Automated 2D–3D registration of portal images and CT data using line-segment enhancement

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In prostate radiotherapy, setup errors with respect to the patient's bony anatomy can be reduced by aligning 2D megavoltage (MV) portal images acquired during treatment to a reference 3D kilovoltage (kV) CT acquired for treatment planning purposes. The purpose of this study was to evaluate a fully automated 2D–3D registration algorithm to quantify setup errors in 3D through the alignment of line-enhanced portal images and digitally reconstructed radiographs computed from the CT. The line-enhanced images were obtained by correlating the images with a filter bank of short line segments, or “sticks” at different orientations. The proposed methods were validated on (1) accurately collected gold-standard data consisting of a 3D kV cone-beam CT scan of an anthropomorphic phantom of the pelvis and 2D MV portal images in the anterior–posterior (AP) view acquired at 15 different poses and (2) a conventional 3D kV CT scan and weekly 2D MV AP portal images of a patient over 8 weeks. The mean (and standard deviation) of the absolute registration error for rotations around the right-lateral (RL), inferior–superior (IS), and posterior–anterior (PA) axes were 0.212° (0.214°), 0.055° (0.033°) and 0.041° (0.039°), respectively. The corresponding registration errors for translations along the RL, IS, and PA axes were 0.161(0.131) mm, 0.096(0.033) mm, and 0.612(0.485) mm. The mean (and standard deviation) of the total registration error was 0.778(0.543) mm. Registration on the patient images was successful in all eight cases as determined visually. The results indicate that it is feasible to automatically enhance features in MV portal images of the pelvis for use within a completely automated 2D–3D registration framework for the accurate determination of patient setup errors. They also indicate that it is feasible to estimate all six transformation parameters from a 3D CT of the pelvis and a single portal image in the AP view. © 2008 American Association of Physicists in Medicine. [DOI: [10.1118/1.2975143](https://doi.org/10.1118/1.2975143)]

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I. INTRODUCTION

In 3D conformal radiotherapy and intensity-modulated radiotherapy of the prostate, accurate and reproducible 3D positioning of the patient during every treatment fraction is crucial if the amount of radiation delivered to the tumor is to be maximized and exposure to surrounding healthy tissue minimized. The use of 3D CBCT imaging or implanted radioopaque fiducial markers with 2D kilovoltage (kV) on-board imaging prior to delivering the treatment fraction can potentially allow reduction of the margins needed for setup errors and interfractional organ motion by aligning the treatment fields directly to the target volume. In many radiotherapy clinics, however, patient positioning is still carried out with

respect to the patient's bony anatomy by aligning megavoltage (MV) portal images acquired during treatment to the 3D kV CT from which the treatment was planned. In this study, we present and evaluate a completely automated 2D–3D registration framework to align 2D MV treatment portal images to a 3D kV planning CT for the quantification of patient setup errors during prostate radiotherapy. Since the prostate is not visible in the portal images, setup deviations are estimated with respect to the patient's bony anatomy of the pelvis. Bony anatomy being nondeformable, it is assumed that setup deviations can be represented by a 3D global rigid transformation comprising three translations and three rotations around three orthogonal axes. It should be noted that

since setup errors are estimated from the pelvic bony anatomy, any anticipated motion or deformation of the prostate with respect to the pelvis should be accounted for by the addition of an adequate internal margin to the clinical target volume. This is particularly important if the magnitude of prostate motion with respect to bony anatomy is large, as may happen in the absence of immobilization techniques such as rectal balloons. In instances where 2D on-board imaging of markers implanted inside the prostate is used for prostate localization, the algorithm proposed would also be useful in decoupling patient setup errors (or motion of the patient) and internal organ motion. Although originally developed for prostate radiotherapy, the methods presented in this study are also applicable to pelvic treatments including pelvic node irradiation.

A number of semiautomated^{1–4} and fully automated^{5–14} 2D–2D registration methods have been developed for the determination of setup errors in prostate radiotherapy. A limitation of 2D–2D registration methods is that they are typically unable to accurately estimate all six transformation parameters, especially in the presence of out-of-plane displacements.^{15,16} 2D–3D registration methods allow for the quantification of all six transformation parameters. Automated 2D–3D registration methods of a CT to MV portal images in prostate radiotherapy are fewer and include bony ridge-based registration by Gilhuijs *et al.*,¹⁷ a hybrid integrated information theoretic segmentation/registration framework by Bansal *et al.*,^{18,19} an evaluation of four intensity-based similarity measures by Clippe *et al.*,²⁰ our previous work^{21–24} which uses intensity-based similarity measures on either line-enhanced images or the raw intensity images and addresses the presence of scattered radiation and Poisson noise in the images, and a hybrid method by Chen *et al.*²⁵ Automated 2D–3D intensity-based registration methods applied to treatment sites other than the pelvis include the work by Kim *et al.*,²⁶ Kim and Fessler,²⁷ and Khamene *et al.*²⁸ Other studies which use kV fluoroscopes instead of MV portal images for the registration include intensity-based methods by Penney *et al.*,^{29,30} feature-based methods by Tomazevic *et al.*,³¹ and the hybrid method by Livyatan *et al.*,³² which was tested with pelvic data.

