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ORIGINAL ARTICLE

# Ultrasound-guided spinal injections: a feasibility study of a guidance system

**Abtin Rasoulian[1] · Alexander Seitel[1] · Jill Osborn[3] · Samira Sojoudi[1] · Saman Nouranian[1] · Victoria A. Lessoway[4] · Robert N. Rohling[1,2] · Purang Abolmaesumi[1]**

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**Abstract**
*Purpose* Facet joint injections of analgesic agents are widely used to treat patients with lower back pain. The current standard-of-care for guiding the injection is fluoroscopy, which exposes the patient and physician to significant radiation. As an alternative, several ultrasound guidance systems have been proposed, but have not become the standard-of-care, mainly because of the difficulty in image interpretation by the anesthesiologist unfamiliar with the complex spinal sonography.
*Methods* We introduce an ultrasound-based navigation system that allows for live 2D ultrasound images augmented with a patient-specific statistical model of the spine and relating this information to the position of the tracked injection needle. The model registration accuracy is assessed on ultrasound data obtained from nine subjects who had prior CT images as the gold standard for the statistical model. The clinical validity of our method is evaluated on four subjects (of an ongoing in vivo study) which underwent facet joint injections.
*Results* The statistical model could be registered to the bone structures in the ultrasound volume with an average RMS accuracy of $2.3 \pm 0.4$ mm. The shape of the individual vertebrae could be estimated from the US volume with an average RMS surface distance error of $1.5 \pm 0.4$ mm. The facet joints could be identified by the statistical model with an average accuracy of $5.1 \pm 1.5$ mm.
*Conclusions* The results of this initial feasibility assessment suggest that this ultrasound-based system is capable of providing information sufficient to guide facet joint injections. Further clinical studies are warranted.

**Keywords** Multi-vertebrae model · Statistical pose + shape model · 3D ultrasound · Spine · Registration

## Introduction

Lower back pain is one of the most common medical problems in the adult population. It is estimated that up to 80 % of adults experience at least one episode of debilitating back pain during their lifetime [16]. Facet joint injections of analgesic agents have been used for patients not responsive to conservative management. This procedure is particularly challenging due to the deep location of the target, its proximity to nerve tissue, the oblique entry angle, and the small and narrow insertion channel between the articular processes of the joint. The current standard-of-care for guiding the injection is fluoroscopy, which has significant drawbacks, including the considerable dose of ionizing radiation and the need for access to a specialized room with fluoroscopy equipment [2]. As an alternative, several ultrasound (US)-based systems have been recently proposed for guidance [3,9,18–20], but have not become the standard-of-care mainly because of the difficulty in image interpretation by anesthesiologists unfamiliar with complex

✉ Alexander Seitel
aseitel@ece.ubc.ca

Purang Abolmaesumi
purang@ece.ubc.ca

1 Department of Electrical and Computer Engineering, University of British Columbia, Vancouver, BC, Canada

2 Department of Mechanical Engineering, University of British Columbia, Vancouver, BC, Canada

3 Department of Anesthesia, St. Paul's Hospital, Vancouver, BC, Canada

4 Department of Ultrasound, BC Women's Hospital, Vancouver, BC, Canada

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spinal sonography. Moore et al. [9] introduced an US guidance system for facet joint injections where the US transducer was tracked using an electromagnetic (EM) tracking system. They showed that integration of a virtual computed tomography (CT)-based model of the spine improved the accuracy in needle placement. A landmark-based registration was used to relate CT and US data. To visualize both the target and the needle, the US transducer had to be oriented in the same plane as the needle, but the ideal spine injection site is easily obscured by the US transducer. To address this problem, Ungi et al. [19] added preoperative US snapshots from the target to allow both the transducer and the needle to be placed at the ideal puncture site, i.e., the skin point with the shortest path to the target. The approach proposed by Chen et al. [3] also used a tracked US transducer to acquire a 3D volume of the spine and enhance it with information from a preoperative CT scan. Registration of US and CT data was accomplished by a GPU-accelerated volume-to-volume approach. For guidance, the position of the EM-tracked needle was shown together with the US and CT data. Another approach for using US for surgical guidance was proposed by Yan et al. [20]. They registered preoperative CT images with the US data by aligning the posterior vertebral surface, extracted from US and CT, using a forward and backward scan line tracing method. In all these guidance systems, models were extracted from preoperative images such as from magnetic resonance imaging (MRI) or CT. However, such preoperative images are not usually available and expose the patient to ionizing radiation in the case of CT. A mechanical approach for guiding spinal needle injections has recently been proposed and uses either 2D US [18] or 3D US [8] combined with a mechanical guide attached to the transducer to constrain the insertion path of the needle to a known trajectory. However, image interpretation still remains an issue.

