Closed-bore XMR (CBXMR) systems for aortic valve replacement: X-ray tube imaging performance

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A hybrid closed-bore x-ray/MRI system (CBXMR) is proposed to improve the safety and efficacy of percutaneous aortic valve replacement procedures. In this system, an x-ray C-arm will be positioned about 1 m from the entrance of a 1.5 T MRI scanner. The CBXMR system will harness the complementary strengths of both modalities to guide and deploy a bioprosthetic valve into the aortic annulus of the heart without coronary artery obstruction. A major challenge in constructing this system is ensuring proper operation of a rotating-anode x-ray tube in the MRI magnetic fringe field environment. The electron beam in the x-ray tube responsible for producing x rays can be deflected by the fringe field. However, the clinical impact of electron beam deflection in a magnetic field has not yet been studied. Here, the authors investigated changes in focal spot resolving power, field of view shift, and field of view truncation in x-ray images as a result of electron beam deflection. The authors found that in the fringe field acting on the x-ray tube at the clinical location for the x-ray C-arm (4 mT), focal spot size increased by only 2%, so the fringe field did not limit the resolving power of the x-ray system. The magnetic field also caused the field of view to shift by 3 mm. This shift must be corrected to avoid unnecessary primary radiation exposure to the patient and the staff in the cardiac catheterization laboratory. The fringe field was too weak to cause field of view truncation. © 2009 American Association of Physicists in Medicine.

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I. INTRODUCTION

Open heart surgical replacement of the aortic valve is an effective treatment for symptomatic aortic stenosis patients.¹ However, typically one-third of patients who require this treatment does not actually receive it due to the risks associated with surgery.^{1,2} Without treatment, the 2 yr mortality rate of aortic stenosis patients is approximately 50%.³ Therefore, cardiologists are developing a minimally invasive catheter-based technique known as percutaneous aortic valve replacement (PAVR) to increase treatment availability to more patients.^{4–10}

To improve the safety and efficacy of PAVR, we are developing a closed bore hybrid x-ray/MRI system (CBXMR) (Ref. 11) in which a C-arm x-ray system is placed close to (≈ 1 m) the entrance of a 1.5 T MRI scanner.¹² This system will harness the complementary strengths of both modalities by providing high quality fluoroscopy and angiography using x-ray imaging and cardiac anatomical imaging using MRI without the risks associated with patient transfers over long distances (several meters).¹² The challenges in the construction of this system have been outlined in previous publica-

tions, most of which have been overcome.^{11–15} The remaining challenge is to overcome x-ray tube electron beam deflection in a magnetic field perpendicular to the electron beam.

Although previous work has described in detail the behavior of an x-ray tube electron beam in a magnetic field,^{13–15} the clinical impact of this behavior has not yet been addressed. This article addresses the clinical consequences of electron beam deflection in a magnetic field perpendicular to the electron beam, which are field of view shift, field of view truncation, and decreased resolving power of the focal spot. Potential solutions for overcoming these problems are also provided.

Electron beam (and focal spot) deflection in a magnetic field can cause an increase in focal spot size perceived by the detector. This decreases the resolving power of the focal spot. The most demanding imaging task during a PAVR procedure is to resolve a guidewire accurately relative to its surroundings. The diameter of a 0.035 in. guidewire used in PAVR procedures referred to the imaging plane is about 1 mm. Therefore, a spatial resolution of about 1 line pair/mm (500 μ m) or better should be used to image the guidewire.

TABLE I. Parameters for the focal spot model.

Symbol	Property	Value
L_{fs}	Apparent focal spot length	See Sec. II A
W_{fs}	Apparent focal spot width	See Sec. II A
L_t	True focal spot length	See Sec. II A
L'_f	Apparent focal spot length at $B=0$ mT	1.0 ± 0.3 mm
α	Anode angle	16°
x_1	Focal spot-to-pinhole distance	72 ± 2 mm
x_2	Pinhole-to-detector distance	72 ± 1 cm
M_D	Image magnification of apparent focal spot	10
W'	Apparent focal spot image width at $B=0$ mT	6 ± 0.2 mm
Z_f	See Fig. 1(b)	Eq. (5b)
Y_f	Horizontal length of apparent focal spot image	10 ± 3 mm
γ	Angle between $\sqrt{Z_f^2 + Y_f^2}$ and Y_f	$=\tan^{-1}(Z_f/Y_f)$
e_1, e_2	Deflection of points P1 and P2 of focal spot	Varies
D_{fa}	Distance from filament to center of true focal spot on anode	Depends on x-ray tube
(Py_1, Pz_1)	Y and Z components of P1 projected onto detector	Eq. (3)
(Py_2, Pz_2)	Y and Z components of P2 projected onto detector	Eq. (3)
M_S	Magnification of star phantom images	2.7

This resolution is easily attainable with current interventional cardiology x-ray systems, but the increase in apparent focal spot size due to a magnetic field must not prevent this resolution from being achieved.

