Intra-Cardiac 2D US to 3D CT Image Registration

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ABSTRACT

Intra-cardiac echocardiography (ICE) is commonly used to guide intra-cardiac procedures, such as the treatment of atrial fibrillation (AF). However, effective surgical navigation based on ICE images is not trivial, due to the low signal-to-noise ratio (SNR) and limited field of view of ultrasound (US) images. The interpretation of ICE can be significantly improved if correctly placed in the context of three-dimensional magnetic resonance (MR) or computed tomography (CT) images by simultaneously presenting the complementary anatomical information from the two modalities. The purpose of this research is to demonstrate the feasibility of multimodality image registration of 2D intra-cardiac US images with 3D computed tomography (CT) images. In our previous work, a two-step registration procedure has been proposed to register US images with MR images and was validated on a patient dataset. In this work, we extend the two-step method to intra-cardiac procedures and provide a detailed assessment of registration accuracy by determining the target registration errors (TRE) on a heart phantom, which had fiducial markers affixed to the surface to facilitate evaluation of registration accuracy. The resultant TRE on the heart phantom was 3.7 mm. This result is considered to be acceptable for guiding a probe in the heart during ablative therapy for atrial fibrillation. To our knowledge, there is no previous report describing multimodality registration of 2D intra-cardiac US to high-resolution 3D CT.

Keywords: multimodal image registration, image-guided cardiac procedures, mutual information

1. INTRODUCTION

Atrial fibrillation (AF) is caused by abnormal electrical pulses producing a disturbance of the heart's normal rhythm, and is estimated to affect over one million people in North America.[1] A new, minimally invasive treatment approach involves introducing an ablation tool directly within the cardiac chamber.[2] Although, minimally invasive epi-cardiac interventions are performed under the guidance of endoscope images, procedures performed inside the beating heart must be guided by other means. We believe that ultrasound (US) has a definitive role to play in the guidance of minimally invasive cardiac therapies.

Two-dimensional (2D) echo images of the heart are routinely acquired during cardiac procedures, but they are not interpreted in their true three-dimensional (3D) context. The ability to correlate real-time position and orientation information provided from intra-cardiac echocardiography (ICE) images with high-resolution anatomical information from previously acquired dynamic 3D magnetic resonance (MR) or computed tomography (CT) images would contribute significantly to both image-guided intra-cardiac therapies and cardiac disease diagnosis. By combining tracking information with US images generated by the intra-cardiac echocardiography, real-time US images of the beating heart can be registered to pre-operative dynamic MR images to provide the surgeon with a comprehensive and detailed dynamic model of the patient's cardiac condition in real time. By referring to composite MR/US or CT/US images the surgeon would immediately be able to visualize and correctly orientate the images within the cardiac chambers for surgical navigation.

Image registration between MR/CT and US datasets is achieved by deriving a coordinate transformation between the coordinate spaces of the two modalities. While there are a number of methods and techniques for multi-modality image registration[3,4,5] in the literature, these approaches have mainly been used in neurosurgical applications,[6] abdominal interventions,[7] orthopedic applications,[8] and offline cardiac analysis.[9] In these situations, registration speed is generally not a critical issue, whereas during cardiac procedures, it is a primary concern, as image registration must be completed rapidly.

In this paper, we demonstrate the feasibility of multimodality image registration of 2D intra-cardiac US images with 3D CT images, with the aim of simultaneously presenting complementary anatomical information from the two imaging modalities. Results from phantom studies indicate that our method can register surgical targets within a phantom with a target registration error (TRE) of 3.7 mm. This result is considered to be acceptable for guiding a probe in the heart during ablative therapy for atrial fibrillation. To our knowledge this is the first study reporting multimodality image registration of 2D intra-cardiac US images and 3D CT images.

2. METHODS

2.1 Cardiac Phantom

A heart phantom (Chamberlain Group, MA, USA) was used in these experiments (Figure 1). Fiducial markers (spherical glass targets with the diameter of 2.5 mm) were attached to the surface of the heart phantom. These fiducial markers could be identified in both imaging modalities and facilitated evaluation of the registration accuracy.

