Evaluation of 3D-2D Registration Methods for Registration of

3D-DSA and 2D-DSA Cerebral Images

Uroš Mitrović[[1]](#footnote-1)a, Žiga Špiclina, Boštjan Likara,b, Franjo Pernuša,b

aUniversity of Ljubljana, Faculty of Electrical Engineering, Laboratory of Imaging Technologies

bSensum Computer Vision Systems, Tehnološki park 21, 1000 Ljubljana, Slovenia

**ABSTRACT**

Recent C-arm systems used for endovascular image-guided interventions enable the acquisition of three-dimensional (3D) and dynamic two-dimensional (2D+t) images in the same interventional suite. The 3D images are used to observe the vascular morphology while the 2D+t images show the current state of the intervention. By spatial alignment of 3D and 2D+t images one can facilitate the endovascular interventions, e.g. by displaying the intra-interventional tools and contrast-agent flow in the augmented 3D+t images. To achieve the spatial alignment several 3D-2D registration methods were proposed that are concerned with finding the rigid-body parameters of the 3D image. Meanwhile, the pose of the C-arm system is usually obtained through a dedicated C-arm calibration. In practice, the calibrated C-arm pose parameters are typically valid only if the imaged object is positioned in the C-arm’s isocenter. To compensate this, the 3D-2D registration should search simultaneously for the rigid-body as well as the C-arm pose parameters. For verification, we tested three 3D-2D registration methods on real, clinical 3D and 2D+t angiographic images of twenty patients, ten of which were imaged with attached fiducial markers to obtain a “gold standard” registration. The results indicate that, compared to searching solely the rigid-body parameters, by searching simultaneously for rigid-body and the C-arm pose parameters significantly improves the accuracy and success rate of 3D-2D registration methods. Among the three tested methods the intensity-based method using mutual information was the most robust, as it successfully registered all clinical datasets, and highly accurate, as the maximal fiducial registration error was less or equal than 0.34 mm.

**Keywords:** 3D-2D image registration, cerebral angiograms, DSA, performance evaluation, gold standard.

1. **INTRODUCTION**

Current state-of-the-art workflow for the treatment of cerebrovascular pathologies involves three phases: 1) diagnosis of the cerebrovascular pathology, 2) planning of the intervention and 3) treatment by minimally invasive endovascular image guided intervention (EIGI).1 The first two phases involve examination of pre-EIGI acquired three-dimensional (3D) images, typically a computed tomography angiography (CTA) or a magnetic resonance angiography (MRA) image or a recently introduced modality called 3D digitally subtracted angiogram (3D-DSA), which is obtained by subtraction of two cone-beam computed tomography (CBCT) images, one with and without a contrast agent. In the 3D-DSA images all non-vascular structures are subtracted, thus enabling higher diagnostic power of detecting cerebrovascular pathologies.2,3 The third phase involves navigation of guide wires and catheters through the vasculature and application of treatment at the site of pathology by using as visual feedback a dynamic two-dimensional (2D+t) fluoroscopic or angiographic images. Performing the navigation presents a difficult task even for the experienced interventional radiologist due to overlapping and foreshortening of the vessels in 2D+t images and due to the lack of depth information. Even though that the vasculature segmentation of the pre-EIGI 3D image is typically displayed to the clinician for selection of working projection and also during EIGI4, he or she needs to mentally reconstruct in 3D the position of the interventional tools and the flow of contrast agent present in the dynamic 2D+t image(s). To enable easier and faster navigation and/or treatment that would also lead to reduction of total ionizing dose and to reduction of contrast medium delivered to the patient, one can augment the pre-EIGI 3D CTA, MRA or 3D-DSA image with the information about the current state of the intervention present in dynamic 2D+t image(s), thus resulting in augmented 3D+t images.

The key enabling technology for obtaining augmented 3D+t images is 3D-2D registration. The purpose of the 3D-2D registration is to position the pre-EIGI 3D image into the best possible spatial alignment with the corresponding dynamic 2D+t image(s). The 3D-2D image registration methods can be categorized as calibration-based or image-based, whereas the latter can be further categorized as either intrinsic or extrinsic.5 The calibration-based methods6,7 rely on geometry of the C-arm imaging device and a dedicated calibration procedure. The main advantage of these methods is low computational time (~ ms), but even small patient movement can invalidate the 3D-2D registration. Conversely the image-based methods can compensate for patient movement during EIGI. The extrinsic image-based methods perform 3D-2D registration by aligning special objects visible both in 3D and 2D images, i.e. fiducial markers. The intrinsic image-based methods rely solely on the image information and can be further classified as intensity-based,5-9 feature-based10-14 and gradient-based methods.16-18 The intrinsic image-based methods are the methods of choice, since they enable the highest level of automation and do not interfere with the current EIGI workflow. However, to apply these methods one has to: 1) initialize the rigid-body parameters of the pre-EIGI 3D image rather close to the correct position and 2) the pose parameters of the C-arm imaging system need to be accurately known. The pose parameters of the C-arm imaging system are usually obtained through a dedicated C-arm calibration. In practice, the calibrated C-arm pose parameters are typically valid only if the imaged object is positioned in the C-arm’s isocenter. To compensate this, the 3D-2D registration should search simultaneously for the rigid-body as well as the C-arm’s pose parameters.

