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RESEARCH ARTICLE

Feature-Based Registration Framework for Pedicle Screw Trajectory Registration Between Multimodal Images

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ABSTRACT Pedicle screw placement for vertebral fixation is a complicated surgery for orthopaedic surgeons. The main challenge is to estimate the accurate trajectory's position to minimize post-operative complications related to pedicle screw placement. Different types of 3D to 2D registration techniques have been employed to avoid the misplacement of the screw during the surgery. However, these techniques cannot be applied directly to MR to X-ray registration due to differences in image intensity and tissue non-correspondence. To overcome these limitations, feature-based 3D to 2D registration technique was developed to map a trajectory position in the intra-operative X-ray image on to the pre-operative MR image. The registration framework validated by generating projection images that perfectly matched simulated X-ray images, then back-projecting the trajectory position on the pre-operative MR image using the estimated transformation parameters. The accuracy of the registered trajectory evaluated by measuring the displacement and directional errors between the registered and planned trajectory. The proposed method successfully registered the trajectory position in the simulated X-ray to pre-operative MR to estimate the trajectory position. A number of experiments are performed on the simulated dataset to assess the effectiveness of the proposed method. The Euclidean distance between the entry and end points and the directional error of the registered trajectory from the planned trajectory were below 1mm in AP, Lateral, and a combination of both planes. The mean trajectory length difference between the planned and registered trajectory was less than 1mm.

INDEX TERMS MR projection, optimization, pedicle screw insertion, similarity metric, trajectory planning.

I. INTRODUCTION

Pathological disorders of the spine like scoliosis, kyphosis, spondylolisthesis, spondylosis, vertebral fracture, intervertebral disc herniation, and spinal tumors cause spinal instability. This can be treated by vertebral fixation involving pedicle screw insertion to minimize vertebral motion, hence preventing the stretching of nerves and surrounding muscles. The insertion of the pedicle screw is challenging due to the limited view of anatomical features, which may cause damage to the patient's spinal cord, nerve roots, or vascular system [2]. So the surgeon has to properly plan Pedicle Screw Insertion (PSI) surgery by considering the pedicle morphological features and choosing the suitable pedicle screw size and insertion path for the patient to avoid the chance of screw misplacement [1].

Free hand-based PSI surgery prefers experienced surgeons to inexperienced ones to have safe surgery [3]. Intra-operative Computed Tomography (CT) 3D imaging has been developed to give a broad idea of the placement of screws during

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surgery [4]. The 3D visualization may lead to additional cost, time, and patient radiation exposure. In the absence of the 3D fluoroscopy imaging system, various Computer Assisted Surgeries (CAS) have been developed specifically to facilitate Pedicle Screw Surgery (PSS) [5]. However, existing CAS still has a variety of limitations such as inconspicuous Field Of View (FOV), the need for a direct line of sight in the presence of optical-based navigational trackers, the cost-ineffectiveness of CAS systems, interference, and lack of robustness problems in magnetic-based tracking systems.

Furthermore, intra-operative radiographs are used to visualize the position of the screw during the surgery. Still, it may lead to the misplacement of the screw due to the limited view provided by the imaging system. Additionally, to avoid misplacement of the screw, multiple radiographic images are taken to navigate the pedicle screw position from the posterior entrance site to the anterior-most targeted point, which results in too much radiation exposure to the patient. Hence to reduce misplacement of pedicle screws and radiation exposure, the 3D to 2D registration method has been developed [7]. It will help to visualize the screw position and orientation in a 3D view with a limited number of intra-operative radiograph images without disturbing the routine clinical settings.

The 3D to 2D registration technique is classified into feature-based and intensity-based registration methods [6]. Several authors developed intensity-based 3D to 2D registration algorithms to register the pre-operative 3D CT with intra-operative 2D X-ray for the 3D visualization of the target region during the surgery [8], [10] [11], [23] [24]. Uneri and team developed a known component 3D to 2D registration algorithm to navigate the surgical tools in the PSS [14]. The position of the surgical tools are identified by registering the known 3D component model present in the multiple radiograph image to the pre-operative CT image. The registration framework visually represents the registered surgical device position in relation to trajectory planning and the pedicle acceptance window. Hence framework provides potentially useful surgical product quality assurance and the accurate detection of pedicle screw breaches. The experiments was conducted on the cadaver and phantom specimens; therefore, it lacks the study on patient data. In a study conducted on spine phantom, the tip displacement and angular deviation were determined to be less than 2 mm and 0.5° respectively.The same work continued by the author to overlay scanned and parameterized models of the probing tool and pedicle screw onto the pre-operative CT through known component-based registration [12]. The algorithm shows a Target Registration Error (TRE) $<$ 2mm in surgical guidance as a function of the projection view having an angular separation of 10° between the reference image [13]. The developed technique offered better TRE than electromagnetic tracker-based localization, and it did so with 95% reliability for angular separations of at least 10°. The near real-time pedicle screw navigation technique was designed by the author to examine the accuracy of screw position. The intra-operative radiographs and the screw models information, such as the shaft's length, diameter, tip, and cap dimensions, are registered with the pre-operative CT to estimate the 3D screw pose. Additionally, experiments were conducted on the simultaneous registration of the multiple pedicle screws. The joint pose estimation of the patient and the pedicle screw increased the convergence time, which led to an increase in calculation time. The study conducted on the spine phantom with simultaneous registrations of 10 screws in the calibrated environment results in a mean Target Registration Error (mTRE) of 1.1±0.1mm and 0.7° ± 0.4° at the screw tip and angular deviation respectively. These metrics were found to be 2.7±2.6 mm and 1.5° ± 0.8° on the clinical dataset.

