Experimental validation of a method characterizing bow tie filters in CT scanners using a real-time dose probe

Sarah E. McKenney and Anita Nosratieh

Department of Radiology, University of California, Davis Medical Center, Rm. 0505, ACC Ellison Bldg., 4860 Y St., Sacramento, California 95817 and Department of Biomedical Engineering, University of California, Davis, Rm. 2303, GBSF, 451 E. Health Science Dr., Davis, California 95616

Dale Gelskey

Diagnostic Imaging Specialists Corporation (DISC), 163 St. Malo Street, St. Malo, Manitoba R0A 1T0, Canada

Kai Yang

Department of Radiology, University of California, Davis Medical Center, Rm. 0505, ACC Ellison Bldg., 4860 Y St., Sacramento, California 95817

Shin-ying Huang, Lin Chen, and John M. Boone^{a)}

Department of Radiology, University of California, Davis Medical Center, Rm. 0505, ACC Ellison Bldg., 4860 Y St., Sacramento, California 95817 and Department of Biomedical Engineering, University of California, Davis, Rm. 2303, GBSF, 451 E. Health Science Dr., Davis, California 95616

(Received 23 June 2010; revised 12 January 2011; accepted for publication 13 January 2011; published 18 February 2011)

Purpose: Beam-shaping or “bow tie” (BT) filters are used to spatially modulate the x-ray beam in a CT scanner, but the conventional method of step-and-shoot measurement to characterize a beam’s profile is tedious and time-consuming. The theory for characterization of bow tie relative attenuation (COBRA) method, which relies on a real-time dosimeter to address the issues of conventional measurement techniques, was previously demonstrated using computer simulations. In this study, the feasibility of the COBRA theory is further validated experimentally through the employment of a prototype real-time radiation meter and a known BT filter.

Methods: The COBRA method consisted of four basic steps: (1) The probe was placed at the edge of a scanner’s field of view; (2) a real-time signal train was collected as the scanner’s gantry rotated with the x-ray beam on; (3) the signal train, without a BT filter, was modeled using peak values measured in the signal train of step 2; and (4) the relative attenuation of the BT filter was estimated from filtered and unfiltered data sets. The prototype probe was first verified to have an isotropic and linear response to incident x-rays. The COBRA method was then tested on a dedicated breast CT scanner with a custom-designed BT filter and compared to the conventional step-and-shoot characterization of the BT filter. Using basis decomposition of dual energy signal data, the thickness of the filter was estimated and compared to the BT filter’s manufacturing specifications. The COBRA method was also demonstrated with a clinical whole body CT scanner using the body BT filter. The relative attenuation was calculated at four discrete x-ray tube potentials and used to estimate the thickness of the BT filter.

Results: The prototype probe was found to have a linear and isotropic response to x-rays. The relative attenuation produced from the COBRA method fell within the error of the relative attenuation measured with the step-and-shoot method. The BT filter thickness estimates resulting from the dual energy scans on the breast CT system were equivalent to the manufacturing specifications. The clinical CT evaluation produced data conceptually similar to previous computer simulations and plausible relative attenuation profiles were observed.

Conclusions: The COBRA method is a fast and accurate method for BT filter characterization, which requires a simple experimental setup in a clinical environment. Because of the ease of data acquisition, multienergy scans can be acquired which allow characterization of the BT filter thickness. © 2011 American Association of Physicists in Medicine. [DOI: [10.1118/1.3551990](https://doi.org/10.1118/1.3551990)]

Key words: bow tie filter, real-time dosimetry, CT

I. INTRODUCTION

CT dosimetry relies increasingly on Monte Carlo simulations. Within the past decade, the high computational demands of Monte Carlo dosimetry techniques are being met with affordable and easily accessible computer hardware.

Despite an increase in processing power, the accuracy of a dosimetry simulation primarily relies on the researcher’s ability to characterize the physical parameters of a CT scanner’s geometry and x-ray beam properties.¹

In commercial whole body scanners, a beam-shaping filter

or “bow tie” (BT) filter is placed in the x-ray beam to equalize the fluence incident on the detector and to reduce the radiation dose to a patient.² The BT-filtered beam reduces the dynamic range demands of a detector and improves the resulting images,³ specifically in terms of contrast-to-noise ratio, scatter-to-primary ratio, CT number accuracy, and uniformity. The added benefit of dose reduction⁴ is particularly noticeable at the periphery of a patient.⁵ While there are beam-shaping filters that intentionally shape the x-ray beam in the z -direction,⁶ all clinical whole body scanners use BT filters with a constant profile in the z -direction, with some minor beam divergence. This is not to say that the x-ray profile is flat along the z -dimension; indeed, most CT beam profiles along the z -axis of the beam show evidence of the heel effect. It is possible to apply filtration to the scanner that reshapes the beam profile along z to compensate for the heel effect; however, the filtration employed for z -dimension beam correction is constant as a function of the fan angle. Therefore, these elements of beam filtration are considered to be part of the x-ray tube inherent filtration; that is, it is constant in the fan angle direction.

