Dose spread functions in computed tomography: A Monte Carlo study

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Purpose: Current CT dosimetry employing CTDI methodology has come under fire in recent years, partially in response to the increasing width of collimated x-ray fields in modern CT scanners. This study was conducted to provide a better understanding of the radiation dose distributions in CT.

Methods: Monte Carlo simulations were used to evaluate radiation dose distributions along the z axis arising from CT imaging in cylindrical phantoms. Mathematical cylinders were simulated with compositions of water, polymethyl methacrylate (PMMA), and polyethylene. Cylinder diameters from 10 to 50 cm were studied. X-ray spectra typical of several CT manufacturers (80, 100, 120, and 140 kVp) were used. In addition to no bow tie filter, the head and body bow tie filters from modern General Electric and Siemens CT scanners were evaluated. Each cylinder was divided into three concentric regions of equal volume such that the energy deposited is proportional to dose for each region. Two additional dose assessment regions, central and edge locations 10 mm in diameter, were included for comparisons to CTDI₁₀₀ measurements. Dose spread functions (DSFs) were computed for a wide number of imaging parameters.

Results: DSFs generally exhibit a biexponential falloff from the $z=0$ position. For a very narrow primary beam input ($\ll 1$ mm), DSFs demonstrated significant low amplitude long range scatter dose tails. For body imaging conditions (30 cm diameter in water), the DSF at the center showed ~ 160 mm at full width at tenth maximum (FWTM), while at the edge the FWTM was ~ 80 mm. Polyethylene phantoms exhibited wider DSFs than PMMA or water, as did higher tube voltages in any material. The FWTM were 80, 180, and 250 mm for 10, 30, and 50 cm phantom diameters, respectively, at the center in water at 120 kVp with a typical body bow tie filter. Scatter to primary dose ratios (SPRs) increased with phantom diameter from 4 at the center (1 cm diameter) for a 16 cm diameter cylinder to ~ 12.5 for a 32 cm diameter cylinder. The SPRs increased dramatically at the center of the phantom compared to the edge. For the three equal area regions, the edge to center SPRs for a 32 cm diameter phantom were ~ 1.8 , 3.5, and 6.3, respectively.

Conclusions: DSFs demonstrate low amplitude long ranging tails which reach considerable distances in cylindrical phantoms. The buildup that results from these long-ranged tails increases at the center of the field (at $z=0$) with increasing scan length. The DSF distributions lend a better understanding of the trends in CT dose deposition over a range of relevant imaging parameters. The DSFs as well as other related data are available to interested parties using EPAPS at <http://www.aip.org/pubservs/epaps.html>. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3223634](https://doi.org/10.1118/1.3223634)]

I. INTRODUCTION

The advent of multidetector computed tomography (CT) scanners, which employ relatively wide collimated x-ray beams during scanning, has created the need to develop new dose measurement procedures in CT. One of the principal concerns associated with these new scanners is that the long tails of scattered radiation extending along the z axis are not accurately measured using current procedures, giving rise to the underestimation of the radiation dose on modern CT scanners. There has been substantial discussion in the literature in regards to inaccuracies associated with the widely used CTDI₁₀₀ measurement techniques.¹⁻⁶

As the medical physics community involved with computed tomography develops new standards for the evaluation of dose in CT, there is much discussion in regards to how large phantoms need to be in order to accurately assess the long tails of scattered radiation in the azimuthal (z) direction.

In addition, there has been increased focus on CT dose reduction in the pediatric patient in recent years. This has become even more of an issue in recent years because as CT scanners have developed faster acquisition capabilities in which patient motion can be better managed, their use in scanning pediatric patients has increased dramatically.⁷

In order to better understand the dose distribution arising from scattered radiation in CT, Monte Carlo simulation techniques were employed for this study. While the widely utilized concept of the line spread function (LSF) defines an imaging system's response to a slit (or line-shaped) input function, dose spread functions (DSFs) are utilized in this work which describe the scattered radiation dose as a function of position given a narrow fan beam incident upon a cylindrical phantom. As with LSFs, DSFs can be used with convolution-based mathematics to derive the scattered radiation dose distribution for CT scan geometry of arbitrary length involving approximately cylindrical patients.

