Physical validation of a Monte Carlo-based, phantom-derived approach to computed tomography organ dosimetry under tube current modulation

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Purpose: To physically validate the accuracy of a Monte Carlo-based, phantom-derived methodology for computed tomography (CT) dosimetry that utilizes organ doses from precomputed axial scans and that accounts for tube current modulation (TCM).

Methods: The output of a Toshiba Aquilion ONE CT scanner was modeled, based on physical measurement, in the Monte Carlo radiation transport code MCNPX (v2.70). CT examinations were taken of two anthropomorphic phantoms representing pediatric and adult patients (15-yr-old female and adult male) at various energies, in which physical organ dose measurements were made using optically stimulated luminescence dosimeters (OSLDs). These exams (chest–abdomen–pelvis) were modeled using organ dose data obtained from the computationally equivalent phantom of each anthropomorphic phantom. TCM was accounted for by weighting all organ dose contributions by both the relative attenuation of the phantom and the image-derived mA value (from the DICOM header) at the same z-extent (cranial–caudal direction) of the axial dose data.

Results: The root mean squares of percent difference in organ dose when comparing the physical organ dose measurements to the computational estimates were 21.2, 12.1, and 15.1% for the uniform (no attenuation weighting), weighted (computationally derived), and image-based methodologies, respectively.

Conclusions: Overall, these data suggest that the Monte Carlo-based dosimetry presented in this work is viable for CT dosimetry. Additionally, for CT exams with TCM, local attenuation weighting of organ dose contributions from precomputed axial dosimetry libraries increases organ dose accuracy. © 2017 American Association of Physicists in Medicine [https://doi.org/10.1002/mp.12461]

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

1. INTRODUCTION

In recent years, the rapid growth of technology utilized in computed tomography (CT) imaging has led to an increase in the complexity of properly performing accurate computational CT patient dosimetry. It is important to model all aspects of the scanner affecting patient dose including proprietary techniques such as tube current modulation (TCM), which is difficult to model with Monte Carlo methods.^{1–5} Previous studies which outline the concern of CT imaging, especially to that of pediatric patients, did not consider TCM in their dosimetry calculations.^{6–8} Additionally, the CT dose index (CTDI), which is a patient exposure metric required to be reported by the Joint Commission has been shown to be unreliable in the presence of TCM.^{9–11} For these reasons, the present study was undertaken to create a methodology that would enhance precomputed axial dosimetry libraries (similar to ImPACT's CTDosimetry and CT-Expo) by accounting for TCM.^{12,13} By weighting axial dose contributions

uniformly, based on computational phantom attenuation, and directly from CT image data, organ dose estimates were compared to measured organ doses within anthropomorphic phantoms undergoing torso exams (with TCM) following common CT protocols.¹⁴

2. MATERIALS AND METHODS

2.A. CT scanner and exam protocols

The work in this study was performed on a Toshiba Aquilion ONE CT scanner (Toshiba America Medical Systems Inc. Tustin, CA, USA) using the vendor's TCM technique (^{SURE}Exposure 3D). For chest–abdomen–pelvis (CAP) exams, the scanner was operated in 64-slice mode with the TCM technique employed over a range of tube potentials (100, 120, and 135 kVp), and a constant rotation time and helical pitch of 0.5 s and 0.828, respectively. These techniques were derived from common protocols clinically used

TABLE I. Summary of CT exam parameters for chest–abdomen–pelvis exams. UF15F and UFADM correspond to the University of Florida (UF) 15-yr female and adult male physical and computational phantoms.

Parameter	UF15F	UFADM
Tube current modulation	Yes	Yes
Collimation	0.5 mm × 64	0.5 mm × 64
Energy (kVp)	120, 135	100, 120, 135
Exam start	Thoracic inlet	Thoracic inlet
Exam end	Lesser trochanter	Lesser trochanter
Filter	Large	Large
Gantry tilt (°)	0	0
Average mA	140 ^a , 120 ^a	265 ^a , 140 ^a , 110 ^a
mA (min, max)	(100, 500)	(100, 500)
Target noise index (SD)	12.5	12.5
Pitch	0.828	0.828
Rotation time (s)	0.5	0.5

^aExam mA is variable due to tube current modulation, reported value is average mA from the CT image-set.

in the UF Health system and are fully summarized in Table I.

The Toshiba Aquilion ONE scanner includes Toshiba's proprietary TCM technique ^{SURE}Exposure 3D. Before an exam, the patient is scanned with two perpendicular “scanograms” with anterior–posterior (AP) and lateral projections.¹⁵ Using this information, the water equivalent thickness of each anatomical region is derived and used as a basis for modulation of the scanner's tube current during the ensuing exam. Additionally, a user inputted target standard deviation in Hounsfield units (HU) is defined and incorporated into the modulation

(Toshiba's standard of 12.5 HU was used for this study). This technique ensures that the beam is modulated in both the patient's z-axis (cranial–caudal direction) as well as within their xy-plane. This combination of z-axis modulation (longitudinal) and xy-plane modulation (angular) is a widely used technique for dose reduction in CT examinations.¹⁶ This technique was used during all torso exams, per common clinical practice.