Esfandiari et al., [21] developed a deep learning-based framework for automatic segmentation and estimation of screw pose. The radiographs are segmented into three regions i.e., screw head, shaft, and background. Then the segmented shaft region is transferred to pre-operative images. The Fully Convolutional Network(FCN) is trained by projecting 3D components of various screw types which are generated by varying rigid transformations. To experimentally validate fiducial marker-based registration, two networks were built, one for the human synthetic X-ray image dataset and the other for the porcine specimen's clinical X-ray image dataset. The registration accuracy of the synthetic experiment was higher than that of the experiment using actual X-ray images. According to claims, the fully automatic framework is strong enough for faster intra-operative applications and can overcome frequent problems with intensity-based registration, such as local optimums and a small capture area. In the case of synthetic and real X-ray images, the deep learning-based network accurately segmented the screw shaft with 93% and 83% accuracy, respectively. On the realistic clinical X-ray images, the TRE measurements were found to be 1.93° ± 0.64° and 1.92±0.55mm respectively.

Newell et al., [20] developed a tool for the 3D visualization of the pedicle screw by registering pre-operative CT with intra-operative images. The two intra-operative images in Anterior Posterior (AP) and lateral directions with approximately 20° tilt are considered as reference images. The framework provided a superior rate of breaching scenario detection and identification when compared to surgeons' examination of 2D X-ray images. Under a variety of experimental circumstances involving screw lengths, materials, and vertebral levels, the method was shown to be reliable. When intra-operative X-ray image distortion was taken into account, the registration inaccuracy was not considerably altered. The angle error was discovered to be 1.3° and the Root Mean Squared Error(RMSE) for head and tail end displacements were measured to be 0.7mm and 0.8mm, respectively.

Naik et al., [22] designed an anatomical landmark-based 3D to 2D Iterative Control Point (ICP) registration to register pre-operative CT with intra-operative X-ray images to estimate the pedicular marker pose. The registration framework was validated by optimizing the projection image using the corresponding control point registration. In the AP and lateral planes, it was determined that the mean Euclidean distances between the Head and Tail end of the reprojected trajectory and the real trajectory were, 0.6mm-0.8mm respectively, and 0.5mm-1.6mm. Similar to this, it was discovered that the equivalent mean angular errors were 4.9° and 20°. It was determined that there was a 2.67mm average trajectory length mismatch between the actual and registered trajectory. The limitation of this work is that it requires manual identification of anatomical landmarks, which requires additional time to start the registration process. Additionally, it was impossible to estimate vertebral pose using uniform landmark selection across multiple levels because of the vertebral distortion.

In diagnosing spinal disorders such as degenerative diseases, deformities, infections, congenital disorders, injuries, tumors, and in surgical planning, Magnetic Resonance (MR) imaging is gaining importance due to its soft tissue characteristics. Hence registering pre-operative MR with the intra-operative X-ray will help us to identify the trajectory pose during the surgery. The approach employed in CT to X-ray registration cannot be adopted directly to the MR to X-ray registration due to variations in image intensity mismatch and tissue non-correspondence. Many techniques have been developed to overcome the mismatch between the image intensity and tissue non-correspondence. De Silva et al., [9] introduced a segmentation technique in the pre-processing stage to prevent the discrepancy between the image intensity. Then the segmented vertebral region in the sagittal MR image is registered with intra-operative radiographs to localize the target vertebral region in intra-operative radiographs. This technique for the vertebral level localization suits well when the images are in the sagittal MR and lateral radiograph protocol. As a result, it's possible that the techniques described in this study won't be directly extendable to images acquired using any scan or positioning protocols.

The proposed framework uses a feature-based multimodal 3D to 2D registration framework to register pre-operative MR with intra-operative X-ray to estimate pedicle screw trajectory pose in the vertebral body during the surgery. The contribution of the proposed work includes identifying the feature similarities between the multimodal images. The identification of the feature includes the segmentation of the particular feature, such as pedicle region, spinous process, vertebral body etc., So to register all the parameters accurately to the change in transformation parameters, the proposed framework identified the vertebral end plate as a better feature for registration. The segmented image is forward projected to generate the projected image and compared with the intra-operative images. In the absence of an intra-operative image, the simulated test images are considered as the ground truth image. These simulated images are generated from the MR image with the known transformation parameters as depicted in the Fig.1. The optimization of the Projected image continues until it finds the best match with the ground truth simulated image to estimate the vertebral pose. Then the trajectory position in the simulated 2D image is back-projected to the trajectory planned pre-operative MR for evaluating the registration framework. The framework performs trajectory mapping between the ground truth simulated 2D dataset and the pre-operative MR in clinical acceptance precision by limiting its study to the simulated radiograph without a real intra-operative radiograph image. In proposed work trajectory registration is preferred in different planes namely, AP, lateral, and combination of AP and lateral planes.

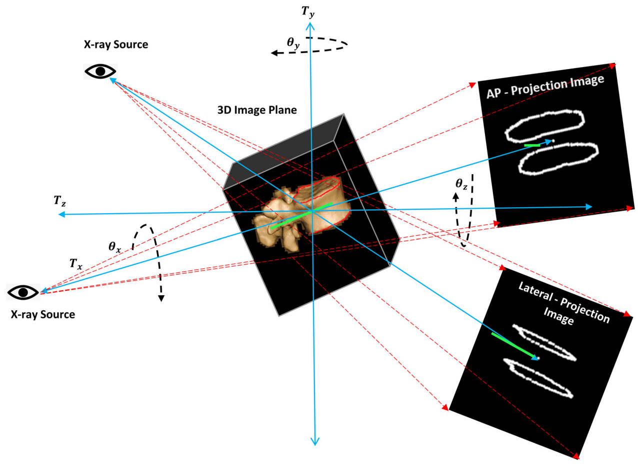

FIGURE 1. The segmented MR image is exposed to virtual c-arm to generate simulated AP and Lateral image with trajectory.