The geometry and composition of a BT filter is typically proprietary information. Unless this information is provided,⁷ medical physicists are unable to accurately model the x-ray beam incident on the patient in commercial CT scanners.⁸ Recognizing this, a method to characterize a CT scanner's BT filter using a real-time exposure meter was previously proposed and demonstrated using computer simulation.⁹ The characterization of bow tie relative attenuation (COBRA) method outlines a protocol to quickly and accurately describe $F(\theta)$, the relative attenuation of a BT filter across the fan angle θ of the scanner.

Accurate Monte Carlo studies in CT rely on knowledge of the x-ray energy-dependent modulation of the x-ray beam along the fan angle. Therefore, knowledge of the overall attenuation of the x-ray beam $F(\theta)$ as a function of fan angle θ for a polyenergetic x-ray spectrum is not sufficient. Consequently, the COBRA method involves a final step, whereby the BT filter characterization of angle-dependent attenuation at different x-ray beam energies (e.g., 80–140 kVp) is used with basis material decomposition techniques to estimate the physical thickness of the bow tie filter. Knowing the estimated thickness of the bow tie filter for one or two basis materials as a function of fan angle then allows the investigator to take the energy dependence of the bow tie filter into account. Even if the selection of basis materials [such as aluminum and polymethyl methacrylate (PMMA)] is not a correct physical depiction of a given vendor's bow tie filter composition, the derived thickness information will provide the same energy-dependent attenuation properties as the characterized BT filter, as this is the basic tenant of basis material decomposition techniques.¹⁰

In this paper, experimental measurements are described using the COBRA technique with a prototypical dose probe in both a custom-designed CT system and a commercial whole body CT system.

FIG. 1. Geometry of the CT scanner in a rotating frame of reference such that the gantry and x-ray tube appear to be stationary while the probe appears to rotate around isocenter. As the fan angle $\theta(t)$ varies, the x-ray beam incident on the probe is modulated by the varied thicknesses of the BT filter. As the x-ray source rotates, the probe's filtered signal train $M_1(\alpha)$ has a maximum signal at $\alpha(t)=0$; a local maxima also occurs at $\alpha(t)=\pi$, since the beam is minimally reduced by the BT filter.

II. THEORY REVIEW

The following is a brief overview of the COBRA method for determining the relative attenuation properties of a beam-shaping filter in CT; for a more detailed review, refer to Boone.⁹ In order for the method to work properly, the entire active area of the x-ray probe must be completely enveloped within the collimated x-ray beam and no other objects, such as a patient table, can be in the beam's path during the acquisition. The geometry of the CT scanner is shown in Fig. 1, where the frame of reference has been centered on the x-ray source such that the source appears to be stationary at position (x_t, y_t) and the real-time dose probe appears to rotate a distance r around isocenter with a position of (x_p, y_p) as determined by the gantry angle $\alpha(t)$. It is assumed that the source-to-isocenter distance s is known and is fixed and that the anode-cathode direction is parallel to the z -direction.

The x-ray fluence detected by the probe at time t is dependent on the distance of the probe from the x-ray tube, denoted by $g(t)$. When no BT filter is present, it is assumed that the x-ray probe's output $M_0(\alpha)$ is independent of the angular incidence of the x-rays. When a bow tie filter is added to the system, the probe's output with filtration $M_1(\alpha)$ is equal to $M_0(\alpha)$ at $\alpha(t)=0$ and $\alpha(t)=\pi$. At these two positions, the relative attenuation of the probe's signal due to the bow tie filter at a fan angle of θ , $F(\theta)$, is unity [i.e., $F(\theta)=1$]. Note that

$$F(\theta) = \frac{M_1(\alpha)}{M_0(\alpha)}. \quad (1)$$

Given $M_1(0)$ and $M_1(\pi)$ and assuming that the effects of focal spot size, off-focal radiation, and detective volume are negligible, it is possible to use the inverse square law to deduce the output in the absence of a BT filter over all $\alpha(t)$ such that

$$M_0 = I_0 \left(\frac{s}{g(\alpha)} \right)^2, \quad (2)$$

where I_0 is the probe's output at isocenter. Consequently, it is possible to measure the relative attenuation of a BT filter

through the analysis of the real-time dosimetric output of a CT scanner as it executes a minimum of one complete rotation.

Assuming that the same BT filter is used when x-rays are generated at different peak voltages, a least-squares algorithm can be used to estimate the filter's thickness. Using dual energy basis decomposition,¹¹ the filter is assumed to consist of one or two basis materials, typically a metal and a plastic,¹⁰ where the linear attenuation coefficient $\mu_{\text{material}}(E)$ is known.¹² The theoretical relative attenuation $\Gamma(\theta)$ of x-rays generated at a single peak energy V is described as

$$\Gamma_V(\theta) = \frac{\int_{E=0}^{E_{\text{Max}}} k(E) \psi^V(E) e^{-\mu_a(E)a(\theta) - \mu_b(E)b(\theta)} dE}{\int_{E=0}^{E_{\text{Max}}} k(E) \psi^V(E) dE}, \quad (3)$$

where $a(\theta)$ and $b(\theta)$ are the thicknesses of the material, $\psi(E)$ is the x-ray energy fluence, and $k(E)$ is a factor that converts photon fluence into the probe's measurement units, such as mGy/s.

