A new technique to characterize CT scanner bow-tie filter attenuation and applications in human cadaver dosimetry simulations

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Purpose: To present a noninvasive technique for directly measuring the CT bow-tie filter attenuation with a linear array x-ray detector.

Methods: A scintillator based x-ray detector of 384 pixels, 307 mm active length, and fast data acquisition (model X-Scan 0.8c4-307, Detection Technology, FI-91100 Ii, Finland) was used to simultaneously detect radiation levels across a scan field-of-view. The sampling time was as short as 0.24 ms. To measure the body bow-tie attenuation on a GE Lightspeed Pro 16 CT scanner, the x-ray tube was parked at the 12 o'clock position, and the detector was centered in the scan field at the isocenter height. Two radiation exposures were made with and without the bow-tie in the beam path. Each readout signal was corrected for the detector background offset and signal-level related nonlinear gain, and the ratio of the two exposures gave the bow-tie attenuation. The results were used in the GEANT4 based simulations of the point doses measured using six thimble chambers placed in a human cadaver with abdomen/pelvis CT scans at 100 or 120 kV, helical pitch at 1.375, constant or variable tube current, and distinct x-ray tube starting angles.

Results: Absolute attenuation was measured with the body bow-tie scanned at 80–140 kV. For 24 doses measured in six organs of the cadaver, the median or maximum difference between the simulation results and the measurements on the CT scanner was 8.9% or 25.9%, respectively.

Conclusions: The described method allows fast and accurate bow-tie filter characterization.

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Key words: CT, bow-tie filter, linear array x-ray detector, in vitro dose, Monte Carlo simulation

1. INTRODUCTION

Radiation dose from a CT scan is associated with multiple scanner and subject related factors, such as CT scanner radiation output, scan length, subject size, and body habitus. CT dose index (CTDI) is useful for characterizing CT scanner radiation output, but does not represent patient dose.¹ *In vivo* CT dose measurement in patients is impractical.² For these reasons, Monte Carlo simulation remains the gold standard for patient-specific CT dose evaluation.

Although Monte Carlo simulation has been widely used in medical physics, its application in patient-specific CT dose evaluation is still limited, because of limited computing speed and inaccessibility of the manufacturer's proprietary scanner information. The first issue may be gradually resolved in the future, with the improving performance of computing power and the adoption of graphics processing unit (GPU) or the Intel many integrated core (MIC) processor.³ To be successful in Monte Carlo based CT dosimetry, the attenuation profile of the bow-tie filter applied in CT scanner has to be determined accurately. Its importance is illustrated by the large variations of the central to peripheral CTDI₁₀₀ ratio with different CT

scanner models and bow-tie filters, as shown in Tables II–V of Li *et al.*⁴

To assist users to construct CT scanner source models in Monte Carlo simulations, Turner *et al.* presented a noninvasive method to generate equivalent energy spectra and filtration models based on the half-value layer (HVL) and bow-tie attenuation measurements on multidetector CT scanners.⁵ With the x-ray tube parked at the 3 or 9 o'clock position and an ion chamber attached to the patient table, relative bow-tie attenuation across a fan beam $[F(\theta)/F(0^\circ)]$, where θ is fan beam angle] was cleverly measured by adjusting the table height. Because the measurements could account for changes in the scanner conditions such as x-ray tube aging, their study revealed that simulations using the source models attained high accuracy and even might have advantages over the use of the CT manufacturer-provided source data of the scanner models. In 2013, McMillan *et al.* demonstrated that the above method can also be applied in kilovoltage (kV) cone-beam CT in radiation therapy.⁶ Such bow-tie attenuation measurement needs about 25 CT scans for a scan field-of-view (FOV) of 50 cm for every tube voltage/bow-tie combination. Thus, the total number of CT scans can exceed 200 when four

