SimDoseCT: dose reporting software based on Monte Carlo simulation for a 320 detector-row cone-beam CT scanner and ICRP computational adult phantoms

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SimDoseCT: dose reporting software based on Monte Carlo simulation for a 320 detector-row cone-beam CT scanner and ICRP computational adult phantoms

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Abstract

This study aims to develop and test software for assessing and reporting doses for standard patients undergoing computed tomography (CT) examinations in a 320 detector-row cone-beam scanner. The software, called SimDoseCT, is based on Monte Carlo (MC) simulation code, which was developed to calculate organ- and effective doses in ICRP anthropomorphic adult reference computational phantoms for acquisitions with the Aquilion ONE CT scanner (Toshiba). MC simulation was validated by comparing CTDI measurements within standard CT dose phantoms with results from simulation under the same conditions. SimDoseCT consists of a graphical user interface connected to a MySQL database, which contains the look-up-tables that were generated with MC simulations for volumetric acquisitions at different scan positions along the phantom using any tube voltage, bow tie filter, focal spot and nine different beam widths. Two different methods were developed to estimate organ- and effective doses from acquisitions using other available beam widths in the scanner. A correction factor was used to estimate doses in helical acquisitions. Hence, the user can select any available protocol in the Aquilion ONE scanner for a standard adult male or female and obtain the dose results through the software interface. Agreement within 9% between CTDI measurements and simulations allowed the validation of the MC program. Additionally, the algorithm for dose reporting in SimDoseCT was validated by comparing dose results from this tool with those obtained from MC simulations for three volumetric acquisitions (head, thorax and abdomen). The comparison was repeated using eight different collimations and also for another collimation in a helical abdomen examination. The results showed differences of 0.1 mSv or less for absolute dose in most organs and also in the effective dose calculation. The software provides a suitable tool for dose assessment in standard adult patients undergoing CT examinations in a 320 detector-row cone-beam scanner.

1. Introduction

The number of computed tomography (CT) examinations increased in recent years due to technological advances and new acquisition techniques. The National Council on Radiation Protection and Measurements reported on the high contribution of CT in the annual radiation exposure of the population in the United States, which represents the 24% of the dose thus becoming the most significant contribution after natural background radiation (Schauer and Linton 2009). A study published by the European Commission demonstrated that CT was associated with more than half of the medical radiation exposure of the European population in 2007-2010 (European Commission 2013). Considering the current interest in radiation exposure from CT examinations, there is a need to develop methods for patient doses assessment in CT examinations that are up to date with current scanner technology and design, current acquisition protocols and the latest reference in computational phantoms. Awareness about radiation exposure in CT contributes to the optimization of the clinical application of CT and the reduction of radiation risks.

Currently, CT scanners display dose information based on operational dosimetric quantities such as the volume CT dose index ($CTDI_{vol}$), which provides the average absorbed dose in a transection of a standard cylindrical CT dose phantom, and the dose-length product (DLP), which also considers the scan length. According to the ICRP (ICRP 1977, ICRP 1991), in radiation protection organ- and tissue doses are the preferred dose descriptors since they can be correlated with radiation risks. The effective dose (E) is calculated from organ- and tissue doses and takes into account the overall radiation-induced health effects and it also allows for practical comparison between radiation exposure from different techniques in diagnostic radiology and beyond.

There are different methods to estimate the organ- and tissue doses or effective dose from CT examinations such as laborious studies with dosimeters embedded in anthropomorphic phantoms or advanced Monte Carlo (MC) simulations. These complex MC studies usually provide numerous and large look-up-tables from which useful dosimetric information can be retrieved, like organ- and tissue doses, with software tools that consist of a relatively simple algorithm and a user interface. From these software tools simple k-factors can be derived that convert the DLP for broadly defined body regions into effective dose or dose in a few organs, which are useful for rough assessment of radiation exposure (ICRP 2007a, Kobayashi *et al.* 2016).

Several dosimetric software tools have been developed to facilitate the retrieval of organ- and tissue dose and effective dose from look-up-tables that were generated with MC simulations, i.e. ImpactDose (Kalender *et al.* 1999), CT-Expo (Stamm and Nagel 2002), Waza-Ari (Ban *et al.*

2011a, Takahashi *et al.* 2011), ImPACT (ImPACT 2012) and VirtualDose (Ding *et al.* 2015). These tools are often used for clinical dose assessment, but also for dose assessment in scientific studies involving CT. Differences between the software tools are significant and the user has to be aware of the limitations (Abdullah *et al.* 2012). Many of the software tools, such as CT-Expo or ImPACT, provide limited functionally due to the use of nowadays outdated mathematical phantoms. In addition, they use outdated look-up-tables, some dating back to the late 1980s and early 1990s. Jansen and Shrimpton (2016) demonstrated for recent scanners potential deviations by up to around 30% when comparing new dedicated MC simulations with results from the ImPACT calculator. A limitation of the Waza-Ari program is that it uses only phantoms based on relatively small Japanese patients and doses can differ around 20% for the same examination compared to ICRP phantoms (Ban *et al.* 2011b). VirtualDose uses anatomically realistic phantoms but not the standard male and female from ICRP110 (ICRP 2009) and only allows dose estimation for 16-slices scanners at the moment. These disadvantages were overcome with ImpactDose since it has gradually incorporated many scanners and phantoms, even those published by ICRP110. The ImPACT calculator is kept up-to-date by scanner matching, this means that current CT scanners are matched with scanners from the late 1980s based on dosimetric characteristics, but the look-up-tables are calculated for scanners from the late 1980s.

In conclusion, contemporary CT scanners, like the 320 multi-slice CT scanner (Aquilion ONE Vision, Toshiba), are not implemented appropriately in currently available dose assessment programs, and they are often based on outdated mathematical phantoms. For example, for the Aquilion ONE current, the cone beam shape and the use of the adequate beam width with its associated overbeaming is not adequately taken into account (McCollough and Zink 1999).

This study aimed at developing and testing software for accurate dose assessment in CT with the Aquilion ONE scanner. The software, called SimDoseCT, is based on look-up tables generated from MC simulation. The dose assessment should take into account all relevant technical characteristics of the scanner, like focal spot size, overbeaming, overranging, beam shaping filter, the cone beam geometry and the heel effect. Dose assessment was performed for the current standard phantoms, namely the ICRP computational adult male and female phantom. The software should also take into account all possible acquisitions within the Aquilion ONE CT scanner, like axial (volumetric), helical and scanogram acquisitions. Selecting the appropriate acquisition technique in the software was facilitated by using a graphical user interface that accurately resembles the user interface of the actual CT scanner.