2.B. Modeling of CT scanner and dosimetry methodology

Computational dosimetry for this project was performed using the Monte Carlo radiation transport code MCNPX (Monte Carlo N-Particle eXtended) v2.70 developed by Los Alamos National Laboratory. The Toshiba Aquilion ONE scanner was modeled by physical measurements acquired from the scanner and mathematical concepts presented by Turner et al. that properly account for all major components of a CT exam including beam energy spectra, beam shape, and output.^{17–19} To further validate the MCNPX custom source subroutine for this scanner, CTDI measurements where compared to simulated CTDI values. Using a typical CTDI body phantom (32 cm), measurements (average of three) were taken at each of the four periphery dosimeter locations (3 o'clock, 6 o'clock, 9 o'clock, and 12 o'clock) and the center dosimeter location for each energy, filter, and relevant collimation setting (32 and 16 mm). The CTDI phantom was then constructed and placed in the MCNPX geometry, centered at the source isocenter. An equivalent set of measured air kerma values where calculated and compared. Table II shows measured, simulated, and percent

TABLE II. Summary of toshiba aquilion ONE CTDI validation study.

Energy (kVp)	Filter	Collimation (mm)	Measured ^a CTDI ₁₀₀ air kerma (mGy) for 100 mAs/rotation			Simulated CTDI ₁₀₀ air kerma (mGy) for 100 mAs/rotation			Percent difference ^b		
			Center	Periphery	CTDI _w ^c	Center	Periphery	CTDI _w ^c	Center	Periphery	CTDI _w ^c
80	M	16	1.53	3.50	2.84	1.61	3.64	2.96	5.32	3.76	4.04
		32	1.35	3.09	2.51	1.41	3.20	2.61	4.73	3.58	3.79
	L	16	1.57	3.97	3.17	1.65	4.03	3.24	4.76	1.54	2.07
		32	1.39	3.50	2.80	1.45	3.56	2.86	4.14	1.66	2.07
100	M	16	3.40	6.85	5.70	3.46	6.97	5.80	1.75	1.78	1.77
		32	2.99	6.02	5.01	3.03	6.14	5.10	1.21	1.94	1.79
	L	16	3.54	7.83	6.40	3.55	7.79	6.38	0.45	−0.40	−0.24
		32	3.11	6.88	5.62	3.11	6.87	5.62	0.02	−0.09	−0.07
120	M	16	6.03	11.23	9.50	6.15	11.36	9.62	2.06	1.12	1.32
		32	5.24	9.79	8.28	5.37	9.94	8.42	2.41	1.53	1.71
	L	16	6.23	12.88	10.66	6.33	12.76	10.61	1.55	−0.91	−0.43
		32	5.44	11.25	9.31	5.51	11.14	9.27	1.38	−0.93	−0.48
135	M	16	8.49	15.28	13.02	8.63	15.26	13.05	1.69	−0.13	0.26
		32	7.32	13.20	11.24	7.52	13.32	11.39	2.74	0.93	1.32
	L	16	8.81	17.58	14.65	8.96	17.33	14.54	1.79	−1.42	−0.77
		32	7.57	15.17	12.64	7.74	14.99	12.57	2.14	−1.22	−0.55

^aAverage of three consecutive measurements in 100 mm ion chamber.

^bCalculated as $100 \times (\text{Simulated Air Kerma} - \text{Measured Air Kerma}) / \text{Measured Air Kerma}$.

^cCalculated as $[(1/3) \times \text{CTDI}_{100, \text{center}} + (2/3) \times \text{CTDI}_{100, \text{peripheral}}]$.

difference for all physical measurements (peripheral data are combined). The average magnitude of the percent difference for the center hole, average of periphery holes, and calculated CTDI_w were 2.38, 1.43, and 1.42 percent, respectively. The median percent difference for the center hole, average of periphery holes, and calculated CTDI_w were 1.92, 1.03, and 1.32 percent, respectively.

CT exams are modeled as the sum of a series of axial rotations performed throughout a computational phantom, which is the same approach used by the IMPACT group.¹³ This “axial summing” method allows for organ doses from multiple exam locations and lengths to be derived from one set of data per computational phantom. For each of the z-locations, an axial exam is simulated, and organ dose contributions to all major organs (in-field or out-of-field) are calculated.

