Development of an x-ray-opaque-marker system for quantitative phantom positioning in patient-specific quality assurance

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Abstract

Purpose: We developed an x-ray-opaque-marker (XOM) system with inserted fiducial markers for patient-specific quality assurance (QA) in CyberKnife (Accuray) and a general-purpose linear accelerator (linac). The XOM system can be easily inserted or removed from the existing patient-specific QA phantom. Our study aimed to assess the utility of the XOM system by evaluating the recognition accuracy of the phantom position error and estimating the dose perturbation around a marker.
Methods: The recognition accuracy of the phantom position error was evaluated by comparing the known error values of the phantom position with the values measured by matching the images with target locating system (TLS; Accuray) and on-board imager (OBI; Varian). The dose perturbation was evaluated for 6 and 10 MV single-photon beams through experimental measurements and Monte Carlo simulations.
Results: The root mean squares (RMSs) of the residual position errors for the recognition accuracy evaluation in translations were 0.07 mm with TLS and 0.30 mm with OBI, and those in rotations were 0.13° with TLS and 0.15° with OBI. The dose perturbation was observed within 1.5 mm for 6 MV and 2.0 mm for 10 MV from the marker.
Conclusions: Sufficient recognition accuracy of the phantom position error was achieved using our system. It is unnecessary to consider the dose perturbation in actual patient-specific QA. We concluded that the XOM system can be utilized to ensure quantitative and accurate phantom positioning in patient-specific QA with CyberKnife and a general-purpose linac.

Key words: patient-specific QA, fiducial marker, CyberKnife, Monte Carlo simulation

1. Introduction

The CyberKnife Robotic Radiosurgery system (CK; Accuray Inc., Sunnyvale, CA, USA) is designed with a robotic manipulator and a lightweight small linear accelerator (linac). This enables realization of non-coplanar beams with sub-

millimeter accuracy and automated image-guided corrections of motion by the robot during a course of radiotherapy, achieving high conformal dose distribution to the tumor. In addition to the conventional fixed and Iris variable aperture collimators, the InCise2-multileaf collimator (MLC) has been implemented in the CK M6 system. The MLC has been reported to have advantages such as reduction in total monitor unit (MU) values and treatment time [1–3]. In addition, five tracking methods are supported by the CK M6 system with Precision treatment planning system (Accuray Inc., Sunnyvale, CA, USA): Fiducial Tracking, Synchrony Tracking, Xsight Lung Tracking, Xsight Spine Tracking, and 6D Skull Tracking.

As radiotherapy treatment delivery becomes more complex, quality assurance (QA) continues to play an increasingly critical role in ensuring the high quality and safety of patient treatments. In particular, it is necessary to perform patient-specific QA for every treatment plan in intensity-modulated radiation therapy and volumetric-modulated arc therapy, as these involve complex procedures that contain many potential errors such as the inverse planning and the MLC movement [4–6]. In addition, AAPM TG-135 recommends that pretreatment delivery QA in stereotactic irradiation should be performed for every treatment plan on a newly installed machine until the treatment team gets a good assessment of what level of accuracy can be achieved [7].

Patient-specific QA is conducted using simple rectangular or cylindrical phantoms with various detectors (e.g., an ionization chamber or radiochromic film) [8–10]. For the CK system design, the patient-specific QA in CK requires the tracking method used in the QA plan to be selected corresponding to the tracking method used in the patient treatment plan [11–13]. The same tracking method can be used for both the patient treatment plan and QA plan; in addition, Fiducial Tracking for the QA plan is compatible with all the tracking methods for patient treatment plans, as summarized in Table 1. When Fiducial Tracking is used in the QA plan, a phantom with attached or inserted fiducial markers must be prepared.

Table 1. Compatibility of tracking methods between patient treatment plan and QA plan in CK. This compatibility is determined by the CK system design and cannot be changed by the user. "Available" indicates that the tracking method corresponding to the one in the patient treatment plan is available in the QA plan. On the other hand, "Blank" indicates that the tracking method is not available in the QA plan.

		QA plan				
	Tracking method	Fiducial	Synchrony	Xsight Lung	Xsight Spine	6D Skull
Patient treatment plan	Fiducial	Available				
	Synchrony	Available	Available			
	Xsight Lung	Available		Available		
	Xsight Spine	Available			Available	
	6D Skull	Available				Available