The performance of existing feature-based methods for the registration of portal images to a CT can be limited by the accuracy and robustness of the feature extraction or segmentation step, which is known to be difficult in portal images. Poor feature localization can result in an objective function with multiple local maxima which is difficult to optimize or with a peak at the wrong pose. Previously,^{22,33} we proposed a method to automatically enhance linear features in 2D digitally reconstructed radiographs (DRRs) obtained from the CT and 2D radiographs. Feature enhancement was accomplished using a matched filtering technique which was based on techniques developed by Lee and Rhodes³⁴ for the enhancement of linear features in noisy and low contrast imagery and by Czerwinski *et al.*^{35,36} for boundary enhancement in ultrasound images. The line-enhanced images were then used within a fully automated 2D–3D intensity-based registration framework with the correlation

coefficient as the similarity measure. It was assumed that the linear features represent mainly bony ridges, which are invariant to photon energy, in the planar images. We chose to enhance rather than segment features, and to use an intensity-based similarity measure computed from all the pixels in the line-enhanced images instead of pixels that might lie along a ridge only, believing this approach to be more robust to the presence of noise in the images.

The results reported in our previous study³³ were for the registration of a kV CBCT and kV radiographs of a phantom of the pelvis. The methods reported were designed, however, to be fully extendable to the registration of kV CT/CBCT images to MV portal images. Because of the lower differential attenuation of soft tissue and bone at higher energies combined with the presence of quantum noise and scattered radiation, MV portal images have a lower signal-to-noise ratio than kV radiographs and are therefore more difficult to register. In this study, we extend the proposed registration framework to MV portal images. We report our results with kV CBCT and MV portal images of a phantom of the pelvis and patient data consisting of a CT image acquired on a conventional scanner and MV portal images of the pelvis. As in all of our previous studies, a single portal image in the anterior–posterior (AP) view was used for the registration to the 3D CT.

II. METHODS AND MATERIALS

II.A. Registration algorithm

II.A.1. Feature enhancement

Linear structures corresponding mainly to bony ridges in the portal images and DRRs were enhanced by filtering the images with a filter bank of short line segments, or sticks, of variable orientation and with circular support as described previously by Munbodh *et al.*³³ and summarized here. Line-enhanced images are obtained by weighting the pixel gray level values in the original image with the “likelihood” of a pixel lying on a ridge as computed from gray level information in the local neighborhood around that pixel. At every pixel, the weights, which range in value from 0 to 1, are computed as an exponential function of the difference between the maximum projection value of the line segments over all orientations and the projection value of the circular region surrounding the line segment which gave the maximum projection value. Pixels which are likely to lie on a ridge have a weight close to one, whereas pixels in regions of almost constant intensity have weights close to zero. Calculating the weights as the difference of projection values of the line segments and circular neighborhood accommodates for additive variations in image brightness due, for example, to overlying soft tissue or the presence of scattered radiation. An important property of the line-enhancement operator is that it is rotation invariant.

A filter bank of 15 line segments evenly oriented between 0° and 180° and of length 11 pixels and width 1 pixel was employed for both the DRRs and portal images. The scale at which linear features are detected can be adjusted by varying

the width of the line segments. In this study, a fixed width of 1 pixel was found to be sufficient and linear feature enhancement was performed on the original unsmoothed images. Preprocessing was constant for all images of a given modality with the value of the constant, C , which affects the contrast of the output image being set to 20 for the DRRs in pose determination experiments on phantom and patient data and 20 and 5, respectively, for portal images of the phantom and of the patient. No other preprocessing was performed on the images.

II.A.2. Similarity measure and optimization procedure

Given a kV CT and a portal image, our task is to estimate the global rigid transformation, ψ , which aligns the CT to the portal image. The transformation ψ is a vector consisting of three translations, t_x , t_y , and t_z , and three rotations, r_x , r_y , and r_z , around the x , y , and z axes. The optimal transformation $\hat{\psi}$ was found by iteratively shifting and rotating the CT until the similarity between the line-enhanced DRR computed from the CT at a particular transformation and the line-enhanced portal image was maximized.

