Registration of 3D Ultrasound Through an Air–Tissue Boundary

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***Abstract*—In this study, we evaluated a new method for registering three-dimensional ultrasound (3DUS) data to external coordinate systems. First, 3DUS was registered to the stereo endoscope of a da Vinci Surgical System by placing a registration tool against an air–tissue boundary so that the 3DUS could image ultrasound fiducials while the stereo endoscope could image camera markers on the same tool. The common points were used to solve the registration between the 3DUS and camera coordinate systems. The target registration error (TRE) when imaging through a polyvinyl chloride (PVC) tissue phantom ranged from 3.85 ± 1.76 mm to 1.82 ± 1.03 mm using one to four registration tool positions. TRE when imaging through an *ex vivo* liver tissue sample ranged from 2.36 ± 1.01 mm to 1.51 ± 0.70 mm using one to four registration tool positions. Second, using a similar method, 3DUS was registered to the kinematic coordinate system of a da Vinci Surgical System by using the da Vinci surgical manipulators to identify common points on an air–tissue boundary. TRE when imaging through a PVC tissue phantom was 0.95 ± 0.38 mm. This registration method is simpler and potentially more accurate than methods using commercial motion tracking systems. This method may be useful in the future in augmented reality systems for laparoscopic and robotic-assisted surgery.**

***Index Terms*—Endoscopy, medical robotics, prostate, registration, surgical guidance/navigation, ultrasound, virtual/augmented reality.**

I. Introduction

Laparoscopic surgery and robotic-assisted laparoscopic surgery have benefits compared to traditional open surgery that include reduced blood loss, improved visualization of the surgical field, and shorter postoperative recovery time. Unfortunately, these approaches greatly reduce haptic feedback to the surgeons, making it difficult for them to identify the mechanical properties of tissues. Although stereo cameras are used for guidance, they do not provide subsurface information. Augmented reality (AR) is one research concept aimed at improving guidance, and can be implemented as the overlay of medical image data onto the surgeon's camera view. This allows the surgeon to visualize subsurface anatomic features (tumors, vasculature, etc.). Prior studies in AR for surgery have incorporated data from a variety of medical imaging modalities, including X-ray [1], computed tomography (CT) [2], magnetic resonance imaging (MRI) [3], [4], and ultrasound (US) [5]–[9].

Robotic-assisted surgery using the da Vinci Surgical System (Intuitive Surgical Inc., Sunnyvale, CA) is an excellent platform for AR, because the da Vinci surgeon views the surgical field through a three-dimensional (3D) computer display and controls surgical instruments using haptic devices. Robotic-Assisted Laparoscopic Radical Prostatectomy (RALRP) is a common application of the da Vinci system. The primary goal of RALRP is the removal of all cancerous tissue. Success in this goal can be measured using postoperative pathology. If a prostate specimen is found to contain cancer cells at its boundary, it reveals a positive surgical margin (PSM), indicating that cancerous tissue may have been left in the patient. Another important goal of RP is avoiding two common side effects: loss of urinary control and sexual impotence. These functional side effects may be a result of damage to neurovascular anatomy that occurs during surgery, specifically the nerves and blood vessels that form plate-like bundles that envelop the lateral aspects of the prostate. In nerve-sparing prostatectomy, surgeons attempt to avoid injuring these neurovascular bundles (NVB) by gently releasing them off the prostate while avoiding the use of electrocautery as much as possible. In a survey of recent case series, Coelho *et al.* [10] found overall PSM rates ranged from 9.3% to 33.3% of RALRP patients. Urinary continence rates were found to range from 47% to 93% of patients at three month follow up, and from 82.1% to 92% of patients at 12 month follow up. Potency rates were found to range from 8.3% to 47% of patients at three month follow up, and from 43.2% to 78% of patients at 12 month follow up.

Even with the da Vinci's superior vision system, it is difficult for surgeons to identify the prostate boundaries, and the location of nerves, blood vessels, etc. The lack of haptic feedback and the limitations of the information provided by preoperative biopsies also make it difficult to identify cancerous lesions. As a result, there exists a fundamental trade off in current clinical practice between removing tissue to prevent PSMs and avoiding injury to periprostatic anatomy such as the NVB. Previous studies have found that intraoperative transrectal ultrasound (TRUS) is useful in prostatectomy for confirming the correct plane of dissection between the prostate base and the

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bladder wall, identifying hypoechoic regions that may represent cancerous tissue near the boundary of the prostate, and visualizing bloodflow in the neurovascular bundles [11]–[13]. An AR system based on TRUS could thus allow the surgeon to perform a more accurate dissection of the prostate, decreasing PSMs while minimizing damage to the sensitive periprostatic anatomy. This could potentially improve both oncological and functional outcomes for the patient.

AR in surgery involves three main technical problems: developing a 3D model of subsurface anatomy based on medical image data, registering the anatomic model to the kinematic or visual frame of the user, and rendering the model to the user [8]. The registration of the medical image data is critical because errors from registration translate directly to errors in anatomic feature localization. This paper investigates the problem of registration between US and robot (camera, manipulator) coordinate systems.

Existing methods for US to camera registration are generally based on external tracking systems. These methods involve three steps. First, optical trackers, magnetic trackers, or robotic manipulators are used to monitor the 3D poses of the US transducer housing and camera housing with respect to a fixed external coordinate system [5]–[7], [14]. Second, the US volume is calibrated to the pose of the transducer housing using a calibration phantom [15], [16]. Third, the 3D camera coordinate system is calibrated to the pose of the camera housing [17]. While these registration methods based on external tracking systems have been applied with success in various clinical settings, they have several shortcomings. The external tracking systems themselves are expensive pieces of equipment that occupy significant space in the operating room (OR). Other pieces of equipment present in the OR can interfere with their operation, for example by blocking the line of sight of optical tracking systems or disrupting the sensitive fields of magnetic tracking systems. Both of these issues are especially problematic in RALRP, where the da Vinci patient-side cart is docked to the OR table, abutting the surgical field. Given that the US transducer poses are tracked with respect to the housing, errors in pose can be magnified by the lever arm effect between the housing and the imaging plane. External tracking systems also typically require modifications to existing transducer and camera housings in order to mount magnetic or optical markers.