Electron beam deflection can also cause the x-ray field of view to shift and decrease in size. A shift in the field of view can prevent primary radiation from interacting with the active region of the detector. This can lead to exposure of patient anatomy that will not be imaged and it can also lead to unnecessary exposure to the operators in the cardiac catheterization laboratory. In a clinical x-ray system, the boundaries of the x-ray field of view must be aligned to the active area of the image receptor to within 2% of the source-to-image distance (SID), as required by federal regulations.¹⁶ This level of error could already be present in x-ray C-arm systems, so additional field of view shift due to the magnetic fringe field can violate these regulations. This violation can be avoided by moving the blades of the x-ray collimator in such that the field of view is aligned within regulations.

Field of view truncation can limit the field of view available to the cardiologist during a PAVR procedure. To encompass all the relevant cardiac anatomy for a PAVR procedure, the field of view should be at least 18 cm on a side. This will ensure that the apex of the heart, the ascending aorta, the coronary arteries, the ventricles, the atria, and other anatomical landmarks (such as the diaphragm and spine) can be visualized in the images to improve the safety and efficacy of the intervention.^{4–10} This will also improve procedural flexibility, enabling the cardiologist to choose between the transapical⁸ and transfemoral^{4,6} approach for replacement of the aortic valve.

Models are developed to predict focal spot size change seen by the detector, focal spot cutoff frequency in the detector plane (from which focal spot resolving power can be determined), field of view shift, and field of view truncation as a result of electron beam deflection in the MRI fringe

field. These models are an extension of previously developed models which predicted focal spot deflection in a magnetic field.¹⁴ Techniques are introduced to measure focal spot deflection using a pinhole. This approach enables focal spot deflection to be visualized directly from the acquired images.

To validate the models, experimental measurements of focal spot cutoff frequency in the detector plane, field of view shift, and field of view truncation in a static magnetic field are obtained. Focal spot cutoff frequency is measured directly using star phantoms,¹⁷ from which the size of the focal spot and its resolving power can also be inferred. Field of view shift is measured as a result of the focal spot deflection using an x-ray collimator, which mimics the clinical situation accurately. In strong magnetic fields (tens of mT), field of view size reduction becomes a concern, so an experiment is performed to simulate this effect using a circular aperture. The strength and direction of the MRI fringe field at the x-ray tube location in the CBXMR system are required to determine whether or not the spatial resolution requirements are met in the fringe field, so they are measured. The models here can be used to predict whether or not an arbitrary rotating-anode x-ray tube can meet specific imaging requirements for any clinical application when placed in a MRI fringe field that causes electron beam deflection.

II. MODELS

In this article, the external magnetic field acting on the x-ray tube and the corresponding deflection of the electron beam will be defined as B and e , respectively. Section II A describes a model that can be used to predict changes in apparent focal spot size and resolving power as a result of electron beam deflection in B . Section II B describes a model to predict field of view shift and truncation as a result of electron beam deflection in B . All parameters for the models are defined in Tables I and II.

TABLE II. Parameters for the field of view model.

Symbol	Property	Value
e	Deflection of the center of the true focal spot on the anode	Measured using technique in Sec. III B
a	Distance from focal spot to collimator blades	17 ± 1 cm
b	Distance from collimator blades to detector	92 ± 1 cm
$h1$	Distance from the focal spot to the off-focal blades	5.1 ± 0.2 cm
$h2$	See Fig. 3(b)	Eq. (13)
S_0	Spacing between the off-focal blades	Varies
$m+n$	$=S_0/2$	Varies
g	Shift of the top boundary of the field of view in the imaging plane	Eq. (10)
j	Shift of the bottom boundary of the field of view in the imaging plane	Eq. (14)
FOV	Field of view dimension before electron beam deflection	Varies
FOV _{new}	Field of view dimension after electron beam deflection	Eq. (15)

II.A. Apparent focal spot resolving power

In this article, the *apparent* focal spot will refer to the focal spot seen by the detector, while the *true* focal spot will refer to the focal spot area produced where electrons collide with the anode. The electrons in the beam collide on a region of the anode which is angled (anode angle α). The length of the true focal spot L_t and the length of the apparent focal spot L'_{fs} are related by¹⁷

$$L'_{fs} = L_t \sin \alpha. \quad (1)$$

When B is applied perpendicular to the electron beam, the true focal spot is deflected in the direction of $E \times B$, where E is the electric field vector applied from the anode to the cathode in the x-ray tube. Well-established models have shown that true focal spot deflection in B can be determined by solving the relativistic Lorentz force equation for electron velocity and integrating the result to determine the deflection in three-dimensional (3D) space.^{13,14,18} This equation is

$$\frac{dp}{dt} = q(E + v \times B), \quad (2)$$

where p is the relativistic momentum of an electron, q is the electric charge, and v is the electron velocity vector.

Figure 1(a) qualitatively describes the change in shape of the true focal spot as a result of electron beam deflection in a magnetic field perpendicular to the beam. The electron beam is incident on the anode at the true focal spot. P1 and P2 show the boundaries of the true focal spot on the anode. Since the anode is angled, electrons incident at P2 travel farther than electrons incident at P1. Therefore, according to Eq. (2), the electrons incident at P2 get deflected more in the magnetic field than electrons incident at P1. This causes the true focal spot to have a parallelogram shape on the anode if it is deflected in the magnetic field. The directions of E and B are also shown. In Fig. 1(b), the parameters of the apparent focal spot image after electron beam deflection in B are also shown (see Table I).

