Monte Carlo simulations of adult and pediatric computed tomography exams: Validation studies of organ doses with physical phantoms

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Purpose: To validate the accuracy of a Monte Carlo source model of the Siemens SOMATOM Sensation 16 CT scanner using organ doses measured in physical anthropomorphic phantoms.

Methods: The x-ray output of the Siemens SOMATOM Sensation 16 multidetector CT scanner was simulated within the Monte Carlo radiation transport code, MCNPX version 2.6. The resulting source model was able to perform various simulated axial and helical computed tomographic (CT) scans of varying scan parameters, including beam energy, filtration, pitch, and beam collimation. Two custom-built anthropomorphic phantoms were used to take dose measurements on the CT scanner: an adult male and a 9-month-old. The adult male is a physical replica of the University of Florida reference adult male hybrid computational phantom, while the 9-month-old is a replica of the University of Florida Series B 9-month-old voxel computational phantom. Each phantom underwent a series of axial and helical CT scans, during which organ doses were measured using fiber-optic coupled plastic scintillator dosimeters developed at the University of Florida. The physical setup was reproduced and simulated in MCNPX using the CT source model and the computational phantoms upon which the anthropomorphic phantoms were constructed. Average organ doses were then calculated based upon these MCNPX results.

Results: For all CT scans, good agreement was seen between measured and simulated organ doses. For the adult male, the percent differences were within 16% for axial scans, and within 18% for helical scans. For the 9-month-old, the percent differences were all within 15% for both the axial and helical scans. These results are comparable to previously published validation studies using GE scanners and commercially available anthropomorphic phantoms.

Conclusions: Overall results of this study show that the Monte Carlo source model can be used to accurately and reliably calculate organ doses for patients undergoing a variety of axial or helical CT examinations on the Siemens SOMATOM Sensation 16 scanner. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4771934>]

Key words: computed tomography, Monte Carlo, organ dose, anthropomorphic phantom

I. INTRODUCTION

Although there are a variety of methods by which one can estimate patient organ doses from computed tomography (CT) examinations, Monte Carlo (MC) simulations have been reported to be the most accurate, reliable, and versatile in accomplishing this task.¹⁻¹⁰ In the Monte Carlo method, the patient and CT scanner are simulated using a computational anatomic model of the patient and an x-ray source model representing the scanner's beam output. However, to ensure ac-

curacy of these calculations, these CT source models must be benchmarked and validated against actual experimental measurements made on the scanners they simulate. In the past, most validation studies were accomplished using standard CT dose index (CTDI) phantoms, but in recent years, anthropomorphic phantoms have been increasingly utilized.^{1,5,11}

In previous work by Lee *et al.*,¹² a Monte Carlo CT source model of a Siemens Sensation 16 had been validated against standard CTDI measurements. However, this validation did not include testing of the model's ability to simulate helical

scans, nor the ability to calculate doses for more complex patient and phantom geometries. The goal of the present study was thus to further validate the source model against organ dose measurements made on two different anthropomorphic phantoms for multiple axial and helical scanning protocols.

II. MATERIALS AND METHODS

II.A. CT scanner description

A SOMATOM Sensation 16 multidetector CT scanner (Siemens Medical Solutions, Erlangen, Germany) was used as the basis for the Monte Carlo source model as well as for all physical dose measurements. The scanner contains 16 rows of detectors and allows beam collimations from 0.75 to 24 mm, has the ability to scan in both axial and variable-pitch helical modes, and features two settings of inherent filtration for head and body scanning along with a single bowtie beam-shaping filter. The fan beam angle is 52° , with a focal-spot-to-isocenter distance of 57 cm. The operator can select tube potentials of 80, 100, 120, or 140 kVp along with varying tube current and gantry rotation speeds. The scanner also allows for use of tube-current modulation during scanning, but this feature was not used in the measurements of this study.

II.B. Modeling of the CT scanner x-ray source

The source model of the scanner was created as a custom source file within a general-purpose Monte Carlo radiation transport code, MCNPX version 2.6.¹³ Material and thickness data for the two inherent filters for head and body scanning were obtained from the manufacturer. This information, in conjunction with the commercial spectrum generation program SPEC78 (Institute of Physics and Engineering in Medicine), was used to create x-ray spectra for the head and body filters for all beam energies.¹⁴ These spectra were incorporated into the MCNPX source model through an input deck for energy sampling. To account for the effects of the bowtie filter on the shape of the fan beam for all beam energy/filter combinations, angular-dependent weighting factors were applied in the source model based on free-in-air lateral dose profile measurements made previously by a pencil ion chamber while the x-ray tube was fixed at the 12 o'clock position in service mode.¹² The effects of overbeaming on the true beam thicknesses for various collimation settings were previously quantified in studies using radiographic film, and were thus taken into account within the source model.¹⁵