Statistical shape models are an alternative for providing anatomical details after registration with US images and have been previously generated for the vertebrae [1,6,13,15]. Boisvert et al. studied the statistical variations of the relative pose of each two adjacent vertebrae separately. They performed a shape analysis of the entire vertebral column and proposed an algorithm for registration of their pose model to radiographic images. Khallaghi et al. [6] built separate shape models for each vertebra and, by incorporating a biomechanical model to constrain the relative pose of adjacent vertebrae, registered the shape models to an US volume. This approach has certain disadvantages. Mainly, the separate reconstruction of each vertebra neglects the many common shape characteristics between different vertebrae of a given subject which may decrease the accuracy of the registration and add to the computational time. To address these problems, we developed techniques for construction of a statistical multi-vertebrae model with a separate statistical analysis of shape and pose of the vertebrae [13]. The pose statistical analysis in contrast to Boisvert et al. is performed on the entire ensemble. The registration of this model with 3D US volumes allows us to augment the US data with anatomical details.

In this paper, we introduce an US guidance system for facet joint injections that allows augmentation of live standard 2D US images with anatomical information from a statistical model of the spine and relating this information to the position of the tracked injection needle. The enabling technology is the fast registration of a statistical shape + pose model of the lumbar spine with an US volume generated from tracked US frames. We evaluate the guidance system in vivo with respect to registration accuracy and clinical validity.

## Methods

The main idea behind the guidance system presented in this work is the augmentation of live 2D US images with detailed anatomical information gained from a statistical model of the lumbar spine and the relation of the augmented image to the EM-tracked injection needle. Figure 1 gives a systematic overview over the guidance approach and its implementation which uses state-of-the-art open-source software libraries to ensure flexibility of the software for dissemination and future extension.

The system is tightly integrated with the US machine, where the "Public software Library for UltraSound imaging research (PLUS)" [7] enables the acquisition of US and tracking data. In section "Materials," it describes the components of the system and the general system setup. The guidance part is implemented as a module within the 3D Slicer framework [12] and features the three main steps that are performed during the guidance procedure and that will be explained in the following sections. The tracked 2D US transducer is used to acquire a 3D volume of the lower back ("US volume acquisition" section), to which a statistical shape + pose model of the lumbar spine is then registered ("Model registration" section). The anatomical information of the model can then be superimposed on the live 2D US frame to guide the tracked needle during insertion ("Guidance" section).

### Materials

An overview of the components involved in the guidance system is given in Fig. 2. The general setup comprises an US machine providing 2D US frames through a curved linear 2D US transducer, an EM tracking system that tracks the US transducer, the injection needle, and a reference sensor, a powerful laptop computer running the guidance interface, and a foot pedal for simple interaction with the guidance system. Calibration of transducer and injection needle—to establish a relation between tracking and US coordinates—

