Closed bore XMR (CBXMR) systems for aortic valve replacement: Active magnetic shielding of x-ray tubes

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Hybrid closed bore x-ray/MRI systems are being developed to improve the safety and efficacy of percutaneous aortic valve replacement procedures by harnessing the complementary strengths of the x-ray and MRI modalities in a single interventional suite without requiring patient transfer between two rooms. These systems are composed of an x-ray C-arm in close proximity (≈ 1 m) to an MRI scanner. The MRI magnetic fringe field can cause the electron beam in the x-ray tube to deflect. The deflection causes the x-ray field of view to shift position on the detector receptacle. This could result in unnecessary radiation exposure to the patient and the staff in the cardiac catheterization laboratory. Therefore, the electron beam deflection must be corrected. The authors developed an active magnetic shielding system that can correct for electron beam deflection to within an accuracy of 5% without truncating the field of view or increasing exposure to the patient. This system was able to automatically adjust to different field strengths as the external magnetic field acting on the x-ray tube was changed. Although a small torque was observed on the shielding coils of the active shielding system when they were placed in a magnetic field, this torque will not impact their performance if they are securely mounted on the x-ray tube and the C-arm. The heating of the coils of the shielding system for use in the clinic caused by electric current was found to be slow enough not to require a dedicated cooling system for one percutaneous aortic valve replacement procedure. However, a cooling system will be required if multiple procedures are performed in one session. © 2009 American Association of Physicists in Medicine.
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I. INTRODUCTION

Percutaneous aortic valve replacement^{1–6} (PAVR) is being developed as an effective alternative treatment for aortic stenosis⁷ patients who are denied open-heart surgical replacement of the diseased aortic valve.^{8,9} PAVR involves replacing the diseased aortic valve of the heart with a bioprosthetic valve using a catheter-based approach, which is minimally invasive compared to open-heart surgery.

We are developing a *closed bore* hybrid x-ray/MRI (CBXMR) system to improve the safety and efficacy of PAVR by harnessing the complementary strengths of both modalities.¹⁰ The CBXMR system is composed of an x-ray C-arm placed near the entrance (≈ 1 m) of a 1.5 T MRI scanner. A flat-panel detector and a rotating-anode x-ray tube are mounted on the C-arm.¹¹

Previous work showed that the electron beam in the x-ray tube used to produce x rays¹² at the focal spot on the anode can be deflected^{13,14} by the magnetic fringe field of the MRI scanner.¹⁵ The electron beam deflection causes a shift in the

irradiated field of view.¹⁶ This shift can lead to exposure of patient's anatomy that is not imaged and, in some cases, pose a safety hazard to the staff in the cardiac catheterization laboratory. Therefore, field of view shift must be corrected in the clinic.

In weak magnetic fields (several mT or less), this correction can be accomplished by moving the blades of the x-ray collimator toward the center axis of the x-ray field by a distance greater than or equal to the shift.¹⁶ The drawback to this approach is that field of view shift is corrected by a reduction in the size of the field of view. This is a problem in strong magnetic fields (tens of mT or stronger), which can shift the field of view by several centimeters or more. The field of view size reduction as a result of moving the collimator blades in several centimeters can lead to a loss of important clinical information.

For strong magnetic fields, it is possible to correct for field of view shift without reducing the field of view size. This can be accomplished by using a larger detector and opening up the collimator blades to encompass the relevant

FIG. 1. (a) Front view of MRI scanner entrance with the x-ray C-arm and scanner isocenters aligned. The magnetic fringe field B acts perpendicular to the electron beam and can be corrected by the shielding coils, which provide B_c . (b) C-arm and MR scanner isocenters misaligned. In this case, a field component B_H is present which is not corrected by the coils. (c) Front view of anode showing a change in circumference of the anode focal track when the focal spot is deflected by B_H . The radial deflection of the focal spot in a magnetic field can increase or decrease the track circumference, depending on the orientation of B_H .

anatomy.¹⁶ The drawback of this approach is that anatomy that does not require imaging for the clinical task is exposed to radiation. It is possible to reposition the x-ray tube instead of opening up the collimator blades, but this adds additional mechanical complexity to the C-arm to facilitate repositioning.

