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Respiratory motion correction for image-guided cardiac interventions using 3-D echocardiography

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Abstract

In this paper we investigate the use of 3-D echocardiography (echo) data for respiratory motion correction of roadmaps in image-guided cardiac interventions. This is made possible by tracking and calibrating the echo probe and registering it to the roadmap coordinate system. We compare two techniques. The first uses only echo-echo registration to predict a motion correction transformation in roadmap coordinates. The second combines echo-echo registration with a model of the respiratory motion of the heart. Using experiments with cardiac MRI and 3-D echo data acquired from eight volunteers, we demonstrate that the second technique is more robust than the first, resulting in motion-correction transformations that were accurate to within 5mm in 60% of cases, compared to 42% for the echo only technique, based on subjective visual assessments. Objective validation showed that the model-based technique had an accuracy of 3.3 ± 1.1 mm, compared to 4.1 ± 2.2 mm for the echo only technique. The greater errors of the echo only technique were mostly found away from the area of echo coverage. The model-based technique was more robust away from this area, and also has significant benefits in terms of computational cost.

Key words: echocardiography, respiratory motion, modelling, MRI

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1 Introduction

The utility and effectiveness of image-guided interventions in the chest and abdomen is currently limited by the problem of respiratory motion. Organs such as the heart can undergo motion of up to 25mm (McLeish et al., 2002) during breathing, causing a misalignment between the static image-derived roadmap and the underlying anatomy. If the organ can be imaged directly using a real-time intraprocedure imaging modality such as echo or X-ray, this motion can be tracked and corrected for. 2-D imaging modalities such as B-mode echo do not provide enough information to compute 3-D motion estimates without some further constraint on the transformation space (e.g. Blackall et al. (2005); Wein et al. (2008)). However, recent advances in 3-D echo imaging and echo-echo similarity measures (Grau et al., 2007) have opened up the possibility of using 3-D echo images to estimate organ motion. We have previously published a preliminary investigation into this possibility (King et al., 2008b). In this paper we perform a more thorough investigation into the use of 3-D echo data for estimating cardiac respiratory motion.

We investigate two possible techniques. The first uses 3-D echo images alone to estimate the motion. The second combines the 3-D echo image data with a previously formed respiratory motion model. The use of respiratory motion models has been demonstrated for a range of image-guided interventions. For example, in Shechter et al. (2005b) a model of cardiac and respiratory motion was formed from biplane contrast-enhanced X-ray images, and used to motion-correct subsequent X-ray images during coronary interventions. In King et al. (2008a, 2009a) a respiratory motion model derived from magnetic resonance imaging (MRI) was used to motion-correct a roadmap during image-guided cardiac catheterisations. In McClelland et al. (2006) a CT-derived roadmap was used for lung radiotherapy. All of this work involved acquiring a surrogate of the true motion whilst acquiring the images to form the motion model, and acquiring the same, or a related, surrogate during the intervention. The motion model estimates the true motion based on the surrogate value. Common surrogates of respiratory motion include the 1-D superio-inferior translation of the diaphragm (Shechter et al., 2005b; King et al., 2008a, 2009a) and the motion of surface points, such as the chest wall (McClelland et al., 2006). Similar models have also been applied to motion-correct image acquisition such as MRI (Manke et al., 2002, 2003; Nehrke and Bornert, 2005).

However, motion estimation using models that feature surrogates is made less accurate by errors and difficulties in measuring the surrogate value. In addition there is sometimes uncertainty that the surrogates being acquired, to form and apply the model, are actually the same. For example, when tracking diaphragm or chest surface motion it can be difficult to make sure that the same physical point is being tracked. This difficulty can be avoided by combining intra-

procedure image data with the motion model. The value of the respiratory surrogate used to form the model is varied to optimise the similarity measure based on the image data, eliminating the need to acquire a surrogate during the intervention. For example, in Blackall et al. (2005) 2-D echo image data was combined with a motion model to estimate liver motion. In this paper we investigate the possibility of using 3-D echo image data to produce cardiac respiratory motion estimates. We investigate whether the echo images alone can provide accurate enough motion estimates or if it is necessary to combine the image data with a motion model.