In this paper, we evaluated the performances of three state-of-the-art 3D-2D registration methods in a realistic scenario, in which the initial rigid-body parameters of the pre-EIGI 3D image and the C-arm’s pose parameters are known only approximately. As recent C-arm imaging systems are capable of acquiring 3D images and dynamic 2D+t images in the interventional suite, the approximate values of these parameters are typically available. Furthermore, we hypothesize that by searching simultaneously for rigid-body and the C-arm’s pose parameters we can improve the accuracy and success rate of the 3D-2D registration methods. Similar studies have already been performed,8,21 however those studies lack a thorough and objective and evaluation, since only a limited number of 3D-2D registration methods were tested and only five or less clinical image datasets were used were used for evaluation. Here, we tested three state-of-the-art 3D-2D registration methods and performed quantitative and qualitative evaluation on twenty clinical image datasets containing different cerebrovascular pathologies.

1. **3D-2D REGISTRATION**

**2.1. Problem statement**

SPIE2013_Geometrical_Setup.tif

Figure 1.Geometrical setup of the C-arm system with world coordinate system **S**w, detector plane coordinate system **S**s, X-ray source **r**s, 3D image coordinate system **S**v and rigid-body parameters of the pre-EIGI 3D image **q**.Thesix pose parameters of the C-arm system: primary angle (PA), secondary angle (SA), source-to-object distance (SOD), source-to-detector distance (SID) and the detector's principle point (*u*0, *v*0) are also shown.

The purpose of the 3D-2D registration is to position the pre-EIGI 3D image into the best possible spatial alignment with the corresponding dynamic 2D+t image(s) (Fig. 1). In general, the optimal alignment process is comprised of two independent steps. In the first step, the *pose* parameters of the C-arm system are determined in a calibration procedure, while in the second step, the *rigid-body* parameters **q** = (*tx*, *ty*, *tz*, *ωx*, *ωy*, *ωz*) of the pre-EIGI 3D image are obtained in a registration procedure. The calibration step results in a 3x4 projective matrix **P** which relates any 3D point in some world coordinate system (**S**w) to its corresponding point on the 2D detector plane (**S**s) with respect to the X-ray source **r**s.

For a typical C-arm imaging system the projection matrix **P** can be described by six pose parameters and a pinhole camera model:

|  |  |
| --- | --- |
|  | (1) |

where SID and SOD denote to the source-to-detector and source-to-object distances, respectively, and (*u*0, *v*0) are coordinates of the principle point on the flat panel detector. and are 3x3 rotation matrices defined by the primary (PA) and secondary (SA) angles of the C-arm gantry:22

|  |  |
| --- | --- |
|  | (2) |

|  |  |
| --- | --- |
|  | (3) |

where:

|  |  |
| --- | --- |
|  | (4) |

In general, the 3D-2D registration methods assume that the pose parameters are known a priori, and therefore focus on optimizing only the rigid-body parameters **q** which define the optimal transform **T**(**q**) from the coordinate system of the pre-EIGI 3D image (**S**v) to **S**w. Such a scenario is not a realistic scenario because the imaged object is not necessarily in the isocenter of the C-arm imaging system, therefore, only approximate values of the pose parameters are known. To obtain the optimal alignment between the 3D and 2D+t images, a 3D-2D registration method needs to be extended to search simultaneously for the rigid-body as well as the C-arm’s pose parameters.

**2.2. Initialization**

As outlined in Section 1., a pre-requisite for successful alignment between the pre-EIGI 3D image and the dynamic 2D+t image(s) is to position the 3D image close enough to the correct position.23 Here, close enough is determined by the capture range of a particular 3D-2D registration method. This is in contrast to typical registration evaluation scenarios, in which where initial guess is not straightforward to obtain, in neuroangiography 3D and 2D images are acquired by the same imaging device, and therefore a coarse initialization can be deduced by the geometry of the C-arm system. As recent C-arm imaging systems used in neuroangiography are capable of acquiring 3D images and dynamic 2D+t images in the interventional suite, a *coarse* initialization of the position of the 3D can be deduced by the geometry of the C-arm system, i.e. by assuming that **S**w is in the center of the pre-EIGI 3D image and that approximate values of the C-arm’s pose parameters can be extracted from the DICOM header of the acquired 2D+t images. However, such an initialization may still lead to large errors in in-plane translations, typically larger than the capture range of the evaluated 3D-2D registration method (Fig. 2 *left*).8 The coarse initialization can be refined by aligning the intensity centroids of the 3D and 2D images , respectively (Fig. 2 *middle*), followed by an exhaustive search of the in-plane translations so as to find the minimal/maximal values of the similarity measure (SM) of the 3D-2D registration method (Fig. 2 *right*). The search range of in-plane translations around the 2D intensity centroid was set to 20% of the maximal size of the 2D image. Compared to coarse initialization, such initialization will be referred to as *fine* initialization.