Figure 2(a) describes a simple pinhole camera model that can be used to explain the change in shape of the apparent

focal spot as a result of deflection in a magnetic field. The origin O is located at the pinhole. With this model, any point in 3D space, $P=(x_1, y_1, z_1)$ (such as a point on the true focal spot) can be projected through the pinhole onto a 2D surface [point $S=(y_2, z_2)$ on the detector] using the geometry of similar triangles.¹⁹ From Fig. 2(b), the coordinates of point S are given by

$$y_2 = \frac{-x_2 y_1}{x_1}, \quad (3a)$$

FIG. 1. (a) Explanation for change in true focal shape on the anode. Since the anode is angled, electrons incident at P2 on the anode have farther to travel in the magnetic field than electrons incident at P1. Therefore, the electrons incident at P2 are deflected more in the magnetic field than the electrons incident at P1. This results in a parallelogram shape for the true focal spot on the anode after it has been deflected by the magnetic field. The distance from the filament to the center of the focal spot on the anode, D_{fa} , is also shown. (b) Apparent focal spot image parameters due to electron beam deflection in a magnetic field (defined in Table I). The magnified values of apparent focal spot length and width are also shown.

FIG. 2. (a) Pinhole camera model for the focal spot. The origin O is located at the pinhole. Any point, $P=(x_1, y_1, z_1)$ on the true focal spot can be projected onto the flat-panel detector at point $S=(y_2, z_2)$. The distance from the pinhole to the detector, x_2 , is shown. (b) A similar triangle approach can be used to predict S . All parameters are defined in Table I.

$$z_2 = \frac{-x_2 z_1}{x_1}. \quad (3b)$$

The magnification is $M_D = x_2/x_1$. In Fig. 1(a), P1 and P2 are the boundary points passing through the center line of the true focal spot on the anode. Using the coordinate system in Fig. 2, the coordinates of these points are given by

$$P1 = \left(x_1 + \frac{L_t \cos \alpha}{2}, \frac{L'_{fs}}{2}, -e_1 \right), \quad (4a)$$

$$P2 = \left(x_1 - \frac{L_t \cos \alpha}{2}, -\frac{L'_{fs}}{2}, -e_2 \right). \quad (4b)$$

These coordinates were obtained by placing the X axis through the center of the true focal spot. The directions of E and B in Fig. 1 are in the Y and $-X$ directions in Fig. 2, respectively, producing a true focal spot deflection in the $-Z$ direction of Fig. 2. e_1 and e_2 are determined by solving Eq. (2) numerically for the components of v in the Y and Z directions and then integrating these components to determine electron displacement in the Y and Z directions. e_1 and e_2 (deflections in the $-Z$ direction) will be determined when the Y components of the electron displacement are $-(D_{fa} - (L_t \sin \alpha)/2)$ and $-(D_{fa} + (L_t \sin \alpha)/2)$, respectively. These deflections will be circumferential shifts on the anode.

At this point, P1 and P2 can be substituted into Eqs. (3a) and (3b) to determine P_{y1} , P_{z1} , P_{y2} , and P_{z2} . Y_f and Z_f are then determined by

$$Y_f = P_{y1} - P_{y2}, \quad (5a)$$

$$Z_f = P_{z1} - P_{z2}. \quad (5b)$$

Finally, from Fig. 1(b),

$$L_{fs} = \sqrt{Y_f^2 + (Z_f + W')^2} / M_D. \quad (6)$$

W' does not change with increasing B . As the true focal spot is shifted down on the anode, W' is unaffected since the entire vertical dimension of the apparent focal spot image is projected onto the detector regardless. However, W_{fs} will change due to an increase in γ . Therefore, using the geometry shown in Fig. 1(b), W_{fs} is given by

$$W_{fs} = W' \cos \gamma / M_D. \quad (7)$$

The measured values of L_{fs} and W_{fs} from star phantoms can be determined using the following equations:²⁰

$$L_{fs} = \left(\frac{\pi \theta}{180} \right) \frac{l}{(M_S - 1)}, \quad (8a)$$

$$W_{fs} = \left(\frac{\pi \theta}{180} \right) \frac{s}{(M_S - 1)}, \quad (8b)$$

where s and l are the shortest and longest lengths of the blurred region measured in a star phantom image and θ is the spoke angle of the star phantom (2°).