Therefore, a technique is required that can correct for field of view shift without truncating the field of view, without repositioning the x-ray tube, or unnecessarily exposing the patient. To accomplish this, we have developed an active magnetic shielding system for the x-ray tube. This shielding system is composed of a pair of electric coils connected to a magnetic field sensing feedback circuit. The shielding system will detect the fringe field applied to the x-ray tube and generate a magnetic counterfield of the same magnitude to cancel it out. This will result in a negligible field acting on the electron beam, so deflection will be corrected. It will be possible to use the active shielding system to correct for all of the deflection in weak magnetic fields (within a specific accuracy) or to use it in combination with the x-ray collimator in strong magnetic fields.

Note that only one pair of shielding coils is required around the x-ray tube in CBXMR systems for PAVR procedures. This coil pair will correct the component of the mag-

netic fringe field B perpendicular to the electron beam along the direction from the output port to the back of the tube [see Fig. 1(a)]. The electric field E from the anode to the cathode of the x-ray tube is shown emerging from the page. The coils will produce a magnetic counterfield B_c to correct the deflection. The parallel component of the fringe field will not introduce any electron beam deflection, so it does not need to be corrected. If the isocenter of the x-ray C-arm is aligned with the isocenter of the MRI scanner, then B can be corrected regardless of the angular orientation of the x-ray tube since the fringe field of the MRI scanner is radially symmetric.¹⁰

Figure 1(b) shows the scenario of misalignment between the isocenters of the x-ray C-arm and the MRI scanner. In this case, an additional component of the fringe field, B_H , will be present, which is also perpendicular to the electron beam. This component is present with oblique angulation of the x-ray C-arm. B_H can cause the electron beam to deflect radially on the anode. While this will not cause a field of view shift, it can change the mean diameter of the anode focal track. This is shown in Fig. 1(c), where the focal spot on the anode is deflected radially toward or away from the center of the anode, changing the track circumference. In the fringe field strengths of the CBXMR system for PAVR pro-

cedures (several mT), the maximum electron beam deflection is about 2 mm.¹⁶ This results in an increase or a decrease (depending on the direction of B_H) in the mean anode track diameter of approximately 4% on a 4 in. anode, which will not adversely impact the clinical performance of the x-ray tube. Isocenter misalignment will be caused by moving the C-arm to facilitate image panning over the patient. Instead of panning with the C-arm, it is possible to fix the x-ray C-arm isocenter such that it is aligned with the MRI scanner isocenter and pan with a moving patient tabletop.

Methods are introduced to characterize the shielding system. A correction technique is introduced and implemented into the shielding system to account for the fact that the magnetic field sensor of the shielding system cannot be directly placed at the electron beam position. This correction will improve magnetic shielding performance.

When the shielding system is operational, electric current flows in the shielding coils and they will experience a torque in the MRI fringe field, which will cause them to twist. This torque could be a safety concern in the clinic if the coils are not securely mounted on the C-arm and the x-ray tube. Therefore, techniques are introduced to measure the torque on the coils. A previously defined model¹⁷ is also used to predict the coil torque in a magnetic field. This model can be used to predict coil torque for a coil of any size in a fringe field of arbitrary strength.

Since the shielding coils will draw dc electric current over periods of tens of minutes as fluoroscopy is being performed, methods are introduced to characterize coil heating. If the shielding system is used for long time periods in the cardiac catheterization laboratory, then coil heating could be a safety concern that must be addressed. A model is developed to explain the coil heating, which can be used to predict heating for an arbitrary coil design. The shielding system can be expanded and developed for other clinical applications of the CBXMR system.

II. MODELS

II.A. Correction technique for feedback sensor position

In the active shielding system, a linear Hall effect sensor is used to detect B acting on the electron beam of the x-ray tube. B will produce a signal in the sensor which will be sent to a feedback circuit. The feedback circuit will then send an electric current to the shielding coils to produce a B_c of the same strength as B to counter B .

For optimal magnetic shielding, the feedback sensor should be placed at the location of the electron beam in the x-ray tube. In practice, this is not possible since the electron beam is within a vacuum sealed glass insert of the x-ray tube. The closest reasonable position for the feedback sensor is at the x-ray tube output port, which is still several centimeters away from the electron beam. Since B_c at the sensor position is different than B_c at the electron beam position, the shielding coils will not accurately correct for B at the electron beam position. However, it is possible to introduce a correction such that B_c at the sensor position is reduced or

TABLE I. Sensor position correction parameters.

