Accuracy of Monte Carlo simulations compared to *in-vivo* MDCT dosimetry

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Purpose: The purpose of this study was to assess the accuracy of a Monte Carlo simulation-based method for estimating radiation dose from multidetector computed tomography (MDCT) by comparing simulated doses in ten patients to *in-vivo* dose measurements.

Methods: MD Anderson Cancer Center Institutional Review Board approved the acquisition of *in-vivo* rectal dose measurements in a pilot study of ten patients undergoing virtual colonoscopy. The dose measurements were obtained by affixing TLD capsules to the inner lumen of rectal catheters. Voxelized patient models were generated from the MDCT images of the ten patients, and the dose to the TLD for all exposures was estimated using Monte Carlo based simulations. The Monte Carlo simulation results were compared to the *in-vivo* dose measurements to determine accuracy.

Results: The calculated mean percent difference between TLD measurements and Monte Carlo simulations was -4.9% with standard deviation of 8.7% and a range of -22.7% to 5.7% .

Conclusions: The results of this study demonstrate very good agreement between simulated and measured doses *in-vivo*. Taken together with previous validation efforts, this work demonstrates that the Monte Carlo simulation methods can provide accurate estimates of radiation dose in patients undergoing CT examinations. © 2015 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4906178>]

Key words: Monte Carlo, *in vivo*, radiation dose, dosimetry, validation, TLD, VC, MDCT

1. INTRODUCTION

The increasing use of computed tomography (CT) in diagnostic radiology is a large contributor of the increasing radiation dose to the public from medical exposures.^{1,2} Concerns about stochastic effects due to increased radiation dose from medical procedures have been a recent motivation to not only report radiation dose from CT but also to improve currently used metrics such as CTDI to patient size-specific dose estimates (SSDEs).³⁻⁸ However, CTDI (Ref. 9) and SSDE do not represent organ dose, which is necessary for estimating individual risk from ionizing radiation. Although these metrics can guide the improvement of our clinical practice,

they should not be used for assessing the risk from diagnostic imaging procedures for individual patients.⁸ It has been shown that the most appropriate metric for assessing this risk to specific subject is the radiation dose to individual organs.^{3,10-14}

One feasible approach to estimating organ dose to patients undergoing CT examinations would be simulations using Monte Carlo radiation transport codes.¹⁵⁻²⁶ However, these Monte Carlo based simulations require a complete knowledge of the simulated CT scanner and the patient, such as the CT x-ray source spectrum (energy and flux), filtration, bowtie, collimation, and imaging parameters for a specific examination such as tube current, kVp, scan length, pitch,

and rotation time. In addition to the required information for simulating a specific scanner, these codes have to be validated and benchmarked against actual experimental measurements to ensure the accuracy of simulated organ doses.

Most of these validation steps, performed by different groups focusing on Monte Carlo simulations, have used the computed tomography dose index (CTDI) phantom and compared actual CTDI measurements with simulated values, and more recent investigations have incorporated anthropomorphic phantoms to improve on the CTDI phantom's homogeneous composition and simple geometric shape.^{27–32} The main limitation of these validation approaches is the use of these phantoms, which do not fully represent the anatomic complexity of a typical patient.

The purpose of this study was to extend the validation of Monte Carlo based simulation dose estimates in patient models by making direct comparisons to *in-vivo* dose measurements. This study used a previously developed Monte Carlo simulation method, which had been validated against CTDI phantoms, and compared *in-vivo* TLD measurements, affixed to the inner lumen of rectal catheters of patients undergoing virtual colonoscopy (VC), with Monte Carlo based simulated dose estimates.

2. MATERIALS AND METHODS

2.A. *In-vivo* dose measurements

2.A.1. Virtual colonoscopy procedure at MD Anderson

MD Anderson Cancer Center Institutional Review Board (IRB) approval was obtained to perform *in-vivo* rectal radiation dose measurements for patients undergoing VC. This cohort consisted of 10 volunteers (five men and five women; age range, 23–87 yr; mean age, 70.4 yr). Effective diameter of these patients ranged from 25.5 to 31.7 cm.