The thicknesses of the filter material can be estimated with the least-squares method from an overdetermined system, consisting of measurements made at multiple x-ray tube peak voltages, such that

$$\chi^2 = \sum_{V=V_1}^{V=V_n} [F_V(\theta) - \Gamma_V(\theta)]^2 \quad (4)$$

is minimized by iterating over possible values of $a(\theta)$ and $b(\theta)$. The end result is an estimation of the bow tie filter's thickness as a function of fan angle. The extent of the utility of the relative attenuation measurements for Monte Carlo simulation depends greatly on the accuracy of the scanner's spectral characterization at isocenter ($\theta=0$). Note that this method assumes that $a(0)=b(0)=0$; consequently, the inherent filtration of the system, including contributions from the bow tie filter at $\theta=0$, is included in $M_0(\alpha)$.

III. MATERIALS AND METHODS

III.A. System setup: Breast CT (bCT) scanner

Probe validation and preliminary relative attenuation data sets were acquired on a prototype dedicated bCT scanner.¹³ This scanner allowed for continuous rotation of the x-ray tube (model MXR-160HP/20, Comet, Flamatt, Switzerland) up to an angular position of $\alpha=415^\circ$; the x-ray tube could also be held fixed at any angular position for a stationary exposure. The beam was filtered with 0.2 mm copper; the x-ray generator (model CP 160/1, Gulmay, Chertsey, United Kingdom) produced a ripple $<0.5\%$.

While standard operation of the bCT scanner did not include a BT filter; a removable Teflon filter was custom-designed and machined for the purposes of reducing radiation dose to the periphery of a breast (see Fig. 2).^{14,15} The BT filter was designed for breasts with a diameter of 14 cm using an in-house simulation. There was no beam modulation in the z -direction and the filter was symmetric about the source beam's central ray. The filter was positioned 153.0 ± 5.5 mm from the x-ray focal spot, as determined from three individual measurements ($N=3$).

FIG. 2. Photograph of the Teflon bow tie filter mounted in the breast CT scanner. The filter has a thickness ranging from 0.2 mm at its center to 62.8 mm along its edge. The filter is bilaterally symmetric and has a uniform thickness along the z -direction.

III.B. System setup: Clinical CT scanner

The COBRA technique was also used to characterize the BT filter of a commercial clinical CT scanner (model AS+, Siemens Medical Systems, Florsheim, Germany). This CT system had $s=590$ mm and its BT filter's shape and composition were proprietary. The beam's ripple was assumed to be 5%.

III.C. Probe characterization

Real-time measurements were obtained using a prototype dose probe from Diagnostic Imaging Specialists Corp. (DISC). DISC also provided a preamplifier, allowing for variable gains of $\times 1$, $\times 10$, $\times 50$, $\times 100$, and $\times 1000$. An analog to digital converter (ADC) (Data Translation model DT9804, Waltham, MA) converted the probe's output into a digital signal and allowed for additional gain levels of $\times 1$, $\times 2$, $\times 4$, and $\times 8$. The digital signal was acquired using custom-written data acquisition software. The probe's active area was a small cylindrical solid-state scintillator with a height of 7 mm and a diameter of 5 mm (see Fig. 3); the manufacturer's reported sampling rate was under 1 kHz.

On the bCT scanner, the probe's position was determined relative to the scanner's rotational isocenter (see Fig. 4). Pre-

FIG. 3. Projection image of the active volume of the prototype x-ray probe at isocenter.

FIG. 4. General experimental scheme of the real-time dose probe with a CT scanner. The bCT x-ray tube and detector were either held stationary or rotated up to 415° . The dose probe had a fixed position at the edge of the imaging field of view for COBRA acquisitions. The probe could also be manually shifted for conventional profile measurements.

viously, a geometric calibration method that estimates the scanner's isocenter and source-to-isocenter distance s was developed for the bCT system.¹⁶ This was used for the initial placement of the probe. A series of 360° scans were made with a 120 kVp/7 mA x-ray source and the probe was manually translated until its output was relatively constant; this position was defined as isocenter. Note that this method for determination of the system's isocenter was not necessary for the COBRA method if the isocenter was previously defined. On the whole body clinical CT scanner, the laser alignment lights were assumed to correctly demark isocenter and the x-ray tube's plane of rotation.

III.C.1. Probe isotropy

To quantify the prototype probe's dependence on the angular position of the x-ray source, the probe was positioned at the scanner's isocenter and then it was manually rotated 360° about its longitudinal axis in 90° intervals. At each position, the probe was exposed for 5 s to a stationary x-ray source generated by a 120 kVp/7 mA tube voltage and current. The probe was rotated a total of four times such that there were four measurements collected at each angle.

