Design of a Monte Carlo model based on dual-source computed
tomography (DSCT) scanners for dose and image quality assessment
using the Monte Carlo N-Particle (MCNP5) code

Stefania CHANTZI1,a, Emmanouil PAPANASTASIOU1, Christina ATHANASOPOULOU1,
Elisavet MOLYVDA-ATHANASOPOULOU1, Panagiotis BAMIDIS1, Anastasios SIOUNTAS1
1Medical Physics Laboratory, School of Medicine, Faculty of Health Sciences, Aristotle University of Thessaloniki, Greece
\textsuperscript{a}E-mail address: chanstef@auth.gr

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Abstract
The purpose of this work was to develop and validate a Monte Carlo model for a Dual Source Computed Tomography
(DSCT) scanner based on the Monte Carlo N-particle radiation transport computer code (MCNP5). The geometry of the
Siemens Somatom Definition CT scanner was modeled, taking into consideration the x-ray spectrum, bowtie filter,
collimator, and detector system. The accuracy of the simulation from the dosimetry point of view was tested by
calculating the Computed Tomography Dose Index (CTDI) values. Furthermore, typical quality assurance phantoms
were modeled in order to assess the imaging aspects of the simulation. Simulated projection data were processed, using
the MATLAB software, in order to reconstruct slices, using a Filtered Back Projection algorithm. CTDI, image noise,
CT-number linearity, spatial and low contrast resolution were calculated using the simulated test phantoms. The results
were compared using several published values including IMPACT, NIST and actual measurements. Bowtie filter shapes
are in agreement with those theoretically expected. Results show that low contrast and spatial resolution are comparable
with expected ones, taking into consideration the relatively limited number of events used for the simulation. The
differences between simulated and nominal CT-number values were small. The present attempt to simulate a DSCT
scanner could provide a powerful tool for dose assessment and support the training of clinical scientists in the imaging
performance characteristics of Computed Tomography scanners.

Key words: Monte Carlo simulation; dual-source CT; dual-energy CT; CTDI; image quality.

Introduction
Computed tomography (CT) is a valuable diagnostic tool used
in modern health care. Due to the rising concerns about
radiation exposure, every effort must be made to ensure that
CT examinations are performed under optimum conditions, in
order to obtain the necessary diagnostic information, while
keeping radiation dose to the patient as low as reasonably
achievable (ALARA).

A number of technical innovations have been introduced
over the last years to meet that challenge (Automatic Exposure
Control system, kVp switching, Adaptive Dose Shield and
beam filtration). In 2004, the introduction of z-Flying Focal
Spot (z-FFS) contributed to the improvement of the spatial
resolution and, hence, diagnostic accuracy. The FFS allows for
a deflection of the focal spot both in the rotation direction (α-
FFS) and in the z-direction (z-FFS), thus doubling the
sampling density [1].

The challenge to improve temporal resolution remained and
it was met by the introduction of the Dual Source CT Scanner
(DSCT). This system has two X-ray tubes and two arrays of
detectors. The acquisition of two projections for each angle of
the gantry, one from the low-energy tube and the other from
the high-energy tube, improves image quality without
increasing dose [2].

Monte Carlo (MC) methods have been used to model a CT
system in order to help evaluate the impact of the various
parameters to image quality and estimate the absorbed dose
according to different examination protocols.

Most of the MC simulation studies focus on comparing
measured and simulated organ absorbed doses from CT helical
and axial scans [3,4]. More specifically, Jarry et al [5]
simulated a Multi-Detector CT (MDCT) scanner using the
MCNP code. In their study, the x-ray source and phantoms
were accurately modeled, for the estimation of the radiation
dose. A complete MC simulation of a single source, single
detector-row CT scanner was carried out by Ay and Zaidi [6]
providing images from different phantoms. Kyriakou and
Kalender [7] investigated the scatter for a DSCT scanner,
simulating the geometry system without the z-FFS technique.

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Wysocka-Rabin et al, in 2011 [8] and Qamhiyeh et al [9], developed a Monte Carlo model for the Siemens SOMATOM Emotion CT scanner, with MC code BEAMnrc/EGSnrc, for producing CT images. This study was later extended to calculate CT numbers with accuracy.

Until recently, an accurate Monte Carlo simulator of a DSCT was not available in the literature. Abadi et al. [10] introduce a realistic CT simulation platform that is compatible with high-resolution 3D voxel-based computational phantoms, accounts for the geometry and physics of a given commercial CT scanner.

The aim of the present study was to create and validate a simulation code of a particular DSCT scanner using image quality parameters measured on simulated phantoms.

Initially, an MC simulator of the scanner was developed using the software package MCNP5 [11]. All the elements of the CT scanner, i.e. x-ray tube, bowtie filter, detector array were thoroughly included in the simulation. Then, four different phantoms were simulated and scanned using the simulated scanner, in order to investigate image quality parameters such as image noise, CT-number linearity, high contrast and low contrast resolution. The projections generated by the simulation were input into a reconstruction algorithm created using the MatLab software (MathWorksInc, Natick, MA, USA), for producing transaxial images of the phantoms.