Parameter	Definition	Value
A	Sensor's conversion factor between magnetic field and voltage	50 mV/mT
k	Sensor's offset voltage value with no magnetic field present	2.5 V
L	Conversion factor relating current in the shielding coils to the magnetic field produced by the coils at the sensor position S	2.25 mT/A
α	Conversion factor relating magnetic field produced by the shielding coils at the sensor position to the field produced at the electron beam position	0.5
$B_c(m)$	Magnetic field produced by the shielding coils at the x-ray tube output port	Variable
$B_c(x)$	Magnetic field produced by the shielding coils at the electron beam position	Variable
I_c	Electric current in the shielding coils	Variable

increased. This will result in accurate B correction at the electron beam position. All parameters used in this correction are defined in Table I.

The voltage produced by the feedback sensor when it is located at the x-ray tube output port is given by

$$v_s = -A(B - B_c(m)) + k. \quad (1)$$

The desired signal produced by the sensor would be obtained if it was positioned at the electron beam, assuming that B at the sensor is equal to B at the electron beam. At this location, the sensor voltage is given by

$$v_f = -A(B - B_c(x)) + k. \quad (2)$$

Therefore, from Eqs. (1) and (2),

$$v_f = v_s - AB_c(m) + AB_c(x). \quad (3)$$

B_c produced by the shielding coils is linearly proportional to I_c . I_c is produced automatically by the feedback circuit. Therefore,

$$B_c(m) = LI_c, \quad (4a)$$

$$B_c(x) = \alpha B_c(m). \quad (4b)$$

Since the magnetic field produced by the shielding coils is linearly proportional to the current, the shielding coils are a linear system and $B_c(x)$ is proportional to $B_c(m)$. Substituting Eq. (4) into Eq. (3) yields

$$v_f = v_s + ALI_c(\alpha - 1). \quad (5)$$

The second term in Eq. (5) is the correction term. This means that the voltage produced by the sensor can be brought to its optimal value for shielding at the electron beam position by adding a voltage term that is directly proportional to the current through the shielding coils. This correction can be implemented in the feedback circuit.

TABLE II. Parameters for shielding coil heating model.

Parameter	Definition	Value
T	Coil temperature	Varies
T_0	Coil temperature at $t=0$	25.5 °C
T_A	Ambient air temperature	24.3 °C
g	Cooling rate constant	0.0257 [Eq. (10)]
t	Coil heating time	Varies
ρ_0	Resistivity of copper at T_0	1.72×10^{-8} Ω m
E	Cross-sectional area of coil wire	2.6×10^{-6} m ²
ℓ	Wire length of each coil	≈ 266 m
R_0	Coil resistance at T_0	1.9 Ω
α_0	Temperature coefficient of resistance of copper	0.0039 °C ⁻¹
c	Specific heat capacity of copper	385 J/(kg °C)
D	Density of copper	8920 kg m ⁻³
m	Coil mass	$=D\ell E$
R	Coil resistance for $t>0$	Equation (7)
ρ	Copper resistivity for $t>0$	Equation (7)

II.B. Shielding coil heating

When in use in a clinical x-ray system, the shielding coils draw current ($\sim 1\text{--}6$ A) for an extended time period ($\sim 10\text{--}30$ min) during a procedure. This will lead to coil heating over time. Heating of the shielding coils can be explained using a simple heat equation involving a heating rate provided by the current and a cooling rate that describes cooling to the air surrounding the coil. The resistance of a coil increases with temperature, so this fact must be included in the heating rate equation. The rate of heating in each coil (in °C/min) can be described by

$$\frac{dT}{dt} = q - g(T - T_A), \quad (6)$$

where all parameters of the model are defined in Table II. q is the heating rate of the coil due to the electric current in the coil ($I_c^2 R$ power dissipation), while the second term is the cooling rate of the coil by the surrounding air, according to Newton's law of cooling.¹⁸

The dependence of R (and ρ) on T is given by¹⁹

$$R = R_0(1 + \alpha_0(T - T_0)) \quad (7)$$

with all parameters defined in Table II. Therefore, Eq. (7) is also an appropriate model for ρ and it is used in q of Eq. (6). Note that¹¹

$$q = \frac{60I_c^2 R}{mc}. \quad (8)$$

The factor of 60 converts the heating rate to °C/min from °C/s. Substituting R and m into Eq. (8) yields

$$q = \frac{60I_c^2 \rho}{E^2 D c}. \quad (9)$$

Equation (9) can be substituted into Eq. (6), which is a differential equation that can be solved analytically. The solution to Eq. (6) is

$$T(t) = M e^{Nt} - P, \quad (10)$$

where M is a constant. N and P are given as

$$N = \frac{60I_c^2 \rho_0 \alpha_0}{E^2 D c} - g, \quad (11a)$$

$$P = \frac{60I_c^2 \rho_0 (1 - \alpha_0 T_A) / E^2 D c + g T_A}{60I_c^2 \rho_0 \alpha_0 / E^2 D c - g}. \quad (11b)$$