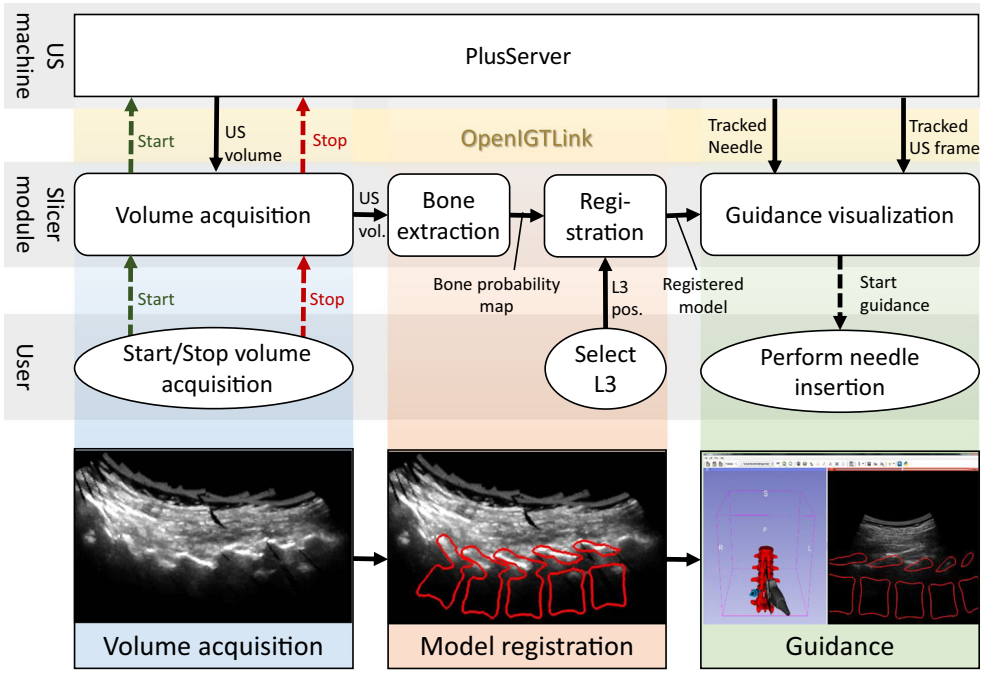


**Fig. 1** Systematic overview of the guidance approach explaining the data flow for the three steps of the guidance workflow: *volume acquisition*, *model registration*, and *guidance*. Image and tracking data retrieval is performed by a server of the "Public software Library for UltraSound imaging research (PLUS)" on the US machine. The guidance software is implemented as a module within the 3D Slicer framework. For volume acquisition, the user triggers a request in the 3D Slicer module, which is forwarded through the OpenIGTLink network protocol to the PlusServer which initiates the reconstruction of an US volume from the electromagnetically tracked 2D US frames until a stop command is issued by the user. The reconstructed 3D volume is then preprocessed by enhancing bone surfaces present in the volume. The resulting bone probability map is the input of the model-based registration algorithm. For guidance, the registered model is shown along with the tracked 2D US frame and the tracked needle in the guidance interface

is performed using the open-source calibration tool fCal[1] provided by the PLUS library. The reference sensor is stabilized in a sensor holder of defined geometry [2] and is attached to the patient's skin 3.5 cm superior of the L1 vertebra in a median position (cf. Fig. 2b). Conventional 2D US is acquired to localize the L1 vertebra by counting up from the sacrum. The reference sensor is used to track any movements of the patient and defines the coordinate system of the volume reconstructed from the 2D US frames, which is mapped into approximately anterior–posterior, lateral, and superior–inferior directions, respectively. The EM field generator of the tracking system is positioned close to the lumbar spine such that all sensors are present in the tracking volume.

### US volume acquisition

Prior to the actual needle insertion, a 3D US volume is reconstructed from the tracked 2D US frames. For this purpose, image and tracking information are continuously acquired by the US machine which acts as a server running the PlusServer application. After the user issues a "Start" signal using the foot pedal, the PlusServer triggers the reconstruction of a 3D US volume using the volume reconstruction functionality[3] available in PLUS and continuously sends the reconstructed volume via the OpenIGTLink network protocol [17] to the 3D Slicer application that visualizes it as an intermediate reconstruction result. The volume reconstruction is stopped by the user of the system by emitting a "Stop" signal using the foot pedal.

### Model registration

Once the volume reconstruction is finished, a statistical shape+pose model of the lumbar spine, presented in our earlier work, is registered to the 3D US volume. We give a brief overview of this method, for a detailed description of the model and its registration to US volumes the reader may refer to [13].

[1] http://perk-software.cs.queensu.ca/plus/doc/nightly/user/ApplicationfCal.html.

[2] http://www.assembla.com/spaces/plus/wiki/Printable_3D_models.

[3] http://perk-software.cs.queensu.ca/plus/doc/nightly/user/ProcedureVolumeReconstruction.html.

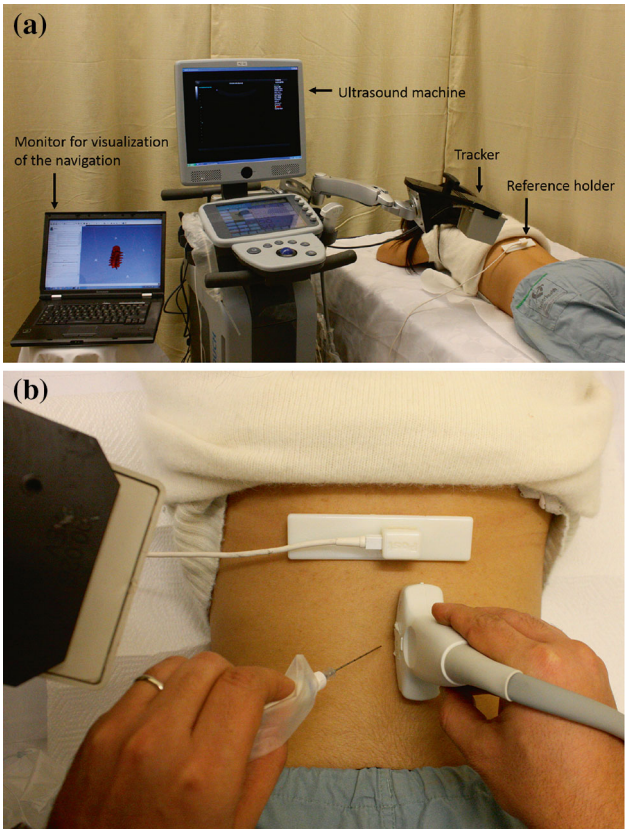


**Fig. 2** Overall setup of the guidance system. **a** The US machine with an attached EM tracking system is positioned close to the patient bed such that the tracking volume covers the area of the lumbar spine. The computer running the guidance software is connected to the US machine through an Ethernet connection. **b** The reference holder is placed superior of the L1 spinous process which is determined by a pre-procedural US scanning