tube voltages and two bow-tie filters are considered. Alternatively, Boone proposed a method to evaluate relative bow-tie attenuation profile from radiation measurements at an off-isocenter position near the scan FOV edge in rotational CT scans.⁷ In his method, $F(\theta)/F(0^\circ)$ can be extracted using the inverse square law and sophisticated analysis of the resultant waveform data. In a subsequent study, McKenney *et al.* further validated the method using a real-time dose probe (active volume 0.14 cm^3) on a dedicated breast CT scanner with a custom-designed teflon bow-tie.⁸ Recently, Whiting *et al.* confirmed that the above method can also be used with an intermediate size real-time dosimeter (active volume 0.6 cm^3) and also developed a method to account for the finite size of the dosimeter.⁹ Randazzo and Tambasco¹⁰ presented a technique to rapidly characterize the CT x-ray beam in terms of fan beam angle-dependent HVL and fluence, using concentric cylindrical aluminum (Al) filters in conjunction with the Boone's method. In addition, Poirier *et al.* characterized a kV x-ray source on a Varian On-Board Imager with in-air relative dose and HVL measurements.¹¹ The results were used as inputs for an in-house dose calculation software and the computed percent depth-dose curves and transverse axis profiles in the homogeneous and heterogeneous block phantoms generally agreed with the measurements on the unit. Whiting *et al.* presented a technique to measure the CT bow-tie attenuation profile relative to the central ray using radiochromic films.¹² A custom-designed aperture was used to expose radiochromic film in a rotating CT gantry. This cost effective technique was applied on a Philips Brilliance 16 CT, and the results were reasonably consistent with the measurements using a real-time dosimeter and the CT scanner detectors, with a maximum difference of 25%.

In this paper, we present a new technique to measure the bow-tie attenuation in CT scanners, using a linear array x-ray detector. When it was placed on a stationary patient table and centered in the scan field, simultaneous and direct detections of the radiation levels across the scan FOV simplified both measurement work and data analysis. The results of two exposures with and without the bow-tie in the beam path allowed a measurement of the absolute bow-tie attenuation $F(\theta)$, which was exempt from the x-ray source to detector distance (SDD) related beam intensity correction (usually called the inverse square law) and thus any uncertainty in SDD. A determination of the attenuation at the bow-tie center is also useful for assessing radiation exposure magnitude or x-ray spectrum.¹² We performed experiments on a clinical CT scanner and GEANT4-based Monte Carlo simulations of human cadaver dosimetry to rigorously test the accuracy of the resultant bow-tie attenuation.

2. MATERIALS AND METHODS

2.A. Measurement of bow-tie attenuation using a linear array x-ray detector

This work used an x-ray detector (model X-Scan 0.8c4-307) by Detection Technology (Piisilta 1, 91100 Ii, Finland). The detector uses $\text{Gd}_2\text{O}_2\text{S}$ (GOS) based ceramic scintillators

and silicon photodiodes for detecting x-rays in diagnostic x-ray energy range (20–160 kV). The GOS scintillator has been used in many CT detectors, and has a high x-ray stopping power, high light output, short primary decay time, and low afterglow. The detector was designed for fast data acquisition with a sampling time as short as 0.24 ms. It contains 384 pixels equally grouped in six modules with a pixel size of $0.7 \times 0.8 \text{ mm}^2$, a pixel spacing of 0.8 mm, and an active detector length of 307 mm. Each readout signal can be digitized with 14 bits. The detector assembly has a length of 332 mm, a width of 335 mm, and a height of 30 mm. A photo of the assembly is shown in Fig. 1. When there is no radiation exposure, detector readout gives the level of nonzero offset. Nonlinear detector gain was examined using a series of exposures made at 80 kV and different tube currents, while the x-ray tube was parked at the 12 o'clock position and there was no bow-tie in the beam path.