Direct US to camera registration, without any external tracking system, requires common features that can be identified in both modalities and used as fiducials. Unfortunately, US and camera data only overlap at boundaries between air and tissue. Therefore, common features must be located at the air–tissue boundary in order to be used as registration targets. This paper describes a method for registering three-dimensional ultrasound (3DUS) to stereoscopic cameras based on this concept. In our method, a registration tool with optical markers and ultrasound fiducials is pressed against the air–tissue boundary so that it can be imaged by both the cameras and the 3DUS, thus providing common points in the two coordinate systems. By eliminating the US transducer calibration and the external tracking systems, this method reduces the possible sources of error in the registration. An initial feasibility study on this approach that did not use an endoscopic camera and had much more limited validation was previously presented [18].

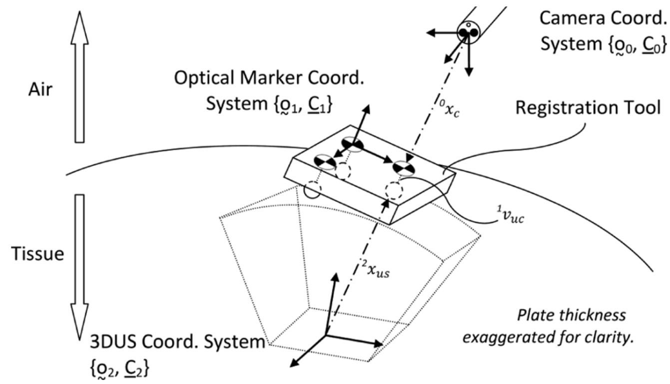


Fig. 1. Registration concept.

Based on the concept of directly identifying common points on an air–tissue boundary, it is also possible to register 3DUS to other coordinate systems, such as the kinematic coordinate system of a surgical robot. This could be used, for example, to generate forbidden-region haptic fixtures [19] based on medical image data. Such registration can also allow real-time US imaging systems to track robotic surgical tools during a procedure, as we describe in [20]. In the case of the da Vinci Surgical System, Intuitive Surgical's Application Programming Interface (API) permits real-time streaming of kinematic data from the da Vinci manipulators [21] relative to a fixed kinematic coordinate system on the da Vinci patient-side cart.

In this study, we examine the accuracy of registering 3DUS to the stereo endoscope and kinematic coordinate system of a da Vinci Surgical System using registration tools pressed against an air–tissue boundary.

II. Method

A. Registration Concept

Fig. 1 depicts our concept for ultrasound to camera registration. The registration tool itself may be implemented in a variety of forms. The form we will focus on in this study is a simple "drop-in" tool with three camera markers on one face and three ultrasound surface fiducials on the opposite face. To perform a registration with this tool we define three coordinate systems: the stereo camera coordinate system $\{\underset{\sim}{o}_0, \underline{C}_0\}$, the optical marker coordinate system $\{\underset{\sim}{o}_1, \underline{C}_1\}$, and the 3DUS coordinate system $\{\underset{\sim}{o}_2, \underline{C}_2\}$. The goal of the registration is to determine the homogeneous transformation ${}^0T_2$ from $\{\underset{\sim}{o}_0, \underline{C}_0\}$ to $\{\underset{\sim}{o}_2, \underline{C}_2\}$. The coordinates of the three camera markers in $\{\underset{\sim}{o}_0, \underline{C}_0\}$, ${}^0x_{c0}$, ${}^0x_{c1}$, and ${}^0x_{c2}$, are determined by stereo triangulation. Likewise, the coordinates of the three US fiducials in $\{\underset{\sim}{o}_2, \underline{C}_2\}$, ${}^2x_{us0}$, ${}^2x_{us1}$, and ${}^2x_{us2}$, are determined by segmenting the fiducials out of the 3DUS volume. The offset between the camera markers and the ultrasound fiducials, ${}^1v_{uc}$, is known from the geometry of the tool. The offset is applied to yield the position of the ultrasound fiducials in $\{\underset{\sim}{o}_0, \underline{C}_0\}$, ${}^0x_{us0}$, ${}^0x_{us1}$, and ${}^0x_{us2}$. There are then three common points known in both $\{\underset{\sim}{o}_0, \underline{C}_0\}$ and $\{\underset{\sim}{o}_2, \underline{C}_2\}$, which means that a standard least squares approach can be used to solve for the transformation ${}^0T_2$. Multiple positions of the registration tool can be imaged and incorporated to increase the number of fiducials, and thus the accuracy of registration.

As an alternative to a purpose-built drop-in tool, existing laparoscopic or robotic instruments can be used to define common points in two coordinate systems. Many existing instruments (forceps, needle drivers, etc.) have a semi-spherical tip profile when their tool jaws are closed. This tip can be pressed into a tissue surface to create a feature that can be segmented in ultrasound. Although the instrument tip itself cannot be seen directly in the cameras when pressed into tissue, a model of the tool's geometry could be constructed to allow features on the tool body such as edges, hinge points, etc., to be used as optical markers for stereo triangulation. The surgical tool could thus be imaged when pressed at multiple positions on a tissue surface, with each position defining one target point for the registration, similar to the registration tool described above.

B. Apparatus

1) Laparoscopic Stereo Cameras: A 12-mm 0° da Vinci stereo endoscope was used for camera imaging. A da Vinci Standard model was used for phantom testing, and a da Vinci Si model was used for *ex vivo* tissue testing. The stereo camera images were captured using two Matrox Vio cards (Matrox Electronic Systems, Dorval, QC, Canada), with the left and right channel DVI outputs from the da Vinci surgical console streamed to separate cards. The capture system ran on an Intel PC with 10 GB memory running Windows XP 32-bit Edition. The images were captured synchronously using the native Matrox API at 60 frames/s and a resolution of 720×486 pixels.