The statistical multi-vertebrae shape+pose model is generated from a training data set consisting of CT scans acquired in supine position which include surface points of the five lumbar vertebrae (L1–L5) acquired from 32 patients. Data acquisition was performed at two institutions, the Vancouver General Hospital (VGH) and Kingston General Hospital (KGH). VGH data comprise seven patients (four males/three females) aged $66.3 \pm 5.8$ years. The KGH data were acquired for other research studies where patients' biometrics was not recorded. Major abnormalities such as strong scoliosis were not present in this training data set. Manual CT segmentation was performed interactively using ITK-SNAP.[4] Pose statistics are separated from the shape statistics since they are not necessarily correlated and do not belong to the same space [11]. Poses are represented by similarity (rigid + scale) transformations which form a Lie group where linear analysis is not applicable. To address this issue, the transformations are projected into a linear space by logarithmic mapping. Next, principal component analysis (PCA) is performed to extract the main modes of variations of the poses. A separate PCA is also used to compute the shape statistics. Note that these analysis are performed on the entire ensemble (including all lumbar vertebrae) which results in common statistics of multiple vertebrae. Such analysis results in a mean shape, $\mu_s$, a mean pose, $\mu_p$, and their modes of variations, $v_s^k$ and $v_p^k$. Linear combination of these modes of variations with the mean shape and mean pose results in a new instance of the ensemble:

$$S = \mathcal{T}(\mu_s, \mu_p, v_s^k, v_p^k, w_s^k, w_p^k), \tag{1}$$

where $w_s^k$ and $w_p^k$ are the weights associated with the corresponding modes of variations. 95 % of the shape and pose variations are captured by the first 25 and seven modes, respectively. The model is capable of reconstructing an unseen observation with distance error below 2 mm using the first 20 modes of variation [14].

Prior to the registration, US images are processed to enhance the bone surface. We follow the technique proposed by Foroughi et al. [4] where pixels with large intensity and shadow underneath are considered as bone surface. Next, the user is asked to locate the center of gravity of the L3 vertebra in the US volume using a single click in the 3D Slicer interface to roughly initialize the model in the 3D US volume. The registration of the model to enhanced US images is performed using a GMM-based registration method. In this iterative technique, the previously generated model boundary points are defined as the centroids of the GMM. The target, i.e., the bone surface enhanced in US images, is considered to be an observation generated by the GMM. The registration is then defined as estimation of proper weights of the modes of variations and a rigid transformation applied to the entire ensemble, to maximize the probability of the GMM centroids generating the target.

The algorithm is parallelized at CPU level, using the Intel Math Kernel Library (Intel, Santa Clara, CA, US). Using this parallelization, the registration, together with the US preprocessing, takes approximately 10 s.

## Guidance

After the model has converged to its optimal, patient-specific shape and pose, the system switches into real-time guidance mode. The navigation display (Fig. 3) then shows a split-screen view with a virtual 3D scene on the left and the live 2D US data on the right. The US slice can now be enhanced with information from the registered model. The virtual 3D scene allows to co-visualize the model position with the position of the US transducer and the injection needle.

[4] www.itksnap.org.

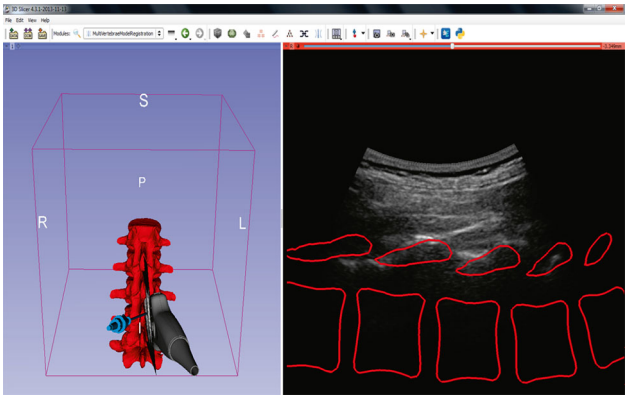


**Fig. 3** Guidance interface implemented as a module within the 3D Slicer application framework. The registered patient-specific model is shown both as an overlay onto the current 2D US frame (*right*) and as a 3D representation in a render window allowing 3D interaction (*left*). The current position of the transducer and its acquired US plane as well as a depiction of the injection needle are displayed with the 3D model