Materials and methods

DSCT scanners

The dual-source scanner that was simulated using MCNP was the Siemens Somatom Definition (Siemens Healthineers, Erlangen, Germany), which is equipped with two separate Straton X-ray tubes. Two slightly different versions of the scanner were simulated (Somatom Definition Flash and Somatom Definition AS) the technical specifications of which are presented in Table 1 [12-14].

| Parameter                      | Dual-source CT Siemens/Definition AS 40 | Dual-source CT Siemens/Definition Flash |
|--------------------------------|----------------------------------------|----------------------------------------|
| X-ray tube                     | 2 Straton MX P                          | 2 Straton MX P                          |
| Anode angle                    | 7°                                     | 7°                                     |
| Detector arrays                | 2                                      | 2                                      |
| detector material              | GOS (Gd₂O₃S)                           | GOS (Gd₂O₃S)                           |
| z-flying focal spot            | Yes, in both tubes                      | Yes, in both tubes                      |
| Number of detector elements per row | 672 (A)                              | 736 (A)                                |
| Field of View                  | 50 cm (A)                              | 50 cm (A)                              |
| Available kV                   | 70, 80, 100, 120, 140                   | 70, 80, 100, 120, 140Sn                |
| maximum                        | 28.8 mm                                | 38.4 mm                                |
| collimation                    | (32x0.6 mm + 8x1.2 mm)                 | (64x0.6 mm)                             |

MCNP5 procedures for DSCT scanner modeling

Simulation of X-Ray Source and X-ray Spectrum

The two different x-ray sources were simulated according to their technical characteristics (Table 1).

One of the key elements in simulating the X-ray source of CT systems is the accurate representation of the x-rays energy spectrum. The energy spectra were calculated using the MCNP5 code. The simulations were run in photon and electron mode (mode: P, E), considering all bremsstrahlung and characteristic x-ray production during electron transport. In the input file, an electron source was defined as a surface source. The anode was a tungsten plate with an angle of 7°. The focal spot size on the target was 0.7mm. The filter was placed in front of the exiting beam [13,14]. The F1 tally was used for calculating the energy spectrum [11]. MCNP5 tally outputs for the calculated spectrum were normalized to the total number of photons in the spectrum.

To validate our X-ray tube model, we compared the MC calculated photon spectra with energy spectral distributions of x-ray source produced using the public software from Siemens website [15] to achieve good agreement. The 80 kV, 100 kV, and 120 kV spectra were calculated with an anode angle of 7° and a filtration of 3.0 mm Al and 0.9 mm Titanium. The 140 kV spectrum was calculated with an anode angle of 7° and a filtration of 3.0 mm Al, 0.9 mm Titanium and 0.4 mm Sn [16,17].

The Siemens Somatom Definition system is equipped with two Straton X-ray tubes which have an electromagnetic beam deflection system for the focal spot [18]. Flying Focal Spot (FFS) was simulated by defining two separate points in the z–direction at the location of the x-ray tube and by defining a “cookie cutter” cell in order to limit the direction of particles to a fan-angle covering the same detector elements in the z–direction. Considering the slice thickness S, the sampling distance at isocenter was S/2 [1]. The z–FFS was applied in Computed Tomography Dose Index (CTDI) measurements.

Simulation of beam-shaping filter

The scanner external filtration consists of the narrow (head) or standard (trunk) filter, which is used to reduce the dose across the lateral parts of the body. Due to the unwillingness of the manufacturer to disclose their exact technical specifications (geometrical characteristics) the beam-shaping filters of the Somatom Definition were modeled using an indirect method: A single-source scanner was simulated with a beam-shaping filter created using the simplified shape of the basic Teflon bowtie filter as described by DeMarco et al [19] (Figure 1). A standard PMMA CTDI phantom was also simulated and was centered at the scanner isocenter. Changing the shape of the bowtie filter, namely the angle θ of the trajectory of a particle that does not enter the CTDI phantom, affected the MCNP5 calculated CTDI values at the center and the periphery of the CTDI phantom. Through trial and error, the angle θ was finally
selected as the one where the CTDI values coincided with published data [19, 20].

**Simulation of the detectors**

The detector elements were simulated across an arc of a circle with a diameter equal to the source-to-detector distance. In Somatom Definition AS, detector A consists of 672x40 elements. Due to the MCNP5 code limitations, the 672x40 elements were simulated in 21 lattices of 32 columns in the x-y direction and 40 rows in the z-direction. Detector B consists of 352x40 elements, which were simulated in 11 lattices of 32 columns in the x-y direction and 40 rows in the z-direction. In Somatom Definition Flash, the 736x64 elements were simulated in 23 lattices of 32 columns in the x-y direction and 64 rows in the z-direction. Detector B consists of 480x64 elements, which were simulated in 15 lattices of 32 columns in the x-y direction and 64 rows in the z-direction. For both scanners, detector material was gadolinium oxysulfide (GOS) with a density of 7.44 g cm$^{-3}$ (Figure 2).

