Long-term stability and mechanical characteristics of kV digital imaging system for proton radiotherapy

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Purpose: To quantitatively evaluate the long-term image panel positioning stability and gantry angle dependence for gantry-mounted kV imaging systems.

Methods: For patient setup digital imaging systems in isocentric rotating proton beam delivery facilities, physical crosshairs are commonly inserted into the snout to define the kV x-ray beam isocenter. Utilizing an automatic detection algorithm, the authors analyzed the crosshair center positions in 2744 patient setup kV images acquired with the four imagers in two treatment rooms from January 2012 to January 2013. The crosshair position was used as a surrogate for imaging panel position, and its long-term stability at the four cardinal angles and the panel flex dependency on gantry angle was investigated.

Results: The standard deviation of the panel position distributions was within 0.32 mm (with the range of variation less than ± 1.4 mm) in both the X-Z plane and Y direction. The mean panel inplane rotations were not more than 0.51° for the four panels at the cardinal angles, with standard deviations $\leq 0.26^\circ$. The panel position variations with gantry rotation due to gravity (flex) were within ± 4 mm, and were panel-specific.

Conclusions: The authors demonstrated that the kV image panel positions in our proton treatment system were highly reproducible at the cardinal angles over 13 months and also that the panel positions can be correlated to gantry angles. This result indicates that the kV image panel positions are stable over time; the amount of panel sag is predictable during gantry rotation and the physical crosshair for kV imaging may eventually be removed, with the imaging beam isocenter position routinely verified by adequate quality assurance procedures and measurements. © 2014 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4868460>]

Key words: digital imaging, patient positioning, protons, panel position, flex map

1. INTRODUCTION

Interest in proton radiotherapy for cancer treatment is currently rising in the world as the number of proton treatment facilities is increasing rapidly.¹ To fully take advantage of the fast dose fall-off and sharp penumbra of proton beams, accurate patient positioning is critical.

Our institutional approaches for patient setup in a proton radiotherapy system utilizes an orthogonal kV x-ray image pair taken immediately before treatments with digital flat-panel imagers. The positions of fiducials or anatomic points on the x-ray images are compared to those on the digital reconstructed radiographs (DRRs) generated from planning x-ray computed tomography (CT) scans.² Good agreement between the kV imaging beam and the treatment beam isocenters is critical for accurate patient positioning. Because of the weight of the gantries, the bending magnets and the counterweights along with the rotating diameter as big as 10 m, the treatment isocenter wobbles as the gantry rotates.³ In addition, the source to imager distance (SID) is large and the imaging panel position and orientation changes with the rotating gantry. A common methodology to overcome such problems is to place two radiopaque wires, usually called crosshairs, in the imaging beams immediately downstream the x-ray source to determine the projection of isocenter on the image panel at

each orientation.² Frequent quality assurance (QA) measurements to verify the coincidence of crosshair and proton beam isocenter are essential.

A crosshair is not necessary in linac photon machines with modern gantry mounted kV imaging systems; instead, a gantry flex map is generated and corrections are applied to compensate for image panel and radiation head sag with gantry rotation. System commissioning as well as routine quality assurance procedures were developed and reported.⁴⁻¹⁰ It has been demonstrated that the panel flex map correction is critical to prevent loss of detail, image shifting, and streaking artifacts in volumetric images reconstructed in cone-beam CT (CBCT).¹¹ The prerequisite of flex map utilization is the repeatable panel position at a specific gantry angle. Such mechanical characteristics have been measured for the linac gantry mounted kV imaging systems,⁶ but there are no published reports for proton gantries. Selby *et al.*¹² reported an automatic image-based method which may be capable for proton radiotherapy imaging system calibration, however, measurements have not been reported for the mechanical characteristics of imaging systems in isocentric proton radiotherapy. For this reason, the goal of the present study is to characterize the long-term stability of image panel positions and quantify the panel flex with gantry rotation in a proton treatment system.

FIG. 1. Schematics of the DIPS for gantry proton radiotherapy systems. The square section indicates the “top-left” pixel of each panel.