II. METHODS

The multimodal 3D to 2D registration framework as shown in Fig.2 is divided into two stages: vertebral pose estimation followed by trajectory registration to estimate the position and orientation of the pedicle trajectory during the surgery. In the absence of the intra-operative images, the registration accuracy is evaluated by generating the simulated X-ray images. Stage I describes the multimodal 3D to 2D registration framework to estimate the vertebral pose in the intra-operative images. Stage II describes the mapping of trajectory position in the pre-operative 3D images using the estimated transformation parameters obtained in stage I. The mapped trajectory position in 3D images are evaluated by measuring the distance between the projected and ground truth images trajectory position.

A. STAGE I: MULTIMODAL 3D TO 2D REGISTRATION

The Stage I framework includes feature extraction and trajectory planning in MR, MR projection, similarity metric, and optimization.

1) FEATURE EXTRACTION AND TRAJECTORY PLANNING IN MR

The pre-operative MR is acquired at different resolutions depending on the hospital setting. The collected dataset is resampled to a resolution of 0.35mm. Prior to the registration, the target vertebral region exposed to the pedicle screw insertion surgery is cropped from pre-operative MR images. After selecting the target vertebrae, the following procedure is employed to choose the suitable screw size and to plan

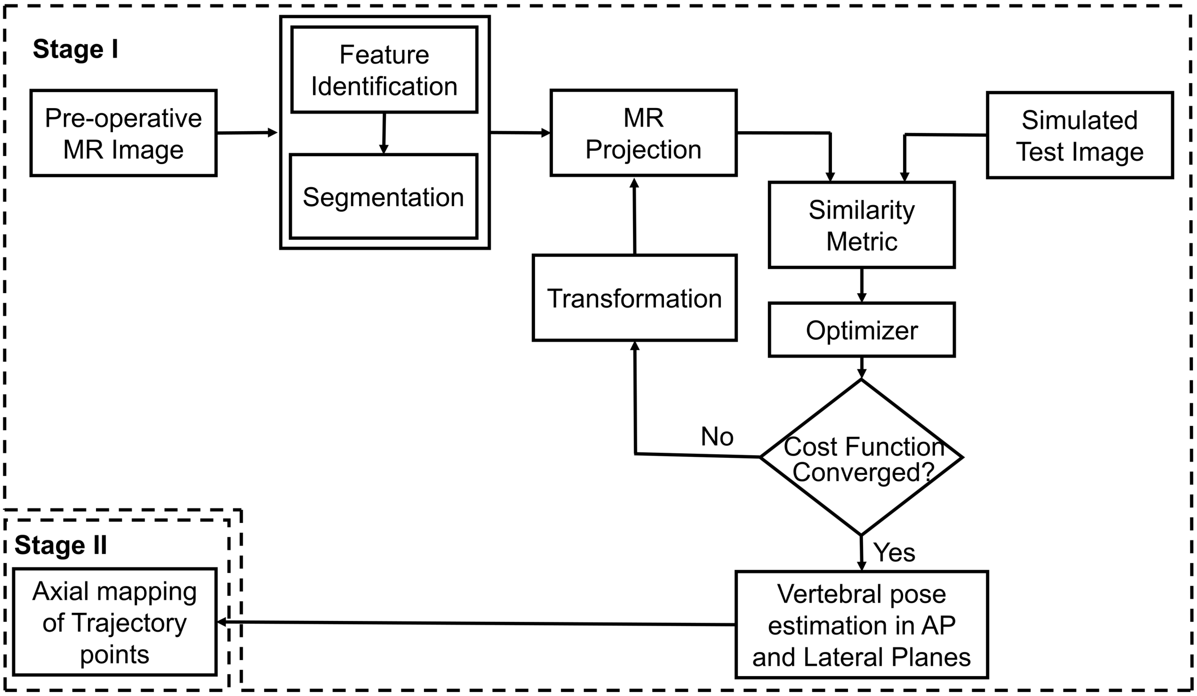

FIGURE 2. The workflow of feature-based 3D to 2D registration framework for pedicle screw registration.

the screw trajectory path for the target vertebrae [22]. The middle axial slice of the target vertebrae is identified, and the midline is drawn on the mid-axial slice from the anterior end to the posterior end such that the line axially divides the vertebrae symmetrically. The pedicle cross point is identified by viewing the pedicle's center point in a slice that has the widest pedicle in all the planes. The line joining the trajectory entry point, pedicle cross point and trajectory endpoint is known as Transverse Pedicular Angle (TPA) [22] for the screw trajectory plan. The trajectory is designed to allow for lateral initiation; as the probe is advanced, the axis is then pierced medially to the appropriate depth, passing through the pedicle cross point. The pedicle angle, which changes with the vertebral level, screw purchase level, and choice of the insertion site, determines the insertion angle.

After the trajectory planning, vertebral end plates are segmented from the target vertebrae using Otsu's thresholding technique. The suggested study assumes a 1mm thickness for the vertebral end plate because the end plate is positioned in multiple slices. The number of slices which have the end plates are identified using the ostu's method. Then each slice is segmented and added as shown in Fig.3(a), then the boundary of the vertebral end plate is extracted as shown in Fig.3(b). Then the extracted boundary is scanned in each slice with endplates to extract the extreme values as shown in Fig.3(c). Similarly, same technique is applied to extract the vertebral bottom endplate.

