Technical Note: Measurement of bow tie profiles in CT scanners using radiochromic film

Bruce R. Whiting^{a)}

Department of Radiology, University of Pittsburgh, Pittsburgh, Pennsylvania 15213

Andreea C. Dohatcu

University of Pittsburgh Medical Center, Pittsburgh, Pennsylvania 15213

Joshua D. Evans

Department of Radiation Oncology, Virginia Commonwealth University, Richmond, Virginia 23298

David G. Politte

Mallinckrodt Institute of Radiology, Washington University, St. Louis, Missouri 63110

Jeffrey F. Williamson

Department of Radiation Oncology, Virginia Commonwealth University, Richmond, Virginia 23298

(Received 17 September 2014; revised 27 April 2015; accepted for publication 29 April 2015; published 18 May 2015)

Purpose: To provide a noninvasive technique to measure the intensity profile of the fan beam in a computed tomography (CT) scanner that is cost effective and easily implemented without the need to access proprietary scanner information or service modes.

Methods: The fabrication of an inexpensive aperture is described, which is used to expose radiochromic film in a rotating CT gantry. A series of exposures is made, each of which is digitized on a personal computer document scanner, and the resulting data set is analyzed to produce a self-consistent calibration of relative radiation exposure. The bow tie profiles were analyzed to determine the precision of the process and were compared to two other measurement techniques, direct measurements from CT gantry detectors and a dynamic dosimeter.

Results: The radiochromic film method presented here can measure radiation exposures with a precision of ~6% root-mean-square relative error. The intensity profiles have a maximum 25% root-mean-square relative error compared with existing techniques.

Conclusions: The proposed radiochromic film method for measuring bow tie profiles is an inexpensive (~\$100 USD + film costs), noninvasive method to measure the fan beam intensity profile in CT scanners. © 2015 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4921120>]

Key words: bow tie filter, CT, radiochromic film

1. INTRODUCTION

Since its invention four decades ago, computed tomography (CT) imaging has become a valuable diagnostic tool.¹ Technically, its performance continues to advance at a rapid pace, in large part due to the application of system modeling for optimization of components and operation. Academic research has been hindered in this area because details of CT systems are often considered proprietary by CT manufacturers.² One example of this is the variation of x-ray intensity across the fan beam exposing an object, which is modified by a bow tie filter that is inserted into the beam to equalize flux reaching the detectors, reduce background scattered radiation and minimize patient dose.³ Knowledge of the profile, or specifically the relative intensity across the fan beam, is necessary in several areas of research: (1) dose-reduction simulators, which add synthetic noise to create an image corresponding to a dose lower than that of the original image,^{4,5} (2) calculation of radiation dose to an individual patient using Monte Carlo techniques, which requires knowledge of incident flux,⁶ and (3) iterative statistical

reconstruction algorithms, which incorporate knowledge of the incident flux in noise models to estimate images with improved noise-resolution tradeoffs.⁷

Such fan beam intensity information could be obtained from mechanical drawings of the bow tie and gantry system containing component materials and dimensions, but this information is often considered proprietary by manufacturers. One could remove the gantry shell and physically measure the components and use an estimate of the central axis x-ray spectrum to compute the relative energy intensity across all detectors, but this is not likely to be appreciated by hospital management. With sufficient access, one could enter the scanner's "service mode" and directly measure the fan beam intensity in a known parked condition or measure intensity during a topogram/scout scan,^{8,9} although some bow tie filters may not be available in the topogram mode. In a different approach, analysis of raw data (sinogram) scans reveal noise (and hence intensity) profiles.¹⁰ While access to such files and data formats may be provided to select researchers, they are not generally available, and the level of access and the data format can vary between CT manufacturers.

FIG. 1. (a) Gantry trajectory-source S rotates about isocenter O at distance R ; at gantry angle Ω , the normal distance from S to the plane X_L - X_R is L , forming fan beam angle ϕ . (b) Apertures allow rays to strike the detectors within an interval of width Δ only within the acceptance angle, e.g., A_L - O - A_R .

