Medical Imaging 1999

Image Processing

Kenneth M. Hanson
Chair/Editor

22–25 February 1999
San Diego, California

Sponsored by
SPIE—The International Society for Optical Engineering

Cooperating Organizations
AAPM—American Association of Physicists in Medicine
APS—American Physiological Society
FDA Center for Devices and Radiological Health
IS&T—The Society for Imaging Science and Technology
NEMA—National Electrical Manufacturers Association/Diagnostic Imaging and Therapy Systems Division
RSNA—Radiological Society of North America
SCAR—Society for Computer Applications in Radiology

Published by
SPIE—The International Society for Optical Engineering

Volume 3661
Part Two of Two Parts

SPIE is an international technical society dedicated to advancing engineering and scientific applications of optical, photonic, imaging, electronic, and optoelectronic technologies.
Fast voxel-based 2D/3D registration algorithm using a volume rendering method based on the shear-warp factorization

Jürgen Weese

Roland Göcke

Graeme P. Penney

Paul Desmedt

Thorsten M. Buzug

Heidrun Schumann

Philips Research Laboratories, Röntgenstraße 24 – 26, D-22335 Hamburg, Germany

Department of Computer Science, University of Rostock, D-18051 Rostock, Germany

Computational Imaging Science Group, Division of Radiological Sciences, Guys Hospital, London Bridge, London SE1 9RT, UK

EVM Advanced Development, Philips Medical Systems Nederland B.V., Veenpluis 4-6, NL-5680 DA Best, The Netherlands

RheinAhrCampus Remagen, University of Applied Sciences, Südallee 2, D-53424 Remagen, Germany

ABSTRACT

2D/3D registration makes it possible to use pre-operative CT scans for navigation purposes during X-ray fluoroscopy guided interventions. We present a fast voxel-based method for this registration task, which uses a recently introduced similarity measure (pattern intensity). This measure is especially suitable for 2D/3D registration, because it is robust with respect to structures such as a stent visible in the X-ray fluoroscopy image but not in the CT scan. The method uses only a part of the CT scan for the generation of digitally reconstructed radiographs (DRRs) to accelerate their computation. Nevertheless, computation time is crucial for intra-operative application and a further speed-up is required, because numerous DRRs must be computed. For that reason, the suitability of different volume rendering methods for 2D/3D registration has been investigated. A method based on the shear-warp factorization of the viewing transformation turned out to be especially suitable and builds the basis of the registration algorithm. The algorithm has been applied to images of a spine phantom and to clinical images. For comparison, registration results have been calculated using ray-casting. The shear-warp factorization based rendering method accelerates registration by a factor of up to seven compared to ray-casting without degrading registration accuracy. Using a vertebra as feature for registration, computation time is in the range of 3-4s (Sun UltraSparc, 300 MHz) which is acceptable for intra-operative application.

Keywords: image guided surgery, 2D/3D registration, volume rendering, similarity measure

1. INTRODUCTION

With 2D/3D registration methods, the location and orientation of a 3D image can be determined with respect to the projection geometry used to acquire an X-ray projection image. In the context of image guided surgery, 2D/3D registration methods have been considered in order to combine the information of pre-operative 3D CT images with intra-operative X-ray fluoroscopy images. On the one hand, the registration result can be used to display the position of surgical instruments, which are being tracked by a localizer, in the pre-operative CT image. On the other hand, the registration result makes it possible to project pre-operative planning information or important anatomical structures from the CT image onto the X-ray fluoroscopy image. The computation time of the registration algorithm is limited for intra-operative application, and it is desirable to perform registration within a few seconds. This poses a problem for gray-value based approaches which require the computation of numerous digitally reconstructed radiographs (DRRs) from the CT image to determine the DRR optimally corresponding with the X-ray projection image.