In addition to knowing organ dose contributions as a function of z-location, TCM was modeled by weighting each dose contribution by the normalized relative attenuation of the computational phantom at the same z-location. The mathematical framework for this attenuation weighting factor was derived based on previous findings by Gies et al. and Schlattl et al.^{4,20} Gies et al. mathematically proved through a simplified problem setup that in order to have a tube current modulation scheme that will provide the lowest uniform theoretical image noise, one must weight the beam projections over the course of an exam relative to one another based upon the inverse square root of the central ray attenuation of each projection (where attenuation is defined as the initial intensity of the central ray divided by the exit intensity from the patient for that projection).²⁰ Schlattl et al. applied this conclusion by explicitly weighing individual projections in full Monte Carlo radiation transport simulations of helical exams.²¹ These results were used in the present study to derive a custom “generic” approach to calculating the relative attenuation value (AV_{average}) of each z-location throughout the computational phantom. This attenuation value takes into account the average of eight equally spaced projections (0° , 45° , 90° , 135° , 180° , 225° , 270° , and 315°) about each z-location of the phantom. With the CT beam “fixed” at each angle, a value of air kerma was calculated computationally on the opposite side of the phantom for each of the eight projections. Previous work in our laboratory determined that a 90° arc detector with a 40 cm radius and 1 cm z-thickness provided the highest level of accuracy when comparing predicted relative attenuation to measured relative attenuation of cadaveric specimens being examined using Toshiba’s ^{SURE}Exposure 3D.²¹ The derivation of each AV_{average} is shown in Eqs. (1) and (2).

$$kerma_{\theta,z} = \text{Simulated Energy Deposition} \left(\frac{\text{MeV}}{g} \right) \quad (1)$$

$$AV_{\text{average}}(z) = \frac{\sum_{\theta_1}^{\theta_8} (kerma_{\theta,z})^{-1/2}}{8 \text{ projection angles}} \quad (2)$$

One disadvantage of this methodology is that organ dose contributions are weighted as a function of z, so the angular (or xy-plane) attenuation is not explicitly modeled into the dosimetry as is performed in other studies.^{1–3,22,23} However, to show that angular information is still implicitly incorporated into the attenuation value, a sensitivity study was performed to calculate the AV_{average} of various computational stylized phantoms with equivalent area in the xy-plane as shown in Fig. 1. Each stylized phantom consists of four tissue types: soft tissue, lung tissue, bone tissue, and air (in-body), with material compositions and densities consistent with ICRP (International Commission on Radiological Protection) Publication 89 on basic anatomical and physiological data.²⁴ Tested in this study were outer body shape (row one), internal organ shape (row two), and internal organ size (row three). As seen in Fig. 1, varying axial extents of both the outer body shape and internal organ shape and size produces different AV_{average} values (numbers below each image). As a result, even though this methodology does not explicitly weight projections in the axial plane, it does take into account axial attenuation information.

One advantage of this method is that both primary and scatter contributions to each organ’s absorbed dose are weighted by the local attenuation of the beam location and not the local attenuation at the organ. In the results reported here, the following terminology is adopted. Organ doses that do not include local attenuation weighting are considered as part of “uniform” dosimetry. Organ doses that are weighted based on the computational phantom’s attenuation are considered “weighted” dosimetry. Finally, organ doses that are

FIG. 1. Stylized computational phantoms with equivalent cross-sectional areas and below their normalized AV_{average} values. (Row 1, varying outer body shape with equivalent area. Row 2, varying inner organ shape with equivalent area. Row 3, varying inner organ size). [Color figure can be viewed at wileyonlinelibrary.com]

weighted based on the scanner reported mA values are considered “image-based” dosimetry.

2.C. Anthropomorphic phantom dose measurements

Previous work by Winslow et al. produced an anthropomorphic physical phantom which represented an adult male (UFADM) per ICRP Publication 89 specifications for height (176 cm) and weight (73 kg).^{14,24} This work was expanded to create a physical size-equivalent 15-yr-old female (UF15F) anthropomorphic phantom (161 cm and 53 kg). Each physical phantom is modeled directly after a corresponding computational phantom that was designed within ICRP Publication 89 specifications. Each physical phantom consists of four tissue types: soft tissue-equivalent substitute (STES), bone tissue-equivalent substitute (BTES), lung tissue-equivalent substitute (LTES), and air spaces.¹⁴ It should be noted that these physical phantoms partially include the phantom’s arm as shown in Fig. 2 with the UF15F phantom placed on the CT table.

Each of the phantom’s critical organs was outlined on the physical phantom’s axial surfaces based on the corresponding axial images of the computationally equivalent phantom. Using these organ reference points, dosimeter locations were created throughout the z-extent of the physical phantom. Within each slit, a NanoDotTM dosimeter (Landauer Inc., Glenwood, IL, USA) of volume $10 \times 10 \times 2 \text{ mm}^3$ was inserted before the phantom was reassembled as shown in Fig. 2. Each optically stimulated luminescence dosimeter (OSLD) was previously calibrated by a 0.6 cm^3 ionization chamber within a CTDI body phantom for every energy and depth that occurs within the actual physical phantom.²⁵ This correction factor for energy and scatter ($C_{E,S}$) was applied to each organ along with a ratio of mass energy absorption coefficients for tissue to that of air (f). The values of f (1.06 for 100 kVp, 1.06 for 120 kVp, and 1.07 for 135 kVp) were taken from