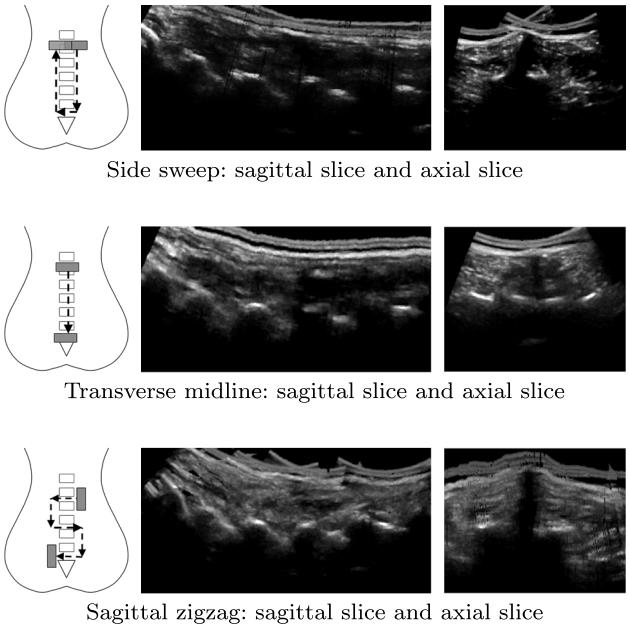


**Fig. 4** Volume scanning protocols as evaluated in this study. In the *side sweep*, the transducer is scanning in transverse orientation and is moved from left of T12/L1 to left of L5/S1 and back to the T12/L1 level on the *right side*. The *transverse midline* scanning acquires the volume by moving the transducer in transverse orientation from the middle of T12/L1 to the middle of L5/S1. The *sagittal zigzag* scan is started at the *left side* of T12/L1 with the transducer in sagittal orientation and moving it in a zigzag pattern down to L5/S1

## In vivo system validation

We present an initial validation of the proposed guidance system including an in vivo evaluation of all three steps of the overall guidance workflow on a total of thirteen subjects from which written consent was obtained (approved under #H13-01968 and #H11-02594). The US volume acquisition and the registration accuracy were evaluated on US data obtained from nine subjects who had prior CT images as a reference measure for the statistical model. The guidance feasibility of the approach is validated on four subjects for which a total of five facet joint injections were performed.

### US volume acquisition

US volumes were acquired from all patients by an expert sonographer using a SonixTouch US machine (Ultrasonix, Medical Corporation, Richmond, BC, Canada) with an integrated EM tracking system (Ascension Technology Corporation, Shelburne, VT, USA) and the tracked curved linear C5-2 transducer (Ultrasonix, Medical Corporation, Richmond, BC, Canada) at 30 frames per second. The calibration of the transducer at different depths yielded an average RMS calibration error of 0.94 mm. Note that the EM tracking system used in this work has a static accuracy of 1.4 mm and 0.5 degrees (Ascension Technology Corporation, Shelburne, VT, USA), which should be considered lower limits, since real-world accuracy will also include some distortion of the EM field from nearby metal objects such as the patient bed.

All US volumes were acquired in prone position which is the most common positioning option for facet joint injections. To provide maximum similarity to the supine position—the subject's posture during CT acquisition—with respect to spine curvature, and also for the subject comfort, a small pillow was placed under the abdomen. Prior to the data collection, a brief sonographic study was performed by the sonographer to tune the imaging parameters such as focus and depth. The sonographer also examined the best possible transducer trajectory for imaging by marking the spinous processes of L1 and S1. To determine the influence of the volume acquisition protocol, three different scanning techniques were performed for the nine subjects where preoperative CT data were available (see Fig. 4 for a graphical illustration):

1. Side-sweep scanning was collected from the subject's top left to bottom left and back to the top right (∼30 s). The transducer angled toward midline and in the transverse plane, 2 cm away from the midline.
2. Transverse midline volume was acquired from top to bottom in the transverse plane (∼15 s).
3. Sagittal zigzag data collection was performed by moving the transducer laterally in the sagittal plane (∼50 s).

For evaluation of the registration accuracy, seven landmarks were picked on the bone surface by one of the authors (AR) in each 2D US image (three on each laminae and one on the spinous process), which took about 20 min for the entire US volume. Note that this step is only needed for validation purposes and is thus not contributing to the overall run time. The landmarks were then transformed to the subject's coordinate

space using the calibration and tracking information which resulted in a set of points resembling the surface of the vertebrae in subject's coordinate space. As expected, only some of the posterior aspects of the vertebrae were visible, i.e., laminae, transverse processes, spinous process, and posterior part of the vertebral body.

### Model registration

For each registration of the model to US volumes, the following measurements were taken:

1. The RMS distance of the manually segmented US points to the model. Although the manual segmentation does not provide a full representation of the vertebrae, it is used to provide a measure of how well the multi-vertebrae model is registered to the US features.
2. The RMS distance between the model and the bone point cloud extracted from the CT images. This measure is calculated to estimate how well the patient-specific shape of the vertebrae registered to the US data matches the patient's anatomy observed in CT images. To this end, each vertebrae of the registered model is separately aligned to the corresponding vertebrae from the segmented vertebrae, using the coherent point drift registration method [10]. Then the RMS surface distance is reported as the shape error.