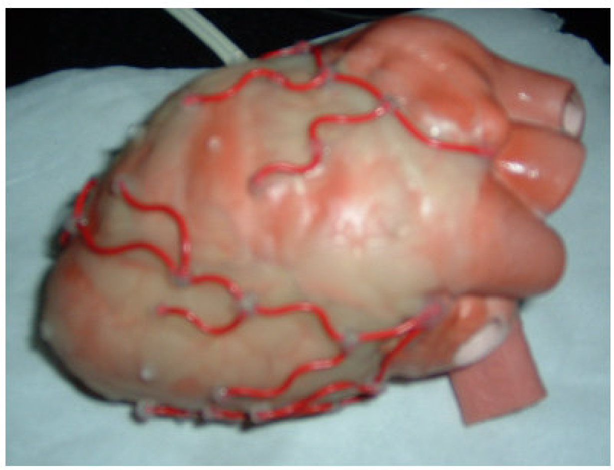


Fig. 1. The heart phantom used in the experiments

2.2 Data acquisition

In this work, all the 3D CT images and 2D/3D intra-cardiac US images were acquired using the heart phantom shown in Figure 1.

3D CT image acquisition

A high-quality 3D CT image of the heart phantom was acquired using a helical CT imaging technique (GE Lightspeed VCT) with the following imaging parameters: slice thickness = 0.625 mm, pitch = 0.0, imaging speed = 3.75 mm/rotation, field of view = 16.2 cm, kVp = 120, mA = 300, total image acquisition time = 1.5 seconds, and in-slice image resolution = 512 x 512.

US image acquisition

2D US images were acquired using an ICE probe (Siemens-Accuson Sequoia, Erlangen, Germany). The ICE probe was tracked with a Polaris optical tracking system (OTS) (Waterloo, ON, CAN). A sample 2D ICE image of the heart phantom is shown in Figure 2.

The first step in the US image acquisition protocol was to calibrate the US images into the OTS coordinate system. US image calibration requires two transformation matrices:

(i) $T_{S \leftarrow 2DUS}$ transforms the 2D US pixel coordinates in to the coordinate space defined by the tracking tool (infrared emitting diode (IRED) sensors attached to the US probe) and;

(ii) $T_{TS \leftarrow S}$ transforms the coordinates from the tracking tool space into the OTS space. The second matrix, $T_{TS \leftarrow S}$, is determined automatically by the OTS. To find the transformation matrix $T_{S \leftarrow 2DUS}$ a Z-bar calibration procedure[10] was used.

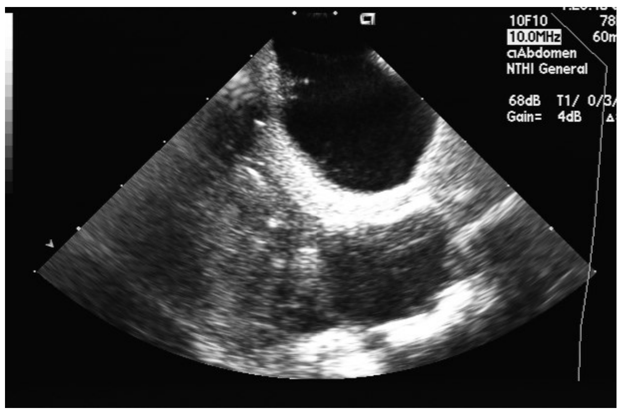


Fig. 2. 2D ICE image of the heart phantom

The US images were captured using a Matrox video frame grabber card (Matrox Electronic Systems Ltd., Dorval PQ, Canada), where each US frame was time stamped. A median filter was applied to the 2D US images to reduce speckle noise and the field of view (FOV) in each US image was restricted to depict only the relevant heart structures and to remove the background in each image.[11] Since the tracking information and 2D US images are not necessarily temporally synchronized, time stamping allows us to match up the US frames with the appropriate tracking information. From the tracked 2D US, the following images were generated.

- **Reconstructed 3D US images**: the tracked 2D US images were transformed into the tracking system space and combined with 3D tracking information to reconstruct 3D US volumes in real time using the freehand reconstruction method.[12]
- **Simulated 2D US images**: a quantitative evaluation of 2D US – 3D CT registration accuracy is not a trivial task as it is difficult to ensure that sufficient fiducial markers appear in the 2D image to assess the registration accuracy. The reconstructed 3D US image was resliced to generate a set of simulated 2D US slices, which are able to employ all the fiducial marker information in the 3D US image for accuracy assessment since we know the spatial relationship between the simulated 2D US images and the 3D US image during the generation of the simulated 2D US slices.

2.3 Registration Method

A two-step registration procedure has been proposed to register 3D US images and 3D MR images of the heart and was validated on patient image datasets.[13] In this work, we aim to extend the two-step method to register 2D intra-cardiac US and 3D CT images and provide a detailed assessment of registration accuracy. In this study, the mutual information (MI) similarity metric[14] was maximized to achieve optimal rigid US – CT image registration using a gradient-based optimization method.