To measure a bow-tie attenuation across a scan FOV on a CT scanner, the detector was placed on the patient table and was centered in the scan field. The detector surface was at the height of isocenter. One edge of the detector was near the isocenter so that the detector could cover half of the scan FOV. The detector was directly connected to a computer through a USB cable. After this stage, the table and the detector were kept stationary during data acquisition. The bow-tie attenuation profile was obtained with the x-ray tube parked at the 12 o'clock position. For each bow-tie/kV combination, two radiation exposures were made, one without bow-tie in the beam path (referred to as air scan in this paper), and the other with the bow-tie in the beam path. Each detector readout signal was corrected for both the background offset and the signal-level related nonlinear gain. For each detector pixel, a fan beam angle θ was calculated using the source to isocenter distance d_{iso} and the horizontal distance d of the pixel to isocenter,

$$\tan \theta = \frac{d}{d_{\text{iso}}}. \quad (1)$$

Bow-tie attenuation was then quantified as a ratio,

$$F(\theta) = \frac{S_{\text{bow-tie}}(\theta)}{S_{\text{air}}(\theta)}, \quad (2)$$

where S was the detected signal with both the offset subtraction and nonlinear detector gain correction. It is worthwhile to note

FIG. 1. Photo of a linear array x-ray detector (model X-Scan c4-307, Detection Technology, Finland). X-ray detection is centered in the black band.

FIG. 2. Attenuations of the added 4–9 mm aluminum in the beam path on a GE Lightspeed Pro 16 CT scanner. The TASMICS spectral model (Ref. 13) based calculations (lines) agreed with the measurements (squares) within 0.47%.

that the pixel-to-pixel variations in both beam intensity and detector gain canceled out in Eq. (2). In this work, we performed experiments on a GE Lightspeed Pro 16 CT scanner, and measured the body bow-tie attenuations at 80, 100, 120 and 140 kV, respectively. The resulting profiles were rigorously tested in human cadaver dosimetry simulations.

2.B. X-ray beam and filtration models

We assessed beam qualities at 80, 100, 120, and 140 kV, respectively, on the GE CT scanner. With the x-ray tube parked at the 6 o'clock position and the body bow-tie in the beam path, the radiation exposure at the isocenter was measured with 0, 4, 5, 6, 7, 8, and 9-mm type-1100 alloy Al added in the beam path, respectively, and the results are shown in Fig. 2. The GE CT scanner technical reference manuals describe that the beam filtrations consist of the x-ray tube assembly equivalent to 4.3 mm Al (at 70 kV) and fixed filtration including the 2-mm graphite and the 0.25-mm Al. When the body bow-tie filter was selected, 76- μ m copper was also added along the

FIG. 3. Comparison of an image slice (slice thickness 2.5 mm) before and after metal artifact correction.

FIG. 4. Variable tube current from an abdomen/pelvis scan of a human cadaver on a GE Lightspeed Pro 16 CT scanner with 120 kV, NT = 20 mm, pitch = 1.375, rotation time = 1 s, and z axis tube current modulation (Ref. 16).

central ray ($\theta = 0^\circ$). The TASMICS spectral model¹³ was used to generate unfiltered x-ray spectra at 80, 100, 120, and 140 kV. With the graphite and copper in the beam path above, we used the x-ray attenuation coefficients from the NIST XCOM database¹⁴ and performed least-squares analyses to determine an equivalent Al thickness at each tube voltage, so that the calculated attenuations by the added 4–9 mm Al agreed with the measurement data in Fig. 2. The optimal Al thickness along the central ray was 6.27, 6.31, 6.35, and 6.45 mm at 80, 100, 120, and 140 kV, respectively, and the calculated attenuations by the added 4–9 mm Al agreed with the attenuation measurements within 0.47%. With the filtration (2 mm graphite, 6.27–6.45 mm Al, and 76 μ m copper) along the central ray, we then used the measured bow-tie attenuation data to

FIG. 5. Offset levels of the X-Scan 0.8c4-307 detector readings when there was no radiation exposure.

FIG. 6. Tests of detector readings at 80 kV with the x-ray tube parked at the 12 o'clock position and no bow-tie in the beam path: (a) detector readings from five repeated exposures at 20 mA, (b) detector readings from eight exposures at 10, 20, 40, 60, 80, 100, 120, and 130 mA; and (c) ion chamber measurements at 10, 20, 40, 60, 80, 100, 120, and 130 mA. Data in (a) and (b) were offset-subtracted and normalized to 10 mA, but were not corrected for the fan beam angle related beam intensity change.

obtain an equivalent Al thickness at each fan beam angle/tube voltage combination. For the x-ray beam at $\theta \neq 0^\circ$ passing a flat filter obliquely, its path length through the filter was larger than the filter thickness, and this effect was accounted for using an effective thickness equal to $\text{thickness}/\cos\theta$.