**2.3 State-of-the-art 3D-2D registration methods**

Three state-of-the-art 3D-2D registration methods were evaluated, two belonging to the class of the intensity-based9,24,25 methods, while one belonging to the class of the gradient-based methods.18 The intensity-based methods are based on 2D-2D matching of real fluoroscopic and simulated images, obtained by casting virtual rays through the 3D image. These method differ with respect to the method of rendering simulated 2D projection images and with respect to the SM, which measures the degree of matching between real and simulated images. In this paper, the simulated 2D projection images were rendered using maximum intensity projection (MIP),8 while either mutual information9,24 (MI) and gradient correlation9,25 (GC) were used as the SM; these two methods will be referred to as MIP-MI and MIP-GC, respectively. The third method is the 3D-2D back-projection gradient-based method (BGB) proposed by Tomaževič et al.18, which is based on matching the 3D gradient vectors representing 3D surface normals and 2D-to-3D back-projected gradient vectors. The volume gradients were computed using 3D Canny edge detector, while the 2D gradients were calculated using a simple central-differences kernel.

SPIE2013_Initialization.tif

Figure 2. Coarse initialization (left), refined by alignment of 3D and 2D intensity centroids (middle) and exhaustive search of the in-plane translations (right).

1. **DATABASE AND EVALUATION CRITERIA**

**3.1 Image database**

For evaluation of the state-of-the-art 3D-2D registration methods, 3D-DSA and 2D-DSA images were obtained from twenty patients undergoing cerebral aneurysm or arteriovenous malformation treatment. Patients had pathologies of various degrees and were in deep anesthesia throughout the image acquisition. The image database was acquired in such way to capture a whole range of different scenarios which may occur during a typical cerebral-EIGI (Fig. 3). At the start of the cerebral-EIGI, two rotational scans of the patient's head were performed resulting in a 512 x 512 x 391 3D-DSA image with voxel size of 0.46 x 0.46 x 0.46 mm. Next, the 2D-DSA images were acquired at two different gantry orientations of the C-arm imaging system, namely the lateral (LAT) and anterior-posterior (AP) position. The LAT and AP 2D-DSA images included in the image database were acquired in different phases of the EIGI, and differ in sizes, gantry orientations and magnification factors. All images were acquired by Siemens Axiom Artis dBA biplane flat detector angiography system. As the flat panel technology was used all 2D images were free of geometric distortions.

SPIE2013_Image_Database.tif

Figure 3. The characteristic 2D-DSA images of the image database. Patient 8 (left) underwent AVM treatment, while patients 12, 13 and 17 (from middle-left to right) underwent aneurysm treatment.

**3.2. Evaluation protocol**

The MIP-MI, MIP-GC and BGB methods were implemented and tested in two different settings. In the first setting, only rigid-body parameters of the pre-EIGI 3D image were optimized, while in the second both the rigid-body and the C-arm’s pose parameters were simultaneously optimized. In both settings, the methods were registering 3D-DSA image to a single 2D-DSA image starting with the rigid-body and the pose parameters as obtained in the initialization, explained in Section 2.2. The methods were run on all twenty clinical image datasets; ten datasets had fiducial-marker based “gold standard” registration that enabled quantitative evaluation, while qualitative evaluation was performed for the registrations on the other ten datasets by visual inspection and assessment.

**3.2.1. Quantitative evaluation**

In order to quantitatively evaluate the performances of the state-of-the-art 3D-2D registration methods an accurate "gold standard" needs to be obtained. During the acquisition of the 3D and 2D images, first ten patients wore an elastic headband with *N*M = 12 attached fiducial markers (steel ball bearings of 2mm diameter). The 2D images were acquired just before the intervention, and then the headband was removed from the patient’s head. The centers of fiducial markers were extracted from the 3D and 2D images using intensity centroid method.26,27 After the centers of fiducial markers were extracted, the markers were erased from the 3D and 2D images and the corresponding intensities replaced by spline interpolation.28The registration accuracy of the 3D-2D registration methods was expressed in terms of fiducial projection error (FPE):

|  |  |
| --- | --- |
|  | (5) |

where and are 2D and 3D centers of fiducial markers, respectively, **P** is the projection matrix defined by the C-arm’s pose parameters and **T**reg is transform matrix defined by the rigid-body parameters **q**reg. Note that, the matrix **T**reg was computed by the 3D-2D registration method, while the matrix **P** was computed either by initialization or by registration method, depending on the test setting described above. 3D-2D registration was executed between 3D-DSA and LAT 2D-DSA and between 3D-DSA and AP 2D-DSA images. Once the 3D-DSA image was aligned to both LAT and AP 2D-DSA images, the reconstructed 3D centers were computed using epipolar geometry.29 Based on and the fiducial reconstruction error (FRE) was computed to measure the 3D-2D registration accuracy as:

|  |  |
| --- | --- |
|  | (6) |

where Tsvd is the rigid-body transform obtained by singular value decomposition.30 The FPEs and FREs obtained by the state-of-the-art 3D-2D registration methods were also compared to the "gold standard" registrations, which are based on the alignment of the fiducial markers. The "gold standard" registrations were obtained by minimizing FRE, where C-arm’s pose parameters of the LAT and AP 2D images and rigid-body parameters of the 3D-DSA image were simultaneously optimized (18 parameters in total). The FPE and FRE values for the "gold standard" registrations ranged from 0.038 and 0.09 mm and 0.038 and 0.06 mm, respectively.