Hence, a tool for an accurately organ- and effective doses estimation in a 320 detector-row cone-beam scanner was offered with the aim of improving the easily dose evaluation for standard adult patients in CT contemporary scanners.

2. Materials and methods

2.1 CT scanner model

The Aquilion ONE CT scanner (Toshiba Medical Systems, Otawara, Japan, software version 4.74ER001) was modelled in great detail for this study. This 320 detector-row cone-beam CT scanner operates at sixteen different beam collimations allowing a z-axis coverage of the beam between 2 and 160 mm (i.e. 160, 140, 128, 120, 100, 80, 60, 50, 40, 32, 20, 16, 12, 8, 4 and 2 mm) and at different tube voltages (i.e. 80, 100, 120 and 135 kVp). Three beam shaping filters (i.e., small, medium and large bow tie filters) can be selected to optimize the x-ray beam intensity inside the field of view (FOV).

The tube current can be selected from 10 to 50 mA in 5 mA steps and from 50 to 580 mA in 10 mA steps. The scanner allows eleven rotation times: 0.35, 0.375, 0.4, 0.45, 0.5, 0.65, 0.75, 1.0, 1.5, 2.0 or 3.0 s. The focal spot size, either large or small, is automatically chosen by the scanner depending on the selected acquisition parameters.

The Aquilion ONE CT scanner has two scan modes for cross-sectional imaging with a rotating x-ray tube, i.e. an axial (volumetric) acquisition or a helical acquisition. In addition the scanner is also able to perform projection imaging with a static x-ray tube for planning purposes, which is referred to as a scanogram.

2.2 *ICRP computational phantoms*

The two adult computational phantoms, the anthropomorphic male (AM) and anthropomorphic female (AF) phantoms, were published by the ICRP (ICRP 2009) and they were used in this study for dose estimation. These two phantoms are regarded as the current international standard for dosimetry. They were constructed from CT images of humans and they represent an average male (length 178 cm, weight 73 kg) and an average female (length 168 cm, weight 60 kg). The voxel size is 2.1 x 2.1 x 8.0 mm^3 for the male and 1.8 x 1.8 x 4.8 mm^3 for the female. The ICRP provides detailed information for the two phantoms, including mass, spatial distribution and composition of each organ or tissue. Each phantom was implemented in a MC simulation program (Salvadó *et al.* 2015*)*, through three representations for each voxel concerning respectively the 141 organs or tissues, the 53 materials and the 29 organs that contribute to the effective dose calculation. A specific subsegmentation was required for tissues containing red bone marrow (RBM) or endosteum (bone surface) because the microscopic structures of the skeleton are smaller than the size of a voxel (Zankl *et al.* 2007). Additionally, for the mentioned tissues, the energy increase due to the secondary particles that are released from mineral bone

components was implemented in the simulation program through an enhancement factor (King and Spiers 1985). Figure 1 shows a surface rendering of the adult phantoms used and different views showing the organs that contribute to the effective dose calculation.

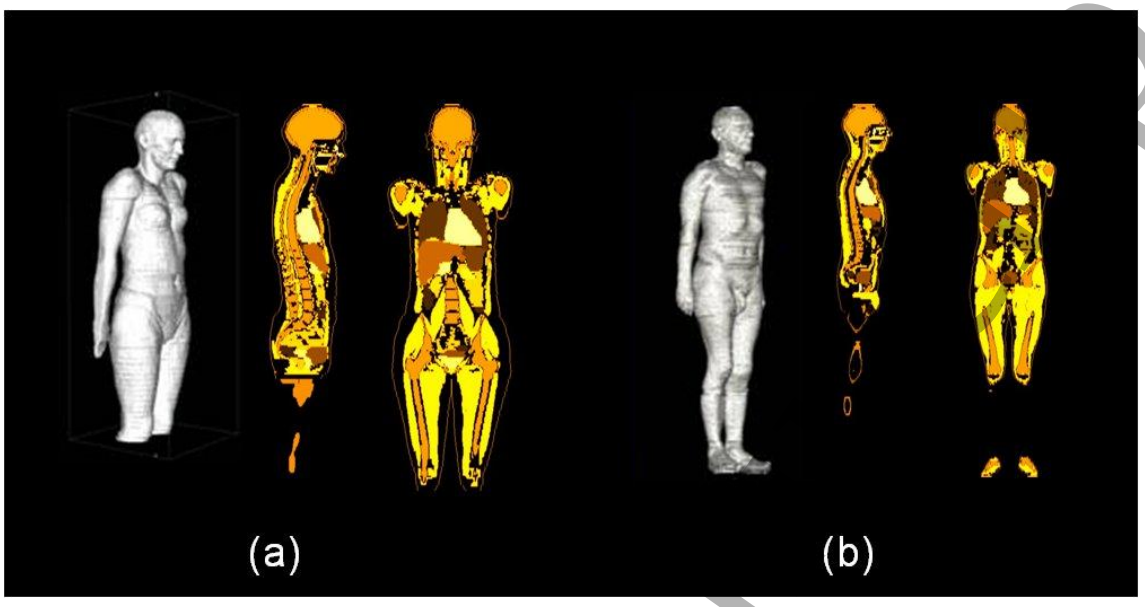


Figure 1. From left to right, 3D surface views and sagittal and coronal views showing the different organs with weighting factors that contribute to the effective dose in different colours for (a) adult female phantom and (b) adult male phantom.

2.3 Monte Carlo simulation

The Aquilion ONE CT scanner was modelled in a MC program according to the specific technical characteristics of this scanner. The MC code simulates all aspects that influence the dose distribution in the phantoms during a CT scan, including the characteristics of the interaction (attenuation and scatter), the x-ray spectrum and x-ray beam (tube voltage and primary filtration, bow tie filter, cone beam geometry, heel effect, beam width, penumbra, rotation and tube charge) and the patient support table (shape, size and material). The CT acquisition parameters can be introduced as an input to the MC code (i.e. tube voltage, tube current, pitch, scan position and scan range). For the calculation of radiation transport, the Electron Gamma Shower V4 (EGS4) code (Nelson *et al.* 1985) in combination with the Low Energy Photon Scattering Expansion (National Laboratory for High Energy Physics (KEK) Japan) (Hirayama *et al.* 2000) was used. The simulation of photons transport in the energy range typical for CT scanners is based on processes of Rayleigh scattering, Compton scattering and photoelectric effect and the associated creation of fluorescent photons or Auger electrons. A cut-off energy of 5 keV was used for photon transport, and a cut-off energy of 30 keV was used for electrons.