The only unknowns in Eq. (10) are M and g . These can be determined by substituting T_0 at $t=0$ min and another measured value of T at a later point in time into Eq. (10). This provides two equations with two unknowns, which can be solved to determine the complete solution for Eq. (10). This model can be used to predict the temperature increase over time for a shielding coil of arbitrary design if at least two measurements of coil temperature are obtained. This model assumes that the wire in the coils heats up uniformly without any spatial heating gradients throughout each coil, which is reasonable since the same direct electric current is present throughout the entire wire length of each coil. Of course, a heating gradient will be present at the boundary between each coil and the surrounding air.

III. MATERIALS

III.A. X-ray tube

A standard radiography rotating-anode x-ray tube (PX1461ES, Dunlee Inc., Aurora, IL) was used in the magnetic shielding experiments. This tube was used to characterize the active shielding system since a rotating-anode x-ray tube will be used in a clinical CBXMR system.

III.B. Air-cored electromagnet

An air-cored electromagnet composed of two coils (Stan- genes Industries, Palo Alto, CA) generated B and has been characterized in previous work.¹¹ The electromagnet was uniform to within $\pm 5\%$ of B at its center over a cubic region located at its center with 12 cm sides. The electromagnet was able to generate a maximum $B=23$ mT at the center between its two coils along its central axis.

III.C. Active shielding coils

Figure 2 shows the geometry of the shielding coils (Filt- ran, Ottawa, ON). The center point between the two coils, C, is shown. The x , y , and z direction definitions are also shown. B_c produced by the coils is directed along the z direction. The spacing between the coils was 23 cm to encompass the diameter of the x-ray tube. Two coils were used to maximize the field uniformity in the gap between them. The locations where the electron beam and sensor would be positioned, F and S, are also shown, including their distances from the inner face of the coil on the right. The parameters for each coil are given in Table III. The coils were capable of producing maximum $B_c=7$ mT at C along their central axis. The calibration factors of the coils at C, F, and S were 1.0, 1.1, and 2.25 mT/A, respectively. B_c was measured with a Hall

FIG. 2. Geometry of the active magnetic shielding coils. The x , y , and z directions are shown. The location of the center between the two coils, C, the electron beam position in the x-ray tube, F, and the Hall effect sensor position, S, are shown along the central axis of the coils. Their distances from the inner face of the coil on the right are shown. The coils were separated by a distance of 23 cm to encompass the diameter of the x-ray tube.

effect sensor probe system (model 4048, F.W. Bell, Orlando, FL). The current in the shielding coils was varied from 0 to 7 A. B_c varied linearly with the coil current at all three positions.

Figure 3 is a plot of the uniformity of B_c in the shielding coils along the x , y , and z directions. The origin $(x,y,z) = (0,0,0)$ is at C. B_c was not uniform in any direction and the strongest variation in B_c was in the z direction. B_c was uniform to within approximately $\pm 30\%$ of its value at C over a cubic region with 10 cm sides centered around C. B_c at the inner coil faces was about three times stronger than B_c at C.

IV. METHODS

IV.A. Active magnetic shielding experiment

Figure 4 shows the apparatus to characterize the magnetic shielding system for the x-ray tube. The x-ray tube was centered in the electromagnet and placed in between the shielding coils such that B from the electromagnet was directed perpendicular to the electron beam in the x-ray tube. A brass mount containing a 30 μm pinhole (Gammex RMI, Middle-

FIG. 3. Uniformity plots of B_c along the z direction (see Fig. 2) of the active shielding coils. The field uniformity was measured in the (a) x , (b) y , and (c) z directions. The origin $(x,y,z)=(0,0,0)$ is located at C between the two coils. Error bars are smaller than the markers on the plots.

TABLE III. Active shielding coil parameters.