To overcome these limitations, the Boone group proposed a method using a dynamic dosimeter, positioned away from the isocenter, to measure the flux as the gantry rotated, from which a profile could be obtained.^{11,12} His experiments used a custom-designed prototype dosimeter that was not publically available. In a related paper,¹³ we reported using Boone's technique with a newly available commercial dosimeter. However, the cost of these dosimeters is not trivial (~\$7 K USD for a digitizer and \$1.5 K for a dynamic air chamber), so there remains a need for a noninvasive, inexpensive, and convenient experimental method for bow tie profile measurements. In this paper, we present such a method, using radiochromic film^{14,15} and commonly available hardware, which is much more economical (~\$100 USD),¹⁶ and we compare the results of using it to the two measurement approaches described above. The methods presented herein make the measurement of CT bow tie profiles accessible to a wider range of researchers.

2. MATERIALS AND METHODS

We present a method for measuring the intensity profile of a CT fan beam using inexpensive hardware and radiochromic film. Details of the implementation are described, and comparisons with two other methods, measurements made by the CT scanner detectors themselves and measurements with a dynamic dosimeter, are given.

2.A. CT scanner

Measurements were performed on a Philips Brilliance 16-detector-row Big Bore CT scanner (Philips Medical Systems, Cleveland, OH) located in Virginia Commonwealth University's (VCU) radiation oncology clinic. For experimental convenience, the film/apparatus was placed on the patient table, necessitating axial (step-and-shoot) protocols to keep the film stationary. Film exposure levels were made by controlling the number of axial rotations, e.g., a single-exposure film would have 15 rotations and a double-exposure film would receive 30 rotations. Prior experiments using other profile measurement methods were performed on the same machine, including real-time dosimeter probes and internal measurements of x-ray intensity incident on the gantry detectors using vendor-supplied software, to provide independent measures of the fan beam profile.¹³

2.B. Fan beam intensity measurement

2.B.1. Aperture apparatus

The sampling scheme is based on the scanner configuration shown in Fig. 1(a), which was modeled as a point source without background scatter. As the source S rotates through angle Ω about the isocenter, a ray (S - x) normal to a line (X_L - O - X_R) through the isocenter samples the fan beam angle $\phi = \arcsin(x/R)$, where x is the distance from the isocenter to the sampling point and R is the source-to-isocenter distance, resulting in an intensity at point x . To isolate the contribution of this ray from the continuous, confounding exposures during the gantry rotation, a metal aperture is physically introduced at the isocenter plane that accepts rays only over a narrow angular range about the normal and attenuates other oblique rays [Fig. 1(b)]. To compute an intensity profile relative to the central ray, the measured intensities along the detection plane must be corrected for their differing distance from the source (inverse-square law) and for the differing dwell time that the source appears over the aperture. The distance from the source-to-isocenter plane (L) is given by

$$L = R \cdot \cos(\phi). \quad (1)$$

The projection of S along the X axis is $x = R \cdot \cos(\Omega)$. Given that the gantry rotates at constant angular velocity $\omega = d\Omega/dt$, the velocity v_x of S along the X axis is

$$v_x = -R \cdot \omega \cdot \sin(\omega t) = -R \cdot \omega \cdot \sin(\Omega) = -R \cdot \omega \cdot \cos(\phi), \quad (2)$$

and the dwell time of the source above an aperture of width Δ is Δ/v_x . The measured intensity $M(x)$ is thus proportional to the fan beam intensity $I(\phi)$ times the dwell time and inverse-square correction,

$$M(x) \propto I(\phi) \cdot \left(\frac{\Delta}{v_x} \right) R^2 / L^2. \quad (3)$$

Solving for the profile relative to the intensity of the central ray gives

$$\frac{I(\phi)}{I(0)} = \frac{M(x) \cdot (R \cdot \cos(\phi))^2 \cdot R \cdot \omega \cdot \cos(\phi) R^2}{M(0) \cdot (R \cdot \cos(0))^2 \cdot R \cdot \omega \cdot \cos(0) R^2} = \frac{M(x)}{M(0)} \cos^3(\phi). \quad (4)$$

To fabricate an aperture, perforated steel tubing with a square cross section was used [Fig. 2(a)]. The tubing was made with 14 gauge steel (wall thickness ~2 mm), having 5/16 in. diameter holes spaced on 1 in. center-to-center intervals, with