The similarity measure used was the Pearson correlation coefficient. It ranges from -1 to 1 with a value approaching 1 indicating a strong positive linear relationship between the random variables, a value of 0 indicating the absence of a linear relationship, and a value approaching -1 indicating a strong negative linear relationship between the random variables. The correlation coefficient was chosen as the similarity measure because even though the DRR and portal image are obtained at different photon energies, the relationship between the different tissue attenuation coefficients at kV and MV energies is similar. It is reasonable therefore to assume that the relationship between the intensity values in the line-enhanced DRR and portal images is linear. The Pearson correlation coefficient, $\rho(U, V_\psi)$, between the line-enhanced portal image and the line-enhanced DRR at transformation ψ was calculated as

$$\rho(U, V_\psi) = \frac{\sum_{i,j} (u(i,j) - \bar{u})(v_\psi(i,j) - \bar{v}_\psi)}{\sqrt{\sum_{i,j} (u(i,j) - \bar{u})^2} \sqrt{\sum_{i,j} (v_\psi(i,j) - \bar{v}_\psi)^2}} \quad (1)$$

where $\bar{x}$ is the mean of the pixels in x , $u(i,j)$ is the gray level value in the line-enhanced portal image at coordinates (i,j) , and $v_\psi(i,j)$ is the gray level value in the line-enhanced DRR under the transformation ψ at location (i,j) . To obtain V_ψ , the CT was first spatially transformed by ψ and a DRR was computed by summing the CT numbers in the spatially transformed CT along the rays from the x-ray source to each of the locations (i,j) on the 2D image plane under projective geometry. The DRR at transformation ψ was then filtered with the line-enhancement operator to give V_ψ .

Estimating the optimal transformation parameters that align the portal image to the CT involves optimizing the similarity measure over the search space. A multiresolution local search technique was used as described in Munbodh *et al.*²⁴ In the local search strategy, each of the six transformation parameters is perturbed alternately by a small fixed

amount (both positive and negative) and the optimization proceeds in the direction of the transformation parameter or transformation parameters which cause the largest increase in the correlation coefficient between the line-enhanced images. This procedure was iterated until the increase in the similarity measure caused by a small change in the transformation parameters was less than a predefined tolerance value.

The multiresolution optimization strategy accelerates convergence to the global maximum and increases the capture range of the registration algorithm. At any given level n , the images were blurred with a Gaussian kernel of standard deviation $0.5n$ and downsampled by a factor of n . The transformation parameters at that level were then used as the starting parameters to the next level until level 1 (the final level) was reached. Three optimization levels were employed in this study in pose-determination experiments on the phantom and two levels were employed when registering patient data.

II.B. Validation

The proposed registration algorithm was validated on images of an anthropomorphic phantom of the pelvis acquired under clinical conditions and on patient data. A single portal image in the AP view was used for the registration to the CT. The data used and pose determination experiments conducted have been described previously in Munbodh *et al.*²⁴ and are summarized here.

An orthogonal, right-handed coordinate system that was fixed with respect to the subject and with the origin at the isocenter was employed throughout this study. The x axis (right lateral, RL) was directed across the subject from right to left, the y axis (inferior–superior, IS) was directed along the subject from toe to head, and the z axis (posterior–anterior, PA) was directed upwards, through the subject and toward the x-ray source.

II.B.1. Phantom data

Pose determination experiments on an anthropomorphic phantom of the pelvis were conducted to quantify the ability of the registration algorithm to estimate known displacements of the phantom from its reference setup position for a range of clinically plausible displacements. In designing the pose determination experiments, particular attention was paid toward ensuring that the imaging geometry and ground truth were accurately known. The data were acquired at the Princess Margaret Hospital on an imaging system which was based on an Elekta SL 20 accelerator and which had both kV and MV imaging capabilities.³⁷ Prior to data collection, the imaging equipment was calibrated to ensure the isocenter of the MV imaging system and that of the kV imaging system were coincident. The imaging system had five degrees of freedom (r_y, r_z, t_x, t_y, t_z).

All CBCT scans were acquired at 120 kVp and all portal images were acquired at 6 MV and a dose of 4 MU. A CBCT scan of the phantom was acquired at the reference position. The origin of the CBCT image was at its geometrical center. The phantom was subsequently displaced from its

reference position by known amounts, as measured by the couch readout, and an MV AP portal image and a matching kV CBCT scan were acquired at each of the new poses. The matching kV CBCT scans were acquired for ground truth determination through the registration of four radio-opaque ball bearings affixed to the phantom using a singular value decomposition method.³⁸ The 15 poses tested are listed in Table I. The reference kV CBCT image used for alignment to the portal image was reconstructed on a $400 \times 400 \times 250$ matrix with a voxel size of $1.04 \times 1.04 \times 1.04$ mm³. The CBCT scans used for ground truth determination, including the reference CBCT, were reconstructed at twice that resolution (half the voxel size). The portal images were downsampled to 1×1 mm² and only a subregion, of size 12×12 cm² at the isocenter and excluding the femurs, of the original 20×20 cm² portal fields was used for registration to the reference CBCT. The origin of the portal images was not at their geometrical center and this was accounted for when reporting the registration results. The source-to-axis distance (SAD) was 1003.1 mm and the source-to-detector distance (SDD) at which the portal images were acquired was 1601.8 mm. Prior to registration, the ball bearings were removed from the reference CBCT image and the portal images using median filtering to ensure that they did not affect the registration.

The portal images contained bright horizontal lines due to defective detector elements in the top half of the image. The kV CBCT contained slight shading artifacts. These were not removed prior to registration.