2 Method and materials

2.1 Method

Our respiratory motion correction approach involves two coordinate systems: those of the MRI and 3-D echo images (see Figure 1). These are related by the known MRI-echo registration, R (see Section 2.2). The roadmap for the image-guided intervention is formed from MRI data. Therefore the motion correction transformation should be applied in this coordinate system. However, we compute the transformation from exhale to the current respiratory position in 3-D echo coordinates. The motion correction transformation in MRI coordinates, S_M , is computed from the transformation in echo coordinates, S_E , as follows,

$$S_M = R^{-1}S_ER. \quad (1)$$

where R is the MRI to echo registration.

Fig. 1. Computing the MRI motion correction transformation using an echo-echo transformation. Breath-hold 3-D echo images are acquired at exhale and the current inhale position. These images are registered, resulting in an echo-echo transformation, S_E . This is combined with the known MRI-echo registration at exhale, R , to compute the MRI-MRI motion correction transformation, S_M .

2.1.1 Technique 1 - echo similarity measure only

We test two techniques for computing the echo-echo transformation, S_E . Both use the echo similarity measure recently proposed by Grau et al. (2007), which quantifies the similarity between echo images based on local orientation and phase differences. The first technique uses 3-D echo image data only and optimises the similarity measure by allowing variation in the six rigid-body transformation parameters. The motion correction transformation, S_E , is estimated as

$$S_E = A_{\tilde{\phi}} \quad (2)$$

where

$$\tilde{\phi} = \underset{\phi}{\operatorname{argmax}} \quad \operatorname{Sim}(A_{\phi}(Echo_{exhale}), Echo_{inhale}) \quad (3)$$

In (2) and (3), $\phi = (T_x, T_y, T_z, \theta_x, \theta_y, \theta_z)$ is a vector containing the estimates of the rigid-body motion parameters, A_{ϕ} is the matrix transformation produced from these motion parameters, $A_{\phi}(\cdot)$ is a function that transforms the image data by the matrix transformation and Sim is the similarity measure (Grau et al., 2007). $Echo_{exhale}$ and $Echo_{inhale}$ are the exhale and inhale 3-D echo images respectively.

2.1.2 Technique 2 - echo similarity measure with motion model

The second technique also optimises the phase-based similarity measure, but constrains the range of permitted transformations using a MRI-derived respiratory motion model.

The motion model is constructed from a short dynamic MRI scan and a high resolution exhale MRI volume. The dynamic acquisitions were acquired at arbitrary respiratory positions during free-breathing but gated to the late diastolic period of the cardiac cycle (see Section 2.2.1). A pencil-beam navigator was applied through the dome of the right hemi-diaphragm to track the inferio-superior translation of the diaphragm. Each dynamic acquisition was automatically registered to the high resolution exhale MRI volume using an affine intensity-based registration algorithm that seeks to maximise the normalised mutual information between the two images. The similarity measure was computed over a manually defined elliptical region of interest covering the four chambers of the heart and the major vessels. The twelve affine motion parameters resulting from the registrations were modelled as polynomial functions of the MRI navigator value. This model is based on that described in King et al. (2008a, 2009a). The model we used in this paper was a slightly simplified version in which the inspiration and expiration phases were not modelled separately, and only first order polynomial functions (i.e. linear fit) were used. This modification was made to simplify the optimisation of the similarity measure. Model formation is automatic and takes about 45 minutes

per model on a Pentium 4 2.16GHz PC

This motion model can predict the affine motion of the heart given the 1-D inferio-superior translation of the diaphragm. Therefore, when optimising the similarity measure only a single term, the diaphragm translation, V , is varied. The motion correction transformation, S_E , is estimated as

$$S_E = M_{\tilde{V}} \quad (4)$$

where

$$\tilde{V} = \underset{V}{\operatorname{argmax}} \quad \operatorname{Sim}(M_V(Echo_{exhale}), Echo_{inhale}) \quad (5)$$

In (4) and (5), M denotes a motion model, V a diaphragm translation value, M_V the affine transformation produced by applying navigator value V to motion model M and $M_V(\cdot)$ is a function that transforms the image data by this transformation.

Note that although the model is formed using the diaphragm translation as a respiratory surrogate, it is not necessary to acquire the surrogate to apply the model. The surrogate value is altered to optimise the value of the similarity measure. If the original version of the motion model were used, in which the inspiration and expiration phases are separated, it would be necessary to optimise two parameters: the navigator value and the breathing direction (i.e. inspiration/expiration).

