2D/3D REGISTRATION AND MOTION TRACKING FOR SURGICAL INTERVENTIONS

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Abstracts

In order to use pre-operative images during an intervention for navigation, they must be registered to the patient’s co-ordinate system in the operating theatre or to an intra-operative image. For the registration to be valid in the case of patient movements, the registration must be updated or the patient movement must be tracked. One problem in this area is the registration of intra-operatively acquired X-ray fluoroscopies with 3D CT images obtained before the intervention as well as motion tracking for this set-up. The result can be used to support the placement of pedicle screws in spine surgery or aortic endoprostheses in transfemoral endovascular aneurysm management (TEAM). The different approaches to 2D/3D registration are discussed and a novel voxel-based method is presented: using a small part of the CT image covering only the vertebra of interest, pseudo-projections are computed and the resulting vertebra template is compared to the X-ray projection using a new similarity measure which is called pattern intensity. Application, performance and registration accuracy are discussed and demonstrated by application to images of a TEAM procedure and of a spine phantom.

Keywords: image-guided surgery, image registration, pattern intensity, similarity measure.

1. Introduction

Pre-operative 3D CT and MR images are used intra-operatively in an increasing number of interventions. One widespread application is neuro-surgery where instruments are tracked by a localizer system and displayed together with the pre-operative image to support navigation [1,2]. For properly aligned overlays, the image co-ordinate system and the intra-
operative co-ordinate system must be matched using registration methods. Various registration methods, based e.g. on markers or surfaces, have been proposed and studied (see. e.g. Refs. [3–6]), because application accuracy as well as clinical acceptance depend strongly on the selected method.

Pedicle screw placement in orthopaedic surgery is a further application where image guidance is used [7,8]. The screws must accurately be drilled into the vertebra to avoid damage, especially to the spinal cord. As in the case of neurosurgery, the drill may be tracked with a localizer system and overlaid onto the pre-operative CT image that is displayed to the surgeon during the intervention. For registration there are several possibilities. The surface of the vertebra can be palpated with a pointing device and registration can be done by fitting the palpated points onto the vertebra’s surface obtained by segmentation from the CT image. Registration can also be done on the basis of intra-operative X-ray projections acquired with a suitable calibrated X-ray device. For that purpose the location and orientation of the vertebra in the CT image with respect to the geometry of the X-ray device is determined using 2D/3D registration algorithms.

2D/3D registration of an X-ray projection with a CT image may also be used in the transfemoral endovascular aneurysm management (TEAM) procedure which is considered within the European Applications in Surgical Interventions (EASI) project [9]. Accurate placement of the aortic endoprosthesis in the endovascular treatment of abdominal aortic aneurysms is very important, and angiograms may be used to visualize the catheter with the endoprosthesis and the aneurysm. In the current practice there is, however, no contrast agent injected during deployment of the endoprosthesis. In this case the 2D/3D registration result can be used to project the aorta and the aneurysm obtained by segmentation from a pre-operative CT (A) image onto the intra-operative X-ray image to visualize aorta and aneurysm without contrast agent.

A 2D/3D registration algorithm suitable for clinical application must satisfy several requirements. These requirements concern registration accuracy, but also computation time. To be suitable for clinical practice, registration must be performed within a few minutes. In the case that registration can be done within a few seconds, the algorithm can also be used for re-registration after patient movements or tracking of patient movements. Furthermore, the algorithm must be able to cope with clinical images and their properties: there are differences in the 3D CT image and the X-ray projection caused by anatomical displacements which originate from different patient positioning in the CT scanner and on the operating table. In addition, if surgical instruments, as e.g. catheters, are used during an intervention, they are visible in the intra-operative X-ray image, but not in the pre-operative CT image.
In this paper we discuss the advantages and disadvantages of 2D/3D registration methods known from literature with respect to the aforementioned requirements and present a new approach which combines important advantages from known methods. With respect to the application, we focus on registration of a vertebra in intra-operative X-ray projections with a pre-operative 3D CT image. The result can be used in the context of pedicle screw placement and the TEAM procedure.