The commercial and non-commercial phantoms with inserted fiducial markers for patient-specific QA are existing, i.e., Stereotactic Dose Verification Phantom (Standard Imaging, Middleton, WI, USA), StereoPHAN with CyberKnife film insert or universal spacer insert (Sun Nuclear Corp., Melbourne, FL, USA), and the spherical radiotherapy phantom for the measurements of mono-isocentric linac-based stereotactic radiosurgery plans reported by McKenna [14]. However, these phantoms are designed for the stereotactic irradiation, which limits the flexibility of the measurement (e.g., fixed placement of measurement devices, small film size, limited beam incident direction). To avoid those limitations and maintain consistency across multiple treatment systems or irradiation techniques, the existing flexible patient-specific QA phantom, I'mRT phantom (IBA Dosimetry, Schwarzenbruck, Germany), with the fiducial markers (X-SPOT; Beekley Medical, Connecticut, USA) attached on the phantom surface has been used for patient-specific QA using Fiducial Tracking in CK in our institution, as shown in Fig. 1 (a). However, this system has remained the following problems: (1) there is a possibility of the marker position on the surface getting shifted unintentionally, and (2) when we perform patient-specific QA using this phantom in a general-purpose linac, the fiducial markers attached on the phantom surface must be detached. This means that if this phantom is to be used again in CK, it must be redone from the Computed Tomography (CT) scanning of the phantom, which is an impractical workflow.

On the other hand, for patient-specific QA in a general-purpose linac, the conventional method of phantom positioning utilizes the alignment of the treatment room lasers with the lines on the phantom surface. This method can potentially cause positioning errors because of the subjective nature of the process, intra- and inter-operator variation, and human errors. Moreover, there is a residual systematic offset between the treatment room laser and the MV beam coordinate systems that cannot be eliminated, which may increase uncertainty for patient-specific QA.

In this study, we developed the x-ray-opaque-marker (XOM) system utilizing fiducial markers that can solve the above problems. The XOM system is designed to be compatible with phantom positioning methods such as CK and the general-purpose linac; moreover, it can be easily inserted or removed from the existing patient-specific QA phantom, as shown in Fig. 1 (b) and (c). For CK, this XOM system can be used for patient-specific QA using Fiducial Tracking, which is compatible with all tracking methods used in patient treatment plans. For general-purpose linac, the XOM system enables the quantification of phantom positioning accuracy by using an image guidance system. To the best of the authors' knowledge, there are no published studies that evaluate the patient-specific QA phantom positioning accuracy for using the phantom with inserted fiducial markers. The phantom positioning accuracy depends on a subject for imaging and image guidance system [15–17], and there is a possibility of obtaining an over- or underestimated dose owing to the dose perturbation from the fiducial markers. Thus, these characteristics must be evaluated before clinical use. This study aimed to assess the clinical utility of our developed XOM system by (1) evaluating the accuracy of the phantom positioning for the XOM system in CK and a general-purpose linac and by (2) estimating the dose perturbation around a fiducial marker through experimental measurements and Monte Carlo (MC) simulations, when irradiated by 6 and 10 MV single-photon beams.

2. Materials and Methods

2.1. Design of XOM system

Seven tungsten carbide spherical markers (AS ONE Corp., Osaka, Japan; 1.0 mm diameter; physical density $\rho = 15.0$ g cm^{-3}; 94% WC; 6% Co) as fiducial markers were inserted in the RW3 solid water plate (IBA Dosimetry, Schwarzenbruck, Germany; 160 mm × 160 mm × 10 mm; $\rho = 1.045$ g cm^{-3}; 98% Polystyrene; 2% TiO_2), as shown in Fig. 2. The position of the seven fiducial markers were logically arranged to be recognized by the target locating system (TLS, Accuray Inc., Sunnyvale, CA, USA; image resolution: 0.28 mm pixel^{-1}) and the on-board imager (OBI, Varian Medical Systems, Palo Alto, CA, USA; image resolution: 0.19 mm pixel^{-1}) image guidance systems that are installed in CK and Clinac iX (Varian Medical Systems, Palo Alto, CA, USA), respectively. The TLS consists of two orthogonal x-ray sources mounted on the ceiling, which make a 45° angle with the plane of the floor, and two flat panel detectors fixed to the floor. The requirements with regard to the position of the seven fiducial markers are as follows: for use with TLS, the markers do not overlap each other when viewed from the direction of 45°, the distance between the markers is ≥20 mm, and the angle between them is >15° [18]. In addition, for use with OBI, the markers do not overlap each other when viewed from the directions of 0° and 90°. The number of inserted fiducial markers were determined to recognize three or more markers on the TLS images without depending on the position of the imaging center (also referred to as "align center") in CK.

2.2. Evaluation of recognition accuracy of the phantom position error for the XOM system

2.2.1. CT scan of the phantom inserted with XOM system

As shown in Fig. 1 (b), our developed XOM system was positioned at the 10th (from the top) of the sixteen RW3 phantom plates in the I'mRT phantom.