2.2 Materials

Our main clinical application is image-guided cardiac catheterisations using the X-ray/MR (XMR) image-guidance system described in Rhode et al. (2005). This system enables a roadmap derived from MRI data to be fused with real-time fluoroscopic X-ray images to assist in guiding cardiac catheterisation procedures. The MRI image to X-ray transformation is known through a combination of system calibration and tracking. In our experiments we do not use the X-ray imaging functionality of the system, but utilise the MRI image to patient transformation only. All of our experiments were carried out in an XMR catheterisation suite, featuring a 1.5 Tesla cylindrical bore Philips Achieva MRI scanner and an Optotrak 3020 optical tracking system (Northern Digital Inc.) for tracking the patient bed.

For 3-D echo acquisition we used an iE33 3-D real-time echocardiography system with a X3-1 3 to 1 MHz broadband matrix array transducer (Philips Healthcare). Infrared light emitting diodes (LEDs) were attached to the echo probe to enable it to be tracked using the Optotrak tracking system and registered to the physical space of the XMR suite. The probe was calibrated using the method described in Ma et al. (2008).

Figure 2 illustrates the coordinate systems involved in the experiments. The MRI-echo registration, R , consists of three transformations. $T_{M \rightarrow P}$ is the MRI-to-patient transformation resulting from tracking of the patient bed and calibration of the XMR image guidance system. $T_{B \rightarrow P}$ is the probe-to-patient transformation, which is found by tracking the echo probe and the patient bed. $T_{E \rightarrow B}$ is the echo-image-to-probe transformation, which is the result of calibrating the echo probe.

Fig. 2. Computing the MRI-echo registration, R . The overall registration consists of the MRI to patient transformation, $T_{M \rightarrow P}$; the echo probe and patient tracking transformation, $T_{B \rightarrow P}$; and the echo probe calibration transformation, $T_{E \rightarrow B}$.

2.2.1 MRI data acquisition

MRI data is used to form the motion model described in Section 2.1.2 and also for validation purposes. Two MRI sequences were used to acquire the images to form the motion model:

- High resolution volume: 3-D balanced TFE scan, respiratory gated at exhale, cardiac triggered and gated at late diastole, typically, 120 slices, $TR=4.4ms$, $TE=2.2ms$, flip angle= 90° , acquired voxel size $2.19 \times 2.19 \times 2.74mm^3$, acquired matrix size 160×120 , reconstructed voxel size $1.37 \times 1.37 \times 1.37mm^3$, reconstructed matrix size 256×256 , acquisition window $\approx 100ms$, navigator window 5mm, scan time approximately 5 minutes.
- Dynamic scan: 3-D TFEPI, cardiac triggered and gated at late diastole, typically, 120 dynamics, 20 slices, $TR = 10ms$, $TE = 4.9ms$, flip angle = 20° , acquired voxel size $2.7 \times 3.6 \times 8.0mm^3$, acquired matrix size 128×77 , reconstructed voxel size $2.22 \times 2.22 \times 4.0mm^3$, reconstructed matrix size 144×144 , TFE factor 26, EPI factor 13, TFE acquisition time 267.9ms, scan time approximately 100 seconds.

The high resolution volume is also used to form the roadmap. During the dynamic scan a pencil-beam navigator was applied through the dome of the right hemi-diaphragm immediately before and after each acquisition. These lead and trail navigator values were averaged to determine a single navigator value for each dynamic acquisition. Subjects were instructed to perform normal, fast and then deep breathing during the dynamic scan to ensure that the resulting motion model was applicable in a range of different breathing

patterns.

For validation purposes (see Section 2.3) five breath-hold MRI images were acquired at normal inhale:

- Breath-hold volume: 3-D balanced TFE scan, cardiac triggered and gated at late diastole, typically, 24 sagittal slices, $TR = 4.7\text{ms}$, $TE = 2.3\text{ms}$, flip angle $= 90^\circ$, acquired voxel size $1.56 \times 1.56 \times 12.0\text{mm}^3$, acquired matrix size 160×160 , reconstructed voxel size $0.97 \times 0.97 \times 6.0\text{mm}^3$, reconstructed matrix size 256×256 , acquisition window $\approx 100\text{ms}$, scan time approximately 20 seconds.

Note that these breath-hold images were used only for validation purposes and were independent from the images used to compute the motion correction transformations, although all images were acquired in the same session. We refer to these breath-hold images as validation MRI images.

2.2.2 Echocardiography data acquisition

3-D echo images were acquired at exhale and normal inhale. For the inhale images, volunteers were instructed to hold their breath at the same inhale position that the breath-hold MRI scans were acquired at. The 3-D echo images were acquired by an experienced echocardiographer from a modified parasternal long-axis view mostly covering the left ventricle and left ventricular in- and out-flow tracts. The 3-D echo images were cardiac gated at late diastole by synchronising image acquisition with the signal from the electrocardiogram.