FIG. 1. (a) The geometry of Monte Carlo simulations is illustrated, showing the x-ray source with bow tie filter rotating around the cylindrical phantom. Five regions are indicated for dose assessment in the phantom, with the center and edge locations being the traditional location for pencil chamber measurements and regions R1, R2, and R3 are equal area regions (a central circle with two outer annuli). (b) The geometry in the z -axis dimension is illustrated, and the dose spread functions were computed along the z axis for each of the five regions center, edge, R1, R2, and R3.

II. METHODS AND MATERIALS

II.A. Monte Carlo simulations

The previously validated SIERRA Monte Carlo code system⁸⁻¹¹ was modified to accommodate the measurement geometry required in this investigation. Figure 1 illustrates the simulated acquisition geometry utilized. For each Monte Carlo assessment, the x-ray source rotated 2π around the cylinder with no table translation (i.e., axial scanning conditions). Each cylinder that was evaluated extended infinitely in the z direction. Cylinders composed of water, polymethyl methacrylate (PMMA), and polyethylene, each with diameters of 10, 15, 20, 25, 30, 35, 40, 45, and 50 cm, were studied. A divergent x-ray beam emanating from a point x-ray source was simulated, and the beam width (along z) at the isocenter was 0.01 mm. Bow tie filters for a General Electric Lightspeed CT scanner (Waukesha, WI) and a Siemens' Definition CT scanner (Florsheim, Germany) were simulated, and the characteristics of these filters were provided to the author under nondisclosure agreements. The source to isocenter distance was set at 54.1 cm for GE simulations and 59.5 cm for Siemens simulations since the bow tie filters are designed for these specific source to isocenter distances. Both head and body bow tie filters were simulated, and simulations were performed in the absence of a bow tie filter as well.

Each cylinder was divided into three concentric regions (see Fig. 1), with region 1 (R1) at the center, R2 is intermediate, and R3 is the outer annulus. Region diameters were selected such that each region (the two outer annuli and the center circular region) had equal two dimensional cross sectional area, and hence had equal three dimensional volumes as well (considering slice thickness). Since the cylinders were homogeneous in composition and density, each region had the same mass, thus the absorbed energy to each region

was linearly proportional to the average dose (energy/mass) to each region. For a cylinder of diameter D , the outer radii of regions R1, R2, and R3 were computed as $D/2 \sqrt{1/3}$, $D/2 \sqrt{2/3}$, and $D/2$, respectively. In addition to the R1–R3 regions, superimposed on each phantom were a center and edge hole, each 10 mm in diameter. The edge hole was centered 10 mm from the edge of each cylinder. Energy deposition was tallied in these holes, which reflect existing CT dosimetry phantoms for comparison with the average dose in the R1–R3 regions. Each cylinder, with the five corresponding subregions, was partitioned in the z -axis dimension into 1 mm thick sections, thereby allowing the dose spread functions (along z) to be measured with 1 mm spatial sampling. The computer simulations used phantoms of infinite length along z , and this geometry differs from the reality of physically measuring dose indices using the widely available 140 mm thick PMMA dose phantoms. The DSFs determined in this investigation should compare well to the physical measurements made on three contiguous 140 mm thick PMMA phantoms, which would span an effective z extent of 420 mm.

Monoenergetic x-ray photon beams were simulated, with x-ray energies ranging from 10 to 140 keV, by 1 keV intervals. A total of 1×10^6 x-ray photons was used for each monoenergetic simulation. The resulting monoenergetic data sets were spectrally weighted to the 80, 100, 120, and 140 kVp x-ray spectra used by several CT scanners. A previously published spectral model¹² was used to generate the x-ray spectra. Additional aluminum was added to the spectra, resulting in half value layers (in aluminum) of 5.5, 6.8, 7.9, and 9.0 mm for the 80, 100, 120, and 140 kVp spectra, respectively. These spectral characteristics are in reasonable agreement with those published by other authors for CT systems.¹³ For the monoenergetic simulations, the use of 10^6 photons yielded precision of 1.5%, 1.1%, 22%, and 14% for regions R1, R3, and the center and edge holes, respectively (calculated in a 32 cm diameter PMMA cylinder at 100 keV). The spectral weighting procedure averages over 71 different energies (at 80 kVp) and 131 different energies (at 140 kVp), resulted in polyenergetic dose spread functions with noise characteristics about an order of magnitude better than the monoenergetic precision.

