Integration of cardiac and respiratory motion into MRI roadmaps fused with x-ray

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Purpose: Volumetric roadmaps overlaid on live x-ray fluoroscopy may be used to enhance image guidance during interventional procedures. These roadmaps are often static and do not reflect cardiac or respiratory motion. In this work, the authors present a method for integrating cardiac and respiratory motion into magnetic resonance imaging (MRI)-derived roadmaps to fuse with live x-ray fluoroscopy images, and this method was tested in large animals.

Methods: Real-time MR images were used to capture cardiac and respiratory motion. Nonrigid registration was used to calculate motion fields to deform a reference end-expiration, end-diastolic image to different cardiac and respiratory phases. These motion fields were fit to separate affine motion models for the aorta and proximal right coronary artery. Under x-ray fluoroscopy, an image-based navigator and ECG signal were used as inputs to deform the roadmap for live overlay. The *in vivo* accuracy of motion correction was measured in four swine as the ventilator tidal volume was varied.

Results: Motion correction reduced the root-mean-square error between the roadmaps and manually drawn centerlines, even under high tidal volume conditions. For the aorta, the error was reduced from 2.4 ± 1.5 mm to 2.2 ± 1.5 mm ($p < 0.05$). For the proximal right coronary artery, the error was reduced from 8.8 ± 16.2 mm to 4.3 ± 5.2 mm ($p < 0.001$). Using real-time MRI and an affine motion model it is feasible to incorporate physiological cardiac and respiratory motion into MRI-derived roadmaps to provide enhanced image guidance for interventional procedures.

Conclusions: A method has been presented for creating dynamic 3D roadmaps that incorporate cardiac and respiratory motion. These roadmaps can be overlaid on live X-ray fluoroscopy to enhance image guidance for cardiac interventions. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4789919>]

Key words: x-ray, MRI, fusion, motion

I. INTRODUCTION

Most cardiovascular interventional procedures are guided by x-ray fluoroscopy. While it provides high temporal and spatial resolution, x-ray fluoroscopy provides only two-dimensional projections, and has poor soft-tissue contrast. In order to enhance image guidance for structural heart interventions, previously acquired roadmaps from other imaging modalities such as magnetic resonance imaging (MRI), computed tomography (CT), and ultrasound may be overlaid or fused with fluoroscopy images.¹⁻⁵ Acquisition and fusion of three-dimensional roadmaps through a rotational CT-like acquisition is a feature now commonly available on commercial x-ray fluoroscopy systems.^{6,7} The roadmap images are usually static volumes and do not depict cardiac or respiratory motion. Interventional targets, such as pulmonary veins, atrioventricular and semilunar valves, or cardiac septal defects, undergo significant motion during both cardiac and respiratory cycles,

and incorporation of physiologic motion into fused roadmaps may increase their utility in guiding diagnostic catheterization, structural heart interventions, and electrophysiology procedures. Use of such motion corrected roadmaps may help to increase procedure success and to reduce procedure time and overall radiation exposure.

Other investigators have studied the motion of the coronary tree using 2D biplane angiograms.^{8,9} Shechter found that while a 3D rigid body transformation was adequate for modeling the respiratory motion of the left coronary tree, an affine deformation better represented the motion of the right coronary artery. Schneider applied principal component analysis to reduce the number of parameters in a rigid 3D motion model. Several studies have focused on correcting for respiratory motion for cardiac electrophysiology procedures. Ector *et al.*¹⁰ described the significance of respiratory motion on left atrial structures, studied on CT images acquired at end-inspiration and end-expiration. He found that