Virtual colonoscopy was first described in 1994 as a minimally invasive means for identifying colorectal polyps and colorectal lesions.³³ The procedure consists of (1) bowel cleansing with cathartic agents, (2) gas insufflation with carbon dioxide, (3) CT scanning of the abdomen pelvis in supine and prone positions, and (4) image analysis for identification of lesions. IRB approval was granted for this study to affix two double-chamber TLDs (i.e., a total of four repeated reading) in the inner lumen of the standard rectal catheter tips (Fig. 1).

FIG. 1. Two views of the rectal catheter tip containing TLD capsules.

The TLDs were placed inside the rectal catheter prior to the CT exam and remained there during the duration of the exam.

All scans were performed on the same model multidetector computed tomography (MDCT) scanner at MD Anderson Cancer Center (LightSpeed VCT, GE Healthcare, Waukesha, WI). For each patient exam, there were six dose contributions to the accumulated dose; four from localizer images and two from one prone and one supine helical CT scans. Table I lists the parameters used for all six scans. It should be noted that these were fixed tube current (mA) scans and that no adjustment in technique was performed based on patient size. For all helical scans, beam width of 40 mm and the large bowtie was used.

2.A.2. TLD readouts and determination of TLD dose response

Once a minimum of 14 days had passed from the actual VC procedure, the TLD readout was performed by the Radiologic Physics Center (RPC) at MD Anderson. RPC provided the following information for the purpose of calculating patient rectal dose: TLD aliquot charge reading Q (in μC), aliquot TLD mass m (in mg), sensitivity correction factor K_s (in $\text{cGy}/\mu\text{C}/\text{mg}$), and fading correction factor K_f (unitless). These values were used in Eq. (1) to calculate dose to TLDs which was reported as dose to muscle,

$$D_{\text{TLD,Uncorrected}} = \frac{Q}{m} \times K_s \times K_f. \quad (1)$$

To further correct the calculated value for linearity, a calibration curve at 120 kVp was generated using identical measurements performed with a 0.6 cm^3 Farmer ionization chamber and TLDs. The Farmer chamber was placed in the center of a 32 cm CTDI phantom and irradiated using the VC protocol shown in Table I. The resulting exposure was corrected for muscle using the f -factor of air to muscle. Next, the Farmer chamber was replaced with TLD-loaded insert rod (four TLD measurements at each mA s, i.e., two capsules with two chambers) and irradiated with the same VC protocol. This procedure was repeated from 20 to 170 mAs tube current with 10 mAs increments for both the Farmer chamber and TLDs. The resulting calibration curve from these measurements is shown in Fig. 2. A linear regression model was used to fit the data and the resulting linear regression equation was used to correct the resulting TLD dose values from Eq. (1) for linearity [Eq. (2)],

$$D_{\text{TLD}} = D_{\text{Farmer chamber}} = 0.8356 \times D_{\text{TLD,Uncorrected}} - 2.5041. \quad (2)$$

A recently published manuscript by scientists at MD Anderson can be consulted for a more detailed description of the TLD readout procedure and determination of TLD dose response.³⁴

2.B. Monte Carlo simulation code

A previously developed and *ex-vivo*-validated (using CTDI_{100} measurements) Monte Carlo based CT dosimetry package was employed to estimate dose to the TLDs from virtual colonoscopy CT.^{15–17,35} MCNPX (Monte Carlo N-Particle eXtended v2.6.0), a general-purpose radiation transport code developed at Los Alamos National laboratory, was modified

TABLE I. MD Anderson CT imaging protocol used for VC scans.

