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Towards real-time, tracker-less 3D ultrasound guidance for spine anaesthesia

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Abstract

Purpose. Epidural needle insertions and facet joint injections play an important role in spine anaesthesia. The main challenge of safe needle insertion is the deep location of the target, resulting in a narrow and small insertion channel close to sensitive anatomy. Recent approaches utilizing ultrasound (US) as a low-cost and widely available guiding modality are promising but have yet to become routinely used in clinical practice due to the difficulty in interpreting US images, their limited view of the internal anatomy of the spine, and/or inclusion of cost-intensive tracking hardware which impacts the clinical workflow.

Methods. We propose a novel guidance system for spine anaesthesia. An efficient implementation allows us to continuously align and overlay a statistical model of the lumbar spine on the live 3D US stream without making use of additional tracking hardware. The system is evaluated *in vivo* on 12 volunteers.

Results. The *in vivo* study showed that the anatomical features of the epidural space and the facet joints could be continuously located, at a volume rate of 0.5 Hz, within an accuracy of 3 mm and 7 mm respectively.

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Conclusions. A novel guidance system for spine anaesthesia has been presented which augments a live 3D US stream with detailed anatomical information of the spine. Results from an *in vivo* study indicate that the proposed system has potential for assisting the physician in quickly finding the target structure and planning a safe insertion trajectory in the spine.

Keywords Guidance · Epidural · Facet joint injection · 3D ultrasound

1 Introduction

Needle insertions in the lumbar spine play an important role in pain management and regional anaesthesia [2, 24]. Two common procedures are facet joint injections and epidural needle insertions (referred to as 'epidurals'). Both procedures require careful placement of the injection needle to ensure effective therapy delivery and to avoid damaging the nerves of the spinal cord [34].

Facet joint injections are commonly used to relieve pain in the lower back, a pain experienced by 80% of the adult population [28]. For these procedures, fluoroscopic guidance is the standard of care to ensure accurate needle placement. This placement is particularly challenging due to the narrow, small channel between the articular processes of the joint, the oblique entry angle, the deep location, and the proximity to nerve tissue. Although fluoroscopic guidance usually results in accurate needle placement and low complication rates [2], the X-ray exposure from these modalities poses a considerable risk and requires that the procedure is performed in a specially equipped clinic.

Epidurals are effective alternatives to general anaesthesia [16], and are often applied to provide anaesthesia for surgery, to ease the pain of labour and delivery, and to inject analgesics before, during, or after surgery. When applied in obstetrics (a procedure performed on over 60% of the pregnant women in the US [23]) fluoroscopy cannot be used for guidance and the insertion of the needle into the spine is based solely on the sense of touch. After choosing an appropriate puncture site the anaesthesiologist must push the needle until the tip enters the epidural space and feel for the associated loss of resistance to saline or air injection.

For both epidurals and facet joint injections, complications associated with inaccurate needle placement, such as accidental dural puncture, are relatively common (2.5%) and in 86% of the cases lead to post-puncture headache [1, 30]. For inexperienced operators the dural perforation rate is even higher (3-5%), indicating that the learning curve for these procedures is steep [7]. The difficulty of the injection increases with the presence of obesity, scoliosis or previous back surgery [34]. Accurate real-time guidance during the intervention, with the end goal to increase accuracy and efficacy of the procedure and to decrease the risk for the patient, is therefore desirable and would provide information of the puncture site, the needle trajectory and the depth of insertion.

Consequently, ultrasound (US) has gained in popularity as a non-ionizing, accessible, and inexpensive imaging modality [34]. Several studies show the feasibility of standard US systems as a real-time modality for guiding these injections [6, 17, 21, 29, 32, 34]. However, the majority of current interventions are not performed using US guidance as difficulty in interpreting the US images or the limited view on internal spinal anatomy cannot yet guarantee a safe and efficient procedure; therefore, US is currently not recommended as the sole imaging technique—particularly for obese patients and those suffering from spine abnormalities [34].

In recent approaches US was enhanced with pre-operative data and/or tracking information. Chen et al. [4] proposed a method in which a tracked US transducer was used to acquire an US volume of the spine. Pre-operative computed tomography (CT) data was related to the acquired US volume via volume-to-volume registration and the electromagnetically (EM) tracked needle was guided towards the target using the information from the CT data during the intervention. Moore et al. [19] proposed and evaluated three guidance approaches for facet joint injections based on intra-operative US and showed that the integration of a virtual CT-based model improved the accuracy of the injection. EM tracking was again used to relate CT and US information after a landmark-based registration was performed, and to provide the location of the needle accordingly. Ungi et al. [33] introduced a guidance system that used tracked US snapshots for guiding needle insertion. This technique allows the physician to visualize the target structure and manipulate the needle intuitively at the same time. Although these approaches achieved promising results regarding the accuracy of the spinal needle injection, they have not replaced conventional approaches and their alteration and complication of the clinical workflow, by introducing additional tracking hardware, still seem to outweigh the benefits regarding radiation exposure and real-time imaging.

Other methods enhancing the US volume with anatomical information, such as fitting a statistical shape model [14], automatic detection of vertebral levels [13], and automatic identification of needle insertion sites [35], have not been found applicable to guidance systems due to their computational complexity. A mechanical approach to the problem in which a needle guide (called EpiGuide) allows live imaging with a standard 3D US transducer of the needle path to the epidural target has recently been introduced [18]; however, image interpretation still remains an issue.

To overcome these limitations, we build upon our previous work on augmentation of US volumes with a statistical shape+pose model [27]. We propose a guidance system that continuously augments standard 3D US with anatomical information and allows for tracking the pose of the lumbar spine throughout the intervention without making use of any additional tracking technology. The core functionality is implemented using an open-source software architecture and the integrated system is evaluated *in vivo* on 12 volunteers in order to test runtime, tracking accuracy, and the anatomical validity of the augmentation.