Capture range experiments were also performed to measure the sensitivity of the algorithm to the initial point selection, i.e., a point around the center of L3, which was marked by the user. To this end, the model is manually well aligned to the target. The center of mass of the L3 vertebra in the model was extracted. Next, a displacement ranging from 0 to 30 mm, in a random direction, was added to the model, followed by registration. Initial displacements were divided into bins with 5 mm width. For each bin, five experiments were performed for each subject and each US volume. The mean distance error between the registered model surface to the manually segmented bone surface points in the US volumes was reported.

### Guidance

The feasibility of the presented approach to guide needle insertions for facet joint injections is evaluated in an ongoing patient study. Until now, data of four subjects have been recorded for a total of five facet joint injections performed by an anesthesiologist following the routinely performed clinical workflow. Our system was only injected into the workflow before the needle insertion to acquire the volumetric US data of the spine and after the injection to record the final position of the needle, as detailed in the following:

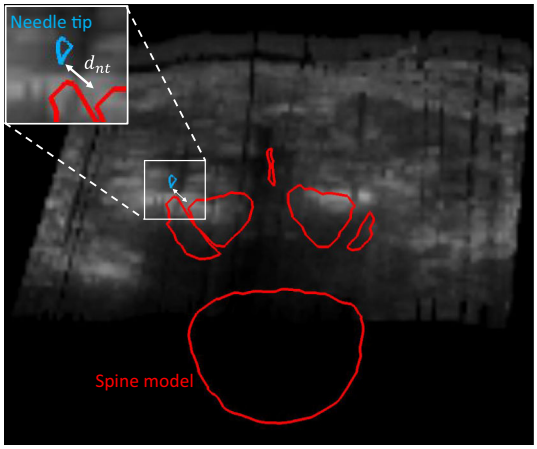


**Fig. 5** Schematic illustration of the accuracy measurement of the anatomical feature of the facet joint. The localization error of the facet joint was measured as the distance between the recorded tip of the needle and the position of the most superior point of the facet joint in the model $d_{\mathrm{nt}}$

1. Place patient in prone position (*conventional*)
2. Determine vertebrae levels using pre-procedure US scans and place reference sensor superior of L1 (*proposed system*)
3. Acquire 3D US volume of the lumbar spine (*proposed system*)
4. Find entry point and insertion angle using fluoroscopic scan (*conventional*)
5. Insert needle to facet joint and confirm with fluoroscopic (*conventional*)
6. Record final needle position (*proposed system*)
7. Inject anesthesia (*conventional*)

Volume acquisition and model registration are performed as described earlier in this paper. For recording the final position of the injection needle, an EM-tracked needle (SonixGPS 0.5 mm needle sensor, Ultrasonix, Richmond, BC, Candada) is inserted into the hollow injection needle. The EM sensor can thus be placed at the tip of the needle during insertion. To determine the feasibility of our proposed approach to correctly locate the target structure (facet joint) in the US volume, we measure the distance of the recorded needle tip position and of a reference landmark picked by an expert sonographer to the middle of the most superior point of the facet joint in the model (see Fig. 5).

## Results

Examples of the registration of the multi-vertebrae model to US volumes are shown in Fig. 6. Distance errors are given in Table 1. The results for the side sweep are significantly better than the other two scans ($p < 0.05$), which makes it the preferred scanning protocol. An RMS error of 2.3 mm

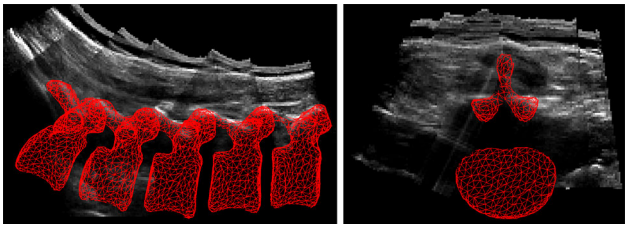

**Fig. 6** Example of the registered model to an US volume

**Table 1** RMS distance (in mm) between the segmented ultrasound and the registered patient-specific model averaged over the nine cases where a CT was available. Results for the side sweep are significantly better ($p < 0.05$)

| | Side sweep | Transverse midline | Sagittal zigzag |
|---|---|---|---|
| RMS | $2.3 \pm 0.4^*$ | $3.2 \pm 0.9$ | $3.0 \pm 0.5$ |
| Max | $8.9 \pm 4.2^*$ | $10.6 \pm 4.4$ | $13.4 \pm 4.9$ |

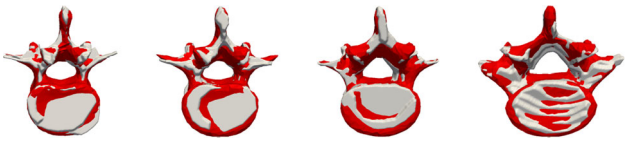

**Fig. 7** A comparison of registered model to US images and the model from CT images. The registered model is highlighted in red, and the white surface shows the CT manual segmentation