The custom source model allows the user to simulate both axial exams and helical exams of varying pitch. For helical exams, this is accomplished by having the source first sample the location of the x-ray focal spot along a mathematically described helix based on the pitch and scan length selected by the user, as well as the previously defined focal-spot-to-isocenter distance of 57 cm. In addition to the specified helical scan length, the source model automatically adds a half rotation at either end of the scan length in order to account for over-ranging. For axial exams, the focal spot location is sampled along a series of circular rings of radius 57 cm spanning

the total specified length of the scan that is spaced apart by the distance of the selected beam collimation. After selecting this starting location, the source samples an angle within the 52° fan beam as well as within the beam collimation thickness, thereby selecting an initial photon direction. Finally, the photon energy is sampled based on the energy spectrum selected for the particular exam. This process is repeated for the total number of particles to be transported. The final version of the source model allows the user to select beam energy, head or body filtration, beam collimation, an axial or a helical exam (with associated pitch), and starting angle of the beam for helical exams.

II.C. Characterization and calibration of PSD dosimetry system

All dose measurements were made using a fiber-optic coupled plastic scintillator dosimetry (PSD) system developed at UF.¹⁶ The detector system measures scintillation photon counts through sampling the signal produced by a photomultiplier tube connected to the detector fibers at a frequency of 20 Hz. These counts correspond linearly to the absorbed dose deposited to the PSDs when properly calibrated to the beam energy of interest. Consequently, calibration factors are needed to convert photon counts to tissue-absorbed dose. This calibration was performed using a Radcal 10×6 -0.6-CT small volume ion chamber and both head and body acrylic CTDI phantoms. Two sets of air kerma measurements were taken with the ion chamber centered within the central and 12 o'clock channels of the CTDI phantoms during a 120 kVp (with body filter for body CTDI phantom and head filter for head CTDI phantom). High-mAs, low-pitch scans of the entire phantom length are taken, which were then rescaled to tissue kerma using tissue-to-air mass energy absorption coefficient ratios. The tissue kerma was assumed to approximately equal tissue absorbed dose due to the use of diagnostic x-ray energies. These ratios were calculated for both soft tissue-to-air and lung tissue-to-air (for lung dose measurements) for effective energies of the 120 kVp head and body energy spectra based on measured half-value layers of both spectra. These exact scans were repeated with the PSDs placed in identical positions and counts received were then recorded. Dose and count information in both channel positions were then used to establish calibration factors that were then averaged to give nominal calibration factors for both the head and body CTDI phantoms. The calibration factor for the head phantom was used for the 9-month-old physical phantom measurements, while the body phantom calibration factor was applied to the adult male physical phantom measurements.

This calibration method was developed in order to provide an easy and efficient means of calibrating the PSD fibers while simultaneously maintaining as close to the actual irradiation conditions present during dose measurements within the anthropomorphic phantoms (see Sec. II.D). However, this calibration method does result in dose measurement uncertainties due to the assumptions made and the inherent energy-dependence characteristics of the PSD fibers as described in Ref. 16. Resultantly, an extensive error propagation analysis

FIG. 1. Frontal view of the (a) UF reference adult male physical phantom, and (b) UF Series B 9-month-old male physical phantom.

was performed, as given in Sec. IV.D, for each dose measurement to better quantify these uncertainties.

II.D. Anthropomorphic phantom dose measurements

In order to validate the use of this Monte Carlo source in calculating patient organ doses, measurements were taken in two custom-built anthropomorphic phantoms, a pediatric 9-month-old male phantom and an adult male phantom, shown in Figs. 1(a) and 1(b), respectively.

Both phantoms were constructed at the University of Florida (UF) with each built as a set of individual 5-mm axial slices. These axial slices were constructed from molds fabricated from cross-sectional images of two computational phantoms using a computerized milling machine.¹⁷ The adult male phantom was built as a replica of the UF computational adult male reference hybrid phantom,^{18,19} and the 9-month-old phantom was built as a replica of the UF Series-B 9-month-old voxel phantom.²⁰ Both phantoms are made with three types of materials developed at UF—soft tissue-equivalent, lung tissue-equivalent, and bone tissue-equivalent substitutes.^{17,21} Each material closely matches the reference densities and x-ray mass-energy absorption coefficients in the diagnostic photon energy range for their respective representative body tissues, as given in Report No. 46 of the International Commission on Radiation Units and Measurements (ICRU).²² Further details on phantom construction are found in Ref. 17.

FIG. 2. (a) Axial view of the 9-month-old male physical phantom with the plastic scintillating dosimeter placed at a selected dose point within the left lung. (b) Corresponding axial view of the 9-month-old male voxel phantom.

