Use of a laser-guided collimation system to perform direct kilovoltage x-ray spectra measurements on a linear accelerator onboard imager

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Purpose: The increased use of image-guided radiation therapy (IGRT) has led to increased use of kV on board imaging (OBI) devices. At present, directly measured OBI beam quality data have only been reported in terms of half-value layers (HVL). However, the HVL metric alone does not give the full OBI energy spectra as needed for accurate beam modeling. Although direct kV spectrometer devices exist they typically suffer from detector pile-up when used with OBI sources. We therefore present, for the first time, a novel laser-guided collimation system that allows direct measurement of the full energy spectrum for clinical OBI systems.

Methods: Several clinically relevant spectra (80, 100, and 125 kVp), with and without the half bow-tie filter, were measured using a thermoelectric cooled cadmium telluride (CdTe) detector paired with a multichannel analyzer. To prevent detector saturation, the photon flux at the detector was reduced by use of an in-house designed laser-guided collimation system. After applying energy bin corrections, direct spectroscopic measurements were compared to Monte Carlo (MC) simulated spectra in order to verify accuracy of collected data. Both percent depth dose (PDD) curves and digitally reconstructed radiographs (DRR) were compared using the measured vs MC spectra.

Results: Measured and MC spectra agree with RMSD between 1.96% and 3.29%. PDD curves generated from the measured and MC spectra were found to match except for in the small buildup region, with an overall match for the six beams ranging between 0.3% and 2.7% RMSD. DRRs matched well with a maximum difference in contrast of 1.1% and RMSD of 0.46% contrast for various materials in DRRs.

Conclusions: The use of a laser-guided collimation system provided a method for quickly obtaining highly accurate kV spectrum data from OBI sources. For kV dose or DRR calculation, it was found that both spectra produced similar results. © 2018 American Association of Physicists in Medicine [<https://doi.org/10.1002/mp.13188>]

Key words: collimation, energy spectrum, kV imaging, laser alignment, linear accelerator (linac), onboard imaging (OBI), x-ray source

1. INTRODUCTION

Modern commercial linear accelerator (linac) systems are increasingly outfitted with onboard-imaging (OBI) devices in order to provide additional information on patient positioning for radiation therapy. In the case of kV x-ray imaging beams, it has been demonstrated that accurate dose simulations requires full knowledge of the kV beam energy spectrum.^{1–4} The recently released AAPM TG180 report on image guidance doses delivered during radiotherapy⁵ provides a review of some of the literature on the dose delivered by the OBI during treatment. For example, older versions of the Varian OBI (Varian OBI 1.3) can deliver 1–9 cGy to soft tissue and 6–29 cGy to bones from a single cone-beam CT scan, although lower doses are possible with advanced dose reduction technologies in modern imaging systems. Imaging dose estimations are also featured in the AAPM TG75 report on the management of imaging dose during image-guided radiotherapy.⁶ The TG75 report lists, among other estimates, dose rates in the order of 10.8 mGy/min at the surface for gantry-mounted kV fluoroscopy, depending on the technique. In addition to dosimetry, spectroscopic beam knowledge is also

required for the generation of highly accurate digitally reconstructed radiographs (DRRs).⁷ Such DRRs may be used as part of a patient positioning or tracking algorithm,^{8,9} or may be used to inform imaging parameter settings.¹ However, despite the importance of accurate spectroscopic information for a variety of imaging tasks associated with the OBI, there have been no direct measurements of the full OBI beam spectrum.

Traditionally, beam quality for kV sources is described in terms of the accelerating potential (kVp), with quality described in terms of half-value layer (HVL).¹⁰ These two quantities are simple to measure empirically, but fail to completely describe the spectra and predict the distribution of interactions that may occur. Many research applications, however, require more detailed information. To fill this gap, Monte Carlo (MC) methods have been used extensively in the literature to generate the necessary kV beam spectroscopic information needed for accurate OBI beam modeling.^{1–4} However, direct experimental verification is still needed as beam spectroscopic quality can vary substantially based on parameters such as the degree of collimation, filtration, and other sources that may have not have been properly

modeled in simulations.¹¹ Although MC-derived spectroscopic data for OBI sources are used for both dosimetric and imaging applications, there have been no reports on direct measurement of the OBI spectra, and this is primarily due to the challenges posed in dealing with the high particle fluence of the OBI.