2) Three-Dimensional Ultrasound System: All ultrasound data for this study were captured using a biplane TRUS transducer in combination with a PC-based ultrasound console (Sonix RP; Ultrasonix Medical Corp., Richmond, BC, Canada). The transducer has both linear sagittal and convex transverse arrays, but the 128-element linear array was used for all imaging. Imaging depth was set at 55 mm. Focus depth was adjusted before testing to produce the best possible image, and afterwards remained constant.

Intraoperative TRUS imaging in RALRP requires a system for holding the transducer itself, since the da Vinci cart is typically positioned immediately between the patient's legs, with the column and arms of the da Vinci greatly limiting access to the patient's rectum. To address this issue, robotic TRUS probe manipulators that allow remote control of the transducer have been proposed [22]–[24]. One such robotic TRUS imaging system (shown in Fig. 2) based on a modified brachytherapy stepper [22] was used to capture 3D data in this study, by rotating the TRUS transducer around its axis and recording ultrasound images at known positions. The range of angles captured by the robot was adjusted according to the position of the registration tool. Images were captured with an angular increment of 0.3° in all testing.

3) Drop-in Registration Tool: Fig. 3 shows the drop-in registration tool used in the first implementation of the registration concept. It consists of a machined stainless steel plate, with angled handles designed to be grasped by da Vinci needle drivers. Optical markers on the top surface are arranged directly above stainless steel spherical fiducials on the opposite face. The

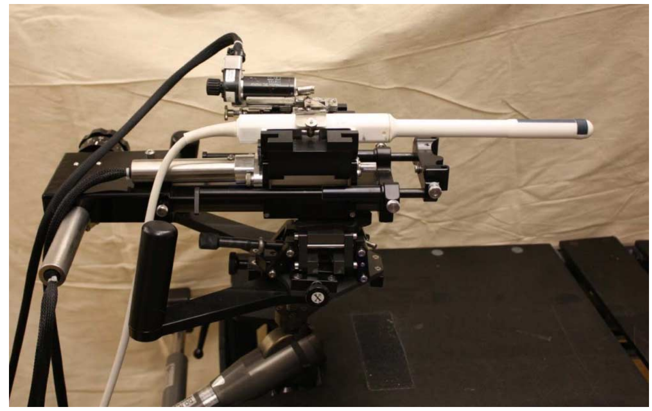


Fig. 2. Robotic TRUS imaging system used to capture 3DUS data.

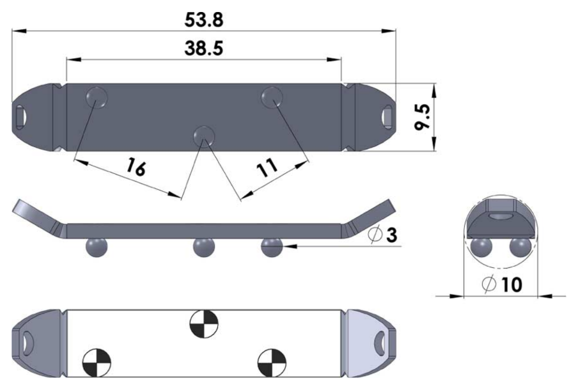


Fig. 3. Schematic of registration tool. Dimensions are shown in millimeters.

spherical fiducials are seated in 1-mm circular holes machined into the plate by a water jet cutter with dimensional accuracy of 0.13 mm, in order to locate them accurately. The fiducials are 3 mm in diameter. Although smaller fiducials would logically seem to provide more precise points to localize as registration targets, we have found that smaller fiducials tend to become lost in the reflections at the air–tissue boundary [18]. The tool was designed to fit through the 10-mm inner diameter of the da Vinci cannulas. It is approximately 9.5 mm wide, with an overall length of approximately 54 mm.

4) Laparoscopic Instrument: A frequently used da Vinci instrument (Large Needle Driver; Intuitive Surgical, Sunnyvale, CA) was used to manipulate the drop-in registration tool. As described in later sections, the instrument itself was also used directly as a second implementation of the registration method by closing the tool jaws, pressing the tip of the closed tool into a phantom tissue surface, and imaging the tip in ultrasound.

C. Registering 3DUS to a Stereo Endoscope

1) Registration Procedure: In this study, we applied our drop-in tool registration method to two different tissue phantoms. A custom-made polyvinyl chloride (PVC) prostate phantom was used to register our 3DUS system to a da Vinci Standard system in a research lab at the University of British Columbia. An *ex vivo* porcine liver was used to register our 3DUS system to a da Vinci Si system in a research lab at Vancouver General Hospital. Both tests followed the same experimental procedure, described below.

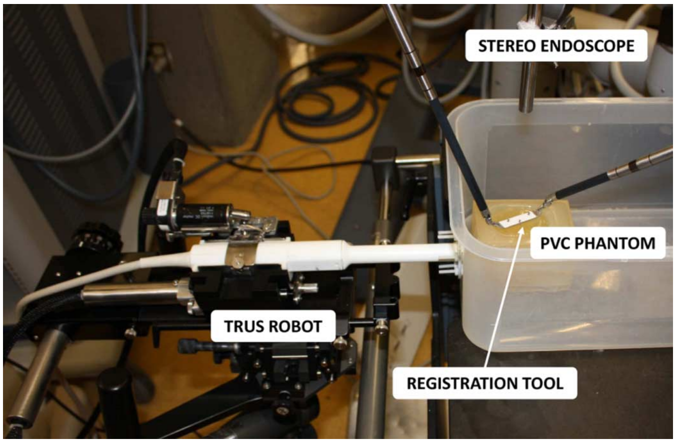


Fig. 4. Experimental setup, showing the TRUS robot, registration tool, stereo endoscope and PVC phantom.

