Two-dimensional ultrasound-computed tomography image registration for monitoring percutaneous hepatic intervention

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(Received 6 August 2018; revised 14 April 2019; accepted for publication 15 April 2019; published 6 May 2019)

Purpose: Deformable registration of ultrasound (US) and contrast enhanced computed tomography (CECT) images are essential for quantitative comparison of ablation boundaries and dimensions determined using these modalities. This comparison is essential as stiffness-based imaging using US has become popular and offers a nonionizing and cost-effective imaging modality for monitoring minimally invasive microwave ablation procedures. A sensible manual registration method is presented that performs the required CT-US image registration.

Methods: The two-dimensional (2D) virtual CT image plane that corresponds to the clinical US B-mode was obtained by “virtually slicing” the 3D CT volume along the plane containing non-anatomical landmarks, namely points along the microwave ablation antenna. The initial slice plane was generated using the vector acquired by rotating the normal vector of the transverse (i.e., xz) plane along the angle subtended by the antenna. This plane was then further rotated along the ablation antenna and shifted along with the direction of normal vector to obtain similar anatomical structures, such as the liver surface and vasculature that is visualized on both the CT virtual slice and US B-mode images on 20 patients. Finally, an affine transformation was estimated using anatomic and non-anatomic landmarks to account for distortion between the colocated CT virtual slice and US B-mode image resulting in a final registered CT virtual slice. Registration accuracy was measured by estimating the Euclidean distance between corresponding registered points on CT and US B-mode images.

Results: Mean and SD of the affine transformed registration error was 1.85 ± 2.14 (mm), computed from 20 coregistered data sets.

Conclusions: Our results demonstrate the ability to obtain 2D virtual CT slices that are registered to clinical US B-mode images. The use of both anatomical and non-anatomical landmarks result in accurate registration useful for validating ablative margins and comparison to electrode displacement elastography based images. © 2019 American Association of Physicists in Medicine [https://doi.org/10.1002/mp.13554]

Key words: computed tomography, microwave ablation monitoring, registration, ultrasound

1. INTRODUCTION

Hepatocellular carcinoma (HCC) is the most common type of primary liver cancer and is the third leading cause of cancer related deaths worldwide.¹ The American Cancer Society reports that mortality rates attributed to liver cancer continue to increase. Of all newly diagnosed liver tumors, 75% will be HCC,² however, only 9%–27% meet the criteria for surgical liver resection.^{3,4} Loco-regional therapies such as microwave ablation (MWA) remain an important option to treat primary liver cancer.^{5–8} During MWA, electromagnetic energy emitted from an antenna inserted directly into a tumor heats the adjacent tissue to achieve local necrosis. In the case of large

tumors, multiple antennae can be used simultaneously to achieve larger ablation volumes.⁹

Patients who undergo percutaneous MWA typically have a diagnostic pretreatment CT or magnetic resonance imaging (MRI) with contrast.¹⁰ The ablation procedure is commonly but not universally performed under general anesthesia. General anesthesia is commonly applied to immobilize the diaphragm, pain and anxiety management, and to facilitate patient positioning aiding in antenna placement.¹¹ Ultrasound (US) is commonly used to guide antenna placement into the tumor because of its direct, real-time visualization capability and lack of ionizing radiation. Following ablation, a contrast-enhanced biphasic CT (CECT) or contrast MRI scan is

performed to verify the effectiveness of the ablation and to ensure a 5 mm ablation margin around the hepatic tumor.^{12–14}

Need for image fusion during percutaneous ablations arises when hepatic tumors cannot be clearly recognized with US because of their size or location.^{14,15} A growing alternative modality for visualizing liver masses and ablated regions utilizing US is electrode displacement elastography (EDE). EDE uses localized perturbations to detect differences in the tissue stiffness between normal liver and the ablated region to determine ablation zone dimensions. EDE has demonstrated high visualization and delineation in strain tensor images on phantoms,^{16–18} animal models,^{19,20} and human patients.^{21–24} In addition, to ablated region visualization using EDE, an important metric to be evaluated is the area of the ablation region. The hyperechoic area on the B-mode image has been previously segmented and utilized for comparison to EDE.²¹ However, the hyperechoic region on B-mode images does not provide an accurate estimate of the ablation area, and can also shift due to bubble movement after ablation.²⁵ Comparison of EDE derived delineations of ablated regions with the current clinical gold standard CECT is therefore necessary to verify EDE viability in clinical settings. Previous reports on area comparisons of ablated region obtained with EDE to pathology have demonstrated a close correspondence in animal models.^{19,26,27} Comparisons of *ex vivo* ablation areas to CT without contrast have also demonstrated close correspondence of ablation areas between strain imaging, CT and pathological measurements.²⁸ However, *in vivo* datasets require the proposed CT-US registration²¹ for a direct comparison of EDE with CECT. In this paper, we present a method for fusing intraoperative CT volumes with US B-mode for potentially evaluating the efficacy of EDE.¹⁶