2.C. GEANT4-based human cadaver dosimetry simulations

In a previous study, we performed *in vitro* dose measurements in a human cadaver (height 183 cm, weight 67.5 kg) scanned on the GE CT scanner.¹⁶ Six 0.6-cm³ thimble chambers were placed in the stomach, liver, paravertebral gutter, ascending colon, left kidney, and urinary bladder. Point doses were measured in nine helical (pitch 1.375) abdomen/pelvis scans with a large scan FOV (50 cm), the body bow-tie, a nominal beam width (NT) of 20 mm, a scan length of 47.4 cm (including the overscans), 100 or 120 kV, constant tube current

or longitudinal tube current modulation (TCM). The results are reported in Ref. 16. A head-to-leg CT scan was also made before inserting thimble chambers in the cadaver torso, and the acquired DICOM images (576 slices, slice thickness 2.5 mm) are used for Monte Carlo simulations of cadaver CT scans.

To simulate the above abdomen/pelvis cadaver scans, we adapted the previously developed GEANT4-based Monte Carlo simulation program¹⁵ to model a GE Lightspeed Pro 16 CT scanner with the x-ray spectra and flat and bow-tie filters described earlier in this paper, a focal spot size of 0.9×0.7 mm², a scan FOV of 50 cm, a source to isocenter distance of 541 mm, and a nominal beam width of 20 mm. A voxelized phantom was constructed using 576 image slices (slice thickness 2.5 mm), among which 179 slices were from one of the above abdomen/pelvis scans, and the others were from the head-to-leg scan. The patient table was included in the DICOM images. The abdomen/pelvis scan images contained metal artifacts due to the metal parts of the thimble

FIG. 7. Detector readings at 80–140 kV, 20 mA with the x-ray tube parked at the 12 o'clock position and no bow-tie in the beam path: (a) detector readings and (b) relative to the isocenter ratio. Data were offset-subtracted, but were not corrected for the x-ray source to detector pixel distance related beam intensity change.

chamber assemblies and connection cables. Before using these images in simulations, we performed artifact correction, based on the head-to-leg scan images. The trial-and-error determined thresholds were used to select the artifact contaminated areas, and their CT numbers were corrected with the average Hounsfield unit (HU) of the same areas in the head-to-leg scan images. Figure 3 shows an image slice before and after the artifact correction.

In constructing the voxelized phantom of $512 \times 512 \times 576$ voxels, the parameterization fast navigation algorithm included in the GEANT4 simulation toolkit (version 9.5)¹⁷ was used to reduce computer memory usage and to speed up the simulation. The CT numbers were converted to materials based on a method by Schneider *et al.*¹⁸ The HU values between -1000 and 1524 were converted to 24 elemental compositions and about 250 mass densities, based on the algorithm described

in Ref. 18. The material compositions and densities used for the other HU values are listed below: (a) those of HU = 1524 were used for HU between 1525 and 2000; (b) Al was used for HU between 2001 and 2650; and (c) copper was used for HU above 2650.

The helical CT scans were simulated in a step-and-shoot manner. The beam was equally irradiated at an interval of gantry rotation of 1° and patient table translation of $76.4 \mu\text{m}$ (pitch $\times$ NT/360). All simulations were performed with the low energy electromagnetic physics with energies above 250 eV (Livermore library).¹⁷ In addition, the active volumes of six thimble chambers were identified from the DICOM images, and were filled with a material of HU = 45 with a mass density of 1.059 g/cm^3 . Energy depositions in the corresponding 203–270 voxels (volumes $0.48\text{--}0.64 \text{ cm}^3$) were accumulated from the simulations. For simulating the TCM scans in Fig. 4, energy depositions were weighted by the tube current at the position of beam irradiation. Each cadaver scan was simulated with a sample of 2.7×10^{10} or more photons. The results at each tube voltage were normalized by a ratio,