One method to reduce photon flux is to scatter the primary x-ray beam off a lucite plane rotated by 45° with respect to the kV source central axis (CAX) and collecting the 90° Compton-scattered x rays using a spectrometer.^{12,13} Unfortunately, such methods provide indirect measurements of the energy spectra that require additional processing steps using the Klein–Nishina formula and Compton energy-angle formula to reconstruct the primary spectrum. Another method to reduce photon flux is the use of collimation in front of the spectrometer.^{14,15} By the use of multiple microdiameter tungsten collimators arranged along the CAX, the photon flux can be reduced by several orders of magnitude allowing direct spectroscopic measurements to be performed. However, the use of extremely small collimators leads to significant issues with angular detector alignment that can take significant time and many trial-and-error measurements before proper alignment is achieved. For example, for two 100 μm diameter tungsten collimation disks that are 1-mm thick and spaced 50 mm from each other, a rotation of less than 0.2° will result in complete occlusion of the source from the detector, as demonstrated in Eq. (1) below. Although the use of such collimator alignment techniques are adequate for laboratory-based experiments, they are not practical in clinical scenarios where machine time is limited and quality assurance (QA) measurements must be made quickly.

In this work, we present a novel method that overcomes the angular alignment problem using an in-house designed, laser-guided collimation method to directly measure the beam spectra for an OBI system using a thermoelectrically cooled CdTe detector. We then compare the measured vs MC-simulated spectra to verify overall accuracy of the technique.

2. MATERIALS AND METHODS

2.A. Laser-guided collimator

As shown in previous works, fluence reduction can be accomplished through the use of a series of collimating plates placed in front of the detector.^{16–18} However, such arrangements can make a direct line of sight of the x-ray source extremely difficult to locate, with solutions including the manual adjustment of positional set screws to achieve maximal fluence,¹⁷ the use of flat panel imaging equipment combined with 3D printing technology,¹⁶ and a system of multiple lasers and collimators placed on precisely machined rails.¹⁸ For a commercially available spectrometer system utilizing two tungsten collimator plates of finite thickness t_1 and t_2 , with apertures of radius r_1 and r_2 , separated by a brass spacer of thickness s , the two apertures will appear as overlapping circles centered from the beams' point of view. As the collimation system is rotated out of alignment, the area of

overlap between the two apertures will decrease, until complete occlusion occurs at an angle as described by Eq. (1),

$$\vartheta = \tan^{-1} \frac{r_1 + r_2}{s + t_1 + t_2}. \quad (1)$$

It is also possible to approximate the degree of source occlusion, as a function of detector-to-source misalignment. The basic geometry of the collimation device is shown in Fig. 1(a). From this figure it is possible to derive Eq. (1) (demonstrating the angle of total occlusion). However, it is more accurate to view the apertures not as line segments, but as circles. Prior to rotation, each collimator opening presents an area πr^2 , where r is either of the previously defined radii r_1 or r_2 . There are then three effects that result in a reduction of solid angle of detector visible to the source. For the first effect (geometric shortening), we note that as the cylinder rotates with respect to the x-ray source by some angle φ , the exposed area becomes ellipsoidal as it is projected on a plane oriented along that angle φ . The ellipse has a long axis equal to the original diameter of the circle, and a short axis shortened by a factor $\cos(\varphi)$. For the second effect (interdisk occlusion), the line of sight between detector and source depends on the overlap of the apertures of the two collimators, and therefore that the solid angle exposed depends on the degree of overlap. As shown in Fig. 1(b), the distance between the centers of the two ellipses is

$$d = (s + t_1 + t_2) \sin \varphi \quad (2)$$

In general, the area of overlap between two ellipses is difficult to solve. The problem can be greatly simplified by treating the elliptical projections as circles (which is a reasonable approximation for small angles of rotation). In this case, for two circles, of radius r_1 and r_2 , the center of each separated by a distance $d > 0$, the area of the overlapping region can be demonstrated to take the value given in Eq. (3).

