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An optically stimulated luminescence dosimeter for measuring patient exposure from imaging guidance procedures

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Received 9 May 2013, in final form 20 June 2013
Published 6 August 2013
Online at stacks.iop.org/PMB/58/5885

Abstract

There is a growing interest in patient exposure resulting from an x-ray imaging procedure used in image-guided radiation therapy. This study explores a feasibility to use a commercially available optically stimulated luminescence (OSL) dosimeter, nanoDot, for estimating imaging radiation exposure to patients. The kilovoltage x-ray sources used for kV-cone-beam CT (CBCT) imaging acquisition procedures were from a Varian on-board imager (OBI) image system. An ionization chamber was used to determine the energy response of nanoDot dosimeters. The chamber calibration factors for x-ray beam quality specified by half-value layer were obtained from an Accredited Dosimetry Calibration Laboratory. The Monte Carlo calculated dose distributions were used to validate the dose distributions measured by using the nanoDot dosimeters in phantom and \textit{in vivo}. The range of the energy correction factors for the nanoDot as a function of photon energy and bow-tie filters was found to be 0.88–1.13 for different kVp and bow-tie filters. Measurement uncertainties of nanoDot were approximately 2–4% after applying the energy correction factors. The tests of nanoDot placed on a RANDO phantom and on patient’s skin showed consistent results. The nanoDot is suitable dosimeter for \textit{in vivo} dosimetry due to its small size and manageable energy dependence. The dosimeter placed on a patient’s skin has potential to serve as an experimental method to monitor and to estimate patient exposure resulting from a kilovoltage x-ray imaging procedure. Due to its large variation in energy response, nanoDot is not suitable to measure radiation doses resulting from mixed beams of megavoltage therapeutic and kilovoltage imaging radiations.

(Some figures may appear in colour only in the online journal)

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1. Introduction

Image-guided radiation therapy (IGRT) plays an increasingly important role in the accurate delivery of highly conformal dose to the target as it makes it possible to see the target and organs at risk before treatment while the patient is in the treatment position. X-ray imaging devices, especially kilovoltage (kV) x-ray imaging devices integrated in a radiation treatment delivery unit, are increasingly available in the clinics. There is a growing interest in determining and documenting patient exposure from imaging procedures used in radiotherapy (Ding and Munro 2011, Ding et al 2010, Walters et al 2009, Spezi et al 2009, Jeng et al 2009, Downes et al 2009, Ding and Coffey 2009, Faddegon et al 2008, Chow et al 2008, Wen et al 2007). Although the Monte Carlo simulation is capable of calculating the dose distributions, it is not routinely available in clinics to calculate dose distributions. The experimental measurements remain the gold standards to validate the accuracy of calculations. A small and easy-to-use dosimeter for in vivo radiation measurements may be an alternative in monitoring imaging dose to patients resulting from repeated x-ray image guidance procedures for patient positioning during IGRT.

Recently a commercial optically stimulated luminescence (OSL) dosimetry system was available for medical and personal dosimetry (Jursinic 2007, 2010, Oster et al 2008, Schembri and Heijmen 2007, Viamonte et al 2008, Yukihara et al 2008, Yukihara and McKeever 2008). A number of studies have been published that describe the use of OSL dosimeters in medicine for radiotherapy therapeutic beams (Schembri and Heijmen 2007, Jursinic 2007, Yukihara and McKeever 2008, Jursinic 2010, Mrcela et al 2011, Reft 2009, Kerns et al 2011) and for low energy x-rays (Al-Senan and Hatab 2011, Yukihara et al 2009). The operation principle of using carbon-doped aluminum oxide (Al2O3:C) has been studied and described in detail (Jursinic 2010, Yukihara et al 2008, Yukihara and McKeever 2008). Although the characteristics including the dose response and energy dependence have been studied for radiotherapy megavoltage photons/electrons and diagnostic kilovoltage photons (Al-Senan and Hatab 2011, Jursinic 2007, Mrcela et al 2011, Reft 2009), there are limited studies on its clinical application to in vivo measurements, especially in estimating the radiation exposure to patients resulting from an image guidance procedure routinely used during IGRT.