While the discussion of DSFs focuses on the relative dose distribution along the z axis, the relative dose is a function of both primary and scattered photon energy deposition. For an appreciation of the relative amount of scatter dose that is deposited anywhere in the phantom relative to the primary dose deposited within the collimated primary beam, the dose SPR_{∞} is useful and is defined as

$$SPR_{\infty} = \frac{\int_{-\infty}^{+\infty} S(z) dz}{\int_{-\infty}^{+\infty} P(z) dz}, \quad (1)$$

where one 2π tube rotation with no table movement was assumed (an “axial” scan). The input function $P(z)$ was essentially a delta function along z , $\delta(z)$, and by analogy to the line spread function, DSFs are defined as

$$\text{DSF}(z) = P(z) + S(z). \quad (2)$$

The DSFs are useful because they can be used to compute the overall dose distribution for a set of scan geometries. For example, for a series of contiguous axial CT scans, the dose distribution along z is given by the convolution of the DSF with the primary beam function $\Pi(z)$ representing a primary beam width w ,

$$\text{Dose}(z) = \int_{-\infty}^{+\infty} \text{DSF}(z - z') \Pi(z') dz', \quad (3)$$

where $\Pi(z)$ is a RECT function of width w defined by

$$\Pi(z) = P_0, \quad z = -w/2 \text{ to } +w/2,$$

$$\Pi(z) = 0, \quad \text{elsewhere.}$$

The above formalisms are strictly valid only when the primary beam fluence is homogeneous in intensity within the collimated beam and is zero outside of this. This applies then to N perfectly contiguous CT scans of collimated beam width L , where $w = NL$. In practice, any fine structure in the primary dose distribution along z in the very narrow input function ($10 \mu\text{m}$) is dwarfed by the larger rectangular sampling bins (1 mm) and the much longer ranged scattered radiation tails.

The SIERRA Monte Carlo code⁸ was written in C (Microsoft C/C++ version 8), and about 40 h of run time on a quad core 2.5 GHz Intel-based PC were used to produce the raw data for this study.

III. RESULTS

The data for the center and edge holes at 80, 100, 120, and 140 kVp for the PMMA cylinders for the GE body (32 cm diameter phantom) and GE head (16 cm diameter phantom) were used to compare the accuracy of the dose computations to the published values. The IMPACT (Ref. 14) published values for the GE 16 slice Lightspeed scanner were used for comparison, and these results are illustrated in Fig. 2. The data for the head and body CTDI_{100} at the center and edge (in mGy per 100 mAs) were divided by the CTDI_{100} measurement in air at the isocenter (in mGy per 100 mAs), providing the normalized CTDI_{100} values (i.e., mGy/mGy). The computed data were scaled to the IMPACT data at one point (see arrow in Fig. 2). The data compared well, with excellent correlation (showing good precision). There was no significant difference between the published and the computed data ($p=0.962$, two sided paired t test with normalization point excluded).