$$A = \frac{1}{2} \sqrt{(-d + r_1 + r_2)(d + r_1 - r_2)(d - r_1 + r_2)(d + r_1 + r_2)} \quad (3)$$

Finally, for the third effect (intradisk occlusion), we note that because of the finite thickness of collimating disks, the front and back faces of each collimating disk will overlap in a manner similar to the overlap of the two separate disks. For a collimation system with the geometry we used (two collimators each 2 mm thick, with 100 μm and 1-mm apertures, separated by a 36-mm spacer, as described below), the independent effects of geometric shortening, interdisk occlusion, and intradisk occlusion are shown in Fig. 1(c). It is not obvious how to combine the three effects; however, it is clear that there is a region dominated by intradisk occlusion, and later, a region dominated by interdisk occlusion, while the effect of geometric shortening is negligible [justifying the use of the circular approximation in the derivation of Eq. (3)].

To help reduce such occlusion issues, an in-house-designed laser alignment method was developed and constructed as shown in Fig. 2. The basic premise of the method

FIG. 1. X-ray source occlusion with a double plate collimator. (a) 1D geometry of occlusion model. (b) Separation between center of collimation disks as a function of angle of misalignment. (c) Calculated source occlusion as a function of angle of misalignment. [Color figure can be viewed at wileyonlinelibrary.com]

is to locate a laser at the spectrometer detector position and shine it through the collimating disks and onto the x-ray source anode. In this way the laser beam passes through both collimator disks indicating the direct line of sight of the collimation system. By following the trajectory of the laser from the collimating assembly to the radiation source, it was possible to correctly align the detector and collimator with the radiation source. Additionally, the high luminosity provided by a laser allows enough photons to pass through the collimation system such that a visually identifiable spot on the x-ray source can be located. Once the collimator is properly aligned, the laser can be swapped out and replaced with the spectrometer.

In this work, a 650-nm laser diode was used as the laser source. To locate the laser behind the collimator disks, and to facilitate easy insertion and removal of the laser into the collimator, a Teflon adapter was machined to precisely fit within the stainless steel housing. To calibrate laser direction, such that the beam runs parallel to the collimator housing, fine threaded adjustment screws were placed in the laser housing such that laser pitch and yaw adjustment could be made (Fig. 2). Angular calibration of the laser was then made by placing the laser in a lathe chuck, shining the laser onto a target located approximately 5 m away, and then rotating the chuck by hand while adjusting the pitch and yaw screws until laser wobble was removed.

Measurements were taken from a Varian Trilogy (Varian, Palo Alto, CA) linac equipped with an OBI (field size

$40 \times 40 \text{ cm}^2$) using a CdTe detector (XR-100T, Amptek, Bedford, MA). Without the collimation, even at the lowest mAs settings and the maximum source-to-detector distance (SDD) as allowed by the linac vault (approximately 4 m), it was found that the x-ray photon flux was high enough to cause complete CdTe detector saturation. Collimation was then performed by using two tungsten disks, each 2 mm thick, with apertures of 100 μm and 1 mm, separated by a 36-mm brass spacer [Figs. 1(a), 2]. The laser was then inserted into the collimator assembly, and the entire assembly moved until the laser spot was located on the x-ray source anode. As the x-ray source was hidden beneath linac covers, knowledge that the kV and MV source CAXs had strong coincidence with machine isocenter, and that linac room lasers were well calibrated, were used to align the laser collimator along the linac x-axis. Upon locating the x-ray source, the laser was removed and the CdTe detector inserted into the collimator housing with special care being taken to not disturb the assembly position. X-ray spectra were then measured for 80, 100, and 125 kVp with and without the addition of a half bow-tie (HBT) filter. All measurements were made by deploying the x-ray source in fluoroscopic mode, with beam settings between 2–4 ms and 10–20 mA per frame, with higher fluence rates used for lower energy beams. These settings were chosen to allow for a high detection rate on the spectrometer without overheating the anode.