The purpose of this work is (a) to investigate the accuracy of using an OSL dosimeter, nanoDot (Landauer, Inc., Glenwood, IL), for measuring radiation doses resulting from kV x-rays commonly used during IGRT and (b) to study feasibility of estimating organ doses by placing the dosimeter on a patient’s skin for in vivo measurements.

2. Methods and materials

2.1. The optically stimulated luminescence dosimetry system

The nanoDot dosimeters used in this study are read out with a MicroStar® reader. It is a portable commercially available system (Jursinic 2010, Kerns et al 2011, Reft 2009) which includes a MicroStar reader, an external laptop and nanoDot dosimeters. The sensitive diameter of the nanoDot detector is a 5 mm, 0.2 mm thick plastic disk infused with aluminum oxide doped with carbon (Al2O3:C). This disk is encased in a 10 mm × 10 mm × 2 mm light-tight plastic holder. It uses aluminum oxide (Al2O3:C) detectors to measure radiation exposure and reads the measurement with OSL technology (Viamonte et al 2008, Yukihara et al 2008, Yukihara and McKeever 2008). There are two types of factory stated accuracies available: 5% with standard nanoDot and 2% with more consistence reproducibility called ‘screened nanoDot’. The almost nondestructive read-out allows for dose verification and intermittent analysis for
total dose accumulation. This capability also makes it possible to measure accumulated total imaging dose to patient resulting from repeated imaging guidance procedures during the entire radiotherapy treatment course. The nanoDot dosimeter requires 8–10 min of stabilization after exposure before read-out. The stated energy dependence is within ±10% over the diagnostic energy range and within ±2% for photons and electrons from 5 to 20 MeV based on manufacture calibration. The nanoDot dosimeters used in this study were screened nanoDot dosimeters and the batch also included a calibration kit provided by the manufacturer for the calibration of the system to read out radiation dose from an 80 kVp x-ray beam.

2.2. The OBI kV-imaging system

The kV x-ray beams in this study were from a Varian OBI system integrated into Varian Trilogy linear accelerator. The x-ray tube is capable of generating photon peak kilovoltage (kVp) values between 40 and 130 kV. The x-ray field size can be adjusted independently in both x and y directions by four collimator blades. The system can be used to acquire projected kV digital radiographs for patient positioning. For volumetric imaging, the Varian OBI 1.4 has six cone beam CT (CBCT) image acquisition modes including standard-dose head, low-dose head, high-quality head, pelvis, pelvis spot light and low-dose thorax. The standard-dose head, low-dose head and high-quality head scans use the same 100 kVp x-rays but with different mAs which result in different dose to patients. Pelvis and pelvis spot light scans use half-fan and full-fan scan modes respectively but use the same 125 kVp x-rays and different mAs. Low-dose thorax scan uses 110 kVp x-rays with half-fan scan mode.

The characteristics and scan parameters of the x-ray source for each imaging mode have been reported in detail (Ding and Coffey 2010). The kV-CBCT beams are produced by the rotating anode tube with a W-Re target (Model G242) (Ding et al. 2007). The photon fluence is modulated by added filters (bow-tie filters). The full bow-tie and half bow-tie are designed to be used with full-fan and half-fan acquisition modes, respectively. The use of bow-tie filters can improve the image quality and reduce image dose. Both filters are made of aluminum with 1.5 mm thickness at the center which are described in a previous studies (Ding et al. 2007, Ding and Coffey 2010). The full-fan bow-tie filter and the half-fan bow-tie filter typically correspond to the 200° (full-fan acquisition modes) and the 360° acquisitions (half-fan acquisition modes), respectively. The source rotation range for a full-fan scan starts from the patient’s left side and ends at 20° beyond the patient’s right side. In the half-fan scan, the x-ray source starts from the patient’s left side and ends at the patient’s left side (Ding et al. 2010).