III.C.2. Probe linearity

Assessment of the linearity of the prototype probe's response to an incident x-ray beam was performed. The x-ray source was held stationary and the probe was positioned at isocenter. For each x-ray tube voltage (80 and 120 kVp), the tube current was increased from 0 mA, in 1 mA intervals, to the maximum current allowed. At each tube current, the probe was exposed for 5 s to the resulting x-ray beam. The probe's output was cropped to exclude the probe's response lag; an observation point was the average of a cropped trace. This method was repeated for permutations of the probe's preamplifier's gain ($\times 50$ and $\times 100$) and the ADC's gain ($\times 1$, $\times 4$, and $\times 8$).

FIG. 5. Experimental setup for step-and-shoot measurements of a fan beam's relative attenuation. The probe's position was incremented from isocenter ($\Delta x = 1$ cm) with steps in the fan angle of $\Delta \theta \approx 1^\circ$. Measurements were acquired at each position with and without the bow tie filter in place.

To convert the raw output signal into accurate values of air kerma, a RadCal 9010 general-purpose ion chamber was fixed at isocenter and the x-ray tube voltage and the current was varied as for the assessment of the probe's linearity. The ion chamber was exposed for approximately 30 s and an exposure rate (in R/min) was measured. Assuming that the chamber's response was linear, a simple transformation constant, nominally $W = 8.76$ mGy/R,¹⁷ allowed the probe's output voltage signal to be converted into air kerma (in $\mu\text{Gy/s}$).

III.D. Bow tie filter characterization

III.D.1. Profile estimation from a stationary source

To validate the COBRA method, dose profiles of the bCT scanner's x-ray beam were measured manually with and without the BT filter (see Fig. 5). The probe was translated ± 12 cm along an axis perpendicular to the beam's central ray and passing through the system's isocenter. To reduce the effects of systematic drift, the probe was first translated in one direction and measurements were taken at $x = 2n$ distances from isocenter, with $0 \leq n \leq 12$. Once the final position was reached, the probe was translated back to its starting position with measurements made at $x = 2n - 1$ distances from isocenter. For each observation, the probe was irradiated 5 s by an x-ray beam generated at 120 kVp, 7 mA. As with the linearity measurements, the probe's output signal trace was cropped and averaged to a single observation point. The relative attenuation at a given fan angle θ was calculated as the ratio of the probe's output with the BT filter present (M_1) to without the BT filter present (M_0) as described in Eq. (1), where $\theta = \tan^{-1}(x/s)$.

III.D.2. Profile estimation from a rotating source

The COBRA method detailed by Boone⁹ was performed on the prototype bCT scanner, described previously (Sec. III A). The DISC probe was positioned 8 cm from isocenter, at the edge of the field of view (FOV) of the bCT scanner, as

depicted in Fig. 4. To estimate the bow tie thickness, dual energy scans were conducted at 120 kVp/7 mA and 80 kVp/11 mA. To assess the accuracy of the unfiltered beam measurements, two more identical scans were made without the BT filter. The maximum allowable source rotation was used (415°) and the probe was irradiated approximately 17.5 s as a scan was executed. Because the probe was highly sensitive to repositioning, the probe's position was held fixed throughout all four runs. The data was collected at 100 samples per second.

While the effect of the inverse square law was measured directly when the BT filter was removed from the setup, the effect was also modeled using only the BT-filtered data, as outlined in Sec. II. The maxima of the scans were estimated using a simple peak-finding algorithm that stepped along the observation points of the probe. Essentially, peak intervals were identified and the maximal value was the average of the data points that were one standard deviation above the average of the interval; this diminished the noise contributions to the maxima while avoiding its underestimation. The position of the maxima was the average position of the data points contributing to the maximal value. The period of rotation $\hat{\tau}$ was computed from the difference of the two maxima. The gantry angle $\hat{\alpha}$ was calculated for each observation point from the period $\hat{\tau}$ and observation time of the first maximum $\hat{\delta}$ such that

$$\hat{\alpha} = \frac{2\pi(t - \hat{\delta})}{\hat{\tau}}, \quad (5)$$

where t is an observation's timestamp and the caret notation distinguishes an estimated value from a direct measurement.

The distance between the probe and isocenter was estimated ($\hat{r}$) using source-to-isocenter distance s and the measured signal $M_1(\alpha)$ at $\hat{\alpha}(t)=0$ and $\hat{\alpha}(t)=\pi$ from

$$\frac{M_1(\pi)}{M_1(0)} = \frac{(s-r)^2}{(s+r)^2}$$

such that

$$\hat{r} = s \left(\frac{1 - \sqrt{\frac{M_1(\pi)}{M_1(0)}}}{1 + \sqrt{\frac{M_1(\pi)}{M_1(0)}}} \right). \quad (6)$$

Using the probe's position, the flux at isocenter $\hat{I}_0$ was estimated as

$$\hat{I}_0 = M_1(\pi) \left[\frac{s - \hat{r}}{s} \right] \quad \text{or, equivalently,} \quad \hat{I}_0 = M_1(0) \left[\frac{s + \hat{r}}{s} \right]. \quad (7)$$

The probe's position was also used to estimate distance from the source $\hat{g}(\alpha)$ by

FIG. 6. Photograph of the experimental setup of the prototype x-ray probe in the clinical CT scanner.

$$\hat{g}(\hat{\alpha}) = \sqrt{s^2 + \hat{r}^2 - 2s\hat{r} \cos(\hat{\alpha})}. \quad (8)$$

Finally, using these estimated values, the unfiltered measurement $\hat{M}_0$ was modeled at all gantry angles as described in Eq. (2).