2 Methods

The workflow of a conventional 2D US-guided spine injection is shown in Fig. 1. The basic steps include: (1) the determination of the vertebra level by counting vertebrae from the sacrum or the first rib either by palpation or by US scanning in a parasagittal plane, (2) the definition of the entry point, the insertion angle, and the insertion depth (after finding the transducer pose that best visualizes the target structure), (3) the actual insertion of the needle (either in a blind manner or by using an in-plane insertion approach), (4) the endpoint confirmation by experiencing a loss of resistance to saline or air injection when the needle tip reaches the target area (which may be supplemented by additional control

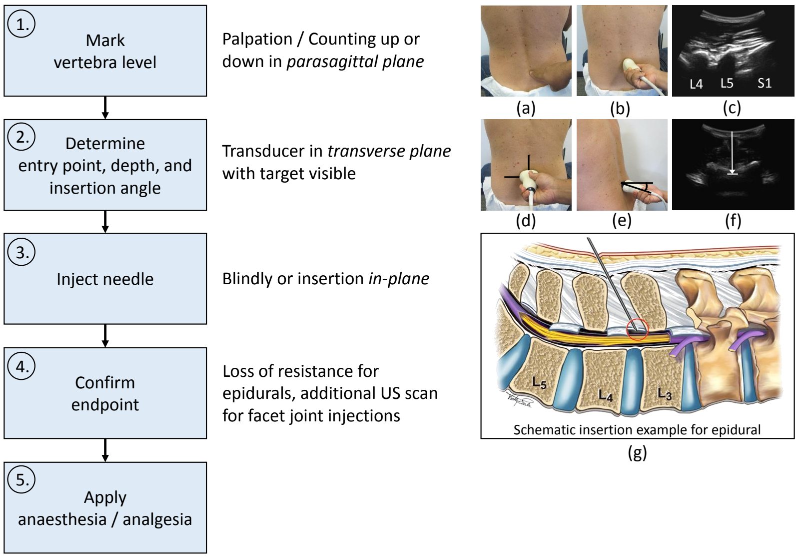


Fig. 1 The workflow of a conventional 2D US-guided spine injection. The level of the vertebra is marked by palpation of the spinous processes (a) or by counting the articulary processes in a parasagittal plane (b+c) from the sacrum or from the ribs. The transverse scanning plane (f) then allows the clinician to find a pose of the transducer that clearly shows the target and enables definition of the entry point (d) and insertion angle (e) and determination of the depth of insertion (f). Needle injection (g) is mostly performed in a blind manner, but in some cases an US guidance technique is used in which the needle is inserted in the US plane. Before the actual anaesthesia/analgesia is applied, loss of resistance to saline or air injection and an US scan is used for placement control.

scans) and (5) the application of the anaesthesia/analgesia. For a more detailed description of this procedure the reader may refer to [3, 18, 20].

Current drawbacks of this procedure, preventing a widespread use of US in clinical routine, are mainly due to the limited image quality and the 2D nature of the current US scanning technique. This method renders image interpretation and visualization of internal and deep structures difficult, which is especially important for safe needle injection and placement control for epidurals and facet joint injections. For improvement of this guidance technique we therefore propose a system that is capable of limiting these deficits by applying the following improvements:

3D Imaging: 3D US is used to allow for continuous US volume acquisition before and during needle insertion, not limited to a specific orientation of the US transducer.

Image Enhancement: The US volumes are enhanced with anatomical information initially not visible in the US (e.g. due to occlusion) by continuously registering a statistical model of the lumbar spine with the US volume.

Continuous Anatomy Localization: The pose of the US transducer with respect to the anatomy of the lumbar spine is determined continuously, without the need for additional tracking hardware, facilitating a continuous image enhancement.

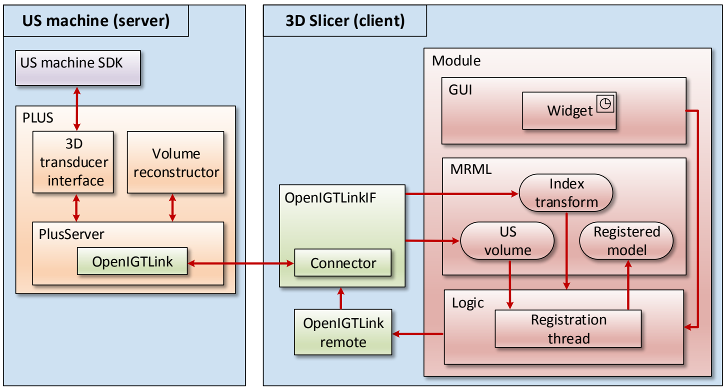


Fig. 2 System architecture of the US guidance system in which blocks represent processing elements and arrows show the direction of data flow. On the US machine, PLUS is responsible for data acquisition, US volume reconstruction, and communicates the reconstructed US volumes (via the OpenIGTLink protocol) to a client. The client application is implemented as a module in 3D Slicer in which the continuous model registration is performed. A timer within the GUI of the module observes the registered model (updated by the registration thread) and the US volume (updated by PLUS) to allow for a synchronized visualization of model and volume. The module uses the publish-subscribe pattern to observe changes and make updates.

The remainder of this section will describe how these improvements are implemented in a practical guidance system for spine anaesthesia (Sec. 2.1) and how the clinical validity of the proposed guidance technique was evaluated (Sec. 2.2).

2.1 3D US Guidance System

To allow for continuous tracking and augmentation of the lumbar spine in the 3D US we propose a system architecture which is tightly integrated into the US machine (see Fig. 2). This system uses state-of-the-art open-source software libraries which allows for a flexible and robust implementation. We use a client-server approach in order to run computationally costly algorithms in parallel. On the US machine, a server program is responsible for data acquisition and US volume reconstruction (Sec. 2.1.1). On the client machine, a 3D Slicer[1] [8] module performs tracking of the spine (Sec. 2.1.2) and provides guidance visualization (Sec. 2.1.3).

2.1.1 Data Acquisition and US Volume Reconstruction

Data acquisition was implemented as part of the open-source "Public software Library for UltraSound imaging research (PLUS)" [15]. PLUS allows for running the acquisition routine on the US machine (server) and sending the US volumes via Ethernet (using the OpenIGTLink protocol [31]) to a client computer running the guidance software (see Fig. 2). Prior to this work PLUS only supported access to US data acquired by 2D US transducers. Here we describe its extension for acquiring US volumes using a motorized 3D US transducer.