Although image fusion tools for this purpose have been under development for 20 yr, the adoption of these tools has been delayed in the clinic due to the extra time required to achieve satisfactory alignment between the image modalities.^{29,30} Intrinsic registration and fusion of preoperative CT/MR images to US^{31–35} has been utilized in some centers,^{31–33} using common landmarks such as the liver surface and/or vasculature,^{31–34,36–44} or electromagnetic antenna tracking.³³ These approaches require appropriate and careful clinical setup.³² Data acquisition during breath holds or respiratory gating is essential to improve the accuracy of image fusion.^{31,34} The registration/fusion of the images varies from real-time,³¹ 15–20 s,⁴⁵ 29–74 s,³⁶ to around 5–10 min.⁴⁶ Recent advancements in computing and automatic image analysis have been able to significantly reduce the extra time required for the image registration so that image fusion becomes a feasible option in ablation procedures.^{47–51} This initial rigid registration can be achieved by defining a minimum of three (3) noncollinear common fiducial points in both datasets, or by defining a common plane and single point.⁵² Classical methods for implementing multi-modality image registration use mutual information to measure differences and a rigid transformation to warp the previously acquired data with the real-time US B-mode image.⁵³

Similar rigid image registration techniques to fuse US with other whole-patient imaging modalities are currently available on several commercial US systems. These systems incorporate CT/MR/PET imaging modalities and fuse them with real-time US. Position tracking for these systems uses an electromagnetic system that incorporates a source transmitter along with sensors to track the position of the transducer and/or ablation antenna.^{29,54} One example is the GE Logiq E9 that uses volume navigation (V-Nav) where the previously acquired diagnostic volume must first be uploaded in DICOM form onto the US machine.⁵² However, manual steps for registration of real-time US with the previously acquired images are required for initial synchronization. The spatial accuracy and clinical efficacy of the V-Nav system from the GE Logiq E9 was evaluated by Hakime et al.³² The experiment specifically compared the choice of landmarks (three anatomical vs. two anatomical and one non-anatomical landmark) and whether anesthesia had any effect on the overall error. The maximum absolute difference between the location of liver nodule in CT and US was used as an accuracy metric. The highest accuracy of 1.9 ± 1.4 mm³² was achieved with the patient under general anesthesia and both US and CT performed immediately one after the other before registration, which reduced to 7.05 ± 6.95 mm with time between the two modalities.³² The choice of landmarks used also had a significant effect on the registration accuracy. When comparing landmark selection, the maximum absolute difference increased by 5.2 mm when no non-anatomical landmarks were included. This is likely because non-anatomical landmarks are more easily identifiable in both US and CT. There was also increased registration error of 7.05 ± 6.95 mm when CT was not performed immediately before registration and this is attributed to patient respiration and involuntary distortion of the liver.

In addition, many other manufacturers have their own implementation of image fusion software such as Siemens eSieFusionTM, Samsung SfusionTM, and Philips AIUSTM. SfusionTM advertises three different modes for data fusion on the US system: manual landmark-based registration, sweeping auto-registration, and positioning auto-registration. The positioning auto-registration is the simplest and fastest method on the Samsung system for automatically registering the two datasets. The image fusion system from Philips uses two methods for performing the automatic registration, which use an US sweep. The sweep provides three-dimensional (3D) information about the liver vessel trees and/or the liver surface. Registration uses this information to automatically find the best fit between the preloaded CT/MRI dataset and the real-time US B-mode image.