2. MATERIAL AND METHODS

2.A. X-ray imaging system

In our proton treatment facility, the proton beam is accelerated using an IBA cyclotron C230 and we use two orthogonal x-ray sources and the imaging panel pairs for patient alignment in each proton gantry beam delivery system. Figure 1 is a schematic diagram showing the geometry of the digital imaging positioning system (DIPS). The directions of the X, Y, and Z axes of treatment room coordinate system are depicted in the center in Fig. 1. The gantry rotates about the Y-axis, the gantry angle equals zero when the beam direction is opposing the +Z axis and increases clockwise when looking along +Y axis. The x-ray sources, the crosshairs, and the image panels are attached to and rotate with the gantry. In Fig. 1, the beam-line (BL) source is located at a gantry angle of 90° and it is coaxial with the proton beam; the crosshair is inserted in the snout for imaging and moved out before treatment. The two wires of the BL crosshair are parallel to the Y and Z axes, and the intersection of the two wires is aligned to the x-ray beam central axis. Ideally, the x-ray beam central axis and the proton beam central axis both pass through the mechanical isocenter so that the projection of the crosshair center defines the proton beam isocenter projection on the BL imaging panel, and this coincidence is routinely verified. When the gantry rotates, one wire is always parallel to the Y-axis, while the other is in the X-Z plane (perpendicular to the Y-axis). The layout of the other x-ray imaging path is similar to the beam-line one except that the x-ray source gantry angle is 90° less, counterclockwise from the beam axis, therefore named G90. The G90 crosshair is permanently attached to the x-ray source; one of the G90 crosshair wires is also always parallel to the Y-axis, and the other wire is parallel to the X-axis as shown in Fig. 1 and in the X-Z plane when the gantry rotates. In this paper, gantry angle refers to the direction of proton beam and the beam-line x-ray source. The distance from the

TABLE I. SAD and SID for the x-ray imaging systems in the two gantry treatment systems. G1: gantry room 1; G2: gantry room 2.

Distances (mm)	G1-beamline	G1-G90	G2-beamline	G2-G90
SAD	1522	2880	1511	2838
SID	2082	3390	2094	3388

x-ray source to the isocenter [Source to axis distance (SAD)] and to the image panel (SID) are different for the four image panels used in the two gantry rooms (G1 and G2), and the values are listed in Table I. The SAD and SID are larger than the typical linac photon gantries, where SAD and SID are about 1000 and 1500 mm, respectively.

The x-ray tubes are Varian A-277 with B15H housing (Varian Medical System) and operating at 70–110 kVp. Inhouse designed image panel motion control systems use Kollmorgen SERVOSTAR CE03550 controllers (Kollmorgen) to move the image panels; the nominal position accuracy is 0.1 mm. The four panels currently used in our gantry treatment rooms are PaxScan 4030R amorphous silicon digital x-ray detectors (Varian Medical Systems). The pristine pixel matrix is 3200×2304 with each pixel 0.127×0.127 mm in size. The images are binned to 1600×1152 pixels of 0.254×0.254 mm before saving to database. In this work, we used the downsized archived images.

2.B. Clinical QA method for the imaging system: Isocenter position accuracy at various gantry angles

The accuracy of the x-ray imaging system is routinely verified by weekly and annual QA procedures. In weekly QA process, a stainless steel BB 1.6 mm (1/16th in.) in diameter embedded in plastic is first aligned to the x-ray isocenter using the x-ray imaging system at 270° gantry angle. An aperture with a 2 mm diameter circular opening is placed in the beam line, and a double-exposed x-ray image is acquired to verify the coincidence of the crosshair center and the proton beam center. Accuracy of the x-ray imaging system is also checked weekly with BBs embedded at four vertices of a cube phantom as shown in Fig. 1. The black dots represent the BBs, which are arranged in such a way that there is no overlapping of the BB projections on any side of the cube.

Annual QA measures the isocenter positions with gantry at 0° , 90° , 180° , and 270° , and an agreement within 1 mm is commonly achieved and considered acceptable. Besides the four cardinal angles, we also measure the isocenter wobbling at multiple gantry angles annually. In this test, we measure the BB's projected positions with respect to the crosshair center in each panel. For this study, the BB was aligned to the isocenter and its position was measured at 15° – 20° increments for both clockwise and counterclockwise rotations, and the BL and G90 variations were used to measure wobble of the isocenter from the relative BB to crosshair displacement. The proton and x-ray isocenter coincidence are also checked on a yearly basis.