The estimated values of $\hat{M}_0$ and the BT-filtered measurements M_1 were used to calculate $F(\theta)$ as in Eq. (1), using the system's geometry (Fig. 1), to transform gantry angle into fan angle. The bow tie filter was assumed to be symmetric, such that $F(\theta) = F(-\theta)$. Finally, the estimates of relative attenuation were binned and averaged in quarter-degree intervals of fan angle; a smoothing spline was then fit to the results to better capture the smooth geometry of the BT filter.

After confirming that the model of the unfiltered source was acceptable, the relative attenuation function was used to estimate the thickness of the bCT bow tie filter. The TASMIP¹⁸ spectral model was used to calculate the relative attenuation at each fan angle. The TASMIP-generated spectra were calibrated from half value layer (HVL) measurements made on the bCT scanner at isocenter with a RadCal 9010 ion chamber. A single basis material least-squares algorithm [Eq. (4)] was used to estimate the thickness of a Teflon (CF₂) bow tie filter. The mass attenuation coefficients for Teflon were estimated using tabulated attenuation coefficients¹² and the photon fluence output was converted into kerma units (μGy). For each independent fan angle, a custom-written program varied the thickness of Teflon in 0.01 mm steps and the thickness that minimized the χ^2 value was selected.

III.D.3. Clinical CT applications

The COBRA method was evaluated using the body BT filter in a commercially available whole body scanner (model AS+, Siemens Medical Systems, Florsheim, Germany). The probe was positioned on a stand on the opposite side of the gantry from the patient table (see Fig. 6). The active area of the probe was placed at the edge of the FOV of the system. Scans at several kVp's were conducted. The scanning protocols for the abdominal scans, used to characterize the body filter, were 80 kVp/100 mAs, 100 kVp/100 mAs, 120 kVp/50 mAs, and 140 kVp/50 mAs. The slip ring construction of the scanner allowed for unlimited rotations of the source; consequently, a 10 s acquisition reflects approximately ten full

rotations of the gantry about the system's isocenter. Measurements were collected at 1000 samples per second.

The method of data analysis of the clinical CT scan was the same as for the bCT data, except that it was necessary to account for the much larger data set. The average period of the gantry rotation τ was computed from the difference between the absolute maxima of the signal train. The measured signal $M_1(\alpha)$, used for estimations of $\hat{r}$ and $\hat{I}_0$, was averaged with $M_1(\alpha+2\pi n)$ such that

$$\bar{M}_1(\alpha) = \frac{1}{N} \sum_{n=0}^{n=N} M_1(\alpha + 2\pi n), \quad (9)$$

where $\alpha = \{0, \pi\}$ and N is the number of gantry rotation periods. Estimates of $F(\theta)$ and BT filter thickness were determined as in Sec. III D 2. Following the work of Lehmann,¹¹ PMMA and aluminum were chosen as two basis materials. BT filter thickness was determined under the assumption that the filter was composed entirely of one material, either PMMA or Al.

IV. RESULTS

IV.A. Probe isotropy

In Fig. 7, the angular response of the prototype probe over 2π is shown to be quite uniform. The average percent error is 0.80% with a maximum of 0.92%, which is acceptable for the purposes of this study. It should be noted that if the probe is held at a fixed angle, the relative error over a single signal train reduces to 0.2%. The variation in probe response at different angular positions is most likely a result of both the positioning method used to obtain the measurements as well as the intrinsic properties of the probe.

IV.B. Probe linearity

The probe's response, as a function of tube current, is linear (Fig. 8), with $r^2=0.999$ at both 80 and 120 kVp. The linear relationship between the prototype probe's raw output signal, S (in V), and the ion chamber's air kerma rate, K (in

FIG. 7. Relative signal of the DISC probe as a function of the probe's angle for a stationary x-ray beam produced at 120 kVp/7 mA, $N=4$.

FIG. 8. Linearity of probe in the bCT scanner at varied values of tube current ($r^2=0.999$); observation points have a standard deviation of ± 1 mV.

$\mu\text{Gy/s}$), was determined to be $K(S)=3993.1S-3.6$ ($r^2=0.999$) at 80 kVp and $K(S)=3512.2S+4.6$ ($r^2=0.999$) at 120 kVp.