2.3. The Monte Carlo simulations and the beam calibration factor

The Monte Carlo user codes BEAMnrc/DOSXYZnrc (Nelson et al. 1985, Kawrakow and Rogers 2002, Rogers et al. 1995, Walters et al. 2006) were used in the simulation of each x-ray source used for kV-CBCT. The detailed x-ray tube geometry was simulated to generate each kV-CBCT x-ray beam with a specific blade setting and a bow-tie filter (Ding et al. 2007, Ding and Coffey 2010). Each simulated x-ray beam specified a scan protocol that was then used in calculations of dose distributions in the phantom and patients (Ding and Coffey 2009, 2010, Ding et al. 2008, 2010). The anode x-ray tube specifications, including target design, beam definition and beam filtration systems and incident electron energy, were obtained from the manufacturer. In the Monte Carlo simulation, the energy thresholds for secondary particle creation (AE, AP) and energy cutoff (ECUT, PCUT) for particle transport (Nelson et al. 1985) were set to AE = ECUT = 0.516 MeV (total electron energy) for electrons and
AP = PCUT = 0.001 MeV for photons as done in previous studies (Ding et al. 2008, 2010, Ding and Coffey 2010).

The beam output of each simulated kV-CBCT beam was individually calibrated using the method described in detail in previous studies (Ding and Coffey 2010, Ding et al. 2008, 2010). The absorbed dose was determined by using the calibrated ion chamber, Exradin Model A14. The air-kerma calibration factors for this chamber specified by the half-value layer (HVL) were obtained from an Accredited Dosimetry Calibration Laboratory (ADCL) (Andreo et al. 2000, Ma et al. 2001). The beam calibration method makes it possible to accurately predict dose to patients from an image acquisition procedure (Ding and Coffey 2009, Ding et al. 2008, 2010) by using Monte Carlo simulations.

The Monte Carlo calculated dose distributions resulting from a specific kV-CBCT scan protocol were calibrated in phantom and validated in RANDO phantom.

2.4. The absorbed dose determination and the phantom

The phantom slabs used in this study for absorbed dose determination using the ionization are made of PLASTIC WATER® low energy range (PW-LR) (CIRS, Norfolk, VA) and one slab has an ionization chamber insert hole which fits an Exradin Model A14 ionization chamber. The phantom materials were chosen because it has been shown to be water equivalent for diagnostic energy photon beams (Ding and Coffey 2010). The dose to water, \( D_{\text{chamber}} \), determined by using the ionization chamber is based on equation (1) (Ma et al. 2001):

\[
D_{\text{chamber}}(kV) = MN_kP_{Q,\text{cham}} \left( \frac{\mu_{\text{en}}}{\rho} \right)^w_{\text{air}},
\]

where \( D_{\text{chamber}} \) is the dose to water in the phantom, \( M \) is the leakage-corrected chamber reading in Coulombs (C) and corrected for other influencing quantities including temperature and pressure, \( N_k \) is the air-kerma calibration factor for the given beam quality, \( Q \), in Gy/C, \( P_{Q,\text{cham}} \) is the overall chamber correction factor that accounts for the change in the chamber response due to the displacement of the phantom by the ionization chamber and the presence of the chamber stem, the change in the energy, etc, and \( \left( \frac{\mu_{\text{en}}}{\rho} \right)^w_{\text{air}} \) the water-to-air ratio of the mean mass energy-absorption coefficients. In this study, the chamber calibration factors for x-ray beam quality specified by HVL and kVp were obtained from an ADCL. \( P_{Q,\text{cham}} \) is taken as unity in this study (Nahum 1996). The air-kerma calibration factors of the ionization chamber were obtained from an ADCL (Andreo et al. 2000, Ma et al. 2001).

2.5. The energy response correction factor for the OSL system

In this study, the microStar reader was calibrated by using the calibration-kit provided by the manufacturer to measure 80 kVp diagnostic energy photon beams. The manufacturer’s kit included a calibrate-set and a QC-set. The calibrate-set contained three nanoDot dosimeters of each dosage: 0.5, 3, 50 and 100 cGy. The QC-set contained three dosimeters with a dosage of 1 cGy and three unexposed dosimeters. The QC-set was used to check the system calibration accuracy.