II.B.2. Patient data

A planning kV CT image and 8 AP and RL portal images were acquired during treatment of a patient over 8 consecutive weeks at the Yale-New Haven Hospital. The patient was immobilized in a rigid foam cast during treatment. The CT was acquired on a GE CT scanner at 140 kVp. The patient was treated on a Varian Clinac2100 C/D machine and the portal images were obtained with an aS 500 Varian amorphous silicon electronic portal imager installed on a Clinac 2100C/D at 6 MV and a dose of 4 MU. The SAD was 1000 mm and the SDD, calculated from the graticule spacing on the portal images, was 1454 mm. Prior to registration, the CT and portal images were downsampled to an isotropic voxel/pixel size of 1 mm. A subregion of the original 15×15 cm² AP portal images of size 13×12 cm² and excluding the femurs was registered to the CT. The RL images were acquired for assessment of registration performance.

II.C. Quantification of registration errors

The total registration error in estimating displacements applied to the phantom was quantified as the maximum distance, $\mathcal{E}$, moved by points on a virtual sphere centered about the simulated isocenter (geometrical center of the CBCT) from their intended position,^{22–24}

$$\mathcal{E} = \max_{x \in S} \| (T_{\hat{\psi}}(x) - T_{\psi}(x)) \|, \quad (2)$$

where $T_{\hat{\psi}}$ and T_{ψ} are the transformation matrices at the estimated and actual transformation parameters, respectively, and x are the coordinates of the points on the surface of a sphere with a radius of 30 mm and at an angular separation of 5°. The magnitude of the displacement of the phantom from its reference position was quantified in a similar fashion by the distance, $\mathcal{D}$:

$$\mathcal{D} = \max_{x \in S} \| (T_{\psi}(x)) \|. \quad (3)$$

Since the ground truth was not known when registering the patient data, the success of the registration was verified visually, through the superposition of the registered line-enhanced DRRs and line-enhanced portal images. Although only AP images were used in the registration, images in the lateral view were used in addition to the AP image to verify that the images were correctly registered since displacements such as r_x and t_z are difficult to visualize in the AP view where they appear as out-of-plane displacements. These show up as in-plane displacements in the lateral view and can be better visualized allowing a better assessment of registration accuracy.

III. RESULTS

The true pose, estimated pose, and registration error in estimating each of the six transformation parameters for poses 1 through 15 are shown in Table I. The total registration error, $\mathcal{E}$, and the correlation coefficient between the line-enhanced DRR and line-enhanced portal image at the estimated pose are also listed along with the magnitude of the displacement, $\mathcal{D}$, applied to the phantom. Absolute translational displacements of the phantom along all three axes ranged up to 10 mm and absolute rotational displacements around the y and z axes were within 5°. In general, the registration errors were very small. Out-of-plane translations along the z axis were the most difficult to estimate among all six transformation parameters and the main contributor to the total registration error, although registration errors along the z axis were still quite small being less than 1 mm for 12 of the 15 poses tested. For the field of view used in the registration, the presence of out-of-plane rotations around the y axis (coupled with rotations around r_z) seemed to be correlated to an increase in the registration error in t_x and in a few cases (three out of five of the poses) in t_z and t_x . The magnitude of errors in estimating r_y itself, however, seemed to be unaffected. In order to draw definite conclusions regarding the correlation between errors in estimating the different parameters, more extensive validation on a larger number of poses would be required. The magnitude of the total registration error, $\mathcal{E}$, was not correlated to the magnitude of the displacement, $\mathcal{D}$. Of 12 of the 15 poses tested, the total registration $\mathcal{E}$ was less than 1 mm. The correlation coefficient at the estimated poses ranged from 0.60 to 0.82 with a mean

TABLE I. Table of displacements and registration errors for poses 1 through 15. The true pose, estimated pose, and registration error in estimating the true pose are listed for each of the six transformation parameters. The correlation coefficient, ρ , obtained at the estimated pose, the total distance moved, $\mathcal{D}$, and the total registration error, $\mathcal{E}$, are also shown.