Outside of commercially available systems, successful reports that automatically register US with other imaging modalities have been discussed in the literature. Wein et al. in Ref. [38] present the foundation for a fully automatic CT to US fusion technique. The method formulates an optimization problem that seeks to maximize the similarity between data from a 3D US sweep and corresponding simulated US data from CT images. The experimental setup included a

magnetic tracking system, which provided information about the transducer orientation. Experimental results across 25 patients showed a mean fiducial registration error (FRE) and target registration error (TRE) of 9.5 and 8.1 mm, respectively using an affine transformation. Here, FRE represents the error associated fiducial landmarks while TRE is associated with the targeted tumor region.

MRI has also shown success in slice registration.^{48,55–57} Real-time US-MRI fusion imaging was evaluated by Kaplan et al.⁵⁸ for use in trans-rectal biopsies with the goal of increasing the yield of the biopsy procedure. Image fusion was performed using six common fiducial markers on both the US and MRI images. An affine transformation was used to minimize the distance between the fiducial points and align the 3D MRI image set with US. The authors presented the results only as a proof of concept and assessed the accuracy of the image fusion quantitatively using a split window of US/MRI data. Schlaier et al.⁵⁹ assessed the fusion of US/MR using a physical model compatible with both imaging modalities. The authors constructed a plastic box that consisted of water, wooden arrowheads, and plastic reflective spheres. A paired point registration method was used by the navigation system to fuse US with MRI and to automatically determine registration accuracy. Overall, the paired point registration method was highly accurate and achieved an overall accuracy of 1.08 ± 0.61 mm (mean $\pm$ SD) for the reflective spheres.

In this paper, we derive an algorithm for extracting, matching and collocating a CT virtual slice⁶⁰ to clinical US B-mode images from 20 patients during ablation procedures under general anesthesia.⁴⁶ We utilize the non-anatomical microwave antenna to initiate and colocate possible planes and finalize the virtual slice⁶⁰ with anatomical landmarks common between the CT and US B-mode image. Finally, an affine transformation is utilized to register the CT virtual slice to the US B-mode imaging plane.

2. MATERIALS AND METHODS

2.A. Human patients

This study was conducted under a protocol approved by the institutional review board at the University of Wisconsin-Madison. Informed consent was obtained from patients scheduled for a minimally invasive MWA procedure prior to data acquisition for EDE. We report on 20 patients with HCC or liver metastases with both US and CT data sets. All patients in this study were placed under general anesthesia prior to the procedure.

2.B. Microwave ablation procedure

Microwave ablation procedures were conducted with a 2.45 GHz MWA system and 17-gauge antennas (Certus 140, Neuwave Medical Inc., Madison, WI, USA). One or more antennas were used depending on the number and size of tumors being treated. Antennas were generally guided using

real-time US if the tumor could be visualized. Most ablations were performed using 65 W for 5 min.

In our facility, MWA procedures are performed on a CT imaging table, such that the patient is in the same position for all treatment and imaging studies performed in the ablation suite. Patient position was determined during pre-ablation treatment planning based on the location of the tumor(s). All 20 patients used for this analysis from this cohort were placed in the supine position.

2.C. Imaging before and after microwave ablation

US B-mode imaging with a C1-6 curvilinear array was utilized for ablation antenna placement (GE Logiq E9, GE Healthcare, Waukesha, WI, USA). After the antenna was located in the tumor based on the treatment planning guidelines, it was locked into place using the cyro-lock feature of the system. CT scanning without contrast was then performed to verify antenna placement accuracy before initiating the ablation procedure (750HD, GE Healthcare).

The standard MWA procedure performed in our facility includes immediate postablation assessment using CECT while the patient is still under anesthesia. For the patients in this dataset, we also acquired radiofrequency (RF) data for EDE-based strain imaging induced through a manual perturbation of the antenna immediately before and after the ablation procedure. We anticipated that tissue deformation due to respiratory and cardiac motion between the capture of both datasets would be present. Based on previous image fusion results, the expected accuracy of registration method in this setting should fall between 1.9 and 6.55 mm.^{32,38}

2.D. CT “Virtual Slice” Selection⁶⁰ and Registration

In this section, we describe computation of the 2D CT virtual slice⁶⁰ that would correspond to the clinical B-mode image used for antenna placement. A flow-chart of the registration processing sequence is shown in Fig. 1. The first registration stage performs a rigid registration to colocate both US B-mode and ablation-suite CT virtual slice along the same imaging plane. The second stage estimates an affine transformation for deformable registration of the CT virtual slice to the US B-mode imaging plane. Although the manual selection process is time-consuming, it provided robust registration using the non-anatomical microwave antenna along with anatomical landmarks for initial transformation irrespective of B-mode US quality, where automatic registration schemes often fail. Under the assumption that patient movement is minimal between the preablation and postablation CT scans, the estimated registration and affine transformations from the preablation CT volume could be applied to postablation CT volume to visualize the ablated tumor dimensions in the same plane as the US B-mode as shown by dashed lines in Fig. 1. This previous statement is not a primary objective of this proposed work but is left for future validation.