Series	Description	Scan type	Area	Rotation time	Image thickness	Pitch	kVp	mA
1	Scout supine	PA and lateral	Midsternum to trochanter				120	10
2	Supine	Helical	Diaphragm to symphysis	0.5	1.25	0.984	120	100
3	Scout prone	AP and lateral	Midsternum to trochanter				120	80
4	Prone	Helical	Diaphragm to symphysis	0.5	1.25	0.984	120	100

to model a MDCT scanner for the simulations.^{36,37} All simulations were performed in photon transport mode with a low-energy cutoff of 1 keV. In photon mode, only photon interactions are tracked, while secondary electrons are assumed to deposit their energy at the photon interaction site. This assumption satisfies the condition of charged particle equilibrium (CPE), in which kerma can be assumed to be equal to absorbed dose.

2.B.1. GE LightSpeed VCT source model

A multidetector row CT scanner (GE LightSpeed VCT, GE Healthcare) was modeled on the platform of MCNPX. The default MCNPX particle source code was modified to model the scanner (geometry, spectrum, and filtration). The CT x-ray source trajectory was reproduced by simulating a helical source path while using a point source to emit photons. The initial position and direction of each simulated photon was randomly selected based on scanner geometry specifications such as source-to-isocenter distance and fan angle. Each simulated photon was provided with a statistical weight for modeling the bowtie filter of the scanner.^{15–17} The “equivalent source” method described by Turner *et al.*³⁵ was used to generate scanner-specific spectrum and filtration description

to model scanner’s x-ray spectrum. Simple validation methods comparing measurements and simulations of CTDI₁₀₀ at the center and periphery of both 32 and 16 cm CTDI phantoms were performed and an average RMS error of approximately 5% between the measured and simulated values was reported.³⁵

2.B.2. Voxelized virtual colonoscopy models

Axial CT images (prone and supine) of all ten virtual colonoscopy patients were collected for creating voxelized models for use in simulations. Collected patient images were reconstructed at 500 mm field of view (FOV) to ensure the coverage of the entire anatomy within images used to create voxelized models. Anonymized images were imported into a workstation³⁸ for the purpose of creating the voxelized models needed for simulation and to specifically identify voxels representing the TLDs. TLDs were identified and manually contoured as a single entity on each image set. The rectal catheter was also identified and manually segmented along with the CT table and contrast media in patient, if seen on the images. Contoured TLDs, rectal catheter, table, and contrast, if present, were set to muscle, plastic, graphite, and water, respectively, using each material’s composition and density. A Hounsfield unit-to-tissue lookup table was used to assign each voxel other than the segmented region to one of the six tissue types (air, water, lung, fat, muscle, and bone) with 17 different tissue density levels as a function of voxel CT number.^{39,40} Figure 3 illustrates the segmented TLDs, catheter, table, and contrast media on a single image. To avoid confusion with bone, the small amount of contrast present in the colon was segmented and its material type was set to water; given the small amount of contrast material present, this should have little to no effect on the simulated absorbed dose in the TLD.

For all but one patient, two sets of voxelized models were generated; one from the prone scan and the other from the supine scan. For the remaining one patient, an additional decubitus scan was performed under the direction of the supervising radiologist. For this specific patient, three sets of voxelized models were created and used to simulate dose to segmented TLDs. Additionally, for this patient, three sets of scouts (AP/PA and lateral) were acquired.

FIG. 2. Linear regression model predicting Farmer ionization chamber response from TLD dose measurements uncorrected for linearity and energy.

FIG. 3. Zoomed view of the rectal region. TLD segmentation is shown in a white dot and the catheter in dark gray.

2.B.3. Simulated VC examination

For each voxelized model, fixed tube current scans representing supine and prone scans were simulated using the virtual colonoscopy scan parameters shown in Table I. Since voxelized models were based on the image data, the z -axis over-ranging distance used for helical reconstruction was missing, not only on the image data but also on the voxelized models. Hence, the dose contribution due to scatter from the over-ranging distance was not taken into account when simulating dose to the TLDs.²⁶ This contribution is assumed to be negligible because the region in which energy is tallied is assumed to be not close to either edge of the helical acquisition or fairly within the scanned region.