the pulmonary veins and mitral valve annulus underwent pronounced motion during respiration. The CT images were not cardiac gated, however, so cardiac motion could not be considered. King *et al.*^{11,12} developed MRI models of respiratory motion incorporated into roadmaps to guide electrophysiology procedures. Respiratory motion was extracted from a single MRI volume acquired at end-expiration registered to 2D dynamic acquisitions at various respiratory phases. An affine motion model for the heart was computed, and the affine parameters were fit to polynomial functions of the diaphragm position. The model was applied by tracking the diaphragm position in the x-ray fluoroscopy imaging and updating the roadmap accordingly. Because the model was only valid at end-diastole, the x-ray images were ECG-gated. In work by Ma *et al.*¹³ the image features tracked in x-ray images (e.g., diaphragm border, heart border, tracheal bifurcation, or coronary sinus catheter) were used for simple translation of the overlaid roadmaps, but no account was made for cardiac motion. Furthermore, the motion of the heart was restricted to a global translation, which did not allow for more complex motion such as rotation or shearing. For cardiac motion correction, Gutiérrez *et al.*,³ described one example of ECG-gated image fusion with cardiac motion compensation. A separate MRI roadmap was segmented for each cardiac phase, and the ECG signal was processed under x-ray to index and display the appropriate roadmap. This required segmentation of the entire roadmap throughout the cardiac cycle (acquired at end-expiration) and did not account for respiratory motion. Brost and co-workers¹⁴ have reported combined cardiac and respiratory motion correction in overlaid roadmaps. They rely on tracking a circumferential mapping catheter when it is in firm contact with a pulmonary vein during ablation for atrial fibrillation and is not generally applicable to other interventional procedures. Metz and co-workers¹⁵ have described registration of 4D coronary CT angiograms with x-ray angiography. The accuracy is sensitive to a manual initialization step, and it is not clear how well the methods would perform in real-time during an intervention.¹⁵ A summary of previous work is shown in Table I.

The goal of this work was to incorporate both cardiac and respiratory motion into dynamic, motion-corrected 3D roadmaps to guide cardiac interventional procedures. Real-time MR images acquired in several planes were used to construct affine models for the individual cardiac structures of interest, to capture variation in motion between different parts of the heart. These models were then used to drive the motion of 3D roadmaps fused onto live x-ray fluoroscopy. *In vivo* experiments were conducted to measure how well the model reflected cardiac-respiratory motion of the aorta and proximal

FIG. 1. Flow-chart showing processing steps for (top) image acquisition and retrospective sorting according to cardio-respiratory phase; and (bottom) image warping and fit to affine model.

right coronary artery (RCA). Under x-ray fluoroscopy, target registration error was measured to determine how closely the motion-corrected contours tracked with regions of interest in real-time.

II. MATERIALS AND METHODS

II.A. MRI: Computation of affine model

An overview of the proposed strategy is shown in Fig. 1. Cardiac and respiratory motion was measured using real-time MR images acquired in multiple planes. This allowed the motion of different anatomical structures to be tracked in three-dimensions throughout the cardiac and respiratory cycles. Nonrigid registration was used to compute an image-based respiratory navigator¹⁶ that was combined with the recorded ECG signal to bin the images according to cardiac and respiratory phase. The images were then registered to the reference frame at end-expiration, end-diastole, and the resulting motion field was used to warp contours from the reference frame to each point in the cardiac-respiratory plane. Finally, these motion fields were fit to affine models.

II.A.1. Animal procedures

All animal experiments were approved by the institutional Animal Care and Use Committee and performed according to contemporary NIH guidelines. Four swine (weight range: 48–56 kg) were anesthetized with atropine, butorphanol, ketamine, and xylazine, and then maintained on isoflurane with mechanical ventilation.

II.A.2. MRI

II.A.2.a. Real-time scan for motion model. MRI was performed on a 1.5 T scanner (Espree, Siemens Medical Solutions, Erlangen, Germany). Details of the real-time image

TABLE I. Summary of previous studies in motion compensation of 3D roadmaps.

Author	Anatomy	Motion	Model	Accuracy
King (Ref. 11)	Pulmonary vein	Respiratory	Affine	2–4 mm
Ma (Ref. 13)	Pulmonary vein	Respiratory	1D or 2D translation	2 mm
Brost (Ref. 4)	Mapping catheter	Respiratory and cardiac	2D translation	2 mm
Metz (Ref. 15)	Coronary tree	Respiratory and cardiac	3D+ <i>t</i>	1–5 mm

FIG. 2. MR image showing three-chamber view of the left ventricle. Arrows point to the RCA, aorta, and diaphragm. Additional slices acquired for motion tracking are depicted as intersecting lines. The dotted line is perpendicular to the long axis of the aorta, and the dashed line intersects the short axis of the proximal RCA. All three slices intersect the diaphragm, allowing tracking of respiratory motion.

acquisition and calculation of an image-based navigator have been described previously.¹⁶ Real-time images were acquired using a steady-state free precession (SSFP) sequence with the following typical imaging parameters: TR/TE = 2.7/1.1 ms; flip angle = 80°; slice thickness = 6 mm; matrix = 192 × 93, GRAPPA rate 3 acceleration, temporal resolution = 84 ms per frame. The field of view ranged from 300 to 400 mm, depending on the size of the subject. Images were acquired free-breathing and the ECG signal was simultaneously recorded using an external monitor (InVivo Corporation, Orlando, FL). The external monitor calculated the trigger signal as the peak of the QRS complex and this signal was connected to the scanner. For each imaging plane, images were acquired continuously for 30 s in order to cover several cardiac and respiratory cycles. Three imaging planes were acquired, as shown in Fig. 2: a standard three-chamber view showing the aorta and RCA, and two slices along the long-axis of the aorta and short-axis of the RCA. Care was taken so that the diaphragm was also visible in each slice.