In the following section the well-known approaches to 2D/3D registration are outlined and discussed. In Sec. 3 the new approach [10] is presented and described in detail. Section 4 includes results gathered by application to images of a TEAM procedure. In addition, registration results for different vertebrae of a spine phantom are presented which give an indication of registration accuracy. Finally, the conclusions are summarized and possible directions for future work are proposed in Sec. 5.

2. Existing 2D/3D registration methods

A 2D/3D registration is used to determine the location and orientation of a 3D image with respect to the geometry of the X-ray device used for intra-operative imaging. The geometrical set-up is shown in Fig. 1 together with the co-ordinate systems of the X-ray device and the 3D image. The projection plane is placed in the $x$–$y$ plane with the origin in its centre and the X-ray source is above this plane.

For 2D/3D registration, there are two approaches available. One approach consists of voxel-based methods. Respective methods are based on
pseudo-projections and correlation as similarity measure. The other approach is based on surfaces and contours which have to be determined by segmentation from the images. Both approaches are outlined and discussed in the following two subsections.

2.1. Voxel-based methods

One possibility for registration with voxel-based methods is to compute pseudo-projections from the CT image and correlate them to the intra-operative X-ray images directly. Alternatively, correlation is evaluated for gradient images derived from the pseudo-projections and the intra-operative X-ray images. In both cases location and orientation of the CT image with respect to the projection geometry are determined by optimization of the correlation value [11,12].

These methods have the advantage that no sophisticated pre-processing of the CT image or X-ray projection such as segmentation is required. They are, however, very slow and time constraints for clinical application or even motion tracking can hardly be satisfied because the entire CT image is used within the computation of pseudo-projections. Furthermore, there are dissimilarities in pseudo-projections obtained from CT images and intra-operative X-ray images which degrade the registration result:

(1) Gantry tilt and other effects can lead to considerable geometric inaccuracies in CT images [13,14].
(2) The X-ray images can be affected by geometrical distortions. The curved screen of the image intensifier leads, e.g., to a pin-cushion distortion and the influence of the earth's magnetic field on the electrons in the image intensifier causes a characteristic S-shaped distortion.
(3) The sharpness of pseudo-projections is very different from that of X-ray images because of the CT slice thickness.
(4) There are several effects leading to different grey-value variations in the X-ray image and the pseudo-projection. For example, the absorption coefficients depend on the energy of the X-ray beam which is in general different for CT scanners and X-ray devices. Furthermore, different look-up-tables are used for contrast enhancement in the post-processing of X-ray images.
(5) If surgical instruments (e.g., catheters) are used during an intervention, they are visible in the intra-operative X-ray image, but not in the pre-operative CT image.
(6) Different patient positioning during image acquisition and the intervention can cause anatomical displacements between the images.
Geometrical distortions arising from the imaging equipment can be eliminated using suitable calibration procedures (see e.g., Refs. [15,16] for X-ray distortion correction). This is a general prerequisite for obtaining useful 2D/3D registration results. If necessary, the sharpness of X-ray images and pseudo-projections can be adjusted by proper smoothing of the images. It is also possible to model some effects leading to grey-value differences and generate more similar pseudo-projections and X-ray images in that way [12]. Overlaid instruments or anatomical displacements cause, however, serious difficulties and it is a major disadvantage of 2D/3D registration methods based on grey-value correlation that they are not robust with respect to them.

2.2. Surface-based methods

Surface-based methods [8,17,18] require segmentation of the vertebra in the 3D CT image as well as outlining the corresponding contours in the projection image. Registration is performed by optimizing the distance between the 3D vertebra surface and the lines connecting contour points with the X-ray source. This shows that proper segmentation and contour outlining is most important for accurate registration results.

These methods are fast enough to be used during an intervention, because no pseudoprojections must be calculated. They can cope with surgical instruments overlaid onto an intra-operative X-ray image if they do not hide the vertebra of interest completely. Furthermore, a possible bending of the spine does not degrade accuracy, because a single vertebra is used for registration.