In this study, the dose measured by using nanoDot dosimeters for kV-CBCT beams is expressed by the following:

\[
D_{w,\text{nanoDot}}(kV) = M_{w,\text{nanoDot}} f_{kV},
\]

where \( D_{w,\text{nanoDot}} \) is the dose measured by the nanoDot dosimeter, \( M_{w,\text{nanoDot}} \) is the dose reading from the microStar reader which is calibrated by using the manufacturer’s kit for an 80 kVp x-ray, and \( f_{kV} \) is an additional energy response correction factor for the kV beams. We
introduced $f_{kV}$ to account for differences in energy responses between 80 kVp x-rays and our kV beams. The additional correction factors for difference kV beams are obtained by comparing the dose, $D_{\text{chamber}}$ determined by using equation (1) as the gold standard for the same irradiation:

$$f_{kV} = \frac{D_{\text{Chamber}}^w}{M_{\text{nanoDot}}^w}. \quad (3)$$

The additional correction factors, $f_{kV}$, account for differences in energy responses between manufacturer’s 80 kVp x-rays and a user’s kV beams. The correction factors, $f_{kV}$, were determined by placing nanoDot dosimeters at the location where the dose was measured by using the calibrated ionization chamber.

The microStar reader system is operated in two modes with different stimulation light power (low and high power light beam) depending on the magnitude of dose received by the dosimeter. In the imaging dose range (0–8 cGy) used in this investigation, the system operates in a high light power stimulation beam in order to get enough signal to measure small dose received by the nanoDot dosimeter.

The dose distributions measured with nanoDot were compared with Monte Carlo calculated dose distributions in the PW-LR slabs ($30 \times 6 \times 30$ cm$^3$). The phantom geometric center was placed at the isocenter of the kV-CBCT scan. The nanoDot detectors were placed with 3–5 cm between each other in the middle plane of the PW-LR slab phantom. Each dosimeter was read five times. The first of five readings was not used in the average because the first reading was often affected by measurement noise. The average of four repeated readings was taken as the measured value. Additional energy correction factors used were manually applied to the average readings by using equation (2).

In order to evaluate feasibility to use nanoDot as a dosimeter in the application to monitor the dose to patients resulting from an image guidance procedure, we used an anthropomorphic RANDO phantom (Alderson et al 1962) which consists of 35 axial segments containing a human skeleton and material with similar properties to soft tissues. The dosimeters were placed at the skin of the RANDO phantom. The measured doses were compared with the Monte Carlo calculated skin dose. The volumetric CT images of the RANDO phantom used in the Monte Carlo calculations were obtained from a CT scanner (Philips Brilliance CT Big Bore, Philips Healthcare, Andover, MA). This is the same CT scanner used for obtaining patient CT images for radiotherapy treatment planning.

In the Monte Carlo dose calculations, each patient, each voxel in the CT images is converted from a CT number to a specific material and density. Four materials, air, lung, tissue and bone, are used in creating a CT-based patient phantom. More detailed descriptions regarding the ramp for converting CT values to material and density can be found in a National Research Council of Canada publication (Walters et al 2006). The voxel size used in the Monte Carlo calculations is 0.25 cm $\times$ 0.25 cm $\times$ 0.25 cm.

The Monte Carlo kV-CBCT dose calculations include three steps: (1) Monte Carlo simulation of realistic kV-CBCT scan sources. Each kV-CBCT scan specific source was stored in a file called phase-space file (Rogers et al 1995) as for image dose calculations (Ding and Coffey 2010, Ding et al 2008). These simulated kV sources were benchmarked against measurements in water phantom (Ding et al 2007, 2008, Ding and Coffey 2010) for the x-ray source accuracy. (2) The x-ray source output of each kV-CBCT scan mode was calibrated for dose calculations, and (3) the calibrated x-rays source was then used to calculate the dose distributions in phantom and CT-images resulting from the specific kV imaging procedure.

The method of the dose calculations resulting from an imaging guidance producers were described in detail in previous investigations (Ding et al 2008, Ding and Coffey 2010).
Figure 1. The additional energy response correction factor, $f_{kV}$, as a function of nominal x-ray tube kilovoltage potential from 60 to 125 kV, which corresponds to HVLs of 2.3 to 5.4 mm Al. The error bars are estimated to be 2–3% based on the experimental reading analyses.