2) MR PROJECTION

The segmented 3D MR volume and its projected 2D image coordinates are parameterized by a projection geometry [15]. The projection geometry is defined by a world coordinate frame and the 3D world coordinate frame defines the position and orientation by translation and rotation of the 3D volume [16]. The 3D position are projected onto 2D detector plane as follows. Consider a source position at (x_s, y_s, z_s), object point O(x,y,z), projected point $P_{proj}(u, v)$, Center of Projection (COP), Source to Object Distance (SOD) and Source to Detector Distance (SDD) is depicted in the Fig.4. The relationship between the 3D volume and the corresponding 2D coordinates on the detector can be expressed as follows;

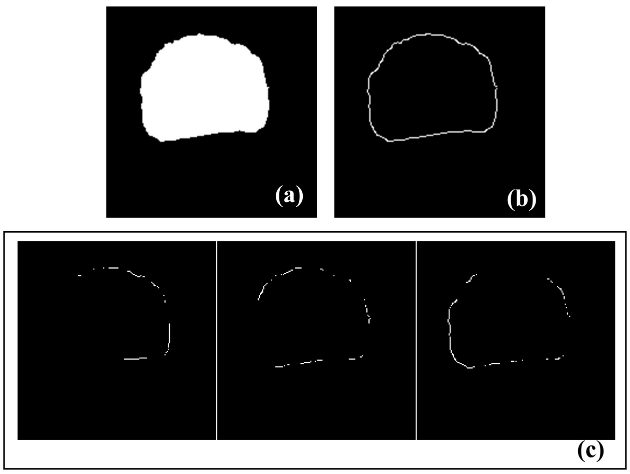

FIGURE 3. (a) Segmented vertebral end plate. (b) Boundary extracted end plate. (c) The extracted boundary values from individual slices.

$$\begin{bmatrix} u \\ v \end{bmatrix} \sim PM_{3\times 4} \begin{bmatrix} x \\ y \\ z \\ 1 \end{bmatrix} \tag{1}$$

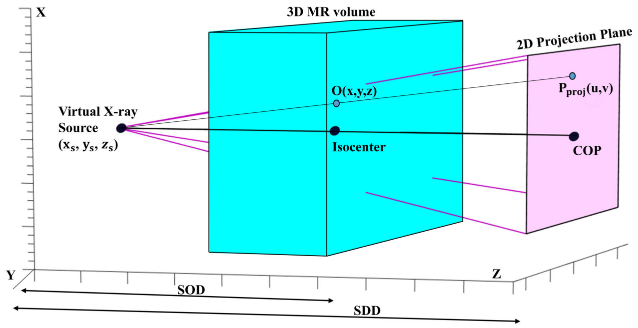

FIGURE 4. Depiction of single 3D point projection on the 2D plane.

where the $[x\ \ y\ \ z]^T$ is the segmented 3D MR volume coordinates and $[u\ \ v]^T$ is the projected point location in the detector coordinate. To match with the size of projection matrix, the size of 3D coordinates is represented in a homogeneous coordinate form by adding the extra scalar. The 3D coordinate is obtained by dividing the coordinate by the last coordinate in the same manner the projected location is obtained by dividing the first and second coordinate by the third coordinate. The projection matrix is expressed employing rigid body transformation and perspective projection in a homogeneous coordinate system and computed using imaging parameters that comprise both the extrinsic and intrinsic characteristics [17].

$$PM_{3\times4} = \begin{bmatrix} SDD & 0 & xs & 0 \\ 0 & SDD & ys & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \begin{bmatrix} & R_{3\times3} & & \begin{matrix} T_x \\ T_y \\ T_z \end{matrix} \\ 0 & 0 & 0 & 1 \end{bmatrix} \tag{2}$$

where xs and ys stand for the source position, T_x, T_y, T_z stand for the 3D volume translation, and R is a 3×3 rotation matrix in the x, y, and z direction. After the multiplication of projection matrix with 3D image coordinate results in a homogeneous coordinates of the projected 2D points as follows;

$$\begin{bmatrix} u \\ v \end{bmatrix} \sim \begin{bmatrix} P(1) \\ P(2) \\ P(3) \end{bmatrix} \tag{3}$$

Then the projected 2D coordinates are obatined by renormalization $u = P(1)/P(3)$ and $v = P(2)/P(3)$. The projection of 3D segmented MR image using a projection matrix genertae a simulated 2D image as shown in Fig. 5.

3) SIMILARITY MEASURE

The major goal of our research is to find similar metrics that can handle various clinical datasets collected from the Kasturba Medical College (KMC), Manipal. The simplest method involves two binary images to perform an exact pixel-by-pixel match by superimposing two shapes on top of one another and measuring how much of one fits into the other. Therefore, based on the position of the pixel values, the proposed work employed a simple binary image-matching technique [18]. The similarity between the fixed and projected image using the binary image matching technique is calculated as follows:

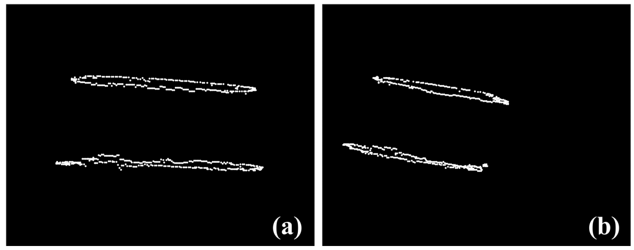

FIGURE 5. The projected image in (a) AP plane (b) Lateral plane.

$$SM = \sum_{i,j=1}^{m} \frac{x_{(i,j)} * y_{(i,j)}}{x_{(i,j)}} \tag{4}$$

The $x_{(i,j)}$ and $y_{(i,j)}$ represent the fixed and projected pixel values at the respective location. The value of the similarity measure will show how similar the floating image is to the fixed image. The higher similarity measure shows a better similarity between the images.

4) OPTIMIZATION

The optimization algorithm continuously tunes the transformation parameters to achieve the best alignment between the fixed and the projected images. The proposed framework employed Covariance Matrix Adaptation Evolution Strategy (CMAES) [19]. In the optimization algorithm, search space parameters are normally distributed and randomly generated about the current estimate based on the mean and covariance matrix. In each stage for a particular generation, similarity between the images is calculated for all the sample points. Based on the similarity value, the mean and covariance matrix values for the next generation are modified in accordance with the distribution value. The unnecessary iteration and premature termination are avoided in the objective function by using the stopping criterion. The stopping criterion was selected so that when the similarity between both images is one or the difference of the last ten objective functions is minimal.