Figure 3 illustrates the scatter component of the dose spread functions (sDSFs) for a number of different conditions. Figure 3(a) illustrates the sDSFs for each of the three regions R1, R2, and R3 within the cylinder, for a 30 cm diameter water cylinder scanned using a 120 kVp spectrum with a GE body bowtie filter. Each sDSF was normalized to unity at the center, maximum, value. The central region (R1) demonstrates a broader sDSF than the other two more peripheral regions. Figure 3(b) illustrates the dose spread functions for the center R1 region with three different cylinder

FIG. 2. The Monte Carlo relative dose results were compared to the published dose values for 16 cm (head) and 32 cm (body) diameter PMMA phantoms. The point with the arrow indicates the data point used for normalization. The Monte Carlo results (excluding the normalization point) were not significantly different from the published values. The solid squares were for the body center, open squares for body edge, solid triangles head center, and open triangles for head edge. The four points for each symbol correspond to 80, 100, 120, and 140 kVp, with these points increasing in this order from left to right for all symbols.

compositions, water, PMMA, and polyethylene. These sDSFs demonstrate slightly different slopes on the semilog plots, reflecting the different attenuation properties of each material. Figure 3(c) illustrates the dose spread functions as a function of the cylinder diameter, and it is seen that the 50 cm diameter cylinder has much broader sDSF than the 30 and 10 cm diameter cylinders, as expected. Figure 3(d) illustrates the dose spread functions for the center region R1 for a 40 cm diameter water cylinder, without bow tie and with two different bow tie filters. The sDSFs, normalized at their maximum, show very similar trends between the two vendor bow tie filters but these results vary from the case with no bow tie filter. Figure 3(e) illustrates the dose spread functions as a function of the x-ray spectrum used (tube voltage), and the higher energy spectrum demonstrates a DSF with a more gradual drop off with distance. Finally, Fig. 3(f) illustrates the sDSFs for the three isovolume regions R1, R2, and R3, along with those from the center and edge locations. These data are for a 32 cm diameter PMMA cylinder with a Siemens body bow tie filter. The collection volume of the center and edge regions are much smaller than the other regions, and thus more Poisson noise is seen in the sDSFs. Although the center hole sDSF tracks that of the R1 region, the smaller diameter of the center hole leads to an average higher sDSF level. The edge region tracks the R3 region; however, it is slightly lower, which is consistent in that the edge is more peripheral than the R3 region overall.

The sDSFs for a 32 cm diameter cylinder of PMMA, using a GE body bowtie filter and a 120 kVp spectrum, were convolved with a number of different scan lengths (RECT

Fig. 3. (a) The sDSFs for each of the three regions R1, R2, and R3 are illustrated. These data are normalized to the maximum value at the center of the sDSF to emphasize differences in shape (distribution). Only scatter data are shown. (b) The sDSFs are shown for three phantom compositions, for the center region (R1). These sDSFs were normalized at the maximum central value. It is seen that the sDSF for water and PMMA are very similar; however, that of polyethylene ("poly") is broader due to its lower density and the corresponding longer range of x-ray photon penetration. (c) The sDSFs (normalized at the center maximum value) are shown for three different phantom diameters. The 50 cm diameter phantom demonstrates a much broader dose distribution than the 30 and 10 cm diameter phantom sDSFs. (d) The sDSFs are shown for three different bow tie conditions: (1) No bow tie filter, (2) a GE body bow tie filter, and (3) a Siemens' body bow tie filter. With these values normalized at the center, it is seen that there is very little difference in sDSF for the two commercial bow tie filter, but both of these reduce the lateral (z dimension) spread of the sDSF compared to not using a bow tie filter. (e) The sDSFs are shown as a function of x-ray beam spectrum (80 and 140 kVp), and the higher energy spectrum demonstrates a slightly broader distribution as expected. (f) The sDSFs are shown for all five regions assessed in this study, the center and edge as well as regions R1, R2, and R3. For this 32 cm diameter phantom, the R1, R2, and R3 (which are equal volume) regions have 341 times the collection volume as compared to the center and edge locations. This is why the quantum noise is so apparent at the center and edge locations compared to R1–R3.

functions) to produce the profiles seen in Fig. 4. The scan lengths run from 10 to 600 mm. Not only does the beam profile become much broader as a function of increasing scan lengths, as expected, the amplitude of the relative dose at the center ($z=0$) of the scan length L increases as L increases until reaching a limiting equilibrium dose for $L > 400$ mm. The increasing amplitude as a function of scan length is a

consequence of the low amplitude, long range tails of the dose spread function. This figure demonstrates that the dose to the patient at the center of the region scanned increases markedly with L until approaching its equilibrium value.