During clinical quality assurance tests (and patient setup), inhouse developed software first detects the crosshair center

FIG. 2. An example of the original DIPS image (a) and the intensity-reversed image (b). The image intensity along the row 640 (solid line) and the Gaussian fitting to determine the center of the vertical wire on this row are plotted in (c).

position and inplane panel rotation and corrects for it, if any. This step removes uncertainty in panel sag and inplane rotation with gantry rotation; therefore, our clinical QA tests characterize the isocenter wobbling caused by the sag of beam delivery nozzle, the x-ray sources, and the crosshair.

2.C. Crosshair automatic detection method

For the purpose of this work, a MATLAB® (version R2010b, MathWorks, Natick, MA) script was written to automatically calculate the crosshair center position and the inplane rotation using a method similar to what is utilized in the clinic DIPS software. The crosshair center position and the panel rotation of each image were then recorded for analysis in this study.

Before the crosshair detection, the intensity of the original kV image, as seen in Fig. 2(a), is reversed for better visualization [Fig. 2(b)]. Crosshair detection is a two-step process: a rough estimation and a fine-tuning, with the positions of the vertical and the horizontal wire determined separately. Taking the vertical line as an example, the rough estimation determines in which row the center of the vertical wire is located in the image. Each image consists of M rows and N columns of pixels, with the top-left one indexed (1, 1) and the bottom-right one (M , N). In the image coordinate system, as depicted in Fig. 2(b), positive “x” and “y” point to the directions of increasing column and row indices, respectively. During the first step, we calculate a score S_V for each column based on the intensities of its $2w$ neighboring columns, where $2w + 1$ is the nominal width (in unit of number of pixels) of the vertical wire projection on the panel. For the n th column: $S_V(n) = \sum_{m=1}^M \{2 \times I(m, n) - \min[I(m, n : n - w)] - \min[I(m, n : n + w)]\}$ where $I(m, n)$ is the intensity of pixel (m, n), $\min[I(m, n : n - w)]$ is the lowest intensity

among pixels (m, n) and w pixel to its left, and $\min[I(m, n : n + w)]$ is the lowest intensity of pixels (m, n) and w pixel to its right. Since we know that the crosshair center should be in the central region of the panel, we only calculate $S_V(n)$ for $n \in \{\frac{1}{3}N : 1 : \frac{2}{3}N\}$, then the median row index of the w rows with highest scores is the rough estimated position of the vertical wire N_{rough} . Similarly, we calculate the score $S_H(m)$ for $m \in \{\frac{1}{3}M : 1 : \frac{2}{3}M\}$ and get the rough estimated position of the horizontal wire M_{rough} .

The next step is to find the precise wire positions by curve fitting. Again, let us take the vertical wire as an example. We first find the center of the wire along each row using the $2w + 1$ pixels near column N_{rough} by the following Gaussian fitting formula: $y = a \cdot \exp(-\frac{(x-b)^2}{c^2}) + d$, with $x = N_{\text{rough}} - w : 1 : N_{\text{rough}} + w$ being the indices of columns; $y = I(m, N_{\text{rough}} - w : 1 : N_{\text{rough}} + w)$ the image intensity of the $2w + 1$ pixels near column N_{rough} on the m th row; and a, b, c , and d are the parameters to be fitted. The fitted parameter b_m is the center of the vertical wire on the m th row. Figure 2(c) shows the image intensity and the fitted function for the pixels along the solid line in Fig. 2(b). By repeating the Gaussian fitting for multiple rows, we can get the central positions b_m of the vertical wire as a function of m . To reduce the computational time, the Gaussian fitting is performed for every five rows for $m \in \{\frac{1}{4}M : 1 : \frac{3}{4}M\}$, and there was no observed difference in the results.

