Ultrasound-based technique for intra-thoracic surgical guidance

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ABSTRACT

Image-guided procedures within the thoracic cavity require accurate registration of a pre-operative virtual model to the patient. Currently, surface landmarks are used for thoracic cavity registration; however, this approach is unreliable due to skin movement relative to the ribs. An alternative method for providing surgeons with image feedback in the operating room is to integrate images acquired during surgery with images acquired pre-operatively. This integration process is required to be automatic, fast, accurate and robust; however inter-modal image registration is difficult due to the lack of a direct relationship between the intensities of the two image sets. To address this problem, Computed Tomography (CT) was used to acquire pre-operative images and Ultrasound (US) was used to acquire peri-operative images. Since bone has a high electron density and is highly echogenic, the rib cage is visualized as a bright white boundary in both datasets. The proposed approach utilizes the ribs as the basis for an intensity-based registration method – mutual information.

We validated this approach using a thorax phantom. Validation results demonstrate that this approach is accurate and shows little variation between operators. The fiducial registration error, the registration error between the US and CT images, was < 1.5mm.

We propose this registration method as a basis for precise tracking of minimally invasive thoracic procedures. This method will permit the planning and guidance of image-guided minimally invasive procedures for the lungs, as well as for both catheter-based and direct trans-mural interventions within the beating heart.

Keywords: multimodal image registration, validation, ultrasound-based registration, US − CT registration

1. INTRODUCTION

Conventional cardiac surgeries are extremely invasive. In order to expose the surgical target (the heart), a sternotomy is preformed. A 10 – 12 inch incision is made down the middle of the chest and the rib cage is cracked open. The time for the rib cage and surrounding soft tissue to mend is significant, and some patients may require up to 8 weeks for full recuperation. During the surgery the heart is arrested and the patient must be attached to a heart-lung machine in order to maintain blood circulation and oxygenation. During the course of the operation the heart-lung machine functions as the patient's heart and lungs and is considered the most invasive component of the operation. The immune system reacts to the heart-lung machine as a foreign invader and serious inflammation and infection may occur. Patients attached to the heart-lung machine are more likely to experience complications, which increases the length of hospital stay and patient recovery time. The recovery period from conventional cardiac surgeries represents an economic burden on the Canadian health care system, and has serious economic and personal implications for the patient.

In the future, many of these issues will be addressed through the use of minimally invasive procedures. During minimally invasive robotic cardiac operations the surgeon works on the beating heart through small incisions (ports) made in between the ribs. Introducing robotic arms and imaging tools (endoscopes) through these ports allows the surgeon to visualize and operate on the beating heart without performing a sternotomy. Also, since the heart is no longer arrested, a heart-lung machine is not used and complications and recovery times are reduced [1]. Patients typically resume normal activities within 2 weeks following minimally invasive procedures compared to 6 – 8 weeks with conventional surgeries.

Minimally invasive robotic interventions offer the potential to perform therapeutic maneuvers both on the surface and within the heart cavities, while avoiding lasting trauma to the patient. Although epi-cardial approaches can be performed minimally invasively under the guidance of endoscope images, procedures performed inside the beating heart must be guided by other means. Ultrasound systems exist that can be introduced into the cardiac chamber but the images

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produced are of limited utility unless they are accurately registered to pre-operative 3D images of the patient's heart. Without such multi-modality integration the context of the acquired intra-operative image is lost. In order to achieve image registration between intra-operative US images and pre-operative MR (or CT) images, the patient must first be accurately matched to the virtual model. This paper describes an US-based method of registering a patient to pre-operative CT images of the thorax.

We also note the potential applications and value of this approach when interpreting trans-thoracic echocardiogram (TTE) images acquired for diagnostic rather than therapeutic purposes.

2. BACKGROUND

2.1 Surgical Guidance Methodology

For minimally invasive techniques to gain widespread acceptance, it is essential that technology be developed to both ensure accurate guidance of surgical tools and to improve visualization of the tools and images. Currently, these minimally invasive procedures offer limited intra-operative surgical guidance as pre-operative virtual models are not easily integrated and used in the operating room. Our surgical guidance approach (Figure 1) utilizes pre-operative MR/CT images to supplement information gained from intra-operative US. To achieve our goal we need to perform the following discrete steps:

1) acquire pre-operative MR/CT images and create a virtual model
2) register this virtual model to the patient in the operating room
3) acquire intra-operative US images with a tracked probe
4) map the intra-operative ultrasound to the virtual model to guide the surgical procedure

The focus of this paper is the accurate registration of a virtual model to the patient (the peri-operative step). In this navigation process, pre-operative datasets are acquired and used for planning and intra-operative visualization. Additionally, these datasets are preprocessed to accelerate the intra-operative registration. All of the pre-operative preprocessing steps are performed prior to surgery and the algorithms used in preprocessing steps are not time-critical. Conversely, the peri- and intra-operative preprocessing steps take place in the operating room and need to be optimized for speed.