Following PSD calibration measurements, the dosimeters were next placed within precut channels at selected dose points of the phantoms. Views of the dosimeters within these channels can be seen in Figs. 2(a) (9-month-old phantom) and 3(a) (adult phantom), respectively. Each point dose was measured three times and averaged in order to yield improved statistics and dose uncertainties. Several point dose measurements were then subsequently averaged to yield an estimate of the volume-averaged organ absorbed dose delivered to the phantom. Depending on their size, organs had anywhere from one to several dose points that were used to calculate an average dose value.

The adult male anthropomorphic phantom was used to take organ dose measurements for six different CT scan protocols: three axial (head, chest, and abdomen/pelvis) and three helical (chest, abdomen, and pelvis). The number of organ doses measured for each scan ranged from four to eight. The pediatric phantom was used for measurements for two full-body scans, one axial and one helical—each with 13 organ point doses measured. For each scan, the phantom was placed in a vacuum-locked immobilization bag and aligned with the CT scanner's axis of rotation. A summary of scan parameters for each CT scan is shown in Table I.

II.E. Computational phantom dose calculations

Organ dose calculations using the Monte Carlo CT source model were performed using the UF Series-B 9-month-old male voxel phantom [Figs. 1(a) and 2(b)] and the UF reference adult male hybrid-voxel phantom [Figs. 1(b) and 2(c)]. Each anthropomorphic phantom was uniquely matched to

FIG. 3. (a) Axial view of the adult male physical phantom with the plastic scintillating dosimeter placed at a selected dose point within the right kidney. (b) Corresponding axial view of the adult male voxel phantom.

TABLE I. Summary of scan parameters used for each CT examination considered in the study.

Scan	Anatomical extent	Beam energy (kVp)	Filter	Beam collimation (mm)	Effective mAs	Pitch
Adult male						
Axial head	Top of head to second cervical vertebra	120	Head	10	100	
Axial chest	Clavicles to dome of diaphragm	120	Body	10	100	
Axial abdomen/pelvis	Dome of diaphragm to femoral heads	120	Body	10	100	
Helical chest	Clavicles to tops of kidneys	120	Body	24	150	0.50
Helical abdomen	Dome of diaphragm to iliac crests	120	Body	24	170	0.75
Helical pelvis	Iliac crest to lesser trochanters	120	Body	24	160	0.75
9-month-old						
Axial full-body	Top of head to mid thigh	120	Body	24	100	
Helical full-body	Top of head to mid thigh	120	Body	24	100	1.40

their corresponding computational twin in terms of body contour, skeletal anatomy, and lung anatomy. Furthermore, the boundaries of all soft tissue organs could be traced on each physical slice in a manner consistent with their definitions in the corresponding computational phantom, thus permitting confidence in the location and interpretation of point dose estimates.

The arms of the computational phantoms were removed to reflect their absence in the physical phantoms during CT measurements. The computational phantoms were voxelized to resolutions of $2 \times 2 \times 2 \text{ mm}^3$ for the adult male and $0.86 \times 0.86 \times 3 \text{ mm}^3$ for the 9-month-old for input into MCNPX v2.6 for simulation. Within the Monte Carlo simulation setups, the voxelized phantoms were placed upon a model of the CT scanner's carbon-fiber patient table. The table was modeled as two concentric cylinders of different radii that were truncated to a width of 40 cm corresponding to the width of the patient table. The source model was next established to reflect the parameters of each physical scan. For all helical scans, a total of eight starting angles (0° , 45° , 90° , 135° , 180° , 225° , 270° , and 315°) were modeled. Organ doses were calculated using the F6 kerma approximation tally in MCNPX; therefore, no secondary electrons were transported. Considering the energy range of the CT x-ray spectra and the subsequent ranges of secondary electrons in tissue, the kerma approximation offers an acceptable approximation in the estimate of average organ absorbed dose. The effects of this approximation were investigated in detail by Chao *et al.*,²³ who showed negligible differences between incorporating and omitting secondary electron transport for diagnostic-energy beams. All Monte Carlo simulations were made using 100×10^6 photon histories on a 12-processor Dell Precision T7500 workstation, resulting in tally uncertainties of less than 0.25% for all adult male exams and less than 0.45% for all pediatric male exams.

Since MCNPX provides calculation results in absorbed dose per simulated photon, the number of photons delivered by the scanner per unit mAs, called the Monte Carlo normalization factor, were used to convert relative MCNPX dose tallies to organ doses in absolute units of mGy. The normalization factors were calculated based on the ratio of pencil ion chamber measurements in free-in-air (mGy/mAs) to the MCNPX-simulated free-in-air ion chamber doses

(mGy/photon) given in previous studies.¹² Each unique beam energy, filter, and collimation combination required its own normalization factor. Absolute organ doses were calculated by multiplying the dose in mGy/mAs by the total mAs delivered during the exam.