Fig. 2. Slice of the CT image with the vertebra of interest (a) and intra-operative X-ray projection acquired during a TEAM procedure (b).
A crucial problem is, however, related to outlining of the vertebra contours which may be done interactively or with a co-operative approach between registration and 2D segmentation [19,20]. This is illustrated in Fig. 2 which shows a slice of a CT image with the vertebra of interest and an intra-operative X-ray image of a TEAM procedure. The transverse processes are clearly visible in the CT slice, whereas they are not visible in the X-ray projection. This shows that accurate contour outlining can be a challenging problem.

3. A new voxel-based approach

In the following subsections a voxel-based 2D/3D registration method is presented which combines important advantages of the methods outlined in the Sec. 2. First, the new method is motivated and its basic concept is introduced. Then the registration algorithm is described. In the remaining subsections the different parts of the algorithm, i.e. pre-processing of the images, computation of pseudo-projections, the similarity measure and the optimization strategy, are discussed in more detail. The algorithm is basically the same as the one described in Ref. [10]. However, several details have been changed, making it more robust and leading to a considerable reduction of computation time for registration.

3.1. Basic concept

The basic idea of the new voxel-based 2D/3D registration approach is to

Fig. 3. X-ray image with the registered vertebra template subtracted (a) and corresponding vertebra template (b). For visualisation the grey-values of the vertebra template have been inverted and scaled.
compute pseudo-projections using only a small part of the CT image covering the vertebra. The resulting vertebra template is subtracted from the X-ray image, and registration is done by rubbing out the vertebra structures in the X-ray image. In the case of optimal registration, these structures will vanish completely and altogether there will be less structures visible.

For illustration, the CT image and the X-ray projection of Fig. 2 have been used. Figure 3 shows the intra-operative X-ray image with the registered vertebra template subtracted, as well as the vertebra template itself. After registration, all structures due to the vertebra-of-interest vanish, indicating proper registration. Of course there are other structures left in the area of this vertebra after subtraction: an organ as well as a catheter are clearly visible.

3.2. Algorithm

Location and orientation of the CT image with respect to the geometry of the X-ray device used for intra-operative imaging are described by a rigid transformation:

\[ x = R(\omega) x' + t. \]

The point \( x \) refers to the co-ordinate system of the X-ray device and the point \( x' \) to that of the CT image (compare Fig. 1). The latter co-ordinate system has its origin in the centre of the CT volume-of-interest with the vertebra. The rotation matrix \( R(\omega) \) defined by:

\[ R(\omega)x = x + \frac{\sin \alpha}{\alpha} \omega \times x + \frac{1 - \cos \alpha}{\alpha^2} \omega \times (\omega \times x), \]

where \( \alpha = \sqrt{\omega_x^2 + \omega_y^2 + \omega_z^2} \) is represented by the vector \( \omega = (\omega_x, \omega_y, \omega_z) \). This vector is aligned along the rotation axis and its Euclidean length \( \alpha \) defines the rotation angle. The vector \( t = (t_x, t_y, t_z) \) defines the translation. The components \( t_x \) and \( t_y \) correspond to a shift parallel to the projection plane, whereas \( t_z \) describes the height above the projection plane.

The 2D/3D registration algorithm determines the parameters \( \omega \) and \( t \) given initial values \( \omega_0 \) and \( t_0 \) and consists of the following steps:

1. Segmentation of the vertebra of interest in the pre-operative CT image.
2. Subtraction of the average grey-value of the tissue around the vertebra from the CT image and further pre-processing (see Sec. 3.3).
3. Computation of a pseudo-projection for the actual parameters \( \omega \) and \( t \) taking only the CT volume with the segmented vertebra into account. Since the average grey-value of the tissue directly adjacent to the vertebra has been subtracted from the CT image, the pseudo-projections show the grey-value variation due to the presence of the vertebra only.
(4) Scaling of the pseudo-projections' grey-values and subtraction from the X-ray image. For a proper grey-value scaling and the correct location and orientation of the CT image, the structures in the X-ray projection corresponding to the vertebra will vanish, and overall there will be less structures visible after subtraction.

(5) Calculation of a similarity measure which characterizes the "structuredness" of the X-ray image after subtraction of the scaled pseudo-projection.

(6) Optimization of the similarity measure with respect to the grey-value scaling and the parameters $\omega$ and $t$. Within each iteration of the optimization, steps (3) to (5) must be repeated.