3. Results

Figure 1 shows the additional energy response correction factor, $f_{kV}$, as a function of the nominal x-ray tube kVp from 60 to 125 kVp, which correspond to the HVLs of 2.3 to 5.4 mm Al, respectively (Ding and Coffey 2010). The correction factors were determined by using equation (3) by placing nanoDot dosimeters in the phantom at the location which was adjacent to the calibrated ionization chamber. The dose was determined by the chamber readings (equation (1)). The increase in energy response correction factors for beams with bow-tie filters can be explained by reduced low-energy photon counts in the spectrum due the bow-tie filter.

Figure 2 shows the comparison between nanoDot measured and Monte Carlo calculated dose distributions resulting from kV-CBCT with high-quality head, low-dose thorax, pelvis and pelvis spot light of full bow-tie filter scan protocols, respectively. These four different dose distributions represent all available dose distributions from Varian OBI kV-CBCT scan protocols. The same relative dose distributions were found for standard-dose head, low-dose head and high-quality head scans and the differences were in the absolute doses because the mAs used in these scans were 145, 72 and 720 mAs, respectively (Ding and Munro 2013). We selected high-quality head scan in comparison, to increase the signal-to-noise ratio in our experimental measurements. The measurement points were in the middle of the plane of the phantom. The differences among kV-CBCT scan protocols in the Varian OBI 1.4 system include x-ray tube kVp, scan rotation angles, size of beam fields and types of bow-tie filters. The agreements for different scan procedures between the measurements and the Monte Carlo calculations are within the experimental and calculation uncertainties. It is worth noting that the accuracy of the Monte Carlo calculated dose distributions were benchmarked against experimental measurements for each specific kV-CBCT scan protocol (Ding and Coffey 2010). Although the statistical uncertainties of the Monte Carlo simulations are less than 0.3% which is smaller than the thickness of the curves shown in figure 2, the estimated uncertainties of the MC calculation are 3–5% considering errors introduced in the Monte Carlo simulated beam calibration.
Figure 2. The comparison between nanoDot measured dose points (diamond symbol) and the Monte Carlo calculated dose distributions (lines) resulting from kV-CBCT with high-quality head, low-dose thorax, pelvis and pelvis spot light of full bow-tie filter scan protocols, respectively. The uncertainties for the measurement points are estimated to be 2–4%. The error bars (reproducibility) of measured points are approximately the size of the symbol.

Figure 3 presents the results of the study using the anthropomorphic RANDO phantom in which the imaging doses result from a high-quality head kV-CBCT scan, where (a) shows a photo of the RANDO phantom indicating the area where nanoDot dosimeters were placed; (b) shows the Monte Carlo calculated dose distributions in color wash in a sagittal and an axial CT image; (c) shows calculated dose profiles along the lines A–B and C–D; the large dose fluctuations in the dose profiles are because the dose to bone (cervical vertebrae) is 2–4 times of the dose to soft tissues as seen in the dose color wash. The measured (MC calculated) radiation doses for dosimeter-1 to dosimeter-5 are 0.31 (0.32), 0.22 (0.21), 0.29 (0.31), 2.64 (2.63) and 1.84 cGy (1.95 cGy), respectively. The estimated uncertainties of the measurements are 2–5% based on the reproducibility and response corrections. The uncertainties of the Monte Carlo calculations are 3–5% although the statistical uncertainties are less than 0.5%. The agreements between the measurements of the nanoDot and Monte Carlo calculations are within the uncertainties of the experiments and the Monte Carlo calculations.

It is noteworthy that the doses to eyes in this RANDO phantom study are five times higher than previously reported (Ding et al 2010) because we used a high-quality head kV-CBCT in OBI 1.4 in this experimental measurement. Three options (standard-dose head, low-dose head and high-quality head) are available OBI 1.4 for selecting a kV-CBCT scan for the head. The
Figure 3. (a) A photo of the RANDO phantom indicating the placement of dosimeters: dosimeter-1 and -2 are above the right and left eye respectively, dosimeter-3 is on the upper lip and dosimeter-4 and -5 are at the right (point C) and the left (point D) side of the neck, respectively. (b) Calculated dose distributions in color wash in a sagittal and axial images; (c) calculated dose profiles along the lines A–B and C–D, respectively.