FIG. 2. (a) View of apparatus formed by perforated steel tubes, which have a square cross section and perforated holes on all four sides, with end caps formed by bolts and washers to block side exposure. Bottom, exposed film strips attached to steel bar base, acting as attenuator to prevent backside exposure. For measurements, the tubes were placed on top of the film base and then positioned in the CT gantry. (b) Cross sectional view of apparatus in place over film. Rays passing through a hole are unattenuated (B), while other normal rays (A) pass through 8 layers of steel. There is no direct line of sight from the base of the hole to adjacent holes, although some rays may pass through fewer layers, e.g., C passes through four layers. Likewise, areas between holes may have rays that are attenuated by fewer layers, e.g., D passes through four layers.

a total length of 45 cm and a total of 16 aperture openings. To increase absorption, two sizes (1, 1.25 in.) of square steel tubes were telescoped inside each other. To narrow the acceptance angle, two of these two-in-one tubes were placed on top of each other, with 4 mm separation spacing, aligned such that there was a vertical clear channel but there was no direct line-of-sight opening between holes on the bottom and adjacent holes on the top of the aperture [Fig. 2(b)]. This gave an aperture acceptance angle of approximately 6.8° and a fan beam angular range of $\pm 0.4^\circ$. As shown in Fig. 2(b), non-normal rays have varying amounts of attenuation during gantry rotation but the dominant exposure is from the normal rays that pass through the holes unattenuated.

To ensure the film is exposed only when the source is above the aperture apparatus, the film and aperture were placed on a 9 mm thick steel bar, and a lead apron of 0.5 mm Pb equivalent thickness was put underneath them on the CT couch. On the end of the tube openings, steel washers of total thickness (diameter) 20 mm were held in place with bolts to attenuate any radiation that might enter from the side as the source rotates down toward 90° .

The aperture apparatus was visually aligned in the CT scanner's plane of rotation using the scanner's patient-positioning lasers, such that the film intersected the isocenter between holes six and seven of the 16 total holes, giving the appearance of an asymmetric profile in Figs. 4 and 5. This provided double measurements at six points on the profile, plus three additional points at larger fan beam angles. A longer aperture apparatus could be fabricated to give double measurements at all points across the profile.

2.B.2. Radiochromic film measurements

To measure the x-ray intensity filling the aperture, radiochromic film was exposed and digitized. Gafchromic XR-QA2 (International Specialty Products, Wayne, NJ) is a self-developing, dye-based active layer on a reflective base. Film strips 16×203 mm were cut and taped under the aperture. CT protocols delivering approximately 20 cGy/revolution (e.g., 120 kVp, 300 mA, 1 s rotation) would produce a

measurable optical density with ~ 10 rotations. When acquiring a complete set of measurements spanning multiple tube potentials and bow tie configurations, x-ray tube heating could become a limitation.

The films were digitized at least 24 h after exposure. A PC document scanner (Laserjet 6500, Hewlett Packard, Palo Alto, CA) scanned the films at 300 dpi spatial resolution with three 8-bit color channels, and the images were saved in JPEG files. The scanner measures the brightness (reflected light) of a film, as verified by digitizing calibrated density step wedges. The decrease in reflectance for the measurements utilized here was typically less than 50%. The XR-QA2 film design includes three separate color dyes, each of which responds to a different intensity range. The red (most sensitive) channel was used for analysis.

The response of Gafchromic film has been studied extensively,¹⁷ with several models proposed to characterize the exposure E as a function of scanned film brightness R . The “double logarithmic” response function, $E(R) = \log(1 + ((\log(R/R_b))/D_s))$, where R_b is the base film reflectance and D_s is the saturated film density, is based on a fundamental physics model and was found to give the best results in this study [Fig. 3(a)]. A data analysis technique was developed to extract these two model parameters from the measurements.