This phantom was scanned by the SOMATOM Confidence RT Pro 64 slice CT system (Siemens Healthineers, Forchheim, Germany). The scanning parameters were 120 kV tube voltage, 300 mm field of view, and 1.0 mm slice thickness. The iterative metal artifact reduction (iMAR) algorithm (Siemens Healthineers, Forchheim, Germany) was used for image reconstruction. At that time, the phantom was intentionally rotated by each known rotational error value to ensure that the recognition accuracy of the phantom position error can be evaluated for various rotations. A digital protractor (MJ-PRO; Sato Shouji Inc., Kanagawa, Japan; detection accuracy: <0.10°) and a lok-bar (HM-bar; Nichigen Co. Ltd., Tokyo, Japan) [19] were used in the yaw rotation, and a digital inclinometer (DP-30XY; Niigata Seiki Co. Ltd., Niigata, Japan; detection accuracy: <0.05°) was used in the roll and pitch rotations. More specifically, the known rotational error values of yaw, roll, and pitch were selected from 0.0°, 1.0°, 2.0°, 3.0°, 4.0°, and 5.0°, where the CT image rotated 0.0° in yaw, roll, and pitch was defined as the reference CT image.

2.2.2. Creation of test treatment plans for image matching

Precision treatment planning system (Ver. 1.1.1.1) and Eclipse treatment planning system (Ver. 13.6; Varian Medical Systems, Palo Alto, CA, USA) were used to design the test treatment plans. The reference CT image in Section 2.2.1 was shifted by each known translational error value to ensure that the recognition accuracy of the phantom position error can be evaluated for various translations. More specifically, the known translational error values of left-right (LR), superior-inferior (SI), and anterior-posterior (AP) were selected from 0.0 mm, 0.5 mm, 1.0 mm, 3.0 mm, 5.0 mm, and 10.0 mm, where the digital reconstructed radiograph (DRR) created from the reference CT image was defined as the reference DRR. For the treatment planning process for CK, Fiducial Tracking was selected as the phantom tracking method.

2.2.3. Measurements of residual position error for TLS

The I'mRT phantom inserted with the XOM system on the couch was initially imaged with two oblique x-ray images using the TLS. The imaging parameters were 135 kVp tube voltage, 200 mA tube current, and 160 ms exposure time, which were the suitable conditions for the visual confirmation and recognition of the fiducial markers by the operators and image guidance system. Subsequently, the TLS images and the reference DRRs were matched with reference to the fiducial markers automatically, and then the phantom position was adjusted by couch translation and rotation according to the matching results. This adjusted position of the phantom was defined as the reference position. Next, the x-ray images with the TLS were acquired at the reference position again. For evaluation of the recognition accuracy of the phantom position error for the XOM system, these TLS images were subsequently matched with the DRRs with the known translational or rotational error values described in Section 2.2.1 and Section 2.2.2. The residual phantom position error was estimated. The above experiments were repeated five times.

2.2.4. Measurements of residual position error for OBI

The I'mRT phantom inserted with the XOM system on the couch was initially imaged with two x-ray images of gantry angles 0° and 90° using the OBI. The imaging parameters were 135 kVp tube voltage, 200 mA tube current, and 160 ms exposure time, which were the same as those mentioned in Section 2.2.3. Subsequently, the OBI images and the reference DRRs were matched with reference to the fiducial markers manually, and then the phantom position was adjusted by the couch translation and rotation (except roll) according to the matching results. This adjusted position of the phantom was defined as the reference position. Next, the x-ray images with the OBI were acquired at the reference position again. These OBI images were subsequently matched with the DRRs with the known translational or rotational error values described in Section 2.2.1 and Section 2.2.2. The residual phantom position error was estimated. The above experiments were repeated five times.

2.3. Evaluation of recognition accuracy of the phantom position error for another commercial phantom

To compare the recognition accuracy of the phantom position error for the XOM system with that for another commercial phantom, the StereoPHAN with CyberKnife film insert for TLS and universal spacer insert for OBI was used. Four gold markers and five aluminum markers are inserted in CyberKnife film insert and universal spacer insert, respectively. For evaluation of the recognition accuracy of the phantom position error for the StereoPHAN with each insert, the residual phantom position errors were estimated in the same way as in Section 2.2.

2.4. Evaluation of phantom positioning accuracy for XOM system with TLS or OBI

The use of tracking method by the TLS is essential for the phantom positioning for CK. For evaluating the phantom positioning accuracy for the XOM system with TLS, the I'mRT phantom inserted with the XOM system settled on the couch was initially imaged with two oblique x-ray images using the TLS. The imaging parameters were the same as those mentioned in Section 2.2.3. Subsequently, the TLS images and the reference DRRs were matched with reference to the fiducial markers automatically, and then the phantom position was adjusted by the couch according to the matching results. Finally, the residual phantom positioning errors for the XOM system with the TLS were estimated by matching the TLS images acquired at this position again with the reference DRRs. The above experiments were repeated five times.