II.A.2.b. Marker scan for registration. The MRI and x-ray fluoroscopy images were registered using external fiducial markers, as previously described.^{3,17} Multimodality contrast-filled beads (PinPoint, Beekley Corporation, Bristol, CT) were attached to an elastic belt placed around the subject. A 3D, T1-weighted scan was used to acquire images for marker segmentation and localization. The 3D spoiled gradient-echo imaging sequence used fat saturation and the following parameters: TR/TE = 2.6/0.9 ms; flip angle = 17°; matrix: 128 × 96; slice thickness = 1.6 mm; 144 slices.

II.A.2.c. Motion model. Using the recorded ECG timing information, the real-time images were retrospectively triggered and linearly interpolated to a fixed number (20) of cardiac phases for each cardiac cycle. The number of phases was chosen empirically, so that it was sufficient to capture cardiac motion and was close to the temporal resolution of the acquisition (e.g., for a heart rate of 70 bpm, 20 phases corresponds to an interpolation factor of 2). In MRI, a special

sequence of “navigator echoes” may be acquired to monitor the motion of the diaphragm in order to trigger or gate the acquisition to the respiratory cycle. These navigator echoes are a subset of data that enable translation in the navigator read-out direction to be monitored during a scan. Acquisition of the navigator is interleaved with the image acquisition, and may be played in 1D, 2D, or 3D, depending on the type of motion to be tracked. For cardiac studies, 1D navigators are most commonly used to track the motion of the diaphragm, since the bulk of respiratory motion of the heart is in the head-foot direction. In this work, the motion of the diaphragm was tracked using an image-based navigator, which means that the motion was tracked directly from the images instead of from a separate navigator acquisition. The entire time series was registered to the first frame, and an image-based navigator was identified as shown in Fig. 3(a), usually as a pixel on the liver or diaphragm.¹⁶ The head-foot motion of this navigator was calculated, and local maxima and minima were identified and used to classify expiratory and inspiratory phases, as shown in Fig. 3(b). Using the ECG timing and respiratory navigator, the images were sorted according to cardiac and respiratory phase. The images were then interpolated onto a grid in the cardiac-respiratory phase plane, ranging from end-expiration to end-inspiration, so that there were an equal number of respiratory phases for each cardiac phase [Fig. 3(c)].

Nonrigid registration was used to compute the motion fields to transform the images at end-diastole, end-expiration, to every other point in the cardiac-respiratory space. The registration method is intensity based and maximizes the local cross-correlation between images (see Ref. 18 for details). The deformation is modeled as a smooth vector field that gives the displacement of each pixel in the reference image to match the corresponding location in the target image. The problem is formed to minimize an energy function equal to the sum of a “dissimilarity” term (between the reference and warped image) and a regularization term (which penalizes fast variations in the displacement field). The optimization uses a gradient descent algorithm. It is applied in a multiresolution approach to improve speed and likelihood of convergence. The registration process calculates motion fields automatically for every image pixel. For a particular cardiac structure, regions of interest were drawn on each imaging plane at end-diastole, end-expiration ($\mathbf{x}_{EE,ED}$). These points from all three imaging planes were warped to each respiratory-cardiac phase ($\mathbf{x}_{r,c}$) using the computed motion fields. The warping was fit to a 4×4 affine transform, $\mathbf{A}_{r,c}$ by solving in a least squares sense the following linear equation:

$$\mathbf{A}_{r,c} \mathbf{x}_{EE,ED} = \mathbf{x}_{r,c}. \quad (1)$$

A distinct affine transform was computed for each cardiac and respiratory phase, and repeated for each anatomic structure of interest. In some cases, the nonrigid registration process did not reliably propagate contours across cardiac phases, so the ROIs were manually adjusted on each cardiac phase at end-expiration and warped across respiratory phases.