III. RESULTS

III.A. Comparisons of both simulated and measured organ doses

Tables II and III show measured and simulated organ absorbed doses for the adult male and pediatric CT scans, respectively. For all simulated helical exams, organ dose results for the eight separate starting angle simulations were averaged to provide the simulated organ dose values used for comparison with measured values. Percent differences in the mean organ absorbed dose given by the MC simulations and the PSD dosimetry were calculated as $[100\% \times (D_{MC} - D_{PSD}) / D_{PSD}]$. Additionally, the calculated uncertainties for both the measured doses and the percent differences can also be found on Tables II and III.

For the adult phantom data given in Table II, percent differences were within $\pm 5\%$ for 5 of 16 organs, within $\pm 10\%$ for 11 of 16 organs, and within $\pm 15\%$ for 15 of 16 organs for the axial scans. For helical scans of the adult phantom, these percent differences were within $\pm 5\%$ for 3 of 15 organs, within $\pm 10\%$ for 11 of 15 organs, and within $\pm 15\%$ for 14 of 15 organs.

As shown in Table III, percent differences in organ dosimetry for axial scans of the pediatric 9-month-old phantom were within $\pm 5\%$ for 9 of 13 organs, within $\pm 10\%$ for 11 of 13 organs, and under 15% for all organs considered. For the helical scans of the 9-month-old phantom, percent differences in organ dosimetry were within $\pm 5\%$ for 4 of 13 organs, within $\pm 10\%$ for 11 of 13 organs, and under 15% for all organs considered.

Table IV summarizes the overall percent differences over all exams, as well as for previous CTDI₁₀₀ validation studies. For the adult male, the absolute percent differences were within -16 to $+15\%$ for axial scans and within -18 to $+14\%$ for helical scans. The average percent difference for all organs considered was 7.8% and 9.1% for axial and helical scans,

TABLE II. Simulated and measured organ doses for CT examinations of the adult male phantom.

Scan	Organ	Measurement points	Simulated dose ^a (mGy)	Measured dose (mGy)	Percent difference
Axial head	Brain	4	13.8	14.2 ± 4.8%	− 3.1 ± 4.6
	Eyeballs	2	17.4	19.0 ± 6.8%	− 8.3 ± 6.2
	Salivary glands	6	16.2	17.2 ± 3.9%	− 6.1 ± 3.7
	Thyroid	1	2.4	2.5 ± 9.5%	− 4.7 ± 9.1
Axial chest	Thyroid	1	7.7	8.7 ± 9.5%	− 11.0 ± 8.5
	Esophagus	6	5.5	5.6 ± 4.2%	− 2.6 ± 4.0
	Lungs	6	7.1	8.4 ± 4.0%	− 15.7 ± 3.4
	Heart	3	7.3	8.0 ± 5.6%	− 8.4 ± 5.1
Axial abdomen/pelvis	Stomach	4	4.2	4.7 ± 5.6%	− 10.9 ± 5.0
	Liver	4	3.6	3.4 ± 6.1%	5.0 ± 6.4
	Kidneys	2	6.9	6.9 ± 6.9%	− 0.2 ± 6.9
	Small intestines	6	6.3	7.3 ± 3.9%	− 13.7 ± 3.4
	Colon	8	7.6	8.1 ± 4.3%	− 5.8 ± 4.0
	Urinary bladder	1	5.5	5.9 ± 9.5%	− 7.3 ± 8.8
	Prostate	1	3.2	2.8 ± 9.6%	14.7 ± 11.0
	Testes	2	0.8	0.7 ± 8.4%	7.8 ± 9.1
Helical chest	Thyroid	1	16.1	15.2 ± 12.0%	6.2 ± 12.8
	Esophagus	6	10.7	11.7 ± 4.3%	− 8.9 ± 3.9
	Lung	6	12.8	13.4 ± 4.0%	− 4.7 ± 3.8
	Stomach	4	11.2	12.0 ± 6.0%	− 6.8 ± 5.6
	Liver	4	11.1	12.0 ± 5.2%	− 7.1 ± 4.8
Helical abdomen	Liver	4	12.8	14.2 ± 5.5%	− 9.8 ± 5.0
	Stomach	4	13.2	13.1 ± 5.1%	1.0 ± 5.1
	Kidneys	2	13.5	11.9 ± 6.8%	13.4 ± 7.7
	Small intestine	6	8.5	9.3 ± 4.2%	− 8.9 ± 3.8
	Colon	8	12.8	11.8 ± 3.9%	− 8.2 ± 4.2
Helical pelvis	Small intestine	6	7.3	8.6 ± 4.6%	− 15.0 ± 3.9
	Colon	8	5.2	6.1 ± 4.9%	− 14.3 ± 4.2
	Urinary bladder	1	11.7	11.3 ± 9.5%	3.7 ± 9.9
	Prostate	1	11.9	13.3 ± 9.5%	− 10.7 ± 8.5
	Testes	2	18.3	22.2 ± 7.7%	− 17.7 ± 6.4

^aAll simulated doses have Monte Carlo errors less than 0.25%.

respectively. For the 9-month-old phantom, the absolute percent differences were within −13 to +15% for axial scans and within −15 to +10% for helical scans. The average percent difference for all organ considered was 5.5% and 7.0% for axial and helical scans, respectively. For the head CTDI₁₀₀ measurements, percent differences between measured and simulated values were between −7.2 and +0.1%, with an average difference of 3.5%. For the body CTDI₁₀₀ measurements, percent differences ranged between −8.6 and +1.1%, with an average difference of 3.9%.