For evaluating the phantom positioning accuracy for the XOM system with the OBI, the I'mRT phantom inserted with the XOM system was settled on the couch and aligned based on the visible lines on the phantom surface using the treatment room

lasers. Subsequently, two x-ray images of gantry angles 0° and 90° using the OBI were acquired. The imaging parameters were the same as those mentioned in Section 2.2.4. The residual phantom positioning errors for the treatment room lasers were estimated by matching these OBI images with the reference DRRs with reference to the fiducial markers manually. After that, the phantom position was adjusted according to the matching results. This adjustment was performed when the residual phantom positioning error for the treatment room lasers was larger than 2 standard deviations (SDs) for the recognition accuracy evaluation measured in Section 2.2.4. Finally, the residual phantom positioning errors for the XOM system with the OBI were estimated by matching the OBI images acquired at this position again with the reference DRRs. The above experiments were repeated five times by four independent operators (one medical physicist and three radiation technologists). All operators were familiar with and proficient in the performance of the procedure.

2.5. Evaluation of phantom positioning accuracy for another commercial phantom with TLS or OBI

To compare the phantom positioning accuracy for the XOM system with that for another commercial phantom, the StereoPHAN with CyberKnife film insert for TLS and universal spacer insert for OBI was used. For evaluation of the phantom positioning accuracy for the StereoPHAN with each insert, the residual phantom positioning errors were estimated in the same way as in Section 2.4.

2.6. Evaluation of dose perturbation around a fiducial marker

2.6.1. Experimental evaluation

The measurements in this study were based on the use of 6 and 10 MV single-photon beams produced by the Clinac iX. Fig. 3 shows a schematic of the experimental setup. The geometry was created using the RW3 solid water plates, a polymethyl methacrylate (PMMA) plate, and a tungsten carbide spherical marker of 1.0 mm in diameter. The tungsten carbide spherical marker was positioned in the PMMA plate, which was located at a depth of 10 cm from the phantom surface. The dose perturbation around the marker was measured by the Gafchromic EBT3 films (Ashland ISP Advanced Materials, NJ, USA). The films were set to a depth of 2.0 cm, 5.0 cm, 9.0 cm, 10.0 cm, 10.1 cm, and 11.1 cm in the center of the phantom. The phantoms shown in Fig. 3 were irradiated with a field size of 10.0 × 10.0 cm^2, a source-to-surface distance of 91.0 cm, and 250 MU.

A dose calibration curve of the Gafchromic EBT3 film was created, ranging from 0 to 4.0 Gy. The films were placed perpendicular to the central beam axis in the Solid Water HE slab phantoms (Sun Nuclear Corp., Melbourne, FL, USA) at a depth of 10.0 cm and irradiated with a field size of 10 × 10 cm^2. The exposed films were scanned at 24 h post-irradiation using an Epson DS G20000 scanner (Epson Corp., Nagano, Japan) to produce 48-bit RGB color images of 100 dots-per-inch resolution [20,21]. The dose calibration curve was created by the film analysis software (DD-System; R-TECH Inc., Tokyo, Japan) and the analyses of the film measurements were performed utilizing the same software.

2.6.2. MC simulations

The particle and heavy ion transport system (PHITS) code version 3.20 [22,23] was used to simulate 6 and 10 MV photon beams delivered by the Clinac iX. The detailed geometry of the Clinac iX head was provided by the vendor. PHITS version 3.20 optionally implements the Electron Gamma Shower version 5 MC code [24], and it was selected in all of the simulations in this study. The other calculation parameters were as follows: the electron cut-off energy was set to 0.521 MeV; the photon cut-off energy was set to 0.010 MeV [25,26]. The field size was set to 10.0 × 10.0 cm^2. The numbers of source electrons were determined such that the statistical uncertainty [22,23] would be almost within 1% for all of the relative dose profiles in the region between the phantom surface and 30 cm in depth.

The MC model was tuned by matching the measured and simulated percentage depth doses (PDDs) and the transversal dose profiles (TDPs) at depths of 1.4 cm, 10 cm, and 20 cm for 6 MV and 2.3 cm, 10 cm, and 20 cm for 10 MV in the water phantom, whilst varying the initial spot electron energy [27]. For the MC simulation, the dose scoring grid (cross-plane × in-plane × depth) for the PDD was set to 12 mm × 12 mm × 1 mm, and that for the TDP was set to 1 mm × 25 mm × 5 mm. The mean energy and full width at half-maximum (FWHM) of the energy distribution and the radial intensity spread of the incident electron beam were optimized and they were expressed as Gaussian distributions. The corresponding measured data were obtained using a BEAMSCAN 3D water phantom (PTW, Freiburg, Germany) and the PTW Semiflex 31010 ion chamber (PTW,

Freiburg, Germany). In this study, the simulated PDD and TDPs were compared with the measured profiles using the gamma index metric [28]. Gamma index criteria used in this study were: dose difference of 2% and distance to agreement of 2 mm. Gamma index values which were ≤1 were defined as in agreement with the measured and simulated dose distribution in the water phantom and the gamma index passing rate was determined.

For evaluation of the dose perturbation around a tungsten carbide spherical marker, the geometries described in Section 2.6.1 were reproduced in the MC simulations, and the relative dose profiles along the beam axis were calculated. The dose scoring grid was set to 0.5 mm × 0.5 mm × 0.2 mm. The simulated profiles were compared with the measured values from the Gafchromic EBT3 films.