previous dissertation work by Jones, Lavoie, and American Association of Physicist in Medicine (AAPM) Report No. 96.^{26–28} The values of $C_{E,S}$ were 0.95, 0.90, and 0.85 for 100, 120, and 135 kVp, respectively. Each dosimeter was measured without an exam for the purposes of knowing background exposure ($D_{\text{background}}$). It should be noted any dosimeter surpassing 1 mGy was excluded from this study. Then, dose readings for each of the clinically relevant phantom-energy combinations (listed in Table I) were measured (D_{reading}) for two consecutive exams (for consistency). Final dose recordings (D_{tissue}) for each dosimeter were computed as:

$$D_{\text{raw}} = \frac{D_{\text{reading}} - D_{\text{background}}}{2} \quad (3)$$

$$D_{\text{tissue}} = \frac{D_{\text{raw}}}{C_{E,S}} \times f \quad (4)$$

Once the dose to each OSLD had been properly measured, the absorbed dose to each critical organ was calculated by averaging the dose to all dosimeters placed within each organ’s presegmented volume within the phantom. This approach is equivalent to that used in Griglock et al. for CT dosimetry in postmortem subjects using NanoDotTM dosimeters in the Toshiba Aquilion ONE Scanner.²⁹

Table III shows a breakdown of individual organs as a function of point doses for the adult male phantom. For this particular phantom, the 15 organ doses measured were derived from a total of 55 point doses: 6 for lung, 3 for stomach, 4 for liver, 6 for esophagus, 4 for kidney, 13 for colon, 10 for small intestine, 2 for gonads, and 1 for the remaining organs. The 15-yr-old female phantom consisted of 14 organs derived from 47 point doses. For the purposes of this study, averaging distributed point doses is used as a surrogate for true organ dose, which is infeasible to measure based on its physical definition. It is believed that the number of dosimeters used per organ based on its axial and z-extent justifies using averaged point doses to estimate the true organ dose. It should also be noted that previous work has shown minimal angular dependence for these particular OSL dosimeters when used for in-phantom CT dosimetry, especially at deeper depths in tissue.^{23,25,30–32}

2.D. Computational dosimetry calculations of organ dose

After the completion of the physical measurements, each exam was modeled using the previously described dosimetry methodology. As each physical phantom was explicitly constructed to be an equivalent size of a corresponding computational phantom, no organ segmentation was required. It should be noted that the computational phantoms were adjusted to include the partial arms present in the anthropomorphic physical phantoms. Each NURBS (nonuniform rational basis spline) model was voxelized using an in-house script written in MATLAB (The Mathworks Inc., Natick,

FIG. 2. The 15-yr-old female anthropomorphic phantom assembled on the CT scanner bed. [Color figure can be viewed at wileyonlinelibrary.com]

TABLE III. Comparison of organ doses and point doses for the adult male phantom at 120 kVp.

Organ	Position ^a	Dose (mGy)	Organ dose (mGy)	Organ	Position ^a	Dose (mGy)	Organ dose (mGy)
Thyroid	Center-middle	12.3	12.3	Pancreas	Center-middle	10.1	10.1
Lung	Right-superior	9.9	11.1	Colon	Right-superior	13.5	11.6
	Left-superior	9.8			Left-superior	12.0	
	Right-middle	10.4			Right-middle-superior	15.2	
	Left-middle	10.5			Left-middle-superior	11.2	
	Right-inferior	13.0			Center-middle-superior	14.0	
	Left-inferior	12.8			Right-middle	14.4	
					Left-middle	11.7	
Thymus	Center-middle	11.1	11.1		Right-middle-inferior	13.9	
					Left-middle-inferior	10.4	
Stomach	Center-anterior	13.1	11.8		Left-inferior	9.7	
	Center-posterior	11.4			Left-inferior	9.9	
	Center-inferior	10.8			Left-inferior	7.2	
					Center-inferior	7.9	
Liver	Right-superior	12.4	12.0	Small Intestine	Right-superior	9.6	11.1
	Left-superior	11.5					
	Center-middle	12.0			Left-superior	12.1	
	Center-inferior	12.0			Right-middle-superior	10.7	
					Left-middle-superior	11.7	
Gallbladder	Center-middle	10.6	10.6		Right-middle	10.1	
					Left-middle	13.0	
Esophagus	Center-superior	9.5	10.0		Right-middle-inferior	11.8	
	Center-middle-superior	10.3			Left-middle-inferior	10.0	
	Center-middle	10.1			Right-inferior	11.1	
	Center-middle	9.8			Left-inferior	10.9	
	Center-middle-inferior	10.1		Bladder	Center-middle	8.1	8.1
	Center-inferior	10.0					
Spleen	Center-middle	11.6	11.6	Prostate	Center-middle	7.3	7.3
Kidney	Right-superior	10.3	10.3	Gonads	Right-middle	11.3	10.8
	Left-superior	10.3			Left-middle	10.2	
	Right-inferior	10.6					
	Left-inferior	9.9					

^aCenter refers to lateral direction and middle refers to inferior–superior direction.