In this contribution, a fast gray-value based method for the registration of X-ray projection images with 3D CT images is presented. The method is based on the concepts proposed in Ref. 2 and differs from other gray-value based approaches, because it uses only a part of the CT image for computing DRRs rather than the whole CT image. This part should contain...
an anatomical structure which is clearly visible in the CT image as well as in the X-ray projection image. In the following, a vertebra is used as an example of such a structure. In the available image data, the entire CT image is about 20 times larger than the minimum enclosing box containing the vertebra. To compute the registration result, a DRR is subtracted from the X-ray projection image after scaling the gray-values of the DRR, and the location and orientation of the CT image as well as the gray-value scaling are varied, until the anatomical structure vanishes in the difference image. This is illustrated in Fig. 1 which shows an X-ray projection image, a DRR of the vertebra used as feature for registration and the X-ray projection image with the gray-value scaled DRR subtracted. After registration the vertebra vanishes almost completely in the difference image. This effect is quantified using the pattern intensity as similarity measure.

Figure 1. X-ray projection image (a), DRR of the vertebra used for registration (b), and X-ray projection image with the gray-value scaled DRR subtracted (c). The vertebra used for registration is marked by an arrow (a). After registration this vertebra vanishes in the difference image (c). For optimal reproduction of the DRR with the vertebra (b), the contrast has been inverted.

The pattern intensity operates on the difference image \( I_{\text{diff}} \) and is defined according to

\[
P = \sum_{i,j} \sigma^2 \sum_{(i-\delta)x,(j-\delta)y} \frac{\sigma^2}{\sigma^2 + (I_{\text{diff}}(i,j) - I_{\text{diff}}(k,l))^2}.
\]

The parameter \( \sigma \) controls whether a gray-value variation is considered to be a structure or not. The parameter \( r \) defines the size of the neighborhood in which gray-value variations are compared. The pattern intensity assigns values approaching unity to pixels in areas showing only little gray-value variation and a value approaching zero to points in the neighborhood of structures such as gray-value edges or lines. Compared to other similarity measures, the pattern intensity is especially robust with respect to additional structures such as soft tissue structures or stents in the area of the vertebra of interest. This property is important for the robust and reliable registration of clinical images.

Experiments with a ray-casting algorithm have shown that almost the entire computation time for registration is spent on the generation of DRRs, although only a small part of the CT image is taken into account. For that reason the suitability of different volume rendering methods for 2D/3D registration has been investigated and a method based on the shear-warp factorization of the viewing transformation has been selected. In the following section, the requirements of a rendering method for 2D/3D registration are summarized, and a method based on the shear-warp factorization is described. The properties of the shear-warp factorization based rendering method allow for a specific optimization method which is presented in Section 3. For testing the registration algorithm, images of a spine phantom and clinical data sets have been used. Section 4 contains information about the image data sets and the experiments which have been performed. The results are discussed in Section 5. In particular, the influence of the rendering method on computation time and registration accuracy is considered. The conclusions are summarized in Section 6.

2. FAST VOLUME RENDERING BASED ON THE SHEAR-WARP FACTORIZATION

For 2D/3D registration, the geometry used to acquire the X-ray projection image must be accurately known, and geometric distortions (e.g. pincushion distortion) must be compensated. The projection geometry is used to accurately compute DRRs, and so volume rendering methods suitable for perspective projections must be used. Rendering methods based on an approximation of the projection geometry or methods computing, for instance, parallel projections, which are magnified to
account for the distance between X-ray source and image volume, are not suitable for 2D/3D registration. Furthermore, the gray-values of the DRR must be calculated accurately, because they are used to evaluate the similarity measure and influence the registration accuracy. This prohibits the application of acceleration methods such as early ray termination which compute the gray-values only approximately.

As mentioned in the introduction, only part of the CT image is used to generate DRRs, and computation of DRRs can be restricted to the minimum enclosing box which contains the anatomical structure used as feature for registration. The minimum enclosing box is, however, only a rough method to take the segmentation into account. If a vertebra is used for registration, about 75% of the volume in the minimum enclosing box does not belong to the vertebra and can be neglected. The computation of DRRs can become particularly fast, if this property is efficiently taken into account during volume rendering.