		r_x (deg)	r_y (deg)	r_z (deg)	t_x (mm)	t_y (mm)	t_z (mm)	$\mathcal{D}$ (mm)	$\mathcal{E}$ (mm)	ρ
1	True pose	0	0	0	0	0	0	0		
	Estimated pose	-0.100	-0.018	-0.021	0.013	0.093	0.495			0.798
	Error	-0.100	-0.018	-0.021	0.013	0.093	0.495	0.557		
2	True pose	0.027	0.018	1.059	-11.427	-8.966	-10.701	18.480		
	Estimated pose	-0.116	0.021	1.097	-11.487	-8.883	-10.254			0.596
	Error	-0.143	0.003	0.038	-0.060	0.083	0.447	0.531		
3	True pose	0.113	0.007	1.081	-5.757	-4.401	-4.380	8.933		
	Estimated pose	-0.006	0.139	1.050	-5.768	-4.539	-4.997			0.606
	Error	-0.119	0.132	-0.031	-0.011	-0.138	-0.617	0.727		
4	True pose	0.085	4.999	1.035	-1.415	0.022	-0.232	4.106		
	Estimated pose	-0.030	4.967	0.987	-1.667	-0.086	1.261			0.789
	Error	-0.115	-0.032	-0.048	-0.252	-0.108	1.493	1.584		
5	True pose	0.000	-0.005	6.466	-1.496	1.817	-0.196	5.740		
	Estimated pose	-0.038	0.045	6.472	-1.565	1.711	-0.176			0.788
	Error	-0.038	0.050	0.006	-0.069	-0.106	0.020	0.159		
6	True pose	0.000	4.995	6.466	-1.496	1.817	-0.196	6.486		
	Estimated pose	-0.655	5.029	6.542	-1.823	1.703	1.394			0.809
	Error	-0.655	0.034	0.076	-0.327	-0.114	1.590	1.958		
7	True pose	0.140	-0.001	1.128	8.570	-0.150	-0.786	9.202		
	Estimated pose	-0.045	0.107	1.159	8.513	-0.242	-0.645			0.812
	Error	-0.186	0.108	0.031	-0.057	-0.092	0.141	0.291		
8	True pose	0.004	0.009	1.082	-1.057	10.591	-0.778	11.237		
	Estimated pose	0.283	0.076	1.222	-1.214	10.679	-0.450			0.768
	Error	0.279	0.067	0.140	-0.157	0.088	0.328	0.541		
9	True pose	0.081	0.024	1.130	-0.724	-2.586	-0.846	3.379		
	Estimated pose	0.033	0.076	1.128	-0.846	-2.680	-1.325			0.777
	Error	-0.048	0.052	-0.002	-0.122	-0.094	-0.479	0.540		
10	True pose	0.068	-0.007	1.130	-0.833	0.735	9.724	9.902		
	Estimated pose	0.080	0.053	1.112	-0.971	0.625	9.347			0.801
	Error	0.012	0.060	-0.018	-0.138	-0.110	-0.377	0.449		
11	True pose	0.094	-0.022	1.105	-0.809	0.777	-4.121	4.421		
	Estimated pose	0.040	-0.049	1.104	-0.862	0.601	-4.715			0.787
	Error	-0.054	-0.027	-0.001	-0.053	-0.176	-0.594	0.653		
12	True pose	-0.052	0.004	1.129	-0.861	1.081	-10.631	10.833		
	Estimated pose	0.048	0.060	1.128	-0.956	1.125	-10.122			0.794
	Error	0.100	0.056	-0.001	-0.095	0.044	0.509	0.580		
13	True pose	0.124	5.022	5.328	9.647	1.607	0.061	13.593		
	Estimated pose	-0.522	5.092	5.378	9.271	1.547	1.456			0.820
	Error	-0.646	0.070	0.050	-0.376	-0.060	1.395	1.773		
14	True pose	-0.095	5.013	5.328	-1.769	9.774	0.165	12.967		
	Estimated pose	-0.280	5.076	5.386	-2.081	9.851	-0.075			0.799
	Error	-0.185	0.063	0.058	-0.312	0.077	-0.240	0.490		
15	True pose	0.152	5.032	5.283	-1.877	0.997	10.304	13.262		
	Estimated pose	-0.342	5.084	5.378	-2.245	0.945	9.847			0.807
	Error	-0.494	0.052	0.095	-0.368	-0.052	-0.457	0.833		

TABLE II. Mean, standard deviation, minimum, maximum, and median of the absolute value of the registration errors for each of the six transformation parameters and also for the total registration error for poses 1 through 15.

Absolute error	r_x (deg)	r_y (deg)	r_z (deg)	t_x (mm)	t_y (mm)	t_z (mm)	$\mathcal{E}$ (mm)
Mean	0.212	0.055	0.041	0.161	0.096	0.612	0.778
s.d.	0.214	0.033	0.039	0.131	0.033	0.485	0.543
Minimum	0.011	0.003	0.001	0.011	0.044	0.019	0.159
Maximum	0.655	0.131	0.139	0.376	0.176	1.590	1.959
Median	0.119	0.052	0.031	0.123	0.093	0.478	0.557

value of 0.77 indicating that the pixel gray level values in the aligned line-enhanced images were strongly linearly correlated.

Table II shows the mean, standard deviation, minimum, maximum, and median of the absolute registration error over all 15 poses for each of the six transformation parameters and of the total registration error. The mean and (standard deviation) of the absolute error for the individual parameters and of the total registration error was r_x : 0.212° (0.214°), r_y : 0.055° (0.033°), r_z : 0.041° (0.039°), t_x : 0.161(0.131) mm, t_y : 0.096(0.033) mm, t_z : 0.612(0.485) mm, and $\mathcal{E}$: 0.778(0.543) mm. It should be noted that since we did not test for rotations around the x axis, the small registration errors obtained for r_x simply indicate that the registration algorithm did not drift from its starting point of 0° for r_x .