The cutoff frequency of the focal spot in the detector plane (in line pairs/mm) is determined by²¹

$$R = 90 / \pi l, \quad (9)$$

where s can be substituted for l to determine the cutoff frequency in the direction of the shortest blurred region in the star phantom images. This cutoff frequency depends on the star phantom image magnification. From this result, the focal spot resolving power in the object plane can be determined if the magnification of the object being imaged is known. The resolving power is simply the result of Eq. (9) multiplied by this magnification.²²

II.B. Field of view

II.B.1. Shift

In a clinical x-ray system, a *collimator* is used to define the field of view (see Fig. 3). The collimator contains three pairs of lead blades. Two pairs of blades in the body of the collimator, one pair perpendicular to the other, are used to define the field of view. These are known as the *collimator blades*. A third pair of blades (the *off-focal blades*) is present outside of the collimator body close to the x-ray tube output port. These blades are used to obstruct off focal¹⁷ and scatter radiation and prevent this radiation from entering the images. In a properly designed collimator, the off-focal blades are backed off away from the x-ray field such that they do not interfere with the field of view defined by the collimator blades. B perpendicular to the electron beam can cause a field of view shift, as shown in Fig. 3(a). Using the geometry of similar triangles, the field of view shift is given by

$$FOV_s = eb/a. \quad (10)$$

FIG. 3. (a) Geometry of the field of view shift caused by a magnetic field B . The field of view on the detector is determined by the collimator blades. After deflection of the focal spot by B , the field of view is shifted on the detector. (b) Field of view truncation occurs if the focal spot deflects beyond the boundary of one of the off-focal blades. All parameters are defined in Table II.

II.B.2. Truncation

In addition to causing a shift of the field of view boundaries, a strong deflection of the electron beam can deflect the focal spot beyond the boundary of one of the off-focal blades. Once the focal spot is behind the off-focal blade, the blade can truncate the field of view. Figure 3(b) shows the model describing this effect. All parameters of the model are defined in Table II. g is determined from Eq. (10). From Fig. 3(b), using similar triangle geometry,

$$m = h1(FOV/2)/(a + b). \quad (11)$$

Here, n can be determined from m and the spacing S_0 between the off-focal blades as follows:

$$n = \frac{S_0}{2} - m. \quad (12)$$

Using similar triangles, $h2$ is given by

FIG. 4. Directions and field definitions for the magnetic fringe field outside the MRI scanner entrance. z is the longitudinal direction along the length of the patient table and r is the radial direction out from the center of the table. The fringe field anywhere outside the scanner entrance can be resolved into components along these directions, defined as B_r and B_z . The origin in this coordinate system $(z, r) = (0, 0)$ is located at the entrance of the MRI scanner along the center line of the patient tabletop.

$$h2 = \frac{e(h1)}{(e - n)}. \quad (13)$$

Here, j is determined by

$$j = \frac{n(a + b - h2)}{(h2 - h1)}. \quad (14)$$

Finally, FOV_{new} is given by

$$FOV_{new} = FOV + g - j. \quad (15)$$

III. MATERIALS AND METHODS

III.A. MRI fringe field

The fringe field of a self-shielded 1.5 T MRI scanner (General Electric Signa Excite 14.0) was measured with a Hall effect sensor probe system (model 4048, F. W. Bell, Orlando, FL). The longitudinal field parallel with the length of the patient table B_z and the radial field perpendicular to the patient table B_r were measured (see Fig. 4). The longitudinal direction z and the radial direction r are shown. In Fig. 4, r and B_r are shown to point in the vertical direction. However, the scanner is a solenoid, so B_r is radially symmetric. This means that r can be directed along any direction perpendicular to the patient table top (not just vertically) and B_r measured along r will produce the same data set regardless of the direction chosen for r . The position $(z, r) = (0, 0)$ is located at the entrance of the MRI scanner ($z = 0$) at the center of the patient table top ($r = 0$) in this coordinate system. Therefore, the x-ray tube would be located near the point $(z, r) = (100, 50)$ cm. Since B_r is radially symmetric, oblique angulation of the C-arm will not change the value of B_r acting on the electron beam, assuming that the isocenters of the C-arm and the MRI scanner are properly aligned. Without cranial or caudal angulation of the C-arm, B_r is the only component of the fringe field capable of causing electron beam deflection since B_z is parallel with the electron beam and unable to deflect the beam. If cranial or caudal angulation is introduced, then both B_r and B_z will contribute com-

FIG. 5. (a) Apparatus to image focal spot deflection. A brass mount with a pinhole is placed on the output port of the x-ray tube, through which a magnified image of the apparent focal spot can be obtained on a flat-panel detector. (b) Apparatus to measure focal spot cutoff frequency at the detector plane using a star phantom. (c) Apparatus to measure field of view shift in B . The standard radiography x-ray tube containing a collimator is centered in the electromagnet. (d) Apparatus to measure field of view truncation. A brass mount containing a circular aperture at its center was positioned at the x-ray tube output port to simulate a collimator with off-focal and collimator blades.

ponents that are perpendicular to the electron beam and cause deflection. However, cranial/caudal angulation is not essential for PAVR procedures.⁶

B_z and B_r were measured from $z=0$ cm in increments of $z=10$ cm out to $z=200$ cm. At each value of z , B_z and B_r were measured from $r=0$ cm in increments of $r=10$ cm out to $r=100$ cm.