**3.2.2 Qualitative evaluation**

In the previous subsection we described a quantitative and objective evaluation of 3D-2D registration that is only possible if fiducial markers are attached to the patient and visible on both pre-EIGI 3D and during-EIGI 2D images. In some cases the fiducial markers may not be visible due to a high magnification factor, while the use of fiducial markers might also obscure the pathology and hamper the intervention. Moreover, the use of dedicated objects for deducing the 3D-to-2D alignment is not desirable during EIGI. In this case, the performances of the state-of-the-art 3D-2D registration methods can be assessed qualitatively by visual inspection and assessment of the 3D-to-2D alignment quality. In ten datasets, the human observer visually inspected the alignment of the 2D-DSA and the overlaid MIP of the registered 3D-DSA image, and marked the registration as "Good", "Fair" or "Bad" (Fig, 4). These ten datasets varied more in appearance than the first ten datasets, since they contained the 2D-DSA images acquired during different phases of the intervention, and used diverse magnification factors so that only small parts of vascular structures were visible in 3D and 2D images, and contained interventional tools.

SPIE2013_Human_Assessment.tif

Figure 4. Examples of "Good", "Fair" and "Bad" registration (left, middle and right, respectively) assessed by the human observer. Dark structures represents the vessels in 2D-DSA image, while the white structures represent the MIP of the 3D-DSA image.

1. **RESULTS**

**4.1 Quantitative evaluation**

Tabs. 1 and 2 report the FPE and FRE values of the evaluated 3D-2D registration methods as obtained by optimizing solely the rigid-body parameters or simultaneously the rigid-body and the C-arm’s pose parameters (All), respectively on datasets with “gold standard” based on fiducial markers. The maximal absolute differences of the rigid-body parameters **q** from the "gold standard" position to initial coarse and fine positions, and final MIP-GC, MIP-MI and BGB positions for datasets with fiducial markers are given in Tab. 3.

Table 1. FPE values in mm for the MIP-GC, MIP-MI and BGB methods applied to 3D-DSA to single 2D-DSA registration of the ten image datasets with fiducial markers. FPE for LAT and AP 2D-DSA registrations are shown in *top* and *bottom* rows for each method, respectively. The FPE values were obtained either by optimizing solely the rigid-body parameters or simultaneously the rigid-body and C-arm’s pose parameters (All).

|  |  |  |  |  |  |  |  |  |  |  |  |  |  |
| --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- |
| Test |  | Method |  | Dataset | | | | | | | | | |
|  |  | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 |
| Rigid-body |  | MIP-GC |  | 0.83 | 7.85 | 0.68 | 0.92 | 1.25 | 0.79 | 1.15 | 0.69 | 0.93 | 0.70 |
|  |  | 2.32 | 0.19 | 0.20 | 0.18 | 0.33 | 9.28 | 1.26 | 0.22 | 0.18 | 0.18 |
|  | MIP-MI |  | 0.80 | 0.60 | 0.65 | 0.96 | 0.99 | 0.71 | 1.38 | 0.74 | 0.93 | 0.63 |
|  |  | 0.65 | 0.32 | 0.23 | 0.42 | 0.51 | 0.52 | 1.62 | 0.44 | 0.31 | 0.16 |
|  | BGB |  | 1.22 | 2.21 | 0.76 | 1.49 | 2.27 | 1.65 | 1.72 | 1.26 | 1.76 | 0.98 |
|  |  | 1.26 | 0.47 | 0.72 | 1.01 | 1.42 | 0.74 | 2.32 | 0.67 | 0.76 | 0.39 |
| All |  | MIP-GC |  | 0.29 | 0.39 | 0.22 | 0.21 | 0.30 | 0.53 | 0.62 | 0.24 | 0.41 | 0.31 |
|  |  | 0.32 | 0.20 | 0.16 | 0.14 | 0.28 | 0.18 | 0.69 | 0.23 | 0.34 | 0.17 |
|  | MIP-MI |  | 0.34 | 0.33 | 0.19 | 0.24 | 0.63 | 0.45 | 0.67 | 0.33 | 0.56 | 0.32 |
|  |  | 0.84 | 0.25 | 0.18 | 0.38 | 0.52 | 0.23 | 0.86 | 0.48 | 0.36 | 0.20 |
|  | BGB |  | 1.01 | 2.38 | 0.39 | 3.44 | 1.73 | 4.62 | 2.90 | 0.94 | 1.74 | 3.03 |
|  |  | 1.23 | 0.49 | 0.41 | 1.03 | 1.42 | 0.73 | 3.09 | 0.85 | 0.80 | 0.47 |

Table 2. FRE values in mm for the MIP-GC, MIP-MI and BGB methods applied to 3D-DSA to LAT and AP 2D-DSA registration of the ten image datasets with fiducial markers. The FRE values were obtained either by optimizing solely the rigid-body parameters or simultaneously the rigid-body and C-arm’s pose parameters (All).

|  |  |  |  |  |  |  |  |  |  |  |  |  |  |
| --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- |
| Test |  | Method |  | Dataset | | | | | | | | | |
|  |  | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 |
| Rigid-body |  | MIP-GC |  | 0.52 | 3.67 | 0.21 | 0.28 | 0.33 | 0.71 | 0.31 | 0.21 | 0.20 | 0.25 |
|  | MIP-MI |  | 0.52 | 0.30 | 0.15 | 0.37 | 0.51 | 0.13 | 0.75 | 0.20 | 0.27 | 0.19 |
|  | BGB |  | 0.20 | 0.31 | 0.30 | 0.58 | 1.53 | 0.85 | 0.56 | 0.23 | 0.50 | 0.42 |
| All |  | MIP-GC |  | 0.10 | 0.11 | 0.06 | 0.08 | 0.12 | 0.26 | 0.24 | 0.10 | 0.12 | 0.07 |
|  | MIP-MI |  | 0.34 | 0.07 | 0.07 | 0.14 | 0.29 | 0.25 | 0.12 | 0.33 | 0.18 | 0.09 |
|  | BGB |  | 0.65 | 0.74 | 0.14 | 1.30 | 0.79 | 1.67 | 1.74 | 0.60 | 0.50 | 1.25 |