Another scan parameter not modeled in these simulations was the tube start angle which was arbitrarily set to 0° for all patients because this information was not available. As described by Zhang *et al.*,⁴¹ tube start angle can have a considerable effect on surface dose, but it has lower impact on more internally/centrally located organs. The TLDs were more internally placed in the patients, and therefore, we assume little or no variation in terms of dose due to start angle effects.

Additionally, projectional simulations using each patient's scan length were performed to estimate dose from the projectional localizer radiographs (scout views). For each patient, except for one, four sets of scouts, two for each helical scan were simulated and added to the simulation results of the prone and supine scans. For one patient, an additional helical scan (in decubitus position) and its associated localizer were also simulated.

2.B.4. Dose calculations

For each patient model, absorbed dose to TLDs was calculated using collision kerma, which is equal to absorbed dose under the assumption of charge particle equilibrium.

For each simulated photon, MCNPX tally type $*F4$ was used to track energy fluence in segmented TLDs and multiplied by mass energy-absorption coefficients (μ_{en}/ρ) to convert to collision kerma. The resulting dose per simulated photon was converted to dose per mA by multiplying the Monte Carlo output by a scanner, kVp, and collimation-independent normalization factor.¹⁵⁻¹⁷ Absolute organ doses were obtained by multiplying dose per mA s by the mAs, used to scan the patient, and the number of gantry rotations. For calculating dose from the localizer radiographs, the Monte Carlo output was simply multiplied by a conversion factor to convert Monte Carlo output from energy deposited per particle to particle per mA s and then multiplied by total mAs using scan time per localizer.

3. RESULTS

Table II summarizes the results obtained from both measurements and simulations. This table also shows the percent differences between these two values for each patient, using the measured value as reference. Percent differences range from a minimum of -23% to a maximum of 5.7% with an average of -4.9% and a standard deviation of 8.7% . For nine out of ten patients, the percent differences were less than 9% .

As mentioned in the methods, due to unavailable information of the tube start angle, all simulations for prone and supine scans used 0° (12'o clock position) as their tube start angle. Since TLDs were placed more internally within the patients, we anticipated a very small effect on dose to TLDs due to tube start angle.

4. DISCUSSION

This work compared physical measurements of dose with Monte Carlo based simulated dose estimates and to our knowledge is the first study to validate a Monte Carlo model of a MDCT scanner using *in-vivo* measurements performed during routine diagnostic/screen CT exams. Comparisons resulted in very good agreements between measurements and simulations; for nine out of ten patients simulated results were within less than 9% of measurements. For patient model 10, percent difference was -23% . Monte Carlo simulations underestimated the measured dose for patient model 10. Investigating the disagreement between measurement and simulation for this model uncovered some possible causes for higher percent difference.

For this particular patient, the prone helical image set did not cover the entire length of the TLDs, i.e., the last image still contained visible TLD capsule. The voxelized model based on these images is not only missing part of the dosimeter, but also since the dosimeter is at the end of the scan acquisition, z -axis over ranging, which is not modeled for these patients, had a larger effect on dose for this specific patient model than for the other patient models. Scatter from this region can substantially contribute to the total dose measured by the TLD.^{42,43} This was the most likely cause for the underestimation of the dose to TLDs in this patient.

TABLE II. Tabulated measurements and simulated doses to TLDs with their percent differences. TLD measurements include all localizer exposures.