III.B. Focal spot deflection

A custom-made nonferromagnetic x-ray tube (Rytech X-ray Inc., Scarborough, ON, Canada) was used initially to evaluate apparent focal spot size and focal spot detector cut-off frequency in B .²³ This x-ray tube was constructed with nonferromagnetic materials for the focusing cup. An air-cored stator was used in place of the iron-cored stator found in standard x-ray tubes. Since this x-ray tube was lacking ferromagnetic components, B at the electron beam position could be accurately determined. The distance from the filament to the center of the true focal spot on the anode, D_{fa} , of this x-ray tube was 1.9 cm.

Once it became available, a standard radiography x-ray tube (Varian A292, Varian Medical Systems, Palo Alto, CA) was used to characterize field of view shift and truncation since a standard x-ray tube with ferromagnetic components will be used in the clinical CBXMR system. This tube had superior induction motor performance in the fringe field compared with the nonferromagnetic x-ray tube¹² since the air-cored stator provided minimal motor torque. The focal

spot behavior of the standard radiography x-ray tube in the fringe field was similar to the nonferromagnetic tube (Sec. IV C). D_{fa} for this x-ray tube was 1.2 cm.

An air-cored electromagnet (Stangenes Industries, Palo Alto, CA) was used to produce B and is described in detail elsewhere.¹² The electromagnet was able to produce a maximum B of 23 mT at its center along its central axis and field uniformity was within $\pm 5\%$ over a cubic region with 12 cm sides around its center. This magnet was able to provide a magnetic field component perpendicular to the electron beam of comparable strength to the fringe field components B_r and B_z of the MRI scanner, both of which can cause electron beam deflection depending on x-ray tube orientation.

Figure 5(a) shows the apparatus for determining the true focal spot deflection in B . A 30 μm pinhole (Gammex RMI, Middleton, WI) was fastened into a brass mount and the mount was placed on the output port of the nonferromagnetic x-ray tube or the standard radiography x-ray tube. Either x-ray tube was placed into the electromagnet such that B was perpendicular to the electron beam. Magnified images ($M_D = 10$) of the apparent focal spot were obtained on a flat-panel detector (FPD14, Anrad Corporation, St. Laurent, Quebec, Canada), which was positioned 72 cm from the pinhole. This distance ensured that the magnified image of the apparent focal spot had adequate detector element (del) coverage. B was varied from 0 to 23 mT. The x-ray tube voltage and current were set to 70 kV and 7 mA, respectively, for both x-ray tubes. Increasing the kV will cause a decrease in deflection for the range of B studied¹⁴ since the increased acceleration of the electrons in stronger E reduces the electron

time of flight available for deflection. Deflection and Z_f for both tubes were measured directly from the resulting images.

III.C. Apparent focal spot cutoff frequency in the detector plane

Figure 5(b) shows the experiment for measuring the apparent focal spot cutoff frequency in the detector plane (and L_{fs} and W_{fs}) of the nonferromagnetic x-ray tube in B . The nonferromagnetic x-ray tube was centered in the electromagnet such that B was perpendicular to the electron beam. The flat-panel detector was placed 91.5 cm from the output port of the x-ray tube. A star phantom (Type 9, Funk, Erlangen, Germany) was placed between the detector and the x-ray tube 60 cm from the detector. The star phantom was placed centered on the central axis of the undeflected focal spot. Therefore, the off-central axis focal spot cutoff frequency was measured after deflection. The 60 cm distance provided adequate magnification ($M_S=2.7$) such that the del size (300 μm) was contained within the blurred region of the star phantom images. This ensured that the del size did not interfere with the measurements of the focal spot. The spoked region of the phantom was 45 mm in diameter. Using the same x-ray tube voltage and current settings as Sec. III B, images of the phantom were obtained with B ranging from 0 to 23 mT.

III.D. Field of view

III.D.1. Shift

Figure 5(c) shows the apparatus used to measure field of view shift of the standard radiography x-ray tube in B . The tube was centered inside the electromagnet with B perpendicular to the electron beam. A collimator (model 150M, Huestis Medical, Bristol, RI) was mounted on the output port of the x-ray tube. Before applying B , the field of view was set to 18 cm on a side using the collimator. The flat-panel detector was placed 1 m from the output port of the x-ray tube. This will be the approximate SID used in the clinical CBXMR system. The detector was used to obtain images of the field of view from the collimator. A bar pattern phantom (Type 21, Funk, Erlangen, Germany) was mounted to the detector and used as a reference object to visualize the field of view shift more easily. Images of the field of view were obtained with B ranging from 0 to 23 mT. The x-ray tube current and voltage settings were identical to those in Sec. III B.

III.D.2. Truncation

Figure 5(d) shows the apparatus to study field of view truncation with the standard radiography x-ray tube. A brass mount with a circular aperture at its center was placed flush with the x-ray tube output port. Since the mount was over 1 cm thick, it simulated a collimator with off-focal and collimator blades using the faces of the aperture closest to and farthest from the x-ray tube output port, respectively. A reference image was obtained with the aperture centered within the x-ray tube output port. The mount was then placed 3 mm

TABLE III. Field of view truncation parameters of the circular aperture.