2.B. Correction of raw spectroscopic data

Raw spectroscopic data acquired with the CdTe detector must be modified to account for detector efficiency. Additionally, sufficiently high-energy photons incident on the detector may, when undergoing the photoelectric effect in the detector, produce K-shell vacancies which may result in the release of K_α and K_β photon emissions. These fluorescent emissions may not be absorbed by the detector, and therefore, the apparent energy absorbed will be lower than the incident energy of the photon, resulting in an incorrectly measured energy spectrum. The manufacturer of the detector provides data on detector efficiency, but in order to correct for the K_α and K_β photon emission (escape photons), we use a stripping method^{19–21} previously described in the literature. The stripping method proceeds channel-by-channel, removing counts from lower energy bins and placing them into an energy bin of the original incident photon prior to interaction in the detector. In order to do so, the fractional amount of escape photons for each incident photon energy is calculated in EGSnrc. The interactions of monoenergetic photons across a range of energies are simulated in a slab of CdTe modeled with the detector geometry (5 $\times$ 5 mm slab of CdTe 1000 μm thick, behind a 100 μm beryllium window). In this way, the fraction of $K_{\alpha 1}$, $K_{\beta 1}$, $K_{\alpha 2}$ and $K_{\beta 2}$ escapes for each incident photon energy is determined, and used in the stripping process. We define the functions $f_i(E)$ as the probability of an escape photon of type i ($K_{\alpha 1}$, $K_{\beta 1}$, $K_{\alpha 2}$ and $K_{\beta 2}$) at energy E . For the raw spectroscopic measurements, we define $N_d(E)$ as the number of photons detected at energy E , and $N_t(E)$ as the true number of photons

FIG. 2. Laser-based collimator assembly used to align a spectrometer with an OBI x-ray source. The collimator consists of two tungsten disks with collimation holes of radius r_1 and r_2 , respectively, that are separated by a distance s and have thicknesses t_1 and t_2 , respectively. Adjustment screws are used for fine tuning of laser beam pitch and yaw angles. Both a diagram to demonstrate the geometry as well as photographs of the detector, collimator, and x-ray source are shown. [Color figure can be viewed at wileyonlinelibrary.com]

at energy E . The difference between the two, $\Delta N(E)$, accounts for both the counts missed at higher energies and counts added at lower energies due to the interactions in the detector. Thus, we have the equation for the difference,

$$\begin{aligned}\Delta N(E) &= N_d(E) - N_t(E) \\ &= \sum_i f_i(E + E_i) N_t(E + E_i) \\ &\quad - \sum_i f_i(E - E_i) N_t(E)\end{aligned}\quad (4)$$

where E_i is the energy of an escape photon associated with the i th peak. Therefore, the stripping process begins at the highest energy channel, using the above equation to strip away counts from the lower channels and adding them back to the original channel, and then proceeding to a lower channel, until reaching the escape edge.

2.C. Comparison of spectra

Monte Carlo data were generated based on the manufacturer specifications of the Varian Trilogy OBI modeled in BEAMnrc. In the Monte Carlo model, a 5-mm diameter electron beam is incident on a tungsten target, and the resultant bremsstrahlung beam passes through an exit window comprised of glass, oil, and polycarbonate, a lead primary collimator, an aluminum filter, and an additional set of secondary lead collimators. For all beams with the HBT filter, the HBT filter was also included in the model. The simulation continues through air to a source-detector distance of 4 m to replicate measurement conditions. For each of the six spectra, 500

million histories were used during the simulation. The output of the MC program is a phase space file, which provides a list of particles, each with an energy, position, and direction.