B. STAGE II: TRAJECTORY REGISTRATION

Stage II includes the axial mapping of the trajectory position on the simulated data to the pre-operative MR image. The trajectory entry point and end point marked in the simulated dataset, the position of the trajectory point is axially mapped onto the pre-operative MR by utilizing the estimated transformation parameters of the vertebral body obtained in the Stage I. The trajectory entry and the endpoints having a length of 3.5cm are back projected onto pre-operative MR. The entry point and target point in pre-operative MR are determined by computing the intersection points of simulated datasets. The feature-based 3D to 2D registration determines the accuracy of the trajectory mapping. The registration error occurs mainly due to the segmentation error; this may cause imprecise trajectory mapping between the AP and lateral simulated image. The trajectory evaluation was performed on

AP, lateral, and a combination of both planes based on the ground truth data.

III. EXPERIMENTAL SETUP

The MR of the vertebral region from L2 to L4 are collected from the KMC hospital, Manipal. The acquired MR dataset has a slice thickness of 3mm and it is resampled to a resolution of 0.35mm. The cropping, trajectory planning of the target vertebrae, and extraction of the endplates are done as explained in section II-A1. Simulated X-ray images are generated in the absence of intra-operative X-rays in order to assess the proposed algorithm's precision, robustness, reproducibility of posture, and processing speed [8]. The simulated X-ray with trajectory positions are generated by varying the transformation parameters in the projection matrix following various clinical settings to achieve the best outcome. The variations in the transformation parameters are set by an experienced surgeon depending on the specific estimates of configuration variability. Since it was assumed that the spinal vertebrae would be seen inside the fluoroscopic field of vision, the extreme value that would place the spine outside of the detector was excluded.

The experiments using the simulated X-ray are carried out in three planes utilizing a few standard settings. The virtual Computer Assisted Radio Monitoring (C-arm) Source and detectors are designed to mimic realistic clinical settings by considering C-arm variability, operator expertise, positional assistance, and the intricacy of the operational setup. The nominal position of C-arm imaging system involves a SDD=1000 mm, SOD=500 mm, source position (xs, ys, zs=0, 0, 0), source to detector position=(0, 0, SDD), and source isocenter position =(0, 0, SDD/2). The translation and rotation parameters are selected randomly by the primary unit vector in 3D space to generate simulated test images.

A. EXPERIMENT 1: TEST ON TRAJECTORY POINTS REGISTRATION IN AP, LATERAL AND COMBINATION OF AP AND LATERAL PLANES

For the generation of simulated AP test images, 10 vertebral poses are selected by varying the translation and rotational parameters around the primary vector are derived from a Gaussian probability distribution around zero mean with three standard deviations (3σ) for each distribution as follows: $3\sigma_{T_x} = 3\sigma_{T_y} = 20\ mm, 3\sigma_{T_z} = 50\ mm, 3\sigma_{\theta_x} = 3\sigma_{\theta_y} = 3\sigma_{\theta_z} = 10^0$. The nominal transformation position to generate simulated X-ray image along AP axis are $(T_x, T_y, T_z, \theta_x, \theta_y, \theta_z) = (0, 0, 0, 0, 0, 0)$.

Similarly, the simulated lateral test images are generated by selecting 10 different vertebral poses with trajectory positions using the same settings as the simulated AP test images except for the rotation in the Y-direction. For the lateral test images θ_y value are generated around the mean 90^0 with variations of $3\sigma_{\theta_y} = 10^0$. The nominal transformation position to generate simulated X-ray image along lateral axis are $(T_x, T_y, T_z, \theta_x, \theta_y, \theta_z) = (0, 0, 0, 0, 90, 0)$.

The proposed work selected 10 vertebral poses in each AP and lateral plane with 50 different trajectory positions. So, the AP and lateral combinations led to an experiment on 5000 simulated images in both the right and left pedicle regions.

B. EXPERIMENT 2: TEST ON TRAJECTORY POINTS MAPPING IN AXIAL PLANES

During the surgery, surgeons continuously capture the images to evaluate the trajectory position to avoid the misplacement of the screw. The vertebral poses vary depending on the position of the C-arm, so each image acquisition has different transformation parameters. To mimic the real clinical scenario, we considered the set of 7 different vertebral poses in AP and lateral planes along with the trajectory length variations from 0.5cm to 3.5cm with 0.5cm increments. So the experiments were conducted on the combination of 7 AP and lateral poses with the right trajectory position to test the accuracy of the registration framework.

C. EVALUATION METHODS

Various regularizations are employed to assess the accuracy of 3D to 2D registration. These standards and rules describe the evaluation process, evaluation standards, and concept of ground truth registration etc., [6]. The pedicle screw placement error was evaluated in the 3D plane by measuring the distance between the registered trajectory to the planned trajectory. Uneri et al., [14] in the clinical PSS navigation system, provided a target localization error of 2mm in the 3D plane. In stage I, the results are validated by measuring the distance between the projected MR position with the simulated test image position in the 2D plane. If the average of all the positions smaller than 2mm [21] indicates that the estimated location was well within this projected boundary of the true vertebral pose.

After the registration of stage I, the trajectory position in the simulated test image is overlaid onto the projected MR image. Then the trajectory position is back traced using the estimated transformation parameter to map in the pre-operative 3D plane to evaluate the registration framework. The evaluation includes identifying the errors in the trajectory entry points, endpoints, trajectory angles in two planes, and the difference in trajectory lengths. The geometrical angle error and Euclidean distance are measured in the 3D plane. The surgeons assess all of the measures for dependability and to support the previously established assessments. The evaluation of the trajectory positions in 3D helps to detect breaches using the intra-operative AP and lateral images without the need for post-operative scans.