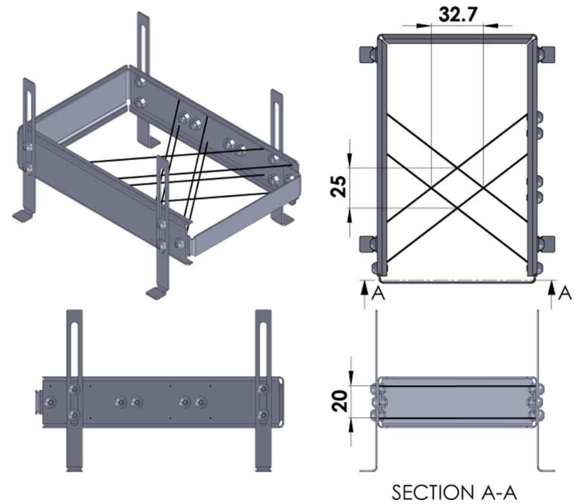


Fig. 5. Schematic of cross wire validation phantom. Dimensions are shown in millimeters.

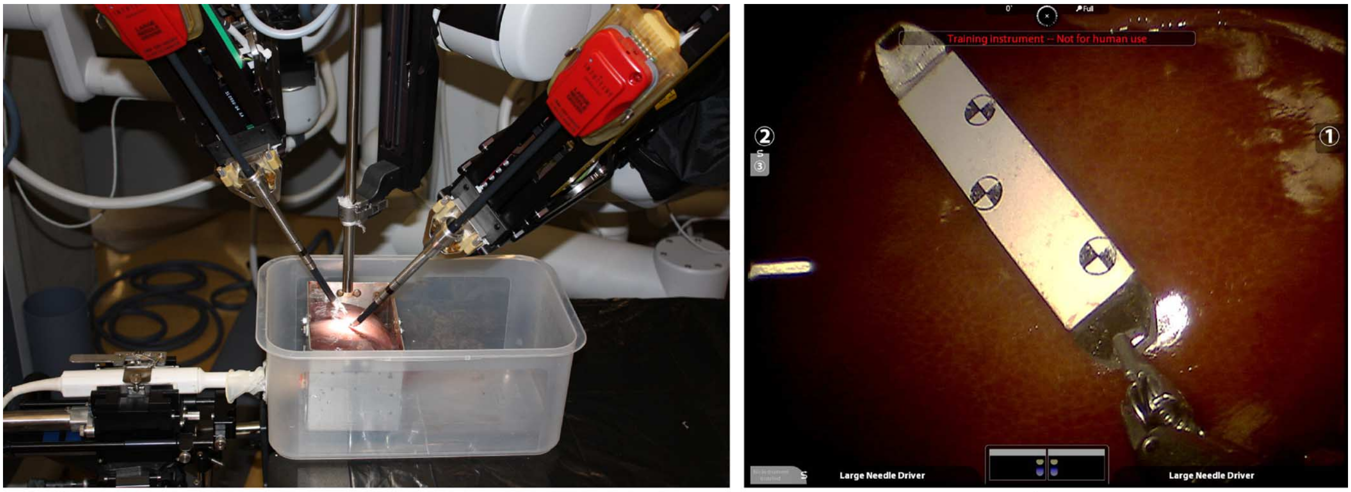
Fig. 4 shows an overview of the setup used in these experiments. The TRUS transducer was installed on a standard operating room table using a brachytherapy positioning arm (Micro Touch; CIVCO Medical Solutions, Kalona, IA). The da Vinci stereo endoscope was positioned so that it could view the sagittal imaging array of the TRUS transducer. A tissue phantom was installed over the TRUS probe, with the top surface visible in the da Vinci camera view. As shown in Fig. 6, the registration tool was applied to the top surface of the phantom using the da Vinci manipulator. The tool was positioned so that the three ultrasound fiducials were visible in the 3DUS and the three optical markers were visible in the da Vinci camera view. The left and right camera images and a 3DUS volume were captured. The registration tool was moved to a new position, and reimaged. A total of 12 positions were imaged for each ultrasound phantom.

A standard stereo camera calibration was performed using Bouguet's camera calibration toolbox for Matlab [17]. The registration tool optical markers were selected in the left and right camera images, and the initial selection was automatically refined to sub-pixel precision using a Harris corner detector. The left and right image points were then used to triangulate the positions of the registration tool optical markers in the 3D camera coordinate system. Similarly the US fiducials were manually localized in the 3DUS volumes. As described above, the common fiducial points on the registration tool were used to solve the homogeneous transformation between the camera and ultrasound coordinate systems using a standard least-squares algorithm [25], minimizing the sum of squared distance error between the common points. Fiducial points from multiple positions of the registration tool were incorporated in order to increase the accuracy of the registration. Between one and four positions of the registration tool were used, with the registration tool translated and rotated at random across the surface of the phantom, which measures approximately 95 mm $\times$65 mm. Fiducial registration error (FRE) was defined as the average residual error between the camera markers and ultrasound fiducials.

2) Validation Procedure: Fig. 5 shows a cross wire phantom used to evaluate the accuracy of our registration method. The phantom was designed to provide points that could be precisely localized by both 3DUS and stereo cameras. It consists of eight intersection points of 0.2-mm nylon wire arranged in a grid approximately 35 mm $\times$ 25 mm $\times$ 20 mm. The wire grid is supported by a custom-built stainless steel frame.

After the registration was determined, without moving either the US transducer or the stereo endoscope, the tissue phantom was removed and the US transducer was immersed in a waterbath. The cross wire phantom was installed in the waterbath and a 3DUS volume of the phantom containing all eight cross wire points was captured. Again without disturbing any of the apparatus, the water in the bath was drained and left and right camera images of the cross wire phantom were captured. This process was repeated for a second position of the cross wire phantom, yielding sixteen target points in all. The cross wire points were localized in the camera coordinate system using stereo triangulation, and in the US coordinate system by manually localizing the points in the 3DUS volumes. All ultrasound data were corrected for differences in speed of sound ($c_{\rm PVC} = 1520$ m/s; $c_{\rm tissue} = 1540$ m/s; $c_{\rm water} = 1480$ m/s at room temperature). To determine registration error, we used the previously found registrations to transform the positions of the cross wire intersections from the 3DUS coordinate system into the stereo camera coordinate system. These transformed points were compared to the triangulated positions of the cross wires, with target registration error (TRE) defined as the distance between the transformed US points and the triangulated camera points. We measured the error for registrations incorporating between one and four positions of the registration tool (i.e., between three and twelve fiducials). A one-way analysis of variance (ANOVA) is a generalization of the student t-test for more than two groups. We used an ANOVA to determine whether registrations incorporating between two and four positions of the tool were significantly more accurate than the single tool case.