2.3 Experiments

To validate the two proposed motion correction techniques we used the validation MRI images described in Section 2.2.1. To compute the motion correction transformations, first a motion model was formed from the high resolution exhale MRI volume and the dynamic MRI scan. Next, exhale and inhale echo images were acquired and used to compute motion correction transformations using the two proposed techniques (i.e. using echo images only and combining the echo images with the motion model). The results were used to warp the exhale MRI image to inhale, which was then compared with one of the validation MRI images. This validation approach makes the assumption that the inhale breath-hold positions for the MRI and echo image acquisition were the same. We used experienced volunteers who were instructed to hold their breath in the same position each time. To validate their ability to do this accurately we acquired five separate validation MRI images and assessed their variability. For both the breath-hold repeatability and the respiratory motion

correction experiments we performed subjective and objective validation.

The following experiments were performed on eight healthy volunteers who all gave informed consent to the procedure.

2.3.1 Breath-hold repeatability

To subjectively test the volunteers' ability to breath-hold repeatably at the same inhale position, seven anatomical landmarks were manually localised in the five validation MRI images. Mean locations were computed for each landmark. The root mean square (RMS) distance away from the mean locations of the landmarks was computed. The seven landmarks used were the left lower and upper pulmonary vein ostia, the right upper pulmonary vein ostium, the junction between the inferior vena cava and the right atrium, the tricuspid valve, the anterior mitral valve and the junction between the left ventricle and the ascending aorta. Where possible the centre of each anatomical feature was localised.

Objective validation was performed by automatically segmenting the four chambers of the heart in all five validation MRI images using the technique described in Zhuang et al. (2008). All possible pairings of the five segmented images were compared. Two different comparisons were made between each pair of images. First, for each chamber we determined the total number of voxels and the number of overlapping voxels between the two segmented images and computed the overall Dice coefficient (i.e. using the statistics from all four chambers) between the two images. Second, we extracted surfaces for each chamber from the segmented images using the marching cubes algorithm implemented in VTK (Schroeder et al., 1997). For each chamber, the distances between each point in the first surface to the nearest point in the second surface were computed. These values were combined to determine an overall RMS point distance between each pair of segmented images. Note that we use the nearest point because we do not know the anatomically corresponding point. This measure will always underestimate the error as the nearest point may be closer than the anatomically corresponding point. The mean/standard deviation over all image pairs were computed for both the Dice coefficients and the point distances.

2.3.2 Motion-correction validation

To validate the performance of the two motion correction techniques, we used one of the validation MRI images described above. For each volunteer, the image in which the anatomical landmarks were closest to their mean locations was selected for this purpose.

Two experiments were carried out for each volunteer. For each experiment, we acquired one 3-D echo image at exhale and one at normal inhale. The echo images were acquired as described in Section 2.2.2. Registrations, S_E , were computed using techniques 1 and 2. The results were used to compute MRI motion correction transformations, S_M , according to (1). These transformations were used to warp the high resolution exhale MRI volume to the inhale position. The warped image was compared with the validation MRI image.

A subjective comparison was made by visually assessing the error between the warped image and the validation image. We do not believe it was reasonable to accurately assess errors in the through-plane direction (i.e. medio-lateral) because of the large slice thickness of the breath-hold MRI images (12mm acquired slice thickness). Therefore error assessments were made only in the sagittal plane. Errors were assessed separately for each of the four chambers of the heart and observers were asked to grade according to the maximum misalignment at any point in the chamber. Three observers independently graded the alignment of the warped images and the validation MRI image according to the following scale:

- 1: maximum error less than 2mm
- 2: maximum error between 2mm and 5mm
- 3: maximum error between 5mm and 10mm
- 4: maximum error greater than 10mm

All assessment was 'blinded', i.e. the observers were presented with the image pairs in a random order and not told which technique they were produced with.

Objective validation of the motion-correction experiments was performed in a similar way to that in the breath-hold repeatability experiments. For each experiment, the warped MRI image was automatically segmented and compared to the segmented validation MRI image. The Dice coefficients and surface point distances were computed as described in Section 2.3.1.