II.A.2.d. MRI roadmap. As shown in Fig. 4, contours for the MRI-derived roadmaps were drawn on breath held retrospectively ECG gated SSFP images with the following

FIG. 3. Respiratory navigator on real-time MR images. (a) Position of respiratory navigator (filled white circle) on real-time images. Open circles represent the sampled grid used for warping. (b) Head-foot position of navigator, with extracted inspiratory and expiratory phases. (c) Interpolation of respiratory and cardiac phases. Acquired images (dots) are sorted according to cardiac phase and respiratory position. They are linearly interpolated to integer mm diaphragm positions, and across cardiac phases (grid). End-expiration and end-inspiration are defined as the maximum and minimum respiratory positions for which all cardiac phases have also been acquired (or interpolated). In this example, end-expiration = 0 mm and end-inspiration = -7 mm.

imaging parameters: TR/TE = 3.2/1.4 ms; flip angle = 80° ; slice thickness = 6 mm; matrix = 192×192 , GRAPPA rate 2 acceleration. For the aorta, contours were drawn on short axis images from the aortic arch to the aortic valve. The centerpoints of these contours were used for error analysis (see below). The contours may be interpolated to a surface for image fusion, as shown in Fig. 5. For the RCA, a single contour was drawn on the cine three-chamber view.

II.A.3. X-ray

II.A.3.a. Marker scan for registration. X-ray fluoroscopy was performed on a single-plane, image-intensifier based system (Axiom Artis FC, Siemens Medical Solutions, Forchheim, Germany). A set of anterior oblique projections (without craniocaudal angulation) were acquired from -30° to $+30^\circ$ in 3° increments. The 3D marker positions were identified and reconstructed in both MRI and x-ray, and the MRI

FIG. 4. Contours on MR images for (left) aorta and (right) RCA. These contours were drawn on breath held cine SSFP images and were used for the overlaid roadmaps on live x-ray.

FIG. 5. Respiratory navigator on angiography images. Aorta (surface) and proximal RCA (line) are also shown. Navigator (plus sign) tracks edge of diaphragm. End-inspiration (a) and end-expiration (b) positions (triangles) are estimated from training phase.

points were registered to the x-ray points using rigid-body (translation and rotation only) registration.^{3,17,19}

II.A.3.b. Cardiac-respiratory motion. In order to drive the motion of the MRI-derived contours under x-ray, the respiratory and cardiac phase must be detected in real-time. The respiratory-cardiac phase was used to index the appropriate set of affine parameters ($\mathbf{A}_{rc}$) for each cardiac structure, and then the new 3D coordinates of the contours were computed using Eq. (1). To measure cardiac phase, the analog ECG output of a commercial hemodynamic recording system (Cathcor, Siemens Medical Systems) was connected to a data acquisition system running LabView (National Instruments, Austin, TX). Cardiac phase was calculated as the time elapsed since the last QRS divided by the time between successive QRS complexes. The respiratory phase (r) was determined directly from the x-ray images, as shown in Fig. 5. An initial point was manually placed on the superior edge of the diaphragm, and after an initial training phase the edge was tracked in real-time and the position was calculated. The edge was tracked by filtering a region around the initial seed point with a Sobel edge operator and looking for the maximum response. For each subsequent image frame, the center of the search region was set to the location of the strongest edge in the previous frame. The training phase consisted of 1–2 respiratory cycles to measure the minimum and maximum excursion of the diaphragm as well as estimate the frequency. End-expiration ($r = 0$) and end-inspiration ($r = \pm 1$) phases were assigned to the most superior and inferior diaphragm positions, respectively. Respiratory phase was calculated as the linear distance between these two positions, with inspiratory phases having negative values and expiratory phases having positive values.

II.A.3.c. Angiography. Standard cine aortography (15 frames/s) used 50 ml of iodinated contrast power injected at 10 ml/s via a pigtail catheter positioned in the aortic root. Tidal volumes were varied to test different breathing conditions between MRI and x-ray. For each animal, three separate injections were done under different ventilator settings: breath hold (BH), low tidal volume (LOW V_T) (8 ml/kg), and high tidal volume (HIGH V_T) (10 ml/kg). The positions for end-expiration and end-inspiration were

recalculated for each ventilator setting. The tidal volumes were always set to the low tidal volume settings under MRI.

MRI derived roadmaps were projected on the live fluoroscopy images using the gantry and table positions to calculate the appropriate projection matrix.¹⁹ The matrix is the product of a perspective projection multiplied by an orientation matrix, which represents the C-arm rotation and table position. The gantry and table position were available in real-time via a dedicated network connection between the angiography system and computer performing the image fusion (see Ref. 3 for details). In the case of the RCA, which was drawn as a single contour on a single 2D plane, the x-ray projection angle was chosen to acquire a similar three-chamber view so that foreshortening of the contour was not too severe.