Patient model	Helical scan	Simulated ^a helical dose (mGy)	Total simulated ^a scouts dose (mGy)	Total simulated ^a dose contribution of localizers (mGy)	TLD measurements (mGy)	Percent difference
Pt. 1	Supine	4.7	0.9	9.3	10.1(±0.8)	-7.6
	Prone	3.7				
Pt. 2	Supine	6.0	1.2	12.1	12.5(±0.5)	-3.2
	Prone	4.9				
Pt. 3	Supine	3.9	1.0	8.9	9.7(±0.1)	-8.3
	Prone	4.1				
Pt. 4	Supine	3.9	0.8	8.3	9.0(±0.4)	-8.1
	Prone	3.6				
Pt. 5	Supine	6.0	1.2	13.6	13.0(±0.6)	4.6
	Prone	6.4				
Pt. 6	Supine	6.0	1.1	11.6	11.3(±0.4)	2.4
	Prone	4.4				
Pt. 7	Supine	3.8	1.1	9.5	10.0(±0.6)	-5.1
	Prone	4.5				
Pt. 8	Supine	5.4	1.3	18.7	17.7(±0.2)	5.7
	Prone	5.7				
	Decubitus	6.3				
Pt. 9	Supine	4.2	1.4	10.6	10.1(±0.3)	4.1
	Prone	5.0				
Pt. 10	Supine	11.2	1.0	18.6	24.1(±0.5)	-22.7
	Prone	6.5				
% average (STD)		5.5 (1.7)	1.1(0.2)	12.6(3.8)	12.8(4.7)	-4.9(8.7)
% STD		2.1	0.2	4.9	4.7	5.6
% min		3.6	0.8	8.3	9.0	-8.3
% max		11.2	1.4	23.3	24.1	5.7

^aFor each simulation number of simulated photons was chosen to ensure a relative error less than 1%.

Other possible reasons for discrepancies between measurements and simulated values is the x - y plane shift in the reconstructed images to center the anatomy in the display field of view. Since the image data are used to create the model, the center of the reconstructed image in the x - y plane will be assumed to be the same as the center of the virtual gantry in the Monte Carlo model of the CT scanner. However, if the images were reconstructed at different x and y positions other than $x = 0$ and $y = 0$, the region of interest (ROI) is no longer simulated at its true position during the actual scan. However, since all image sets in this study were reconstructed at 500 mm FOV, we can safely assume that none of these models were shifted in the x - y plane during reconstruction. For cases when images are not reconstructed at 500 mm FOV, any shift in x - y plane can be corrected by shifting the entire geometry to the actual position at which the scan was performed, assuming the magnitude of the actual shift in both directions, x and y , is known.

Another parameter of importance is the z -axis over-ranging distance, which was not included in these voxelized models but as shown in patient number 10, can have an effect on dose depending of where the region of interest is located within the scan length. A possible approximation of the dose contribution

of this effect could be an adjustment using interpolation of dose with respect to distance for a desired location. This could be performed for multiple regions of interest and create a dose lookup table based on the distance of the ROI to the end of image data.

A more immediate solution for this patient (patient model 10) could be doubling the simulation result of the supine helical dataset. As seen in Table II, for all the other patient models, simulated values for prone and supine helical scans are very similar in magnitude. This observation can be used to justify that dose to TLD from the prone helical scan for patient model 10 is approximately the same as the dose from the supine scan. Using this approach, simulated total dose to TLDs for patient model 10 increases to 23.3 mGy compared to a measured dose of 24.1 mGy, a -3.2% difference.

The purpose of this work was the validation of Monte Carlo simulations against physical measurements. In this particular work, the *in-vivo* measurements performed by Mueller *et al.*³⁴ provided an extremely unique opportunity to validate the Monte Carlo simulations. These scans were performed with a relatively straightforward scanning protocol that used a helical scan of pitch 1 with a fixed tube current. Some additional work that describes a series of validation conditions for scans in

which tube current modulation has been used is described here.⁴⁴

5. CONCLUSION AND DISCUSSION

This study is believed to be the first attempt to benchmark Monte Carlo simulated MDCT dose estimates to internal *in-vivo* dose measurements. Overall Monte Carlo simulated rectal doses and *in-vivo* measured rectal doses demonstrated very good agreement.

