0$) and force [$\mathbf{F} = \nabla(\mathbf{m} \cdot \mathbf{B}_0)$]. If the dipole is aligned parallel (or antiparallel) to $\mathbf{B}_0$, the net torque on the dipole should be zero, as is the

case for our shim magnets. The force on the dipole depends on the gradient of $\mathbf{B}_0$ at the location of the dipole. Because the x-ray detector (and thus the shim magnets) is centered near the mid-plane between the two MR magnetic windings (Fig. 2), $\mathbf{B}_0$ is sufficiently uniform and therefore the net force on the shim magnets should be negligible. When the magnets (always securely placed inside the holders) were moved in and out of the MR scanner, we tried to move them along the mid-plane while maintaining the antiparallel orientation in order to minimize transient torques and forces.

For each group of the shim magnets, the dipole field approximation is equivalent to expanding the magnetic field of the magnets in terms of Legendre functions and keeping the second-order term in the local coordinate system. Higher-order terms are negligible if the overall size of the group of the magnets is small compared to the distance from the magnets to the imaging volume. This condition may not always be well satisfied, e.g., when a large number of magnets are needed. In these cases a more sophisticated model that includes higher-order Legendre terms should be used to characterize the field of a group of shimming magnets instead of dipole fields (second-order terms).

In field measurement, two GRE scans with different TE were performed sequentially, effectively assuming that the phase images are temporally stable. However, this assumption may not be true due to system instability. Phase shifts due to instability will be interpreted as an artificial additional field. Sequences that interleave the acquisition of images at two echo times within a single scan may be helpful in reducing field errors resulting from the system drift.

V. CONCLUSIONS

When a flat-panel x-ray detector is placed under the patient cradle of an MR scanner, magnetic components inside the detector create an additional and undesired magnetic field that may not be correctable by linear shimming. In order to compensate for the field due to the detector, we used permanent magnets placed below the detector to provide passive shimming. A systematic procedure was developed to optimize the strengths and positions of two sets of high-powered magnets (NdFeB). Experimental results showed great promise for this technique, which can be more generally used to reduce degradation of the MR field from devices that are slightly magnetic but need to be used inside or in the vicinity of the MR scanner.

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