Coil property	Coils described in Sec. III C
Inner radius	7 cm
Outer radius	10 cm
Mean radius	8.5 cm
Wire gauge	AWG No. 13
Wire diameter	1.83 mm
Coil thickness	4.9 cm
Spacing between coils	23 cm
No. of turns per coil	486 (27×18)
Mass per coil	6.2 kg
Coil material	Copper
Resistance per coil	1.9 Ω

ton, WI) at its center was attached to the x-ray tube output port. A linear Hall effect feedback sensor (A1321, Allegro MicroSystems Inc., Worcester, MA) was mounted on the front surface of the brass mount just below the pinhole at the x-ray tube output port. This location was used to bring the feedback sensor as close to the electron beam as possible and to ensure that the sensor was perpendicular to B for optimal measurement. The pinhole was used to obtain images of the x-ray tube focal spot on a flat-panel detector. The detector was placed 115 cm away from the pinhole for sufficient magnification of the focal spot image ($M \approx 16$). Focal spot images were obtained for $B=0-7$ mT with and without B_c applied. The x-ray tube voltage and current were set to 70 kV and 7 mA, respectively.

IV.A.1. Feedback circuit

Figure 5 shows a block diagram to explain the principle of operation of the feedback circuit. A reference signal is sent in to an amplifier and compared to a feedback signal to provide the necessary output to drive the shielding system to produce B_c . The amplifier always attempts to produce an output such that the input feedback and reference signals are equivalent.

FIG. 4. Apparatus to characterize the active shielding system for the x-ray tube. The x-ray tube was centered between the two coils of the air-cored electromagnet, which provided B . The active shielding coils were placed around the x-ray tube once it was mounted inside the electromagnet. A brass mount with a $30\ \mu\text{m}$ pinhole was mounted on the x-ray tube output port. A linear Hall effect sensor was placed just below the pinhole. Magnified images of the x-ray tube focal spot were obtained through the pinhole on a flat-panel detector 115 cm away.

This is known as negative feedback.²⁰ The output from the amplifier drives a current source (such as a transistor) which delivers the necessary electric current to the shielding system for B_c . The shielding system, composed of the shielding coils and the linear Hall effect sensor, produces an output proportional to the sensed magnetic field. Therefore, the shielding system produces a sensor signal (from the Hall effect sensor) and a correction signal (from a potentiometer). The correction signal is sent through an inverter and added (using an op-amp summer) with the sensor signal [Eq. (5)]. The sum of the two signals is inverted and delivered to the amplifier as the feedback signal. The output from the amplifier and the feedback will always be at the appropriate level to ensure that current to the shielding coils is sufficient to produce $|B_c| = |B|$.

IV.B. Shielding coil torque measurements

Figure 6(a) shows the apparatus to measure torque on one of the shielding coils in B . The shielding coil was centered in the electromagnet such that B was perpendicular to the magnetic field produced by the shielding coil. This ensured that a torque acted on the shielding coil after it was turned on. To avoid friction, the shielding coil was hung by a piece of string from a wooden beam mounted across the top of the electromagnet. Since the shielding coil was not resting on a

FIG. 5. Feedback circuit algorithm.

FIG. 6. (a) Apparatus to measure magnetic torque acting on the shielding coils in B . One of the shielding coils was centered in the air-cored electromagnet. A wooden beam was placed on top of the electromagnet and the coil was hung from the beam using string. (b) Technique to obtain the torque measurements. The air-cored electromagnet produced B which produced moment forces F_m on the shielding coil with a magnetic dipole moment μ . A blocker was used to fix the left side of the coil by providing a counterforce $-F_m$. On the right side of the shielding coil, a spring scale was used to provide $-F_m$ such that no net torque acted on the coil.

flat surface, this prevented friction from producing a torque to oppose the torque produced by B . Therefore, the measured torque was due to B only.

Figure 6(b) shows the technique to measure the torque on the shielding coil. A view from above the coil in the electromagnet is shown. When the coil was turned on, a magnetic dipole moment μ was produced. $B = 21\ \text{mT}$ was then applied from the electromagnet, and two moment arms acted on either side of the coil at its mean radius r in an attempt to align μ with B .¹⁷ The moment arms are given by the forces F_m . A blocker was placed in front of the coil on one side to produce $-F_m$. On the opposite side of the shielding coil, a 30 N spring scale was used to produce $-F_m$ such that no net force (and therefore no torque) was acting on the coil and it was unable to rotate in B . Therefore, the spring scale provided a measurement of F_m . The measured torque acting on the shielding coil in B is then given by

$$\tau_m = 2rF_m. \quad (12)$$

The current in the shielding coil was varied from 0 to 7 A.