[1] 3D Slicer, `http://www.slicer.org`

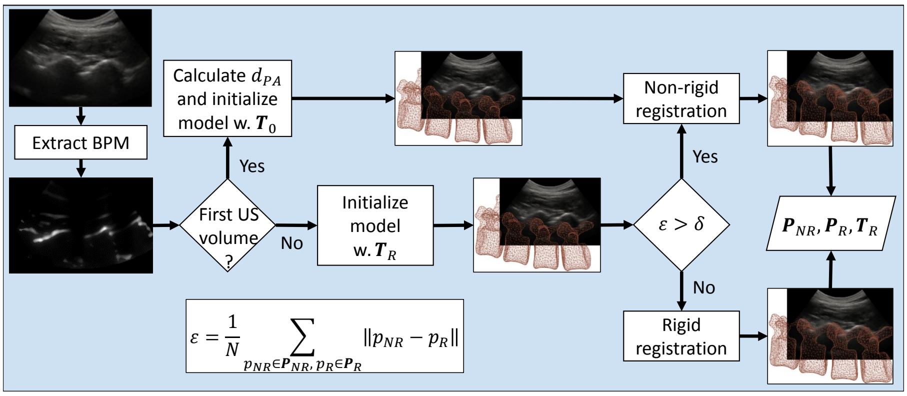


Fig. 3 Flowchart of the algorithm running within the registration thread of the 3D Slicer module. For each US volume received from the server the BPM is extracted. For the first US volume the model is initialized at a certain vertebra level (where the depth d_{PA} is automatically calculated). The non-rigid registration step then comprises the optimization of the shape and the pose coefficients of the model, as well as its rigid transformation. For subsequent US volumes a rigid registration is performed, unless the mean surface error ϵ is above a certain threshold δ, in which case the registration update will be non-rigid.

Access to the data from the US machine was implemented as a C++ class within PLUS. The main function of the class was to interface between the US machine SDK and PLUS by connecting to the motorized transducer, setting the acquisition parameters, calculating the pose and the position of the individual 2D frames with respect to the actual motor movement, and storing the individual frames (including pose and position) in PLUS's internal buffer. By defining a motor to image transform that takes into account both the pose and the position of the individual frames, the data stored in the internal buffer is reconstructed into US volumes using PLUS's volume reconstructor[2]. During this reconstruction a spatiotemporal integration smoothes the US volumes. To account for missing frames during the acquisition, a hole filling algorithm is applied to each US volume, interpolating missing voxels from a Gaussian-weighted average of the cubic neighborhood with a diameter of 3 voxels and a standard deviation of the Gaussian of 0.667. The reconstructed US volumes are then sent from the US machine to all listening clients using the software PlusServer (included in PLUS).

2.1.2 Continuous Anatomy Localization of the Lumbar Spine

The continuous localization of the lumbar spine in the 3D US is performed on the client machine within a 3D Slicer module (see Fig. 2). The localization is accomplished by continuously registering a statistical shape+pose model of the lumbar spine to the reconstructed US volumes received from the server. This localization repeatedly comprises the three steps of bone feature extraction, model initialization, and model registration (see Fig. 3).

[2] PLUS user manual: Volume reconstruction, `http://perk-software.cs.queensu.ca/plus/doc/nightly/user/ProcedureVolumeReconstruction.html`

Extraction of the Bone Features from the US Volumes. Foroughi et al. [9] proposed a technique for computing a so-called bone probability map (BPM) assigning each voxel in an US volume a probability value of being a bone surface voxel. They make use of the fact that bone highly reflects US waves and creates a distinct shadow behind the bone due to the large impedance mismatch between bone and soft tissue. This knowledge is transferred into an image containing a reflectivity number calculated from the Laplacian of Gaussian of the US image and a shadow map where each pixel value corresponds to a weighted sum of the pixel values below the actual pixel. The BPM is calculated as the product of the shadow map and the image containing the reflection numbers.

The runtime of the original implementation was reported as 0.5 seconds per US frame which is not sufficient for our continuous tracking application concerned with US volumes not US frames. However, the pixel-based nature of the calculations of this algorithm allows us to improve processing speed drastically by computing the resulting BPM in a parallelized manner. Our fast implementation uses the Intel Math Kernel Library[3] (MKL, Intel Corp., Santa Clara, CA, USA) for efficient memory access and effective computations of mathematical operations. OpenMP[4] allows us to perform the per-pixel operations in parallel to increase system performance even further.

Model Initialization. In order for the optimization of the statistical shape+pose model to converge to the correct target features, the model has to be initialized roughly at the correct position in the US volume. We use a semi-automatic approach in which the physician initially has to chose the correct vertebrae level of interest, approximately midsagittal. As the localization is started, a rigid transform $\boldsymbol{T}_0$ initializes the model. The depth d_{PA} (the translation in the posterior-anterior direction) of $\boldsymbol{T}_0$ is calculated using the information contained in the BPM

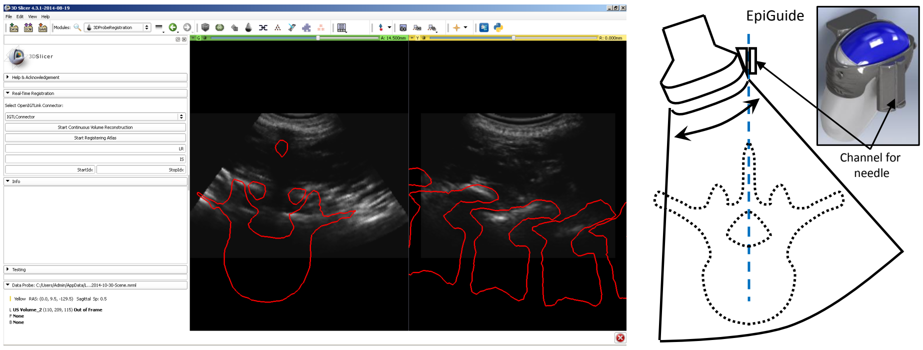
$$d_{PA} = \frac{\sum_{f=l}^{F-r} \left\lfloor \frac{\prod_{y=c}^{n_y} y \cdot [\sum_{x=0}^{n_x} p_f(x,y)]}{\sum_{y=c}^{n_y} \sum_{x=0}^{n_x} p_f(x,y)} \right\rfloor}{F - r - l}, \tag{1}$$

where $p_f(x, y)$ is the pixel value of frame f in the BPM consisting of F frames, n_x and n_y are the number of pixels in the x- and y-direction, c is a constant defined to disregard pixels close to the transducer, and r and l are thresholds defined in order to disregard outer frames of the BPM consisting of mainly zero intensity pixels. For subsequent US volumes the optimized rigid transform $\boldsymbol{T}_R$, calculated during the registration, is used in order to initialize the model.