IV. RESULTS

The optimization continuously tunes the transformation parameter until the objective function stagnates at its ideal value, which improves the accuracy of vertebral pose estimation. After the vertebral pose estimation, the projected and feature extracted MR image closely matches the simulated ground truth datasets shape and structure. The erroneous registration in the AP and lateral direction lead to non-overlap of head and tail ends of trajectories on back-projection.

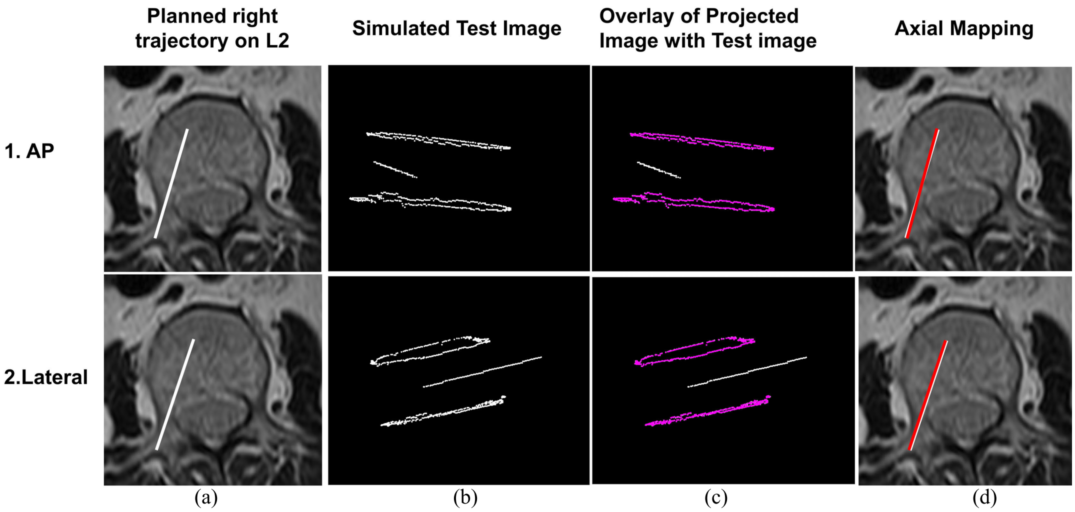

FIGURE 6. (a) Trajectory planned Pre-operative MR image. (b) Simulated Test image. (c) Overlay of projected MR with simulated test image. (d) Axial mapping of the trajectory position on the planned trajectory.

TABLE 1. The table shows the trajectory registration error in AP, lateral planes and combination of both the planes between the registered and planned trajectories.

Vertebral Level	Trajectory Position	AP Plane			Lateral Plane			AP+Lateral Plane			
		Head (mm)	Tail (mm)	Angle(°)	Head (mm)	Tail (mm)	Angle(°)	Head (mm)	Tail (mm)	Angle(°)	Trajectory Length Error(mm)
L2	Right	0.232	0.172	0.17	0.34	0.21	0.18	0.252	0.066	0.16	0.05
	Left	0.78	0.95	0.32	0.83	0.95	0.42	0.313	0.047	0.18	0.07
L3	Right	0.025	0.011	0.012	0.089	0.087	0.007	0.021	0.005	0.004	0.006
	Left	0.02	0.015	0.007	0.087	0.085	0.005	0.011	0.012	0.0084	0.003
L4	Right	0.031	0.032	0.0067	0.036	0.042	0.004	0.03	0.02	0.036	0.008
	Left	0.04	0.035	0.008	0.038	0.046	0.006	0.08	0.019	0.044	0.015

Fig.6(a) displays the trajectory planning in the right pedicle region of the L2 vertebrae. The trajectory entry and end points are positioned in multiple MR slices. For the display purpose, respective axial MR slices were intensity-wise cumulated, and the trajectory positions were overlaid on it. The trajectory planned MR image are projected with known transformation parameters to generate simulatd test image as shown in Fig.6(b). Accuracy of the vertebral pose estimation are measured by overlaying the projected MR image with the simulated test image as shown in Fig.6(c). Fig.6(d) displays the axially mapping of the trajectory positions along with the planned trajectory positions in the pre-operative MR.

Table 1 summarizes the absolute errors in the trajectory entry and end points and trajectory angle in both the AP and lateral planes, as explained in experiment 1 for each vertebral region. It also summarizes the absolute errors in the trajectory entry, endpoints, trajectory angle and trajectory length difference while estimating the vertebral pose by combining the AP and lateral poses as explained in experiment 2 for each vertebral region. The trajectory entry, end points and trajectory angle are measured by overlaying the registered trajectory with the planned trajectory positions. On optimization, the cost function value reaches one, indicating that the entire vertebra is registered correctly and there is no misalignment between the projected and simulated test images. The trajectory length differences are measured by calculating the difference between the actual and planned trajectory length. The accurate estimation of the vertebral poses in both planes results in a nonzero trajectory length difference.

The Fig.7 shows the axial mapping of the registered trajectory position on the planned trajectory in pre-operative MR image as explained in experiment 1 and the projected MR image in the AP and lateral planes to evaluate the estimated vertebral pose for each vertebral region on both the right and left pedicle region. The Fig.8 shows the axial mapping of the registered position on the planned trajectory in the pre-operative MR image as explained in experiment 2.