As a reference algorithm for the generation of DRRs, a simple and fast ray-casting method has been used, because this method is suitable for perspective projections and it leads to accurate gray-values if a sufficiently large sampling rate along the rays is used. Of course, computation of the DRRs has been restricted to the minimum enclosing box surrounding the anatomical structure used as feature for registration. This method, however, neglects that a large part of the minimum enclosing box does not contribute to the DRR, because suitable methods for skipping empty parts such as ray-acceleration by distance encoding did not lead to a speed up.

A volume rendering method based on the shear-warp factorization of the viewing transformation was found to be especially suitable for 2D/3D registration. This method allows to construct perspective projections, to accurately compute the gray-values, and also to skip empty regions in the minimum enclosing box efficiently. It generates a DRR in two steps. In a first step, a 2D intermediate image with coordinate axes parallel to the coordinate axes of the CT image is computed (see Fig. 2a). The gray-values of the intermediate image are defined by the integral of the CT gray-values along the ray starting at the X-ray source and going through a pixel of the intermediate image. The intermediate image is, however, not obtained by computing the corresponding integral for one pixel after the other as it is done in the case of ray-casting. Instead, the CT image is considered to be a stack of slices which are parallel to the intermediate image plane, and one slice after the other is added to the intermediate image. This is done using a run-length encoded data set of the anatomical structure used as feature for registration, and empty regions not contributing to the intermediate image can thus efficiently be skipped (see Fig. 2b). In a second step, the final DRR is obtained from the intermediate image by interpolation.

![Diagram](image)

**Figure 2.** Sketch illustrating the principle of the rendering method based on the shear-warp factorization. Firstly, an intermediate image with coordinate axes parallel to the coordinate axes of the CT image is calculated (a). A run-length encoding allows to consider only those parts of the CT image during rendering which contribute to the intermediate image (b). Secondly, the DRR is computed from the intermediate image by interpolation.
The plane of the intermediate image is dependent on the viewing direction (see Fig. 2a), which is defined by the vector from the X-ray source to the center of the minimum enclosing box of the CT image. The coordinate axis, for which the dot product of the viewing direction and the vector describing the coordinate axis has its largest absolute value, is perpendicular to the intermediate image plane whereas the other two coordinate axes define this plane. To perform the computation of the intermediate image and the addition of the CT slices to the intermediate image as fast as possible, nearest neighbor interpolation is used. A sufficient quality of the resulting intermediate image for registration purposes is achieved by using a data set resampled to a higher resolution instead of the CT image itself. This resampling can be done in a pre-processing step after segmenting the anatomical structure used as feature for registration. Run-length encoding of the resampled data set is also performed within this pre-processing step. Because the intermediate image plane can have three different orientations with respect to the coordinate system of the CT image, three run-length encoded data sets are generated to enable a fast computation of DRRs for all viewing directions. The entire pre-processing of the CT image can take a few minutes, but it can be performed before the intervention, when time is less crucial.

The gray-values of the DRR are obtained from the intermediate image using bi-linear interpolation. The size of a pixel in the intermediate image is determined according to

\[ c = \frac{a}{f} \]  

where \( a \times a \) denotes the size of a pixel in the DRR and \( c \times c \) the size of a pixel in the intermediate image. The factor \( f \) defines the resolution of the intermediate image. It controls the interpolation error within the construction of the DRR and can, therefore, also influence the registration accuracy.