The MC program was developed and validated for a beam collimation of 160 mm in a previous study (Salvadó *et al.* 2015). In this prior work, the validation was performed with a set of CTDI measurements and the corresponding simulations within two standard CT dose phantoms (150 mm long, 160 mm diameter for the head phantom and 320 mm diameter for the body phantom) and using a 100 mm long CT ionization chamber. Agreement between measurements and MC results was within 5%. Similar procedure was used in this study to validate the MC program for all the available collimations in the scanner by comparing the results from simulations in standard CT dose phantoms with the CTDI measurements provided by Toshiba's manufacturer under the same conditions. The CTDI was calculated according to the definitions of the IEC (IEC 2009) and the IAEA (IAEA 2011) at the centre position and as average at the peripheral positions of the CT dose phantoms. The validation was performed for each available collimation, tube voltage, FOV and focal spot size. Once validated, the MC code calculates the absorbed energy in each voxel of the phantom. The mean absorbed dose in each organ or tissue was computed as the total absorbed energy in the voxels corresponding to this organ or tissue divided by the total mass. The mean absorbed dose in each material was also calculated.

MC simulations were performed for cross-sectional imaging of the two anthropomorphic adult reference computational phantoms concerning all possible combinations of the following acquisition parameters: tube voltage (i.e. 80, 100, 120 and 135 kVp), FOV (small, medium and large; this is associated with the beam shaping filter), focal spot size (small or large) and nominal beam width (160, 128, 60, 50, 40, 16, 8, 4 and 2 mm). The simulations were performed as separate axial scans of one rotation (360 different angles) at different positions each 4 mm along the phantom from head to feet for all collimations except for narrowest beams. Simulations were performed each 2 mm for 4 mm-collimation and each 1 mm for 2 mm-collimation. Thus, the overlap in the simulations ranges from a 2-fold overlap for the 2, 4 and 8 mm beam widths to a 40-fold overlap for the 160 mm beam width.

Additionally, simulations were performed for the projection imaging of the two phantoms (scanogram) using contiguous simulations, each 1 mm along the phantom, for a beam width of 2 mm, for all tube voltages and FOV sizes, a large focal spot size and four tube angles (0, 90, 180 and 270º). MC simulations provided doses in all organs and doses in all different materials in the phantom, normalized per mAs.

In most CT acquisition protocols of the body, the arms of the patient are positioned along the head and in CT scans of the head, along the body. Accordingly, the arms were removed from the phantoms during all the simulations. The mass of the arms was taken into account for the calculation of the absorbed dose.

The MC simulations were performed in a supercomputing center (CSUC, Consorci de Serveis Universitaris de Catalunya) and were carried out using 2.0×10^7 photon histories for each cross-sectional simulation and 2.0×10^6 for each projection simulation (scanograms). The computation times for a complete cross-sectional simulation, e.g. one full rotation in each scan position along the phantom, were on average 5 days for all the collimations, except for narrowest beams that were 10 days for 4mm-collimation and 20 days for 2mm-collimation. The computation time for the scanograms was on average 2 days.

2.4 SimDoseCT software

SimDoseCT consists of a graphical user interface connected to a MySQL database. The application runs on a PC and shows the same interface as the scan console of the Aquilion ONE CT scanner (software version 4.74). The user can design an acquisition protocol with the same options and user interface as on the scanner console. Once the acquisition protocol is designed the user has the option to visualize the dose values either by selecting a dose tab in the interface or by generating a PDF dose report. Both options will trigger the application to retrieve dose values from the database. The database can either be stored local or accessed as an online database by performing a secure query. The graphical user interface has read-only privileges in the database for security reasons. The user can change any acquisition parameter and the dose will be updated in real-time to show how it affects the organ dose and effective dose. This software is coded using the object-oriented programming language C# (Microsoft, USA).

2.5 Dose calculation algorithm in SimDoseCT

Doses were stored, for specific scan positions along the phantom, in the database (each 4, 2 mm and 1 mm for the narrowest beam widths). Doses at other scan positions can be linearly interpolated by SimDoseCT in steps of 0.5 mm. Since MC simulations were performed only for nine of the sixteen available beam widths of the Aquilion ONE scanner, two methods were used to estimate doses from acquisitions using other available beam widths. Table 1 shows the method used for each beam width.

Table 1. Nominal beam width of reference and method used to estimate the absorbed dose for the seven collimations not included in database.

b (mm)	b_{ref} (mm)	Method used
140	128	1
120	60	2
100	50	2
80	40	2
32	16	2
20	16	1
12	16	1

For beam widths close to one of the sixteen available beam widths in the database, the following correction factor based on actual beam widths was used to calculate absorbed doses (*method 1*):

$$D=D_{ref}\cdot\frac{b_{actual}}{b_{ref,actual}} \qquad (1)$$

where D is the absorbed dose to be estimated for beam width b and b_{ref} is the beam width corresponding to dose values, D_{ref}. The subscript *actual* indicates that the overbeaming was also considered in the beam width. The overbeaming was calculated from the actual shape of the x-ray beam profile along the axis of rotation from the scanner. For beam widths equal to or below 80 mm, overbeaming was measured using the Gafchromic XR-QA Dosimetry Film (International Specialty Products Inc. (ISP), Wayne, NJ, USA) and for larger beam widths, the Piranha 657 dose profiler (RTI Electronics, Fairfield, NJ, USA) was used.

For beam widths that are substantially smaller or wider compared to beam widths available in the database, a different method was used. The dose values were calculated using a reference beam width that is twice as small as the beam width of interest. The sum of the absorbed doses from two scan positions with an interval equal to the beam width of reference was used together with a correction factor to take into account the effect of overbeaming (*method 2*):

$$D=D_{ref}\cdot\frac{b_{actual}}{2b_{ref,actual}} \qquad (2)$$

where D is the absorbed dose to be estimated for beam width b, and D_{ref} is the absorbed dose for the beam width b_{ref}. The subscript *actual* indicates that the overbeaming was also considered in the beam width.