FPE values of the MIP-GC method obtained by optimizing solely rigid-body parameters were between 0.18 and 9.28 mm, while FPE values obtained by optimizing all parameters (rigid-body + C-arm pose) were between 0.14 and 0.69 mm. The MIP-MI method obtained FPE values between 0.16 and 1.62 mm for optimizing rigid-body and FPE values between 0.18 and 0.84 for optimizing all of the parameters. The BGB method was consistently less accurate than the other two methods with FPE values between 0.39 and 2.32 mm for optimizing rigid-body parameters and with FPE values between 0.41 and 3.44 mm for optimizing all parameters.

FRE values for the MIP-GC method were between 0.20 and 3.67 mm by optimizing solely rigid-body parameters, and between 0.06 and 0.26 mm by optimizing all parameters. The MIP-MI method obtained FRE values between 0.13 and 0.75 mm by optimizing rigid-body parameters and FRE values between 0.07 and 0.34 by optimizing all parameters. Similarly to FPE-based analysis the FRE values for the BGB method were consistently less accurate, i.e. between 0.2 and 1.53 mm by optimizing rigid-body parameters and between 0.5 and 1.74 mm by optimizing all parameters.

Table 3. Maximal absolute differences of the rigid-body parameters q = (*tx*, *ty*, *tz*, *ωx*, *ωy*, *ωz*) from the "gold standard" position to initial coarse and fine, and final MIP-GC, MIP-MI and BGB positions for datasets with fiducial markers. Note that *tx*, *ty* and *ωz* denote the in-plane parameters, while *tz*, *ωx* and *ωy* denote the out-of-plane parameters.

|  |  |  |  |  |  |  |  |
| --- | --- | --- | --- | --- | --- | --- | --- |
| Method |  | tx [mm] | ty [mm] | tz [mm] | wx [o] | wy[o] | wz [o] |
| initial coarse |  | 8.61 | 24.94 | 58.36 | 0.83 | 0.56 | 0.57 |
| initial fine |  | 1.94 | 1.42 | 58.36 | 0.83 | 0.56 | 0.57 |
| MIP-GC |  | 3.64 | 10.42 | 48.23 | 5.18 | 1.20 | 1.26 |
| MIP-MI |  | 1.27 | 1.62 | 4.27 | 0.78 | 0.58 | 0.58 |
| BGB |  | 3.33 | 2.09 | 10.04 | 12.04 | 1.26 | 10.98 |

**4.2 Qualitative evaluation**

The qualitative evaluation of the 3D-2D registration methods performed by visual assessment is given in Tab. 4, both for optimizing solely rigid-body parameters or simultaneously the rigid-body and the C-arm’s pose parameters (All). The maximal absolute differences of rigid-body parameters **q** with respect to the final position obtained by the MIP-MI method to initial coarse and fine positions, and final MIP-GC and BGB positions are given in Tab. 5.

Table 4. Evaluation of the 3D-2D registrations obtained by the MIP-GC, MIP-MI and BGB methods for 3D-DSA to LAT and AP 2D-DSA registrations of the ten image datasets used for qualitative evaluation. The values "G", "F" and "B" in the table correspond to visual assessment of registration as "Good", "Fair" and "Bad".

|  |  |  |  |  |  |  |  |  |  |  |  |  |  |
| --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- |
| Test |  | Method |  | Dataset | | | | | | | | | |
|  |  | 11 | 12 | 13 | 14 | 15 | 16 | 17 | 18 | 19 | 20 |
| Rigid-body |  | MIP-GC |  | B | G | G | G | G | B | G | G | F | G |
|  |  | B | G | G | G | G | B | G | G | G | G |
|  | MIP-MI |  | F | G | G | G | G | B | G | G | G | G |
|  |  | G | G | G | G | G | G | F | G | G | G |
|  | BGB |  | F | G | G | G | F | B | F | G | F | G |
|  |  | B | G | F | G | F | B | F | G | G | G |
| All |  | MIP-GC |  | G | G | B | G | G | B | G | G | G | G |
|  |  | G | G | G | G | G | F | B | G | G | G |
|  | MIP-MI |  | G | G | G | G | G | G | G | G | G | G |
|  |  | G | G | G | G | G | G | G | G | G | G |
|  | BGB |  | G | G | B | G | G | B | F | G | G | G |
|  |  | G | G | G | G | B | B | B | G | G | G |

Table 5. Maximal absolute differences of rigid-body parameters q = (*tx*, *ty*, *tz*, *ωx*, *ωy*, *ωz*) from the final position obtained by the MIP-MI method to initial coarse and fine, and final MIP-GC and BGB positions for datasets used for qualitative evaluation. Note that *tx*, *ty* and *ωz* denote the in-plane parameters, while *tz*, *ωx* and *ωy* denote the out-of-plane parameters.