$$\text{NF} = \frac{D_{\text{in-air,measured per mAs}}}{D_{\text{in-air,simulated per photon}}} \times \text{mAs}_{\text{total}}, \quad (3)$$

and were compared with the measurement data,¹⁶ where $\text{mAs}_{\text{total}}$ was the total tube-current product (mAs) of a helical CT scan. In Eq. (3), $D_{\text{in-air}}$ was the air cavity dose assessed in a single axial scan with a CT pencil chamber (active length 10 cm, volume 3 cm^3) placed at the isocenter and with the body bow-tie in the beam path, but without phantom or patient table in the beam path. The nominal beam width was 20 mm.

3. RESULTS

3.A. Detection of bow-tie attenuation profile

Figure 5 shows typical offset levels of the linear array detector readings when there was no radiation exposure. The offset changed slightly across six modules and 384 pixels.

FIG. 8. Primary beam profiles measured using the linear array x-ray detector placed on the rotation axis of the GE Lightspeed Pro 16 CT scanner.

Fig. 9. Illustration of bow-tie attenuation measurements using Eq. (2) with the x-ray tube parked at the 12 o'clock position: (a) a time sequence of the readings at the isocenter pixel without bow-tie in the beam path (air scan), (b) a time sequence of the readings at the isocenter pixel with the body bow-tie in the beam path, (c) radiation signals of 384 pixels in the bow-tie and air scans, and (d) radiation signal ratios of the bow-tie to air scans. The radiation exposure technique was 140 kV, 20 mA, and 1 s exposure time. All data were not corrected with nonlinear detector gain, but those of (c) and (d) were corrected with offset subtraction.

Figures 6(a) and 6(b) illustrate the offset-subtracted detector readings at 80 kV and 10–130 mA with the x-ray tube parked at the 12 o'clock position and no bow-tie in the beam path. The tube current linearity was verified with ion chamber (10x5-3CT, Radcal, Monrovia, CA) measurements at the isocenter, as shown in Fig. 6(c). For comparison, all data in Figs. 6(a) and 6(b) were normalized to 10 mA. For five repeated exposures at 20 mA, Fig. 6(a) shows small (up to 1%) fluctuations in the detector readings. When the tube current was changed, Fig. 6(b) shows that the detected radiation signals were approximately proportional to tube current, and the deviations were up to 10%, so that the signal-level related nonlinearity in detector gain was small. The nonlinearity was further examined at 80–140 kV and 20 mA without a bow-tie in the beam path. Figure 7(a) shows that the detector readings substantially increased with tube voltage. When the data were normalized by the isocenter pixel readings, the relative to isocenter ratios in Fig. 7(b) had small variations with 80–140 kV. Compared to the central ray, the peripheral beam obliquely passed through flat filters and was attenuated more. As x-ray attenuation usually decreases with the increased energy, we would expect that

the relative intensity in Fig. 7(b) should increase with tube voltage. However, two groups of 80 and 100 kV were somehow flipped. Their difference (up to 2.1%) might be an indication of a signal-level related nonlinearity in detector gain. In addition, the linear array x-ray detector was used to measure primary beam profiles on the axis of rotation of the GE Lightspeed Pro 16 CT scanner and the results of two nominal beam widths are shown in Fig. 8 with the heel effect clearly in evidence. The full width at half maximum (the magnitude at $z = 0$) was, respectively, 12.8 and 21.6 mm for NT = 10 and 20 mm, close to the CT manufacturer's specifications of 11.7 and 20.5 mm for these beam collimations.