IV.C. Coil heating measurements

To measure heating in the shielding coils, one of the coils was connected to a power supply and 6 A of dc was sent through it. A digital thermometer (HH81, Omega Engineering Inc., Stamford, CT) was used to measure the coil temperature over time by mounting it on the surface of the coil.

Thermal compound (Wakefield Engineering Inc., Pelham, NH) was used to improve the thermal connection between the coil and the thermometer.

Concurrently, the electrical resistance of the coil was also measured since resistance changes with temperature. These measurements were needed to verify a resistance temperature model, which is required to predict the heating rate of the shielding coils accurately. 6 A of dc was delivered directly to the coils and the voltage required to deliver this current was measured periodically every minute over a total time period of 30 min. Since the current and the voltage were known, coil resistance and its corresponding temperature dependence could be determined.

V. RESULTS

V.A. Active magnetic shielding experiment

Figure 7 shows images of focal spot deflection. The crosshairs indicate the center of the focal spot image location with no B applied. By including the sensor position correction, electron beam deflection was corrected to within an accuracy of 5% for $B=0-7$ mT.

V.B. Shielding coil torque measurements

Figure 8 is plot of the measured torque acting on the shielding coil versus the shielding coil current. The predicted values of the torque are also shown (Sec. VI B). The torque increased linearly with shielding current.

V.C. Coil heating measurements

Figures 9(a) and 9(b) show the coil resistance and heating measurements, respectively. The predicted values for coil resistance and temperature are also shown (see Sec. II B). The coil resistance increased linearly with temperature. The coil temperature rose linearly at the early stages of heating and the rate of temperature rise decreased over time.

VI. DISCUSSION

VI.A. Active magnetic shielding

With sensor position correction, the shielding system corrected electron beam deflection in the x-ray tube to an accuracy of $\pm 5\%$. In Fig. 7(b) the apparent focal spot image appears larger than with no B applied because B causes more deflection of one part of the electron beam than the other.¹⁶ Electrons that travel farther before colliding with the anode are deflected more than electrons that travel a shorter distance before collision. This shielding technique requires an initial calibration when the x-ray tube is placed in an arbitrary B , but no further calibration is necessary if the x-ray tube is positioned into different B . Therefore, the calibration only needs to be performed once. The calibration will account for the different values of B_c at the electron beam and sensor positions (see Fig. 3(c) and Sec. II A). The feedback will ensure that sufficient current is delivered to the coils if B changes.

FIG. 7. Focal spot images showing shielding performance. The crosshairs show the location of the center of the focal spot image at $B=0$ mT. (a) Focal spot image in $B=0$ mT. (b) Focal spot image deflection in $B=7$ mT. (c) Focal spot image position with shielding.

Previous work showed that B perpendicular to the electron beam will not exceed 5 mT if the C-arm is placed in the fringe field about 1 m from the entrance of the MRI scanner.¹⁶ Also, a collimator connected to the output port of the x-ray tube would be used to set the field of view for the procedure. Therefore, the shielding coils would need to be designed to adequately correct for this value of B and they

FIG. 8. Shielding coil torque in B . B was set to 21 mT. The error bars show the measured torque values and the solid line is the expected torque, which was predicted using Eq. (13).

would need to be of larger radius to fit around the collimator such that they are flush with the x-ray tube. In the clinical system, the shielding coils would only need to be activated during periods of x-ray exposure during the procedure.

If the C-arm needs to be placed closer to the scanner entrance for other clinical applications or near a weakly shielded scanner, then coils with more turns and/or more electric current are required to correct the stronger fringe field. Sensor position correction in stronger B would further increase the power requirements. It is also possible to combine the active shielding approach with collimation, which will reduce the required collimation and minimize field of view truncation. If a larger detector is used instead to adapt to field of view shift, then the shielding system will reduce

the degree to which the collimator blades need to be opened, minimizing unnecessary exposure to the patient.

In addition to sensor position correction, other options are possible to improve the active magnetic shielding system for the x-ray tube. First, the feedback sensor can be mounted inside the x-ray tube housing (but not the vacuum insert) such that it is closer to the electron beam for a more accurate measurement of B . Second, since the electron beam is not centered in the x-ray tube, it may be possible to improve the uniformity of B_c at the electron beam using an asymmetric coil geometry, such that each shielding coil in the coil pair has a different design.²¹ With this approach, the spatial derivative of B_c at the electron beam can be minimized to correct for electron beam deflection even if the feedback sensor and electron beam are separated by several centimeters.