Figure 5(a) illustrates the increase in dose at the center of the scan length (i.e., at $z=0$), which occurs with increasing scan lengths, for all five regions (R1–R3, center and edge)

FIG. 4. This figure illustrates the relative dose is a function of position for nine different scan lengths; 10, 50, 100, 150, 200, 300, 400, 500, and 600 mm. In addition to becoming wider, the amplitude of the dose distributions becomes greater as the width increases due to the influence of the very long low amplitude scatter tails in the DSFs, as shown in Fig. 3.

assessed. The increase seen in this figure is the “rise to the equilibrium dose,” as discussed by Dixon.¹⁵ A 120 kVp spectrum on a 32 cm diameter water phantom with a Siemens body bow tie filter was simulated. As the scan length increases, the dose increases and finally asymptotes to the

(a)

(b)

FIG. 5. (a) The relative dose at the center of the scan field of view, as a function of scan length, is illustrated for the five regions in the phantom for a 32 cm water phantom imaged at 120 kVp with a Siemens scanner. (b) The rise to the equilibrium dose as a function of scan length is shown for a number of phantom diameters.

equilibrium value. The increase in the relative dose values for each region from the 100 mm scan length (corresponding to CTDI₁₀₀ measurements) demonstrate one of the limitations of the use of a 100 mm pencil chamber for CT dosimetry. The increases in dose from a 100 mm scan length to a 600 mm scan length (32 cm water phantom at 120 kVp) were 43%, 27%, and 16% for regions R1–R3, respectively. For the center and edge values, the increases were 56% and 14%, respectively. Figure 5(b) illustrates the influence of scan length with the family of curves reflecting the dose at the center location over a range of different phantom diameters. This relative dose increase at $z=0$ with increasing scan length is due to its augmentation by scatter reaching the central dose point from outlying rotations and there is simply more scatter (higher SPR) in larger diameter phantoms.

Figure 6(a) illustrates the scatter to primary dose ratio (SPR) as a function of photon energy in a monoenergetic beam for a 32 cm PMMA phantom using a GE body bow tie filter. The relative scatter dose in all five regions (and for the total phantom volume shown as the mean of regions R1–R3) increases relative to the primary dose up to about 60 keV, where the SPR doses then drop off as energy increases. Figure 6(a) shows high SPRs of up to 9 for region R1 for the 32 cm diameter phantom, and these values are lower for the 16 cm diameter phantom results shown in Fig. 6(b). In this smaller cylinder, the maximum SPR for the R1 region is about 4.5. Although the data for the center region (“center”) have much greater quantum noise due to its smaller collection volume (and the fact that this monoenergetic data is not averaged over an entire x-ray spectrum), the dashed trend line demonstrates the mean SPR. Both Figs. 6(a) and 6(b) show the much higher average SPR for the center region as compared to even the R1 region, which is also central to the phantom but is larger in diameter. Figure 6(c) shows the SPR values for the five regions as a function of cylinder diameter, here for a water phantom at 120 kVp. Again, the center hole dose values are substantially higher than even the R1 (center) region.

The well-behaved curves seen in Figs. 5(a) and 5(b) suggest that correction factors can be computed from the sDSF data to allow measurements made over a scan length (or integration distance) of 100 mm (or 300 mm) to the equilibrium doses measured at a 600 mm scan length, when such measurements are made in 600 mm long phantoms. These tables cannot be used with complete accuracy to correct measurements made in the standard 140 mm long phantoms. Table I shows such correction factors for water [Table I(a)], PMMA [Table I(b)], and polyethylene [Table I(c)] for each of the five regions evaluated (R1, R2, R3, center, and edge) over a range of phantom diameters (16–50 cm). These correction factors were computed as the ratio of the doses for the 600 mm scan length (considered the equilibrium dose) to the dose for the 100 mm scan length. Thus, using these correction factors, doses could be measured with 100 mm scan length (e.g., a 100 mm pencil chamber for the center and