The ultimate objective of our project is to improve image guidance for minimally invasive cardiac surgery by integration of intra-cardiac US images with high quality CT/MR images. The purpose of the two-step registration procedure is to accelerate intra-operative registration of intra-operative US images with pre-operative CT/MR images. The first step involves registering the reconstructed 3D US and 3D CT images to provide a near-optimal starting point for intra-operative 2D US – 3D CT image registration. In the second step, intra-operative registration is accelerated by using the initial transformation from the first step along with tracking information, since the MI-based registration speed depends strongly on the distance from the starting point to the optimal solution. For 2D – 3D registration, we considered two cases: 1) simulated 2D US – 3D CT image registration and 2) tracked 2D US – 3D CT image registration.

3D US – 3D CT registration

The 3D US and 3D CT images were first viewed in custom software (OCCIViewer)[15] and the two image sets were manually translated and rotated until the volumes were approximately aligned to provide the 'initial' guess for our intensity-based registration. The mutual information (MI) similarity metric was then maximized to obtain the optimal registration transformation $T_{CT \leftarrow 3DUS}$, which is employed to map the 3D CT image to the heart phantom.

Simulated 2D US – 3D CT registration

After the simulated 2D US slices were extracted from the 3D US volume, the transformation $T_{2DUS \leftarrow 3DUS}$, responsible for mapping from the 3D US image coordinate space to the simulated 2D US image coordinate space, was determined. The starting point of this 2D – 3D registration was derived from the transformation $T_{2DUS \leftarrow 3DUS}$ and the 3D US – 3D CT registration result $T_{CT \leftarrow 3DUS}$. The MI similarity metric was then maximized to find the optimal registration between the simulated 2D US and 3D CT images.

Tracked 2D US – 3D CT registration

In this case, each tracked 2D US image was associated with a transformation $T_{S \leftarrow 2DUS}$ from tracking information, which was combined with the 3D US – 3D CT registration transformation $T_{CT \leftarrow 3DUS}$ to obtain a near-optimal initial transformation for the tracked 2D US – 3D registration. The MI similarity metric was then maximized to obtain the optimal registration between the tracked 2D US and 3D CT images.

2.4 Registration Accuracy Assessment

We used the target registration error (TRE) of fiducial markers to assess registration accuracy. The TRE is calculated as the root-mean-square distance between the positions of the reference fiducial markers in the CT image and the corresponding fiducial markers in the US image space. In this study, the fiducial markers were located in both the 3D CT and 3D US images shown in Table 1.

3D US – 3D CT registration

After registering 3D CT and 3D US images using the MI-based method, we obtain the optimal registration transformation $T_{CT \leftarrow 3DUS}$, which was employed to transform the positions of fiducial markers in the 3D US space to the 3D CT space for calculating TRE. Because fiducial markers were identified in both 3D CT and 3D US images, the TRE is calculated as follows:

$$TRE = \sqrt{\frac{1}{N} \sum_{k=1}^{N} \left\| P_{CTk} - T_{CT \leftarrow 3DUS} P_{3DUSk} \right\|^2} \qquad (1)$$

where N is the number of fiducial markers on the heart phantom, $\{P_{CTk}, k = 1,2,\cdots,N\}$ the positions of the fiducial markers in the 3D CT image, $\{P_{3DUSk}, k = 1,2,\cdots,N\}$ the positions of the fiducial markers in the 3D US image.

Fiducial marker#	CT space (mm)			US space (mm)		
	x	Y	z	x	y	z
1	-7.46	35.19	30.63	33.37	-77.38	-1.31
2	5.71	47.94	-11.08	37.13	-40.53	26.41
3	30.29	29.74	-23.02	21.47	-12.17	15.22
4	7.92	42.83	-40.00	24.87	-19.53	46.51
5	-47.47	46.17	-0.63	21.77	-77.14	65.07
6	-47.45	37.81	-8.75	3.53	-52.59	90.95
7	-43.03	52.76	-8.41	20.74	-55.98	85.28

Table 1. Fiducial markers in 3D CT and 3D US spaces

Simulated 2D US – 3D CT registration

Since all fiducial markers do not appear in each simulated 2D US image, we cannot directly calculate the TRE of simulated 2D US – 3D CT registration based on fiducial markers. However, based on the transformation $T_{2DUS\leftarrow 3DUS}$ between the simulated 2D US image and the 3D US image, we can map all the fiducial marker positions in the 3D US image space to the simulated 2D US image space. Therefore we can calculate the TRE with equation (2):

$$TRE = \sqrt{\frac{1}{N}\sum_{k=1}^{N}\left\|P_{CTk} - T_{CT\leftarrow 2DUS}\left(T_{2DUS\leftarrow 3DUS}P_{3DUSk}\right)\right\|^2} \tag{2}$$

where the transformation $T_{CT\leftarrow 2DUS}$ was obtained from simulated 2D US – 3D CT registration, and the transformation $T_{2DUS\leftarrow 3DUS}$ was obtained during extraction of the 2D US image from the 3D US image.