The final step is to fit the center positions to a straight line that is used to represent the precise location of the wire. This is performed using the $v = \text{robustfit}(m, b_m)$ function of Matlab curve fitting toolbox, where b_m is the fitted peak positions of the m rows. Using the fitted parameter v_1 and v_2 , the vertical line can be expressed as $x = v_1 + v_2 \cdot y$. Similarly, the expression $y = h_1 + h_2 \cdot x$ was obtained for the

horizontal line. The final position of the two wires was also visually inspected. Occasionally, the crosswire positions were incorrectly detected due to line artifacts or edges in the image, so our software allows visual verification and manual correction, if needed, in both steps.

The inplane rotation of the vertical and horizontal wires are $\theta_v = \arctan(v_2)$ and $\theta_h = -\arctan(h_2)$, where positive rotation is counterclockwise while looking at the panel from the x-ray source; the crosshair rotation is the average of θ_v and θ_h . We calculate the interception of the two wires by solving equations of the two wires and the coordinates of the crosshair center (in unit of pixel index) are: $C_x = (v_1 + v_2 \cdot h_1) / (1 - v_2 \cdot h_2)$ and $C_y = h_1 + v_1 \cdot h_2 / (1 - v_2 \cdot h_2)$. The coordinates are converted to the distance to the top-left pixel by multiplying the pixel sizes in the x and y directions, respectively. All distances were then projected to the isocenter plane by multiplying the SAD/SID ratio for the specific image panel.

2.D. Patient image data

Although the crosshair positions in the kV images detected by DIPS are saved in our clinical database, such data may not be appropriate for panel position repeatability analysis, because the image panels were sometimes intentionally shifted for a better view of the patient geometry or for calibration procedures. Due to the enormous amount of total images in the clinical database (e.g., 47 118 images in year 2012), it is impractical to hand-pick all of the useful images. Instead, for the purpose of this work, we used a subset of these images acquired for patient setup not involving panel shifts.

Under an institutional review board approved protocol, a total of 2744 x-ray images from randomly selected patients treated for various disease sites in the two gantry rooms from January 2012 to January 2013 were collected and analyzed to study the long-term stability of the image panel positions. Among these, 2527 images were taken at the four cardinal angles, i.e., 0° , 90° , 180° , and 270° , from the four panels; the numbers of images for each panel and gantry angle are listed in Table II.

For beams with oblique incidence, sometimes the patients were initially set up with the orthogonal BL and G90 image pair at one of the cardinal angles, and then with the BL only after the gantry was rotated to the treatment position. As a result, there were more BL images than G90 ones.

TABLE II. Number of x-ray images at the four cardinal angles.

Gantry angle (deg)	G1		G2	
	BL	G90	BL	G90
0	252	165	115	88
90	270	215	116	84
180	91	67	118	79
270	352	269	142	104
Total	965	716	491	355

To investigate the amount of crosshair position variation during gantry rotation, we used the kV x-ray images taken at all available angles from randomly selected patients treated within one month (December 2012). A total of 420 images at 13 different gantry angles, including the cardinal angles, were analyzed; and there were at least four images for each gantry angle, panel (BL or G90), and treatment room combination.

3. RESULTS

3.A. Isocenter position vs gantry angle: Clinical QA results

Figures 3(a)–3(d) show isocenter position (measured from the crosshair center to a BB aligned to the x-ray isocenter) variation in the X-, Y-, and Z-directions at different gantry angles for the four kV image source/panel combinations: G1-BL, G1-G90, G2-BL, and G2-G90. These variations are caused by beam delivery nozzle and x-ray source sag but not panel flex, because the crosshair center position was determined at each gantry angle on each panel separately. The isocenter wobbling in Y-direction were all within 0.5–1 mm. In the X- and Z-direction, the magnitude of wobbling was within about ± 1 mm, except for G2-G90 which was about ± 2 mm.