Pre-processing of the CT image in steps (1) and (2) can be done prior to the intervention where time is not crucial. Registration itself is done in steps (3) to (6) and most time is spent in the computation of pseudo-projections from the CT volume-of-interest with the vertebra. Nevertheless, this approach allows for a much faster implementation than other voxel-based 2D/3D registration methods because they use the whole CT image for computing pseudo-projections. In contrast to contour-based 2D/3D registration methods, the new approach does not require outlining of the vertebra's contours in the X-ray projection.

3.3. Pre-processing of the CT image

Segmentation of the vertebra of interest can be done, for instance, with the automatic algorithm based on a grey-value profile parser proposed in Ref. [21]. In that way a binary image of the segmented vertebra is obtained in addition to the CT image. Given both images, the average grey-value of the tissue around the vertebra can be computed using elementary image-processing operations. First, two iterated dilation operations are applied to the binary image. Second, the surrounding of the vertebra is determined by performing a bitwise exclusive-or operation of the dilated image and the original binary image. Third, the grey-values of all voxels in the CT image which belong to the surrounding of the vertebra are averaged.

Further pre-processing is necessary to tune the quality of pseudo-projections computed from the CT image for registration purposes. On the one hand, pseudo-projections have an anisotropic sharpness which is due to the rather large slice thickness of CT images compared to the resolution in a slice. This anisotropy may lead to a systematic bias in the registration result and each slice of the CT image is smoothed with a uniform filter to compensate for this effect. On the other hand, the large slice-to-slice distance of CT images degrades the quality of pseudo-projections if they are computed
without interpolation between adjacent slices. These effects cause small oscillations of the similarity measure which complicate optimization and increase the danger of being trapped in small local optima. Interpolation of the CT grey-values may be done within the computation of pseudo-projections which prolongs, however, time for registration during the intervention.

Another possibility consists of supersampling the CT image. This can be done before the intervention when time is not crucial, but the memory consumption of the algorithm is increased in that way. Nevertheless, supersampling is used, because the volume of interest with the vertebra is rather small and typically only about $250 \times 250 \times 25$ voxels large for a CT image with a voxel size of $0.5 \text{ mm} \times 0.5 \text{ mm} \times 2 \text{ mm}$.

3.4. X-ray projection and pseudo-projection

Pseudo-projections are calculated by simply summing up the grey-values inside the segmented volume along the ray from the X-ray source to a pixel in the projection plane. The pseudo-projections obtained in that way are not as sharp as an X-ray projection because the slice thickness as well as the slice-to-slice distance of CT images are rather large. To adjust the sharpness, a uniform moving average filter is applied to the X-ray image as a preprocessing step. The size of the optimal filter mask depends on the size of a CT voxel and on the projection image’s pixel size.

Experiments indicate that the resolution of the X-ray image and the pseudo-projection can be reduced without significantly degrading the accuracy of the final registration. This observation is used to speed up registration because most computation time is spent during the computation of pseudo-projections. Using e.g. $256 \times 256$ images instead of $512 \times 512$ images, the computation time is approximately reduced by a factor four. Depending on the size of the vertebra in the X-ray image, it is also possible to use $128 \times 128$ images leading to a total time reduction by a factor of about 16.

Finally, it should be noted that we have not attempted to use a detailed model to correct for the grey-value variations between the X-ray and pseudo-projections. Different look-up-tables used for contrast enhancement in the post-processing of X-ray images and different beam energies of CT scanners and X-ray devices are only two of the numerous reasons for these differences which make correction with a detailed model a challenging task. Instead, the grey-values in the pseudo-projection are scaled with a proper factor. This factor is a parameter which has to be estimated in addition to the parameters describing location and orientation of the CT image.
3.5. Similarity measure

Correlation [22], entropy [23–25] and mutual information [26] are typical quantities to be used as similarity measures in medical image registration. These similarity measures turned out to be insufficient and a new measure which we call pattern intensity has been defined to successfully solve the registration problem considered.