Model Registration. The statistical lumbar spine model was constructed in an earlier study with training data collected from 32 patient volumes (note that this data was only used for model construction and is independent from the data set used for evaluation) [27]. Our continuous registration algorithm is an extension of the technique presented in the same study. Once US volumes are received from the server, registration is started and the model is initialized and non-rigidly registered to the BPM. This non-rigid registration results in the transform $\boldsymbol{T}_R$ as well as

[3] Intel Math Kernel Library, `https://software.intel.com/en-us/intel-mkl`

[4] The OpenMP API specification for parallel programming, `http://openmp.org/wp`



(a) (b)

Fig. 4 (a) Guidance interface implemented as a module in 3D Slicer. The continuous registration of the shape+pose model with the US volume is shown as an augmentation for both the transverse and the sagittal plane. (b) Schematic illustration of the guidance procedure for an epidural showing the overlaid spine model and the needle path when using a guidance tool such as the EpiGuide, where a needle can be easily attached and detached.

optimized shape and pose coefficients $\widetilde{\boldsymbol{w}}^s$ and $\widetilde{\boldsymbol{w}}^p$, best fitting the model to the US volume. For subsequent US volumes—unless the mean surface error ϵ between the previous model $\boldsymbol{P_R}$ and the last non-rigidly registered model $\boldsymbol{P_{NR}}$ is greater than a threshold δ—only the rigid transformation $\boldsymbol{T}_R$ is optimized

$$Q = \sum_{l=1}^{L} \sum_{m,n=1}^{M,N} P(\boldsymbol{x}_n^l|\boldsymbol{y}_m) \left\| \boldsymbol{y}_m - (\boldsymbol{R}\Phi(\boldsymbol{x}_n^l;\widetilde{\boldsymbol{w}}^s,\widetilde{\boldsymbol{w}}^p) + \boldsymbol{t}) \right\|^2, \tag{2}$$

where L is number of objects, M is the number of surface points in the target, N is number of surface points in the lth object, $P(\boldsymbol{x}_n^l|\boldsymbol{y}_m)$ is the correspondence function, $\Phi(.)$ the similarity transformation of shape and pose, and $\boldsymbol{R}$ and $\boldsymbol{t}$ are the rotation and translation components of $\boldsymbol{T}_R$. The threshold δ is necessary since an US volume may contain new anatomical information as the transducer moves. Finally, to allow for an efficient implementation, Intel MKL is again used to reduce computation time for mathematically complex operations.

2.1.3 Guidance Visualization

With the lumbar spine model continuously registered to the 3D US we can provide the user with anatomical estimates where US fails to generate image information. The epidural space and the facet joints are of particular interest for the case of spinal needle injections. The guidance interface is part of the 3D Slicer module running on the client machine which inherently allows for intuitive interaction with the US image data (scroll through different slice views, change level/window). In accordance with our clinical partners we determine a two window view showing a transverse view on the left side and a sagittal view on the right side, of the US volume, as these planes contain most of the important anatomical features.

A registration is labeled successful if the mean surface error of two subsequent models is below a certain threshold in which case the anatomical information of the model is superimposed onto these image planes for guidance (see Fig. 4(a)).

Once the target is visualized with the proposed software, the needle insertion can be performed. As a helpful tool for real-time imaging during insertion by a single operator, we propose the use of the recently presented EpiGuide [18]. The EpiGuide provides a calibrated channel for the needle and an easy release mechanism for the needle from the guide once the needle tip has reached the target (see Fig. 4(b)).

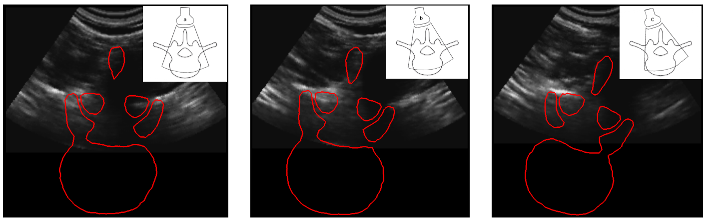
2.2 System Validation

The evaluation of the guidance approach was conducted *in vivo* on 12 young, healthy volunteers of average weight, from which informed written consent was obtained. US scanning of each subject was performed by an expert sonographer using a SonixTouch US machine (Ultrasonix, Medical Corp, Richmond, BC, Canada) as it is one of few systems which allow for running custom software directly on the US machine, as well as providing a research interface for controlling data acquisition. A convex 4D motorized curvilinear array transducer (m4DC7-3/40, Ultrasonix Medical Corp, Richmond, BC, Canada) was used. Such transducers are common in mid-tier ultrasound machines. The transducer was operating at 4.5 MHz, with a depth of 7.0 cm, a 45% gain and acquired US volumes at a volume rate of 0.5 Hz. Each US volume consisted of $221 \times 146 \times 197$ pixels and had a 0.5 mm spacing. An optical tracking system (Polaris Spectra®, Northern Digital Inc., Waterloo, Ontario, Canada) was used to acquire reference data of the transducer movement during data collection. The tracked US transducer was calibrated using the open-source calibration tool fCal[5] and a double N-wire phantom [5]. The calibration RMS error was recorded as 0.57 mm.

US volumes were acquired in prone and sitting position. For each subject the body of L3 was identified as the midpoint between the L2-3 and L3-4 interlaminar spaces and marked on the skin. The continuous registration was started and 15 US volumes were obtained as the transducer was moved from a perpendicular midline sagittal plane to a sagittal plane, with an approximately 15 degree medial angle, 15 mm lateral to the midline (see Fig. 5). The quantitative evaluation of the guidance system was conducted with respect to the following evaluation parameters:

Anatomical Feature Identification Accuracy: The correct identification of anatomical features for facet joint injections and epidural needle insertions was evaluated as shown in Fig. 6. The clinically acceptable tolerance is defined as 3 mm for epidurals (defined by the width of the ligamentum flavum covering the epidural space [22]), and 5 mm for facet joint injections (considered sufficient for anaesthetic block to be effective [10]). All annotations were made by an experienced sonographer. For each subject, one US volume was chosen (in both prone and sitting position). The reference point representing the epidural space in the US volume was determined by calculating the midpoint between two annotations placed on both sides of the ligamentum flavum (LF) of L3. The LF was identified as a short, bright, single or double linear echo immediately lateral to the base of the spinous process (SP). Two annotations were placed at both midpoints of the leading edge of the LF echo complex on each

[5] Freehand tracked ultrasound calibration application (fCal), `http://perk-software.cs.queensu.ca/plus/doc/nightly/user/ApplicationfCal.html`



(a) (b) (c)

Fig. 5 Schematic illustration of the data collection procedure. As US volumes were obtained, the continuous registration was started. The transducer was then moved from a perpendicular midline sagittal plane to a sagittal plane with an approximately 15 degree medial angle, 15 mm lateral to the midline, resulting in 15 US volumes with corresponding registrations. Fig. (a), (b), and (c) show slices augmented with the model at different transducer positions.