3. Results

3.1. Evaluation of recognition accuracy of the phantom position error for the XOM system

Examples of the kV x-ray images of the phantom including the XOM system provided by TLS and OBI, which were matched with the reference DRRs, are shown in Fig. 4. Fig. 5 compares the residual position errors for the XOM system measured by matching the images with the TLS or the OBI for each known translational or rotational error value. Table 2 presents the root mean squares (RMSs), SDs, and maximum values of the residual position errors. The experiments showed that the RMSs of the residual position errors in translations and rotations with the TLS were 0.07 mm and 0.13°, and the SDs were 0.07 mm and 0.13°, respectively. The RMSs of the residual position errors in translations and rotations with the OBI were 0.30 mm and 0.15°, and the SDs were 0.29 mm and 0.15°, respectively. More specifically, the RMSs of the residual position errors in translations with the TLS were 0.08 mm in LR, 0.06 mm in SI, and 0.08 mm in AP. The corresponding RMSs of the residual position errors in rotations were 0.14° in yaw, 0.11° in roll, and 0.12° in pitch. The RMSs of the residual position errors in translations with the OBI were 0.22 mm in LR, 0.30 mm in SI, and 0.36 mm in AP. The corresponding RMSs of the residual position errors in rotations were 0.14° in yaw and 0.15° in pitch.

Table 2. Residual position errors for the XOM system measured by matching the images with the TLS or the OBI.

	Translation (mm)			Rotation (deg)		
	LR	SI	AP	Yaw	Roll	Pitch
TLS in CK						
RMS	0.08	0.06	0.08	0.14	0.11	0.12
SD	0.08	0.05	0.08	0.10	0.10	0.13
Maximum value	-0.20	-0.10	0.20	-0.30	±0.20	0.30
OBI in Clinac iX						
RMS	0.22	0.30	0.36	0.14		0.15
SD	0.20	0.30	0.32	0.11		0.14
Maximum value	-0.50	1.00	-1.00	-0.20		0.30

Abbreviations: TLS = target locating system; OBI = on-board imager; CK = CyberKnife; RMS = root mean square; SD = standard deviation; LR = left-right; SI = superior-inferior; AP = anterior-posterior; Yaw = rotation around AP direction; Roll = rotation around SI direction; Pitch = rotation around LR direction.

3.2. Evaluation of recognition accuracy of the phantom position error for another commercial phantom

Table 3 presents the RMSs, SDs, and maximum values of the residual position errors for the StereoPHAN with CyberKnife film insert or universal spacer insert measured by matching the images with the TLS or the OBI for each known translational or rotational error value. The experiments showed that the RMSs of the residual position errors in translations and rotations with the TLS were 0.10 mm and 0.20°, and the SDs were 0.10 mm and 0.20°, respectively. The RMSs of the residual position errors in translations and rotations with the OBI were 0.22 mm and 0.15°, and the SDs were 0.22 mm and 0.10°, respectively. More specifically, the RMSs of the residual position errors in translations with the TLS were 0.11 mm in LR, 0.09 mm in SI, and 0.10 mm in AP. The corresponding RMSs of the residual position errors in rotations were 0.17° in yaw, 0.21° in roll, and

0.22° in pitch. The RMSs of the residual position errors in translations with the OBI were 0.22 mm in LR, 0.22 mm in SI, and 0.22 mm in AP. The corresponding RMSs of the residual position errors in rotations were 0.08° in yaw and 0.19° in pitch.

Table 3. Residual position errors for the StereoPHAN with CyberKnife film insert or universal spacer insert measured by matching the images with the TLS or the OBI.

	Translation (mm)			Rotation (deg)		
	LR	SI	AP	Yaw	Roll	Pitch
TLS in CK						
RMS	0.11	0.09	0.10	0.17	0.21	0.22
SD	0.10	0.09	0.10	0.14	0.13	0.16
Maximum value	±0.20	0.20	0.20	0.30	0.40	-0.50
OBI in Clinac iX						
RMS	0.22	0.22	0.22	0.08		0.19
SD	0.23	0.22	0.23	0.07		0.07
Maximum value	±0.50	±0.50	±0.50	-0.20		-0.30

Abbreviations: TLS = target locating system; OBI = on-board imager; CK = CyberKnife; RMS = root mean square; SD = standard deviation; LR = left-right; SI = superior-inferior; AP = anterior-posterior; Yaw = rotation around AP direction; Roll = rotation around SI direction; Pitch = rotation around LR direction.

3.3. Evaluation of phantom positioning accuracy for XOM system with TLS or OBI

The mean residual phantom positioning errors in translations for the XOM system with the TLS were -0.0 ± 0.1 mm (1 SD) in LR, -0.0 ± 0.1 mm in SI, and 0.0 ± 0.1 mm in AP. The corresponding mean residual phantom positioning errors in rotations were 0.0 ± 0.1° in yaw, -0.0 ± 0.1° in roll, and -0.1 ± 0.1° in pitch.