FIG. 6. (a) The SPR is shown as a function of x-ray energy at each of the five regions (plus the mean SPR for the whole phantom) for a 32 cm PMMA phantom. Trend lines were included to smooth the noisier data of the center and edge regions. The SPR has a maximum around 60 keV. (b) The SPR is shown for a 16 cm PMMA phantom. The same trends in SPR are seen as in Fig. 6(a); however, the overall SPR magnitude is lower in this smaller diameter phantom. (c) The SPR is shown as a function of the cylinder diameter for the five regions R1, R2, R3, center, and edge. The SPR here is integrated over infinite z [Eq. (1)]. Typical body imaging conditions apply, with a water composition at 120 kVp.

edge regions) and corrected to the equilibrium dose using the factors in Table I(a), Table I(b), or Table I(c). An additional set of tables (Table II) provide the similar correction factors except the values are the ratio of the 600 mm scan length to the 300 mm scan length values.

IV. DISCUSSION

The dose spread functions over a number of parameters were evaluated using Monte Carlo techniques, and the figures illustrate some of the dependencies that were observed. In addition to the data illustrated, the raw data compiled in this investigation have been deposited electronically using the Electronic Physics Auxiliary Publication Service (EPAPS).¹⁶ The graphs provided in this paper are meant primarily to provide a small sample of the parameter space that was evaluated, and the electronically stored data allow interested parties to make use of the bulk of the sDSF data computed.

The raw scatter component of the dose spread functions are illustrated in Fig. 3, and the effects of region, phantom composition, phantom diameter, bow tie filter, and x-ray spectrum were demonstrated. Figure 3 shows only the scatter distribution (sDSF), while the more practical result using realistic scan lengths for CT (Figs. 4 and 5) include the contribution from both scatter and primary radiation (DSF).

The dose in CT is dominated by scattered radiation, which can be >14 times that of primary radiation for a 32 cm diameter PMMA phantom at 60 keV at the center of the phantom [Fig. 6(a)]. The center regions of the phantom (along its axis) receive more scattered radiation dose deposition due to solid angle effects than more peripheral regions, but they also receive the least primary radiation due to attenuation. Both of these trends tend to maximize the SPR at the center of the symmetrical phantoms evaluated in this study.

The correction factors contained in Table I(b) (PMMA phantom) for the 16 and 32 cm can be compared to the previously determined values published as Fig. 4 in Ref. 6. The four comparison values in Table I(b) are bolded, and the average difference in these comparative values is 0.8%. For the body phantom (32 cm PMMA), the peripheral and center values in Table I(b) are within 2% of those reported by Dixon and Ballard (Table VI in Ref. 4).

The main thrust of this study was to demonstrate differences in the relative dose distribution, and not on absolute dosimetry per se. However, it should be noted that the concept of the dose spread function lends itself well to absolute dosimetry.

There are limitations in this study. A point x-ray source was simulated, while the intensity distribution, which results from a real CT x-ray tube focal spot, is much broader. Nevertheless, since the scale of the current study addresses scatter distribution distances measured in multiple centimeters, it is likely that a finite focal spot of the order of 2 mm will have a negligible influence on the data presented. The phantom geometry simulated in this work assumed an infinite phantom along the z dimension, and this will lead to slightly higher scatter levels (i.e., higher amplitude sDSFs) due to the backscattered x rays from more distance (away from $z=0$) scatter media.

The collimated primary beam incident upon the mathematical phantoms in this simulation study was much narrower than that found in modern CT scanners. The largest concern with this point relates to differences in angular di-

TABLE I. Correction factors are provided which convert central dose measurements performed with a scan length of 100 mm to that of a 600 mm scan length. All data are for 120 kVp.