The drop-in tool registration was also used to create example ultrasound image overlays in the da Vinci view. After performing a registration using the method described in the previous section and without moving either the TRUS robot or



(a) (b)

Fig. 6. 3DUS and da Vinci stereo endoscope arranged to image the *ex vivo* air–tissue boundary (a) and da Vinci camera view of the tool pressed against the surface of the porcine liver (b).

the da Vinci stereo endoscope, a prostate elastography phantom (CIRS, Norfolk, VA) was placed so that the TRUS could image simulated anatomy within the phantom, while the endoscope imaged the top surface of the phantom. An ultrasound sweep and corresponding camera images were recorded, and the simulated anatomic features (urethra and seminal vesicles) were manually segmented from all TRUS images in which they appeared. The segmented phantom anatomy was then overlaid in the correct position and orientation on the camera images.

D. Registering 3DUS to a Surgical Robot

1) Registration Procedure: In this experiment, we used the tips of an existing da Vinci surgical instrument to register our 3DUS system to the kinematic coordinate system of a da Vinci Standard system, while imaging through a PVC phantom. Four registrations were performed. For each registration, three points were identified on the surface of the tissue phantom using a da Vinci instrument as follows. The jaws of the instrument were closed, and the tip of the tool was pressed into the surface of the phantom. Using our robotic TRUS probe manipulator, the position of the ultrasound imaging plane was adjusted until the ultrasound image contained the tip of the da Vinci tool. The tip of the tool was then selected in the image, and the positions in the kinematic coordinate systems of the TRUS manipulator and the da Vinci were recorded. The da Vinci API directly provides the position and orientation of the instrument tip relative to the da Vinci kinematic coordinate system, making this straightforward. Previous studies have found the API's position data to be accurate to within approximately 1 mm [26], [27]. After three points were recorded, they were used to solve the rigid transformation between the two coordinate systems.

2) Validation Procedure: After each registration was performed, the da Vinci instrument was imaged at eight different additional positions on the phantom surface, with the location in the kinematic coordinate system of the da Vinci recorded for each point. Series of 2DUS images with position data were captured for each position, and processed into volumetric 3DUS data. The tool tips were manually segmented out of the resulting volumes. Using the calculated registrations, the corresponding points recorded from the da Vinci API were transformed into the 3DUS coordinate system, and compared to the segmented 3DUS points to determine the registration error.

III. Results

Fig. 7(a) and (b) shows the appearance of spherical fiducials pressed against the surface of the PVC ultrasound phantom and an *ex vivo* liver sample, respectively. Fig. 7(c) shows the appearance of the tip of the da Vinci instrument pressed against the surface of the PVC ultrasound phantom. Fig. 7(d) shows the method used for manually localizing the fiducials in the ultrasound images.

A. Registering 3DUS to a Stereo Endoscope

Table I lists TRE (cross wire intersections) and FRE (surface fiducials) when registering through the PVC prostate phantom. Between one and four positions of the registration tool were used to determine the transformation, so results are averaged over multiple possible combinations (e.g., 12 choose 4 combinations, 12 choose 3 combinations, etc.). Asterisks indicate a statistically significant difference in sample mean compared to the single tool results ($p < 0.05$).

Table II similarly lists registration accuracy results when imaging through the *ex vivo* porcine liver.

Fig. 8 shows an example of an overlay image produced using our registration method. The left da Vinci camera image, captured while the stereo endoscope imaged the prostate elastography phantom, is shown. The segmented simulated urethra and seminal vesicles are shown superimposed in the correct position and orientation on the camera image.

B. Registering 3DUS to a Surgical Robot

Table III lists registration accuracy results for registering the 3DUS data to the da Vinci kinematic coordinate system. Errors are presented in the anatomic frame of the patient corresponding to RALRP (i.e., superior–inferior error, e_{S-I}, medial–lateral error, e_{M-L}, and anterior–posterior error, e_{A-P}).

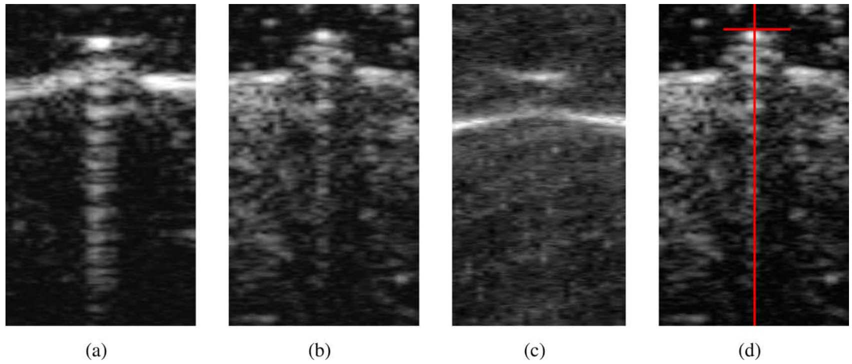


Fig. 7. Surface fiducial against an air–tissue boundary and imaged through a PVC phantom (a) and an *ex vivo* liver tissue sample (b). Laparoscopic instrument tip pressed against an air–tissue boundary and imaged through a PVC phantom (c). Illustration of method for localizing fiducial tip (d): The axis of the reverberation is identified, and the tip is selected along that line.