2.3.3 *Rigid vs. affine model comparison*

Our final experiment compared the motion-correction performance of the model-based approach using an affine or a rigid motion model. The motion-correction experiments described in Section 2.3.2 compare a rigid echo-only technique with an affine model-based technique. To assess whether any differences in performance between these two approaches are because of the greater number of degrees of freedom in an affine motion description, we repeated the objective validation experiments using only the rigid component of the motion model (i.e. all scaling parameters were fixed to 1.0 and all skew parameters were fixed to 0.0).

3 Results

3.1 Breath-hold repeatability

Table 1 shows the subjective and objective breath-hold repeatability errors. The figures obtained from subjective and objective validation are in general agreement. All volunteers apart from volunteer G were able to hold their breath repeatably with an accuracy of 2.1mm or less. No objective validation was possible for volunteer H because visual inspection of the segmentations of the validation MRI images showed that the automatic segmentation algorithm had failed. This was probably due to the deeper inhale breath-hold position of this volunteer.

Subject	Subjective Validation	Objective Validation	
	RMS landmark error (mm)	Dice coefficient, mean +/- s.d.	Surface point distance (mm), mean +/- s.d.
Vol. A	1.1	0.91 +/- 0.03	1.4 +/- 0.5
Vol. B	0.6	0.94 +/- 0.01	1.1 +/- 0.1
Vol. C	2.1	0.88 +/- 0.04	1.9 +/- 0.7
Vol. D	1.5	0.9 +/- 0.05	1.6 +/- 0.7
Vol. E	0.9	0.9 +/- 0.04	1.7 +/- 0.6
Vol. F	0.6	0.94 +/- 0.01	1.2 +/- 0.3
Vol. G	3.8	0.8 +/- 0.09	3.2 +/- 1.6
Vol. H	1.1	*	*

Table 1

Subjective and objective breath-hold repeatability results. For each volunteer, subjective validation was performed by localising seven anatomical landmarks in five validation MRI images. The RMS distance from the mean locations was computed. Objective validation was performed by automatically segmenting the validation MRI images and computing the Dice coefficients and distances between the extracted surfaces.

* The automatic segmentation algorithm failed for this volunteer's breath-hold images so no objective validation was performed.

3.2 Motion correction

3.2.1 Subjective validation

The subjective visual assessment results are summarised in Figure 3. To give an indication of the difference between the inhale and exhale states, observers were also asked to assess the alignment between the exhale MRI volume and the validation inhale MRI volume. Our main clinical application is cardiac catheterisations, for many of which the accuracy requirement is 5mm. This figure is based on the size of the pulmonary veins, which are one of the smallest structures likely to be encountered. These are normally at least 10mm in diameter for most adults (Anselme et al., 2006). This means that a grading of 1 or 2 is within this accuracy requirement.

The field of view of the echo images mostly covered the left ventricle. Therefore it is interesting to assess whether there was any difference in the accuracy of the registrations within the left ventricle and within the other three chambers. Table 2 shows the percentage of the gradings that were graded at either 1 or 2 (i.e. within the clinical accuracy requirement) for the left ventricle, the other three chambers and over all four chambers. Overall the model-based technique performs better than the echo only technique (61% compared to 42%). However, in the left ventricle the results are similar (63% compared to 58%). The model-based technique performs much better in the other three chambers (60% compared to 37%). This suggests that the echo only technique performs reasonably in the area of coverage of the echo images, but the registration outside of this area can be poor. The model-based technique, on the other hand, constrains the registration to physiologically 'realistic' motions, making the registration much more accurate outside of the echo field of view.

Figure 4 illustrates sample visual assessments carried out by one observer. Figures 4(a)-(d) show the comparison without motion correction and Figures 4(e)-(f) show the same comparison after motion correction using the model-based technique. For each of the four chambers of the heart, the figure shows sagittal slices from the validation MRI image (in grey) fused with the test image (in green). For the top row (i.e. without motion correction), the test image is the original high resolution exhale MRI image. For the bottom row (i.e. with model-based motion correction) the test image is the high resolution exhale MRI image warped to inhale using the model-based registration result. In each case the cursor and reformatted sections are positioned to illustrate the maximum estimated misregistration.

Fig. 3. Visual assessment of motion correction transformations using a validation MRI image. For each experiment, the echo-only and echo+model approaches were tested. The test transformation was used to warp an exhale MRI image which was then visually compared with the validation MRI image. Comparisons were made at the boundaries of each of the four chambers of the heart. Grading system: 1 = error < 2mm; 2 = error between 2mm and 5mm; 3 = error between 5mm and 10mm; 4 = error > 10mm. The charts show the number of gradings (1-4) made for each of the techniques. For comparison the observers also graded the alignment before any motion correction was applied. The figures show the aggregate results over the three observers, separately for each chamber and also the combined overall statistics.