Figure 9 shows the results obtained at 140 kV, 20 mA, 1 s exposure time, and the x-ray tube parked at the 12 o'clock position. Two time sequences of the isocenter pixel showed sharp signal rises and falls at the starts and ends of radiation exposures, and flat signal readings during the exposures. As expected, the radiation signal with the body bow-tie was less than that without a bow-tie. Although radiation signals decreased substantially near the boundaries of detector modules, this pattern nicely canceled out in the bow-tie to air scan

FIG. 10. Body bow-tie attenuations evaluated at 80, 100, 120, and 140 kV on a GE Lightspeed Pro 16 CT: (a) attenuation [see Eq. (2)] versus fan beam angle and (b) equivalent Al thickness versus fan beam angle.

ratios [see Eq. (2)], in which the fan beam angle related beam intensity change also canceled out. Figure 9(d) was close to the body bow-tie attenuation profile across the scan FOV, although the data were not corrected for the signal-level related detector gain in Fig. 6(b).

On the GE CT scanner, we performed both air and body bow-tie scans at 80, 100, 120, and 140 kV with the x-ray tube parked at the 12 o'clock position. The radiation detections by the linear array x-ray detector were corrected with offset subtraction and nonlinear detector gain. As shown in Eq. (2), the ratio of the radiation signals of the bow-tie to air scans was the bow-tie attenuation. The results at 80–140 kV are shown in Fig. 10(a). As expected, the attenuation decreased with tube voltage. We evaluated an equivalent Al thickness at each kV/fan beam angle combination, based on the TASMICS spectral model¹³ and the NIST XCOM data.¹⁴ The results are presented in Fig. 10(b), and were used in GEANT4-based Monte Carlo simulations.

3.B. Point doses sampled in the human cadaver

In a previous study, we found that the dose measurement using 0.6 cm³ volume thimble chambers placed in the human cadaver scanned with a large helical pitch (1.375) was sensitive to the x-ray tube starting angle.¹⁶ Therefore, in this study, the angle was taken into account in the cadaver simulations. We then obtained the soft tissue doses sampled in the stomach, liver, paravertebral gutter, ascending colon, left kidney, and urinary bladder of the cadaver scanned at 100 or 120 kV, using either constant tube current or z axis TCM (Fig. 4). The results were then compared with the dose measurements reported in Ref. 16. In Table I, the measurement data of the TCM scans¹⁶ were recalculated with the average tube current (435.5 mA) of the entire scans. For 18 dose entries in (a), (c), and (d), the measurement data of 2 or 3 tube start angles had an absolute deviation (from the average reported in Table I) of 0.4% (median) or 3.8% (maximum).¹⁶ For 24 entries, the difference (in magnitude) between the simulation results and the dose

measurements¹⁶ was 8.9% (median) or 25.9% (maximum). These deviations included the measurement errors of seven dosimetry units [six for point-dose measurements and one for $D_{\text{in-air}}$ measurements used in Eq. (3)], the dose variations of 2 or 3 tube start angles, the uncertainty in the f -factor (9.37 mGy/R) used in Ref. 16, and the errors involved in the simulation chain, which were difficult to assess with the current computing power. The statistical errors of the simulation results were within 0.6%, evaluated with multiple (>50) repeated simulations.

4. DISCUSSION AND CONCLUSIONS

This study demonstrates a new method to measure CT bow-tie attenuation profile using a linear array x-ray detector, which is generally available from Detection Technology (<http://www.deetee.com>) with a cost of a typical CT dosimeter. This method eases both measurement work on CT scanner and data analysis. In addition, the use of Eq. (2) does not require the SDD related beam intensity correction, and the pixel-to-pixel detector gain variation canceled out. The signal-level related detector nonlinearity is small [up to 10%, see Fig. 6(b)] and can also be taken into account in data analysis. Consequently, the absolute bow-tie attenuation $F(\theta)$ can be accurately evaluated.

This method was applied on a clinical CT scanner, and the obtained body bow-tie attenuations were used in GEANT4-based cadaver dosimetry simulations. For 24 point doses sampled in six organs of the cadaver scanned with three protocols using 100 or 120 kV, a large helical pitch (1.375), distinct x-ray tube starting angles (resulted from the actual CT scans), constant tube current or TCM (Fig. 4 with drastic mA change), the difference between the simulation results and the measurements on the CT scanner¹⁶ was 8.9% (median) or 25.9% (maximum).