2.B.3. Data analysis

Data from the digitized film images were acquired with ImageJ software (Rasband, W.S., ImageJ, U.S. National Institutes of Health, Bethesda, MD, <http://imagej.nih.gov/ij/>, 1997–2014). Using a circular region of interest with a diameter small enough to avoid penumbra effects (but at least 50 pixels in diameter), the reflectance under the aperture holes, as well as baseline reflectance between aperture openings, was measured. Note that higher kVp measurements can introduce a background baseline due to increased penetration of the apparatus [peaks observable in Fig. 3(b)]. This baseline is estimated for each kVp and subtracted from the hole-exposure measurements during the computation of the bow tie profile.

FIG. 3. (a) Brightness-dose response for the double logarithmic model for a typical scanned film. The range of digitized brightness values in the experiments reported here was 140–220. (b) Scanned profile of film response at 140 kVp; the varying peak heights show the presence of a background baseline.

In order to estimate the fan beam intensity profile, the measured film reflectance must be converted into relative radiation exposure units. In normal usage, to measure absolute radiation dose for treatment planning or equipment characterization, radiochromic film undergoes an extensive calibration procedure, with reference dosimeters, explicit multiple controlled exposures, etc. In our case, we are interested in the relative energy fluence profile of the fan beam and needed a convenient technique to quantify relative intensity without requiring separate absolute calibration measurements. To do this, multiple measurements at different exposures were made with a known relationship, e.g., two exposures differing by a factor of 2. In this case, each aperture has 16 holes and 17 attenuated regions between holes, so there are 33 pairs of film measurements spanning

the range of interest, each pair having a single- and double-exposure level [Fig. 4(a)]. These data are used to numerically solve (Nelder–Mead Simplex method, *fminsearch*, MATLAB Optimization Toolbox, Mathworks, Natick, MA) for the parameters of the film model by minimizing the quadratic relative error between the double-exposure measurement and twice the corresponding single-exposure measurement, according to

$$\text{Cost} = \sum_{i=1}^N \left(\frac{E(R_{i,2x}) - 2 \cdot E(R_{i,1x})}{E(R_{i,2x})} \right)^2, \quad (5)$$

where $E(R)$ represents the exposure calculated by the response model and N is the number of measurement pairs. The results of using the model to transform to exposure space are shown in

FIG. 4. (a) Digitized film values for single and double exposures at 120 kVp, with the upper curves being the background baseline measurements and the lower curves being the hole exposures. Corresponding measurement pairs are used to determine parameters of the film response model. (b) Conversion of film density to exposure, comparing double-exposure measurements to twice the single-exposure measurement.

FIG. 5. Plots of bow tie profiles measured with the presented film method for energies 90, 120, and 140 kV, left to right, respectively. Error bars represent the estimated uncertainty.

Fig. 4(b). This approach was found to be robust and is similar to a method used to calibrate holographic film response.¹⁸

In analyzing the data, two questions were addressed: (1) how precisely does the film response model measure the bow tie profile? and (2) how well does the radiochromic film method compare with other measurement methods that compute bow tie profiles? For the first question, the standard deviation of the difference for all pairs of exposures at twice the single and the double exposure ($2 \cdot E_{1x} - E_{2x}$) was computed, indicating the precision of the exposure calibration method. Also, the standard deviation of the relative error ($2((2 \cdot E_{1x} - E_{2x}) / (2 \cdot E_{1x} + E_{2x}))$) for hole measurements was computed. (The background floor measurements were excluded, since they have mean values ~ 0 and the relative error would be dominated by the terms in the denominator.)

To characterize the quality of the bow tie profiles, use was made of the fact that the experimental technique produces several sets of redundant data (single and double exposures, plus two holes that are symmetrically positioned about the isocenter) that can be used to estimate precision. For the six central profile positions, the standard deviation of the four equivalent measurements (two symmetric holes, each with standard and double exposures) was computed, as well as the relative error. Likewise, for the three peripheral profile positions, the standard deviation of single- and double-exposure measurements was calculated.