3.2.2 Objective validation

A summary of the objective validation results for volunteers A-G is shown in Table 3. No objective validation was possible for volunteer H because of the failure of the automatic segmentation algorithm. A comparison of Tables 2 and 3 shows that the subjective and objective validation results suggest the same conclusions. Compared to the model-based technique the echo only technique seems to perform reasonably well in the area of echo coverage (in the

Region	Registration	Percentage of gradings at 1 or 2 (i.e. within clinical accuracy requirement)
Left ventricle	Before	0%
	Echo only	58%
	Echo + model	63%
Other 3 chambers	Before	2%
	Echo only	37%
	Echo + model	60%
Overall (i.e. 4 chambers)	Before	1%
	Echo only	42%
	Echo + model	61%

Table 2

Percentage of gradings that were within the clinical accuracy requirement of 5mm before motion correction, and after motion correction using the two proposed techniques. The results are shown for the left ventricle (i.e. within the echo image field of view), for the other three chambers, and over all four chambers. Compared to the echo only technique, the model based technique has much improved results outside the echo field of view, suggesting that it constrains the registration result to physiologically 'realistic' motions.

left ventricle there was a point distance of $3.4 \pm 1.2\text{mm}$ compared to $3.4 \pm 1.3\text{mm}$, and Dice coefficients of 0.78 ± 0.07 compared to 0.81 ± 0.06 . However, in the other three chambers the echo only technique performs worse (point distances of $4.3 \pm 2.3\text{mm}$ compared to $3.3 \pm 1.1\text{mm}$ and Dice coefficients of 0.65 ± 0.15 compared to 0.73 ± 0.11). In a two-tailed paired t-test both measures showed a statistically significant improvement ($p < 0.05$) for the model-based approach over the echo only approach in the other three chambers and also over all four chambers. No statistically significant difference ($p > 0.05$) was found in the left ventricle.

3.3 Rigid vs. affine model comparison

Table 4 shows the objective results for the model-based approach using the rigid component of the motion model. Both the Dice coefficient and the surface point distance measures give comparable results to those shown in Table 3 for the affine model-based approach. In a two-tailed paired t-test no statistically significant difference ($p > 0.05$) was found between the rigid and affine model-based approaches for both the Dice coefficients and the surface point distance measures. This demonstrates that the better performance of the model-based

Fig. 4. Visual assessment of registration results in the sagittal plane at the boundaries of: (a)(e) left ventricle; (b),(f) left atrium; (c),(g) right ventricle; (d),(h) right atrium. The top row shows sample images before motion correction, and the bottom row shows the same images after motion correction using the model-based technique. The grey image is the validation MRI image and the green image is the test image. In each case the cursor and reformatted sections are positioned to illustrate the maximum estimated misregistration.

approach seen from the results in Section 3.2 is mainly because of the greater constraints on likely transformations that the motion model provides, rather than the higher degrees of freedom in an affine transformation compared to a rigid one.

4 Discussion and conclusions

We have presented an investigation into techniques for respiratory motion correction of roadmaps for image-guided cardiac interventions using 3-D echocardiography. Although our main clinical application is image-guided cardiac catheterisations in an XMR suite, the methods we describe have potential application in a range of interventions.

We validated our technique on 8 volunteer datasets. Acquiring these datasets was a complex procedure and required input from a range of clinical, scientific and technical staff, both during data acquisition and postprocessing. This precluded our being able to acquire large numbers of datasets on a routine basis. Nonetheless, the data enabled us to draw some interesting conclusions regarding the use of 3-D echocardiography for cardiac respiratory motion cor-

Region	Registration	Dice coefficients, mean +/- s.d.	Surface point distances, mean +/- s.d.
Left ventricle	Before	0.65 +/- 0.11	7.3 +/- 2.8
	Echo only	0.78 +/- 0.07	3.4 +/- 1.2
	Echo + model	0.81 +/- 0.06	3.4 +/- 1.3
Other 3 chambers	Before	0.62 +/- 0.11	5.8 +/- 1.7
	Echo only	0.65 +/- 0.15	4.3 +/- 2.3
	Echo + model	0.73 +/- 0.11	3.3 +/- 1.1
Overall (i.e. 4 chambers)	Before	0.62 +/- 0.11	6.2 +/- 2.1
	Echo only	0.68 +/- 0.15	4.1 +/- 2.2
	Echo + model	0.75 +/- 0.11	3.3 +/- 1.1

Table 3

Objective assessment of respiratory motion correction predictions - summary of results over all experiments for volunteers A-G*. Results are computed before motion correction, and after motion correction using the two proposed techniques. The Dice coefficient values represent the proportion of overlapping voxels between the segmented validation MRI image and the segmented high resolution exhale MRI image warped with the test transform. The surface distance values represent the distances between surfaces extracted from the same two segmented images.