3.B. Long term crosshair position stability

The distribution of the crosshair center positions relative to the top-left pixel in each panel of the 2527 images taken at four cardinal angles is shown in Figs. 4(a)–4(d) for the four panels. Due to maintenance services (x-ray tube and panel replacement) being performed during the period of our investigation, the relative position of the panel may be shifted slightly; consequently, the crosshair center positions are separated into two clusters, such as clearly seen in Fig. 4(c) for G1-G90 panel at 0° (AP). Figures 4(e) and 4(f) show the trend of the crosshair center position variation in the Y and X-Z direction grouped into weeks for G1-G90 panel, and the black lines indicate when the panel or tube was replaced. It is obvious that the crosshair position shifted systematically after the panel replacement. Similar patterns were also observed for the other three panels; therefore, we evaluate the long-term stability of the crosshair by comparing the crosshair positions relative to the mean positions between maintenance.

Since the mean panel positions at different gantry angles can be included in a flex map, we calculated the panel position variation relative to its mean position at each cardinal angle separately. The histograms of the crosshair positions relative to the mean position at all four cardinal angles combined are shown in Figs. 5(a)–5(d) for the four individual panels and in (e) for the four panels, and the standard deviation and range of the crosshair positions are listed in Table III.

Statistics from our raw clinic data showed standard deviations of panel positions at the cardinal angles of 1–1.5 mm. These values are higher than the results derived from the selected images, because factors that artificially increase the errors, such as intentional panel shifts, systematic panel

FIG. 3. (a)–(d) Isocenter positions with respect to the crosshair center measured at multiple gantry angles for the four panels used in the two gantry treatment systems.

FIG. 4. (a)–(d) Distances from the crosshair center to the up-left corner of the panel backprojected to the isocentric plane at the four cardinal angles for the four imagers in the two gantry rooms. (e) and (f) For G1-G90 panel, the trends of the crosshair center positions are plotted as functions of time (weeks). AP, LL, PA, and RL correspond to the geometries where the BL source (and proton beam nozzle) located at gantry angle of 0° , 90° , 180° , and 270° , respectively. The solid black lines indicate time for panel or tube replacement.

FIG. 5. Histogram of the crosshair position variations at the four cardinal angles for the four panels (a)–(d) and all four panels (e).

position change after panel or source replacement, etc., were not properly accounted for.

3.C. Crosshair rotations

The mean and standard deviation of the crosshair inplane rotations are tabulated in Table IV for the four panels at the four cardinal angles. The mean rotations were within 0.5° , and the standard deviations were all below 0.26° .

TABLE III. The standard deviation (σ) and the range of the relative crosshair positions.

Panel	X-Z (mm)		Y (mm)	
	σ	Range	σ	Range
G1-BL	0.27	[−1.00 0.79]	0.28	[−0.84 1.34]
G1-G90	0.20	[−0.87 0.83]	0.21	[−0.81 1.10]
G2-BL	0.20	[−0.61 0.68]	0.21	[−0.97 0.46]
G2-G90	0.32	[−1.26 1.10]	0.20	[−0.79 0.81]
All 4	0.25	[−1.26 1.10]	0.24	[−0.97 1.34]

3.D. Crosshair position variation vs gantry angles

The crosshair positions are plotted as a function of gantry angle in Figs. 6(a)–6(d) for G1-BL, G1-G90, G2-BL, and G2-G90, respectively, where the bars represent the standard deviation of multiple measurements. The standard deviation is substantially smaller than the variations at different gantry angles, which indicates good repeatability of panel positions at each gantry angle. The variations among different gantry angles are in general in the order of a few millimeters in both Y and X-Z directions and are different for each panel. As previously mentioned G90 images are usually only acquired for

TABLE IV. Statistics of the crosshair/panel rotations of the four panels at the cardinal angles (unit: degree).

Panels	AP		LL		PA		RL	
	Mean	$\pm$ std	Mean	$\pm$ std	Mean	$\pm$ std	Mean	$\pm$ std
G1-BL	−0.01	0.16	0.09	0.07	0.11	0.17	−0.23	0.16
G1-G90	−0.14	0.12	−0.32	0.08	−0.38	0.09	−0.20	0.11
G2-BL	−0.43	0.19	−0.51	0.10	0.47	0.15	−0.27	0.26
G2-G90	0.37	0.07	0.27	0.09	0.11	0.05	0.26	0.13

FIG. 6. The crosshair center relative positions (backprojected to the isocentrical plane) at different gantry angles.

setup fields at the cardinal angles; therefore, there were limited G90 images at noncardinal angles for our analysis.

