A Global CT to US Registration of the Lumbar Spine

Simrin Nagpal[a], Ilker Hacihaliloglu[b], Tamas Ungi[a], Abtin Rasoulian[b], Jill Osborn[c], Victoria A. Lessoway[d], Robert N. Rohling[b], Daniel P. Borschneck[e], Purang Abolmaesumi[b], Parvin Mousavi[a]

[a] School of Computing, Queen's University, Kingston, ON, Canada;
[b] Department of Electrical and Computer Engineering, University of British Columbia, Vancouver, BC, Canada;
[c] St. Paul's Hospital, Vancouver, B.C., Canada
[d] RDMS Department of Ultrasound, BC Women's Hospital, Vancouver, B.C., Canada
[e] Department of Orthopedic Surgery, Kingston General Hospital, Kingston, ON, Canada

ABSTRACT

During percutaneous lumbar spine needle interventions, alignment of the preoperative computed tomography (CT) with intraoperative ultrasound (US) can augment anatomical visualization for the clinician. We propose an approach to rigidly align CT and US data of the lumbar spine. The approach involves an intensity-based volume registration step, followed by a surface segmentation and a point-based registration of the entire lumbar spine volume. A clinical feasibility study resulted in mean registration error of approximately 3 mm between CT and US data.

Keywords: registration, lumbar spine, ultrasound, computed tomography

1. INTRODUCTION

Spinal pain is a common discomfort in the average workforce with dramatic effects on performance, leading to significant cost to the society. First-line therapies focus on conservative non-operative care through custom physical therapy programs, and medications. Although most patients with chronic pain are adequately managed with adjusting lifestyle or oral medications, adequate pain reduction is only achieved in half of the patients, and a full 21% are non-responsive to opioid therapy[1]. Long-term results are even less favorable reporting only 16.7% sustained pain relief. While other studies reported better results[2] with analgesic therapy, the important point is that oral medication is not an appropriate or effective therapy for a large fraction of patients. When solo oral medication fails, some form of spinal injection is considered. With most spinal injections, a steroid medication is injected in order to reduce inflammation in target regions surrounding spine.

The current gold standard for these procedures is fluoroscopy or CT for guiding the injection. It is now universally acknowledged that under radiological-guidance, spinal injections are safe and efficacious for treating pain emanating from and around the spine. There is also general agreement that CT provides better imaging of the target anatomy owing to better visualization of soft tissue. Fluoroscopy, however, is often faster and more flexible in the hands of many interventionalists, and so it may provide a higher clinical throughput. However, radiological needle guidance for anesthesia delivery has significant drawbacks, particularly the use of ionizing radiation and the need for a specialized pain management clinic with access to fluoroscopy.

In search of a more accessible and non-ionizing imaging alternative, ultrasound (US) guidance has enjoyed a recent resurgence, with research looking into pre-puncture US to measure the depth from skin to epidural space[3], help decide the puncture site to reach the desired intravertebral level[4], and for real-time guidance of needle insertion[5,6,7]. Current evidence[8,9] suggests that while ultrasound-guided procedures have a similar success rate to those of conventional percutaneous spinal interventions, they lead to several patient-oriented benefits, such as reduced procedure time, much less radiation, fewer needle passes, and less discomfort.

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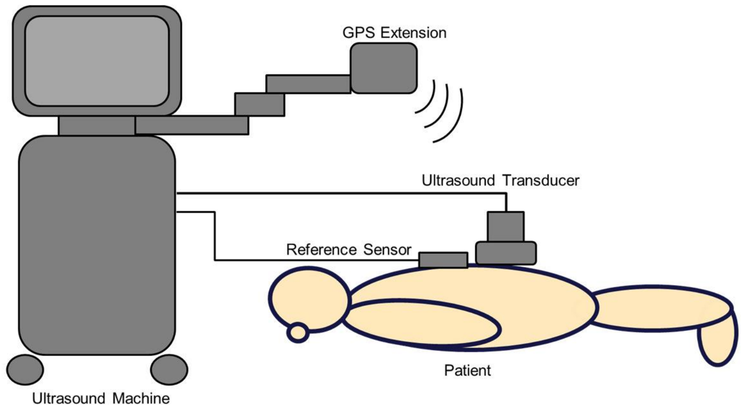

Figure 1: Data acquisition system for tracked US. SonixTouch (Ultrasonix, Richmond, BC) ultrasound scanner; GPS extension (Ascension DriveBay electromagnetic tracker); tracked C5-2 curvilinear transducer (Ultrasonix); 3D position sensors affixed to the transducer and patient. This figure is modified from[19].