IV.C. Bow tie filter characterization

IV.C.1. Profile estimation from a stationary source

The fan beam profiles of the bCT scanner are depicted in Fig. 9. The dose rate precision is relatively good with an average standard deviation of $5.1 \mu\text{Gy/s}$ across all observation points. There was a translational positioning error of ± 1 mm, resulting in a fan angle error of $\pm 0.1^\circ$. The asymmetry of the beam profile, at angles greater than 12° , is the result of asymmetric collimation of the cone beam; this was verified by measuring the profile of the uncollimated beam. Relative attenuation analysis was limited to the imaging FOV, extending $\pm 8.5^\circ$. The relative attenuation of the bow tie filter was used as the gold standard for the evaluation of the COBRA method.

IV.C.2. Profile estimation from a rotating source

Figure 10 plots the registered measured air kerma rates for the four runs as a function of gantry angle. As expected, there are only two angular positions of the source where the filtered beam is unattenuated and matches the unfiltered

FIG. 9. Profile traces of the bCT scanner using the conventional method depicted in Fig. 5, corrected for inverse square law effect. M_1 traces the profile of the bow tie filter, while M_0 traces the unfiltered beam. Note that the interval of interest is $\pm 8.5^\circ$, the imaging field of view.

FIG. 10. Raw output of the probe's signal train at (a) 80 and (b) 120 kVp on the bCT scanner with the Teflon bow tie filter (black) and without the bow tie filter (gray). The unfiltered signal train was also estimated (not depicted in the figure) from the peaks of the measured signal with the bow tie filter. The estimated unfiltered signal was compared to the measured signal train in Fig. 11.

beam, when $\alpha(t)=0$ and π . As the source rotates through the rest of the gantry positions, the bow tie filter dramatically attenuates the x-ray beam.

Using the method described in Sec. III D 2, the probe's estimated position $\hat{r}$ from the two scans was 80.3 ± 0.5 mm from isocenter; this is in very good agreement with the manually measured position of 80.0 ± 1 mm from isocenter. Figure 11 is a plot of the correlation between the measured M_0 and estimated $\hat{M}_0$ air kerma rates at 80 and 120 kVp; additional Bland–Altman plots of the residuals are inset. While the estimated air kerma rates seem to be systematically higher than the measured rates, the average relative difference between the estimated and measured data is 1.25% at 80 kVp and 1.26% at 120 kVp. The estimates exhibit adequate correlation with the measured data; at 80 kVp, the $r^2=0.997$ and at 120 kVp, the $r^2=0.995$.

It is observed in Fig. 12 that the relative attenuation estimated using the COBRA method falls within the uncertainty of the measurements made using the more traditional method (summarized in Sec. III D 1). Figure 13 is a comparison of the estimated thicknesses to the known thickness as a function of fan angle. The COBRA estimates of BT filter thickness compare well to thicknesses estimated from (1) the computer-aided design (CAD) drawings of the filter or (2)

FIG. 11. Plot of the correlation between the measured and estimated air kerma rates for unfiltered beams at both (a) 80 and (b) 120 kVp. The line signifies $y=x$, the ideal relationship between the model and measurement of the unfiltered beam. Bland–Altman plots of the residuals are displayed as insets.

from the simulation used as a basis for the CAD drawings.¹⁴ The thickness estimates from the COBRA method had an average error of ± 1 mm. Because Teflon is highly scattering, the scatter-to-primary ratio dramatically increases with the thickness of the filter; consequently, the difference between the COBRA estimate and CAD thickness climbs to 12 mm at fan angles larger than 8° . Again, it should be noted that this phenomena occurred near the edge of the FOV (8.5°) of the detector.

FIG. 12. Comparison of the relative attenuation $F(\theta)$ of the two BT filter characterization methods for the bCT scanner at a tube voltage of 120 kVp.

FIG. 13. Comparison of BT filter thickness from the COBRA method to the computed thickness used for the filter's design as well as its final CAD drawing.

IV.D. Bow tie filter characterization: Clinical CT applications

An example of the raw signal train of the probe as the clinical CT gantry makes multiple revolutions is plotted in Fig. 14; the modeled signal without a BT filter $\hat{M}_0$ is also shown. The periodicity of the signal train reflects the rotation of the x-ray gantry; the consistency of peak maxima is indicative of the probe's stationary positioning. The probe position during the body scan was computed as $\hat{r} = 240.5 \pm 2.5$ mm. Figure 15 depicts the bow tie attenuation properties derived for the body BT filter and the data are shown for 80, 100, 120, and 140 kVp. The continuous gantry rotation provides a multitude of samples, all within a single 10 s acquisition; an average of 10,400 data points are plotted at each tube voltage. Figure 16 is a plot of the thickness of the body BT filter as estimated from the attenuation data.

V. LIMITATIONS

While relative attenuation measurements from the COBRA method match those made with the conventional step-and-shoot method, there are some limitations of the method. First, the algorithm that models M_0 could be more robust. Currently, $\hat{M}_0$ may differ significantly from the M_0 measurements, particularly for scans with a limited number of gantry rotations. A linear regression plot of the model vs measure-

FIG. 14. Raw output from the prototype probe in the clinical CT scanner for the body filter at 120 kVp (gray). The model unfiltered signal train is also included (black); samples were acquired at 1000 Hz.