* No data was available for volunteer H because the automatic segmentations failed.

Region	Dice coefficients, mean +/- s.d.	Surface point distances, mean +/- s.d.
Left ventricle	0.81 +/- 0.06	3.5 +/- 1.2
Other 3 chambers	0.73 +/- 0.11	3.2 +/- 0.8
Overall (i.e. 4 chambers)	0.75 +/- 0.11	3.3 +/- 0.9

Table 4

Summary of objective assessment of respiratory motion correction predictions for a rigid model-based technique. These results can be compared directly with the affine model-based results shown in Table 3 (the rows where the registration is 'Echo + model'). The figures are comparable and show a similar improvement over the echo only approach in areas away from the left ventricle, demonstrating that this improvement is because of the greater constraints introduced by the motion model rather than the higher degrees of freedom of an affine motion description.

rection. The results presented in Figure 3 and Tables 2 and 3 suggest that, in the heart, using the echo only registration technique may not be robust or accurate enough to produce reliable respiratory motion correction transformations using 3-D echo images. The echo only technique resulted in motion correction transformations within our clinical accuracy requirement of 5mm

in only 42% of cases, compared to 61% for the model-based approach. The reason for the poor performance of the echo only technique appears to be less accurate motion predictions in areas of the heart that were not covered by the 3-D echo images. The echocardiographer sometimes found it difficult to acquire good quality images, especially at inhale, because of a number of factors: some commonly used acoustic windows were compromised at inhale in some subjects; the subjects had to remain supine during echo acquisition to avoid invalidating the MRI to patient registration, $T_{M \rightarrow P}$ (routine diagnostic echo usually involves positioning the subject on their left side); and optically tracking the echo probe required line of sight between the probe LEDs and the tracking cameras, further restricting the range of possible movement. The fact that a relatively small area of the image contained strong echo signals, mostly in the left ventricle, meant that the echo-only registration sometimes contained significant rotational errors, causing large misalignments away from this area. These errors were much less apparent in the left ventricle itself. However, the constraints on the allowable range of transformations imposed by the motion model seemed to overcome this problem in most cases. Many cardiac catheterisation procedures, such as radio-frequency ablation of the right and/or left atria, take place outside of the left ventricle. Therefore an accurate registration is essential in these areas.

We used an accuracy requirement of 5mm which was based on the sizes of one of the smallest structures likely to be encountered during a cardiac catheterisation procedure (the pulmonary veins). However, there number of different types of cardiac catheterisation procedure, including radio-frequency ablations, pulmonary vascular resistance studies and pacing studies, each of which has different guidance requirements. For example, ablation procedures are commonly used to treat cardiac arrhythmia (Huang and Wood, 2006). Guidance technology might be employed to guide the ablation catheter to its target(s) and to record ablation points for subsequent review by the cardiologist. For most atrial ablations an accuracy of 5mm is sufficient. However, for many ventricular ablations the source of the arrhythmia is much more localised and greater accuracy is necessary. Similarly, when making electrical measurements in a pacing study, greater accuracy may offer clinical benefit. Therefore accuracy requirements depend to a large degree on the type of intervention and there is no single figure that is valid for all types of intervention. Our figure of 5mm is applicable for general navigation purposes. However, it should be remembered that further increases in accuracy beyond this requirement would open up a wider range of clinical applications than is currently possible in image-guided cardiac catheterisations.