As explained and illustrated in Sec. 3.1, the structures due to the vertebra of interest will vanish in the case of proper registration, and overall there will be less structures visible. A suitable similarity measure should, therefore, characterize the "structuredness" of the difference image $I_{\text{diff}}(x,y)$, i.e. of the X-ray projection $I_{\text{x-ray}}(x,y)$ after subtraction of the grey-value scaled pseudoprojection. This can be done by assigning a small value to points in the neighbourhood of structures such as grey-value edges or lines, and a large value to points in areas showing only little grey-value variation.

A quantity leading to the desired effect is the pattern intensity:

$$P_{r,\sigma}(I_{\text{diff}}) = \sum_{x,y} \sum_{(x-v)^2+(y-w)^2 \leq r^2} \frac{\sigma^2}{\sigma^2 + (I_{\text{diff}}(x,y) - I_{\text{diff}}(v,w))^2},$$

which depends on two parameters. The parameter $r$ defines the size of the neighbourhood in which grey-value variations are taken into account. The parameter $\sigma$ is a kind of sensitivity controlling whether a grey-value variation is considered to be a structure or not. To avoid summation over all pixels within evaluation of the pattern intensity, the difference

$$\Delta P_{r,\sigma}(I_{\text{diff}}) = P_{r,\sigma}(I_{\text{diff}}) - P_{r,\sigma}(I_{\text{x-ray}})$$

can be used: for evaluation of $\Delta P_{r,\sigma}(I_{\text{diff}})$ only the pixels belonging to the projected vertebra and pixels with a distance less or equal to $r$ from any pixel of the projected vertebra have to be taken into account. In that way evaluation of the pattern intensity becomes much faster, because the projected vertebra is much smaller than the pseudoprojection itself.

3.6. Optimization

The parameters of the rigid transformation are varied separately according to the sequence $t_x$, $t_y$, $\omega_x$, $\omega_y$, $\omega_z$, and $t_z$. This is done repeatedly, until six subsequent maximizations do not further increase the similarity measure. It should be mentioned that variation of the parameter $t_z$ is done by shifting along the direction given by the X-ray source and the centre of the CT volume-of-interest and, therefore, also affects the translations $t_x$ and $t_y$. 

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Furthermore, it should be noted that the grey-value scaling is adjusted, whenever one of the other parameters is changed by varying this quantity between zero and a maximal value with a predefined increment.

Maximization with respect to a single parameter is done by going with an adjustable step size in the direction where the similarity measure increases until the maximum is found. The step size is increased, if a predefined number of successive steps in the same direction increase the similarity measure. If neither a step forwards nor a step backwards increases the similarity measure, the step size is reduced as long as it is larger than a minimum value.

For each parameter, separate values of the step size and the minimum step size are used. At the beginning of the optimization the step size is initialized by a multiple of its minimum value.

4. Examples

To illustrate the performance of the 2D/3D registration algorithm presented in the previous section, the algorithm has been applied to clinical data of a TEAM procedure. Registration has been performed for different sets of start values and a plot of the similarity measure as a function of the rotation and translation parameters has been generated to show that the algorithm has a proper convergence range and finds a well-pronounced optimum of the similarity measure. These results are included in Sec. 4.1. In addition, 2D/3D registration has been performed for images of a spine phantom and results for different vertebrae have been compared to get an indication of registration accuracy (see also Ref. [27]). The results are included in the Sec. 4.2.

4.1. Clinical data set

A slice of the CT image with the vertebra used for registration and the intra-operative X-ray projection are shown in Fig. 2. The image has 512 x 512 x 123 voxels, a resolution of 0.488 mm in a slice, a slice-to-slice distance of 2 mm, and a slice thickness of 5 mm. The X-ray projection refers to a geometry defined by an image intensifier of 12" diameter and an X-ray source 1 m above it. To get pseudo-projections with isotropic sharpness, each slice of the CT image was smoothed with a uniform 5 x 5 pixel filter. Smoothing of the X-ray projection with a uniform 5 x 5 pixel filter was also suitable for adjusting its sharpness to that of the pseudo-projection. In addition, the X-ray projection's resolution has been reduced from 512 x 512 to 128 x 128 to speed up registration.