FIG. 15. Estimated relative attenuation of the body filter on the clinical CT scanner before data have been binned and averaged.

ment data from the breast CT signal train (Fig. 11) reveals that the slope m differs significantly from $m=1$ at 120 kVp and that the intercept differs significantly from zero at both tube voltages when using a student's t -test. Ideally, the model and measurements should match at each data point. The most observable evidence of an imperfect model is a trend in the residuals of nonlinearity and non-normality in the Bland-Altman plots. These differences indicate errors in the estimates of both $\hat{I}_0$ and $\hat{g}(\alpha)$; these errors primarily result from slight errors in identifying the peak of the measured waveforms. Fortunately, the residual differences using the COBRA method are small enough to have only a minor impact on the estimate of $F(\theta)$. To achieve more robust results, the signal train can be measured over a larger number of gantry rotations. Alternately, an iterative model-based solution to the entire waveform may be sought, equivalent to one-dimensional iterative reconstruction techniques.

While estimations of the BT filter's thickness were accurate, these estimates are dependent on a well-characterized x-ray spectrum at $F(\theta=0)$. One limitation of the experimental setup was an inability to directly assess the HVL of the whole body scanner. Optimized characterization of BT filter thickness across multiple CT scanners would be more robust if accurate kVp and HVL measurements were available.^{19,20} While basis decomposition allows any material to be de-

FIG. 16. Estimated thickness of the bow tie filter for a body scan on the clinical CT scanner. Thicknesses were estimated for a filter made entirely of either PMMA (black circles) or Al (white circles).

scribed as a function of basis materials, the thickness estimation will likely be more accurate when the composition of the BT filter is known. Moreover, care must be given to avoid inaccurate solutions, from describing materials with a high atomic number, or unrealistic solutions, such as negative material thicknesses, when choosing basis materials.¹⁰

The COBRA method computes the BT filter thicknesses along the fan angle, but assumes that the scanner's filtration in the z -dimension is constant as a function of fan angle. Most CT scanners have a measurable heel effect in the z -dimension, but the shape of this along z is the same at all fan beam angles θ . Filters to compensate for the heel effect can be used and these would not compromise the accuracy of the COBRA method, as they would be invariant as a function of fan angle. In general, a consistent beam profile along z , as a function of fan angle, is a necessary requirement for artifact free CT imaging.

VI. DISCUSSION

This study describes the experimental validation of the COBRA method, where the theory was developed previously and was illustrated using computer simulation.⁹ The results in the current study demonstrate that a practical characterization of the BT filter attenuation and of the angle-dependent thickness can be made in the clinical environment.

It is anticipated that the primary utility of the COBRA method will be to characterize the angle-dependent thickness of one or two component BT filters for the purposes of Monte Carlo simulation studies. Another key advantage of this measurement technique is that since no proprietary knowledge is used in the characterization, the beam-shaping characteristics of bow tie filters across scanner models and manufacturers can be measured, discussed, and compared in the open literature.

The COBRA technique requires that a real-time probe be used with temporal resolution on the order of approximately 200 Hz–2 kHz. There are a number of high bandwidth dosimeter systems that are currently available commercially; in general, these systems are capable of importing the measured waveform directly into spreadsheet software. Currently available real-time dosimeters include both air ionization and solid-state systems. All dosimeters have energy dependencies and techniques for correcting for their energy dependence are necessary to achieve similar results across dosimeters. Historically, dose measurement has been performed using air ionization chambers; nevertheless, it is noted that the energy dependence of solid-state (scintillator) based dosimeters better represents the energy dependence of the detectors in the CT scanners. Thus, for solid-state detector systems, the computed attenuation curve $F(\theta)$ will be a good match to the actual response of CT detectors to the BT filter. The fact that the measured data are normalized with respect to each other in the COBRA method means that only relative values, not absolute x-ray beam intensity values, are used. This reduces but does not eliminate the energy dependence of the computed $F(\theta)$ functions.

The COBRA method, as described here, requires the sensitive region of the dosimeter to be fully contained within the width of the CT beam along the z -dimension. Given the increasing beam width of modern CT scanners, this should not be a problem. We are also interested in characterizing the x-ray beam profile along the z -dimension of the scanner and, combined with the $F(\theta)$ measurements determined from the COBRA method, achieving a comprehensive understanding of the beam properties emitted from the x-ray tube assembly in CT systems.

VII. CONCLUSIONS

The COBRA method of characterizing the angle-dependent attenuation properties of a bow tie filter was compared against time-consuming step-and-shoot measurement techniques and excellent agreement was demonstrated. Without revealing proprietary information, the attenuation features of a CT beam-shaping filter are described from a series of 10 s scans performed at four different tube voltages. Additionally, the thickness of a bow tie filter can be estimated to within 10% of the manufacturing specifications. While this technique can noninvasively and rapidly characterize the BT properties of any commercial whole body CT scanner, it relies on a dependable definition of the scanner's distance from source to isocenter (or probe to isocenter) as well as state of the art, real-time dosimetric hardware. In particular, this method requires a linear-response dose probe with high temporal frequency, real-time output, and an isotropic detection volume that fits well within the collimated width of the scanners' x-ray beam. Further experimental work is necessary to extend characterization in the z -direction.