The dose from the scanogram acquisition is calculated by directly summing the dose from contiguous positions within the range of the scanogram, using a beam width of 2 mm.

For the axial mode with a single rotation, the program provides the doses from the database according to the CT input parameters and concerning the scan position to cover the scan range with the selected beam collimation. When contiguous axial scans are required to cover the scan range with the selected collimation, the program obtains the organ doses by summing the dose corresponding to the actual positions and actual beam width.

The dose values that are stored in database are normalized to mAs per rotation. These values have to be corrected for the nominal mAs (Q_{nom}), the pitch (p) and the overlap between the successive positions taking into account the distance between the successive positions (d_s) and the beam width (b). The corresponding correction factor is Q_{eff}.

$$Q_{eff} = \frac{1}{p} \cdot \frac{d_s}{b} \cdot Q_{nom} \qquad (3)$$

The extra rotations at the start and the end of the helical acquisition (overranging) were also considered and added to the planned scan range. The overranging was assessed as one entire extra rotation at the start and the end of the acquisition following the same procedure described in previous works (van der Molen and Geleijns 2007).

In addition to organ doses, the software also calculates the effective dose for the male and female phantom using tissues weighting factors from ICRP Report 103 (ICRP 2007b). For the gender-specific calculation, the weighting factor for female breast tissue was taken 0.24 and no tissue weighting factor was applied for male breast tissue. SimDoseCT also provides the $CTDI_{vol}$ that corresponds to the designed acquisition protocol, for all combinations of tube voltage, FOV and beam collimation. The dose length product (DLP) was calculated by multiplying the $CTDI_{vol}$ with the nominal scan length. SimDoseCT shows the dose results (i.e. organ doses, effective dose, $CTDI_{vol}$ and DLP) in real time for the whole protocol as well as for each individual scan.

2.6 Validation of SimDoseCT algorithm

Validation of the software algorithm was achieved by comparing the dose estimation from the SimDoseCT tool with dose results obtained from dedicated MC simulations for specific acquisitions. The dose values taken into account in the validation were the effective dose and the absorbed doses in organs that contribute to the effective dose. In addition, the absorbed dose in other relevant organs such as eye lens, testis and ovaries were also evaluated. The testing was performed using three different axial volumetric protocols for head, thorax and abdomen. The

reported doses for the axial acquisitions are associated with only one rotation. For each protocol, the validation was performed for six collimations for which the look-up-table was not included in the database (140, 120, 100, 80, 20 and 12 mm) and for the maximum and minimum available beam widths of the scanner (160 and 2 mm). A 64 slice, 32 mm-collimation, abdomen protocol was used to validate the helical mode. The CT acquisition parameters of the protocols that were used for the validation are provided in Table 2.

Table 2. Acquisition parameters of the CT protocols used for the SimDoseCT algorithm validation.

CT Examination	Tube voltage (kVp)	Rotation time (s)	Tube current (mA)	Pitch	Field of view (FOV)	Focal spot size	Beam width (mm)	Scan position from the top of the head (mm)	
								AM	AF
Volume - Head	80	0.5	200	_	Small	Small	160, 140, 120, 100, 80, 20, 12, 2	80	80
Volume - Thorax	120	0.5	200	_	Medium	Large		436	402[a]
Volume - Abdomen	100	0.5	200	_	Large	Large		524	494[a]
Helical - Abdomen	100	0.5	200	0.828	Large	Large	32	444-604	414-574[a]

[a]Scan positions that required interpolation in SimDoseCT.

2.7 Dose comparison with results from other dosimetric software tools

Most of the existing dosimetric software tools today are based on pre-calculated tables with doses from outdated generic scanners from the late 1980s. Since SimDoseCT provides accurate organ dose reporting in a specific CT contemporary scanner, organ doses in SimDoseCT were compared with results from other CT patient dosimetry calculators, like ImPACT and ImpactDose, in order to demonstrate the functionality of the SimDoseCT tool for a 320 detector-row cone-beam scanner. ImPACT uses mathematical phantoms and Impactdose tool uses ICRP computational adult phantoms. Dose differences were calculated in the three volumetric examinations (head, thorax and abdomen) using a scan range of 160 mm.

3. Results

3.1 Monte Carlo validation

To validate the MC simulation for all the available collimations in the 320 detector-row cone-beam scanner, differences in $CTDI_{centre}$ and $CTDI_{surface}$ from manufacturer measurements and from MC simulation were calculated. Differences were up to 9% for all the available collimations in the scanner and for all tube voltages and bow tie filters, confirming that dose calculations from MC simulation were in agreement with the actual exposure conditions. Table 3 provides the CTDI comparison for six different collimations using 80 kV for the head

phantom with the small bow tie filter (S) and small focal spot size (SFS) and using 120 kV for the body with the medium filter (M) and large focal spot (LFS).

Table 3. CTDI measurements from manufacturer and CTDI results from MC simulation for six different collimations available in the Aquilion ONE scanner for a head and a body examination. Differences were also calculated.

	$CTDI_{center}$ (mGy/mAs)			$CTDI_{surface}$ (mGy/mAs)		
Beam width (mm)	Measurements	MC simulation	Δ (%)[a]	Measurements	MC simulation	Δ (%)[a]
HEAD 80kV,S ,SFS						
160	0.069	0.065	-6.7	0.075	0.077	3.1
80	0.063	0.058	-7.9	0.068	0.071	3.9
40	0.071	0.066	-5.9	0.076	0.079	3.0
32	0.079	0.074	-5.8	0.085	0.088	3.1
16	0.089	0.084	-5.1	0.096	0.098	2.2
2	0.224	0.212	-5.1	0.241	0.247	2.4
BODY 120kV, M ,LFS						
160	0.061	0.056	-8.2	0.112	0.114	2.2
80	0.052	0.048	-8.5	0.092	0.095	2.9
40	0.059	0.055	-6.2	0.105	0.107	2.2
32	0.065	0.062	-5.0	0.116	0.118	1.4
16	0.078	0.073	-5.9	0.138	0.140	1.7
2	0.229	0.218	-4.9	0.406	0.412	1.4

[a] $\Delta(\%) = \frac{Simulated - Measured}{Measured} \times 100$

3.2 SimDoseCT

Look-up-tables were calculated for 445 scan positions along the AM phantom and 306 positions along the AF phantom for 264 combinations of scan parameters (kVp, FOV, focal spot size and collimation) available in the Aquilion ONE CT scanner. For the narrowest collimations (4 mm and 2 mm), the number of positions, and thus look-up-tables, is even two and four times higher respectively. The look-up-tables allow for dose calculation at any tube position through interpolation, even if the exact position is not included in the look-up-tables. At each scan position, and for all scan parameters the look-up-tables provide the normalised absorbed dose for 170 organs and tissues and for 54 materials. In total 378504 look-up-tables were calculated in almost 1.0×10^5 hours of simulation time.