**Table 2** RMS distance (in mm) between the shape of the manual segmentation of vertebrae in the CT and the shape of the registered patient-specific model vertebrae in the ultrasound images averaged over all nine cases where a CT was available

| | Side sweep | Transverse midline | Sagittal zigzag |
|---|---|---|---|
| L2 | $1.3 \pm 0.3$ | $1.3 \pm 0.3$ | $1.3 \pm 0.3$ |
| L3 | $1.5 \pm 0.4$ | $1.4 \pm 0.2$ | $1.3 \pm 0.3$ |
| L4 | $1.5 \pm 0.4$ | $1.5 \pm 0.4$ | $1.5 \pm 0.4$ |
| L5 | $1.5 \pm 0.3$ | $1.5 \pm 0.3$ | $1.5 \pm 0.3$ |
| All | $1.5 \pm 0.4$ | $1.4 \pm 0.3$ | $1.4 \pm 0.3$ |

(maximum 8.4 mm) to the surface points extracted from the US volume was recorded for the model registration using this scanning technique.

The RMS distance errors between the manual segmentation of the CT and the registered model are given in Table 2. Interestingly, there are no significant differences between the US acquisition techniques. All vertebrae give errors of ~1.5 mm, suggesting that all registered models can accurately generate patient-specific vertebrae anatomical shapes. Figure 7 shows a comparison of the registered model to the reference CT surfaces.

Results on the capture range experiment show that the error remains essentially unchanged for variations under 10 mm. This covers a reasonable area (20 mm) within the vertebrae, suggesting the method is robust to initialization errors.

Figure 8 illustrates the results of our model-based guidance visualization for the five cases where a facet joint injection was performed. The registered model is overlaid onto the US volume and shown at the axial position of the target facet joint together with the recorded final needle position. Quantitative results of how accurate the model augmentation can localize the target facet joint are shown in Table 3. The average distance of the model-predicted facet joint position to the final tip position of the needle from which anesthesia was successfully performed was determined as $5.1 \pm 1.5$ mm.

The total run time for the guidance approach was recorded at around 2 min with volume reconstruction contributing for 50 s, manual model initialization for 60 s, and the model registration for 10 s. Note that needle insertion was not part of this study and is thus not included in the procedure time.

## Discussion and conclusion

In this study, we presented a guidance system for facet joint injections that allows augmentation of the live standard 2D US image with anatomical information from a statistical model of the lumbar spine and relating this information to the position of the tracked injection needle. An initial evaluation on in vivo data yielded an RMS registration accuracy to the bone structures in the US volume of 2.3 mm and an RMS surface distance error compared to a CT reference in the magnitude of 1.5 mm. Preliminary results of an ongoing study on patients actually treated by facet joint injections show that the facet joint can be identified using the overlaid model with an accuracy of around 5.1 mm, which is equivalent to the postulated accuracy considered to be sufficient for the anesthetic block to be effective [5].

There promise to be several potential benefits of the presented approach over conventional guidance techniques. The main drawbacks of the widely used fluoroscopic guidance can be seen in its radiation exposure to both patient and anesthesiologist, its limited portability, the limited ability of visualizing nerves and vessels, and the projective 2D nature of the images acquired during the guidance process [21]. The proposed guidance approach alleviates these issues by providing live US imaging during needle insertion that can be augmented with information generated from the registered 3D model and shown along with the accurate tracking data of the insertion needle. Although the introduction of tracking hardware alters the workflow and setup of the procedure generally used for US-guided spine injections, the additional information provided through the model improves the interpretability of the US images and promises to allow physicians inexperienced with US imaging to quickly learn to perform the procedure. With many US manufacturers already providing tracking modules with their standard equipment, workflow adaption should not be a major problem. We fur-