Previous work has highlighted the differences between the respiratory motion of different regions of the heart. For example, Shechter et al. (2005a) reported that the motion of the left coronary artery was largely rigid-body, whereas that of the right coronary artery was better represented by an affine trans-

form. Therefore, one possible cause for the echo only technique performing worse away from the left ventricle is that the motion on the right side of the heart has a larger affine component. However, in our case we believe that the effect of these differences is small compared to the sometimes large rotational errors introduced in the echo only registration away from the area of echo coverage, i.e. the left ventricle. This conclusion is supported by the results presented in Section 3.3. This could also have been tested by running the echo only approach using an affine motion description, i.e. 12 degrees of freedom. However, in our experience the relatively poor quality of inhale echo images causes a lack of robustness in echo-echo registration. Echo registration using inhale images appears to need some constraints to make the robustness acceptable. Constraining it to 6 degrees of freedom (rigid body) improves the robustness over an affine approach (12 degrees of freedom). Introducing the motion model (1 degree of freedom) produces the most robust results of all. We prefer to use an affine motion model because cardiac respiratory motion is known to be reasonably well approximated by an affine model (Manke et al., 2002).

Although we have showed that the model-based approach performs better than the echo only approach, the results for the model-based approach (DICE coefficient of 0.75 ± 0.11) still represent a significant misalignment. However, there are a number of sources of uncertainty in these experiments. We cannot be certain that volunteers held their breath in the same position for the inhale echo image and the inhale MRI breath-hold image, although the breath-hold repeatability results presented in Table 1 suggest that this error will be less than 2mm in most cases. Also, it is well-known that breath-hold and free-breathing positions of the heart can differ (Keegan et al., 2002). The motion model used for technique 2 was formed from a dynamic MRI scan acquired during free-breathing, so this may introduce extra errors into the validation for this technique. Other possible error sources include errors in calibration of the echo probe, which we estimate to be approximately 2mm (Ma et al., 2008); errors in the MRI to physical space transformation, which again we estimate to be approximately 2mm (Rhode et al., 2003, 2005); segmentation errors, which are also of the order of 2mm (Zhuang et al., 2008); and errors in optically tracking the echo probe and patient bed, which we estimate to be small compared to the other error sources. Bearing in mind these possible sources of error, the fact that the model-based technique results were within our clinical accuracy requirement in 60% of cases is encouraging.

We used a simplified motion model in the work presented in this paper, in which the well-known hysteresis effect (Keegan et al., 2002) was not modelled. The main reason for this was to simplify the optimisation of the similarity measure in the model-based technique. If we used a motion model that captured hysteresis effects then the breathing direction (inspiration or expiration) would need to be introduced as a further parameter in the optimisation,

doubling the optimisation time. Our previous work in the heart (King et al., 2009a) has suggested that the hysteresis effect, although present, would be small compared to other error sources in this work.

The model-based approach has the advantage of increased efficiency. Although the current implementation does not work in real-time, it is a factor of four faster than the echo-only technique. In addition, in principle it has greater potential for parallel execution: because only a single parameter is being optimised, the similarity measure for many possible values of this parameter could be computed in parallel, greatly speeding up the optimisation process.

We performed our visual assessment of registration accuracy in the sagittal plane only. This was a necessary restriction due to data acquisition limitations resulting in a large slice thickness (12mm) of the breath-hold images. This would make any assessment of through-plane (i.e. medio-lateral) errors unreliable. We believe that this validation was justified as it has been reported in the literature that there is relatively little medio-lateral motion of the heart during respiration (Shechter et al., 2004; McLeish et al., 2002; King et al., 2008a). The dominant cardiac respiratory motion parameters are normally the inferio-superior translation and scaling, the antero-posterior translation, and the medio-lateral axis rotation, all of which are observable and quantifiable from sagittal slices.

In conclusion, we have presented a possible solution to the problem of respiratory motion correction of roadmaps that has potential application to a range of image-guided interventions. Our results suggest that the relatively poor quality of 3-D echo data acquired at inhale means that 3-D echo data alone may not contain enough information to allow accurate motion correction estimates to be determined in the heart. Further research is required to investigate whether the same conclusion applies to other abdominal organs. However, results for the model-based approach are encouraging, and suggest that in the future motion correction based on real-time 4-D echocardiography could be possible. We are currently developing techniques for using 3-D echo data to compute the initial MRI to patient registration (Ma et al., 2009; King et al., 2009b), eliminating the need for a full XMR catheterisation suite. If such a technique were adopted, a 3-D echo system would be routinely present so using it for respiratory motion correction would not add any extra expense or inconvenience to the procedure. In future work we will address the incorporation of real-time 4-D echo images into our technique. We will also investigate whether our scheme can be adapted to allow some deviation from the average breathing cycle described by the motion model, enabling the technique to capture the inter-cycle variability in breathing motion.

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