V. DISCUSSION

Due to the images acquired in different imaging modalities, the registration framework requires extraction of the features to mitigate the image intensity mismatch and tissue non-correspondence between the MR and X-ray image for better vertebral pose estimation. The proposed work extracts the vertebral end plate as a better feature for vertebral pose estimation. Additionally, feature extraction in the images will speed up the projection image generation during the optimization stage, supporting the framework's suitability for speeder intra-operative applications that help with clinically acceptable registration accuracy. This feature-based registration approach is especially appropriate for surgical operations where surgeons use their intuitive 3D to 2D landmark mapping knowledge to navigate surgical equipment. Each

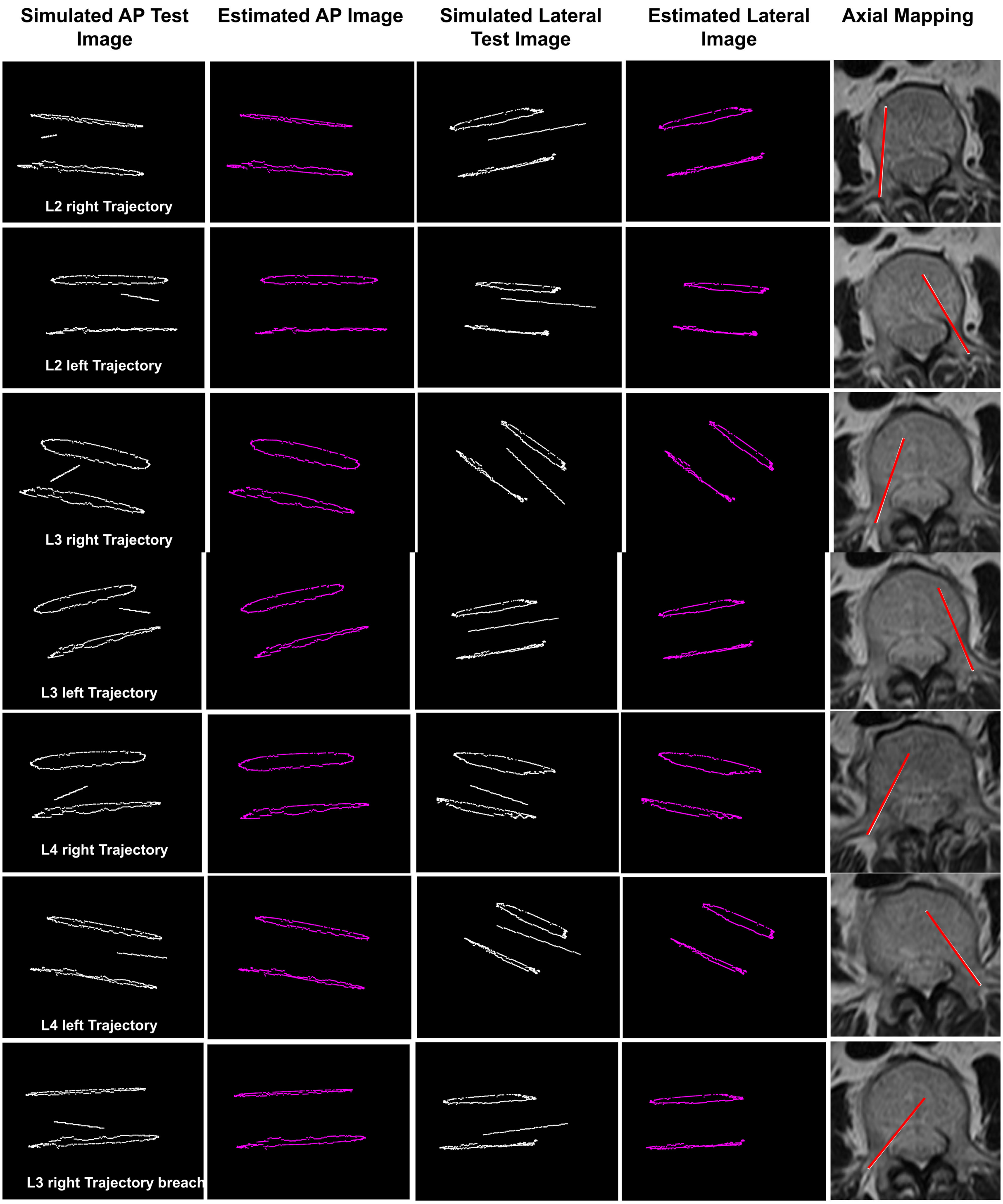

FIGURE 7. Validation of trajectory registration on the planned trajectory. Column 1 and 3 represents the simulated AP and Lateral test image genertaed with known transformation parameters. Column 2 and 4 displays the projected MR image genertaed by optimizing the cost function in AP and Lateral plane respectively. In column 5 red and white line indicates the true and estimated trajectory position on the pre-opeartive MR.

vertebra was subjected to deformable registration because of the vertebral deformation that occurred between the intra-operative and pre-operative environments. Due to this, the individual vertebra is cropped and closer registration parameters are estimated for individual vertebrae.

In simulated datasets, the proposed framework mean displacement error of head and tail ends in AP and lateral planes were found to be less than 1mm and the mean angular error in AP and lateral planes were found to be less than 1^0 in the 3D plane, which is better compared to