|  |  |  |  |  |  |  |  |
| --- | --- | --- | --- | --- | --- | --- | --- |
| Method |  | tx [mm] | ty [mm] | tz [mm] | wx [o] | wy[o] | wz [o] |
| initial coarse |  | 41.03 | 36.87 | 48.08 | 4.81 | 7.63 | 7.08 |
| initial fine |  | 3.67 | 3.70 | 48.08 | 4.81 | 7.63 | 7.08 |
| MIP-GC |  | 16.96 | 4.74 | 43.53 | 1.44 | 7.69 | 3.59 |
| MIP-MI |  | 0 | 0 | 0 | 0 | 0 | 0 |
| BGB |  | 12.57 | 10.57 | 65.66 | 3.17 | 3.77 | 3.44 |

When optimizing solely rigid-body parameters the MIP-GC method achieved 75% "Good", 5% "Fair" and 20% "Bad" registrations. The MIP-MI method seemed to be more successful with 85%, 10% and 5% registrations marked as "Good", "Fair" and "Bad", respectively. The least successful was the BGB method for which 50%, 35% and 15% of registrations were marked as "Good", "Fair" and "Bad", respectively.

When all parameters were optimized simultaneously, the MIP-GC and the MIP-MI methods were also more successful than the BGB method. The MIP-GC method had 80%, 5% and 15% marked registrations as "Good", "Fair" and "Bad", respectively, while for the MIP-MI method all registration were labeled as "Good". The BGB method had 70% "Good", 5% "Fair" and 25% "Bad" registrations.

1. **DISCUSSION & CONCLUSION**

In the last decade, a number of 3D-2D registration methods were proposed particularly for the application in cerebral-EIGI. In general, these methods are concerned with finding six rigid-body parameters of the pre-EIGI 3D image, while assuming that the pose of the C-arm imaging system is known and follows from the C-arm calibration. Evaluation of the 3D-2D registration methods usually requires some reference or "gold standard" registrations,31,32 about which a set of random rigid-body displacements of the pre-EIGI 3D image are generated. Then the 3D-2D registration is run repeatedly, each time using as initialization one random rigid-body displacement from the set. Based on the final displacement from the "gold standard" registrations, the registration accuracy, capture range and success rate are computed. This scenario, however, does not mimic the real initial conditions seen in practice for two reasons. First, the C-arm’s pose parameters are usually not accurately known, because the imaged object may not necessarily be in the isocenter of the C-arm, and second, the displacement of the 3D image from the "gold standard" is not known a priori and can far exceed the capture range of the 3D-2D registration method. Initial rigid-body parameters and the C-arm’s pose parameters obtained from the DICOM header may lead to initial displacements from the actual "gold standard" position up to 40 mm for in-plane translations and up to 7o for in-plane rotations (Tabs. 3 and 5). Such a large initial displacement is beyond the capture range of all of the tested state-of-the-art 3D-2D registration methods. Therefore, we proposed an approach to refine the initial rigid-body and the C-arm’s pose parameters that reduced the error of the in-plane translations below 4 mm, which should be sufficient in most cases to obtain a successful 3D-2D registration. Using the initial coarse and fine positions, we analyzed the performances of three state-of-the-art 3D-2D registration methods under real conditions, using twenty real clinical image datasets. Furthermore, we demonstrated that besides the rigid-body parameters of the pre-EIGI 3D image the C-arm’s pose parameters also need to be optimized to achieve high registration success rates.

Among the twenty real clinical image datasets, the first ten datasets contained images with fiducial markers acquired at the start of the cerebral-EIGI, which were then used to measure the actual accuracy of the registration methods, i.e. quantitative analysis. The last ten datasets contained the images of different cerebral-EIGI scenarios, which were used to measure the robustness of the registration methods to various registration-hampering phenomena. In these cases, the final alignment of the 3D-DSA and corresponding 2D-DSA images was visually assessed, i.e. qualitative analysis. The image database of twenty patients undergoing cerebral aneurysm or arteriovenous malformation treatment, containing pathologies of various degrees and the 2D+t images acquired in different phases of the EIGI that differ in sizes, gantry orientations and magnification factors, while some even contain interventional tools, seem to be representative of the real clinical situation and enables a thorough evaluation of the performances of the 3D-2D registration methods, from which

reliable conclusions can be drawn.

Three state-of-the-art intrinsic image-based methods, namely the intensity-based MIP-MI and MIP-GC and the gradient-based BGB were tested in two different settings. In the first setting all the methods were run optimizing solely rigid-body parameters of the pre-EIGI 3D image, while in the second setting the rigid-body and the C-arm’s pose parameters were optimized simultaneously. Based on obtained results, the two intensity-based methods performed best both in terms of accuracy and success rate when all the parameters were optimized simultaneously. Regardless of the setting the BGB method was consistently less accurate and generally less successful; for 3D-DSA to LAT 2D-DSA registration in patient 2 and 6 it was more successful than the MIP-GC (cf. Tab. 2). Overall, the MIP-MI method seems to significantly outperform the other methods in terms of success rate as it successfully registered all 3D-DSA to the corresponding 2D-DSA images. Meanwhile the MIP-GC and the BGB methods had 80 and 70% of successful registrations, respectively. On the other hand, when the initialization was within the capture range of the MIP-GC method, it turned out to be the most accurate with maximal FPE and FRE of 0.69 and 0.26 mm. The MIP-MI were somewhat less accurate with maximal FPE and FRE of 0.84 and 0.34 mm, while the BGB method was the least accurate with maximal FPE and FRE of 2.32 and 1.53 mm. Most importantly, the results indicate that compared to searching solely the rigid-body parameters, by searching simultaneously for rigid-body and the C-arm pose parameters significantly improves both the accuracy and success rate of 3D-2D registration methods.