III.B. Variations in organ dose estimates due to starting angle

As an additional component to this study, the individual simulated organ doses for the eight chosen starting angles were compared to their corresponding average values. The percent difference results for the adult male helical exams and the 9-month-old helical exam are given in Tables V and VI, respectively. As shown in Table VII, the range of percent differences in organ doses [defined as $100\% \times (D_{\text{starting angle}} - \bar{D}_{\text{starting angle}}) / \bar{D}_{\text{starting angle}}$] for helical scans of the adult male phantom ranged from −19.8% to

+33.6% (average percent difference of 10.4%). The corresponding percent differences for helical scans of the 9-month-old phantom were substantially less and ranged from only −4.5% to +4.2% (average percent difference of only 1.4%).

IV. DISCUSSION

IV.A. Percent differences of inphantom doses and CTDI₁₀₀ measurements

Analyzing the inphantom percent difference summaries yields two observations about the data: (1) the helical scans had larger average percent differences than their respective axial scans, and (2) the adult male scans had larger differences than the 9-month-old scans. The first observation is most likely explained by the fact that helical scans introduce over-ranging and starting angle effects, which in turn introduce uncertainties in simulating the scans due to incomplete knowledge of the exact dosimetric effects of both. The second observation can be explained by two possible reasons. First, the scans performed on the adult male had specific anatomical landmarks governing the scan length, whereas the 9-month-old scans were full-body. Small placement errors can occur

TABLE III. Simulated and measured organ doses for CT examinations of the 9-month-old phantom.

Scan	Organ	Measurement points	Simulated dose ^a (mGy)	Measured dose (mGy)	Percent difference
Axial full-body	Brain	2	11.6	12.0 ± 5.7%	− 3.3 ± 5.5
	Thyroid	1	16.2	16.5 ± 8.0%	− 1.9 ± 7.9
	Thymus	1	13.9	13.5 ± 8.0%	2.8 ± 8.2
	Lungs	4	14.3	14.7 ± 4.0%	− 3.3 ± 3.9
	Heart	2	14.7	14.8 ± 5.7%	− 0.5 ± 5.7
	Stomach	2	13.4	14.3 ± 8.0%	− 6.2 ± 7.5
	Liver	4	13.6	14.1 ± 8.1%	− 3.2 ± 7.8
	Gall bladder	1	13.6	13.2 ± 8.0%	2.7 ± 8.2
	Pancreas	1	13.0	14.9 ± 8.0%	− 13.0 ± 7.0
	Adrenals	2	12.1	13.2 ± 8.0%	− 8.9 ± 7.3
	Kidneys	2	13.3	13.2 ± 8.0%	0.5 ± 8.0
	Urinary bladder	1	14.2	15.8 ± 8.0%	− 10.2 ± 7.2
	Testes	2	14.1	12.3 ± 5.7%	14.5 ± 6.5
Helical full-body	Brain	2	11.5	11.9 ± 9.7%	− 3.6 ± 9.4
	Thyroid	1	15.5	16.8 ± 9.5%	− 7.9 ± 8.8
	Thymus	1	13.4	13.8 ± 9.8%	− 2.9 ± 9.5
	Lungs	4	14.0	14.9 ± 5.2%	− 5.9 ± 4.9
	Heart	2	14.4	15.0 ± 7.1%	− 4.1 ± 6.8
	Stomach	2	13.7	13.8 ± 11.3%	− 0.6 ± 11.2
	Liver	4	13.2	14.0 ± 13.3%	− 5.6 ± 12.6
	Gall bladder	1	13.0	14.2 ± 9.6%	− 8.6 ± 8.7
	Pancreas	1	12.9	14.9 ± 9.6%	− 13.3 ± 8.3
	Adrenals	2	12.2	14.2 ± 9.6%	− 14.3 ± 8.2
	Kidneys	2	13.3	14.2 ± 9.6%	− 6.5 ± 8.9
	Urinary bladder	1	13.7	14.8 ± 9.9%	− 7.7 ± 9.1
	Testes	2	13.4	12.2 ± 7.9%	9.8 ± 8.7

^aAll simulated doses have Monte Carlo errors less than 0.45%.TABLE IV. Summary of percent differences between simulated and measured organ doses, as well as for previous validation of CTDI₁₀₀ measurements.