While discrepancies can be due to the Monte Carlo model itself, such as modeling of the spectra, filtration, and source movement, some discrepancies can be caused by inadequate information available from patient images, as seen with patient number 10, patient modeling, and unavailable imaging parameters, such as tube start angle.

There are essential requirements that images have to meet to be useable for creation of accurate voxelized models. These requirements include images reconstructed at the largest possible FOV to ensure complete visibility of the exposed anatomy and reconstructed images at $x = 0$ and $y = 0$ to ensure the transfer of actual patient position within the gantry to the virtual scanner. In addition to these requirements, there are scanning parameters that are equally important to be known for the simulations, such as tube start angle in helical, axial, and the projectional radiograph mode. In case of tube current modulation algorithm utilization in the scan, individual tube current along with the corresponding table location and tube angle is necessary for a correct representation of the actual scan and accurate simulated doses.

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¹F. A. Mettler, Jr., B. R. Thomadsen, M. Bhargavan, D. B. Gilley, J. E. Gray, J. A. Lipoti, J. McCrohan, T. T. Yoshizumi, and M. Mahesh, "Medical radiation exposure in the U.S. in 2006: Preliminary results," *Health Phys.* **95**, 502–507 (2008).

²National Council on Radiation Protection and Measurements, "Ionizing radiation exposure of the population of the United States," NCRP Report No. 160 (2009).

³American Association of Physicists in Medicine, "The measurement, reporting, and management of radiation dose in CT," AAPM Report No. 96. Report of AAPM Task Group 23 of the Diagnostic Imaging Council CT Committee (2008).

⁴US Food and Drug Administration, "FDA unveils initiative to reduce unnecessary radiation exposure from medical imaging," Food and Drug Administration Silver Spring (2010). Available at <http://www.fda.gov/NewsEvents/Newsroom/PressAnnouncements/ucm200085.htm>.

⁵M. M. Rehani and D. P. Frush, "Patient exposure tracking: The IAEA smart card project," *Radiat. Prot. Dosim.* **147**, 314–316 (2011).

⁶R. D. Neumann and D. A. Bluemke, "Tracking radiation exposure from diagnostic imaging devices at the NIH," *J. Am. Coll. Radiol.* **7**, 87–89 (2010).

⁷State of California, Radiation control: Health facilities and clinics: Records, Bill number AB-510, (2012). Available at <http://leginfo.ca.gov/faces/billVotesClient.xhtml>.

⁸J. M. Boone, K. J. Strauss, D. D. Cody, C. McCollough, M. McNitt-Gray, and T. L. Toth, "Size-Specific Dose Estimates (SSDE) in Pediatric and Adult Body CT Examinations," AAPM Report No. 204 (2011).

⁹C. H. McCollough, S. Leng, L. Yu, D. D. Cody, J. M. Boone, and M. F. McNitt-Gray, "CT dose index and patient dose: They are not the same thing," *Radiology* **259**, 311–316 (2011).

¹⁰National Research Council, *Health Risks from Exposure to Low Levels of Ionizing Radiation. Beir VII Phase 2* (National Academies, 2006).

¹¹International Commission on Radiological Protection and Measurements, "The 1990 recommendations of the International Commission on Radiological Protection. ICRP publication 60," *Ann. ICRP* **21**, 1–32 (1991).

¹²International Commission on Radiological Protection, "The 2007 Recommendations of the International Commission on Radiological Protection. ICRP publication 103," *Ann. ICRP* **37**, 1–332 (2007).

¹³D. Brenner and E. Hall, "Computed tomography—An increasing source of radiation exposure," *N. Engl. J. Med.* **357**, 2277–2284 (2007).

¹⁴E. J. Hall and D. J. Brenner, "Cancer risks from diagnostic radiology," *Br. J. Radiol.* **81**, 362–378 (2008).

¹⁵J. 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).