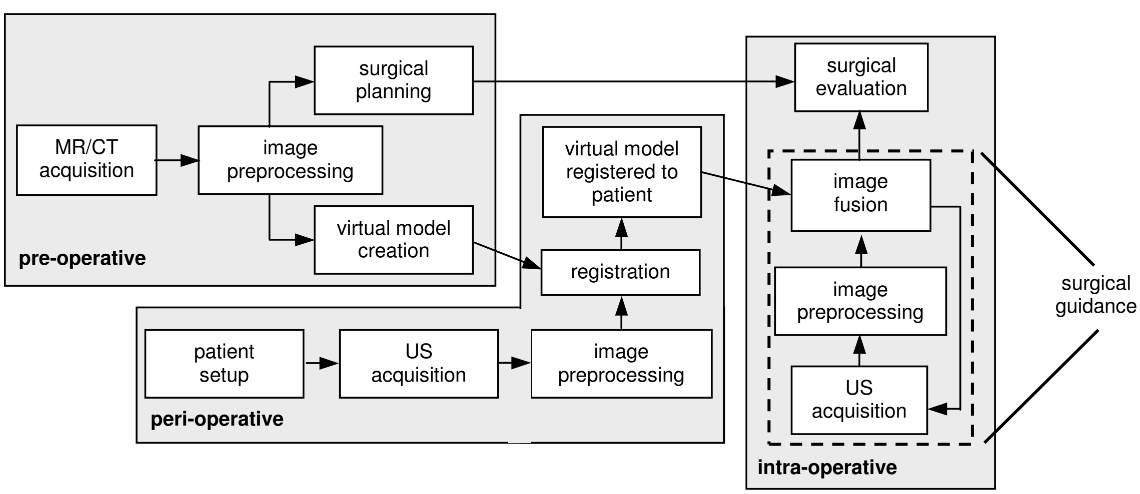


Figure 1. Schematic of the proposed surgical navigation procedure. The pre-opeative MR/CT images are acquired and used to create a patient-specific virtual model of the thorax, which consists of the rib cage and heart. Following patient setup in the OR, peri-operative US images are acquired and used to register the virtual model to the patient within

the OR. When this procedure is adapted to clinical practice, the virtual model will be registered to the patient, and intra-operative US images will be acquired during the surgery and used to update the model through a process of image fusion.

2.3 Ultrasound-based Registration

Both the anatomical structure and unique motion of the heart pose difficult challenges and place special constraints on the registration process. Frame-based techniques (commonly used in brain image registration) [2] or techniques that rely on the placement of external markers on the patient's body [3] assume a rigid underlying anatomy and a fixed spatial relationship of this anatomy with respect to the external markers. Frame-based stereotactic methods are inappropriate for thoracic surgeries and skin markers are too unreliable. To avoid the complications and limitations of using skin markers, we propose a method that involves acquiring intra-operative images and matching them to pre-operative virtual models.

For an intra-operative imaging technique to be acceptable, the benefit it provides must outweigh the disruption it causes to normal operating room activities. Intra-operative methods such as interventional MR or intra-interventional CT restrict the surgeon's access to the patient. Also, much of the standard operating-room equipment (surgical instruments, anaesthesia units and life-support equipment) are not compatible with the strong magnetic fields used during interventional MR. Intra-operative US is attractive because it is safe, comparatively low in cost, simple to use, minimally disruptive and there are no compatibility problems between ultrasound imaging and standard operating room equipment. The use of ultrasound as an aid to image registration is not new. Bass *et al.* [4] used an ultrasonic 'wand' to identify bony landmarks beneath the skin surface. The method described here differs from Bass' approach in that our approach is imaged-based rather then point-based.

There are a number of methods used to perform image registration. Surface-based registration and intensity-based registration are two main approaches used in medical imaging. The accuracy of surface-based registration relies on the accurate segmentation of anatomical structures in the images to be registered. Segmentation of ultrasound images presents a difficult challenge and usually requires manual intervention for optimal robustness, making the accuracy of surface-based registration user-dependent. Currently, segmentation of the bone surface from an US dataset is difficult to implement in a robust way, due to the quality of the images and the extensive time required to segment a complete 3D dataset. Conversely, intensity-based registration provides the best method for ultrasound registration.