4. DISCUSSION

In general, there have been few publications on patient setup imaging system QA of proton radiotherapy systems. Many gantry delivered proton radiotherapy facilities use crosshairs placed in the path of proton and x-ray beams to define the isocenter position on the image panels, since this historical practice can eliminate image panel position variations, which may be substantial for proton gantry imaging panels as they are frequently extended from and retracted to the rotation gantry.

To our knowledge, there is no published result about the stability of proton image panel positioning. However, this information is critical for crosshair-free imaging guidance system. For example, there exists commercial software (e.g., VeriSuite® Patient Positioning and Verification System) that can be implemented for x-ray imaging systems in particle radiotherapy systems similar to ours. A crosshair-free imaging system usually uses a machine-specific flex-map look-up table to correct for gantry, x-ray source, and panel sag. Mechanical characteristics of the system are necessary to determine the appropriate calibration procedure and test frequency for each component of the system.

We investigated the long-term panel position stability at the four cardinal angles for the four panels used in our two gantry proton treatment systems for a period of a year. Using the automatic crosshair detection algorithm, we were able to analyze a large amount of real patient x-ray images collected before treatment beam deliveries. The overall crosshair

position relative to the panel is quite stable for the same gantry angle. For all four panels, the position variation standard deviation is between 0.20 and 0.32 mm, and the maximum variation from the mean position is 1.34 mm in either Y (inline) or X-Z (cross-line) direction among all the 2527 images taken over a period of 13 months. In the meantime, the panel rotation variations were not greater than 0.26° (standard deviations). This demonstrates that the image panel positions are highly reproducible, despite the fact that they are frequently moving into and out of the imaging positions. Note that analysis from our clinic data indicates the panel position variations were 1–2 mm (standard deviation). As previously mentioned, this is an overestimation due to intentional panel shifts.

Based on the highly reproducible crosshair positions over such a long term, we believe it is reasonable that we do not need to detect the crosshair positions in each and every image because they can be corrected by the gantry flex map. To take it one step further, the crosshair may not be necessary for patient setup imaging. Removal of the crosshair can not only speed up the patient setup procedure but also improve the visualization of the images and possibly reduce the uncertainties in fiducial or anatomic feature markups, because they could be blocked by the crosshair wires. Furthermore, imaging without crosshairs would allow CBCT for proton gantry systems. Of course, without the crosshair, adequate quality assurance tests are needed and appropriate procedures should be designed and implemented in order to verify the panel position and the overall coincidence of the imaging system geometry to the proton treatment beam geometry. Such QA procedures have been developed for linac delivered external photon beams,^{4-7,11} and with appropriate modifications, can be used for proton imaging systems.

The other purpose of our work is to characterize the image panel sag while the gantry rotates. As shown in Fig. 6, the crosshair position variation (standard deviation) at a specific gantry angle is much smaller than that at various angles. Combined with the demonstrated long-term panel position's stability at the four cardinal angles, we expect that the positions at all other gantry angles are also highly reproducible, even though the images we analyzed were collected within only 1 month.

The pattern of the crosshair position variation at different gantry angles is panel and source-specific. As seen in Fig. 6, the Y-direction variation was about ± 2 mm for the G2-G90 panel and within ± 1 mm for the other three; in the X-Z direction wobbling varied from ± 2 to ± 4 mm. So the gantry angle dependent sag for each panel should be measured separately if such information is needed.

Although the image panels tend to sag several millimeters, the standard deviation of flex at a specific angle is small compared to the overall variations. This means that as long as the panel flex is correctly measured and accounted for, the panel position can be reliably calculated. With consistent panel and source sag at different gantry angles, it would be possible to develop gantry mounted x-ray volumetric imaging techniques, such as cone-beam CT (Ref. 11) or tomosynthesis¹³ that are available to linac photon treatment machines. With the volumetric imaging data, patient positioning uncertainty can be reduced as well as the planning margin, which in turn can potentially spare more surrounding normal tissue and reduce treatment toxicity.