	CTDI Phantoms		9-month-old phantom		Adult male phantom	
	Head CTDI ₁₀₀	Body CTDI ₁₀₀	Axial scans	Helical scans	Axial scans	Helical scans
Range	(−7.2%, 0.1%)	(−8.6%, 1.1%)	(−13.0%, 14.5%)	(−14.3%, 9.8%)	(−15.7%, 14.7%)	(−17.7%, 13.4%)
Average magnitude	3.5%	3.9%	5.5%	7.0%	7.8%	9.1%

TABLE V. Percent differences for individual simulated organ doses in the adult male for each beam starting angle as compared to the corresponding averaged value.

Scan	Organ	Beam starting angle							
		0°	45°	90°	135°	180°	225°	270°	315°
Helical chest	Thyroid	− 15.2	− 11.8	14.6	33.6	18.6	− 10.2	− 15.7	− 13.8
	Esophagus	− 0.5	− 3.0	− 0.9	7.7	1.6	− 3.9	0.5	− 1.5
	Lung	0.3	− 2.2	− 1.2	3.2	1.7	− 0.6	0.3	− 1.6
	Stomach	− 7.7	8.7	20.8	11.9	− 9.5	− 18.2	− 11.6	5.7
	Liver	− 4.7	− 14.0	− 7.2	11.0	17.3	12.8	− 2.0	− 13.2
Helical abdomen	Stomach	13.0	− 18.7	− 8.5	11.8	24.4	11.3	− 13.4	− 19.8
	Liver	− 13.5	8.7	− 8.5	− 17.4	− 8.6	11.9	16.2	11.3
	Kidneys	− 9.7	13.5	20.6	4.3	− 12.3	− 14.0	− 11.9	9.4
	Small intestine	3.8	− 7.0	− 11.1	− 10.2	6.4	17.5	6.3	− 5.7
	Colon	8.0	− 11.2	− 18.5	− 8.9	10.8	17.4	10.0	− 7.7
Helical pelvis	Small intestine	11.2	11.3	− 5.4	− 13.4	− 15.4	− 9.9	8.4	13.1
	Colon	8.3	13.2	1.8	− 15.4	− 17.7	− 8.8	5.5	13.2
	Urinary bladder	9.5	9.0	− 10.3	− 8.0	− 7.7	− 10.9	6.1	12.3
	Prostate	7.4	8.1	− 12.0	− 4.4	− 1.3	− 11.8	3.0	11.1
	Testes	15.8	15.6	− 9.3	− 19.8	− 17.3	− 16.0	14.6	16.3

TABLE VI. Percent differences for individual simulated organ doses in the 9-month-old male for each beam starting angle as compared to the corresponding averaged value.

Organ	Beam starting angle							
	0°	45°	90°	135°	180°	225°	270°	315°
Brain	−0.6	−0.2	0.3	0.7	0.3	0.4	−0.4	−0.6
Thyroid	2.3	1.2	−0.5	−0.7	−2.1	−0.4	−1.4	1.6
Thymus	0.9	1.2	−1.5	−1.0	−1.1	−1.3	1.5	1.4
Lungs	0.8	0.0	−1.1	0.6	−0.1	−0.5	0.1	0.3
Heart	1.5	0.5	−2.0	−1.2	−1.0	−0.3	1.4	1.1
Stomach	2.3	0.5	−2.9	−2.6	−1.8	0.9	2.4	1.1
Liver	0.0	1.2	1.9	0.4	−1.6	−3.2	−0.3	1.5
Gall bladder	2.6	2.4	0.4	−3.8	−4.5	−1.4	1.6	2.7
Pancreas	1.2	0.1	−3.6	−0.6	1.0	0.2	1.8	0.0
Adrenals	0.0	−1.0	−0.4	4.2	0.5	−0.8	−1.0	−1.3
Kidneys	−0.7	−0.8	−1.4	3.3	1.7	−1.1	−0.5	−0.5
Urinary bladder	1.9	2.6	−1.1	−2.2	−3.0	−2.4	1.6	2.6
Testes	2.7	2.4	−3.8	−1.5	−3.7	−1.1	2.6	2.3

when trying to exactly replicate the anatomical location of the scan length of the physical exam onto the simulated phantom, which can result in larger organ dose discrepancies, especially with organs located at the edges of the scan length. Since the 9-month-old scans covered the entire body, such discrepancies caused by scan length placement would be negligible as compared to the adult male scans. The second reason for larger differences in the adult male phantom is the fact that the adult male phantom is much wider and larger than the 9-month-old. Since the 9-month-old is so narrow, the effects of any discrepancies in the lateral beam profile during simulation will be much less than that of the adult male, as the 9-month-old would be situated mostly within the central axis of the beam (where the beam shape does not vary substantially). The adult male sits across an area where the beam shape changes significantly, thus any discrepancies in the physical and simulate beam shapes would have a larger impact on organ doses than on the 9-month-old.