Consequently, by retrieving data from the look-up-tables the organ- and effective doses from any CT Aquilion ONE protocol (working with software version 4.74) and any scan range can be obtained for the male and female computational phantoms, for each scan mode (i.e. volume, helical or scano) and for the CT acquisition parameters specified by the user in the SimDoseCT user interface (Table 4). The calculation takes into account the dose gradients in the beam e.g.

due to the beam shaping filter, the heel effect and the penumbra due to the finite focal spot size. The effects of overbeaming and overranging are also taken into account.

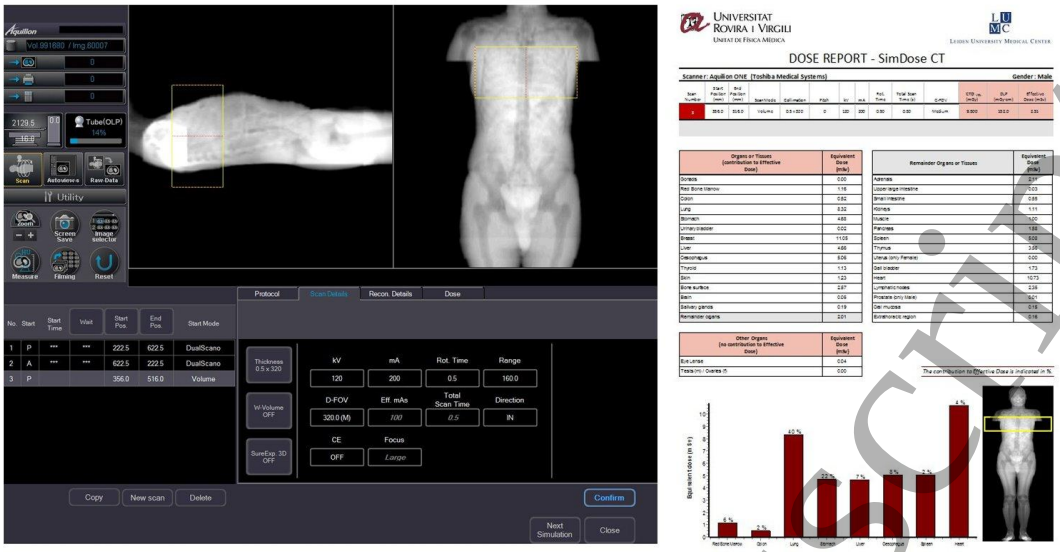
Table 4. Available phantoms and CT scan parameters in SimDoseCT.

Scan mode	Phantom	Tube voltage (kVp)	Field of view (FOV)	Focus spot size	Beam width (mm)
Axial (Volumetric)	AM, AF	80, 100, 120, 135	Small, Medium, Large	Small, Large	160, 140 [a], 128, 120 [a], 100[a], 80[a], 60, 50, 40, 32 [a], 20 [a], 16, 12 [a], 8, 4, 2
Helical	AM, AF	80, 100, 120, 135	Small, Medium, Large	Small, Large	80 [a], 50, 40, 32 [a], 20 [a], 16, 8, 4, 2
Tube-fixed irradiation (Scanogram)	AM, AF	80, 100, 120, 135	Small, Medium, Large	Large	2

[a] Dose values based on calculation methods for dose estimation.

After entering the acquistition protocol in SimDoseCT, the dose results are displayed immediately in the user interface. A patient dose report can be generated and saved; an example is shown in

Figure 2. The selected scan parameters for the protocol and the corresponding effective dose, $CTDI_{vol}$ and DLP are reported in a table. The absorbed dose in all the organs that contribute to the effective dose, including the remainder organs, is listed in a table together with the absorbed dose of other organs or tissues of interest, such as eye lens and testis or ovaries. The report shows also a graph with the organ doses of the eight organs that have the highest contribution to the effective dose. A graphical representation of the scanned region in the phantom is shown to provide a visual control of the selected anatomical region for the scan. The described information is presented for the entire CT examination and also for each scanogram and each individual scan in the whole protocol.



(a) (b)

Figure 2. (a) User interface of SimDoseCT software and (b) an example of the dose report generated from SimDoseCT.

3.3 Validation of SimDoseCT algorithm

Table 5. lists the $CTDI_{vol}$, the DLP and the effective doses in a head, a thorax and an abdomen CT examination for the eight selected collimations to validate the volumetric mode of the SimDoseCT software. The results are associated to only one rotation, so DLP and effective dose decrease with decreasing beam collimation. The differences between dose values obtained directly from dedicated MC simulations and SimDoseCT are presented. The results show that most differences in effective doses were below 6.5%, being negligible in most cases. For some protocols of thorax and abdomen the relative differences were slightly higher representing only absolute differences up to 0.1 mSv.

The differences in organ doses obtained from the comparison between both dose calculation tools were lower than 10% in most organs. In the case of using a beam width that used the method 1 or method 2 to estimate doses (i.e. 140, 120 100, 80, 20 and 12 mm), the differences could represent more than 10% in some organs or tissues, such as eye lenses or extra thoracic region in head protocols; the thymus in thorax examinations or the breast in abdomen scans. In these organs, the differences were 0.5 mSv at the most and 1 mSv in the case of eye lenses (between 15 and 27%).

In the case of beam widths for which the dose data are included in look-up-tables, these differences were negligible in all organs. Dose differences of 0.1 mSv were found if the selected tube position involves an interpolation of the dose data in the SimDoseCT software.

Table 5. Effective doses obtained using SimDoseCT tool and from a dedicated MC simulation program for the different volume CT examinations used for the SimDoseCT validation. The results are for one single axial rotation. The percentage dose differences between both methods were calculated. The $CTDI_{vol}$ and the DLP are also indicated in each case.