¹⁶J. J. DeMarco, C. H. Cagnon, D. D. Cody, D. M. Stevens, C. H. McCollough, M. Zankl, E. Angel, and M. F. McNitt-Gray, "Estimating radiation doses from multidetector CT using Monte Carlo simulations: Effects of different size voxelized patient models on magnitudes of organ and effective dose," *Phys. Med. Biol.* **52**, 2583–2597 (2007).

¹⁷G. Jarry, J. J. DeMarco, U. Beifuss, C. H. Cagnon, and M. F. McNitt-Gray, "A Monte Carlo-based method to estimate radiation dose from spiral CT: From phantom testing to patient-specific models," *Phys. Med. Biol.* **48**, 2645–2663 (2003).

¹⁸E. Angel, C. V. Wellnitz, M. M. Goodsitt, N. Yaghamai, J. J. DeMarco, C. H. Cagnon, J. W. Sayre, D. D. Cody, D. M. Stevens, A. N. Primak, C. H. McCollough, and M. F. McNitt-Gray, "Radiation dose to the fetus for pregnant patients undergoing multidetector CT imaging: Monte Carlo simulations estimating fetal dose for a range of gestational age and patient size," *Radiology* **249**, 220–227 (2008).

¹⁹E. Angel, N. Yaghamai, C. M. Jude, J. J. DeMarco, C. H. Cagnon, J. G. Goldin, A. N. Primak, D. M. Stevens, D. D. Cody, C. H. McCollough, and M. F. McNitt-Gray, "Monte Carlo simulations to assess the effects of tube current modulation on breast dose for multidetector CT," *Phys. Med. Biol.* **54**, 497–512 (2009).

²⁰A. C. Turner, M. Zankl, J. J. DeMarco, C. H. Cagnon, D. Zhang, E. Angel, D. D. Cody, D. M. Stevens, C. H. McCollough, and M. F. McNitt-Gray, "The feasibility of a scanner-independent technique to estimate organ dose from MDCT scans: Using CTDI[sub vol] to account for differences between scanners," *Med. Phys.* **37**, 1816–1825 (2010).

²¹A. 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).

²²M. Caon, G. Bibbo, and J. Pattison, "An EGS4-ready tomographic computational model of a 14-year-old female torso for calculating organ doses from CT examinations," *Phys. Med. Biol.* **44**, 2213–2225 (1999).

²³C. Lee, C. Lee, R. J. Staton, D. E. Hintenlang, M. M. Arreola, J. L. Williams, and W. E. Bolch, "Organ and effective doses in pediatric patients undergoing helical multislice computed tomography examination," *Med. Phys.* **34**, 1858–1873 (2007).

²⁴A. Khursheed, M. C. Hillier, P. C. Shrimpton, and B. F. Wall, "Influence of patient age on normalized effective doses calculated for CT examinations," *Br. J. Radiol.* **75**, 819–830 (2002).

²⁵W. Chen, D. Kolditz, M. Beister, R. Bohle, and W. A. Kalender, "Fast on-site Monte Carlo tool for dose calculations in CT applications," *Med. Phys.* **39**, 2985–2996 (2012).

²⁶M. Khatonabadi, D. Zhang, K. Mathieu, H. J. Kim, P. Lu, D. Cody, J. J. DeMarco, C. H. Cagnon, and M. F. McNitt-Gray, "A comparison of methods to estimate organ doses in CT when utilizing approximations to the tube current modulation function," *Med. Phys.* **39**, 5212–5228 (2012).

²⁷B. Schmidt and W. A. Kalender, "A fast voxel-based Monte Carlo method for scanner- and patient-specific dose calculations in computed tomography," *Phys. Med.* **18**, 43–53 (2002).

²⁸P. Deak, M. van Straten, P. C. Shrimpton, M. Zankl, and W. A. Kalender, "Validation of a Monte Carlo tool for patient-specific dose simulations in multi-slice computed tomography," *Eur. Radiol.* **18**, 759–772 (2008).