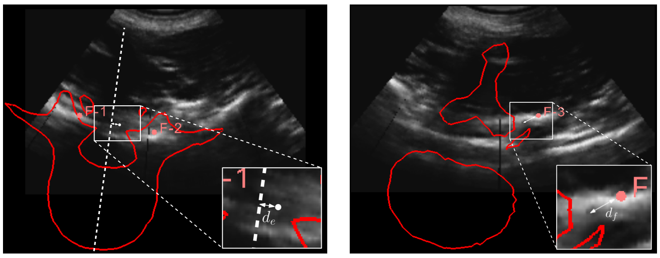
side of the SP. The midpoint of these two annotations was calculated and the Euclidean distance of this midpoint to the midsagittal plane of the registered model was used to quantify the correctness of the model overlay. The overlay was considered successful for visualizing the epidural space area when the reference point was not occluded by bony structures of the registered spine model. The reference position of the facet joints was identified by their characteristic 'thumb-print' echo pattern consisting of an acoustic shadow and a bright echo arising from the symmetrically curved surface located immediately lateral and slightly posterior to the laminae. Annotations were placed in the middle of the most superior points of the symmetrical surface echo/shadow complexes. The accuracy of correctly visualizing the facet region was calculated as the mean of the Euclidean distances from these annotated points, to the middle of the most superior point of the corresponding facet joint in the model.

Tracking Accuracy: For the 15 US volumes (for each subject in both prone and sitting position) and their corresponding registrations, the accuracy of continuously locating the position and orientation of the spine is determined by calculating the relative deviations between the first registered model and subsequent registrations, and comparing them to the measurements obtained by tracking the transducer. Of particular interest is the error of rotation around the inferior-superior axis, and the error in lateral movement, as they are the common transducer movements during a needle insertion in the spine.

Runtime: The runtime was measured for all parts of the workflow (volume reconstruction, BPM calculation, non-rigid registration, rigid registration) using the `vtkTimerLog`[6] provided by the open-source Visualization Toolkit[7] (VTK). The server computer (US machine) was operated with an Intel® Core™ i5-2400 CPU @ 3.10 GHz processor with 1.52 GB of RAM, Windows XP Service Pack 3, Visual Studio 2008, and Plus 2.1.2.3645. The client, machine running the guidance software, featured an Intel® Core™ i7-4470 CPU @ 3.40 GHz processor with 16.0 GB of RAM, Windows 7 Service Pack 1, Visual Studio 2008, 3D Slicer 4.3.1, OpenMP 2.0, and Intel® MKL 11.0.

[6] vtkTimerLog, `http://www.vtk.org/doc/nightly/html/classvtkTimerLog.html`

[7] The Visualization Toolkit (VTK), `http://www.vtk.org`



(a) Epidural space (b) Facet joint

Fig. 6 (a) The error for epidural localization d_e was calculated as the distance between the midpoint of the ligamentum flavum positions annotated by the sonographer and the midsagittal plane through the registered model. (b) The error for the facet joint localization d_f was measured as the mean distance between the landmarks annotated by the sonographer and their corresponding points of the facet joints in the registered model.

Table 1 Average runtimes of the steps of the 3D US guidance system based on the data collected from the 12 volunteers averaged over $n = 360$ samples (given as mean±std).

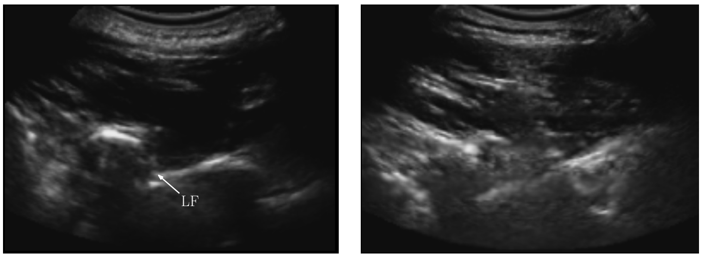
	Operation	Average time (s)
US machine	US volume reconstruction	2.10 ± 0.03
Guidance machine	Model registration	1.40 ± 0.20
	- Extract BPM	0.60 ± 0.04
	- Non-rigid registration ($n_{nr} = 60$)	1.31 ± 0.08
	- Rigid registration ($n_r = 300$)	0.24 ± 0.04

3 Results

The epidural space could successfully be identified with the registered model in 12 out of 12 cases for prone position, and in 10 of 12 cases for sitting position. In one of the unsuccessful sitting cases the registration failed, in the other, the sonographer could not locate the anatomical features of interest in the US volume (see Fig. 7). The accuracy of the epidural localization is summarized in Fig. 8(a) and averaged to 3.0 mm for the prone position and to 2.7 mm for the sitting position. The accuracy of the facet joint location, where a binary success metric was difficult to define, is shown in Fig. 8(b) and averaged to 7.1 mm for the prone position and to 6.9 mm for the sitting position. The model registration on the client machine could be performed in 1.40 ± 0.20 s on average. Fig. 9 shows the errors in continuously localizing the spine for rotations along the inferior-superior axis and translations in the left-right direction for prone and sitting positions. They averaged to $3.4 \pm 3.2°$ and 4.5 ± 3.2 mm over all subjects in both prone and sitting position. Runtime results are summarized in Table 1. Averaged over all US volumes, acquired and processed for the 12 subjects (n=360), the time for reconstructing the US volume on the US machine was 2.10 ± 0.03 s.