FIG. 1. Strategy for colocalizing and registering a computed tomography (CT) virtual slice from the 3D CT volume to the ultrasound B-mode imaging plane.

A plane in 3D space can be represented by the point-normal formula shown in Eq. (1)

$$\bar{n}_{VS} \cdot (r_1 - r_2) = 0 \quad (1)$$

where $\bar{n}_{VS}$ denotes the normal vector to the desired plane and r_1, r_2 are arbitrary points on that plane. $\bar{n}_{VS}$ can be calculated from the cross-product of any two vectors passing through a common reference point r_0 on the plane, as:

$$\bar{n}_{VS} = (r_1 - r_0) \times (r_2 - r_0) \quad (2)$$

To locate the imaging plane from the CT volume slices, we need to derive two vectors on the US B-mode image plane that encompasses the antenna as illustrated in Fig. 2. An example of this approach on 3D CT volume data is shown in Fig. 3(a) and 3(b) with Fig. 3(c) showing a blown-up CT volume encompassing the antenna highlighted in green. For vector selection, the most intuitive choice was one vector along the orientation of the antenna and another vector along the direction of the antenna to transducer. The vector denoting antenna orientation, $\bar{N}_1$, was derived based on the antenna tip-point and centroid of the antenna. The second vector, $\bar{N}_2$, was derived by the projection of $\bar{N}_1$ onto the normal vector of the xz-plane (see Fig. 2) using the following vector reduction formula,

$$\bar{N}_2 = \bar{N}_1 - \langle \bar{N}_1, \bar{n}_{xz} \rangle \bar{n}_{xz} \quad (3)$$

The reasoning is that US transducer position will be parallel to the xz-plane under the assumption that the US transducer was positioned perpendicular to the patient surface.⁶¹ This slice was obtained using the normal vector parallel to the xz-plane, also representing the plane tangential to the

FIG. 2. Strategy for obtaining the computed tomography (CT) virtual slice from the 3D CT volume. Potential slice planes were obtained by varying a normal vector around antenna to acquire the plane corresponding to the ultrasound B-mode image. [Color figure can be viewed at wileyonlinelibrary.com]

patient surface where $\bar{N}_1$ and $\bar{N}_2$ are shown by green and blue arrows, respectively as illustrated in Fig. 2 and reproduced with actual data in Fig. 3(e). A vector selection example on patient data is shown in Fig. 3(e) where $\bar{N}_1$ and $\bar{N}_2$ are shown by green and blue arrows, respectively. The virtual slice generated by these two vectors in the CT image space does not necessarily define the US B-mode image plane. The final virtual slice was identified by rotating the antenna plane until three clearly visible anatomic landmarks (e.g., undeformed liver surface or vasculature points) match in both images.

Landmarks are selected based on common visualization in both the US and CT slices. The virtual slice selected is primarily based on ablation antenna visualization over the 3D CT volume followed by rotation to align these common anatomical landmarks in both CT and US slices. This task does not require significant proficiency for the operator. For affine transformation, although the accuracy of selected points may vary based on operator's proficiency level in viewing these images, more than three set of noncollinear points can be selected so that the affine transformation can be found using least squares formulations. Example images for landmark selection are illustrated in Fig. 3(d) and 3(e) indicated by the red sectors.