MA, USA) that discretizes each phantom into a three-dimensional array of organ tags. To properly model the physical phantoms, all materials were assigned to one of four tissues used: soft tissue, bone tissue, lung tissue, and air (in-body), for which compositions and densities were taken directly from ICRP Publication 89. Starting from the top of the head to the mid-thigh, MCNPX simulations of an axial CT exam were computed with a beam width of 3.2 cm (64 slices $\times$ 0.5 mm) every 1.6 cm (for increased starting position precision) and organ doses to all in-field and out-of-field organs were calculated (per starting photon). Each organ dose was then normalized to mGy mAs⁻¹ using a normalization factor (NF) measured at each energy-collimation combination as shown in Eqs. (5) and (6). This methodology is consistent with previous CT dose reporting software such as CTDosimetry and CT-Expo which rely on precomputed Monte Carlo dosimetry on a computational anthropomorphic phantom.^{12,13}

$$NF_{E,C} \left(\frac{\text{photons}}{\text{mAs}} \right) = \frac{\text{Air Kerma}_{\text{measured}} \left(\frac{\text{mGy}}{\text{mAs}} \right)}{\text{Air Kerma}_{\text{simulated}} \left(\frac{\text{mGy}}{\text{photon}} \right)} \quad (5)$$

$$\text{Organ Dose} \left(\frac{\text{mGy}}{\text{mAs}} \right) = \left[\sum_{i=Z_{\text{exam start}}}^{Z_{\text{exam end}}} \text{Organ Dose}_i \left(\frac{\text{mGy}}{\text{photon}} \right) \right] \times NF_{E,C} \left(\frac{\text{photons}}{\text{mAs}} \right) \quad (6)$$

For torso exams, each exam range was determined using the anatomical landmarks (from Table I) which are used as guidelines for clinical practice. In addition to the visible exam range, helical over-ranging must be taken into account as it greatly influences dose values to organs that lie at the boundary of the exam. Previous work on the Toshiba Aquilion ONE scanner determined that CAP exams on average over-range

by one additional full rotation on both ends of the exam.²⁸ Using this information and the organ doses as a function of z-location, the appropriate exam range was determined for each organ. To convert relative organ doses (per mAs) to absolute organ doses, each organ dose must be multiplied by an effective mAs as derived in Eq. (7):

$$\text{Effective mAs} = \frac{(\text{Exam Average mA}) \times \text{Rotation Time (s)}}{\text{Exam Pitch}} \quad (7)$$

For uniform dosimetry, which does not account for local attenuation, each organ dose contribution is multiplied by the exam's effective mAs and then summed. For weighted dosimetry, each organ dose contribution (function of z-location) is multiplied by the normalized attenuation-based weighting factor (for that same z-location), WF, as given in Eqs. (8) and (9):

$$WF(z) = \frac{AV_{\text{average}}(z)}{\sum_{i=Z_{\text{exam start}}}^{Z_{\text{exam end}}} AV_{\text{average},i}} \quad (8)$$

$$\text{Organ Dose (mGy)} = \left[\sum_{i=Z_{\text{exam start}}}^{Z_{\text{exam end}}} \text{Organ Dose}_i \left(\frac{\text{mGy}}{\text{mAs}} \right) \times WF_i \right] \times \text{Effective mAs} \quad (9)$$

For image-based dosimetry, the reported mA from each image in the anthropomorphic phantom's image-set was recorded directly from the DICOM header. Figures 3 and 4 show a comparison of the mA distribution for the UF15F and UFADM phantoms undergoing a 120 kVp chest–abdomen–pelvis exam, respectively. The image-based mA data were resampled to the size of the axial dosimetry data. This was achieved by comparing the anatomical landmarks (“Exam Start” and “Exam End” from Table I) of both the CT image-

set and spatial extent of each axial beam center within the computational phantoms.

It should be noted that for these phantoms, no vendor specified maximum mA value (500 mA) was reached during these exams and thus was not taken into account in the derived WF values. However, a minimum mA value of 100 was used as shown in Figs. 3 and 4. Minimum or maximum mA settings can be taken into account within Eq. (9) by restricting the value of WF_i so that its product with the effective mAs (which would have to be known) satisfies the mA constraint.

3. RESULTS

The UF15F and UFADM physical phantom dosimetry results are given in Tables IV and V, respectively. Measured organ dose, calculated organ dose, and root mean square percent difference (RMS) are included for each phantom for: (a) uniform weighting (*uniform*), (b) computational phantom-derived attenuation-based weighting (*weighted*), and (c) CT image-set-derived mA weighting (*image*), all in units of mGy. Table VI summarizes these percent differences (minimum, maximum, median of the magnitude, and RMS) as a function of beam energy, phantom, as well as overall results for all exams.