Figure 1 shows the superimposed line-enhanced DRRs and portal images in the AP view before and after registration for all 8 weeks. In these figures, linear features in the portal images are green in color, while those in the DRR are red. At alignment, linear features corresponding to bony

ridges in both images should overlap appearing yellow. As can be seen from Fig. 1, the images were correctly aligned over all 8 weeks. After registration of the planning CT to the portal images and after correcting for the shift in the treatment field isocenter with respect to the geometric center of the portal image, we found that estimated in-plane translational setup deviations with respect to the planned position of the patient ranged from -3.5 to 2.5 mm along the x and y axes. Out-of-plane translations along the z axis ranged from -2.1 to 3.3 mm. Estimated out-of-plane rotations around the y axis were all less than 1.3° . A systematic setup deviation of approximately 2.5° around the x axis (out of plane) and -1° around the z axis (in plane) was observed during all 8 weeks. The systematic rotation around the x axis was possibly due to couch sag during the planning CT acquisition. Even though the line-enhancement procedure also enhanced the outlines of collections of bowel gas, these did not affect the registration as can be seen in the registered images.

RL DRRs and portal images of the patient were used to verify that translations along the z axis and rotations around

FIG. 1. Estimation of patient setup deviation with the lines algorithm using images in the AP view. An overlay of the portal image and the DRR in its original pose for weeks 1–8 is shown in the first and third columns. Images of the fused portal image and the DRR computed at the estimated pose for weeks 1–8 are shown in the second and fourth columns. Linear features from the portal image are green in color and features from the DRR are red. At alignment, linear features in both images should overlap minimizing the amount of red and green structures in the fused registered images and maximizing the amount of yellow. Note that the bony anatomy was registered correctly for all 8 weeks. The images displayed here do not include the correction for the treatment field offset.

FIG. 2. An overlay of the portal image and the DRR in its original pose for weeks 6 and 7 is shown in the first column. Images of the fused portal image and the DRR computed at the estimated pose for these 2 weeks using the AP and RL image separately are shown in the second and third columns, respectively. Linear features from the portal image are green in color and features from the DRR are red. The AP-only registration algorithm was able to recover the pelvic tilt present in week 6 accurately, while the RL-only registration algorithm incorrectly aligned the femurs instead of the pubic symphysis and sciatic notch. The registration with the lateral image was wrong by approximately 2° and 2 mm in its estimation of the rotation around the right-lateral axis and the translation along the posterior–anterior axis, respectively. No femur motion or pelvic tilt was observed in week 7. In this case, estimates of setup deviations obtained with the AP image and lateral image were comparable.

the x axis were estimated correctly as illustrated for weeks 6 and 7 in the second column of Fig. 2. Interestingly, in the process of verifying that these two out-of-plane deviations were correctly estimated, it was observed that the pelvis was slightly tilted with respect to the femurs during weeks 3, 5, and 6, despite the patient being immobilized in a rigid foam cast during treatment. The third column of Fig. 2 shows an overlay of the RL portal image and DRR with the registration performed with the RL image only for weeks 6 and 7. No femur motion was observed in week 7 and the registration with the AP and RL images were both correct. For week 6, however, the AP-only registration algorithm was able to recover the pelvic tilt accurately, whereas the lateral-only registration algorithm incorrectly aligned the femurs instead of the pubic symphysis and sciatic notch. This suggests that in the presence of femur motion or pelvic tilt use of lateral images in the registration might be a source of errors. Unless the femurs are excluded from the lateral images, it is preferable to use the AP image only in the registration.

The registration computational time depends on several factors, namely the size of the DRRs, the resolution of the images, the magnitude of the displacements being estimated, and the optimization strategy used. It was found, here, that the main bottleneck to speed was the time to compute the DRRs. In this study, full field of view DRRs were computed from the CT, even though the similarity measure was computed on only a subregion of those DRRs. On a 2.6 GHz Dell Precision 650 workstation, the time for the 2D–3D registration algorithm with three optimization levels to converge to the final solution was about 15 min for a 200×200 DRR with a resolution of $2 \text{ mm} \times 2 \text{ mm}$ computed from a CT with a resolution of $2 \text{ mm} \times 2 \text{ mm} \times 2 \text{ mm}$. For a DRR of size 400×400 pixels and a resolution of $1 \text{ mm} \times 1 \text{ mm} \times 1 \text{ mm}$,

the registration time increased about eightfold to about 2 h. The time to compute a DRR at a resolution of $1 \text{ mm} \times 1 \text{ mm}$ was about 40 s and every iteration in the local search optimization strategy required the computation of 13 DRRs. The main purpose of this study was to establish that accurate registration was feasible with the algorithm presented and a single portal image rather than to achieve computational speed. The speed of the registration could be improved vastly by using new techniques for fast DRR computation,^{39–42} which would allow computation of the DRRs in milliseconds, and also by employing a more efficient optimization strategy such as conjugate gradient ascent or the optimization strategy used in the study by Munbodh *et al.*³³ for the alignment of kV CT data to kV radiographs. We chose to use this optimization strategy here for consistency with our previous publication in which we investigated the effect of filtering on the performance of intensity-based registration of kV CT to MV portal images.²⁴