This report details the experience with proton gantries from a single vendor at a single institution and the x-ray imaging system uses an inhouse designed panel motion control system. Systems using different configurations will have different experiences; however, our results may be applicable to similar proton gantry designs. Nonetheless, the mechanical characteristics of the imaging system are essential for development of accurate and efficient image-guidance techniques in proton and other particle radiotherapy systems.

5. CONCLUSION

We demonstrated the long-term position stability of the gantry mounted x-ray image panels by analyzing 2744 patient images acquired in two systems over a period of 13 months. The position variations at a specific gantry angle were less than 0.32 mm and inplane rotation variations were within 0.26° for the four panels. The image panel sag at different gantry angles was characterized and the patterns

were panel-specific, indicating the necessity to measure individual panels. Highly reproducible image panel positions indicate potential possibilities of onboard volumetric imaging techniques for gantry proton treatment systems.

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¹PTCOG, <http://www.ptcog.ch/index.php/facilities-in-operation>.

²M. Engelsman, A. Mazal, and D. A. Jaffray, "Patient positioning and set-up verification for planning and treatment," in *Proton and Charged Particle Radiotherapy*, edited by T. F. Delaney and H. M. Kooy (Lippincott, Philadelphia, PA, 2008), pp. 57–69.

³S. Devicienti, L. Strigari, M. D'Andrea, M. Benassi, V. Dimiccoli, and M. Portaluri, "Patient positioning in the proton radiotherapy era," *J. Exp. Clin. Cancer Res.* **29**(1), 47 (2010).

⁴J. Lehmann, J. Perks, S. Semon, R. Harse, and J. A. Purdy, "Commissioning experience with cone beam CT for image guided radiation therapy," *J. Appl. Clin. Med. Phys.* **8**(3), 21–36 (2007).

⁵W. Du, J. N. Yang, E. L. Chang, D. Luo, M. F. McAleer, A. Shiu, and M. Martel, "A quality assurance procedure to evaluate cone-beam CT image center congruence with the radiation isocenter of a linear accelerator," *J. Appl. Clin. Med. Phys.* **11**(4), 15–26 (2010).

⁶M. B. Sharpe, D. J. Moseley, T. G. Purdie, M. Islam, J. H. Siewerdsen, and D. A. Jaffray, "The stability of mechanical calibration for a kV cone beam computed tomography system integrated with linear accelerator," *Med. Phys.* **33**(1), 136–144 (2006).

⁷J.-P. Bissonnette, "Quality assurance of image-guidance technologies," *Semin. Radiat. Oncol.* **17**(4), 278–286 (2007).

⁸S. Yoo, G.-Y. Kim, R. Hammoud, E. Elder, T. Pawlicki, H. Guan, T. Fox, G. Luxton, F.-F. Yin, and P. Munro, "A quality assurance program for the on-board imager," *Med. Phys.* **33**(11), 4431–4447 (2006).

⁹W. Mao, L. Lee, and L. Xing, "Development of a QA phantom and automated analysis tool for geometric quality assurance of on-board MV and kV x-ray imaging systems," *Med. Phys.* **35**(4), 1497–1506 (2008).

¹⁰H. Guan, R. Hammoud, and F. Yin, "A positioning QA procedure for 2D/2D (kV/MV) and 3D/3D (CT/CBCT) image matching for radiotherapy patient setup," *J. Appl. Clin. Med. Phys.* **10**(4), 273–280 (2009).

¹¹D. A. Jaffray, J. H. Siewerdsen, J. W. Wong, and A. A. Martinez, "Flat-panel cone-beam computed tomography for image-guided radiation therapy," *Int. J. Radiat. Oncol., Biol., Phys.* **53**(5), 1337–1349 (2002).

¹²B. P. Selby, G. Sakas, S. Walter, and U. Stilla, "Geometry calibration for x-ray equipment in radiation treatment devices and estimation of remaining patient alignment errors," *Proc. SPIE* **6913**, 691355 (2008).

¹³D. J. Godfrey, F.-F. Yin, M. Oldham, S. Yoo, and C. Willett, "Digital tomosynthesis with an on-board kilovoltage imaging device," *Int. J. Radiat. Oncol., Biol. Phys.* **65**(1), 8–15 (2006).