doses for the standard-dose head, low-dose head and high-quality head scans can be scaled by their x-ray exposures parameters in mAs (400, 200 and 2000), respectively (Ding et al. 2010). The default settings in OBI 1.4 are to rotate the x-ray source below the patient that
An OSL dosimeter for measuring patient exposure

Figure 4. (a) An axial view of planning CT images showing Monte Carlo calculated dose distributions resulting from the low-dose thorax scan and in vivo dose measurements where three nanoDot dosimeters were placed at point-R, point-A and point-L. The lung tumor PTV is also shown; (b) an axial view of actual kV-CBCT images acquired during the patient treatment positioning; (c) a dose profile along the line E–F where the peaked dose in the dose profile corresponds the dose to bone (rib). (d) Calculated organ doses expressed by dose–volume histograms (DVH).

starts from the patient’s left side and ends at 20° past the patient’s right side (total 200°) (Ding et al 2010). When the patient is in the supine position on the treatment table, the x-ray beam only exits from the patient’s face when using the head scan. This results a significant dose reduction to patient radiation sensitive organs located in the anterior part of the head, such as eyes (Ding et al 2010). The slightly higher dose to the right eye compared to the left eye was due to the head scan x-ray source rotation geometry (Ding et al 2010). We also performed dose measurements for imaging dose resulting from a pair of orthogonal kV radiographs and a similar agreement was obtained and the doses to patients resulting from a pair of orthogonal head kV radiographs were less than 0.1 cGy.

Figure 4 presents an in vivo patient study of the dose patient skin resulting from a low-dose thorax kV-CBCT scan during IGRT, where figures 4(a) and (b) show the calculated dose distributions in color wash on an axial image from the treatment planning CT and a corresponding image reconstructed from the actual kV-CBCT scan, respectively. The three nanoDot dosimeters were placed at point-R, point-A and point-L. For this patient the isocenter of the kV-CBCT scan was set to the isocenter of the treatment target which is at the left side of the lung. It can be seen that part of the patient’s body was missing from the kV-CBCT images because the maximum axial view diameter of the kV-CBCT scan is limited to 45 cm. The comparisons of measured (MC calculated) doses at point-R, point-A and point-L were 0.45 (0.42), 1.24 (1.21) and 1.46 cGy (1.41 cGy), respectively. Here we observed up
to 6% discrepancies between measurements and calculations. The discrepancies between measurements and calculations are expected to increase for in vivo experiments due to a potential patient positioning variation because the Monte Carlo simulations are based on planning CT images instead of actual kV-CBCT images. Fortunately, the dose to skin was insensitive to the small variations of the location as shown in the measured doses at point-A and point-L. Figure 4(c) shows the calculated dose profile along the lines E–F. The dose to bone is 2–4 times of the dose to soft tissues as seen in the dose to the ribs in the plotted dose profile. Because the scanned isocenter was placed at the center of the PTV in the left lung, the resulting dose to the left lung is higher than the dose to the right lung as seen in the dose–volume histograms (DVH) plots. It is noteworthy that the dose to skin exposed to primary x-ray beam ranges from 0.3 to 1.4 cGy, while the dose to soft tissues were also within the same range of the skin dose. This may indicate the possibility of using measured skin dose to serve as a surrogate to estimate patient internal organ doses when patient dose distributions resulting from a specific image guidance procedures are available (Spezi et al 2009, Downes et al 2009, Ding et al 2010, Nelson and Ding 2012).

4. Discussions

Because the source of the radiation rotates around the patient for a kV-CBCT scan, studies have shown that the variation of the dose to soft tissues between skin and organs within the imaged volume is predictable (Downes et al 2009, Spezi et al 2009, Ding and Coffey 2009, Ding et al 2010, Chow et al 2008, Doucet et al 2003) and the values may be tabulated (Ding and Malcolm 2012). It is useful to have an experimental method to monitor and to estimate additional patient exposure resulting from image guidance procedures, such as kV-CBCT scan. There are other advantages of using experimental method to estimate or monitor the imaging dose to patient. The measured dose is able to detect abnormal imaging dose that the patient received while calculated dose cannot. This is because the calculated dose resulting from an image procedure that was calibrated based on the specified scan protocol. If an imaging procedure were not performed appropriately by selecting wrong scan protocol or by not using bow-tie filters, the in vivo patient measurements would show unexpected dose values. An easy of use and noninvasive dosimeter can provide an alternative method to estimate patient exposure as well as detect accidental incorrect calibration of an imaging system.