Beam width (mm)	CTDIvol (mGy)	DLP (mGy·cm)	E (mSv)					
			AM			AF		
			SimDoseCT	MC	Δ (%)[a,b]	SimDoseCT	MC	Δ (%)[a,b]
Head								
160	7.3	116.3	0.15	0.15	0.0	0.19	0.19	0.0
140	7.1	99.3	0.13	0.13	0.0	0.16	0.17	6.3
120	6.8	82.1	0.11	0.11	0.0	0.14	0.14	0.0
100	6.3	63.0	0.09	0.09	0.0	0.11	0.11	0.0
80	6.7	53.3	0.07	0.07	0.0	0.09	0.09	0.0
20	8.2	16.4	0.02	0.02	0.0	0.03	0.03	0.0
12	9.4	11.3	0.01	0.01	0.0	0.02	0.02	0.0
2	23.5	4.7	0.01	0.01	0.0	0.01	0.01	0.0
Thorax								
160	9.5	152.0	2.51	2.51	0.0	6.26	6.27	0.2
140	9.2	128.8	2.19	2.21	0.9	5.72	5.66	1.1
120	8.5	102.0	1.93	1.90	1.6	5.03	5.04	0.2
100	7.5	75.0	1.63	1.58	3.2	4.46	4.36	2.3
80	7.9	63.2	1.29	1.27	1.6	3.74	3.70	1.1
20	10.1	20.2	0.43	0.39	10.3	1.40	1.28	9.4
12	11.9	14.3	0.29	0.28	3.6	0.93	0.91	2.2
2	34.7	6.9	0.14	0.14	0.0	0.44	0.44	0.0
Abdomen								
160	6.6	105.1	2.10	2.10	0.0	2.60	2.59	0.4
140	6.3	87.6	1.90	1.89	0.5	2.22	2.24	0.9
120	5.8	70.0	1.68	1.66	1.2	1.96	1.93	1.6
100	5.2	51.6	1.47	1.43	2.8	1.68	1.62	3.7
80	5.4	43.2	1.22	1.21	0.8	1.36	1.34	1.5
20	6.9	13.9	0.41	0.38	7.9	0.44	0.41	7.3
12	8.2	9.8	0.28	0.27	3.7	0.30	0.29	3.5
2	23.8	4.8	0.13	0.13	0.0	0.14	0.14	0.0

[a] $\Delta(\%) = \frac{E_{SimDoseCT} - E_{MC}}{E_{MC}} \times 100$

[b] Differences lower than 0.005 mSv were considered 0%.

For the abdomen CT protocol using a beam width of 32 mm in helical mode, the effective dose was 3.53 mSv in SimDoseCT and 3.64 mSv in the dedicated MC simulation for AM and 4.71 mSv and 4.74 mSv for AF, respectively. Hence, maximum differences of 3% were found. Figure 3 presents the organ absorbed doses in AM and AF phantoms for the abdomen study in helical mode obtained by SimDoseCT and MC simulation. The selection of organs was based

on an exceeded limit of 1 mGy and their contribution to the effective dose is more than 99%. Differences in doses between both methods were below 5.5%, with maximum absolute differences of 0.3 mSv.

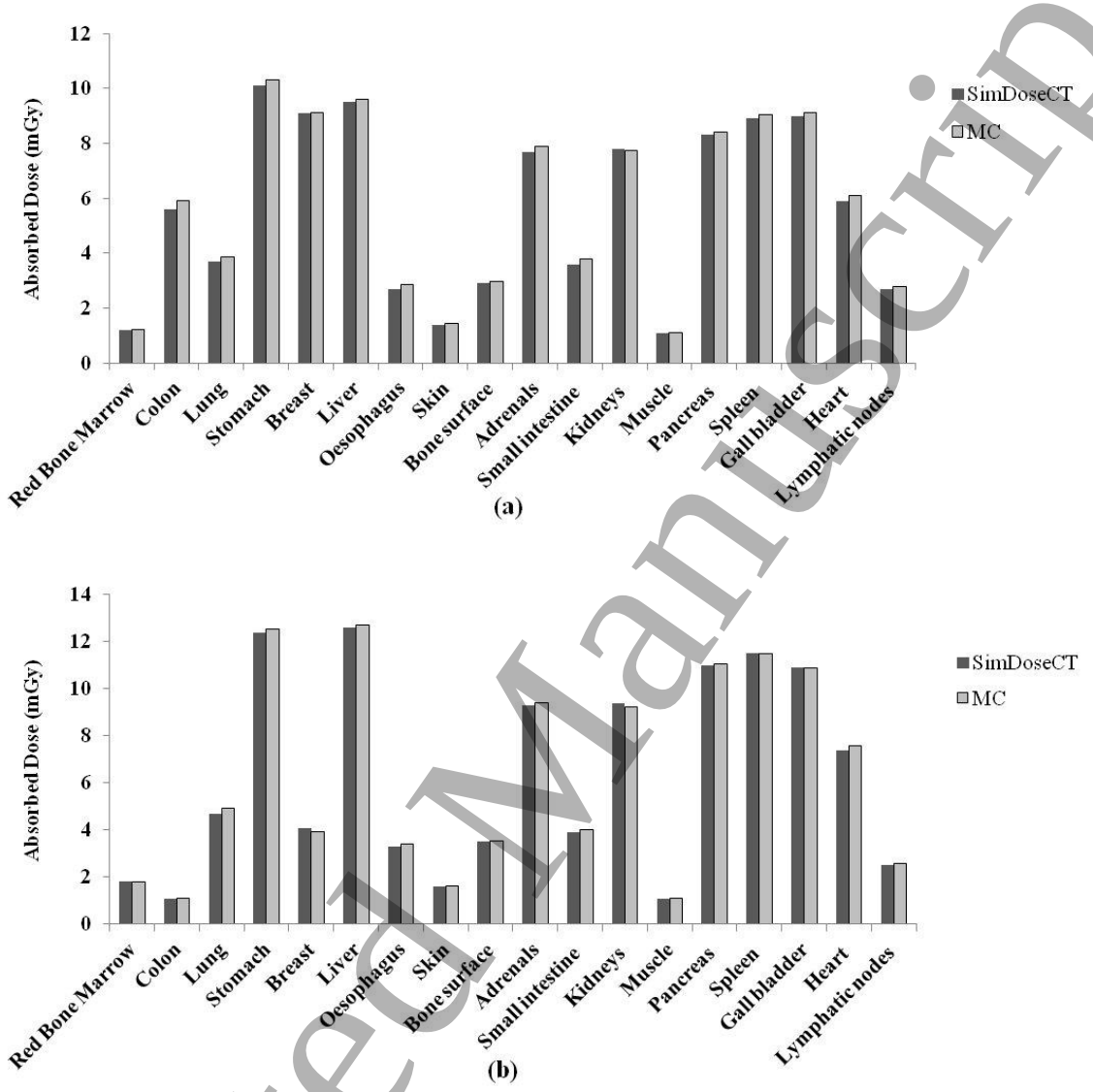


Figure 3. Absorbed doses for (a) adult male phantom and (b) adult female phantom in a helical abdomen protocol obtained from SimDoseCT tool and a dedicated MC simulation.