The mean residual phantom positioning errors in translations for the treatment room lasers in the general-purpose linac were 0 ± 0 mm in LR, 0.2 ± 0.4 mm in SI, and 0 ± 0 mm in AP. The corresponding mean residual phantom positioning errors in rotations were -0.2 ± 0.0° in yaw and 0.2 ± 0.0° in pitch. The mean residual phantom positioning errors in all translations for the XOM system with the OBI were 0 ± 0 mm. The corresponding mean residual phantom positioning errors in rotations remained the same because the results of the residual phantom positioning errors in rotations for the treatment room lasers were within 2 SDs for the recognition accuracy evaluation.

3.4. Evaluation of phantom positioning accuracy for another commercial phantom with TLS or OBI

The mean residual phantom positioning errors in translations for the StereoPHAN with CyberKnife film insert with the TLS were 0.0 ± 0.3 mm (1 SD) in LR, 0.1 ± 0.1 mm in SI, and 0.1 ± 0.1 mm in AP. The corresponding mean residual phantom positioning errors in rotations were 0.0 ± 0.1° in yaw, -0.1 ± 0.2° in roll, and -0.0 ± 0.1° in pitch.

The mean residual phantom positioning errors in all translations for the StereoPHAN with universal spacer insert with the OBI were 0 ± 0 mm. The corresponding mean residual phantom positioning errors in all rotations were 0 ± 0 mm.

3.5. Evaluation of dose perturbation around a fiducial marker

Fig. 6 shows comparisons of the measured and simulated profiles in the water phantom. The source parameters were determined by matching the measured PDDs and TDPs, and the results indicated a mean energy of 6.2 and 10.3 MeV with 3% FWHM and a radial intensity spread with FWHM of 2.0 mm for the nominal energy of the 6 and 10 MV beams, respectively. The gamma index passing rate was found to be 100% for all PDDs and TDPs for 6 and 10 MV.

Fig. 7 shows the measured and simulated relative dose profiles obtained with and without a tungsten carbide spherical marker for a single-photon beam, which yielded mean differences between the measured and simulated values of 0.32 ± 1.77% (1 SD) for 6 MV and 0.83 ± 1.20% for 10 MV along the beam axis. In this study, relative dose profiles were normalized to the dose at a depth of 5.0 cm because it was under conditions of secondary electron equilibrium and was less affected by the

scattered radiation from the marker. According to the results of the MC simulations along the beam axis, the entrance dose at the tungsten carbide spherical marker-phantom interface increased by 17.2% for 6 MV and 17.0% for 10 MV, while the exit dose at the tungsten carbide spherical marker-phantom interface decreased by 14.6% for 6 MV and 14.2% for 10 MV, which were compared to the case without the tungsten carbide spherical marker in the RW3 solid water phantom. The dose enhancement at the upstream region was observed within approximately 0.4 mm for 6 MV and 0.6 mm for 10 MV from the marker, while the dose reduction at the downstream region occurred within approximately 1.5 mm for 6 MV and 2.0 mm for 10 MV from the marker.

4. Discussion

To assess the clinical utility of our developed XOM system with inserted tungsten carbide spherical markers in CK and a general-purpose linac, we evaluated the system based on two criteria: phantom positioning accuracy and dose perturbation around a marker. The phantom positioning accuracy was quantified using the I'mRT phantom inserted with the XOM system with the TLS or the OBI. The dose perturbation around a marker was evaluated for a 6 and 10 MV single-photon beam produced by the Clinac iX. Yan et al. [29] reported that the CT slice thickness had a non-negligible effect on the phantom positioning accuracy. As they stated that a thin CT slice thickness enhances the phantom positioning accuracy, a CT slice thickness of 1.0 mm was used for the test treatment plans in this study.

For evaluation of the recognition accuracy of the phantom position error for the ExacTrac X-ray 6D system (BrainLAB AG, Feldkirchen, Germany) and the cone-beam CT, Chang et al. [30] reported that the residual position errors were measured using an anthropomorphic phantom containing a lumbar spine structure. They reported that the RMSs of the residual position errors in translations and rotations with the ExacTrac X-ray 6D system were less than 1.3 mm and 0.6°, respectively. In contrast, the RMSs of the residual position errors in translations and rotations with the cone-beam CT were less than 0.8 mm and 0.2°, respectively. The results of this study demonstrated that the feasibility of superior recognition accuracy of the phantom position error by using the XOM system, compared with the results of Chang et al. [30].