4. DISCUSSION

4.A. Comparison of organ dose estimates

As shown in Table VI, weighting axial-slice organ dosimetry based on relative attenuation (under either the *weighted* and *image-based* schemes) resulted in overall better organ dose accuracy across all five phantom-energy combinations. *Uniform* dosimetry exhibited a root mean square percent difference of 21.2%, compared to 12.1% and 15.1% for the

FIG. 3. Comparison of the mA distribution (WF multiplied by Average mA) for each of the three dosimetry methods on the UF15F phantom (120 kVp exam shown). Note: Smaller Z values correspond to the top (cranial–caudal direction) of the phantom.

FIG. 4. Comparison of the mA distribution (WF multiplied by Average mA) for each of the three dosimetry methods on the UFADM phantom (120 kVp exam shown). Note: Smaller Z values correspond to the top (cranial–caudal direction) of the phantom.

TABLE IV. Summary of measured and estimated organ doses for the UF15F phantom at 120 and 135 kVp.

Organ	Measured	Uniform	% Diff ^a	Weighted	% Diff ^a	Image	% Diff ^a
120 kVp organ doses (mGy)							
Thyroid	13.9	24.2	74.5	18.1	30.1	21.3	53.3
Lung	12.2	12.1	−1.2	13.6	11.4	12.4	1.5
Thymus	11.7	13.6	16.4	12.6	7.9	12.5	6.6
Stomach	14.6	14.4	−1.5	13.8	−5.6	13.7	−6.5
Liver	14.9	13.6	−8.4	13.7	−8.0	13.6	−8.6
Gallbladder	14.0	12.8	−8.3	12.3	−11.8	12.1	−13.4
Esophagus	12.1	12.5	3.2	12.6	3.8	12.4	1.8
Spleen	14.5	13.6	−6.2	13.2	−9.3	13.1	−10.2
Kidneys	12.7	13.9	9.4	12.6	−1.0	12.6	−1.3
Pancreas	12.6	13.6	7.4	12.5	−1.3	12.3	−2.7
Colon	13.6	14.6	7.5	13.7	0.9	13.8	1.6
SI	14.6	14.7	0.2	13.7	−6.2	13.8	−5.8
Bladder	11.5	11.5	−0.1	12.5	9.0	12.7	10.5
Gonads	11.3	10.1	−10.0	11.3	0.2	11.4	0.9
RMS difference (%)			21.2		10.5		15.7
135 kVp organ doses (mGy)							
Thyroid	18.3	27.4	50.2	22.7	24.3	24.1	32.1
Lung	14.7	14.3	−2.5	16.0	9.2	14.7	0.1
Thymus	14.6	16.0	9.8	15.2	4.0	14.7	0.7
Stomach	17.0	17.0	−0.3	16.1	−5.6	16.1	−5.3
Liver	18.3	16.1	−11.6	16.0	−12.4	16.1	−11.9
Gallbladder	17.5	15.3	−12.4	14.6	−16.8	14.5	−17.2
Esophagus	14.8	14.9	0.5	15.2	2.9	14.7	−0.8
Spleen	17.8	16.0	−10.0	15.3	−14.0	15.4	−13.7
Kidneys	15.5	16.4	5.9	14.7	−5.3	14.8	−4.3
Pancreas	15.7	16.1	2.6	14.7	−6.9	14.7	−6.9
Colon	15.8	17.1	8.0	15.8	0.1	16.2	2.3
SI	17.2	17.3	0.3	16.0	−7.3	16.3	−5.6
Bladder	13.8	13.7	−1.2	14.7	6.2	15.1	9.1
Gonads	13.2	12.2	−8.0	13.3	0.9	13.6	3.1
RMS difference (%)			15.1		10.4		11.6

^a[(calculated dose − measured dose)/measured dose] × 100%.

weighted and image-based dosimetry, respectively. This trend points to the need to account for patient (or phantom) attenuation when computing organ doses from CT exams performed under TCM. There is also no clear dependence on beam energy nor phantom size as the *uniform* dosimetry yielded inferior results in all direct comparisons. As expected, the thyroid (lower than average attenuating region) and the gonads (higher than average attenuating region) experienced the highest estimation percent difference with regard to the uniform dosimetry (no attenuation is considered). It should be noted that this effect is compounded by the starting angle of the helix which makes peripheral organ doses more erratic, especially for the lower doses observed in the thyroid. This trend is supported visually (especially for the thyroid) in Figs. 3 and 4 as the average mA of the exam is overestimated in the neck region of each of the two phantoms. Additionally, the range of the percent differences is smaller (%min and %max) for the attenuation-based approaches. It should be noted

that the median (magnitude) of organ dose is comparable for three studies.

When comparing *weighted* to *image-based* dosimetry, *weighted* dosimetry is superior by a small margin. This is surprising as the scanner reported mA map should be the exact representation of the scanner's output. This discrepancy could have been caused by having to discretize this information to the resolution of the dosimetry data. Image data are available at a finer resolution and had to be resampled. Additionally, the construction of the anthropomorphic phantom is based on using 5 mm slices, which results in small air gaps within the phantom. These gaps can cause streaking artifacts in the image-set as well as affecting the ^{SURE}Exposure 3D TCM algorithm. This artifact is not something that would appear in the computational attenuation measurements as the phantom geometry has no discontinuities within the Monte Carlo geometry.