2.E. Affine transformation for virtual slice refinement

The final estimated virtual slice, shown in Fig. 3(g), may not coregister well to the US B-mode image plane due to distortion from the US transducer placement (US imaging deforms the skin surface to make skin contact without an air interface) as well as respiratory and physiological motion. To

FIG. 3. An example workflow using patient image data illustrating the ultrasound (US) B-mode – computed tomography (CT) registration process. (a) Transverse slice with the antenna location highlighted in blue. (b) CT volume as a stack of transverse slices with the antenna location highlighted within the blue cube. (c) Zoomed in region of the antenna location in the CT volume with the antenna shown in green. (d) Comparable US B-mode image from patient after ablation with outline of sector image highlighted in red. (e) Zoomed in CT volume with the two vectors creating the virtual slice plane highlighted in green and blue for the vector along the direction of the antenna and the vector projected onto the xy-plane, respectively. An example sector US outline is transposed onto this plane which is rotated about the antenna to match three clearly visible anatomic landmarks. (f) The resultant CT virtual slice after affine transformation in green transposed onto the red US B-mode image. (g) The entire CT virtual slice collected preablation. (h) The CT virtual slice estimated post-ablation. [Color figure can be viewed at wileyonlinelibrary.com]

correct for this distortion, an affine transformation was used in the second stage of the registration pipeline. Mathematically an affine transformation can be represented by Eq. (4).

$$\begin{bmatrix} x' \\ y' \\ 1 \end{bmatrix} = \begin{bmatrix} s_x \cos(\theta) & -h_x \sin(\theta) & t_x \\ h_y \sin(\theta) & s_y \cos(\theta) & t_y \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} x \\ y \\ 1 \end{bmatrix} \quad (4)$$

where s is scale, h is shear effects, and t is translation for the coordinates in CT images, respectively, and θ allows for

rotation correction. The parameters of the transformation matrix were estimated by registering the anatomical points on both CT and US images used in the virtual slice selection, and the transformation matrix was applied on the CT virtual slice to refine the registration to the US B-mode image plane. The resulting transformed CT virtual slice is shown overlaid in green on the underlying red US B-mode in Fig. 3(f). The ablated region shape and dimension on the registered CT images is considered as the gold standard for assessment of a successful ablation procedure and used for comparison to the

US B-mode and EDE results. Assuming no gross patient movement, the estimated CT virtual slice and affine transformation parameters can be utilized on the postablation CECT to visualize the postablation region shown in Fig. 3(h).

2.F. Registration error analysis

The TRE is a measure used to validate the performance of point-based image registration accuracy.^{62,63} TRE was measured by comparing the Euclidean distance between corresponding identified points in both the image sets using Eq. (5).

$$TRE = \sqrt{(p_{xy} - T(q_{xy}))^2} \quad (5)$$

where p_{xy} and q_{xy} are the x and y coordinates of a target registration point in the fixed image (US) and moving image (CT), respectively. T is the estimated affine transform that attempts to register the CT image plane with US. Therefore, TRE quantifies how close features points in the transformed CT image are to the same features in the US B-mode image after registration. The points used for the registration error analysis include points near the antenna along with other anatomical landmarks visible in both the US and CT images. A minimum of three feature points were used for each pair of registered images.

3. RESULTS

Figure 3(d), 3(f)–3(h) illustrates the image fusion results for a patient starting with the B-mode image in Fig. 3(d) and the corresponding CT virtual slice in Fig. 3(g). Affine transformation to warp the virtual CT slice and merge it with the B-mode image is shown in Fig. 3(f). Note the complete overlap of the ablation antenna in Fig. 3(f); the green map shows the CT virtual slice while the red image denotes the B-mode image. The final image shows the corresponding postablation CECT virtual slice in Fig. 3(h), which is obtained using the same slicing parameters used for the CT-US registration and used to obtain Fig. 3(g), respectively.

In a similar manner, Fig. 4 presents image fusion results for a second patient, with the B-mode image in Fig. 4(a) and the corresponding CT virtual slice in Fig. 4(b). The affine transform used to align and fuse the virtual CT slice with the B-mode image is shown in Fig. 4(c). Note that the ablation antenna in Fig. 4(c) lies completely on the ablation antenna visualized in Fig. 4(a). The liver surface seen in Fig. 4(a) and 4(b) also coincide. The green image depicts the CT virtual slice while the red image depicts the US B-mode image. The corresponding postablation CECT virtual slice is shown in Fig. 4(d), which is obtained using the same slicing parameters used for the CT-US registration to obtain Fig. 4(b), respectively. Both results demonstrate registration by the agreement of both anatomical and non-anatomical features.

The identified landmarks used for the registration error analysis in both image data sets include points near the antenna along with other anatomical landmarks visible in

both the US B-mode and corresponding CT virtual slices. A minimum of three different (independent) manually selected feature points were used for each pair of registered images for both registration and validation. The mean $\pm$ SD registration error was 1.85 ± 2.14 mm using both anatomical and non-anatomical landmarks.