IV. DISCUSSION

We have evaluated a completely automated method to register a 3D kV CT to a single 2D MV portal image. Linear features in the kV DRRs and MV portal images were automatically enhanced using a filtering technique which involved correlating the images with a filter bank of line segments at different orientations. Line enhancement simultaneously served to reduce additive low-frequency artifacts from soft tissue and scattered radiation in the images. The line-enhanced images were then registered using the Pearson correlation coefficient as the similarity measure. The registration algorithm was validated on kV CBCT and MV AP portal images of an anthropomorphic phantom of the

pelvis and kV CT and MV AP portal images of a patient. Excellent registration accuracy was obtained. The mean and (standard deviation) of the absolute error in estimating each of the six transformation parameters in pose determination experiments with the phantom was r_x : 0.212° (0.214°), r_y : 0.055° (0.033°), r_z : 0.041° (0.039°), t_x : 0.161 (0.131) mm, t_y : 0.096 (0.033) mm, t_z : 0.612 (0.485) mm. The mean and (standard deviation) of the overall registration error was 0.78 (0.54) mm. Registration of the patient data was successful in all cases as confirmed visually. These results are very important as they indicate that (1) it is possible to automatically enhance linear features in the portal images for robust and accurate registration and (2) for the range of displacements tested, the AP view of the portal image contains features which are sensitive to and can be used to determine out-of-plane setup deviations (r_x, r_y, t_z) and can be reliably used within an automated 2D–3D registration framework.

This study is an extension of our previous work³³ where the line-enhancement algorithm was first proposed and used within a completely automated registration framework to align a kV CBCT image to kV AP radiographs of the same pelvic phantom used here. There, the mean and (standard deviation) of the absolute registration error for each of the six transformation parameters was r_x : 0.029° (0.016°), r_y : 0.030° (0.028°), r_z : 0.011° (0.009°), t_x : 0.021 (0.023) mm, t_y : 0.020 (0.020) mm, and t_z : 0.134 (0.096) mm. Those results are better than the results obtained in this study, which is to be expected given that the kV radiographs had a higher signal-to-noise ratio than the MV portal images. As in this study, the main source of registration inaccuracy was the estimation of translations along the z axis. This is because translations along the z axis cause but a small change in scale in the projection images resulting in an objective function with a relatively flat global peak. Although not reported here, the registration algorithm has also been validated on simulated data of an anthropomorphic phantom of a pelvis and a kV CT and 8 MV portal images on film of a pelvic phantom and found to be accurate.^{21,22} To date, the proposed algorithm has been tested on images acquired on a range of imaging equipment spanning technical developments from film, through electronic portal imaging devices and kV on-board imagers to CT and CBCT and on both phantom and patient data.

Despite the relatively larger error in estimating translations along the z axis, the accuracy achieved in this study with a single portal image in the AP view compares very favorably to the accuracy reported in other studies which use two orthogonal portal images^{17,20,19} and which are described in the following.

In the approach by Gilhuijs *et al.*,¹⁷ bony ridges were either automatically extracted or manually delineated in the transmission images. The method was tested by aligning a randomly displaced 3D CT to pairs of simulated transmission images. These comprised pairs of simulator images and portal images artificially created from the patients' 3D CT (field sizes of approximately 13×13 cm²). Validation was also performed on three clinical data sets from three prostate pa-

tients. The method was found to be fast and accurate to within 1 mm and 1°. In a subsequent study by Remeijer *et al.*,¹⁶ however, the performance of the method proposed by Gilhuijs *et al.* was found to be limited by the segmentation step of bony anatomy in lateral images of the pelvis.

Clippe *et al.*²⁰ tested four similarity measures: The Pearson correlation coefficient, chi square, mutual information, and the correlation ratio within an automated 2D–3D registration framework. The algorithms were validated on a 3D CT and orthogonal portal images of an anthropomorphic phantom for seven displacements with a field of view of 15×15 cm². The registration error was quantified as the root mean square distance of 1000 points spread inside a cube of 15 mm³ centered at a simulated isocenter from their desired position. The mean of the registration error for the seven poses tested was, respectively, 2.3, 6.0, 6.3, and 6.8 mm for the correlation ratio, correlation coefficient, chi square, and mutual information.

Bansal *et al.*¹⁹ presented an integrated 2D–3D information theoretic segmentation/registration framework to align a 3D CT and orthogonal portal images. Their method was tested on simulated data with additional testing being performed on a set of open-field images of an anthropomorphic phantom and a set of open-field clinical images (field of view of approximately 20×20 cm²). The accuracy obtained on phantom and clinical data was of the order of a few millimeters.