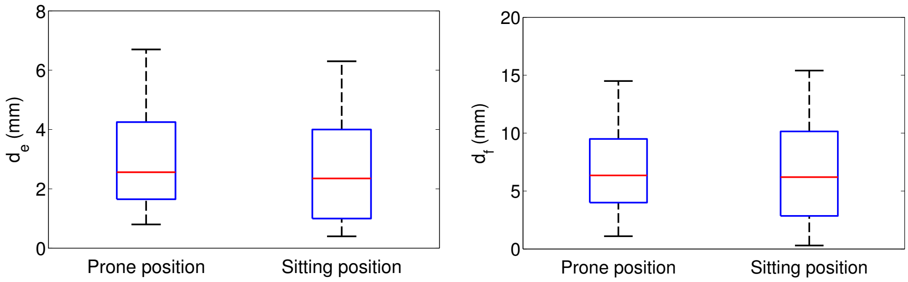
4 Discussion

The presented approach to guide epidural needle insertions and facet joint injections is to the authors knowledge the first of its kind, able to continuously and



(a) Subject 10 sitting (b) Subject 12 sitting

Fig. 7 Comparisson of two US volumes from different subjects. For Subject 10 the image quality was sufficient to allow the sonographer to confidently locate the anatomical features of interest, e.g. the ligamentum flavum (a). For Subject 12 the US volume quality was not sufficient to allow the sonographer to confidently find the anatomical features of interest (b).

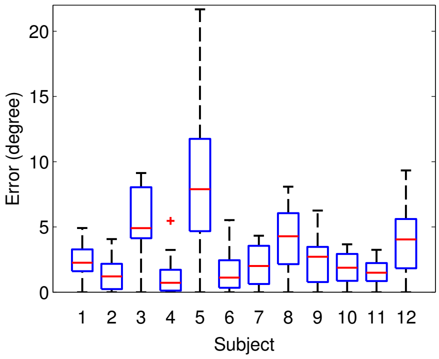


(a) Epidural space (b) Facet joints

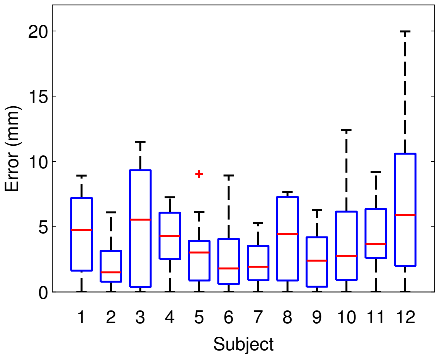
Fig. 8 Accuracy of the anatomical location of the epidural space (a) and the facet joints (b), for prone and sitting position averaged over all subjects. The box plots feature the median, upper (q3)/lower (q1) quartile, outliers ($> q3 + 1.5(q3 - q1)$ or $< q1 - 1.5(q3 - q1)$), and the sample minimum/maximum excluding outliers.

automatically localize features of the spine in 3D US without making use of any additional external tracking hardware. An *in vivo* study on 12 volunteers showed a localization accuracy for the epidural space that is within the clinical tolerance and an accuracy for the facet joints that is close to the clinically acceptable margin. For epidurals, where we could define a binary success metric (such that an insertion trajectory to the predicted target area could be found that did not traverse any bony structures), a success rate of 92% was achieved with an average error of 3 mm. For the facet joints, where a binary success metric was difficult to define, we could predict the location of the joint with an average accuracy of 7 mm. Tracking errors around 3.5° in rotation and 4.5 mm in translation of the midsagittal plane throughout the continuous registration indicate the ability of our approach to track the lumbar spine in 3D US. The time needed to calculate this model based image augmentation was recorded at 2.10 ± 0.03 seconds per US volume. These results will now be discussed in more depth.

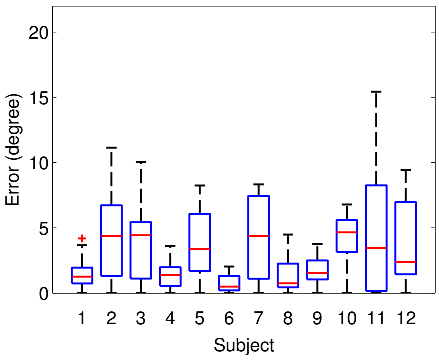
Anatomical Feature Identification Accuracy. For epidurals, an average error of 3 mm is within the clinically acceptable tolerance. For facet joint injections, an accuracy of 7 mm is close to the postulated 5 mm. One reason that the error in locating the facet joints was higher than in the localization of the epidural space



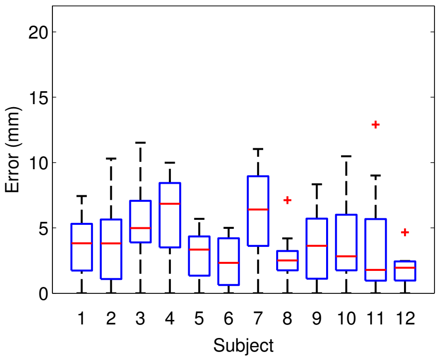
(a) Prone IS rotation



(b) Prone LR translation



(c) Sitting IS rotation



(d) Sitting LR translation

Fig. 9 Tracking accuracies for the transducer movement and rotation for each subject: (a) rotation around the inferior-superior (IS) axis in prone position, (b) translation along the left-right (LR) axis in prone position, (c) rotation around the IS axis in sitting position and (d) translation along the LR axis in sitting position. The box plots feature the median, upper (q3)/lower (q1) quartile, outliers ($> q3 + 1.5(q3 - q1)$ or $< q1 - 1.5(q3 - q1)$), and the sample minimum/maximum excluding outliers.

may be that the epidural space is located at a rather central position in the model and thus less subject to rotational alignment errors than the facet joints. Another reason could be that we defined the error for the facet joints using only one point of the facet region (*cf.* a plane for the epidural space). This error metric has the benefit of being uncomplicated, but the difficult interpretation of the US volume (especially in this region) renders an accurate and precise determination of the annotated reference landmark challenging and may therefore affect the result of our accuracy measurement. Furthermore, the position of the facet joints at the edge of the US volume may affect the clarity of the bone features needed for an accurate registration; therefore, applying a bone detection approach that is able to distinguish bone surfaces from image artifacts more clearly should increase the accuracy of the registration. As an example, local phase-based US image enhancement approaches promise to improve bone detection in US images [11, 26].