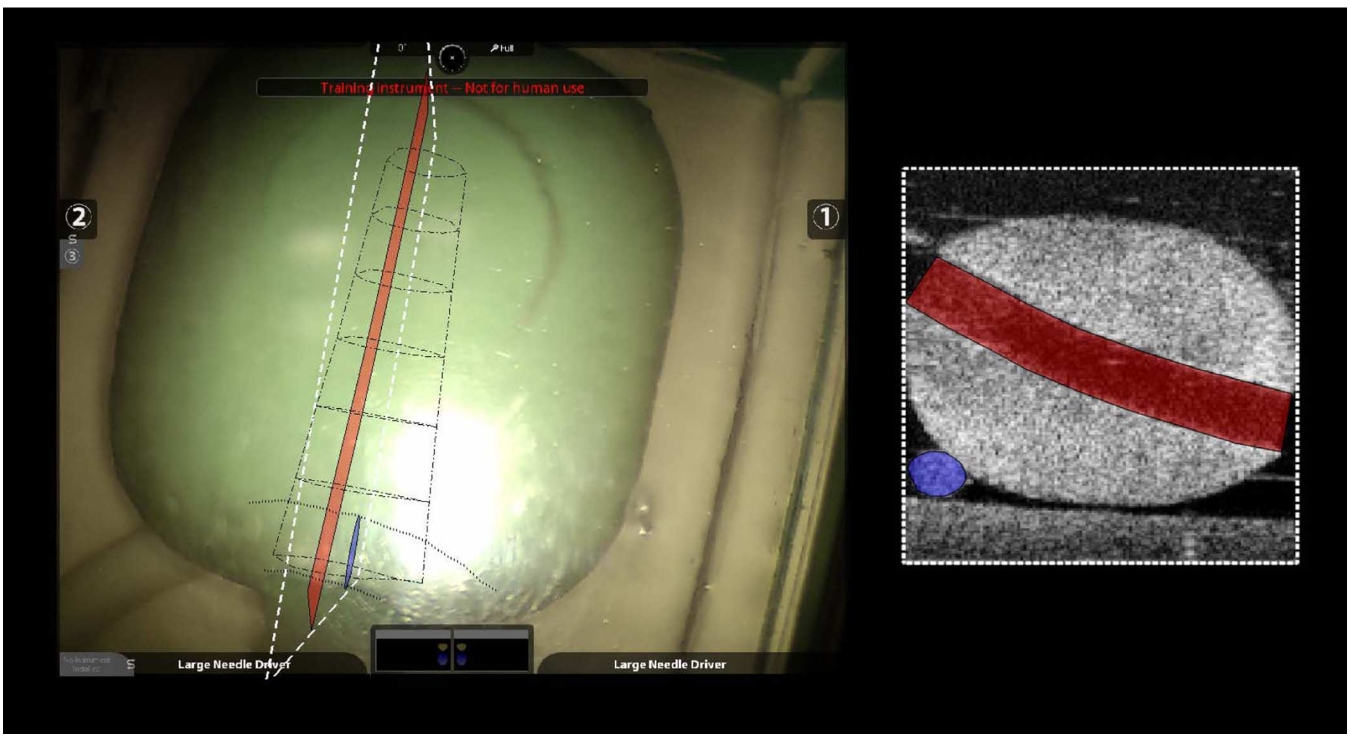


Fig. 8. Example display of TRUS information on da Vinci camera view based on prior registration. The TRUS image on the right shows the manually segmented urethra (red) and seminal vesicles (blue), as well as the image border (white dashed). The segmented anatomy and image border are overlaid in the da Vinci camera image on the left. The left image also shows the tubular outlines of the urethra and seminal vesicles (black dotted), based on the segmentation from other image slices (not shown).

TABLE I
Registration Accuracy Imaging Through PVC Phantom

tool poses	fiducials	targets	FRE (mm)	TRE (mm)
1	3	16	0.20 ± 0.09	3.85 ± 1.76
2	6	16	0.75 ± 0.38	2.16 ± 1.16*
3	9	16	0.81 ± 0.55	1.96 ± 1.08*
4	12	16	0.85 ± 0.62	1.82 ± 1.03*

TABLE II
Registration Accuracy Imaging Through *Ex-Vivo* Liver Tissue

tool poses	fiducials	targets	FRE (mm)	TRE (mm)
1	3	16	0.54 ± 0.20	2.36 ± 1.01
2	6	16	0.82 ± 0.29	1.67 ± 0.75*
3	9	16	0.91 ± 0.32	1.57 ± 0.72*
4	12	16	0.95 ± 0.34	1.51 ± 0.70*

IV. Discussion

In our previous feasibility study on 3DUS to camera registration, we found an average TRE of 1.69 ± 0.60 mm using a single registration tool position and imaging through a PVC tissue phantom [18]. In that experiment, we used stereo cameras with 120-mm disparity, and a relatively large registration tool. In this experiment, we used a da Vinci stereo endoscope with 3.8-mm disparity, and a smaller registration tool designed to fit through a 10-mm cannula. Based on the differences in camera and tool geometry, it is not surprising that we found the equivalent error measure in this study to be greater. In this study, using a single registration tool position, we found an average TRE of

TABLE III
ACCURACY OF 3DUS TO DA VINCI KINEMATIC FRAME REGISTRATION

Trial	e_{S-I} (mm)	e_{M-L} (mm)	e_{A-P} (mm)	e_{Total} (mm)
1	0.46 ± 0.52	0.42 ± 0.51	0.34 ± 0.15	0.79 ± 0.33
2	0.29 ± 0.39	0.48 ± 0.47	0.38 ± 0.10	0.75 ± 0.14
3	0.58 ± 0.68	0.47 ± 0.56	0.53 ± 0.32	1.03 ± 0.37
4	0.82 ± 0.83	0.55 ± 0.63	0.50 ± 0.17	1.23 ± 0.44
Avg.	0.54 ± 0.65	0.48 ± 0.52	0.44 ± 0.21	0.95 ± 0.38