For the second question, a comparison of the film method to other intensity profile measurement methods (either measurements directly from the CT scanner detectors¹⁹ or using a high frequency response dosimeter¹³) was performed, using data acquired in previous studies. For image reconstruction tasks, such as dose-reduction simulators or statistical reconstruction algorithms, the intensity profile measured by the CT scanner can be considered as the “gold standard” because it is involved with image creation. For other tasks, such as material decomposition of the bow tie filter, the gold standard would necessitate knowledge of scanner components and is beyond the scope of this paper. The root-mean-square relative error between profiles was calculated to measure similarity. Also, a chi-square statistic²⁰ (χ^2) for the respective measurements was calculated and tested against the hypoth-

esis that the two techniques are equivalent using a complementary incomplete gamma function, with a significance level of $\alpha > 0.05$. (The chi-square computation requires noise estimates for each measurement method, which were computed in previous studies.¹³)

3. RESULTS

3.A. Film method precision

The bow tie profiles measured with the film method described here are shown in Fig. 5 and an analysis is presented in Table I. The precision of the film calibration process is seen to be 2.5%–5%, while relative errors in the hole measurements are 5%–6%. The profile errors are on the order of 1%–2% (because raw data measures are combined, reducing the magnitude of the errors), while the relative errors are on the order of 2%–4%.

3.B. Comparison with other bow tie measurement methods

Profiles for the three methods are plotted in Fig. 6 and Table I reports statistics of agreement between the three. Visually, all profiles are close, with some divergence at higher fan beam angles. The relative error between film and the other two methods varies between 4% and 11%, with maximum differences being 15%–25%. The corresponding differences between the scanner and dosimeter methods were 3%–5%, with maximums of 8%–16%. Applying a chi-square

TABLE I. All data are for relative measurements, nominally in the range 0–1. Calibration process results (a) and (b) indicate precision of acquisition method, while (c) and (d) show precision of computed profiles.

kVp	90	120	140
Film method precision			
(a) Precision of calibration process (%)	2.6	3.1	4.7
(b) Relative error for hole calibration (%)	6.3	5.0	5.9
(c) Error in profile estimate (%)	1.6	1.3	2.3
(d) Relative error in profile estimate (%)	2.4	2.9	3.6

Fig. 6. Top plots are bow tie profiles at different tube potentials (left to right, 90-120-140 kV) for three measurement methods [radiochromic film (●), CT scanner (—), real-time dosimeter (---)]. Bottom plots are corresponding relative differences for film-CT scanner (✕) and film-dosimeter (✕).

test for the relative differences between methods indicated that all were statistically significant. It should be noted that the precision of the bow tie profile estimates made by the other methods are substantially better than the film approach, with the scanner having noise about two orders of magnitude less (presumably due to high DQE solid state detectors and several thousand samples at each of hundreds of fan beam angles) and the dosimeter acquiring tens of thousands of samples. Therefore, the noise of the film method dominates the stochastic part of the comparison statistics. Because the comparison was made with normalized bow tie profiles (scaled to 1.0 at the isocenter), the largest discrepancies occurred toward the periphery of the fan beam (Fig. 6).

4. DISCUSSION

The results indicate that the radiochromic film method of determining the fan beam intensity profile has a precision of approximately 5% and is reasonably consistent with other methods of bow tie profile measurement (root-mean-square relative errors ranged between 4% and 11%, with a ~25% maximum relative error).

A limitation of these three measurement methods is that they acquire a relative intensity profile and normalization of the central axis ray removes a degree of freedom, such that differences between methods will only be apparent toward the periphery of the fan beam. Also in many cases, the intensity at the bow tie center needs to be anchored to an absolute property (exposure magnitude, spectrum at the isocenter, etc.) The differences between methods may be partly attributable to the difference in energy response of the detectors, i.e., film vs solid state detectors vs air chambers, which would need to be accounted for when applying the profiles for particular tasks.

If one has access to information from CT manufacturers or can operate a scanner in service mode, less effort is required to characterize the fan beam profile. For example,

most modern scanners perform scout/topogram scans with a stationary gantry, during which a direct measurement could be made. However, in some scanners, certain filters may not be used for scout/topogram scans, so a method for dynamic acquisition, such as the one introduced here, would be required. DICOM tag (0018, 1160) records which filter was used in all scans, so the presence of the bow tie filter in the CT and topogram scans can be confirmed.