Information theoretic registration methods such as mutual information make no assumptions about the intensity values in the images, but match the two images based on a statistical relationship between the two datasets. Mutual information involves iteratively transforming one image (the floating image) until it is optimally aligned with another image (the reference image), based on the maximum of the mutual information metric. The mutual information, $I(A,B)$, between two data volumes A and B is a function of the individual probability density functions $p(a)$ and $p(b)$ and the joint probability density function $p(a,b)$ of voxel intensities in the overlapping zones of A and B.

$$I(A,B) = \sum_a \sum_b p(a,b) \log\left(\frac{p(a,b)}{p(a)p(b)}\right) \qquad \text{equation 1}$$

3. METHODOLOGY

3.1 Data Acquisition

The plastic rib cage frame, from the Pulsatile Heart Model (Limbs and Things Inc.) was used in all experimentation. Fiducial markers (aluminum toriods, height = 1.73 ± 0.08 mm, and radius 1.28 ± 0.08mm) were affixed to the surface of the ribs. The rib cage [Figure 2] was placed in a 7% (by weight) glycerol-water solution to mimic the speed of sound in human tissue.

Phantom image data sets were acquired using both CT and US. The imaging modalities used to acquire the pre- and peri-operative images were specifically selected in order to address inter-modal registration issues. CT was used to acquire the pre-operative images and US was used to acquire the peri-operative images. The registration target in this research (the rib cage) is both highly echogenic and clearly visible in CT, resulting in high-contrast images of the bone surfaces in both modalities.

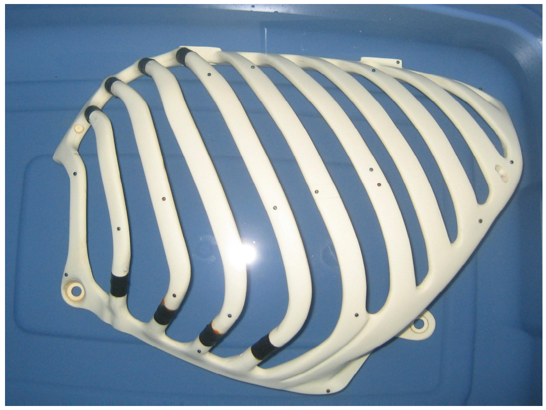


Figure 2. The rib cage phantom. The rib cage phantom was used in all experimentation. Fiducial markers were affixed to the surface of the ribs. For the US acquisition, the rib cage phantom was placed in a 7% (by weight) glycerol-water solution, to mimic the speed of sound of human tissues.

3.1.1 Pre-operative Image Acquisition

A high-resolution pre-operative CT data set was acquired using a helical CT imaging technique with the following imaging parameters: slice thickness = 1.25 mm, pitch = 1.75, imaging speed = 3.75 mm/rotation, field of view = 40 cm, kVp = 140, mA = 160, imagine time = 54 sec, and image resolution = 512 x 512.

3.1.2 Peri-operative Image Acquisition

Two-dimensional (2D) US images were acquired using an Aloka SSD 1700 US machine. The US images were acquired using a 5MHz curved-array probe. The US probe, affixed with a set of infrared emitting diodes (IREDs), was tracked using an optical tracking system (OTS) (Polaris, NDI, Waterloo ON, Canada). The first step in the US image acquisition was to calibrate the US images into this OTS coordinate system. This requires two transformation matrices. The first matrix, $T_{tt \leftarrow us}$, mapped the US imaging coordinate system to the coordinate system defined by the tracking tool. The second matrix $T_{ots \leftarrow tt}$, mapped the coordinates from the tracking tool space into the OTS space. Note the subscript tt denotes the coordinate space defined by the tracking tool.

A Z-bar calibration procedure [5] was used to determine the transformation matrix mapping the US image space to the tracking tool space. US images of a custom made phantom, consisting of strings arranged in four Z-bar patterns, supported by a Lucite frame were acquired. The cross sections of each Z-bar pattern were visible as three bright spots in the US images. Using custom software, a 2.5 mm diameter region-of-interest was manually placed around each bright spot and used to identify the coordinates of each spot. Using the four Z-bars, four sets of homologous points in both the phantom and the image coordinate space are obtained. This fiducial marker configuration permits the location of the centre of the US slice to be determined, and is relatively immune to variations in slice thickness. A calibration program was used to generate the matrix, $T_{tt \leftarrow us}$. The second matrix, $T_{ots \leftarrow tt}$, was determined automatically by the OTS.