3.85 ± 1.76 mm imaging through PVC and 2.36 ± 1.01 mm imaging through *ex vivo* liver. To improve the results, we compensated for the geometry changes by adding one or more additional positions of the tool to the registration. For the *ex vivo* liver testing, two positions of the tool produced an average TRE of 1.67 ± 0.75 mm. This is comparable to our previous result and represents a statistically significant reduction ($p < 0.05$) in TRE over a single tool position. Previous studies have reported accuracies of 3.05 ± 0.75 mm based on magnetic tracking [6] and 2.83 ± 0.83 mm based on optical tracking [5], although these studies used different accuracy measures. Adding more registration tool positions further reduced the average TRE. Incorporating four tool positions, for example, produced an average TRE of 1.51 ± 0.70 mm for the *ex vivo* liver test. While the increase from one tool position to two tool positions produced a statistically significant improvement, no other additional tool position produced an incremental improvement that was statically significant. In this experiment, the registration tool was randomly repositioned over the phantom surface, so adding a second tool position to the first would be equivalent to using a larger registration tool. Incorporating two registration tool positions would likely be the best choice for a clinical system, as this provides equivalent accuracy to more tool positions without the time needed for additional ultrasound scans.

Based on the previous studies that have considered intraoperative TRUS in RALRP, an AR system based on TRUS would potentially be useful for identifying the correct planes of dissection at the prostate base, at the prostate apex, and medial to the neurovascular bundles [11]–[13]. Identifying the correct plane between the prostate and the NVB is the most critical step, and requires the highest level of accuracy. Ukimura *et al.* [11] found that the mean distance between the NVB and the lateral edge of the prostate ranged from 1.9 ± 0.8 mm at the prostate apex to 2.5 ± 0.8 mm at the prostate base. This is suggestive of the required accuracy for an AR system in RALRP. Our system for 3DUS to camera registration approaches this accuracy when incorporating two positions of the registration tool.

In our previous feasibility study we found the average error for manually localizing surface fiducials to be approximately 1 mm [18]. The appearance of the spherical fiducials in US, and the ability to localize them accurately, clearly has an important effect on the overall accuracy of our method. Because our 3D TRUS system uses sweeps of 2D images to construct 3D data, the resolution is lowest in the elevational direction of the array. Altering the incremental angle between images might thus affect registration accuracy significantly. Altering the depth of the US focus relative to the fiducials might also affect the overall accuracy, as the boundaries of the fiducials would become less clear. The boundaries of the fiducials raise another issue, as it is uncertain whether the high-intensity response at the top of the fiducials represents the actual edge of the spheres or reflections from within the metal spheres. Hacihaliloglu *et al.* also considered this problem when using a stylus pointer with a spherical edge in US. They imaged a row of spheres with different known diameters, and compared the differences between the edges in the ultrasound images with the known differences in diameter. They concluded that the edge of the high-intensity response was in fact the edge of the sphere, perhaps with a small constant offset [28]. The force applied to hold the registration tool against the air–tissue boundary would also appear to be important to the appearance of the fiducials, but we qualitatively observed that varying the applied force did not greatly affect the appearance of the fiducials, as long as the fiducials were in contact with the phantom or tissue.

As seen in Tables I and II, average 3DUS-to-camera registration errors were higher for the PVC phantom than the liver tissue sample with the same number of registration tool positions included, although the differences were small compared to the sample standard deviations. As shown in Fig. 7, the US fiducials appear differently when imaged through the liver tissue compared to the PVC. The tips of the surface fiducials were generally easier to identify through the liver sample, as there was less high-intensity reverberation. This may explain the slight decrease in registration error. Two other sources of error, the manual segmentation of the nylon-wire registration targets and slight deflections of the TRUS robot caused by the weight of the phantoms, may also have led to a small level of variability between registration tests.

The average registration error using a da Vinci instrument to directly identify common points in the 3DUS and da Vinci kinematic coordinate system was approximately 1 mm. Manually localizing the fiducials is likely the main source of error in this registration. Error in the da Vinci kinematic output due to deflection of the manipulators could also potentially affect this accuracy; however, we believe the effect of deflection was minimal in our testing for several reasons. First, the registration accuracy is close to the accuracy of the da Vinci API itself based on previous reports [26], [27]. Second, the force required to create recognizable features on a tissue surface is small compared to the force required for surgical tasks such as suturing. Third, our da Vinci instruments were not constrained by endoscopic ports, so no bending was introduced at these points. When using our registration method in a clinical situation, care must be exercised to minimize misalignment between the instrument arms and the endoscopic ports. Otherwise, as described in [29], the accuracy of the da Vinci API output would be significantly reduced, and some type of model-based correction might be required.

For registration of 3DUS to laparoscopic stereo cameras, we have focused mainly on using a purpose-built drop-in registration tool. It may also be possible to use existing surgical instruments, rather than a special tool, to create registration targets for this purpose. We believe it should be feasible to accurately locate the instrument tip in the camera by creating a geometric model of the surgical instrument, detecting and stereo-triangulating identifiable points on the instrument, and fitting the model to these triangulated points. Although methods for tracking laparoscopic tools using stereo cameras have been previously described [5], [30]–[32], this is a challenging problem for future research.

Performing a registration currently requires manual segmentation and triangulation of registration tool fiducials. Once we have identified the common points, finding the transformation itself requires only a simple least-squares calculation that takes milliseconds on a typical PC. Once fiducial detection and localization in both the ultrasound and camera coordinate systems is made automatic, the registration process will take no longer than the time required to capture a 3DUS volume. Capturing a 45° sweep of sagittal B-mode images using our TRUS robot takes approximately 3 s.

Automatic detection of markers in a camera image is a well studied problem in computer vision (many commercial optical tracking systems are based on this) so we do not believe this step will present an obstacle. We have recently described a method for automatic localization of surface fiducials in 3DUS [33] that takes advantage of the regular appearance of surface fiducials, and the fact that their distinctive comet-tail reverberations (see Fig. 7) distinguish them from surrounding features. In the described method, fiducials are detected in the 2D TRUS images using a circular Hough transform, and a clustering algorithm is used to group detections across images and remove outliers. In a similar experiment to the one described above in Section II-D, a TRUS robot was automatically registered to the da Vinci kinematic coordinate system with an average error of 1.80 ± 0.32 mm [33]. The algorithm is currently implemented in Matlab and takes approximately 55 s to process an 80° sweep of TRUS images. Future implementation in C++ and optimization should reduce the runtime to an acceptable level.