It is worth discussing the uncertainties of dose measurements using nanoDots. The measurement errors are contributed by several factors which include the accuracy of dose determination using an ionization chamber in phantoms and the determination of the energy response correction factor for nanoDot detectors. Although the air-kerma calibration factors for the chamber specified by HVL were obtained from an ADCL, there are variations in beam spectra even with the same HVL between user’s beam and that of the calibration lab. In addition, it is often convenient that the x-ray source output determination is performed in a plastic phantom instead of liquid water phantom. It has been shown that the potential errors of determining the x-ray source output of a kV-CBCT beam in non-water equivalent plastic materials for diagnostic energy x-rays, such as solid water and PMMA, may approach 8% and 20%, respectively (Ding and Coffey 2010). Therefore a water equivalent plastic phantom for diagnostic photon energy beams, such as PW-LR, should be used. In our study the benchmarking of the MC calculations against the experimental measurements were performed in water phantom by comparing ionization measured dose profiles and Monte Carlo calculated dose profiles for each kV-CBCT source (Ding et al 2007, Ding and Coffey 2010).

It can be seen that the correction factor for 80 kVp is not unity even when we used the calibration set for 80 kVp x-rays because of different filtration and scattering conditions
between the x-rays used by the manufacturer and that by users. Although the energy response correction factors shown in figure 1 may not be accurately applicable to other x-rays produced from different kV sources or to a different batch of nanoDot detectors manufactured at different time, the methodology and technique presented in this study can be used to determine the energy response corrections in order to measure the dose resulting from image guidance procedures accurately.

For in vivo measurements using nanoDot dosimeters, the measurements uncertainties are also affected by the location of the detector’s placement. Although the dose at the skin is not very sensitive to the small variation of dosimeter placement positions, the measurement uncertainties may increase if the dosimeter is not placed directly on the skin or placed on the mask.

Although reproducibility of measurements presented in this study is 2–5% and the estimated uncertainties of the MC calculation are 3–5%, the overall uncertainties for the whole method can be larger considering other factors discussed above.

It is worth mentioning that nanoDot dosimeter is not suitable to measure radiation doses resulting from mixed beams of megavoltage therapeutic and kilovoltage imaging radiations because the radiation response of the nanoDot between kV and MV is large. Reft (Reft 2009) studied the response of the detectors as a function of energy for photons with energies from kilovoltage to megavoltage energy photon beams and showed that nanoDot’s energy response relative to the 6 MV photons increases from 1.05 for 60Co to 3.5 for x-rays with an equivalent energy of 35 keV. In clinical practice for in vivo imaging dose measurements, detectors are to be removed after image guidance procedures are completed and before the therapeutic megavoltage beams are delivered to patients.

5. Conclusions

We have tested a commercially available OSL dosimeter, nanoDot, for measuring dose resulting from kV-CBCT imaging using high-quality head, low-dose thorax, pelvis and pelvis spot light. Additional energy response corrections are necessary in order to improve the measurement accuracy for a user’s kV-CBCT beam after the microStar reader has been calibrated to measure 80 kVp photon beams. The range of the energy response corrections has been found to be 0.88–1.13 for measuring radiation doses from kV x-rays ranging from 60 to 125 kVp with/without bow-tie filters in the Varian OBI 1.4 system.

The test on RANDO phantom in the head region and in vivo patient in the thorax region have demonstrated the feasibility of using nanoDot as an useful dosimeter for measuring patient dose from an image guidance procedure. The dosimeter placed on a patient’s skin has potential to serve as a method to monitor and to estimate patient exposure resulting from a kilovoltage x-ray imaging procedure.

The nanoDot is suitable for in vivo dosimetry due to its small size and noninvasiveness. The almost nondestructive read-out of the nanoDot dosimeter also allows the measurement of dose from each imaging acquisition as well as accumulated total imaging guidance dose from repeated imaging procedures during the entire treatment course with a single dosimeter.

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