3. OPTIMIZATION AND VOLUME RENDERING

To perform registration, the pattern intensity is optimized with respect to the rigid transformation describing location and orientation of the CT image and the gray-value scaling of the DRR. The gray-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 step size. For optimization of the rigid transformation, six pre-defined search directions are used. Time consuming computation of the intermediate image must be done only for three of them, because the other three search directions can be defined in such a way that the intermediate image remains unchanged. This is illustrated in Fig. 3. A rotation of the CT image around an axis passing through the X-ray source and being parallel to the projection plane results approximately in a shift parallel to that plane. A rotation around the viewing direction is approximately a rotation of the projection image in the projection plane. In either case the intermediate image remains unchanged. In contrast to this, a rotation around an axis, which passes through the center of the CT volume used for registration and which is perpendicular to the viewing direction, or a variation of the distance between X-ray source and CT volume change the perspective in such a way that the intermediate image varies.

![Figure 3. Transformations leaving the intermediate image unchanged (a) or causing changes of the intermediate image (b).](image-url)
In order to introduce the search directions, the rigid transformation describing location and orientation of the CT image with respect to the projection geometry is written as

\[
\tilde{x} = R \tilde{R} \tilde{x} - \begin{pmatrix} 0 \\ 0 \\ d \end{pmatrix} + \tilde{s}.
\]  

(3)

The points \( \tilde{x} \) and \( \tilde{R} \) refer to the coordinate system of the X-ray device and to the coordinate system of that part of the CT volume used during registration. Both coordinate systems are depicted in Fig. 2a. The point \( \tilde{s} \) defines the location of the X-ray source. The quantity \( d \) defines the distance between the X-ray source and the CT volume. The matrix \( \tilde{R} \) is a rotation matrix which describes the shifts parallel to the projection plane. It is represented by a rotation axis \( \hat{\phi} = (\varphi_x, \varphi_z, 0) \) in the \( x-y \) plane according to

\[
\tilde{R} \tilde{x} = \tilde{x} + \frac{\sin \gamma}{\gamma} \hat{\phi} \times \tilde{x} + \frac{1 - \cos \gamma}{\gamma^2} \hat{\phi} \times (\hat{\phi} \times \tilde{x}), \quad \gamma = \sqrt{\varphi_x^2 + \varphi_z^2}.
\]  

(4)

The rotation matrix \( \tilde{R} \) describes the orientation of the CT volume. To adjust the orientation, eq. (3) is replaced by

\[
\tilde{x} = R' \tilde{R} \tilde{x} - \begin{pmatrix} 0 \\ 0 \\ d \end{pmatrix} + \tilde{s}
\]  

(5)

with the matrix \( R' \) being one of the matrices

\[
R'_x = \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos \beta_x & -\sin \beta_x \\ 0 & \sin \beta_x & \cos \beta_x \end{pmatrix}, \quad R'_y = \begin{pmatrix} \cos \beta_y & 0 & \sin \beta_y \\ 0 & 1 & 0 \\ -\sin \beta_y & 0 & \cos \beta_y \end{pmatrix} \quad \text{or} \quad R'_z = \begin{pmatrix} \cos \beta_z & -\sin \beta_z & 0 \\ \sin \beta_z & \cos \beta_z & 0 \\ 0 & 0 & 1 \end{pmatrix}.
\]  

(6)

After optimizing the orientation with respect to one search direction, the rotation matrix \( \tilde{R} \) is updated (\( R' \tilde{R} \rightarrow \tilde{R} \)).

The sequence of the search directions during optimization is given by \( \varphi_x, \varphi_y, \beta_x, \beta_y, \beta_z, d \) and \( \beta_z \), and computation of a new intermediate image is only required if \( \beta_x, \beta_y \) or \( d \) are varied. Maximization for a given search direction is done using an adjustable step size \( \Delta \). The step size \( \Delta \) is increased (\( N_{\text{max}} \Delta \rightarrow \Delta \)), if \( N_{\text{max}} \) successive steps in the same direction increase the similarity measure. If neither a step forward nor a step backward increases the similarity measure, the step size \( \Delta \) is reduced as long as it is larger than a minimal value \( \Delta_{\text{min}} \) (\( \max(\Delta/N_{\text{max}}, \Delta_{\text{min}}) \rightarrow \Delta \)). For each search direction separate values of the step size \( \Delta \) and the minimal step size \( \Delta_{\text{min}} \) are used. At the beginning of the optimization the step size \( \Delta \) is initialized by a multiple of its minimal value \( \Delta_{\text{min}} \).