With the parameters of the pattern intensity given by \( r = 3 \) and \( \sigma = 10 \), registrations have been performed using 14 different sets of start values \( \omega_0 \).
and $t_0$. They have been generated by adding a rotation of ±0.1 rad, i.e. ±6°, to each of the rotation parameters obtained by manual registration or ±0.2 rad, i.e. ±12°, to one of them. The translation parameters have been selected in a way that a rough overlay of the projection template with the vertebra of interest in the X-ray projection was obtained. The deviations from the manually determined translations were 2 to 5 mm for the shifts parallel to the projection plane and up to 50 mm for the height above the projection plane.

Computation time for registration was about 60 to 90 s on a SparcStation 20 with a 85 MHz SuperSparc II CPU. After registration, the pattern intensity had values around 1970 and there was only one set of start values for which optimization was trapped in a local maximum with much smaller values. When not trapped in a local maximum, the rotation and translation parameters were $\omega_x = 1.559 \pm 0.002$ rad, $\omega_y = -0.072 \pm 0.004$ rad, $\omega_z = 0.106 \pm 0.003$ rad, $t_x = -18.51 \pm 0.06$ mm, $t_y = -14.98 \pm 0.11$ mm, and $t_z = 300.56 \pm 1.48$ mm. This shows that the registration method has a proper convergence range and that consistent parameter values are obtained if the start values are inside the convergence range. The X-ray projection with the registered vertebra template subtracted as well as the vertebra template itself are shown in Fig. 3. All structures of the vertebra of interest vanish, which

![Graphs showing pattern intensity in dependence of rotation and translation parameters.](image)

Fig. 4. Pattern intensity in dependence of the rotation and translation parameters. The centre of each plot corresponds with the optimal parameter values.
indicates that the rotation and translation parameters correspond with a good registration. These results also show that the new 2D/3D registration method is robust with respect to additional structures such as catheters or organs in the area of the vertebra of interest.

Furthermore, plots of the pattern intensity as a function of the rotation and translation parameters have been generated. The grey-value scaling has been optimized whenever the similarity measure was evaluated. The results are shown in Fig. 4. The plots show that the pattern intensity has a pronounced maximum corresponding to the optimal registration parameters and that all six parameters characterizing location and orientation of the CT image can be determined from one projection. The pattern intensity is very sensitive to the rotational parameters \( \omega_x, \omega_y \), and \( \omega_z \) and the shifts \( t_x \) and \( t_y \) parallel to the projection plane, whereas dependence on the height \( t_z \) above the projection plane is weaker. This effect can be attributed to the projection geometry and a lack of information in the projection direction.

4.2. Images of the spine phantom

The CT image with the spine phantom has a size of 320 \( \times \) 320 \( \times \) 94 voxels and the voxel size is 1.094 mm \( \times \) 1.094 mm \( \times \) 3 mm. X-ray projections showing a frontal and a lateral view of the spine are shown in Fig. 5. To reduce the influence of geometrical inaccuracies, a pin-cushion distortion correction has been performed. The projections refer to a geometry defined by an image intensifier of 15" diameter and an X-ray source 1125 mm above it. For registration, the X-ray projections have been smoothed with a uniform 3 \( \times \) 3 pixel

![Fig. 5. X-ray images of the spine phantom showing a frontal (a) and a lateral (b) view.](image-url)
TABLE I
Rotation and translation parameters after registration

<table>
<thead>
<tr>
<th>Vertebra</th>
<th>$\omega_x$ (rad)</th>
<th>$\omega_y$ (rad)</th>
<th>$\omega_z$ (rad)</th>
<th>$t_x$ (mm)</th>
<th>$t_y$ (mm)</th>
<th>$t_z$ (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>L2 (frontal view)</td>
<td>1.723</td>
<td>0.025</td>
<td>0.090</td>
<td>29.19</td>
<td>-27.24</td>
<td>202.19</td>
</tr>
<tr>
<td>L3 (frontal view)</td>
<td>1.720</td>
<td>0.030</td>
<td>0.090</td>
<td>28.85</td>
<td>-27.50</td>
<td>202.78</td>
</tr>
<tr>
<td>L4 (frontal view)</td>
<td>1.719</td>
<td>0.029</td>
<td>0.089</td>
<td>28.74</td>
<td>-27.77</td>
<td>199.32</td>
</tr>
<tr>
<td>L2 (lateral view)</td>
<td>1.553</td>
<td>-0.574</td>
<td>0.574</td>
<td>6.52</td>
<td>-30.17</td>
<td>331.67</td>
</tr>
<tr>
<td>L3 (lateral view)</td>
<td>1.533</td>
<td>-0.574</td>
<td>0.600</td>
<td>6.84</td>
<td>-30.03</td>
<td>331.00</td>
</tr>
<tr>
<td>L4 (lateral view)</td>
<td>1.554</td>
<td>-0.574</td>
<td>0.594</td>
<td>7.12</td>
<td>-29.96</td>
<td>332.55</td>
</tr>
</tbody>
</table>