Directly comparing the mA maps of the three methodologies (Figs. 3 and 4) show agreement in terms of broad trends,

TABLE V. Summary of measured and estimated organ doses for the UFADM phantom at 100, 120, 135 kVp.

Organ	Measured	Uniform	% Diff ^a	Weighted	% Diff ^a	Image	% Diff ^a
100 kVp organ doses (mGy)							
Thyroid	10.7	21.5	101.8	14.7	37.6	13.7	28.7
Lung	12.0	10.1	-15.5	11.8	-1.2	12.0	-0.2
Thymus	12.2	11.8	-3.4	13.0	7.0	12.9	6.0
Stomach	11.0	12.0	9.1	12.5	13.9	11.2	1.8
Liver	13.0	11.1	-14.5	11.8	-9.2	10.6	-19.0
Gallbladder	10.8	10.5	-2.5	10.7	-0.5	9.6	-11.0
Esophagus	10.5	9.5	-9.7	10.3	-1.8	10.0	-4.8
Spleen	10.9	11.6	5.8	12.1	10.5	10.8	-1.2
Kidneys	10.0	11.5	14.5	11.5	14.6	10.3	3.2
Pancreas	9.8	11.1	12.9	11.2	13.7	9.8	-0.5
Colon	12.5	12.8	2.9	12.7	1.9	12.3	-1.0
Small intestine	11.5	12.3	6.3	12.3	6.6	13.0	12.5
Bladder	7.7	10.1	31.2	9.6	25.8	10.7	39.2
Prostate	6.6	8.9	34.7	7.7	16.6	7.4	12.4
Gonads	8.9	6.7	-24.3	5.6	-37.1	5.3	-40.0
RMS difference (%)			30.8		17.5		17.9
120 kVp organ doses (mGy)							
Thyroid	12.3	18.8	52.7	13.1	6.2	11.9	-3.4
Lung	11.1	9.9	-10.9	11.6	5.1	11.7	5.5
Thymus	11.1	11.3	1.9	12.6	14.0	12.1	9.4
Stomach	11.8	11.6	-1.8	12.2	3.8	11.0	-6.5
Liver	12.0	10.8	-9.5	11.6	-2.9	10.5	-12.0
Gallbladder	10.6	10.4	-1.3	10.8	1.9	9.7	-8.2
Esophagus	10.0	9.5	-5.2	10.4	4.5	10.0	0.8
Spleen	11.6	11.2	-3.7	11.8	1.8	10.5	-9.0
Kidneys	10.3	11.1	8.6	11.3	9.9	10.0	-2.6
Pancreas	10.1	10.9	7.6	11.1	9.6	9.7	-4.0
Colon	11.6	12.1	4.2	12.1	4.2	11.4	-1.7
Small intestine	11.1	11.8	6.3	12.0	7.7	12.0	7.7
Bladder	8.1	10.1	25.3	9.8	20.8	11.8	46.5
Prostate	7.3	9.6	30.5	8.3	13.2	9.0	22.4
Gonads	10.8	15.0	39.0	12.0	11.2	11.7	8.2
RMS difference (%)			20.5		9.3		14.8
135 kVp organ doses (mGy)							
Thyroid	14.6	19.6	34.0	13.7	-6.3	12.5	-14.7
Lung	11.1	10.9	-2.1	12.8	15.5	12.8	15.9
Thymus	10.8	12.3	13.9	13.8	27.5	13.3	22.3
Stomach	12.5	12.7	1.8	13.4	7.7	12.1	-2.9
Liver	13.9	11.9	-14.3	12.8	-7.9	11.6	-16.5
Gallbladder	13.5	11.6	-14.3	12.0	-11.5	10.8	-20.2
Esophagus	10.6	10.5	-0.5	11.6	9.8	11.2	6.0
Spleen	11.4	12.2	6.7	12.9	12.8	11.6	1.0
Kidneys	13.4	12.2	-8.6	12.4	-7.4	11.0	-17.9
Pancreas	12.4	12.0	-3.4	12.2	-1.6	10.7	-13.6
Colon	14.0	13.1	-6.2	13.1	-6.1	12.4	-11.3
Small intestine	12.7	12.9	1.4	13.1	2.7	13.1	2.8
Bladder	10.3	11.2	9.0	10.8	5.2	13.1	27.4
Prostate	9.2	10.6	16.0	9.2	0.8	10.0	9.0
Gonads	12.7	15.7	23.3	12.6	-1.2	12.3	-3.7
RMS difference (%)			13.7		10.6		14.5

^a[(calculated dose - measured dose)/measured dose] × 100%.

TABLE VI. Summary of overall percent differences for all phantoms and energies using each of the three dose weighting schemes.