After registration the parameters introduced above have been converted into a rigid transformation according to

\[
\tilde{x} = R \tilde{x}' + \tilde{t}.
\]  

(7)

The point \( \tilde{x}' \) in this equation refers to a coordinate system with origin in the center of the CT image whereas the point \( \tilde{x} \) in Eq. (3) refers to a coordinate system with origin in the center of that part of the CT volume which is actually used for registration. The rotation matrix \( R \) is represented by the vector \( \phi \). This vector defines the rotation axis and its Euclidean length defines the rotation angle. The components \( t_x \) and \( t_y \) of the translation vector \( \tilde{t} \) describe the shift of the CT volume parallel to the projection plane and the component \( t_z \) its height above the projection plane.
4. EXPERIMENTS

The results included in this paper refer to two different data sets. The first data set represents images of a spine phantom. The phantom consisted of a lumbar spine and pelvis embedded in acrylic resin. Twelve fiducial markers were placed on the phantom prior to image acquisition. A CT scan (Fig. 4a) with voxel sizes 1.094mm × 1.094mm × 1.5mm and image dimensions 320 × 320 × 119 voxels was acquired. A cubic phantom of known dimensions was also scanned and used to correct for any geometric scaling errors in the CT scan. Furthermore, an anterior-posterior X-ray projection image was taken (Fig. 4b). It has a resolution of 1024 × 1024 pixels, the fluoroscopy field of view (FOV) was 15", and the source to image distance (SID) was 1125mm. A distortion correction phantom and software were used to correct for pincushion distortion. The marker in the images were used for the accurate determination of the projection geometry as well as for the computation of the “ground-truth” registration. In addition, a thorough accuracy analysis of the “ground-truth” registration has been performed. The “ground-truth” registration itself and information about its accuracy can be found in Table 1.

Figure 4. Slice of the CT image representing the spine phantom (a) and anterior-posterior X-ray projection image (b). The vertebra used for registration is marked by an arrow.

The second data set represents clinical images of a TEAM (Transfemoral Endovascular Aneurysm Management) procedure for which computer assistance has been considered within the EASI project. The CT image has a dimension of 512 × 512 × 123 voxels. The voxel-sizes are given by 0.488mm in a slice, a slice-to-slice distance of 2mm, and a slice thickness of 5mm. The X-ray projection (Fig. 1a) shows an anterior-posterior view of the spine and has been corrected for pincushion distortion. The fluoroscopy FOV was 12" and the SID 995mm.

For clinical images starting estimates of the rotation parameters and the height above the projection plane can be derived using knowledge of the patient position in the scanner and in the operating theater. Given these values a DRR is computed and two roughly corresponding points are interactively indicated on the DRR and the X-ray projection. These points are used to derive starting estimates for the shifts parallel to the projection plane. In accordance with this procedure, a set with 81 different starting estimates has been generated. Starting with the “ground-truth” registration in the case of the phantom data and the result of a single registration in the case of the clinical data, the starting estimates for the rotation parameters and the height above the projection plane have been obtained from all the combinations of \((\alpha_o - \Delta\alpha, \alpha_o + \Delta\alpha, (\alpha_o - \Delta\alpha, \alpha_o, \alpha_o + \Delta\omega), (t_x - \Delta t, t_x + \Delta t)\) with \(\Delta\omega = 5.6\degree (= 0.1\text{rad})\) and \(\Delta t = 30\text{mm}\). For each combination a DRR has been computed and the shifts parallel to the projection plane have been determined from manually indicated corresponding points on the DRR and the X-ray projection.