Despite this surge of interest, ultrasound has not become the standard-of-care for pain management and obstetrics/general surgery, because of issues related to poor quality depiction of anatomical features, poor depiction of the needle, and the difficulty in interpreting images by novice ultrasound operators (i.e. many anesthesiologists). These inherent limitations of ultrasound imaging are even more pronounced in the case of obese patients, in whom the greater distance from skin to target leads to lower image quality at the target location.

To overcome some of these problems and provide improved guidance, US images have been registered to a virtual representation of anatomical targets obtained from pre-procedure data such as CT[6,10,11,12]. While these methods have demonstrated great potential in registration of US and CT images of phantom and animal models, and human cadaver tissue, to best of our knowledge, no prior technique has been evaluated on data obtained in vivo from human subjects. The only exception is the work by Winter *et al.* [13] where registration of a single vertebra is achieved using clinical data.

In this paper, multiple-vertebrae of the lumbar spine in the preoperative CT is globally registered to the intraoperative US followed by point-based registration. We provide details of the methodology, and present experimental results on the registration of CT and US data obtained from three patients. We report a registration accuracy of about 3 mm, which should be sufficient to guide the needle to the targets surrounding the spine anatomy, such as the facet joints.

2. MATERIALS AND METHODS

2.1 Data Collection

Three human data sets of the lumbar spine are used to test the proposed registration approach. Approval was obtained from Institutional Research Boards (IREB). Informed consent was acquired from the subjects that participate in the study, and we only recruited subjects that have preoperative CT data. We use the methodology presented by Rasoulian *et al.*[14] for automatic labeling of the lumbar levels in CT. The hardware used to acquire the US images and associated tracking data is composed of a SonixTouch US scanner (Ultrasonix, Richmond, BC, Canada) with a Guidance Positioning System (GPS) extension (Ascension DriveBay EM position tracker, Burlington, VT) and a C5-2 GPS curvilinear US transducer (Ultrasonix). Finally, a Model 800 EM tracking sensor (Ascension) is used as the patient coordinate reference. This reference sensor is affixed on the patient's skin above the T12 vertebra. The GPS extension has an adjustable arm that attaches to the US machine. This allows the position of the EM transmitter to be as close to the reference sensor as possible. The US transducer is tracked by the EM transmitter through an embedded pose sensor. Figure 1 shows the schematics of data acquisition. The setup time is minimal (around five minutes) in order to adhere to the clinical environment and avoid any changes to the current clinical practice. US data acquisition is performed with the subject in a prone position, in a sagittal zigzag pattern. The ultrasound transducer is positioned at the left L1 transverse process moving slowing across to the right; it is then moved down to L2 and across in the opposite direction. US data is acquired smoothly and slowing, while keeping contact with the subject's skin.

2.2 CT to US Registration Pipeline

The registration pipeline in this paper includes an intensity based registration followed by surface extraction and point-based registration of CT and US volumes.

An automatic intensity-based registration, that uses voxel information, is applied to align the entire lumbar spine volumes from CT and US, hence referred to as a global registration approach. Images of CT and US volumes of the spine differ greatly, especially in terms of intensity values. Given an intensity-based registration, where correspondence is found through the voxel values, a preprocessing step is taken to minimize the variability of the intensity values in the CT and US images as much as possible local phase bone features are automatically extracted from both CT and US data sets to enhance the bone surfaces using the novel approach developed by Hacihaliloglu et al.[15]. The processed volumes with enhanced bone surfaces are the input to the intensity-based registration. Only the CT bone surfaces visible in the US are used for registration.

The algorithm used for intensity-based registration is the general registration module in the open-source medical imaging software 3D Slicer[16], BRAINS. Rigid transformation is selected to perform a rigid intensity-based registration. Mutual information is used as the similarity metric for registration and the other parameters of the algorithm are set to the defaults within 3D Slicer BRAINS module. Optimal alignment is achieved when the amount of pixel intensity information the input images contain about each other is maximized.