To compare spectra from MC and direct measurement, first, the spectra were set to the same scale. After rebinning the spectra from MC and measurements to occupy the same energy channels, with a variable scaling factor α , the mean square error between the spectra may be described by the dot product equation

$$(s_{MC} - \alpha s_m) \cdot (s_{MC} - \alpha s_m) \quad (5)$$

where s_{MC} is the spectrum from MC, and s_m is the measured spectrum. Taking the derivative of the mean square error with respect to the scaling term, and setting it to zero, the scaling term which minimizes the mean square error is given by Eq. (6) below

$$\alpha = \frac{s_{MC}^T s_m}{s_m^T s_m}. \quad (6)$$

The physical interpretation of the scaling term is that it accounts for the change in scale from the normalized Monte Carlo data and the nonnormalized measured data. The magnitude of the measured data depends on the beam-on time, the mAs/frame, collimation, and the frames per second used while collecting data, all of which were varied in order to efficiently collect data. To quantitatively assess the agreement between measured and MC, a root mean square deviation (RMSD) metric was used [the square root of the MSE as defined in Eq. (5)]. To account for detector noise, a moving average filter

was used to smooth the measured data before the RMSD calculation. Because the characteristic peaks are ideally sharp in the MC results, but spread out in the measured spectrum due to the finite size of the energy bins, RMSD values were recalculated after omitting comparisons of counts between 57 and 71 keV, in order to eliminate the influence of this effect.

2.D. PDD and DRR comparisons

To evaluate the importance of accurate keV energy spectra, both simulated percent depth dose (PDD) curves and digitally reconstructed radiographs (DRR) were compared using the measured vs MC spectra. PDD curves were generated in DOSXYZnrc from MC-derived spectra, and compared to PDD curves generated in DOSXYZnrc from our experimentally measured spectra (80, 100, 125 keVp, with and without HBT filter). Although the MC process used to simulate spectra produces full phase space files modeling the changes in spectrum off-axis, our measured beam spectra are measured at a single point on the central axis. Therefore, the MC PDD calculation for both MC spectra and measured spectra use a point source. While a dose calculation from a full phase space file would be more accurate, it is infeasible to produce a phase space file from measured data; therefore, in order to compare similar data, a point source with the spectrum along the CAX obtained from the phase space file was used instead.

DRRs were generated based on a previously reported model⁷ by Moore *et al.* using different beam spectra and a phantom comprised of seven pegs of different material from the Gammex RMI 465 phantom (Gammex, Middleton, WI). The seven Gammex/RMI inserts used were LN-300 Lung (electron density relative to water $[\rho_e]^{m,w} = 0.29$), LN-450 Lung ($[\rho_e]^{m,w} = 0.44$), adipose equivalent ($[\rho_e]^{m,w} = 0.90$), solid water ($[\rho_e]^{m,w} = 1.00$), muscle equivalent ($[\rho_e]^{m,w} = 1.02$), inner bone equivalent ($[\rho_e]^{m,w} = 1.08$), and cortical bone equivalent ($[\rho_e]^{m,w} = 1.71$). These inserts were chosen to provide a cross section of clinically relevant tissue types found in the body. Six spectra were used for the beam qualities described above (80, 100, 125 keVp, with and without HBT filter), and the difference in DRR contrast for the seven pegs by the various beams were measured and compared.

3. RESULTS

A comparison of the experimentally measured spectra and the MC-simulated spectra is shown in Fig. 3. For the majority of the energy bins, there was excellent agreement between measured and MC-simulated results. For higher energy bins, a slight disagreement was found between measured and MC spectra in that the measured spectra typically had less counts than the MC data.

The PDDs for the filtered and unfiltered 80, 100, and 125 keVp spectra are shown in Fig. 4. Visually the curves look identical; however, subtraction of the two curves reveals a slight difference. For the PDDs, the RMSD are found to be 0.3%, 0.6%, 0.8%, 1.0%, 2.0%, and 2.7% for 80 keVp, 100

keVp, 125 keVp, 80 keVp (HBT), 100 keVp (HBT), and 125 keVp (HBT), respectively.