VI.B. Shielding coil torque

The predicted torque acting on the shielding coil is given by¹⁷

$$\tau_p = \pi n I_c r^2 B, \quad (13)$$

where n is the number of turns in the coil. The coil parameters are given in Table III and I_c and B were set using the method in Sec. III C. Using these parameters, τ_p agreed with τ_m to within 10%. The dominant source of error between the predicted and measured torques occurred at low shielding coil currents. At low shielding coil currents, the torque on the shielding coil was weak and very little tension was present in the spring scale, so there was significant fluctuation in the F_m reading on the spring scale. This introduced additional error. At higher shielding coil currents, the tension in the spring scale was high and repeated measurements of F_m were more consistent with less fluctuation. Equation (13) can be used in designing the mechanical constraints for mounting the shielding coils on the x-ray C-arm.

Figure 8 shows that for clinically usable shielding coils of comparable size to those described in Table III, the torque acting on them does not exceed several N m. Therefore, shielding coils can be used safely in the MRI fringe field with secure mounting. In other applications that require closer proximity of the C-arm to the MRI scanner where the fringe fields are stronger, more robust mounting systems would be required to prevent coil rotation due to the larger magnetic torques produced.

VI.C. Coil heating

From Fig. 9, the coil temperature increased with time, but the rate of temperature rise decreased over time. The model in Eq. (7) was used in Fig. 9(a) and agreed with experiment to within 1%. Equation (10) was used as the model to fit the data in Fig. 9(b) and it agreed with experiment to within 1%, which is within the error of the measurement of T . In addition to T_0 , T at $t=20$ min was used as the second data point to solve for the unknowns in Eq. (10). The model predicts that the coil heating rate is strongest at the early stages of coil operation and that the heating rate decreases over time. This was observed experimentally.

FIG. 9. (a) Shielding coil resistance dependence upon temperature. The circles are the experimental results. The solid line is the predicted resistance based on Eq. (7). (b) Coil temperature measured over time as 6 A of dc was applied. The circles are the experimental data and the solid curve is based on the heating model of Eq. (10).

From Fig. 9(b), the shielding coil reached 50 °C after approximately 25 min, at which point the coils become painful to the touch. This duration is an adequate amount of fluoroscopy time for most PAVR procedures, although up to 45 min is required in some cases.²² However, it is possible to use a coil with a thicker wire diameter to reduce its heating rate, so a cooling system to reduce the heating rate of the coils will not normally be required if only one procedure is performed in a session. However, if multiple procedures are performed in succession, then a cooling system will be required to prevent overheating of the coils. This can be accomplished by surrounding the coils with additional heat sinks, implementing a water cooling system or even using ac cooling fans.

VII. CONCLUSION

We have developed an active shielding system that can correct for field of view shift in an MRI fringe field for percutaneous aortic valve replacement procedures. This shielding system will prevent field of view shift in the x-ray tubes of CBXMR systems without truncating the field of view or increasing the primary radiation exposure to the patient. The shielding system uses feedback to automatically adjust to changes in fringe field strength if the x-ray tube is moved into a different position during the procedure. The shielding correction can be improved by placing the feedback sensor closer to the electron beam using an asymmetrical shielding coil geometry to improve field uniformity at the electron beam position or by including a sensor position correction in the feedback circuit. The latter approach was investigated in this paper.

The torque acting on the shielding coils was found to be low in magnetic fields and this will not be a concern in the clinical CBXMR system if the shielding coils are securely mounted on the C-arm and the x-ray tube. Coil heating as a result of using shielding coils in the CBXMR system for PAVR procedures will not be a concern for a single procedure because the heating rate is too low to overheat the coils over the duration of the procedure. However, a cooling system for the heating coils will be required if multiple procedures are performed in one session.

Therefore, the active shielding system is an effective and simple approach for safely integrating x-ray tubes into CBXMR systems. This shielding system can be used to correct the electron beam deflection to within an accuracy of 5%, and it can also be used in combination with an x-ray collimator or a larger detector to correct for field of view shift in stronger magnetic fields (tens of mT). This combined approach will reduce field of view truncation and unnecessary exposure to the patient.

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