Linear regression analysis was performed for the feature points to determine correlation between the results for landmarks on US and CT. For each image set, the Euclidean distance from the origin was calculated for each feature point location in both the CT virtual slice and the US B-mode image as shown in Fig. 5. The results show that points with the highest correlation come from non-anatomical landmarks [Fig. 5 (a)] due to the rigid structure of the ablation antenna. Anatomical feature points also show good agreement on US and the transformed CT image but suffer from small differences due to minor tissue deformations in the US B-mode as displayed in Fig. 5(b). Figure 5(c) shows the high correlation ($R^2 = 0.98$, $P < 0.001$) between the complete set of feature points on corresponding CT and US B-mode images. These results are anticipated since the non-anatomical landmark (antenna) is clearly visualized in both modalities and does not deform with its dimensions remaining constant in both modalities. In addition to Fig. 5, Figure 6 depicts the registration error reduction obtained using the affine transformation after the initial rigid registration (virtual slice) using five independent feature points for each registered data set. In addition, Fig. 6 presents registration errors incurred after the rigid registration (virtual slice) stage, after the corresponding rigidly registered image pairs were appropriately scaled. Fig. 6 also presents registration error results after rigid translation, rotation and the combination of scaling, rotation and translation (SRT). Note that registration errors present after rigid registration are significant when compared to registration errors after the affine registration stage. Utilizing the

FIG. 4. An example of ultrasound (US) B-mode – computed tomography (CT) registration result for a second patient (a) US B-mode prior to ablation with antenna visible. (b) Colocated CT virtual slice. (c) The registered CT virtual slice after affine transformation in green transposed onto the red. (d) The co-located CT virtual slice estimated post-ablation. [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 5. Computed tomography (CT) virtual slice registration accuracy to ultrasound (US) B-mode. (a) Plot of non-anatomical registration points (antenna) from US and CT. $R^2 = 0.98$, $P < 0.001$. (b) Plot of anatomical registration points on US and CT. $R^2 = 0.98$, $P < 0.001$. (c) Final plot of both anatomical and non-anatomical points on CT and US. $R^2 = 0.98$, $P < 0.001$. [Color figure can be viewed at wileyonlinelibrary.com]

affine transformation allows for significantly lower registration errors.

4. DISCUSSION

We present a registration method for fusing CT and US images from percutaneous liver ablation procedures. The method is based on a virtual slicing technique, which yields several possible CT image matches containing the ablation antenna. The slice corresponding to the US B-mode image plane is determined by performing additional control point registration using anatomical markers and an affine transform. Fiducials used to estimate registration errors included antenna tip and anatomical landmarks (liver surface, vasculature) visible on both US and CT images. A minimum of three fiducials were used for each pair of registered images.

Registration of US to CT images can be directly applied to ablation area delineation results obtained with electrode displacement elastography for comparative performance analysis in human subjects. Yang et al.²¹ used the contrast and contrast-to-noise ratio to measure the feasibility of EDE for patients with HCC since no ground-truth images were available. These metrics, by definition, require a defined region of interest (ROI) in and around the ablation region on the image. For EDE, this region is often chosen to be elliptical based on the expected shape of a typical ablated region. However, this is not always the case. This variation in the calculation method for CNR makes it difficult to compare the performance of elastography algorithms across research studies and consequently imposes limitations on the reproducibility of this evaluation process. The method proposed in this paper offers a solution to this problem by obtaining a ground-truth “virtual” CT image, which can be compared with results from EDE. Comparison between ablation areas and dimensions between EDE and CECT images can therefore be performed, enabling a comparison to the current clinical gold standard.

As the implementation in this paper is a proof of concept, the computational requirement currently is not significant when compared to the time required for the manual selection of the virtual slice and feature points for registration. Most of the processing time is determined on operator’s experience for selecting the virtual slice and feature points for registration. To improve registration accuracy, users can select more than the minimum three points to obtain a least squares solution to the affine transformation. Approximate processing time for registration was primarily dictated by manual selection of features used for rigid registration of the virtual slice at 15 min, while the rigid registration itself and affine transformation steps take less than 1 min combined.