²⁹R. Schmidt, J. Wulff, B. Kastner, D. Jany, J. T. Heverhagen, M. Fiebig, and K. Zink, "Monte Carlo based calculation of patient exposure in x-ray CT examinations," *IFMBE Proc.* **22**, 2487–2490 (2009).

³⁰X. 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 I. Development and validation of a Monte Carlo program," *Med. Phys.* **38**, 397–407 (2011).
- ³¹T. Fearon, H. Xie, J. Y. Cheng, H. Ning, Y. Zhuge, and R. W. Miller, "Patient-specific CT dosimetry calculation: A feasibility study," *J. Appl. Clin. Med. Phys.* **12**, 196–209 (2011).
 - ³²D. 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).
 - ³³G. D. Vining, D. J. Bechtold, R. E. Scharling, E. S. Grishaw, and E. K. Shifrin, "Technical feasibility of colon imaging with helical CT and virtual reality," *AJR, Am. J. Roentgenol* **162**(Suppl. 104) (1994).
 - ³⁴J. W. Mueller, D. J. Vining, A. K. Jones, D. Followill, V. E. Johnson, P. Bhosale, J. Rong, and D. D. Cody, "In vivo CT dosimetry during CT colonography," *AJR, Am. J. Roentgenol.* **202**, 703–710 (2014).
 - ³⁵A. Turner, "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).
 - ³⁶E. Waters, "MCNPX version 2.5.C," Los Alamos National Laboratory Report LA-UR-03-2202 (2003).
 - ³⁷E. Waters, "MCNPX user's manual version 2.4.0," Los Alamos National Laboratory Report LA-CP-02-408 (2002).
 - ³⁸M. S. Brown, R. Pais, P. Qing, S. Shah, M. F. McNitt-Gray, J. G. Goldin, I. Petkovska, L. Tran, and D. R. Aberle, "An architecture for computer-aided detection and radiologic measurement of lung nodules in clinical trials," *Cancer Inf.* **4**, 25–31 (2007).
 - ³⁹J. J. DeMarco, T. D. Solberg, and J. B. Smathers, "A CT-based Monte Carlo simulation tool for dosimetry planning and analysis," *Med. Phys.* **25**, 1–11 (1998).
 - ⁴⁰E. Angel, C. V. Wellnitz, M. M. Goodsitt, N. Yaghmai, J. J. DeMarco, C. H. Cagnon, J. W. Sayre, D. D. Cody, D. M. Stevens, A. N. Primak, C. H. McCollough, and M. F. McNitt-Gray, "Radiation dose to the fetus for pregnant patients undergoing multidetector CT Imaging: Monte Carlo simulations estimating fetal dose for a range of gestational age and patient size1," *Radiology* **249**, 220–227 (2008).
 - ⁴¹D. Zhang *et al.*, "Variability of surface and center position radiation dose in MDCT: Monte Carlo simulations using CTDI and anthropomorphic phantoms," *Med. Phys.* **36**, 1025–1038 (2009).
 - ⁴²A. Schilham, A. J. van der Molen, M. Prokop, and H. W. de Jong, "Over-ranging at multislice CT: An underestimated source of excess radiation exposure," *RadioGraphics* **30**, 1057–1067 (2010).
 - ⁴³C. J. Tien, J. F. Winslow, and D. E. Hintenlang, "A methodology for direct quantification of over-ranging length in helical computed tomography with real-time dosimetry," *J. Appl. Clin. Med. Phys.* **12**, 350–361 (2011).
 - ⁴⁴M. Bostani, K. McMillan, J. J. Demarco, C. H. Cagnon, and M. F. McNitt-Gray, "Validation of a Monte Carlo model used for simulating tube current modulation in computed tomography over a wide range of phantom conditions/challenges," *Med. Phys.* **41**(11), 112101 (10pp.) (2014).