II.A.3.d. Data analysis. We measured accuracy by comparing motion-corrected MRI-derived contours with manual contours on the x-ray fluoroscopy images. Nine representative frames from different respiratory-cardiac phases were chosen for analysis. Centerlines were derived from manually drawn contours and MRI-derived contours of the aorta and proximal RCA, projected onto the x-ray images. The 2D root-mean-square error (RMSE) between n MRI points (m_i) and x-ray points (x_i) was calculated as

$$\text{RMSE} = \sqrt{\frac{\sum_{i=1}^n (x_i - m_i)^2}{n}}. \quad (2)$$

The RMSE was computed for both static contours (drawn at end-expiration, end-diastole, no affine motion correction applied) and motion-corrected contours, which were deformed using the computed affine transforms. To estimate the contribution of registration error, the RMSE of static contours on breath hold angiograms at end-diastole were recorded. For each ROI and ventilator setting, the errors were pooled across animals. Data are presented as median $\pm$ interquartile range. The RMSE for static and motion-corrected contours were compared using a paired, two-sided Wilcoxon signed rank test, with $p < 0.05$ considered to be statistically significant.

III. RESULTS

Plots of the affine motion components for one subject are shown in Fig. 6. The motion is relative to the position at end-expiration, end-diastole. In this figure, the effects of respiratory and cardiac motion have been separated. The respiratory phases run from end-inspiration to end-expiration. The cardiac phases range between 0 and 1 (with 0 and 1 representing successive QRS complexes). The x-y-z axes of motion are equivalent to the left-posterior-head axes for a supine subject. The figure indicates that the aorta and RCA structures exhibit little respiratory motion but considerable cardiac motion.

The centerline tracking is shown in Fig. 7. The regions of interest for the aorta and RCA are shown in the top row, for context. The bottom row shows tracking results for the aorta (d)–(e) and RCA (f)–(g). In this example, the ventilator was set to the low tidal volume setting. The respiratory and cardiac phase is shown at the top of each frame. The “true” centerline, calculated from manually drawn contours of vessel borders

FIG. 6. Affine transform components for aorta (left) and proximal RCA (right) for one subject. From top to bottom, plots are of scale, shear, rotation, and translation. For respiratory motion (first and third columns), cardiac phase is fixed to end-diastole, and for cardiac motion (second and fourth columns) respiratory phase is fixed to end-expiration. For both aorta and RCA, the magnitude of cardiac motion is larger than respiratory motion.

on x-ray images, is shown as a solid white line. The static MRI contour is shown as open white circles, and the motion-corrected MRI contour is shown as solid white circles. For the aorta, the errors were small for both the motion-corrected and static contours, and they both are similar to the manually drawn centerline. The difference was more significant for the RCA, which moved more during through the cardiac cycle. The errors are most apparent during systole, when the RCA moves most. There was still some residual error in the motion-corrected contours for both structures.

The RMSE across all animals is listed in Table II and graphed in Fig. 8. The estimated registration error for the aorta was 1.6 ± 1.0 mm, and for the RCA was 1.1 ± 1.0 mm. For the aorta, the static and dynamic, motion-corrected errors steadily increased with tidal volume. The difference in RMSE between static and motion-corrected contours reached statistical significance in all cases except for the low tidal volume setting. For the RCA, the differences between the static and motion-corrected errors were larger, and the differences reached statistical significance at all three ventilator settings.

IV. DISCUSSION

This work presents a method for integrating cardiac and respiratory motion into MRI-derived roadmaps fused with live x-ray fluoroscopy. Anatomic localization is critical for minimally invasive structural heart procedures (e.g., annular plane or coronary artery position for percutaneous aortic valve implantation²⁰) and electrophysiology procedures (e.g., pulmonary vein isolation). Fusion of dynamic soft tissue imaging

with fluoroscopy may make these interventions safer, easier, and more effective. The novel contribution of this work is the modeling and correction of cardiac and respiratory motion simultaneously using a simple affine model.