An important question to be addressed is how accurately must the bow tie profile be determined? This depends on what use the profile is to be put and is an open area of research for several potential applications. For dose-reduction simulations that use the intensity profile to add synthetic noise to projection data, a just-noticeable difference in image noise level for human viewers is typically on the order of 10%–15% in noise magnitude,²¹ so adding sinogram variances in quadrature with a 5%–25% error would be barely noticeable. Using the profile information for material decomposition of the bow tie components involves operations that may amplify errors.²² In previous studies,²³ we found such amplification to be on the order of a factor of 2, which would make the uncertainties of this film method larger than the 10% reported when using the dynamic dosimeter.¹² Determining how the uncertainty of the fan beam profile propagates in patient dose calculation by Monte Carlo methods^{11,24–26} is an active area of research, as is the influence of bow tie profile uncertainty in statistical iterative reconstruction algorithms.¹⁹ If more accurate bow tie profiles are required, improving radiochromic precision by repeated measurements could be required, as well as energy-response calibration.

5. CONCLUSIONS

An inexpensive, noninvasive, and fast technique to measure CT fan beam intensity profiles for research purposes is presented, which would be useful when alternative measurement

methods are not available. The method described here uses radiochromic film, readily available hardware, and personal computer peripherals, making it economical, and utilizes a simple self-calibrating film procedure. The proposed method was compared to methods based on real time dosimeters or service-level access to a clinical CT system. Relative differences in profiles between the systems were statistically significant, with maximum relative errors on the order of 25%. Applications such as dose-reduction simulations would be able to tolerate this level of uncertainty; however other quantitative applications may require additional detector response corrections.

ACKNOWLEDGMENT

This work was supported in part from grants (No. R21 EB013384, B. Whiting, PI; No. R01 CA149305, J. Williamson, PI) awarded by the National Institutes of Health.

^{a1}Author to whom correspondence should be addressed. Electronic mail: whitingbrucer@gmail.com; Telephone: 314-458-6383; Fax: 412-641-2582.

¹W. A. Kalender, "Computed tomography," in *Fundamentals, System Technology, Image Quality, Applications*, 3rd ed. (Publicis Corporate Publishing, Erlangen, 2011).

²C. H. McCollough, G. H. Chen, W. Kalender, S. Leng, E. Samei, K. Taguchi, G. Wang, L. Yu, and R. I. Pettigrew, "Achieving routine submillisievert CT scanning: Report from the summit on management of radiation dose in CT," *Radiology* **264**, 567–580 (2012).

³J. Hsieh, *Computed Tomography* (SPIE, Bellingham, WA, 2003).

⁴P. Massoumzadeh, S. Don, C. F. Hildebolt, K. T. Bae, and B. R. Whiting, "Validation of CT dose-reduction simulation," *Med. Phys.* **36**, 174–189 (2009).

⁵S. Zabic, Q. Wang, T. Morton, and K. M. Brown, "A low dose simulation tool for CT systems with energy integrating detectors," *Med. Phys.* **40**, 031102 (13pp.) (2013).

⁶J. M. Boone, "Dose spread functions in computed tomography: A Monte Carlo study," *Med. Phys.* **36**, 4547–4554 (2009).

⁷J. F. Williamson, B. R. Whiting, J. Benac, R. J. Murphy, G. J. Blaine, J. A. O'Sullivan, D. G. Polite, and D. L. Snyder, "Prospects for quantitative computed tomography imaging in the presence of foreign metal bodies using statistical image reconstruction," *Med. Phys.* **29**, 2404–2418 (2002).

⁸A. C. Turner, D. Zhang, M. Khatonabadi, M. Zankl, J. J. DeMarco, C. H. Cagnon, D. D. Cody, D. M. Stevens, C. H. McCollough, and M. F. McNitt-Gray, "The feasibility of patient size-corrected, scanner-independent organ dose estimates for abdominal CT exams," *Med. Phys.* **38**, 820–829 (2011).

⁹W. Skrzynski, "Measurement-based model of a wide-bore CT scanner for Monte Carlo dosimetric calculations with GMCTdospp software," *Phys. Med.* **30**, 816–821 (2014).

¹⁰B. R. Whiting, P. Massoumzadeh, O. A. Earl, J. A. O'Sullivan, D. L. Snyder, and J. F. Williamson, "Properties of preprocessed sinogram data in x-ray computed tomography," *Med. Phys.* **33**, 3290–3303 (2006).