For each of the 81 starting estimates, a registration has been performed. In the case of the phantom images vertebra L4 was used as feature for registration (Fig. 4b). The X-ray projection was smoothed with a 3 × 3 uniform filter and resampled to a resolution of 256 × 256 pixels. In the case of the clinical data set vertebra L3 was used (Fig. 1), the X-ray projection was smoothed with a 5 × 5 uniform filter and resampled to a resolution of 128 × 128 pixels. All other parameters were the same for the phantom images and the clinical data set. The CT image was resampled to a voxel size of approximately 0.25mm × 0.25mm × 0.25mm. The parameters of the pattern intensity were given by \(r = 3\) and \(\sigma = 10\). During optimization the gray-value scaling was adjusted using an increment of \(\Delta f = 0.00002\). The minimum step size \(\Delta\alpha\) was 0.057\degree (= 0.001\text{rad}) in the case of the rotation parameters \(\beta_x, \beta_y,\) and \(\beta_z, 0.01\text{mm divided by the SID in the case of the parameters} \varphi_x \text{ and} \varphi_y\).
corresponding to the shifts parallel to the projection plane, and 0.2mm in the case of the parameter \( d \) defining the distance between the X-ray source and the CT volume. The maximum number of successive steps \( N_{\text{max}} \) was 5. At the beginning of an optimization the step size \( \Delta \) was initialized with 125 \( \Delta_{\text{max}} \) which is of the order of the deviations between the starting estimate for the rotation and translation parameters and their final result after registration.

Given all 81 registration results, the maximum of the similarity measure has been determined and a registration has been classified as successful if the similarity measure after registration was larger than 0.75 times the similarity measure’s maximum. Otherwise registration was rated as unsuccessful. For the successful registrations mean rotation parameters, mean translation parameters, the mean similarity measure, and the mean computation time have been computed as well as the corresponding standard deviations. Table 1 (phantom data) and 2 (clinical data) contain the results for various values of the parameter \( f \) which defines the resolution of the intermediate image. For comparison, registration results obtained with a ray-casting algorithm instead of the rendering method based on the shear-warp factorization are also included. The number of unsuccessful registrations was small. A maximum of 1 or 2 failures per experiment occurred which can be attributed to an awkward manual picking of the corresponding points. Finally, it should be mentioned that the times listed in Table 1 and 2 refer to the computations which must be performed intra-operatively. They do not include the time for pre-processing the CT image (i.e. segmentation of the anatomical feature, resampling, and run-length encoding), because this pre-processing can be done before the intervention. All computations have been performed on a Sun UltraSparc with 300 MHz CPU.

5. DISCUSSION

Registration accuracy is very important for clinical applications. For that reason, particular attention is paid to the influence of the rendering method on the registration accuracy within the following discussion. We do not want to speed up our algorithm at the cost of degrading the registration accuracy.

In the case of the phantom images, the registration algorithm leads to results which are almost independent of the starting estimates. This is depicted by the standard deviations of the rotation and translation parameters in Table 1. For both rendering methods as well as for the different resolutions of the intermediate image, the standard deviations of the rotations and the shifts parallel to the projection plane are smaller than 0.15° and 0.1mm. The standard deviation of the height above the projection plane is larger and can have a magnitude up to 1.1mm. This effect is due to the projection geometry, because of which this parameter cannot be determined as accurately as the others. The mean registration results agree very well with the “ground-truth” registration. The differences between the mean parameter values and the “ground-truth” registration have a size of 0.44° for the rotations, 0.33mm for the shifts parallel to the projection plane, and 5.4mm for the height above the projection plane. They are larger than the differences between the mean registration results for the shear-warp factorization based rendering method with different resolutions of the intermediate image and the ray-casting algorithm, which are up to 0.22°, 0.16mm, and 0.6mm large. This shows that neither the rendering method nor the resolution of the intermediate image affects the registration accuracy. The shear-warp factorization based rendering method with the smallest resolution tested \( f = 1 \) can thus be used making it possible to perform registration within 4s. This corresponds to an acceleration by a factor of almost seven compared to ray-casting algorithm.