ACKNOWLEDGMENTS

This research was funded from the National Institutes of Health Grant No. R01 EB002138 and also by a grant from the UC Davis Health system. The authors would like to thank Dr. J. Anthony Seibert, Ph.D. (University of California, Davis) for his advice and assistance with the use of the clinical CT scanner.

^{a)}Electronic mail: jmboone@ucdavis.edu

¹M. R. Ay and H. Zaidi, "Development and validation of MCNP4C-based Monte Carlo simulator for fan- and cone-beam x-ray CT," *Phys. Med. Biol.* **50**, 4863–4885 (2005).

²R. G. Walter and R. W. Carlson, U.S. Patent No. 4, 288,695 (1981).

³N. Mail, D. J. Moseley, J. H. Siewerdsen, and D. A. Jaffray, "The influence of bowtie filtration on cone-beam CT image quality," *Med. Phys.* **36**, 22–32 (2009).

⁴S. A. Graham, D. J. Moseley, J. H. Siewerdsen, and D. A. Jaffray, "Compensators for dose and scatter management in cone-beam computed tomography," *Med. Phys.* **34**, 2691–2703 (2007).

⁵J. E. Tkaczyk, Y. F. Du, D. Walter, X. Y. Wu, J. Y. Li, and T. Toth, "Simulation of CT dose and contrast-to-noise as function of bowtie shape," *Proc. SPIE* **5368**, 403–410 (2004).

⁶J. Hsieh, U.S. Patent No. 6,647,095 (2003).

⁷J. J. DeMarco, C. H. Cagnon, D. D. Cody, D. M. Stevens, C. H. McCollough, J. O'Daniel, and M. F. McNitt-Gray, "A Monte Carlo based method to estimate radiation dose from multidetector CT (MDCT): Cylindrical and anthropomorphic phantoms," *Phys. Med. Biol.* **50**, 3989–4004 (2005).

⁸J. Gu, B. Bednarz, P. F. Caracappa, and X. G. Xu, "The development, validation and application of a multi-detector CT (MDCT) scanner model

- for assessing organ doses to the pregnant patient and the fetus using Monte Carlo simulations," *Phys. Med. Biol.* **54**, 2699–2717 (2009).
- ⁹J. M. Boone, "Method for evaluating bow tie filter angle-dependent attenuation in CT: Theory and simulation results," *Med. Phys.* **37**, 40–48 (2009).
- ¹⁰L. A. Lehmann, R. E. Alvarez, A. Macovski, W. R. Brody, N. J. Pelc, S. J. Riederer, and A. L. Hall, "Generalized image combinations in dual kVp digital radiography," *Med. Phys.* **8**, 659–667 (1981).
- ¹¹R. E. Alvarez and A. Macovski, "Energy-selective reconstructions in x-ray computerized tomography," *Phys. Med. Biol.* **21**, 733–744 (1976).
- ¹²J. M. Boone and A. E. Chavez, "Comparison of x-ray cross sections for diagnostic and therapeutic medical physics," *Med. Phys.* **23**, 1997–2005 (1996).
- ¹³J. Boone, T. Nelson, A. Kwan, and K. Yang, "Computed tomography of the breast: Design, fabrication, characterization, and initial clinical testing," *Med. Phys.* **33**, 2185 (2006).
- ¹⁴J. M. Boone, N. Shah, and T. R. Nelson, "A comprehensive analysis of DgN(CT) coefficients for pendant-geometry cone-beam breast computed tomography," *Med. Phys.* **31**, 226–235 (2004).
- ¹⁵A. L. C. Kwan, J. M. Boone, and N. Shah, "Evaluation of x-ray scatter properties in a dedicated cone-beam breast CT scanner," *Med. Phys.* **32**, 2967–2975 (2005).
- ¹⁶K. Yang, A. L. C. Kwan, D. F. Miller, and J. M. Boone, "A geometric calibration method for cone beam CT systems," *Med. Phys.* **33**, 1695–1706 (2006).
- ¹⁷J. T. Bushberg, J. A. Seibert, E. M. Leidholdt, and J. M. Boone, *The Essential Physics of Medical Imaging*, 2nd ed. (Lippincott Williams and Wilkins, Philadelphia, 2002).
- ¹⁸J. M. Boone and J. A. Seibert, "Accurate method for computer-generating tungsten anode x-ray spectra from 30 to 140 kV," *Med. Phys.* **24**, 1661–1670 (1997).
- ¹⁹A. F. Maia and L. V. E. Caldas, "A simple method for evaluation of half-value layer variation in CT equipment," *Phys. Med. Biol.* **51**, 1595–1601 (2006).
- ²⁰R. L. Kruger, C. H. McCollough, and F. E. Zink, "Measurement of half-value layer in x-ray CT: A comparison of two noninvasive techniques," *Med. Phys.* **27**, 1915–1919 (2000).