When comparing the percent differences of the inphantom doses to those of the CTDI₁₀₀ measurements of the previous validation study, it is clear that the inphantom measurements yield larger disagreements; both in the range of the percent difference values as well as the average magnitude of those differences. These discrepancies are significant especially for the adult male, with ranges and average differences being more than twice those of the CTDI measurements. Not surprisingly, this increase in average percent differences seemed

TABLE VII. Summary of percent differences between simulated and measured organ doses in the starting angle study.

	9-month-old phantom	Adult male phantom
	Helical scans	Helical scans
Range	(−4.5%, 4.2%)	(−19.8%, 33.6%)
Average magnitude	1.4%	10.4%

to follow with the increasing complexity and heterogeneity of the phantoms employed.

IV.B. Uncertainty in organ dose estimates due to starting angle

As evidenced by the results of the starting angle study, there are significant differences between organ doses of specific starting angles and organ doses averaged over all angles. The greatest differences tend to correspond to the more superficially located organs such as the thyroid, kidneys, adrenals, and testes; a trend also seen in a study by Zhang *et al.*²⁴ It should be noted that the percent differences for the 9-month-old male were far less dramatic than that of the adult male, owing to the fact that the dimensions of this patient phantom are small enough such that dose gradients between superficially and deeply located organs are smaller than those found in the reference adult male. However, both datasets reaffirm the fact that beam starting angle variations can have substantial influences on organ doses delivered in helical CT examinations.

IV.C. Percent differences of inphantom doses in similar studies

The inphantom organ dose percent difference results (including both the ranges and average magnitudes for each scan) are very comparable to the study on GE CT scanners by Li *et al.*,¹¹ in which the authors detail the first set of inphantom validation measurements for helical CT scans. As mentioned in that study and another by Deak *et al.*,⁵ percent differences within 20% are considered acceptable matches due to the overall complexity of CT examinations and the large number of factors that influence both the measurement and simulation results. However, in our study the majority of percent differences for organ doses fell within $\pm 10\%$ for all scans, and the average magnitudes were less than half of the 20% quoted in the study by Li *et al.*¹¹ Based on these results, the validation of the source term for more complicated scans and phantom geometries could be considered seen. However, it is necessary to perform a thorough uncertainty analysis on the simulated and measured doses before final conclusions can be drawn.

IV.D. Uncertainties of simulated and measured organ doses

The two largest sources of uncertainty for the organ doses determined from the simulated exams come from the Monte Carlo organ dose tally errors and the lack of knowledge of the starting angles of the helical exams being simulated. As mentioned previously, each simulation involved 100×10^6 particle histories, resulting in tally uncertainties of less than 0.25% for all adult male exams and less than 0.45% for all 9-month-old male exams. These errors are small enough to be considered negligible, especially in comparison to errors associated with the physical dose measurements. Although the starting angle study results do show large organ dose variations, the

method of measuring doses in the phantoms helps to alleviate the potential errors associated with the lack of starting angle information for each scan. Since each point dose is an average of three measurements, and multiple dose points are averaged to arrive at an approximate average organ dose, 3–24 separate scans are used to calculate each organ dose. Each scan has a random starting angle and thus the effects of individual starting angles on dose will tend to average out over the course of the measurements. Resultantly, using the average of the simulated organ doses over eight starting angles as the final simulated helical organ doses is a reasonable approximation. Thus, for the purposes of this uncertainty analysis, the errors on the simulated doses are assumed to be those due to Monte Carlo tally errors.

The uncertainties associated with the measured organ doses are higher and more complex than those of simulated doses. As discussed in Sec. II.C, the calibration of the PSD fibers as well as the characteristics of the fibers themselves are sources of error creating uncertainties ranging from 3.9% to 12.0% for the adult male and 5.2% to 13.3% for the pediatric male organ dose measurements. There were five major sources of error identified: (1) the uncertainty of the ion chamber measurements, (2) the use of acrylic phantoms for calibration instead of soft tissue phantoms, (3) the energy dependence of the PSD fibers, (4) the choice of effective energy at which to calculate tissue-to-air mass energy absorption coefficient ratios so as to convert air kerma to tissue kerma after calibration, and (5) the individual variation of the three repeated measurements per dose point.

Ion chamber measurements used during PSD calibration were subject to a 4% calibration error that was factored into the dose uncertainty using error propagation on the equation to calculate the calibration factor. The uncertainty of the calibration factors associated with using acrylic instead of soft tissue phantoms was addressed in a separate study. Here, CTDI phantoms were created using the soft tissue-equivalent material, after which fibers were calibrated in both the acrylic and soft tissue CTDI phantoms with their calibration factors compared in units of counts per mGy of air kerma. The results showed agreement within 3.5%, and nearly identical depth dependence behavior between center and peripheral calibration factors for both materials. These results were treated as an uncertainty term applied to the calibration factors.