Based on the presented results, we conclude that by using the two intensity-based methods a highly accurate 3D-DSA and 2D-DSA registrations can be obtained, surpassing the requirements of accuracy for the cerebral-EIGI. However, a pre-requisite for translating either of these two methods into the interventional suite requires real-time execution, which could be achieved by implementation on graphics processing unit (GPU),33,34 and, more importantly, an indication of registration failure. Our future research efforts are therefore directed towards the development of fast method implementations and of methods for automatic detection of registration failure.

**ACKNOWLEDGMENTS**

This research was supported by the Ministry of Higher Education, Science and Technology, Republic of Slovenia, under grants L2-2023, L2-9758, J2-0716, J2-2246, and P2-0232.

**REFERENCES**

[1] Rudin, S., Bednarek, D.R., and Hoffmann, K.R., “Endovascular image-guided interventions (EIGIs),” Medical Physics 35(1), 301 (2008).

[2] Van Rooij, W.J., Sprengers, M.E., De Gast, A.N., Peluso, J.P.., and Sluzewski, M., “3D Rotational Angiography: The New Gold Standard in the Detection of Additional Intracranial Aneurysms,” American Journal of Neuroradiology 29(5), 976–979 (2008).

[3] Zhang, X.-Q., Shirato, H., Aoyama, H., Ushikoshi, S., Nishioka, T., Zhang, D.-Z., and Miyasaka, K., “Clinical significance of 3D reconstruction of arteriovenous malformation using digital subtraction angiography and its modification with CT information in stereotactic radiosurgery,” International Journal of Radiation Oncology Biology Physics 57(5), 1392–1399 (2003).

[4] Strobel, N., Meissner, O., Boese, J., Brunner, T., Heigl, B., Hoheisel, M., Lauritsch, G., Nagel, M., Pfister, M., et al., [Imaging with Flat-Detector C-arm Systems] , in Multislice CT, M. F. Reiser, C. R. Becker, K. Nikolaou, and G. Glazer, Eds., Springer Berlin Heidelberg, 33–51 (2009).

[5] Markelj, P., Tomaževič, D., Likar, B., and Pernuš, F., “A review of 3D/2D registration methods for image-guided interventions,” Medical Image Analysis 16(3), 642–661 (2012).

[6] Ruijters, D., Homan, R., Mielekamp, P., Van de Haar, P., and Babic, D., “Validation of 3D multimodality roadmapping in interventional neuroradiology,” Physics in Medicine and Biology 56(16), 5335–5354 (2011).

[7] Söderman, M., Babic, D., Homan, R., and Andersson, T., “3D roadmap in neuroangiography: technique and clinical interest,” Neuroradiology 47(10), 735–740 (2005).

[8] Kerrien, E., Berger, M.-O., Maurincomme, E., Launay, L., Vaillant, R., and Picard, L., [Fully Automatic 3D/2D Subtracted Angiography Registration] , in Medical Image Computing and Computer-Assisted Intervention - MICCAI 1999, Springer, London, UK, 664–671 (1999).

[9] Hipwell, J.H., Penney, G.P., McLaughlin, R.A., Rhode, K., Summers, P., Cox, T.C., Byrne, J.V., Noble, J.A., and Hawkes, D.J., “Intensity-based 2-D - 3-D registration of cerebral angiograms,” IEEE Transactions on Medical Imaging 22(11), 1417–1426 (2003).

[10] Byrne, J.V., Colominas, C., Hipwell, J., Cox, T., Noble, J.A., Penney, G.P., and Hawkes, D.J., “Assessment of a Technique for 2D–3D Registration of Cerebral Intra-Arterial Angiography,” British Journal of Radiology 77(914), 123–128 (2004).

[11] McLaughlin, R.A., Hipwell, J., Hawkes, D.J., Noble, J.A., Byrne, J.V., and Cox, T.C., “A comparison of a similarity-based and a feature-based 2-D-3-D registration method for neurointerventional use,” IEEE Transactions on Medical Imaging 24(8), 1058–1066 (2005).

[12] Chung, A.C.S., Wells III, W.M., Norbash, A., and L. Grimson, W.E., [Multi-modal Image Registration by Minimizing Kullback-Leibler Distance] , in Medical Image Computing and Computer-Assisted Intervention - MICCAI 2002, Springer, Berlin, Heidelberg, 525–532 (2002).

[13] Feldmar, J., Ayache, N., and Betting, F., “3D–2D Projective Registration of Free-Form Curves and Surfaces,” Computer Vision and Image Understanding 65(3), 403–424 (1997).

[14] Liu, A., Bullitt, E., and Pizer, S.M., [3D/2D Registration Via Skeletal Near Projective Invariance in Tubular Objects] , in Medical Image Computing and Computer-Assisted Interventation - MICCAI 1998, Springer, Berlin, Heidelberg, 952–963 (1998).