3.4 Comparison with other dosimetric software tools

The organ- and effective doses calculated in this study with SimDoseCT for a 160mm-volumetric acquisition were compared with results obtained from ImPACT and ImpactDose CT patient dosimetry calculators**.** Table 6 shows the absorbed doses in the eight organs that received the highest dose in each examination (head, thorax and abdomen) and the effective dose. The organ doses from SimDoseCT and ImpactDose are presented as an average for males

and females. The relative differences between doses obtained from the different tools are also indicated.

Table 6. Absorbed doses in some organs and effective dose, averaged for males and females, obtained from SimDoseCT, ImPACT and ImpactDose dosimetry calculators for a head, thorax and abdomen examination. The relative differences on dose were also calculated.

	SimDoseCT	ImPACT		ImpactDose	
	Dose (mGy or mSv)	Dose (mGy or mSv)	Δ (%)[a]	Dose (mGy or mSv)	Δ (%)[a]
HEAD					
Eye lense	6.34	6.30	0.6	n.a[b]	n.a[b]
Brain	4.35	6.70	-54.2	4.53	-4.2
Extra thoracic region	2.49	0.48	80.7	n.a[b]	n.a[b]
Bone surface	2.26	1.90	15.7	2.36	-4.6
Salivary glands	2.24	6.70	-199.8	n.a[b]	n.a[b]
Oral mucosa	0.96	6.70	-601.6	2.03	-112.2
Red Bone Marrow	0.24	0.48	-100.0	0.19	20.0
Skin	0.44	0.48	-9.1	0.33	24.1
Effective Dose	0.17	0.31	-82.4	n.a[b]	n.a[b]
THORAX					
Breast	12.32	9.90	19.6	n.a[b]	n.a[b]
Heart	12.07	10.00	17.1	11.95	1.0
Lung	9.59	9.30	3.0	9.91	-3.3
Oesophagus	5.74	12.00	-109.1	6.34	-10.4
Liver	5.24	1.20	77.1	n.a[b]	n.a[b]
Spleen	5.17	0.89	82.8	7.24	-40.0
Stomach	4.59	0.77	83.2	5.42	-18.1
Thymus	4.39	12.00	-173.7	n.a[b]	n.a[b]
Effective Dose	4.39	3.50	20.2	n.a[b]	n.a[b]
ABDOMEN					
Stomach	7.15	4.80	32.9	8.02	-12.2
Liver	7.14	5.60	21.6	n.a[b]	n.a[b]
Spleen	6.92	5.40	21.9	6.67	3.5
Gall bladder	6.06	2.60	57.1	7.67	-26.6
Pancreas	5.44	6.50	-19.6	n.a[b]	n.a[b]
Adrenals	5.31	8.10	-52.5	n.a[b]	n.a[b]
Kidneys	4.62	3.30	28.5	6.21	-34.6
Heart	3.71	6.50	-75.2	3.19	13.9
Effective Dose	2.35	2.00	14.9	n.a[b]	n.a[b]

[a] $\Delta(\%) = \frac{\text{Dose}_{\text{ImPACT/ImpactDose}} - \text{Dose}_{\text{SimDoseCT}}}{\text{Dose}_{\text{SimDoseCT}}} \times 100$

[b] Not available in Demo mode

The results showed that SimDoseCT doses in organs directly irradiated agreed with the values for ImpactDose within 13% such as brain in head examination, heart or lung in thorax study and

stomach or spleen in abdomen protocol. However, significant discrepancies were observed between doses in organs partially irradiated due to their location at the border of the scan range. This was the case of oral mucosa in head study, spleen in thorax scan and kidneys in abdomen examination. When results from SimDoseCT were compared with those from ImPACT, effective dose and absorbed dose in most organs showed large differences (between 15 and 85%) and even the dose was between two and six times higher in some organs, like salivary glands or oral mucosa in head scan and oesophagus and thymus in thorax examination.

4. Discussion

A software tool, called SimDoseCT, has been developed for dose assessment in computed tomography based on Monte Carlo simulation of the radiation distribution for the Aquilion ONE 320 detector-row cone-beam scanner, in combination with two ICRP computational adult phantoms, one male and one female. The program consists of a graphical user interface, similar to the interface of the console of the Aquilion ONE scanner, and retrieves data from 378504 look-up-tables. With the software organ doses and effective dose can be accurately calculated for any acquisition protocol of the Aquilion ONE CT scanner. MC simulation was validated by comparing simulated and measured CTDI values for the CT dose phantoms using all the available collimations in the scanner. Additionally, a comparison between dose estimation from SimDoseCT and dose results obtained from dedicated MC simulations was performed to validate the methodology applied in the software.

The CTDI values obtained by simulations were on good agreement with those measured and provided by the manufacturer. The differences which were within 9% could be due to uncertainty in measurements and in the implementation of the scanner model for the simulation. Similar discrepancies were published in other studies (Jarry *et al.* 2003, DeMarco *et al.* 2005, Deak *et al.* 2008, Gu *et al.* 2009, Morant *et al.* 2012, Salvadó *et al.* 2015).

Good correspondence was found between the effective doses calculated using SimDoseCT and the results obtained from dedicated MC simulations. As can be observed in Table 5, the relative differences were 0.1 mSv or less for the absolute dose values. No relationship was observed between these small deviations and any beam width or method used to estimate the dose. Since the scan range selected in AF for thorax and abdomen protocols implied a tube position with interpolated dose data from different look-up-tables in SimDoseCT, most differences in Table 5 can be related to deviations caused by this interpolation.