Once the US images were mapped into the OTS space, tracked US images were acquired. In order to test the influences of the operator on the quality of the scans, two operators each performed five scanning trails. Scanning a number of regions and combining them to create a large sparse volume ensured adequate spatial coverage of the rib cage. This approach provides the necessary robustness without the necessity of using US to scan the entire rib cage. Five 3D US volumes, representing different regions of the rib cage phantom, were acquired. By using the tracking information, the 2D US images were reconstructed in real-time into 3D US volumes, using the freehand reconstruction method [6].

3.2 Inter-modal Image Registration

The composite US volume was imported into a customized multi-modality image viewer [OCCIviewer, Atamai Inc., London ON, Canada] along with the 3D CT rib volume. The rib images have similar structures repeated throughout both image sets; therefore unsupervised registration algorithms are likely to converge to local optima resulting in false registration. In order to correctly register the two images without getting trapped in local minima, the two images were manually aligned to provide a “first guess” or initial orientation for the registration program. The rib cage is a unique structure and coarsely aligning the two volumes was robust and demonstrated little user dependence.

In this study, the mutual information metric was computed using the method of Mattes *et al.* [7]. This approach is fast and can directly calculate the derivatives for the mutual information metric with respect to the transformation parameters allowing for the use of a fast gradient-based optimization method.

3.3 Registration Accuracy Assessment

We used the fiducial registration error (FRE) measure [8] as our quality reference for registration accuracy. Static FRE represents the mean discrepancy between the position of the reference fiducial points in the CT volume and the corresponding fiducial markers on the patient after registration.

The fiducial markers used in this research were visible in CT images and could easily be identified in the OTS by placing an active tracking pointer (a calibrated pointer with mounted IREDs) into the inner diameter of the toroid. Directly following the acquisition of the US volumes for each trial, the active tracker tool was positioned into the centre of the fiducial marker and the location in OTS space was recorded.

The centre of mass of each fiducial marker was determined by first identifying the approximate centre with a cursor, thresholding the image to isolate the markers from the other features in the CT images, and then using a standard centre-of-mass (COM) calculation (equation 2) to determine the COM in CT coordinate space.

$$(x, y, z)_{com} = \frac{\sum_{i=1}^{M} (x, y, z)_i \cdot m_i}{\sum_{i=1}^{M} m_i} \qquad \text{equation 2}$$

where $(x,y,z)_{com}$ is the coordinate of the COM, m_i is the pixel intensity, and M is the number of neighboring pixels.

The final step in determining the registration accuracy was to compare the fiducial marker locations. The transformation matrix responsible for mapping the CT dataset into the OTS space was applied to the CT fiducial marker set. The root mean square (RMS) distance between the centres of matched fiducial markers was measured to determine the fiducial registration error (FRE). The target registration error (TRE) was also calculated. The TRE for this registration method was determined using equation 3 from Fitzpatrick *et al.* [8].

$$\left\langle TRE^2(r) \right\rangle > \approx \frac{\left\langle FRE^2 \right\rangle}{(N-2)} \left(1 + \frac{1}{3} \sum_{k=1}^{3} \frac{d_k^2}{f_k^2} \right) \qquad \text{equation 3}$$