The comparison of the residual position errors with the TLS showed that the residual position errors with the OBI were larger in translations, even though the image resolution of the OBI was smaller than that of the TLS. The residual position errors in rotations showed similar results. This is because different number of significant digits are used in the image matching software for TLS and the OBI [30]. More specifically, the units of millimeters and degrees are used for TLS, while centimeters and degrees are used for OBI, all of which display values to the first decimal place. Thus, the residual position errors with the OBI became particularly large for the known translational error values of 0.5 mm. In contrast, the effect of the image resolution was negligible on the residual position errors between the two image guidance systems in this study.

The RMSs of the residual position errors for the XOM system with the TLS and the OBI were almost the same as that for the StereoPHAN with CyberKnife film insert for TLS and universal spacer insert for OBI. Similarly, the residual phantom positioning errors for the XOM system with the TLS and the OBI were almost the same as that for the StereoPHAN with each insert. These results demonstrated that the feasibility of sufficient accuracy of phantom positioning by using the XOM system, compared with using the existing commercial phantom with inserted the fiducial markers.

For CK in our institution, the residual systematic offsets between the TLS and the robotic manipulator coordinate systems, also referred to as "DeltaMan", were -0.3 mm in LR, -0.3 mm in SI, and 0.6 mm in AP, which were derived from a sequence of end-to-end (E2E) tests using radiochromic films. For Clinac iX in our institution, the residual systematic offsets between the treatment room laser and the MV beam coordinate systems were -0.2 mm in LR, 0.1 mm in SI, and -0.4 in AP, which were derived from a sequence of Winston-Lutz tests. Similarly, those between the treatment room laser and the OBI were -0.2 mm in LR, 0.4 mm in SI, and 0.1 mm in AP. Those between the OBI and the MV beam were 0.0 mm in LR, 0.2 mm in SI, and 0.5 mm in AP. At the moment, these residual systematic offsets between the coordinate systems cannot be eliminated, and they can potentially affect the results of dose verification in patient-specific QA. In this regard, the phantom positioning with the treatment room lasers in a general-purpose linac such as Clinac iX needs consideration of the very complex spatial coordinate systems due to the involvement of room lasers. Thus, it is possible to construct the dose verification system, including end-to-end geometric verification, by performing patient-specific QA using the XOM system with the image guidance system used for patient positioning.

In this study, it was shown that almost the same or even a better accuracy of phantom positioning can be achieved by using the TLS and OBI, compared to the recognition accuracy of the phantom position error for the XOM system with the TLS and OBI. The residual phantom positioning errors in the actual patient-specific QA should be as small as possible. However, the

correction accuracy of the phantom position is limited by the recognition accuracy of the TLS or OBI and the adjustment accuracy of the phantom by either moving phantom manually or with remote couch shift. Thus, the tolerance limits of the phantom positioning accuracy in clinical situations should be determined at each facility. It was shown that sufficient accuracy of phantom positioning can be achieved by using the treatment room lasers in a general-purpose linac. However, what is important is that the XOM system makes it possible to perform dose verification after the phantom positioning accuracy is checked, which helps minimize intra- and inter-operator variation as well as human errors. On the other hand, because TrueBeam (Varian Medical Systems, Palo Alto, CA, USA), the latest generation of Varian linacs, has one more significant digit for the position verification in translations compared with Clinac iX used in this study, there is a feasibility of superior phantom positioning accuracy by using the XOM system in TrueBeam. In addition, the XOM system will be useful for performing end-to-end test that evaluates the clinical workflow from the CT imaging to the delivery of the treatment plan, especially for a newly installed treatment system.

To evaluate the dose perturbation around a tungsten carbide spherical marker, measurements and MC simulations were performed in this study. The measured and simulated values were in agreement, with the mean difference being 0.32 ± 1.77% for 6 MV and 0.83 ± 1.20% for 10 MV. These results indicate that the MC simulations truly reflect the actual behavior of the radiation scattered and transmitted by the tungsten carbide spherical marker.

For a single-photon beam, some researchers have reported that the dose enhancement was generally observed in the upstream region of high-density material, whereas the dose reduction was found in the downstream region of that material .[25,31–33]. Our results are consistent with these previous works. The dose enhancement is caused by the backscatter radiation from the tungsten carbide spherical marker. On the other hand, the dose reduction is largely due to the decrease in forward-directed secondary electrons generated from the tungsten carbide spherical marker, and it is more rapid than the buildup of secondary electrons generated in the RW3 solid water phantom. Chow and Grigorov [34] have represented the dose enhancement and dose reduction information around a gold seed in the water phantom by performing a MC simulation, for irradiation by a 6 MV single-photon beam. The size of the gold seed used in their work was 1.2 mm × 1.2 mm × 3.2 mm. The relative dose ranged from 88% to 164% at the edge between the gold seed and the water. In general, the dose perturbation from high-density material depends on the effective atomic number, size, and shape of the material [31,35]. Our study demonstrated a similar tendency; however, the magnitude of the dose enhancement decreased compared with the work of Chow and Grigorov [34]. This is because the effective atomic number and size of the tungsten carbide spherical marker used in our study were smaller than those of the gold seed used in their work.