[15] Kita, Y., Wilson, D.L., and Noble, A., [Real-time registration of 3D cerebral vessels to X-ray angiograms] , in Medical Image Computing and Computer-Assisted Interventation - MICCAI 1998 1496, Springer, Berlin, Heidelberg, 1125–1133 (1998).

[16] Sundar, H., Khamene, A., Xu, C., Sauer, F., and Davatzikos, C., “A novel 2D-3D registration algorithm for aligning fluoro images with 3D pre-op CT/MR images,” in Proceedings of SPIE 6141, 61412K–61412K–7 (2006).

[17] Florin, C., Williams, J., Khamene, A., and Paragios, N., [Registration of 3D angiographic and X-ray images using Sequential Monte Carlo sampling] , in Computer Vision for Biomedical Image Applications, First Int’l Workshop, CVBIA ’05, Springer, Berlin, Heidelberg, 427–436 (2005).

[18] Tomaževič, D., Likar, B., Slivnik, T., and Pernuš, F., “3-D/2-D registration of CT and MR to X-ray images,” IEEE Transactions on Medical Imaging 22(11), 1407–1416 (2003).

[19] Markelj, P., Tomaževič, D., Pernuš, F., and Likar, B., “Robust Gradient-Based 3-D/2-D Registration of CT and MR to X-Ray Images,” IEEE Transactions on Medical Imaging 27(12), 1704–1714 (2008).

[20] Livyatan, H., Yaniv, Z., and Joskowicz, L., “Gradient-based 2-D/3-D rigid registration of fluoroscopic X-ray to CT,” IEEE Transactions on Medical Imaging 22(11), 1395 –1406 (2003).

[21] Hentschke, C.M., and Tönnies, K.D., “Automatic 2D/3D-Registration of Cerebral DSA Data Sets,” in Bildverarbeitung für die Medizin 2010 - Algorithmen - Systeme - Anwendungen 574, 162–166 (2010).

[22] Shechter, G., Shechter, B., Resar, J.R., and Beyar, R., “Prospective motion correction of X-ray images for coronary interventions,” IEEE transactions on medical imaging 24(4), 441–450 (2005).

[23] Van der Bom, M.J., Bartels, L.W., Gounis, M.J., Homan, R., Timmer, J., Viergever, M.A., and Pluim, J.P.W., “Robust initialization of 2D-3D image registration using the projection-slice theorem and phase correlation,” Medical physics 37(4), 1884–1892 (2010).

[24] Pluim, J.P.W., Maintz, J.B.A., and Viergever, M.A., “Mutual-information-based registration of medical images: a survey,” IEEE Transactions on Medical Imaging 22(8), 986 –1004 (2003).

[25] Van der Bom, M.J., Klein, S., Staring, M., Homan, R., Bartels, L.W., and Pluim, J.P.W., “Evaluation of optimization methods for intensity-based 2D-3D registration in x-ray guided interventions,” in Proceedings of SPIE 7962, 796223–796223–15 (2011).

[26] Bose, C.B., and Amir, I., “Design of Fiducials for Accurate Registration Using Machine Vision,” IEEE Transactions on Pattern Analysis and Machine Intelligence 12(12), 1196–1200 (1990).

[27] Chiorboli, G., and Vecchi, G.P., “Comments on ‘Design of Fiducials for Accurate Registration Using Machine Vision’,” IEEE Transactions on Pattern Analysis and Machine Intelligence 15(12), 1330–1332 (1993).

[28] Demirci, S., “New Approaches to Computer Assistance for Endovascular Abdominal Aortic Repairs,” Technische Universität München (2011).

[29] Dumay, A.M., Reiber, J.C., and Gerbrands, J.J., “Determination of optimal angiographic viewing angles: basic principles and evaluation study,” IEEE Transactions on Medical Imaging 13(1), 13–24 (1994).

[30] Fitzpatrick, J., West, J., and Maurer, C., “Predicting error in rigid-body point-based registration,” IEEE Transactions on Medical Imaging 17(5), 694–702 (1998).

[31] Van de Kraats, E., Penney, G., Tomaževič, D., Van Walsum, T., and Niessen, W., “Standardized evaluation methodology for 2-D-3-D registration,” IEEE Transactions on Medical Imaging 24(9), 1177–1189 (2005).

[32] Markelj, P., Likar, B., and Pernuš, F., “Standardized evaluation methodology for 3D/2D registration based on the Visible Human data set,” Medical Physics 37(9), 4643–4647 (2010).

[33] Gendrin, C., Furtado, H., Weber, C., Bloch, C., Figl, M., Pawiro, S.A., Bergmann, H., Stock, M., Fichtinger, G., et al., “Monitoring tumor motion by real time 2D/3D registration during radiotherapy,” Radiotherapy and Oncology: Journal of the European Society for Therapeutic Radiology and Oncology 102(2), 274–280 (2012).

[34] Furtado, H., Gendrin, C., Bloch, C., Spoerk, J., Pawiro, S.A., Weber, C., Figl, M., Stock, M., Georg, D., et al., “Real-time 2D/3D registration for tumor motion tracking during radiotherapy,” in Proceedings of SPIE 8314, 831407–831407–8 (2012).

1. E-mail: uros.mitrovic@fe.uni-lj.si; telephone: +386 (0) 1 4768 873; web: http://lit.fe.uni-lj.si/ [↑](#footnote-ref-1)