Few studies use a single radiograph to estimate setup deviations in radiotherapy. Lujan *et al.*⁴³ showed that AP radiographs of the skull contain anatomic features that are sensitive in position and orientation to out-of-plane rotations. Bijhold⁴⁴ investigated the feasibility of registering a single portal image to a 3D CT to estimate out-of-plane parameter variations. Their approach was based on the identification of anatomical match points in the 3D CT and portal image. Jabbari and Pistorius⁴⁵ presented a method to estimate out-of-plane rotations from a single portal image and its 2D reference image (a simulator image or the first portal image). The rotations were estimated from the apparent distortion of the anatomy in the portal image which can be mathematically predicted. It was shown that out-of-plane rotations could be recovered from the scaling of the image which is different in perpendicular directions.

In general, registration accuracy and precision will be a function of (1) the similarity measure, feature space, and search strategy employed in the registration, (2) experimental design and execution and knowledge of the ground truth, (3) knowledge of the imaging geometry, and (4) data complexity. Approximations introduced in the experimental design or during the execution of the experimental protocol, or in the completeness with which the imaging geometry is known will introduce additional degrees of freedom into the registration problem and compromise the accuracy and reliability of the registration results. We have now validated the proposed registration algorithm comprehensively and programmatically on kV CBCT and kV radiographs and MV portal images of a phantom and on CT and portal images of a patient, in order of increasing data complexity. The pose determination experiments with CBCT data of the phantom

were designed in such a way as to allow accurate determination of the ground truth. Care was also taken to ensure that the parameters describing the imaging geometry for use in the registration algorithm were accurately known. All of these have contributed to the high level of registration performance reported in this and previous studies.

This study is the last part of a larger study that we conducted to determine an appropriate feature space and class of similarity measures for 2D–3D registration of 2D radiographs (kV radiographs or MV portal images) to a 3D CT (CBCT or conventional CT).^{22,33,24} Multiresolution gradient ascent was used as the optimization strategy throughout in order to facilitate the comparative evaluation of the different features and similarity measures proposed. Previously, we used the spectral characteristics of bone and no-bone components of the 2D radiographs to guide the development of intensity-based registration methods which avoid soft tissue structure and low-frequency artifacts such as scattered radiation and focus on the entire bony anatomy of the pelvis for the registration.²⁴ As in this study, it was found that avoiding those low-frequency components improved registration performance significantly. Further, registration accuracy when aligning a kV CT scan to MV portal images was excellent when the higher spatial frequencies in the images were used despite the difference in photon energies at which the images were acquired. We also addressed the presence of Poisson noise in 2D radiographs and formulated a new similarity measure derived from a statistical model of the relationship between pixel intensity values in a 3D CT and 2D radiographs.²² The registration accuracy achieved in our studies was comparable, leading to the following conclusions.

First, it is clear that there is a signal which is common to CT and kV radiographs and to CT and MV portal images, and that this common signal can be used to achieve accurate registration. The signal useful for accurate registration is present at spatial frequencies other than the lowest frequencies irrespective of photon energy. Second, the high correlation values obtained at alignment indicate that the relationship between kV DRRs and kV radiographs and also between kV and MV portal images, at the higher spatial frequencies, can be considered to be linear. Third, while the noise present in kV and MV radiographs, in all likelihood, is Poisson in nature, similar performance of the correlation coefficient to similarity measures which address Poisson noise suggests that the nonstationary noise characteristics of these images have limited impact on registration accuracy at current imaging doses. That is, correlation-based similarity measures appear to be robust to the presence of Poisson noise in the images at current imaging doses. Our results suggest that, at current imaging doses, future methods could be constructed to further exploit linear and stationary Gaussian assumptions on signals at the higher spatial frequencies of the CT and portal images. At lower imaging doses, more sophisticated modeling of the relationship between intensity values in the two modalities may be necessary.

The methods presented in this study should, in theory, also be applicable to other rigid or piecewise rigid treatment

sites such as the cranial vault, skull base, and the spine. In such cases, we would have to judiciously select a viewing angle to maximize the feature presentation and its dependence on projection angle. For example, we may need to use the lateral or an oblique radiograph in place of the traditional AP radiograph for applications in the head region.

V. CONCLUSIONS

We have evaluated a fully automated 2D–3D registration framework whereby linear features in 2D kV DRRs, computed from a 3D CT, and 2D MV portal images are automatically enhanced and then aligned using the correlation coefficient as similarity measure. The mean and (standard deviation) of the overall registration error was 0.78(0.54) mm in pose determination experiments on a phantom. In experiments with patient data, the registration was successful in all instances. The results indicate that it is feasible to automatically enhance features in portal images of the pelvis for the accurate and robust estimation of patient setup errors in prostate radiotherapy. They also indicate that it is possible to estimate all six transformation parameters, including out-of-plane rotations and translations, from a 3D CT of the pelvis and a single portal image in the AP view.

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