In this study, the dose perturbation along the beam axis was observed within approximately 1.5 mm and 2.0 mm from the marker for a 6 and 10 MV single-photon beam, respectively. The results showed that the range of influence of dose perturbation from the marker increased as the photon energy increased. This is because the mean energy of the radiation generated from the marker for 10 MV photon beam is higher than that for 6 MV photon beam. These results are consistent with the results of Shimozato et al. [35] and Kawahara et al. [36]. Similarly, when irradiated by a lower-energy photon beam (i.e., 4 MV), the range of influence of the dose perturbation along the beam axis is expected to be shorter than 1.5 mm. Furthermore, the use of a flattening filter free beam could mitigate the dose perturbation from high-density materials [36]. This is because the energy spectrum becomes softer and the mean energy of photon energy spectrum is decreased by removing the flattening filter [37].

Because the distance between the markers and the surface of the XOM system is 1.5 mm or more, it is unnecessary to consider the dose perturbation from the markers when we perform patient-specific QA for 6 MV or lower-energy beam. On the other hand, for 10 MV, slight dose perturbation was observed outside the surface of the XOM system; thus, it is recommended to keep a distance at least 0.5 mm between the surface of the XOM system and the detectors when using a 10 MV single beam. However, in previous works [31,38], it was reported that the use of multiple beams or wide gantry rotation angles could mitigate the dose perturbation from high-density materials, compared with the use of a single beam. In the actual patient-specific QA using our XOM system, because multiple beams or wide gantry rotation angles are used, the influence of dose perturbation from the markers can be neglected.

In this study, we only evaluated the tungsten carbide spherical marker with a diameter of 1.0 mm. The small marker with a diameter of 1.0 mm made of tungsten carbide was selected on the basis of the artifact from the marker on the CT images, the image resolution of the TLS and the OBI, the dose perturbation around the markers, and the cost and ease of availability. In addition, the reason why the spherical marker was selected is because the dose perturbation does not depend on the orientation of the marker and the direction of the incident beam [26]. As mentioned above, sufficient accuracy of phantom positioning was shown by using the XOM system under static conditions. However, it is possible that a larger marker (e.g., 1.5 mm or 2.0 mm

in diameter) is preferable when the phantom under dynamic conditions is used for the QA of the tumor-tracking radiotherapy. The characteristics of the dose perturbation from the marker can be changed by changing the size of the marker. Thus, the evaluation of those characteristics under dynamic conditions is an issue that should be investigated further.

5. Conclusion

We developed an XOM system with inserted tungsten carbide spherical markers. The results of this study demonstrated that (1) the XOM system enables accurate phantom positioning in patient-specific QA with CK and a general-purpose linac, and (2) the dose perturbation from the markers is negligible in actual patient-specific QA. The XOM system can be utilized to ensure accurate and quantitative phantom positioning in patient-specific QA with CyberKnife and a general-purpose linac.

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Conflict of interest

The authors have no relevant conflicts of interest to disclose.

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Figure captions

Fig. 1. (a) Patient-specific QA phantom with fiducial markers attached on the surface; this phantom has been used in our institution for CK. (b) Patient-specific QA phantom including our developed XOM system. (c) Sample of our developed XOM system with seven tungsten carbide spherical markers.
Fig. 2. Schematic of the arrangement of seven tungsten carbide spherical markers as fiducial markers inserted in the RW3 solid water plate: (a) coronal, (b) axial, and (c) sagittal views. *Abbreviations*: LR = left-right; SI = superior-inferior; AP = anterior-posterior.
Fig. 3. Experimental setup for the measurement of the dose perturbation around a marker (a) without and (b) with a tungsten carbide spherical marker. The field size was set to 10.0 × 10.0 cm^2. *Abbreviations*: SSD = source-to-surface distance; PMMA = polymethyl methacrylate.
Fig. 4. Examples of the kV x-ray images of the phantom including the XOM system provided by (a), (b) TLS and (c), (d) OBI which were matched with the reference DRRs.
Fig. 5. Residual position errors for the XOM system measured by matching the images with the TLS or the OBI in (a) LR, (b) SI, (c) AP, (d) yaw, (e) roll, and (f) pitch, respectively.
Fig. 6. Measured and MC simulated (a) PDD for 6 MV, (b) TDPs at a depth of 1.4 cm, 10 cm, and 20 cm for 6 MV, (c) PDD for 10 MV, and (d) TDPs at a depth of 2.3 cm, 10 cm, and 20 cm for 10 MV in the water phantom. The gamma index passing rate was 100% for all PDDs and TDPs for 6 and 10 MV using 2%/2 mm criteria.
Fig. 7. ((a) and (c)) Relative dose profiles with and without the tungsten carbide spherical marker and ((b) and (d)) their enlarged images for a 6 and 10 MV single-photon beam along the beam axis, respectively. The profiles were normalized to the dose at a depth of 5.0 cm.