The difference in DRRs for various materials is shown in Fig. 5. The materials are (in order of increasing electron density) as follows: (a) lower density lung, (b) higher density lung, (c) adipose, (d) water, (e) muscle, (f) inner bone, and (g) cortical bone. Among the set of measured contrast values, the maximum, the difference in observed contrast between DRRs derived from measured and MC spectra is less than the difference in observed contrast between nonfiltered and HBT filtered spectra. The maximum of the observed differences was 1.1%. The root mean square deviation (the average taken in quadrature of the difference between contrasts obtained from MC spectrum DRRs and measured spectrum DRRs) was 0.46%.

4. DISCUSSION

This study presents the use of a laser-guided collimation system to perform direct measurements of the full energy spectrum for a clinically used OBI system. Good agreement between measured and MC-simulated data confirms the collimator's ability to precisely locate and collimate photons originating directly from the OBI source. The small differences observed between the measured and MC-simulated spectra were most likely due to limitations of the MC simulations and calibration of the spectrometer.

As shown in Fig. 3, measured and MC-simulated spectra agree well, however, discrepancies at and around the characteristic peaks are visible. These are expected, as the measured spectra have lower energy resolution than the MC spectra. Therefore, there is a broadening and reduced intensity of the characteristic peaks in the measured spectra, which results in differences when compared to the MC-simulated spectra near the characteristic peaks. Because the peaks increase in prominence from the 80 keVp spectra to the 100 keVp spectra and from the 100 keVp spectra to the 125 keVp spectra, the peak broadening affects the RMSD between MC and measured spectra more dramatically in the higher keVp spectra. Another difference was that the MC spectra, especially in the HBT filtered spectra, show a number of lower peaks at lower energies that are not present in the direct measurement. These discrepancies may be due to the inability to model the collimating array used for direct measurement in the MC system. Modeling the collimating array would require an impractically large computation time in order to generate enough histories for any kind of statistical certainty in the results. Additionally, the error in energy sensitivity calibration of the detector may have contributed to the discrepancies and was not accounted for.

As seen in Figs. 4 and 5, the resulting PDD curves and contrast measurements also show good agreement. PDD agreement for all beams ranged from 0.3% to 2.7% RMSD, with the greatest disagreement for the 125 keVp beam with HBT filter and the differences being most prominent in the buildup regions. The contrast measurements all agree to within 1.1%, with RMSD of 0.46%. The small differences in

FIG. 3. Comparison of MC and measured beam spectra for 80, 100, and 125 kVp beams, with and without half bow-tie filter. MC results normalized to minimize RMSE with measured counts. Uncertainty bars provided based on summation in quadrature of statistical uncertainty from both measured and MC data. [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 4. PDD curves in water for MC spectra (dashed) and measured spectra (solid), with the difference below. [Color figure can be viewed at wileyonlinelibrary.com]

PDD and contrast when comparing MC to measured spectra suggest that the use of MC-calculated spectra for DRR generation and dose calculations is justified.

Although the laser-guided calibration system allows for rapid alignment of multiple collimators and location of a direct line of sight for photons in order to reach the

FIG. 5. Contrast for measured and MC-simulated spectra for (a) 80 kVp beams, (b) 100 kVp beams, and (c) 125 kVp beams, each with and without a HBT filter, for seven materials (low-density lung, high-density lung, adipose, solid water, muscle, inner bone, cortical bone). [Color figure can be viewed at wileyonlinelibrary.com]

spectrometer, the issue of where to point the laser on the OBI source remains. Often the x-ray source of the OBI is hidden behind HBT filters, linac covers, or other accessories making visual location of the x-ray source spot on the anode difficult. In this work, knowledge that the kV and MV source CAXs had strong coincidence with machine isocenter, and that linac room lasers were well calibrated, were used by aligning the laser collimator along the linac x-axis through isocenter and into the OBI.

5. CONCLUSION

A laser-guided collimation system was demonstrated to allow direct spectroscopic measurements of a high-fluence clinical kV OBI system. Spectra for a variety of clinically useful x-ray beams and configurations were measured directly and compared to MC-simulated results. Overall, there was good agreement between measured and MC spectra for all energies. In terms of using measured vs MC-simulated spectra for kV dose or DRR calculation, it was found that both spectra produced similar results for the beams evaluated in this work.

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