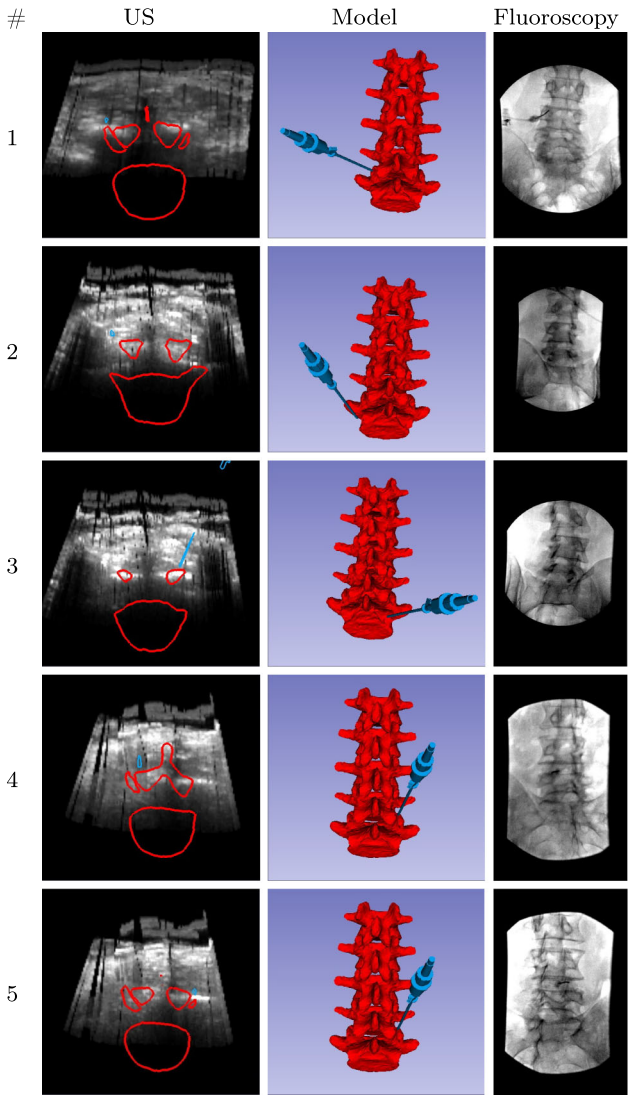


**Fig. 8** Illustration of the guidance results of the five facet joint injections, where US volumes were acquired before the injection, and the final needle position was recorded after the injection. The registered statistical shape + pose model (*red*) is shown relative to the pose of the needle (*blue*) which was inserted under conventional fluoroscopic guidance by the anesthesiologist. For comparison, the fluoroscopic control scan after the needle insertion is shown for every case

thermore believe that the procedure time which currently accounts for around 5 minutes for both fluoroscopy and US guidance [21] will not increase with the proposed method where augmented US images promise to result in a faster insertion process due to the simplified image interpretation.

Previously presented US-based techniques where preoperative tomographic data were related to the live US images reached registration accuracies better than 3 mm [3,20]. These accuracies are comparable to the registration accuracies recorded for our generic model-based approach. Compared to previously presented guidance systems for facet joint injections, where targeting accuracies as low as 0.6 mm (phantom study by Moore et al. [9]) and 1.0 mm (phantom study by Ungi et al. [19]) could be reached, the proposed system allows for an easy integration into the clinical workflow without major changes being necessary. It can be considered as the evolution of the methods presented by Moore et al. [9] and Ungi et al. [19] by providing the augmentation of the US with a statistical model. The initial alignment of the model to the US volume requires minimal interaction, i.e., selection of the L3 vertebra in the US volume, and is insensitive to small initial misalignments which is an improvement compared to these related methods, where initial registration is performed by selection of multiple fiducials in both CT and US. For even better workflow compliance, the tracked US transducer could be used for identifying the initial model position, thus reducing the amount of manual interaction with the volume data.

The current statistical model included the shape and pose variations of all lumbar vertebrae. In two of the five cases (cases 2 and 3) of our ongoing patient study, however, the targeted facet joint was the one between the L5 and the S1 vertebra. The missing model information in this area might then result in an inaccurate localization of the targeted anatomy. Furthermore, the sacrum as a connected bone complex leads to large bright echos in the US image which can eventually be used for automatic model initialization. The statistical model should thus be extended to include all parts of the spine that are relevant for the clinical procedure—in this case the sacrum. Furthermore, the training data used for model construction are segmented vertebrae from the CT images, which are all captured in a supine position. Therefore, the model may not be able to capture all possible spine curvature especially the curvature, when US and needle insertions are performed. Incorporation of training data acquired in the same prone position in which the intervention is performed may further increase registration accuracy.

Future work should further focus on prospective evaluation of the needle injection using this image-guided system and its potential improvement over the conventional manual or fluoroscopy-based technique.

**Table 3** Localization accuracy (in mm) of the facet joint as provided through the registration of the statistical shape + pose model with the US volumes for the five cases where a facet joint injection was performed. The accuracy is given as the distance from the tip of the conventionally inserted needle to the middle of the facet joint in the model as defined in Fig. 5

| Case | 1 | 2 | 3 | 4 | 5 | Average(±SD) |
|---|---|---|---|---|---|---|
| Distance to needle tip | 4.0 | 3.2 | 6.5 | 6.7 | 5.3 | 5.1 ± 1.5 |

The promising results of this initial feasibility assessment indicate that this US-based system is capable of providing accurate guidance information for facet joint injections. The prospect of replacing fluoroscopy as main guidance modality let this approach become extremely valuable.

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**Conflict of interest** A. Rasoulian, A. Seitel, J. Osborn, S. Sojoudi, S. Nouranian, V. A. Lessoway, R. N. Rohling, and P. Abolmaesumi declare that they have no conflict of interest.

**Ethical standard** All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki Declaration and its later amendments or comparable ethical standards. Informed consent was obtained from all individual participants included in the study.

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