FIG. 7. Centerline tracking in the aorta and proximal RCA. Top row shows regions of magnification for context. (a)–(c) Aortogram indicating regions of interest. (d)–(g) show overlaid centerlines. The respiratory phase (R) and cardiac phase (C) are shown at the top of each frame. Respiratory phases are from 0 (end-expiration) to ± 1 (end inspiration), with $(-/+)$ indicating inspiratory/expiratory phases, respectively. Cardiac phases are from 0 to 1. The manually drawn centerline is drawn as a solid white line, the motion-corrected centerline with filled white circles, and the static centerline with open white circles. During systole, (d) and (f), the motion-corrected centerline is closer to the manual contour than the static centerline. At end diastole, (e) and (g), the centerlines overlaid one another, as expected. The differences between the static and motion-corrected contours for the aorta were less compared with differences for the RCA.

TABLE II. Errors in mm between manually drawn centerlines on x-ray and MR-derived centerlines. Root-mean-square error $\pm$ interquartile range is shown. Values are shown for breath hold, Low V_T , and Hight V_T ventilator settings.

Structure	Breath hold			Low V_T			High V_T		
	Static	Dynamic	p -value	Static	Dynamic	p -value	Static	Dynamic	p -value
Aorta	2.0 ± 1.4	1.6 ± 1.0	0.004	2.2 ± 1.5	1.8 ± 1.3	0.08	2.4 ± 2.0	2.2 ± 1.5	0.048
RCA	8.1 ± 11.2	4.5 ± 5.0	<0.001	7.1 ± 8.1	4.6 ± 2.7	<0.001	8.8 ± 16.2	4.3 ± 5.2	<0.001

In this study, the motion was measured from free-breathing real-time MR images to model cardiac and respiratory motion simultaneously. Complex motion of different cardiac structures was reduced into simple models through separate affine transforms. The driving parameters for the model were the ECG waveform and diaphragm position. ECG is routinely measured during both MRI and x-ray. One potential source of error is the lag between x-ray image acquisition and recording of the ECG waveform. Further investigation is needed to quantify this lag and evaluate its significance. In this study, the diaphragm position was extracted directly from images; alternatively it could be tracked using bellows or a capnograph.

We validated our method in an *in vivo* animal study by tracking the motion of animal aorta and proximal RCA under different ventilator tidal volume settings. Motion-correction reduced the roadmap registration error in all cases. The improvements were greater for the proximal RCA than for the aorta. In this pilot study, the observed effects of respiratory motion were relatively small, compared with cardiac motion. This is consistent with the results obtained by Shechter et al.,²¹ who observed a greater 3D cardiac (14.4 mm) than respiratory motion (5.0 mm) of the RCA origin in free-breathing adults. We expect cardiac and respiratory motion to be more dramatic in planned clinical interventional targets, such as muscular ventricular septal defect.

We tested our approach only in animals during mechanical ventilation. Pediatric interventional cardiology procedures usually are performed using mechanical ventilation. The primary limitation of this animal model is that the magnitude of respiration is small and is regular, especially compared to free-breathing human subjects. An attempt was made to model variable breathing by adjusting the end tidal volume settings on the ventilator, but it is likely that the clinical scenario will be much more variable. We plan to validate our approach in free-breathing adults in an upcoming clinical research protocol, and will specifically assess error related to different respiratory patterns in MR and x-ray.

In this study, the motion was measured from multiple 2D MRI slices. For structures that undergo a large degree of through-plane motion, one could explore techniques such as slice tracking²² or methods from the tagging literature^{23,24} to measure 3D motion more accurately. Through-plane motion may also be a consideration when tracking the diaphragm in oblique slices, and alternatives such as respiratory navigators²⁵ will be investigated in future studies.

Future technical developments include MR imaging and reconstruction strategies focused on improving the motion measurements. In order to improve the workflow of the method, the extraction of the motion field may be integrated into the MRI reconstruction process.²⁶ The motion field could

FIG. 8. Root-mean-square error for (a) aorta and (b) right coronary artery. Values are shown for BH, LOW V_T , and HIGH V_T settings. Bar height is median value, and error bar is interquartile range. Static errors (black bars) are compared with motion-corrected errors (white bars). Asterisks (*) indicate statistically significant difference between static and motion-corrected errors

then be used directly to build the affine motion models. We also aim to conduct clinical studies to study how the motion-corrected roadmaps affect overall procedure time and radiation dose for actual interventional procedures.

V. CONCLUSIONS

We have presented a method for incorporating cardiac and respiratory motion into MRI derived roadmaps that can be fused with live x-ray fluoroscopy to improve image guidance for cardiovascular interventional procedures. We demonstrated that affine motion models outperformed static roadmaps in tracking the motion of the aorta and proximal right coronary artery *in vivo*.

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