Tracked 2D US – 3D CT registration

Similar to the simulated 2D US – 3D CT registration, it is difficult to identify a sufficient number of fiducial markers in one tracked 2D US image. In this case, the TRE of tracked 2D US - 3D CT registration was computed indirectly. In the experiment, the tracking information associated with the 2D US image was employed to transform all the fiducial marker positions in the 3D US image space into the tracked 2D US image space using the transformation matrix $T_{2DUS\leftarrow 3DUS} = (T_{TS\leftarrow S}T_{S\leftarrow 2DUS})^{-1}T_{TS\leftarrow 3DUS}$. Then we computed the TRE based on the positions of the paired fiducial markers. One assumption of this approach is that the tracking information is accurate and any error associated with the tracking system can be ignored. In our experiments, this approach is reasonable since the Polaris tracking system has an accuracy of ~ 0.3 mm. The TRE is calculated by equation (3),

$$TRE = \sqrt{\frac{1}{N}\sum_{k=1}^{N}\left\|P_{CTk} - T_{CT\leftarrow 2DUS}\left((T_{TS\leftarrow S}T_{S\leftarrow 2DUS})^{-1}T_{TS\leftarrow 3DUS}P_{3DUSk}\right)\right\|^2} \tag{3}$$

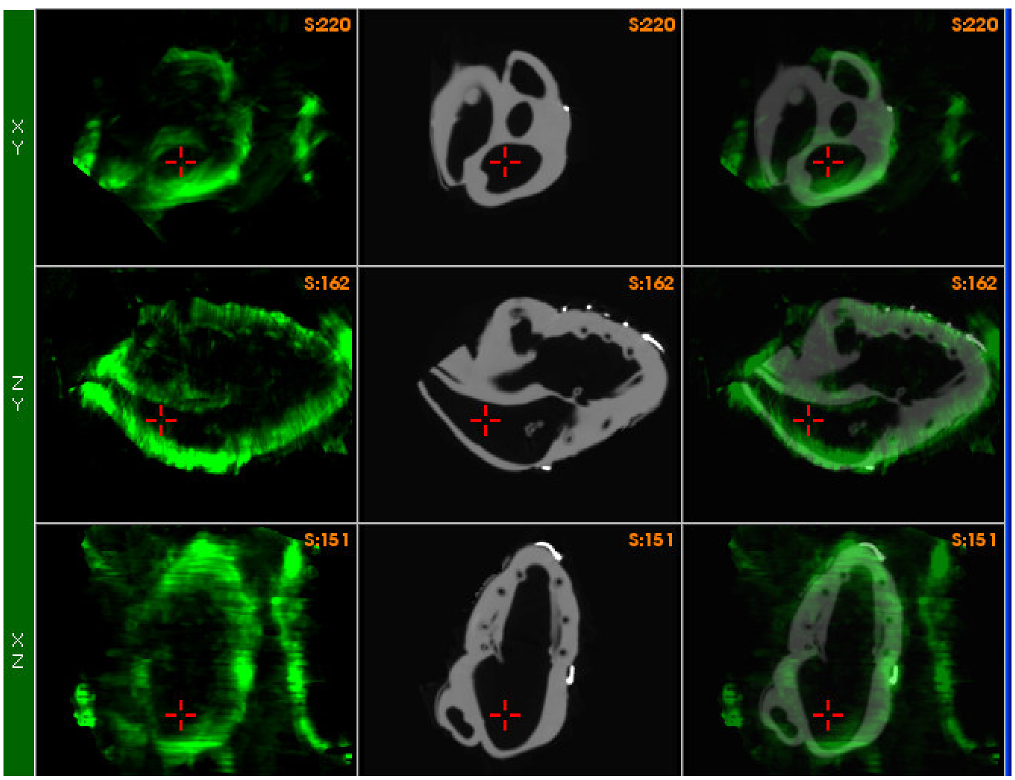
where the transformation $T_{CT \leftarrow 2DUS}$ is obtained from the tracked 2D US – 3D CT registration, the transformation $T_{TS \leftarrow S}$ represents tracking information measured directly by the optical tracking system, $T_{S \leftarrow 2DUS}$ is the calibration transformation matrix, and $T_{TS \leftarrow 3DUS}$ is the transformation from the 3D US image to the tracking system space obtained during the 3D US image acquisition.