Following intensity-based registration, a point-based registration of the CT and US data is performed to further improve the alignment of the two volumes. For point-based registration, point sets are required from both volumes. The CT point set is obtained from the CT bone surface enhanced through local phase filtering as discussed above. However, to extract the US point set, bone surface segmentation is required. We use an existing algorithm by Foroughi et al.[17] for this purpose. In this algorithm, the bone surface pixels are enhanced by a combination of high acoustic impedance and acoustic shadowing, followed by optimization of a cost function using dynamic programming[16]. Due to the high acoustic impedance of bone in US, a pixel that is most likely bone will have a high intensity value, since there is a strong reflection when the US beam hits a bone surface. Due to this reflection, acoustic shadowing should also be present below the bone. Pixels with low intensities are therefore expected below a pixel of high intensity, if the pixel of high intensity is a part of the bone surface. Foroughi *et al.* tested their algorithm on cadaver US data. For clinical data, this algorithm results in substantial noise in bone surfaces. We modified this approach so that instead of using intensity values of bone surfaces for segmentation, we use the intensity values from the phase filtered US image of the bone surface. Local phase filtering is performed using the approach developed by Hacihaliloglu *et al.*[15], as above. This results in less noise in the segmented bone surfaces, and makes it possible to register the CT and US volumes in the next step.

After point sets are selected from CT and US data, a global and a piece-wise point-based registration of the CT and US volumes is performed using point set correspondences, solely. The Coherent Point Drift (CPD) method developed by Myronenko *et al.*[18] is applied for this purpose. This approach uses probability density estimation to find corresponding points between the CT and US rather than finding the closest points between the two point sets as is done in Iterative-Closest-Point-based (ICP) approaches. One of the benefits of this choice of point-based registration is that it has a closed-form solution and can be solved in a finite number of operations[17].

2.3 Validation

Anatomical landmarks on the lamina of each vertebra are selected and placed on the US images, and are used as gold standard in lieu of fiducial markers (subjects have pre-existing CT volumes hence fiducial marking was not possible). Two operators with spine anatomy expertise chose these anatomical landmarks and the results were pooled. We assume the CT and US have the optimal alignment following registration, and that the landmarks points chosen in US correspond to the same landmarks in the CT. Such correspondence is visually confirmed following registration.

To determine the capture range for the registration, 20 tests are performed where the CT and the points representing the lamina landmarks are perturbed by a transformation selected randomly from a uniform distribution of 5º rotation about each axis and 5 mm translation along each axis, and applied to the entire lumbar spine that is visible in the CT. The initial misalignment is determined by calculating the target registration error (TRE) between the original position of the

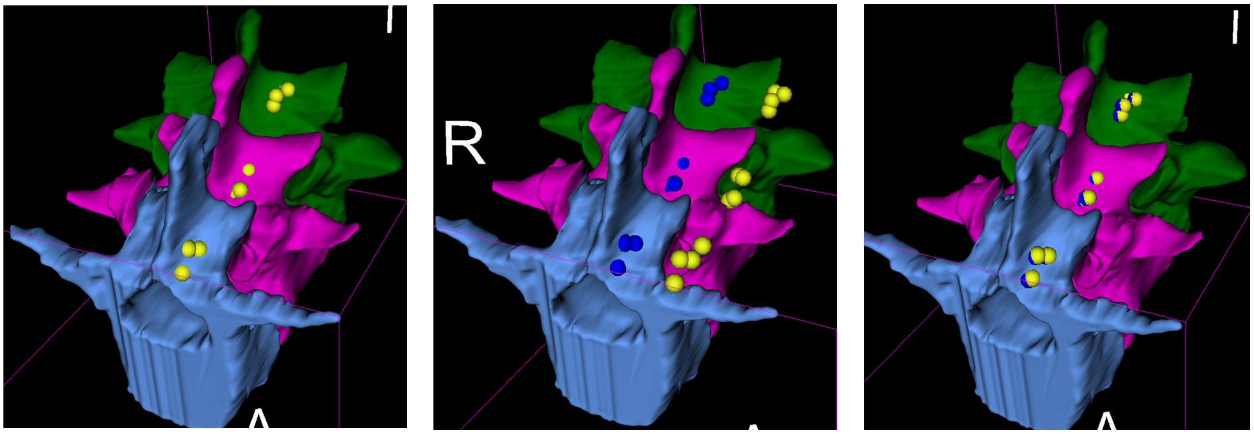

Figure 2: An example of the validation workflow. Original positions of landmarks, following registration, are shown in yellow whereas the randomly transformed positions are shown in blue. (left) CT aligned to original position of US lamina landmarks; (middle) CT and lamina landmarks transformed by initial perturbation; (right) CT and lamina landmarks transformed by CT to US transformation found through registration.

lamina landmark points and the position of the landmarks after the initial perturbation. Registration is then performed and the final TRE is calculated as the root mean square between the transformed lamina landmark points and their original positions. A test run out of the 20 capture range experiments for one subject is shown as an example in Figure 2 to demonstrate how this validation works pictorially.