Symbol	Value
a	6.4 ± 0.2 cm
b	103 ± 0.2 cm
$h1$	5.1 ± 0.2 cm
S_0	4 ± 0.1 mm
$m+n$	2 ± 0.05 mm
FOV	6.7 ± 0.2 cm

above the center of the output port to ensure that the focal spot was beyond the boundary of the aperture. B from the electromagnet was not strong enough to deflect the electron beam sufficiently to cause field of view truncation, so the mount was raised above its initial position in increments of 1 mm to shift the aperture relative to the focal spot. This is equivalent to shifting the focal spot down on the anode in 1 mm increments. At each position, images of the aperture were obtained on the flat-panel detector positioned 1 m away from the x-ray tube output port. Table III lists the parameters of the circular aperture, which are needed for the model of Sec. II B 2. The x-ray tube current and voltage settings were identical to those in Sec. III B.

IV. RESULTS

IV.A. MRI fringe field

Figures 6(a) and 6(b) are plots of B_r and B_z at $z=1.4$ m, which is within the proper clinical location range for the x-ray tube in the CBXMR system for percutaneous aortic

FIG. 6. Plots of (a) B_r and (b) B_z of the self-shielded 1.5 T MRI scanner at $z=1.4$ m from the scanner entrance. These field components are described in Fig. 4. The circled regions highlight the fringe field at the location of the x-ray tube underneath the patient table in the CBXMR system.

FIG. 7. Images of focal spot deflection for the nonferromagnetic x-ray tube for $B=0$ to 23 mT. The crosshairs show the location of the center of the magnified apparent focal spot at $B=0$ mT. The apparent focal spot image moved up on the detector as B was increased. The change in apparent focal spot shape described in Fig. 1 is seen in the images.

valve replacement procedures. B_r was negligible at the patient table and $|B_r|$ increased sharply with increasing r up to $r=70$ cm. $|B_r|$ decreased as r was increased beyond this. $|B_z|$ gradually decreased as r was increased. The circled regions in Fig. 6 show B_r and B_z at the location of the x-ray tube focal spot position under the patient table in the CBXMR system, which is $r=50$ cm.

IV.B. Focal spot deflection

Figure 7 shows images of the apparent focal spot deflection of the nonferromagnetic x-ray tube. In Fig. 7, the crosshairs provide the location of the original apparent focal spot image before deflection. B perpendicular to the electron beam caused the apparent focal spot image to be deflected up on the detector. The apparent focal spot image assumed a more pronounced parallelogram shape with increasing B , resulting in increased Z_f .

Figure 8 is a plot of deflection on the anode with B perpendicular to the electron beam for (a) the nonferromagnetic x-ray tube and (b) the standard radiography x-ray tube. As expected, deflection increased with B for both tubes. Figure 8(c) is a plot of Z_f dependence upon B for the nonferromagnetic x-ray tube. Z_f increased with B as the model in Sec. II A predicts. The predicted values of Z_f are also shown, which are based on Eq. (5b). No change in Y_f was observed over the full range of B .

IV.C. Apparent focal spot cutoff frequency in the detector plane

Figure 9 shows the star phantom images for B perpendicular to the electron beam. The solid and dashed arrows in the images indicate s (shortest length of blurred region) and l (longest length of blurred region). With no B applied, s and l were measured to be 3.0 ± 0.1 cm ($R=0.9$ line pair/mm) and 4.9 ± 0.1 cm ($R=0.6$ line pair/mm), respectively. l increased to 9.7 ± 0.2 cm ($R=0.3$ line pair/mm) and s decreased to 2.1 ± 0.06 cm ($R=1.4$ line pairs/mm) as B was increased to 23 mT. The error in these quantities is the standard error of the mean of three measurements. Also, the orientation of s and l rotated counterclockwise as B was increased.

Figure 10 shows the dependence of W_{fs} and L_{fs} (Table I) on B for the nonferromagnetic x-ray tube, with B applied perpendicular to the electron beam. These values were measured with the star pattern. L_{fs} increased and W_{fs} decreased as B was increased to 23 mT. The predicted values of L_{fs} and W_{fs} are also shown using the model in Sec. II A with the predicted Z_f values.

IV.D. Field of view

IV.D.1. Shift

Figure 11 shows images of field of view shift for the standard radiography x-ray tube for B at (a) 0 mT and (b) 23

FIG. 8. True focal spot deflection as a function of B for (a) the nonferromagnetic x-ray tube and (b) the standard radiography tube. The solid lines are the deflection models for each x-ray tube, based on Eq. (2). (c) Plot of measured Z_f [see Table I and Fig. 1(b)] as a function of B for the nonferromagnetic x-ray tube. Z_f increased as B was increased, causing a more pronounced parallelogram shape for the apparent focal spot image. The “x” marks (connected with a dashed curve) are the predicted Z_f values based on Eq. (5b).

mT. The field of view shifted up on the detector, which is apparent from the horizontal reference lines that have been added above and below the images. Figure 12 is a plot of FOV_S for the standard radiography x-ray tube as a function of B . The predicted values for FOV_S are also shown using the model in Sec. II B 1. FOV_S was 1.3 cm on the detector at $B=23$ mT.