A similar pattern was found for organ doses. The excellent agreement between SimDoseCT and dedicated MC simulations for acquisitions for which the look-up-tables were stored in the

database demonstrated the correct general operation of the program. For acquisitions that required the use of calculation methods using the available look-up-tables, the minor differences in organ dose estimations demonstrated the good methodology of calculation employed by SimDoseCT. The organs that received the higher absorbed doses in each protocol, which were brain in head studies, breast, lung and heart in thorax examinations and stomach, liver and spleen in abdomen protocols, showed dose differences below 6% for collimations above 20 mm. The dose differences exceeding the 10% corresponded to the organs that absorbed a very low dose (< 0.5 mSv), representing absolute dose deviations of only 0.1 mSv. This percentage was also surpassed in organs with very low mass but within the direct beam, such as eye lenses and thymus, or organs partially irradiated in the direct beam, such as extra thoracic region or salivary glands in head protocols and breast in abdomen protocols. In these cases, doses from SimDoseCT were always higher than results from MC simulations.

The results presented in Figure 3 for the helical abdomen scan further support the good concordance between results from SimDoseCT and MC program. This protocol implies the use of several calculation methods in SimDoseCT, i.e. beam width correction (method 2) and helical correction for both phantoms, and also interpolation in AF due to the tube position. Consequently, the organ dose differences obtained in this case, below 5.5%, give an idea of the excellent consistency of the developed dose assessment methodology.

For the abdomen protocol in helical mode, it is important to take notice of the high doses received by the patient in comparison to the doses for a volume scan in the same range (160 mm-volume abdomen acquisition). The quotient between both effective doses were 1.7 and 1.8 for AM and AF respectively. This increase is due to the overlap caused by the pitch, the use of a narrower beam width (and the corresponding overbeaming) and the overranging, which was also implemented in the program for the helical mode. In this case, overranging resulted in an increase of 27.4% on sex-average effective dose and around 33% as average on organ doses. These results were in agreement with studies from other authors, who affirm that overranging may cause excess dose up to 30% (Kalender 2014). Regarding to the overbeaming, other authors (McNitt-Gray 2002, Smith *et al.* 2007) stated that narrower degree of collimation results in a greater penumbral effect, more overbeaming, and, therefore, an increase on the $CTDI_w$ values by as much as 55% in a head phantom and 65% in a body phantom with the higher doses coming when narrower beam collimation is used. In the present study, MC simulations for the abdomen examination confirmed that differences on effective dose due to overbeaming were 5.8% when a collimation of 160 mm was used and 64.7% for the narrowest beam width (i.e. 2mm-collimation). Similar differences were found in organ doses, which were as average 7.5% and 64.8%, respectively.

Dose results from SimDoseCT showed relevant differences between AM and AF. The effective dose for female was between 1 and 2 times higher than that for adult male in head and abdomen examinations. These differences increased in thorax protocols, in which the ratio between effective dose in female and male was between 2.5 and 4 depending on collimation. Similar tendencies can be observed in organ doses, which were slightly higher in female than in male. This can be mainly explained because the smaller size of the women compared to men makes higher the absorbed dose in most organs or tissues under the same exposure conditions. It has to be also noted that no tissue-weighting factor for breast in men was used, while a weighting factor of 0.24 was used for breast in women. The SimDoseCT results for the collimation of 160 mm in volumetric mode were in agreement with those presented in previous studies (Salvadó *et al.* 2015, Geleijns *et al.* 2015, Cros *et al.* 2016) for head, thorax and abdomen protocols.

Several dosimetric software tools to report patient doses in CT examinations are available today. However, results obtained from comparison between organ- and effective doses obtained from SimDoseCT and with other dosimetry calculators showed significant discrepancies on dose due to the use of outdated scanner models and stylized phantoms. From results presented in Table 6, dose differences up to 40 % can be associated to the low specificity of the generic scanners available in most dosimetric software tools. The differences became larger in the comparison with ImPACT calculator because further than an unspecified scanner, hermaphrodite mathematical phantom is used. These results are consistent with those reported by other authors who confirm the significant differences due to anatomical variations between phantoms and deviations between scanner models (Ban *et al.* 2011b, Gu *et al.* 2013, Ding *et al.* 2015, Jansen and Shrimpton 2016). The results demonstrated that SimDoseCT represents a step in increasing accuracy in assessing doses from CT examinations with a 320 detector-row cone-beam scanner as it takes into account the specific technical characteristics about the CT scanner.

SimDoseCT extends the dose data that is shown on the clinical CT scanner console, being $CTDI_{vol}$ and DLP, with organ- and tissue doses and effective dose. With SimDoseCT the goal was to allow for dose assessment that is tailor made for the Aquilion ONE scanner and that is based on the ICRP computational phantoms. Of course the methodology can also be adapted to other CT scanners models. There are some limitations that must be taken into account. Firstly, the dose data provided by SimDoseCT apply only to standard adult patients, with no dose assessment in paediatric patients or patient with a physique that is not standard. Secondly, the angular and longitudinal tube current modulation employed by the scanner is not implemented in the program. However, the software presents a wide flexibility to overcome these limitations. Previously published studies (Salvadó *et al.* 2005, DeMarco *et al*. 2007) demonstrated that patient size has a significant impact on both effective dose and organ doses, leading to

differences more than a factor of 2. In future research, size-specific dose estimates (SSDE) can be considered by including conversion factors in the database to calculate doses from $CTDI_{vol}$, such as conversion factors tabulated in AAPM Report 204 (AAPM 2011). Regarding to the tube current modulation, the tool could be adjusted to take into account a set of tube current values (mAs) depending on the z-tube position (longitudinal modulation). Unfortunately, the implementation of the x-y tube current variation (angular modulation) could introduce an uncertainty since the mAs per angle information is not provided in the DICOM files by the CT scanner. In summary, the current version of SimDoseCT becomes a suitable tool to estimate the organ- and effective doses in patients for Aquilion ONE CT examinations.

5. Conclusion

Software based on MC simulation for dose assessment in a 320 detector-row cone-beam CT scanner and ICRP adult phantoms was developed and validated. Through an interface, the user can select the acquisition parameters for any available protocol in the Aquilion ONE CT scanner to obtain the corresponding organ- and effective doses. MC code was validated by comparing dose measurements within standard CT dose phantoms with results from simulation. The good agreement between dose results from SimDoseCT for head, thorax and abdomen CT examinations and results from dedicated MC simulations for the same protocols demonstrates the accurate methodology of SimDoseCT and its usefulness and clinical application.

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