3. RESULTS

Results for the three patients involved in the study are shown in Figure 3 and Table 1. The subplots of Figure 3 represent the capture range experiments for each patient. The plots show the final TRE between CT and US data, versus the initial TRE following the misalignment. The reported TRE values represent the average of the TRE for each individual vertebrae that is registered for each patient. The mean TRE and the standard deviation, in Table 1, are the average and the standard deviation of the TRE values from the 20 runs of the capture range experiments for each patient, respectively.

Table 1. Mean TRE (mm), and maximum point distance (mm) for CT to US registration.

Dataset	Mean TRE ± std (mm)	Maximum Point Distance (mm)
Patient 1	3.26 ± 0.86	14.99
Patient 2	0.98 ± 0.01	1.20
Patient 3	2.17 ± 1.48	7.06

As seen from the table, the mean for all three patients is approximately 3 mm or less; the standard deviation of mean TRE in patients 1 and 2 is small compared to patient 3. As seen in Figure 3, the final TRE plots for all three patients has a horizontal trend regardless of the initial TRE. The proposed approach is able to register US and CT data with initial misalignments of up to 22 mm. It should be noted that the maximum point distance in Table 1 refers to the maximum distance calculated for any vertebrae, following registration, during the experiments for a patient; this value will not be reflected in Figure 3 as every point in the figure represents the average of TRE from all vertebrae in one capture range experiment of a patient.

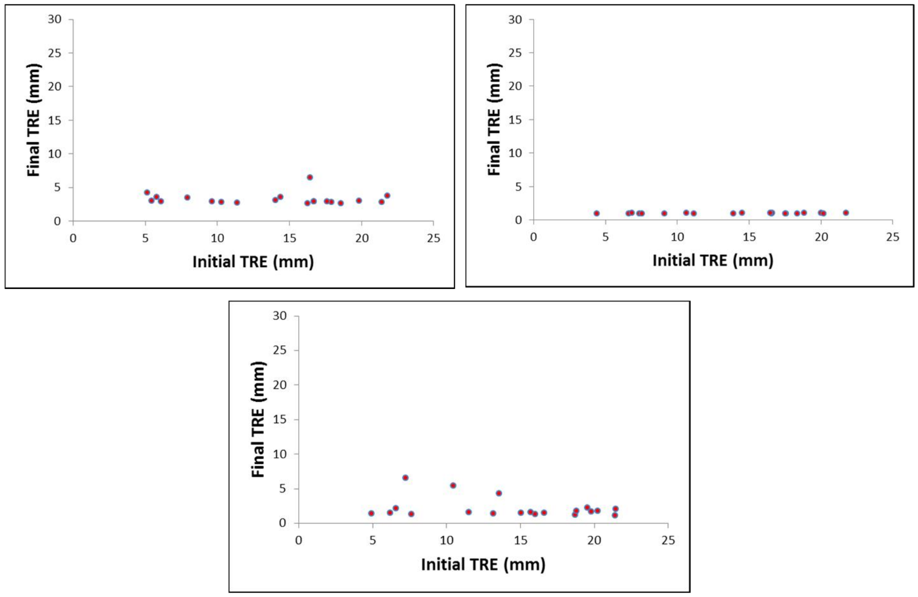

Figure 3: Plot of the final TRE (mm) between the CT and US data after registration given the initial TRE (mm) following the random initial misalignment; (top left) patient 1, (top right) patient 2, and (bottom) patient 3.

4. CONCLUSION

A registration pipeline for the lumbar spine was developed that aligns preoperative CT to intraoperative US using three clinical datasets. By aligning the CT with the US, anatomical information that is not visible in US is provided to the clinician to guide spine needle interventions without having to expose the physician or patient to more radiation intraoperatively. As such, the many benefits of US, such as its real-time imaging, portability and low cost can be capitalized. Further improvements to the registration approach include validation on a larger patient dataset. This is currently in progress. In addition, a GPU or multi-CPU implementation of the proposed methodology could enable us to achieve near real-time performance for clinical application. In addition, a registration module for the pipeline in the open-source software 3D Slicer can be built, where visualization and analysis are already built-in.

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