IV.D.2. Truncation

Figure 13 shows images of the field of view of the circular aperture (a) centered with respect to the x-ray tube output port and (b) shifted 4 mm above the center of the output port. In addition to the shift of the field of view boundaries defined by g and j , a significant truncation of the field of view is apparent. In these images, FOV and FOV_{new} are defined as the diameter of the aperture image taken from the center of

FIG. 9. Star phantom images to measure focal spot cutoff frequency at the detector for different values of B for the nonferromagnetic x-ray tube. B was varied from 0 to 23 mT. The dashed double arrows define l , the length of the longest blurred region in the star phantom images. The solid double arrow is s , which is the length of the shortest blurred region in each star phantom image. These lengths are used to determine apparent focal spot size. l increased and s decreased with increasing B .

FIG. 10. Plot of apparent focal spot length L_{fs} and apparent focal spot width W_{fs} as a function of the magnetic field B for the nonferromagnetic x-ray tube. The circles are the measured values, obtained with the star pattern. The x marks connected by the dashed curve are the predicted values of L_{fs} given by Eq. (6) and the x marks connected by the solid curve are the predicted values for W_{fs} given by Eq. (7).

FIG. 11. Images of the field of view for the standard radiography x-ray tube at B of (a) 0 mT and (b) 23 mT. A field of view shift was observed relative to the bar phantom in the images as B was applied. Horizontal lines have been applied above and below the images to highlight the shift.

FIG. 12. Plot of the field of view shift relative to B . The solid line is the predicted field of view shift from Eq. (10).

the top blurred region to the center of the bottom blurred region. Figure 13(c) is a plot of measured FOV_{new} for $3 \text{ mm} \leq e \leq 7 \text{ mm}$. The predicted values for FOV_{new} are also shown.

FIG. 13. Image of a circular aperture (a) aligned with the center of the x-ray tube output port and (b) with the focal spot shifted down 4 mm with respect to the aperture. (c) Plot of vertical FOV_{new} with focal spot deflection relative to the aperture. The solid line is the predicted vertical FOV_{new} based on Eq. (15). As a reference point, the vertical FOV at no deflection is shown on the plot.

V. DISCUSSION

V.A. MRI fringe field

In Fig. 6 at $z=1.4 \text{ m}$, B_r was negligible at $r=0$ because the MRI scanner is a superconducting solenoid, and the field through the center of a solenoid has a B_z component exclusively. B_r increased with increasing r as the curvature of the field was directed perpendicular to the table up to $r=70 \text{ cm}$. Beyond $r=70 \text{ cm}$, B_r decreased as the field curved back toward the scanner entrance. B_z decreased as r increased due to increased radial spread of the field beyond the center axis of the MRI scanner.

V.B. Apparent focal spot resolving power

In Fig. 7, the apparent focal spot image was deflected up on the detector with increasing B . This means that the true focal spot was deflected down on the anode. E was directed from the anode to the cathode and B was directed from the back of the x-ray tube toward the output port. Therefore, $E \times B$ and the electron beam deflection are directed down, which agrees with the results in Fig. 7. In Figs. 8(a) and 8(b), the experimental results agree with the deflection model based on Eq. (2) to within 7%, which is within the measured error of the focal spot deflection. In Fig. 8(c), the predicted Z_f (Table I) agreed with the measured values to within 8%, which is within the error of the measurement of Z_f .

In Fig. 9, the increase in l and decrease in s (Sec. IV C) was caused by the change in L_{fs} and W_{fs} of the apparent focal spot as a result of deflection (see Fig. 7). l and s rotated counterclockwise with increasing B due to the increased angulation of L_{fs} and W_{fs} as B was changed. In Fig. 10, the predicted L_{fs} and W_{fs} agreed with the experimental values to within 12%, which is within the error of the measurement of Z_f , L_{fs} , and W_{fs} . No change in Y_f was observed as B was changed because the true focal spot was deflected in the $-Z$ direction (down) of Fig. 2(a) exclusively.

In the CBXMR system, the true focal spot in the x-ray tube will be placed at $r=50 \text{ cm}$ below the patient table top at about $z=1 \text{ m}$ from the entrance of the MRI scanner. This will provide a source-to-image distance of approximately 1 m, which is typical in interventional cardiology.²⁴ Therefore, from Fig. 6(a), the x-ray tube electron beam will be exposed to $B_r=4.0 \text{ mT}$ at $z=1.4 \text{ m}$. Even with oblique angulation of the C-arm, this value of B_r will not be exceeded. From Fig. 10, $L_{fs}=1.02 \text{ mm}$ in $B_r=4.0 \text{ mT}$. $L_{fs}=1.02 \text{ mm}$ yields $R=4.9 \text{ line pairs/mm}$ [from Eqs. (8) and (9)]. At the magnifications used in interventional cardiology (about 1.2), this produces a resolving power of 5.9 line pairs/mm in the object plane. Even in the fringe field, this resolving power exceeds what is required for aortic valve replacement procedures and no corrections are needed to reduce the size of the apparent focal spot.