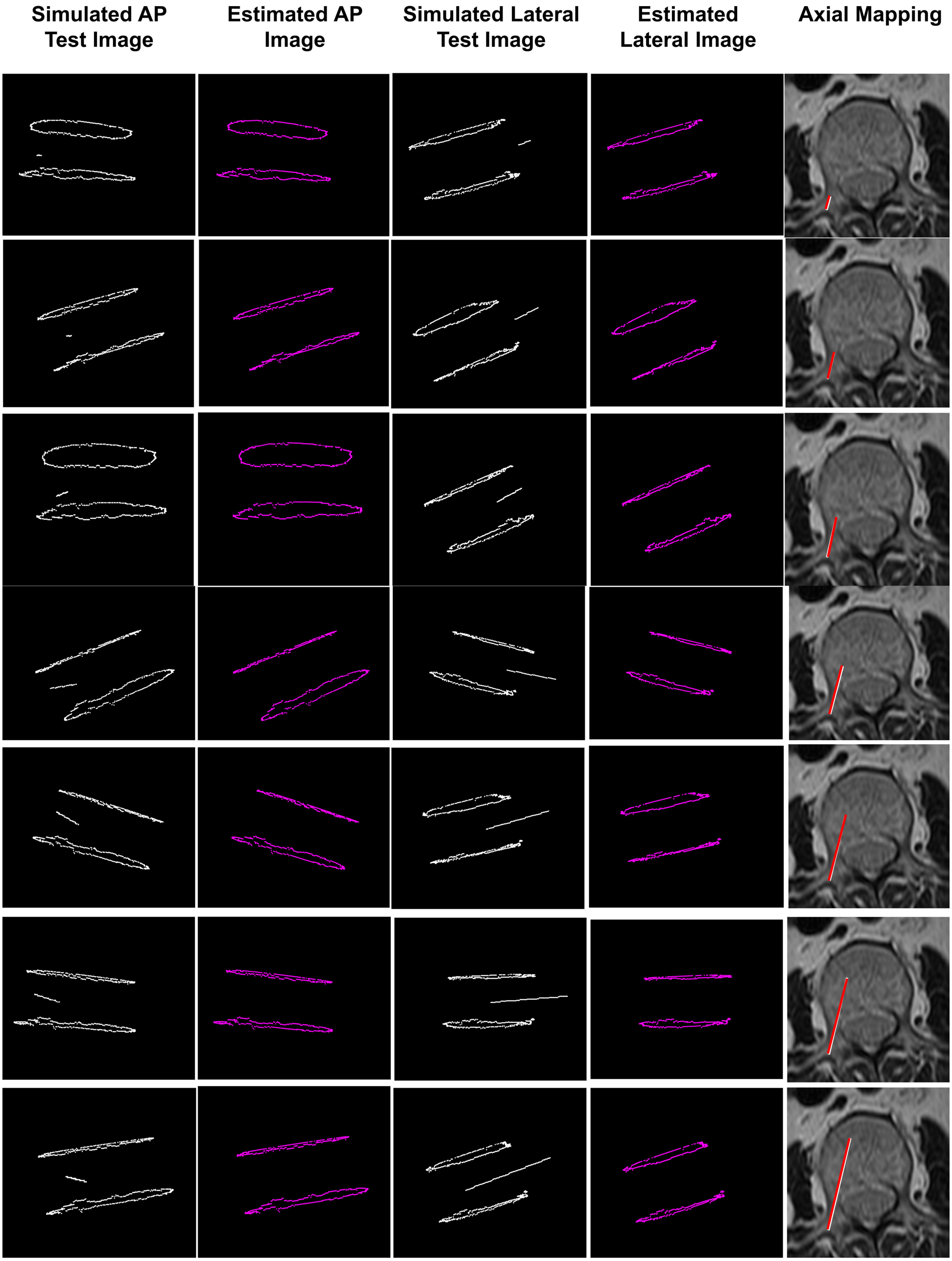

FIGURE 8. Validation of trajectory registration based on the planned trajectory length variations. Column 1 and 3 represents the simulated AP and Lateral test image genertaed with known transformation parameters. Column 2 and 4 displays the projected MR image genertaed by optimizing the cost function in AP and Lateral plane respectively. In column 5 red and white line indicates the true and estimated trajectory position on the pre-opeartive MR.

Naik et al., [22] where the results are evaluated in the 2D plane. The combination of the AP and lateral planes results in the mean displacement error of head and tail ends were found to be less than 0.5mm and the mean angulation error was found to be less than 0.5^0. The trajectory error length between the ground truth and the estimated trajectory was found to be less than 0.5mm. The trajectory mapping on the axial plane is done by considering the AP and lateral images with the angular separation of 90^0, whereas compared to Newell et al., [20] and Uneri et al., [13] the trajectory mapping on the 3D plane is done with angle separation between the images 20° and 10° respectively. The trajectory mapping on the axial plane can be further tested by considering the smaller angular separation between the images.

The trajectory registration accuracy depends on the vertebral pose estimation between the projected MR and simulated test image. The accuracy of the vertebral pose estimation relies on the feature extraction of the pre-operative MR image. The feature extraction includes segmenting the particular features such as pedicle region, spinous process or vertebral body, etc.,. In the developed framework, the vertebral end plate is extracted to register all the transformation parameters accurately. In the absence of an intra-operative X-ray image, the experiment is conducted by generating the simulated test image along with the trajectory positions. The extracted MR image is projected optimally to match it with the simulated test image to estimate the vertebral pose in all the planes. The accurately estimated transformation parameters are used to map the simulated test image's trajectory position on the axial plane.

VI. CONCLUSION

The developed feature-based multimodal registration accurately maps the trajectory position in the simulated test image to the pre-operative MR image. This registration procedure fits well in the routine clinical setup practiced in the local hospital and does not require additional equipment. The trajectory registration mainly depends on the vertebral pose estimation in the 2D plane. Vertebral pose estimation faces a challenge due to the images acquired from various sensors. So the designed framework extracts the particular feature to avoid the intensity mismatch between the pre-operative MR image and intra-operative X-ray. The extraction of the features has made registration faster for intra-operative applications. The vertebral pose estimation are validated by overlaying the projected MR with the simulated test image. The experiment was performed in the AP, Lateral, and combination of both AP and Lateral plane to test the accuracy of the trajectory registration framework. The current study focuses on the simulated data, so further steps to assess registration accuracy on the clinical data.

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CONFLICT OF INTEREST

The authors declare no conflict of interest.

ETHICS APPROVAL

The Indian Council of Medical Research (ICMRinstitutional)'s and national research committee's ethical guidelines were followed in all procedures carried out in studies involving human subjects. Ethics Committee: Kasturba Medical College and Kasturba Hospital Institutional Ethics Committee. CTRI Number: CTRI/2020/04/024490 and IEC No. 162/2020. Review date: March 02, 2020.

INFORMED CONSENT

Informed consent was obtained from all participants involved in the study.

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