Relative motion of the ultrasound transducer and external coordinate system invalidates the registration generated by our method. This is likely not an issue when registering a TRUS probe manipulator to the da Vinci kinematic coordinate system, as both robots generally remain stationary throughout the RALRP procedure. The da Vinci stereo endoscope does move throughout the procedure, however its motion relative to the TRUS system can be resolved using the kinematic information streamed by the da Vinci API, and this can be used to update the registration. This necessitates calibrating the camera coordinate system of the stereo endoscope to the da Vinci's kinematic coordinate system, which can performed using existing camera calibration tools [17]. Our method can also be made to provide registrations very quickly (i.e., in 5 s or less), so surgeons could simply reapply our registration tool if inaccuracy was suspected.

We propose the following clinical procedure for applying our method to generate camera overlays in RALRP. Before the surgery, a standard stereo camera calibration of the da Vinci endoscope would be performed. After the patient was secured and sedated, a robotic TRUS probe holder or other 3D TRUS system would be installed on the table and positioned to image the patient's prostate. Early in the surgery, for example before dissection of the prostate off the bladder wall, the registration tool would be inserted, applied to the surface of the prostate using the da Vinci manipulators, and imaged to generate a registration. The tool could be removed, or left in the patient away from the surgeon's active working area. Robot kinematics would then be used to update the registration through surgery. If the surgeon desired, the tool would be reapplied and used to generate a new registration.

Although the drop-in tool we have described in this paper was intended to be as clinically realistic as possible, some design modification will be necessary before it can be used in an actual clinical setting. The tool will likely need to be constructed as a single solid part, in order to avoid the use of any adhesives or welding. Optical features that can be localized by the stereo cameras will also need to be machined into the tool, or robust biocompatible optical markers will be needed. Different strategies for introducing and manipulating the tool in the body will also need to be evaluated. The drop-in tool was initially conceived as a small passive device that would be carried into the surgical cavity and manipulated by the da Vinci instruments, similar to sutures. For better control with the da Vinci manipulators, it might be desirable to add a flexible tether or other attachment to the tool, although this would make the tool larger and more cumbersome. Another possibility is to attach the registration tool to a standard laparoscopic tool body, allowing it to be manipulated directly by the patient-side surgeon. This would make removing and reinserting the tool between applications simpler. We plan to develop multiple clinical prototypes, and compare them in a future study.

The registration method we have described requires that the 3DUS images a tissue surface from the opposite side of the stereo endoscope and laparoscopic tools. This makes it well suited to intraoperative TRUS in RALRP. Another possible application is the registration of external abdominal 3DUS during Robotic-Assisted Partial Nephrectomy. Unfortunately, the geometric requirements of our method make it less suited to the registration of laparoscopic ultrasound (LUS). While in some cases it might be possible to utilize our registration method with LUS, for example by applying the registration tool and LUS probe to opposite lateral aspects of the prostate, previously described methods based on optical markers attached to the LUS probe [5], [34] likely hold more promise. It is also worth noting that while our focus in this paper has been registration to the stereo endoscope of the da Vinci system, our method can also potentially be applied to conventional single-channel laparoscopic vision systems. This assumes that the optical markers on the registration tool can be localized accurately enough in 3D using a single laparoscopic camera.

The registration accuracy results presented in this paper appear promising; however, many aspects of an actual clinical environment may have an important impact on the accuracy of registration or the usability of the method. The lighting within the surgical field may be too intense or too low for accurate camera marker detection, and the ultrasound fiducials may appear differently when imaged through living tissue. Follow up *in vivo* tests will be able to answer these questions.

We plan to proceed in two primary directions with our *in vivo* testing. First, in animal models, we will apply our drop-in tool method to provide TRUS image overlays in a simulated RALRP procedure. This will allow us to evaluate the effectiveness of the method, as well as gain insight needed to finalize a clinically appropriate design. Second, in an initial patient trial for which we have already obtained ethical approval, we will use existing surgical instruments to register our TRUS robot to the kinematic coordinate system of a da Vinci robot during RALRP. This will require only minimal modification to the existing surgical procedure (essentially the addition of the TRUS robot and the short delays required for data collection). This initial patient trial will allow us to evaluate the accuracy of registration *in vivo*.

It should be noted that the registration work described in this paper represents part of a larger research objective: the development of an augmented reality system for RALRP. Further work in anatomical model construction and user interface design will be required before a complete AR system can be applied and evaluated in a clinical scenario. On the other hand, our registration method does have more immediate clinical applications which we will evaluate sooner, such as the surgical instrument tracking described in [20].

V. Conclusion

We have presented a new method for registering 3DUS data to the stereo camera and kinematic coordinate system of a surgical robot. Compared to existing methods, our identification of common points on an air–tissue boundary eliminates the need for external tracking systems in the operating room. It also does not require any modifications to existing US or camera systems. For US to camera registration, the only additional equipment required is a simple, inexpensive tool which can be made sufficiently compact to fit through a cannula. Validation shows an average TRE of 3.85 ± 1.76 mm and 2.36 ± 1.01 mm when imaging a single tool pose through PVC phantom and liver tissue respectively, which are considered ideal conditions. Incorporating two poses of the registration tool significantly improves TRE to 2.16 ± 1.16 mm and 1.67 ± 0.75 mm for PVC and liver, respectively. For registration of US data to the kinematic coordinate system of a surgical robot, existing surgical manipulators are sufficient to generate registration features, and no additional equipment is required. Validation shows an average TRE of 0.95 ± 0.38 mm when imaging through PVC phantom. After further developing methods for automatic ultrasound fiducial localization and optical marker triangulation, we plan to apply our registration method to augmented reality systems in clinical trials.

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