V.C. Field of view

V.C.1. Shift

The measured FOV_S values agreed with the predicted values to within 8%, which is within the error of the measure-

ment of the location of the collimator blades relative to the x-ray tube output port. B adds additional misalignment of the field of view relative to the active region of the detector. From Fig. 12, a magnetic field of $B_r = 4$ mT at the location of the x-ray tube focal spot will cause a FOV_S of 3 mm using the standard radiography x-ray tube. To ensure compliance with safety regulations and to minimize unnecessary exposure to the patient, the collimator blades can be brought in ≥ 3 mm.

Although this approach is effective in weak magnetic fields (several mT), it is unacceptable in strong B (tens of mT or stronger) since the collimator blades will need to be moved in a large distance (centimeters) to correct for the field of view shift. This will lead to a strong reduction in the field of view size available to the cardiologist, possibly omitting important cardiac anatomy visualization that would be needed for the procedure.

For interventional cardiology applications, the standard detector size is about 22 cm on a side,²⁴ so the collimator can be opened to encompass this area. If the field of view deflects beyond 2% of the source-to-image distance (about 2 cm) in B , moving the collimator blades in more than 2 cm will reduce the field of view size by ≥ 4 cm in the direction of the shift since the blades move in pairs. This can cause an unacceptable loss of clinical information in the images. Therefore, in strong B , a different technique is required to correct for field of shift without reducing the size of the field of view.

To correct for large shifts (several cm), it is possible to use a larger detector and to open up the collimator blades. However, the disadvantage of this approach is that anatomy that does not require imaging for the clinical task will be exposed to primary radiation, resulting in unnecessary exposure to the patient. Ideally, it would be beneficial to correct for field of view shift without truncating the field of view or increasing exposure to the patient.

To correct for this problem, it is possible to reposition the x-ray tube to correct for the shift instead of opening up the collimator blades. However, during an intervention, the angulation of the x-ray C-arm is constantly changed and the x-ray tube is moved into magnetic fields of different strength and orientation. Therefore, it is not practical to manually reposition the x-ray tube every time a new angulation is needed since this would increase procedure time and complexity. However, it might be possible in the future to automate repositioning of the x-ray tube based on feedback that provides information about the amount of shift to the repositioning system.

A fourth possible approach is to use a technique known as active magnetic shielding to correct for the shift. With this approach, a circuit is used to sense B and deliver an electric current to a pair of shielding coils around the x-ray tube. These coils will produce a magnetic counterfield of the same strength as B but reversed direction to negate B applied to the electron beam. This approach uses real-time feedback, so if B changes as a result of changing the position of the x-ray tube, the counterfield will automatically adjust to oppose the

new value of B . We are currently investigating this technique. It is also possible to combine these different techniques.

V.C.2. Truncation

The predicted values for x-ray field of view truncation agreed with the experimental results to within 4% in Fig. 13(c), which was within the error of the measurements. No field of view truncation was observed with the clinical collimator for $0 \leq B \leq 23$ mT. For a field of view of 18×18 cm² at the source-to-image distance shown in Fig. 5(c), S_0 was ≈ 1 cm for the standard radiography x-ray tube. According to the model in Fig. 3(b), truncation would not be observed until $e \geq S_0/2$ (5 mm for the standard radiography x-ray tube). In Fig. 8(b), e never exceeded 3 mm, so field of view truncation was not observed. Based on the results in Fig. 8(b), B of about 50 mT would be required to deflect the focal spot by 5 mm. This field strength is an order of magnitude stronger than the fringe field in the vicinity of the C-arm for this application. This x-ray tube can be placed in B of several mT and no truncation will take place.

Field of view truncation is a more serious concern in x-ray tubes with a long anode-cathode distance placed in strong B (tens of mT or stronger). This scenario could be present if a CBXMR system is developed using a weakly shielded MRI scanner, which will result in a stronger fringe field in the vicinity of the C-arm. Truncation could also be a concern if the x-ray C-arm needs to be placed closer to the scanner entrance for other clinical applications, where the fringe field is stronger.

VI. CONCLUSIONS

We have investigated the imaging performance of a rotating-anode x-ray tube in a magnetic field for CBXMR systems. Specifically, the impact of electron beam deflection (due to the magnetic fringe field) on focal spot resolving power, field of view shift, and field of view truncation was evaluated. Models were described to predict changes in focal spot resolving power, field of view shift, and field of view truncation in an external magnetic field perpendicular to the electron beam. These models were validated by experimental measurements of these effects. Although focal spot resolving power of the x-ray system will not be affected when the x-ray tube is placed in the fringe field of the CBXMR system for PAVR procedures, field of view shift will occur in the magnetic field. This shift can be corrected by further collimation of the field of view to ensure the safety of the operators and the patient in the cardiac catheterization laboratory when using CBXMR systems for PAVR procedures. If scanners with stronger fringe fields (tens of mT) are used, then additional measures may be required to correct for field of view shift, including the use of a larger detector, tube repositioning, and/or active magnetic shielding.

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