Filter to adjust the sharpness. In addition, the resolution has been reduced from 512 \times 512 to 256 \times 256 in order to save computation time.

Registration has been performed for the vertebrae L2, L3 and L4. Before registration, the rotation was 0.17 to 0.25 rad, i.e. 10 to 15°, away from the optimum. The deviation of the translation was 3 to 5 mm for $t_x$ and $t_y$, and 40 to 70 mm for the height $t_z$ above the projection plane. Computation time for registration was about 60 to 90 s. The parameters of the pattern intensity were $r = 3$ and $\sigma = 10$. The values of the parameters $\omega$ and $t$ after registration are listed in Table I. They refer to a CT co-ordinate system with its origin in the centre of the CT image and not in the centre of the CT volume-of-interest, to make a comparison of the translation parameters for the different vertebrae possible. Figures 6 and 7 show the X-ray projections with the vertebra template subtracted.

Looking at the results for the frontal view (Fig. 6), the structures due to the
vertebra of interest vanish completely after registration. This is a strong indication that the proper location and orientation of the CT volume with respect to the projection geometry of the X-ray device has been found. Comparing the values of the parameters $\omega$ and $t$ (Table I), the registration accuracy can be estimated. The deviation in each rotational parameter is below 0.01 rad, i.e. 0.6°, the translation parameters $t_x$ and $t_y$ agree within 1 mm, while shifts $t_z$ perpendicular to the projection plane correspond within only 3 mm.

The results for the lateral view are as good as the results for the frontal view: the vertebra of interest vanishes completely after registration (Fig. 7) and the values of the parameters $\omega$ and $t$ (Table I) confirm high accuracy for five of the six parameters, whereas the accuracy of the height $t_z$ above the projection plane is significantly lower. This effect can be attributed to the projection geometry.

5. Conclusions

A new voxel-based 2D/3D registration method has been presented which combines advantages of the methods known from the literature and avoids their disadvantages. Compared to contour-based 2D/3D registration methods, the new approach does not require accurate outlining of the vertebra contours in the X-ray projection which is a challenging task. Compared to other voxel-based 2D/3D registration methods, the new method is much faster because pseudo-projections are computed using only a small part of the CT image covering the vertebra of interest. The approach overcomes the problems with anatomical displacements, because the vertebra is a rigid structure and anatomical displacements in other regions do not affect the registration. Furthermore, the new approach is robust with respect to additional structures such as catheters or organs in the area of the X-ray projection showing the vertebra of interest.
With the new registration method all six parameters characterizing location and orientation of a CT image can be determined from one projection. Results for a spine phantom indicate high accuracy for five parameters. Because of the projection geometry, the accuracy of the height above the projection plane is significantly lower. This poses no problem if registration is performed to superimpose information from a CT(A) image onto the X-ray projection, as may be done during the TEAM procedure to visualise the aorta and the aneurysm without injecting contrast agent. High accuracy for all parameters, as of required e.g. for spine surgery, may be obtained by using two X-ray projections showing different views simultaneously for registration.

In the current implementation, time for registration is about 60 to 90 s which is acceptable if registration or re-registration need to be performed only a few times. Most time is spent in the computation of pseudo-projections for which a simple ray-casting approach is used. This time can be reduced with refined algorithms for volume rendering which promise a considerable speed-up. This indicates one way to reduce time for registration and meet the more restrictive time constraints for tracking.

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