3. EXPERIMENTAL RESULTS

The ultimate objective of our research is to acquire the ICE images from within the cardiac chamber. However, for the purpose of our validation experiments, since it was difficult to place fiducial markers on the inner wall of the heart phantom, we instead fustered the markers on the outside of the heart, and acquired the images from the outside of the heart phantom instead.

3D US – 3D CT registration

The MI metric was employed to register 3D US and 3D CT images of the heart phantom. The registration result was visually satisfactory with a TRE of 3.7 mm (Figure 3).

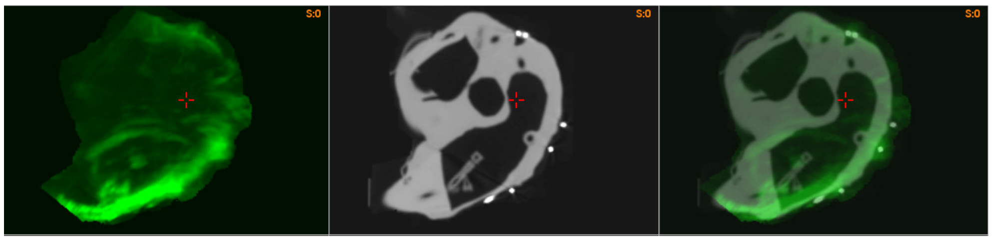


(a) (b) (c)

Fig. 3. 3D US – 3D CT registration (a) three orthogonal slices of the reconstructed 3D US image (b) three orthogonal slices of the 3D CT image (c) the image registration results using MI. The three panels in each column display the 3D object in three orthogonal views.

Simulated 2D US – 3D CT registration

Figure 4 illustrates one of the simulated 2D US-CT registration results. The measured TRE was 4.85 mm.

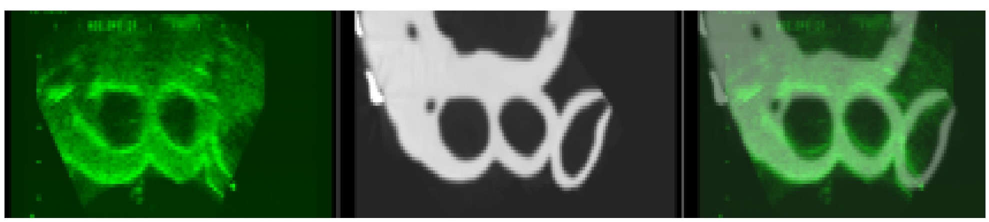


(a) (b) (c)

Fig. 4. Simulated 2D US – 3D CT registration. (a) Simulated 2D US image (b) resampled CT image after registration (c) overlay of two images

Tracked 2D US – 3D CT registration

The result of the tracked 2D US - 3D CT registration are shown in Figure 5 with the registration accuracy (TRE) of 3.70 mm.



(a) (b) (c)

Fig. 5. Tracked 2D US – 3D CT registration (a) 2D US image (b) resampled CT image after registration (c) overlay of two images.

4. DISCUSSION AND CONCLUSION

In this work, we demonstrated the feasibility of 2D intra-cardiac US – 3D CT registration on a heart phantom and lay the groundwork for subsequent clinical experiments. The TRE is comparable to other work in US-MR and US-CT registration using standard US probes. We believe that such US-CT (or MR) registration will be a critical component of new surgical approaches that perform surgical procedures inside the chamber of the beating heart.

In the future, we will build a heart phantom with interior fiducial markers. While the current work has been based upon static registration of the heart, it has the potential to be adapted to the dynamic case, where the ECG signal is employed to synchronize cardiac phases of a pre-operative dynamic CT volume to an intra-operative real-time US image, and ECG-gated multiple 2D US slices at one cardiac phase instead of a reconstructed 3D US image will be registered to the corresponding CT image to shorten peri-operative procedures.

Trans-esophageal echocardiography (TEE) is another widely used US imaging modality suitable for intra-cardiac procedures. Since the target is more readily accessed with TEE than ICE, it is worth investigating the performance of TEE – CT/MR registration using the same approach we have presented in the future.

Minimally invasive intra-cardiac techniques[16,17] have been developed to perform cardiac procedures such as mitral valve replacement inside the beating heart without cardiopulmonary bypass, which significantly reduce surgical trauma and implications and improve cosmetic results compared to traditional open-chest procedures. It has been demonstrated that combining ultrasound and virtual reality is significantly better than using only the 2D ultrasound image for image guidance of mitral valve implantation. In the future, we will integrate virtual reality, intra-operative intra-cardiac US and pre-operative CT images to further improve image guidance for intra-cardiac procedures.

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