| Table 1. Registration results for the spine phantom images in dependence on the parameter \( f \) which defines the resolution of the intermediate image. For comparison, results obtained with a ray-casting method and the “ground-truth” registration are included. |
|---------------------------------|-----------------|-----------------|
|                                | shear-warp factorization based rendering method | ray-casting | “ground-truth” |
|                                | \( f = 1 \)     | \( f = 2 \)     | \( f = 3 \)     |  
| rotation \( \hat{\omega} \) (in deg) | 95.41±0.08 | 95.46±0.15 | 95.60±0.06 | 95.62±0.08 | 95.45±0.09 |
|                                | -0.09±0.08  | 0.07±0.03  | 0.09±0.03  | 0.02±0.04  | 0.09±0.09  |
|                                | 1.39±0.13   | 1.40±0.06   | 1.21±0.05   | 1.18±0.07   | 1.62±0.09   |
| translation \( \hat{t} \) (in mm) | 0.95±0.10 | 0.97±0.05 | 0.82±0.04 | 0.81±0.06 | 1.14±0.15 |
|                                | -28.63±0.03 | -28.55±0.06 | -28.51±0.03 | -28.47±0.06 | -28.72±0.15 |
|                                | 176.2±0.5 | 175.7±0.6 | 176.1±0.4 | 175.6±1.1 | 170.8±1.3 |
| similarity measure            | 3828 ± 3 | 4012 ± 3 | 4056 ± 2 | 4077 ± 5 |
| computation time (in s)       | 3.9 ± 0.7 | 5.8 ± 1.3 | 8.4 ± 1.8 | 26.5 ± 6.4 |
In the case of the clinical data set, the standard deviations characterizing the variation of the registration results for different starting estimates are also small. They are up to 0.22° large for the rotations, 0.06mm for the shifts parallel to the projection plane, and 1.1mm for the height above the projection plane, which is the same order of magnitude as in the case of the phantom images. A comparison of the mean registration parameters shows that the rotation $\omega$ and the height $t_z$ for the shear-warp based rendering method with $f = 1$ differ by about 0.7° and 4mm from the other results. Apart from this, the differences between the mean registration results have a magnitude up to 0.14°, 0.26mm, and 0.6mm for the rotations, the shifts parallel to the projection plane, and the height above the projection plane, respectively. Although there is no "ground-truth" for the clinical data set available which allows for an objective rating of the results, this is a clear indication that the interpolation within the computation of the DRR from the intermediate image degrades the registration accuracy. For a higher resolution of the intermediate image ($f = 2$) this interpolation error is reduced, and there is no indication for a degradation of the registration accuracy due to the rendering method. Registration can, therefore, be carried out within 3.3s which is about four times faster compared to the results for ray-casting.

**Table 2.** Registration results for the clinical data set in dependence on the parameter $f$ which defines the resolution of the intermediate image. For comparison, results obtained with a ray-casting method are included.