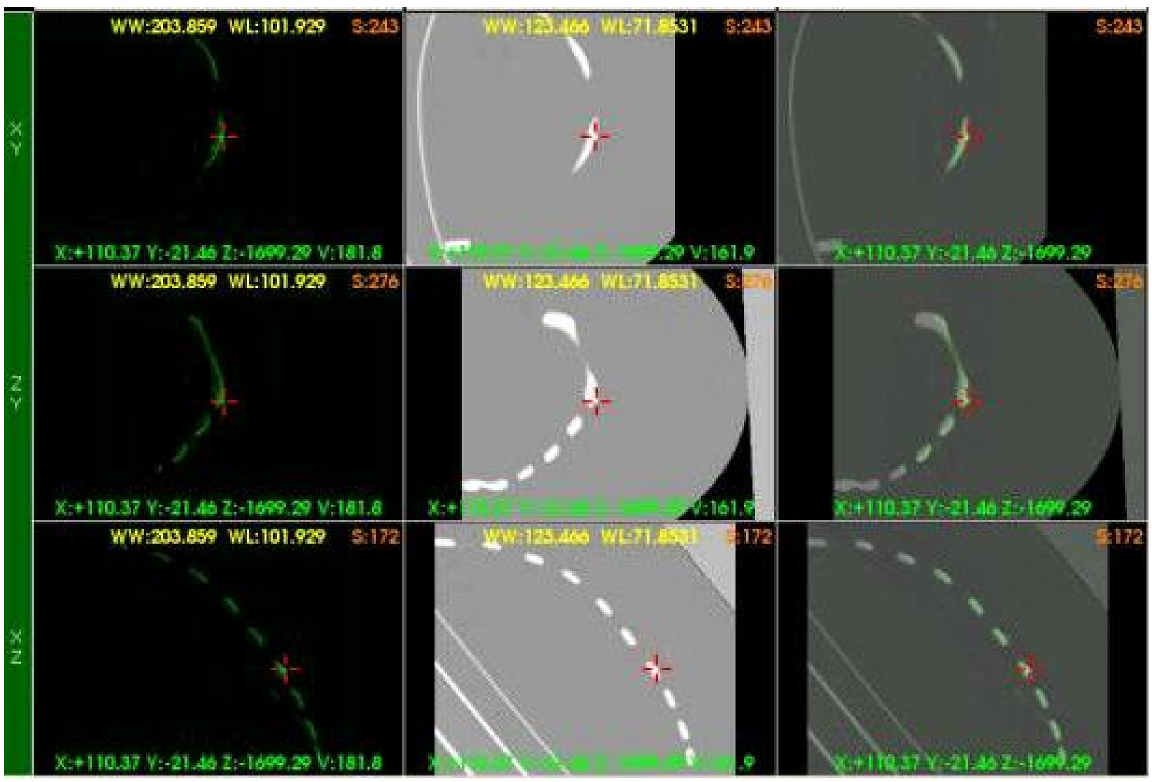
where r is the distance from the target to the centroid of the fiducial markers, FRE is the fiducial registration error, N is the number of fiducials, f_k is the RMS distance of the fiducials from the principal axis k, and d_k is the distance of the target from principal axis k. The target distance, r, used to calculate the TRE was 50 mm, the typical distance from the rib cage surface to the centre of the heart.

A point-based registration method was used to register the fiducial marker locations in CT space to the OTS coordinate space and the RMS distance from matched fiducial markers was measured and compared to the FRE results.

4. RESULTS

4.1 Inter-modal Image Registration

The accuracy of this method to register 3D US images to 3D CT images was found to be < 1.5 mm. A student's t-test, used to analyze the data between the two operators demonstrated that there was no significant difference between the two sets of measurements ($p = 0.22$). In Figure 3, the user interface displays three simultaneous orthogonal views of the registered data sets. The intersection of the three planes defines the current position in the data volume, and the crosshairs are positioned on a matched fiducial marker.



(a) (b) (c)

Figure 3. Registration between 3D US and 3D CT images. (a) orthogonal slices of the US volume of the rib cage phantom; (b) the CT volume of the rib cage phantom; (c) the overlay of the two image sets after registration.

4.2 Registration Accuracy

The registration accuracy of this approach was quantified using both FRE and TRE measures and was compared to a point-based registration method. The FRE demonstrated the discrepancy between the registered CT volume and the true patient anatomy. The TRE quantified the accuracy of this approach to locate a target within the rib cage. The fiducial registration errors obtained for each of the 5 US scanning sessions for both operators is summarized in Table 1. We calculated the FRE of the point-based registration method using 6 fiducial markers, and using equation 3 we calculated the TRE of an assumed target 50 mm beneath the rib cage to be 1.37 ± 0.37 mm. Since the FRE using our mutual information approach is 1.46 mm, we compute (through linear scaling) that the resulting TRE at the same assumed target is 2.24 ± 0.81 mm.

Trial #	Point-based Registration Error (mm)	Mutual Information-based FRE (mm)
Operator 1		
1	1.05	1.67
2	0.75	1.49
3	0.99	1.14
4	0.73	1.00
5	0.76	0.93
Operator 2		
1	0.93	1.72
2	1.47	1.58
3	0.93	2.40
4	0.72	2.02
5	0.64	0.67
Average (± SD)	0.90 ± 0.24	1.46 ± 0.53

Table 1. Fiducial registration errors for the mutual information registration method and the point-based registration errors. The fiducial registration error (± SD) for each operator. The US-based registration method was able to accurately register the CT and US images to within 1.46 ± 0.53 mm [mean ± SD].