The extent of energy-dependence uncertainty of the PSDs was also addressed in a separate study performed after the dose measurements were completed. For this study, several PSD calibration factors were calculated for different irradiation schemes of the CTDI phantoms. Instead of using only the full helical scans of the phantoms as described in Sec. II.C,

smaller partial scans of the phantoms were performed to simulate “scatter” and “primary” spectra reaching the PSDs. The calibration factors from these partial exams as well as from the full helical exams for both top and center CTDI channels were averaged and the associated standard deviations were calculated. The results found that the values of the averages of all these calibration factors were very close to the values of the average calibration factors used in this study and described in Sec. II.C. Therefore, the standard deviations calculated in this energy-dependence study were assumed to closely match those that would have been associated with the fibers used in the phantom dose measurements. The maximum standard deviation was found to be around 8%, which was conservatively assigned as the energy-dependent uncertainty for every fiber used in this study. This result was consistent with what was previously determined for PSD energy dependence in Ref. 16.

The ratio of the mass energy absorption coefficients used to convert air kerma to tissue kerma (and to therefore approximate tissue absorbed dose) was calculated using representative effective energies based off of the measured half-value layers of the 120 kVp head and body energy spectra. However, since at different depths within the phantoms, the effective energy of the spectra will change, the ratios of the coefficients will vary as well. In order to quantify the potential degree of this change, the calibration scans for the soft tissue CTDI phantoms were simulated in MCNPX, with various points being selected to serve as average energy fluence tally sites. The results of these tallies gave approximate energy spectra present within different points of the phantoms. The half-value layers and associated effective energies of these spectra were then calculated using Spektr.²⁵ The mass energy absorption coefficients ratios were then calculated using this range of effective energies. In the end, all the calculated ratios were within 1.3% of the originally used soft tissue and lung ratios, so this value was taken as the associated uncertainty.

Finally, the uncertainty associated with the reproducibility of repeat point dose measurements was calculated by taking the standard deviation of the three individual measurements that were averaged to report the final dose point measurement at each phantom location. This uncertainty was certainly the most random in that its magnitude for each dose point did not follow any discernible trend.

Using the above information, uncertainties for each individual measurement were calculated, and the total uncertainty for each organ dose was calculated using error propagation. As expected, the more dose points used to calculate an average organ dose resulted in improved statistics and therefore less total uncertainty. These measured dose uncertainties and those from the Monte Carlo simulations were then used to

TABLE VIII. Summary of maximum percent differences between simulated and measured organ doses based on dose uncertainty analysis.

	9-month-old phantom		Adult male phantom	
	Axial scans	Helical scans	Axial scans	Helical scans
Range	(−20.0%, 21.0%)	(−22.5%, 18.5%)	(−19.5%, 25.8%)	(−24.0%, 21.1%)
Average magnitude	12.4%	15.9%	14.0%	15.1%

calculate relative uncertainties in the percent differences between the two sets of organ doses.

IV.E. Impact of uncertainty on validation

In order to evaluate the maximum impact the dose uncertainties could have on the results of the validation study, the percent differences for every organ dose were given their maximum value based on the calculated uncertainties. Table VIII summarizes the ranges and average magnitudes of this “worst case” situation. In this analysis, all but five percent differences remain under 20%; with the five all under 26%. The average magnitudes ranged from 12.4% to 15.9%. These results thus continue to be consistent with agreements seen in previous studies of physical phantom dose validation of Monte Carlo source models for CT scanning.

V. CONCLUSIONS

In this study, a Monte Carlo-based energy and geometric source model for the Siemens SOMATOM Sensation 16 CT scanner was further validated for helical and axial scanning using measured point estimates of organ dose in two custom-built physical phantoms representing the ICRP reference adult male and a 9-month-old child. Point measurements of organ dose were then averaged and compared to Monte Carlo simulations of average organ doses for a variety of axial and helical computed tomography scans. On average, organ doses from the Monte Carlo simulations were shown to predict those physically measured to within 8%–9% for axial and helical imaging of the reference adult male, and to within 6%–7% for the 9-month-old child. Individual organ doses were generally found to be within 15% of measurements of organ dose for both phantoms. The influence CT x-ray tube starting angle was found to be relatively insignificant for organ dosimetry of the 9-month-old phantom, but to have a substantial influence on organ dosimetry in adult patients (10% dose uncertainty on average with ranges of between –20% and +30%). An extensive uncertainty analysis was also performed in order to quantify the errors associated with both simulated and measured organ doses. The results of this analysis showed that even in the statistically worst-case scenario involving differences between measured and simulated doses, the validation study could be considered successful. Future work will include a thorough investigation into creating correction factors to lessen the uncertainties associated with PSD measurements, expanding the source model library to include different CT manufactures and scanner models, as well as the incorporation of algorithms for tube current modulation. The availability of a library of physical phantoms that exactly match the body shape and internal organ anatomy of the UF series of reference hybrid phantoms will be a significant asset in validating these models and algorithms.

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