Schlaier et al.⁵⁹ on a phantom achieved an overall accuracy of 1.08 ± 0.61 mm (mean $\pm$ SD) for the registration of

FIG. 6. Registration error distribution with translation, rotation, scaling rotation and translation (SRT) and affine transformation after the initial rigid registration (virtual slice) stage. Error is plotted on a log scale along the y-axis. The ** denotes statistically significant improvement with affine transformation.

reflective spheres between US and MR. However, these results were for a phantom with no motion errors. Wein et al.³⁸ across 25 patients showed a mean FRE and TRE of 9.5 and 8.1 mm, respectively. Hakime et al.³² using the GE Logiq E9 V-Nav system obtained their best accuracy of 1.9 ± 1.4 mm with the patient under general anesthesia and both US and CT performed immediately (no motion artifacts). However, the registration accuracy reduced to 7.05 ± 6.95 mm when some time had elapsed (with motion artifacts) between the two modalities. In addition, the maximum absolute difference increased by 5.2 mm when no non-anatomical landmark was included in the registration process. Additionally, Banerjee et al.⁴³ reported an automated registration error of 3.6 mm using 300 matching points with a mean TRE of 3.9 mm for affine registration and 4.2 mm for rigid registration. For the patients studied in this paper, postablation CT was performed under general anesthesia after the ablation procedure with US for antenna placement performed prior to the CT procedure used to verify antenna position. We obtained a mean registration error of 1.85 ± 2.14 mm (mean $\pm$ SD) over 20 independent data sets using a minimum of three fiducials for each registered image pair (CT and US). This registration error lies between the errors reported in Hakime et al.³² and compares favorably with the registration errors reported by Wein et al.^{32,38}

We report that our low registration error was achieved by our two-stage registration process; first utilizing rigid, non-anatomic landmarks provided by the antenna to colocate images from the two modalities, thereafter incorporating visible anatomic landmarks for affine transformation and final registration. This is visualized in the merged images where the combined anatomic and non-anatomic landmarks depict the close registration of the common image features such as the antenna and liver surface. We obtained a mean

error of 1.85 ± 2.14 mm (mean $\pm$ SD) over 20 independent data sets.

The research presented in this paper describes utilization of the “CT virtual slice” concept previously described by Wu et al.⁶⁰ The CT virtual slice is obtained from a 3D CT volume by collocating the imaging planes corresponding to a 2D clinical US B-mode image used to guide placement of the ablation antenna into the liver tumor to be ablated. Following computation of the colocated virtual slice from the 3D CT volume comprised of cross-sectional images or slices through the body, we account for local deformations present in these due to differences in the respiratory and cardiac phases and local deformation present with the US transducer. An affine transformation is used for deformable registration of the colocated CT virtual slice to the B-mode image. This work paves the way for direct comparison of ablation areas determined using US based electrode displacement elastography to CECT, the current gold standard for assessing minimally invasive thermal ablation of tumors in patients.

4.A. Limitations

Manual selection of the anatomical landmarks is a limitation our approach, and we will pursue automated approaches for the same. Future work will investigate automating the entire registration process, investigating accuracy of estimated registration and transformation parameters for postablation CECT volumes, and comparison between EDE and CT virtual slices.

5. CONCLUSION

This paper presented a registration method for fusing CT and US images from percutaneous liver ablation procedures. We used a virtual slicing technique to find and colocate the slice corresponding to the US B-mode image plane, followed by affine registration using the antenna and anatomical markers. We obtained a mean registration error of 1.85 ± 2.14 mm (mean $\pm$ SD) over 20 independent data sets using a minimum of three fiducials for each registered image pair (CT and US). This work is directly applicable to EDE performance analysis by providing the gold standard clinical CECT imaging plane for measuring the accuracy of EDE results.

ACKNOWLEDGMENTS

This work was funded by NIH grant 2R01 CA112192. Mr. Robert Pohlman, M.S. is funded on T32 CA009206.

CONFLICTS OF INTEREST

The authors have no conflicts to disclose.

FINANCIAL DISCLOSURES

Dr. Christopher Brace, Ph.D is a consultant for Johnson & Johnson (Neuwave) and a shareholder of Elucent Medical

and Symple Surgical. Dr. Ziemlewicz M.D is a consultant for Ethicon, and a consultant and stockholder for Histosonics, Inc. Dr Varghese, Ph.D has a research agreement with Siemens Medical Solutions for the use of the ultrasound research interface.

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