	Uniform	Weighted	Image	Uniform	Weighted	Image	Uniform	Weighted	Image
	100 kVp			120 kVp			135 kVp		
% Min	−24.3	−37.1	−40.0	−10.0	−2.9	−12.0	−12.4	−11.5	−17.2
% Max	101.8	37.6	39.2	52.7	20.8	46.5	34.0	24.3	27.4
% Median	12.9	10.5	6.0	7.51	6.23	6.53	7.97	6.85	9
% RMS	30.8	17.5	17.9	20.9	9.9	15.2	14.4	10.5	13.2
	UF15F			UFADM			All		
% Min	−12.4	−16.8	−17.2	−24.3	−37.1	−40.0	−24.3	−37.1	−40.0
% Max	74.5	30.1	53.3	101.8	37.6	46.5	101.8	37.6	53.3
% Median	7.4	6.5	5.7	9.03	7.66	8.23	7.98	7.27	6.61
% RMS	18.4	10.5	13.8	22.8	13.0	15.8	21.2	12.1	15.1

but are not perfectly aligned. The UF15F phantom's *weighted* and *image-based* attenuation maps show general agreement in the top, middle, and bottom of the exams. One discrepancy is the faster rise in mA seen in the image-set in the shoulder region. It is hard to determine the cause of this but it could be from the scanner ramping up the mA in anticipation of the highly attenuating shoulder-chest region (caused by the breast and the partial arms). For the UFADM phantom's attenuation maps, the plateau in mA in the chest region is smaller for the simulated attenuation and shows less of a spike in the pelvic region. Although the general trend is similar, the gradient seen from the image-set is not present. This could be caused by the attenuation data being calculated at discrete 1.6 cm intervals which causes the impact of the presence of pelvic bone to be minimized. Regardless, the visual mA map is less important as the direct comparison of organ dose, and it should be noted that this methodology can incorporate attenuation weighting from any source (image-derived or calculated on a computational phantom).

4.B. Sources of error

There is a wide range of possible sources of error in this experiment. From a physical measurement perspective, both the physical phantom and the dosimetry measurements have uncertainties. For the physical phantoms, one source of error is in the construction process of the phantoms themselves as they are not exact physical representations of the computational phantom after which they are modeled (in terms of size, density, and exact organ shape). The proprietary materials used do not have an exactly known material composition (simply measured effective atomic number), hence ICRP Publication 89 tissue composition are assumed in the corresponding computational phantom. In terms of the measurements, Landauer states an accuracy of $\pm 2\%$ for their "screened" nanoDotsTM system which was used for phantom dose measurements.

In terms of the computational modeling, the MCNPX source subroutine that models the scanner was derived from measurements by both a Radcal 10×6 -6 general purpose ion chamber and a 10×6 -3CT pencil ion chamber (Radcal,

Monrovia, CA, USA) which each have a calibration uncertainty of $\pm 4\%$, and an energy dependence of $\pm 5\%$. In addition to the measurements, the implicit modeling of the scanner geometry, beam shape, beam size, and energy spectra have uncertainties. All Monte Carlo calculations were performed with sufficient particle histories for less than 1% reported uncertainties for in-field organs.

In terms of exam modeling, modeling a helical scan as a sum of axial scans affects the overall error. Simplifying the helical geometry to a series of axial rotations minimizes starting angle and pitch effects, especially for superficially located organs. Although pitch is embedded into the effective mAs value [Eq. (7)], it is not explicitly modeled. One limitation of this study is the fact that only a pitch below 1.0 (not above) was tested, which would cause uncertainty for exams with larger pitch values. One major source of error in the attenuation calculation is the assumption of one AV_{value} per z-location. Ideally, organ dose contributions would be a function of specific beam location and not only z-location as shown in other studies.²² Nevertheless, the mathematical complexity and the computational burden to achieve this level of accuracy in attenuation information does not seem reasonable for the scope of this project — establishment of a precomputed CT dose library for the UF series of computational phantoms.³³ Regardless of the method used for CT dosimetry, there also exists error in modeling a specific CT exam as its spatial extent must be inferred from the CT image-sets and a constant over-ranging value must be assumed. This method has only been tested against Toshiba's ^{SURE}Exposure 3D TCM technique and thus it would require further validation for organ dosimetry on other CT vendors.

5. CONCLUSIONS

The purpose of this study was to quantify and validate the use of precomputed axial dosimetry for estimating organ doses from CT exams. Even when patient (or phantom) attenuation is not accounted for, the axial summing of organ dose contributions yields acceptable organ dose estimates. Furthermore, incorporating patient attenuation from either the image-set or a computational estimate has a positive impact

on organ dose estimates. This methodology is not vendor-specific and can incorporate attenuation data from any source. This lends itself well to retrospective CT dosimetry as it is unlikely the image-set can be processed for each patient in a cohort. In that situation, this methodology allows for an estimate of patient attenuation by computing it on a computational phantom of similar size in the same region of the body the exam took place. Although attenuation is only redistributed with respect to the patient's z-axis, this methodology improves and supports the continued use of precomputed axial CT dosimetry.

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