Dia	R1	R2	R3	Center	Edge
(a) Correction factors for 100 mm scan length water phantom					
16	1.197	1.143	1.100	1.231	1.108
20	1.264	1.182	1.121	1.319	1.124
25	1.339	1.223	1.140	1.423	1.136
32	1.431	1.268	1.156	1.557	1.141
40	1.522	1.308	1.166	1.706	1.139
50	1.627	1.356	1.174	1.895	1.132
(b) Correction factors for 100 mm scan length PMMA phantom					
16	1.217	1.155	1.106	1.256	1.117
20	1.289	1.196	1.127	1.350	1.132
25	1.369	1.238	1.145	1.460	1.142
32	1.465	1.283	1.160	1.608	1.145
40	1.560	1.324	1.170	1.765	1.140
50	1.669	1.375	1.180	1.966	1.142
(c) Correction factors for 100 mm scan length polyethylene phantom					
16	1.239	1.179	1.130	1.274	1.139
20	1.327	1.236	1.164	1.386	1.167
25	1.428	1.296	1.195	1.512	1.190
32	1.552	1.363	1.223	1.691	1.202
40	1.674	1.421	1.239	1.875	1.202
50	1.811	1.483	1.249	2.083	1.194

vergence across the collimated field (in z). In these simulations, the divergence was virtually nil, whereas the maximum divergence angles for a hypothetical 50 cm source to isocenter distance scanner with 20, 30, and 40 mm wide colli-

ated x-ray beams at the isocenter are 1.1° , 1.7° , and 2.3° , respectively. These angular differences are likely to play a second order role (or less) in the sDSFs demonstrated in this work. However, for the extremely wide cone beam scanners

TABLE II. Correction factors are provided which convert central dose measurements performed with a scan length of 300 mm to that of a 600 mm scan length. All data are for 120 kVp.

Dia	R1	R2	R3	Center	Edge
(a) Correction factors for 300 mm scan length water phantom					
16	1.013	1.010	1.007	1.015	1.008
20	1.020	1.015	1.010	1.024	1.011
25	1.030	1.020	1.013	1.037	1.013
32	1.042	1.026	1.016	1.055	1.014
40	1.056	1.032	1.017	1.079	1.015
50	1.073	1.037	1.018	1.107	1.014
(b) Correction factors for 300 mm scan length PMMA phantom					
16	1.013	1.010	1.007	1.015	1.008
20	1.020	1.015	1.010	1.025	1.010
25	1.030	1.020	1.013	1.037	1.012
32	1.042	1.025	1.015	1.054	1.014
40	1.056	1.030	1.016	1.080	1.013
50	1.071	1.035	1.017	1.113	1.013
(c) Correction factors for 300 mm scan length polyethylene phantom					
16	1.019	1.015	1.012	1.022	1.013
20	1.031	1.024	1.017	1.036	1.018
25	1.045	1.033	1.023	1.054	1.022
32	1.065	1.044	1.029	1.083	1.025
40	1.085	1.053	1.032	1.106	1.029
50	1.108	1.062	1.033	1.147	1.026

which have z coverage of 15 cm (8.5° beam divergence) to 20 cm (11.7° beam divergence) or more, the angular divergence in z is appreciable and the results of this study may not be applicable.

The methods used in this study assumed the geometry of an axial CT scan—with no table motion during the 2π rotation of the gantry. While the results at the center of the cylinder will likely compare well to helical scans with a pitch of unity, the peripheral region results may have differences. One way in which to accommodate these differences in future work using the same general approach would be to break up the computation to a single fan angle. Such an increase in granularity would allow dose spread functions to be determined for each region as described in this work; however, a 2π rotation of the dose deposition map, with appropriate adjustment for the z position changes which take place during helical scan, would be needed.

V. CONCLUSIONS

Monte Carlo simulations were used to compute the dose spread functions in CT over a range of parameters including object composition and diameter, x-ray beam energy, position in the phantom, and bow tie filter. The DSFs were used to compute relative dose as a function of scan length, and from these data an array of correction factors was computed which allows computation of the equilibrium dose (central dose in the limit of a very long scan length) from point measurements of dose in 100 or 300 mm scan lengths. The DSF and other data computed in this study are available to interested parties at (<http://www.aip.org/pubservs/epaps.html>)

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