Increasing the statistical variability of the lumbar shape+pose model by adding more distinct training samples is likely to improve the registration accuracy even further. As mentioned earlier, patient obesity increases the difficulty of needle insertion. As our *in vivo* study was carried out on subjects of average weight, further studies should assess the registration accuracy in obese cases. All future studies should further include data annotated by more than one sonographer in order to avoid intra-observer errors.

Tracking Accuracy. The errors associated with the continuous model-based registration technique were found to be within those reported for a single registration to an US volume in our earlier study [27]. For Subject 2, where the reference epidural position was occluded by bone structures of the registered model in a sitting position, high error in the rotation accuracy of the model registration could be seen. This high error may be due to initial misalignment of the model, wrong bone features detected by the BPM algorithm, or insufficient statistical power of the underlying model training. For Subject 3, the laminae left of the SP are partially occluded which may be due to the anatomy of the subject or bad contact between the transducer and the skin. This occlusion leads to an increased error in translation and rotation. Subject 5 shows a high error in rotation around the inferior-superior axis in prone position. This high error may be attributed to a large initial misplacement of the transducer which was in this case not properly centered above the L3 SP. The resulting inaccurate initialization of the statistical model and the missing parts of the vertebrae in the US volume led to an unsuccessful registration. For Subject 12 (see Fig. 7(b)) the US volume quality was very low. This low quality is indicated by higher errors in the model registration.

Since the registration algorithm is sensitive to initial misalignments, automatic US plane detection, such as recently proposed by Pesteie et al. [25], could help during the initialization process, to confirm the correct placement of the transducer over a defined spine structure. This method is able to determine a 2D US plane in an US sweep containing special spine structures such as laminae, spinous processes, and facet joints or transverse processes, with high accuracy. Further, to ensure that physicians have confidence in the guidance system, the augmented US image should be associated with an uncertainty measure, including the confidence of feature detection in the US volume [12], and the quality of the model-based registration.

Runtime. The guidance system's volume rate was measured as 0.5 Hz. This volume rate, although not real time in the true sense of the word, is sufficient for smooth US enhancement and assisting the physician in finding anatomical information in the 3D US. The computational bottleneck of our parallelized approach is the US volume reconstruction from the motorized transducer. The sweeping motor results in a definite limit on the speed of US volume reconstruction and is also a cause of motion artifacts. Modern matrix arrays, which are able to acquire volumetric US data in real time, promise to increase processing speed. Their improved images, which are less susceptible to motion artifacts or reconstruction errors, should also improve registration quality.

To investigate the influence of the US volume reconstruction on the registration accuracy, future studies should be conducted. Furthermore, our registration algorithm was optimized to balance speed and registration accuracy. Therefore, we predict a higher accuracy running the guidance system with a lower volume rate, and vice versa.

In conclusion, the presented guidance system is capable of quickly assisting in finding the target structure, as well as the insertion corridor, during needle insertions in the lumbar spine. Despite its technically sophisticated nature, including an automated approach of analyzing the acquired US data, this system is still meant to be an aid, and not a replacement, for image interpretation by the physician. As the system was designed to complement and further facilitate the efficiency of the clinical workflow during spine anaesthesia, and avoid introduction of additional external tracking hardware, it has the potential to be easily integrated into a commercial US machine and standard clinical practice.

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References

1. Abdi, S., Datta, S., Lucas, L.F.: Role of epidural steroids in the management of chronic spinal pain: a systematic review of effectiveness and complications. Pain Physician **8**(1), 127–143 (2005)
2. Boswell, M.V., Colson, J.D., Sehgal, N., Dunbar, E.E., Epter, R.: A systematic review of therapeutic facet joint interventions in chronic spinal pain. Pain Physician **10**(1), 229–253 (2007)
3. Carvalho, J.C.A.: Ultrasound-facilitated epidurals and spinals in obstetrics. Anesthesiol Clin **26**(1), 145–58 (2008)
4. Chen, E.C.S., Mousavi, P., Gill, S., Fichtinger, G., Abolmaesumi, P.: Ultrasound guided spine needle insertion. In: Proc. SPIE, vol. 7625, pp. 762,538–1–762,538–8 (2010)
5. Chen, T.K., Thurston, A.D., Ellis, R.E., Abolmaesumi, P.: A real-time freehand ultrasound calibration system with automatic accuracy feedback and control. Ultrasound Med Biol **35**(1), 79–93 (2009)
6. Conroy, P.H., Luyet, C., McCartney, C.J., McHardy, P.G.: Real-time ultrasound-guided spinal anaesthesia: a prospective observational study of a new approach. Anesthesiol Res Pract **2013**, 525,818–525,824 (2013)
7. de Oliveira Filho, G.R.: The construction of learning curves for basic skills in anesthetic procedures: an application for the cumulative sum method. Anesth Analg **95**(2), 411–416 (2002)
8. Fedorov, A., Beichel, R., Kalpathy-Cramer, J., Finet, J., Fillion-Robin, J.C., Pujol, S., Bauer, C., Jennings, D., Fennessy, F., Sonka, M., Buatti, J., Aylward, S., Miller, J., Pieper, S., Kikinis, R.: 3D Slicer as an Image Computing Platform for the Quantitative Imaging Network. Magn Reson Imaging **30**(9), 1323–1341 (2012)
9. Foroughi, P., Boctor, E., Swartz, M., Taylor, R., Fichtinger, G.: Ultrasound bone segmentation using dynamic programming. IEEE Ultrasonics Symposium **13**(4), 2523–2526 (2007)
10. Greher, M., Scharbert, G., Kamolz, L.P., Beck, H., Gustorff, B., Kirchmair, L., Kapral, S.: Ultrasound-guided lumbar facet nerve block: a sonoanatomic study of a new methodologic approach. Anesthesiology **100**(5), 1242–1248 (2004)
11. Hacihaliloglu, I., Guy, P., Hodgson, A.J., Abugharbieh, R.: Volume-specific parameter optimization of 3D local phase features for improved extraction of bone surfaces in ultrasound. International Journal of Medical Robotics and Computer Assisted Surgery (2014). URL `http://dx.doi.org/10.1002/rcs.1552`. Published online
12. Karamalis, A., Wein, W., Klein, T., Navab, N.: Ultrasound confidence maps using random walks. Med Image Anal **16**(6), 1101–1112 (2012)
13. Kerby, B., Rohling, R., Nair, V., Abolmaesumi, P.: Automatic identification of lumbar level with ultrasound. Conf Proc IEEE Eng Med Biol Soc **2008**, 2980–2983 (2008)
14. Khallaghi, S., Mousavi, P., Gong, R.H., Gill, S., Boisvert, J., Fichtinger, G., Pichora, D., Borschneck, D., Abolmaesumi, P.: Registration of a statistical shape model of the lumbar spine to 3D ultrasound images. In: MICCAI 2010, Part II, vol. LNCS 6362, pp. 68–75 (2010)