¹¹J. M. Boone, "Method for evaluating bow tie filter angle-dependent attenuation in CT: Theory and simulation results," *Med. Phys.* **37**, 40–48 (2010).

¹²S. E. McKenney, A. Nosrati, D. Gelskey, K. Yang, S. Y. Huang, L. Chen, and J. M. Boone, "Experimental validation of a method characterizing bow tie filters in CT scanners using a real-time dose probe," *Med. Phys.* **38**, 1406–1415 (2011).

¹³B. R. Whiting, J. D. Evans, A. C. Dohatcu, J. F. Williamson, and D. G. Polite, "Measurement of bow tie profiles in CT scanners using a real-time dosimeter," *Med. Phys.* **41**, 101915 (7pp.) (2014).

¹⁴S. Devic, N. Tomic, S. Aldelaijan, F. DeBlois, J. Seuntjens, M. F. Chan, and D. Lewis, "Linearization of dose-response curve of the radiochromic film dosimetry system," *Med. Phys.* **39**, 4850–4857 (2012).

¹⁵D. Lewis, A. Micke, X. Yu, and M. F. Chan, "An efficient protocol for radiochromic film dosimetry combining calibration and measurement in a single scan," *Med. Phys.* **39**, 6339–6350 (2012).

¹⁶B. R. Whiting and A. C. Dohatcu, "Method for measuring the intensity profile of a CT fan-beam filter," *Proc. SPIE* **9033**, 9033–9072 (2014).

¹⁷N. Tomic, C. Quintero, B. R. Whiting, S. Aldelaijan, H. Bekerat, L. Liang, F. DeBlois, J. Seuntjens, and S. Devic, "Characterization of calibration curves and energy dependence GafChromic XR-QA2 model based radiochromic film dosimetry system," *Med. Phys.* **41**, 062105 (9pp.) (2014).

¹⁸H. Kasprzak, N. Sultanova, and H. Podbielska, "Nonlinear effects of the recording material on the image quality of a Fourier hologram," *J. Opt. Soc. Am. A* **4**, 843–846 (1987).

¹⁹J. D. Evans, B. R. Whiting, D. G. Polite, J. A. O'Sullivan, P. F. Klahr, and J. F. Williamson, "Experimental implementation of a polyenergetic statistical reconstruction algorithm for a commercial fan-beam CT scanner," *Phys. Med.* **29**, 500–512 (2013).

²⁰W. H. Press, S. A. Teukolsky, W. T. Vetterling, and B. P. Flannery, *Numerical Recipes in C*, 2nd ed. (Cambridge University Press, NY, 1994).

²¹D. Zwick and D. L. Brothers, "RMS granularity: Determination of just-noticeable differences," *Photogr. Sci. Eng.* **19**, 235–238 (1975).

²²L. 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).

²³J. F. Williamson, S. Li, S. Devic, B. R. Whiting, and F. A. Lerma, "On two-parameter models of photon cross sections: Application to dual-energy CT imaging," *Med. Phys.* **33**, 4115–4129 (2006).

²⁴X. Li, E. Samei, W. P. Segars, G. M. Sturgeon, J. G. Colsher, G. Toncheva, T. T. Yoshizumi, and D. P. Frush, "Patient-specific radiation dose and cancer risk estimation in CT: Part II. Application to patients," *Med. Phys.* **38**, 408–419 (2011).

²⁵A. C. Turner, D. Zhang, H. J. Kim, J. J. DeMarco, C. H. Cagnon, E. Angel, D. D. Cody, D. M. Stevens, A. N. Primak, C. H. McCollough, and M. F. McNitt-Gray, "A method to generate equivalent energy spectra and filtration models based on measurement for multidetector CT Monte Carlo dosimetry simulations," *Med. Phys.* **36**, 2154–2164 (2009).

²⁶D. J. Long, C. Lee, C. Tien, R. Fisher, M. R. Hoerner, D. Hintenlang, and W. E. Bolch, "Monte Carlo simulations of adult and pediatric computed tomography exams: Validation studies of organ doses with physical phantoms," *Med. Phys.* **40**, 013901 (10pp.) (2013).