<table>
<thead>
<tr>
<th></th>
<th>shear-warp factorization based rendering method</th>
<th>ray-casting</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$f = 1.0$</td>
<td>$f = 2.0$</td>
</tr>
<tr>
<td>rotation $\tilde{\omega}$ (in deg)</td>
<td>$93.22 \pm 0.22$</td>
<td>$93.36 \pm 0.07$</td>
</tr>
<tr>
<td></td>
<td>$-6.57 \pm 0.05$</td>
<td>$-6.50 \pm 0.08$</td>
</tr>
<tr>
<td></td>
<td>$7.06 \pm 0.14$</td>
<td>$6.31 \pm 0.09$</td>
</tr>
<tr>
<td>translation $\tilde{r}$ (in mm)</td>
<td>$4.32 \pm 0.03$</td>
<td>$4.12 \pm 0.05$</td>
</tr>
<tr>
<td></td>
<td>$-18.04 \pm 0.05$</td>
<td>$-17.81 \pm 0.02$</td>
</tr>
<tr>
<td></td>
<td>$170.6 \pm 1.0$</td>
<td>$174.7 \pm 0.3$</td>
</tr>
<tr>
<td>similarity measure</td>
<td>$1980 \pm 2$</td>
<td>$2238 \pm 1$</td>
</tr>
<tr>
<td>computation time (in s)</td>
<td>$2.3 \pm 0.3$</td>
<td>$3.3 \pm 0.4$</td>
</tr>
</tbody>
</table>

**6. CONCLUSIONS**

A fast voxel-based algorithm for the registration of 3D CT images with 2D X-ray projection images has been presented. The speed of the algorithm is due to three characteristic features. Firstly, the algorithm uses only part of the CT image for computing DRRs rather than the entire CT image. Secondly, a rendering method based on the shear-warp factorization has been used for generating DRRs. This rendering method is especially suitable for 2D/3D registration, because it makes it possible to efficiently skip regions of the CT image which do not contribute to the DRR. Thirdly, a specific optimization strategy has been developed by which the number of DRRs computed directly from the CT image is reduced by 50%. The other DRRs are obtained by interpolation from previously computed projection images.

The algorithm has been applied to images of a spine phantom, for which an accurate "ground-truth" registration has been derived from fiducial markers, and to a clinical data set. In the case of the phantom images, the shear-warp based rendering method leads to an acceleration by a factor of almost seven compared to a ray-casting algorithm, and registration could be carried out within 4s (Sun Ultrasparsc, 300 MHz). The rendering method does not affect the registration accuracy, and a comparison with the "ground-truth" registration results in an accuracy around 0.5° for the rotations, 0.5mm for the shifts parallel to the projection plane, and 5-6mm for the height above the projection plane. The relatively large deviation of the latter parameter can be explained by the projection geometry. The application to the clinical data set has shown that interpolation errors arising within the computation of the DRR from the intermediate image can degrade the registration accuracy. This problem can be avoided by increasing the resolution of the intermediate image. Registration of the clinical data set could be carried out within 3.3s (Sun Ultrasparsc, 300 MHz) which is four times faster than registration using ray-casting.

Computation time and registration accuracy depend, of course, on the properties of the images and on the anatomical structure used as feature for registration. Nevertheless, the results presented demonstrate that the 2D/3D registration algorithm is capable of satisfying clinical requirements concerning both, registration accuracy and computation time.
ACKNOWLEDGMENT

The authors would like to thank Prof. Dr. W. P. Th. M. Mali, Prof. Dr. B. C. Eikelboom and Dr. J. D. Blankensteijn, University Hospital Utrecht, for providing the images of the TEAM procedure. We are also grateful to the radiographers at Guy's Hospital for their time and effort in producing images of the spine phantom.

The algorithm was implemented on an experimental version of the EasyVision workstation from Philips Medical Systems and we would like to thank Easy Vision Modules Advanced Development, Philips Medical Systems, Best, for helpful discussions and their support.

The work was done within the context of the EASI project "European Applications for Surgical Interventions", supported by the European Commission under contract HC1012 in their "4th Framework Telematics Applications for Health" RTD programme. The partners in the EASI consortium are Philips Medical Systems Nederland B.V., Philips Research Laboratory Hamburg, the Laboratory for Medical Imaging Research of the Katholieke Universiteit Leuven, Utrecht University & University Hospital Utrecht, The National Hospital for Neurology and Neurosurgery in London, and the Computational Imaging Science Group at UMDS of Guy's and St. Thomas' Hospitals in London.

REFERENCES