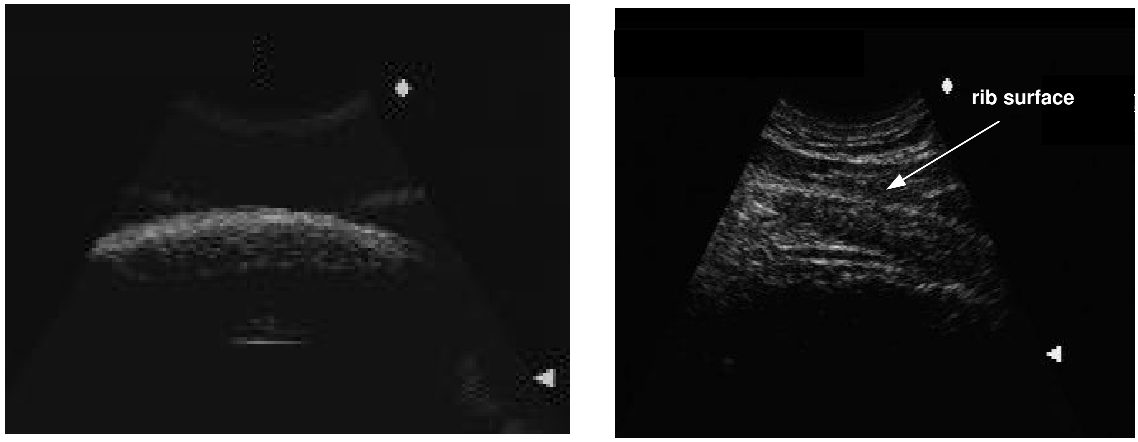
5. DISCUSSION

In this paper we discussed a novel approach used to register a virtual model to a patient within the operating room. This method maps high-resolution CT images of the thorax with US images using the rib cage as a basis for a mutual information-based registration procedure. This method was found to be accurate and showed little operator-dependence.

A plastic phantom rib cage was used to validate all of this research. Although the rib cage used in this preliminary work was plastic, the US images obtained from the actual rib surfaces, even in the presence of overlying soft tissue, are sufficiently similar to those we obtained in this experiment (Figure 4) and strongly suggests that similar registration accuracy would be achieved *in vivo*, provided the starting point for the MI algorithm is within millimeters of the final position. By manually aligning the two image data sets close to the global minima the registration algorithm will ignore any other boundaries present in the images and map the correct rib surfaces together. Depending on the features in the images manual alignment can be difficult, however, that was not the case in this research as the rib cage is a unique object and can be easily aligned near the final position.

This preliminary study produced encouraging outcomes. Although this approach has yet to be optimized for accuracy, the resultant FRE < 1.5 mm and TRE < 2.3 mm are especially promising since the point-based registration algorithm, which is mainly influenced by the accuracy of finding the fiducial marker locations, produced a FRE = 0.9 mm and TRE = 1.37 mm. The more accurately we can locate the fiducial markers in the OTS the more accurate our mutual information method.

This registration procedure is aimed at providing surgical navigation for intra-cardiac procedures using tracked US. Since US is used during the surgical procedure to acquire intra-operative images, expanding the role of US to include registering the virtual model to the patient is a natural choice. In practice both patient respiration and heart motion will be confounding factors and any model-based navigation must address these motion issues. A future study will be to expand the current phantom model to represent the dynamic nature of the *in vivo* environment.



(a) (b)

Figure 4. Visual comparison of the phantom rib cage to an *in vivo* situation. (a) the phantom rib cage used in this study (b) US images of human rib cage in an in vivo environment. The US signal reflects off of the anterior surface of the ribs and produce a boundary feature in the US images.

6. CONCLUSIONS

A novel US-based registration method used for registering a virtual CT model to a patient was tested. This method was validated using a thoracic phantom and was able to register CT and US images with an accuracy of < 1.5 mm. This method showed no significant variation between operators and will provide a robust method of registering a virtual thoracic model to a patient.

Future investigations include extending the proposed method to provide a reference coordinate system for use with a dynamic virtual model of the beating heart, which will then be registered with and synchronized, to the patient. This approach will also be used to provide image-guided support to a new procedure that treats atrial fibrillation by introducing electro- and cryo-ablation tools into the left atrium directly through the wall of the beating heart.

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