15. Lasso, A., Heffter, T., Rankin, A., Pinter, C., Ungi, T., Fichtinger, G.: PLUS: open-source toolkit for ultrasound-guided intervention systems. IEEE Trans Biomed Eng **61**(10), 2527–2537 (2014)
16. Liu, S.S., Strodtbeck, W.M., Richman, J.M., Wu, C.L.: A comparison of regional versus general anesthesia for ambulatory anesthesia: a meta-analysis of randomized controlled trials. Anesth Analg **101**(6), 1634–1642 (2005)
17. Loizides, A., Peer, S., Plaikner, M., Spiss, V., Galiano, K., Obernauer, J., Gruber, H.: Ultrasound-guided injections in the lumbar spine. Med Ultrason **13**(1), 54–58 (2011)
18. Malenfant, P.A., Gunka., V., Beigi, P., Rasoulian, A., Rohling, R., Dube, A.: Accuracy of 3d ultrasound for identification of epidural needle skin insertion point in parturients; a prospective observational study. In: Society for Obstetric Anesthesia and Perinatology (SOAP) 46th Annual Meeting, p. 308. Toronto, ON, Canada (2014)
19. Moore, J., Clarke, C., Bainbridge, D., Wedlake, C., Wiles, A., Pace, D., Peters, T.: Image guidance for spinal facet injections using tracked ultrasound. In: MICCAI 2009, Part I, vol. LNCS 5761, pp. 516–523 (2009)
20. Narouze, S., Peng, P.W.H.: Ultrasound-guided interventional procedures in pain medicine: a review of anatomy, sonoanatomy, and procedures. Part II: axial structures. Reg Anesth Pain Med **35**(4), 386–396 (2010)
21. Niazi, A.U., Chin, K.J., Jin, R., Chan, V.W.: Real-time ultrasound-guided spinal anesthesia using the sonixgps ultrasound guidance system: a feasibility study. Acta Anaesthesiol Scand **58**(7), 875–881 (2014)
22. Nickalls, R.W., Kokri, M.S.: The width of the posterior epidural space in obstetric patients. Anaesthesia **41**(4), 432–433 (1986)
23. Osterman, M.J., Martin, J.A.: Epidural and spinal anesthesia use during labor: 27-state reporting area, 2008. National vital statistics reports: from the Centers for Disease Control and Prevention, National Center for Health Statistics, National Vital Statistics System **59**(5), 1–13 (2011)
24. Parr, A.T., Diwan, S., Abdi, S.: Lumbar interlaminar epidural injections in managing chronic low back and lower extremity pain: a systematic review. Pain Physician **12**(1), 163–188 (2009)
25. Pesteie, M., Abolmaesumi, P., Ashab, H.A.D., Lessoway, V.A., Massey, S., Gunka, V., Rohling, R.: Automatic Recognition of the Target Plane in 3D Ultrasound with EpiGuide. In: 7th NCIGT and NIH Image Guided Therapy Workshop. Boston, MA, USA (2014)
26. Rahmatullah, B., Papageorghiou, A.T., Noble, J.A.: Integration of local and global features for anatomical object detection in ultrasound. In: MICCAI 2012, Part III, vol. LNCS 7512, pp. 402–409 (2012)
27. Rasoulian, A., Rohling, R., Abolmaesumi, P.: Augmentation of Paramedian 3D Ultrasound Images of the Spine. In: IPCAI 2013, vol. LNCS 7915, pp. 51–60 (2013)
28. Rubin, D.I.: Epidemiology and risk factors for spine pain. Neurol Clin **25**(2), 353–371 (2007)
29. Shaikh, F., Brzezinski, J., Alexander, S., Arzola, C., Carvalho, J.C.A., Beyene, J., Sung, L.: Ultrasound imaging for lumbar punctures and epidural catheterisations: systematic review and meta-analysis. The BMJ **346**, 1720–1731 (2013)
30. Sprigge, J.S., Harper, S.J.: Accidental dural puncture and post dural puncture headache in obstetric anaesthesia: presentation and management: a 23-year survey in a district general hospital. Anaesthesia **63**(1), 36–43 (2008)
31. Tokuda, J., Fischer, G.S., Papademetris, X., Yaniv, Z., Ibanez, L., Cheng, P., Liu, H., Blevins, J., Arata, J., Golby, A.J., Kapur, T., Pieper, S., Burdette, E.C., Fichtinger, G., Tempany, C.M., Hata, N.: OpenIGTLink: an open network protocol for image-guided therapy environment. Med Robot **5**(4), 423–434 (2009)
32. Tran, D., Kamani, A.A., Al-Attas, E., Lessoway, V.A., Massey, S., Rohling, R.N.: Single-operator real-time ultrasound-guidance to aim and insert a lumbar epidural needle. Can J of Anaesth **57**(4), 313–321 (2010)
33. Ungi, T., Abolmaesumi, P., Jalal, R., Welch, M., Ayukawa, I., Nagpal, S., Lasso, A., Jaeger, M., Borschneck, D.P., Fichtinger, G., et al.: Spinal needle navigation by tracked ultrasound snapshots. IEEE Trans Biomed Eng **59**(10), 2766–2772 (2012)
34. Yoon, S., OBrien, S., Tran, M.: Ultrasound guided spine injections: Advancement over fluoroscopic guidance? Current Physical Medicine and Rehabilitation Reports **1**(2), 104–113 (2013)
35. Yu, S., Tan, K.K., Sng, B.L., Li, S., Sia, A.T.H.: Automatic identification of needle insertion site in epidural anesthesia with a cascading classifier. Ultrasound Med Biol **40**(9), 1980–1990 (2014)