When Eq. (2) is used to measure bow-tie attenuation, two separate radiation exposures with and without bow-tie in the beam path are required. Or, one can use just one radiation exposure to derive the relative bow-tie attenuation,

TABLE I. Comparison of soft tissue doses (mGy/100 mAs) sampled in six organs of the human cadaver scanned with the large FOV, the body bow-tie, NT = 20 mm, and helical pitch = 1.375 on a GE Lightspeed Pro 16 CT scanner. Scan length was 47.4 cm, including the overscans. The measurement data are reported in Ref. 16, and those of the TCM scans were recalculated with the average tube current (435.5 mA) of the entire scans. The x-ray tube starting angles occurred in the measurements and used in simulations are listed in parentheses.

Entry	Liver	Kidney	Colon	Stomach	Bladder	Gutter
(a) 100 kV, constant mA						
Measurement (293°, 294.1°, 294.8°)	5.91	4.03	5.88	4.85	6.99	6.22
Simulation (294°)	5.50	5.07	6.29	5.69	5.74	5.53
Difference (%)	-7.0	25.9	7.0	17.3	-17.9	-11.1
(b) 120 kV, constant mA						
Measurement (290°)	9.89	7.69	9.72	8.40	11.38	10.22
Simulation (290°)	9.09	8.58	10.20	9.40	9.18	9.23
Difference (%)	-8.1	11.6	5.0	11.9	-19.3	-9.7
(c) 120 kV, constant mA						
Measurement (109.6°, 114.3°)	9.19	7.41	8.72	6.70	10.30	10.25
Simulation (112°)	8.94	8.80	8.73	8.05	9.92	9.57
Difference (%)	-2.8	18.7	0.1	20.1	-3.7	-6.6
(d) 120 kV, z axis TCM (Fig. 4)						
Measurement (106.8°, 108.3°, 109°)	7.71	7.38	7.27	5.90	14.41	8.84
Simulation (108°)	7.15	7.05	6.82	6.54	13.92	7.84
Difference (%)	-7.3	-4.5	-6.1	10.7	-3.3	-11.3

$$\frac{F(\theta)}{F(0^\circ)} = \frac{S_{\text{bow-tie}}(\theta)}{S_{\text{bow-tie}}(0^\circ)}, \quad (4)$$

with both bow-tie and linear array x-ray detector in the beam path. X-ray tube is in a parked position or rotates one or more revolutions. Data analysis for the parked tube is straightforward. For the rotating x-ray tube, one can use a CT scan with a long gantry rotation time, such as 1 or 2 s/rotation. Then one can identify an exposure time window in which the source is facing the detector's x-ray window and the central ray is perpendicular to the detector surface. As the detector sampling time can be as short as 0.24 ms, there are 11.6 or 23 samples within a rotation angle window of 1° when rotation time is 1 or 2 s, respectively. The data analysis steps for both the parked and rotational x-ray sources are listed below: (a) perform detector offset subtraction, (b) perform nonlinear gain correction, (c) perform pixel-by-pixel gain correction, (d) perform the source to detector pixel distance correction using the inverse square law, and (e) compute relative bow-tie attenuation $F(\theta)/F(0^\circ)$. This may be explored in a future study.

Our work also demonstrates multiple challenges in developing a patient dose evaluation software for CT. One needs to assess beam quality, characterize the bow-tie filter, and model the patient's body. While a voxelized phantom may be accurate in patient modeling, it requires a large computer memory. The simulation results need to be verified with the measurements on CT scanner. Ion chambers are accurate for this purpose, but the related metal artifacts in the reconstructed images need to be corrected for the images to be used in simulation. The peripheral CT dose profile presents oscillation because of beam divergence and subject attenuation. Consequently, point dose in the peripheral region may be sensitive to x-

ray tube trajectory and the tube start angle needs to be taken into account in the simulation. Our experience may provide insights for the community.

In conclusion, the described method allows noninvasive and fast bow-tie filter characterization. This study demonstrates the accuracy of the method and its application in Monte Carlo simulation based CT dose calculations.

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