**CTDI test phantoms**

In this study, body and head CTDI dosimetry phantoms were used. Both phantoms were made of polymethyl-methacrylate (PMMA) with a density of 1.19 g cm$^{-3}$ and had a length of 15 cm. The diameter of the head phantom was 16 cm and of the body phantom 32 cm. Each phantom incorporated five air-filled holes in which pencil ion chambers were placed. The ion chamber was modeled as a set of four concentric cylinders with a length of 100 mm with C552 air-equivalent walls and electrode, polyacetal exterior cap, and a 3 cm$^3$ active volume [21,22].

**Image quality test phantoms**

Four image quality test phantoms were simulated and were used to produce test images in order to validate the CT scanner simulation.

**Homogeneous water phantom.** A simple homogeneous cylindrical (30 cm in diameter, 6 cm long) water phantom was simulated, in order to measure image noise in a reconstructed transverse slice (Figure 3a).

**Low contrast phantom.** A cylindrical (16 cm in diameter, 6 cm long) water phantom was initially simulated (Figure 3b). Two semi-cylindrical blocks of PMMA, one 2 mm thick and the other 20 mm thick, were then inserted inside the water phantom. Each of these blocks had four holes, measuring 1.5 cm, 1.0 cm, 0.5 cm and 0.2 cm in diameter, which were filled with water. Scanning this phantom with 10 mm slices would produce low contrast regions in the thin semi-cylindrical block.

**High contrast phantom.** A cylindrical (16 cm in diameter, 6 cm long) PMMA phantom was simulated (Figure 3c) with six sets of air thru-holes (five holes per set). Diameter of holes is 2.0, 1.4, 1.2, 1.0, 0.6, and 0.4 mm. The distance between two consecutive holes in each set was equal to the respective hole diameter.

![Figure 1. Geometry of basic beam-shaping filter.](image1)

![Figure 2. The DSCT detector elements, as designed with the MCNP code.](image2)

![Figure 3. Schematic representations of (a) the homogeneous water phantom, (b) the low contrast resolution phantom, (c) the high contrast resolution phantom and (d) the CT number linearity phantom.](image3)
Water phantom with inserts of different materials. A cylindrical (16 cm in diameter, 6 cm long) water phantom was initially simulated (Figure 3d). Four cylindrical blocks (2 cm in diameter, 1 cm long) of different materials (air, PMMA, polyethylene and teflon) were then inserted towards the periphery of the water phantom.

Test object validation and Monte Carlo simulation aspects

The F6 tally was used for calculating energy depositions [11]. MCNP5 tally outputs for dose are in units of MeV/g per source particle. In order to convert the results to a more meaningful dose quantity (mGy/mAs), a CTDI method was applied. The method of CTDI estimation used in this study is similar to previous works by Jarry et al [5] and DeMarco et al. [19].

For CTDI calculations (free-in-air, in head phantom and in body phantom), a single axial 360° scan was simulated by a rotating source placed in a circle with a radius equal to the distance from the focal spot to isocenter. The rotation was performed in discrete 5° angular steps. The MC calculations with only an ionization chamber (IC) at isocenter, were used to compute the normalization factor. The normalization factor described previously is calculated by the following equation:

\[
(NF)_E = \frac{CTDI_{\text{[100,air,measured, per, 100 mAs]}}}{CTDI_{\text{[100,air,simulated, per, particle]}}} \tag{1}
\]

where \(CTDI_{\text{[100,air,measured, per, 100 mAs]}}\) is the air kerma per 100 mAs at the scanner isocenter given by IMPACT [12] for a given beam energy \(E\) and a collimator width and \(CTDI_{\text{[100,air,simulated, per, particle]}}\) is the Monte Carlo calculated air kerma per particle by simulating the ion chamber at the scanner isocenter for the same scanner settings [5].

The second set of calculations in the body and the head CTDI phantoms was performed under the same technical parameters as with the ionization chamber. The absolute dose at the center and the periphery of body phantom is calculated by the following equation:

\[
D = D_{E,\text{mV}} \cdot CF \tag{2}
\]

where \(D_{E,\text{mV}}\) is the Monte Carlo simulated dose MeV/photon and CF is the normalization factor for a given kV [5].

For the normalization simulations and CTDI calculations, single axial scans were simulated using scan parameters as shown in Table 2. CTDI values in head phantom were calculated and compared with measured values of head phantom scanned at 120 kV (Dual Source mode). CTDI Monte Carlo simulations were made using 2x10^6 photon histories, resulting in tally uncertainties of less than 2% (Figure 4).

For phantom images, an axial (sequential) scan was performed to give reconstructed slices. 360 views were simulated at 1° between each view. The *F8 tally was used for calculating energy deposited on each detector element per tracked particle [11]. Because of MCNP5 does not simulate gantry rotation, the geometry of each view is created in separate files. The initial intensity was calculated, by simulation without phantom in the FOV.

To improve the efficiency of the simulation the variance reduction method was used. The Surface Source Write (SSW) option was used to increase the speed of MC simulations [11] (Figure 5).

Table 2. Typical CT acquisition parameters used in the simulation.

| Scan                | Tube voltage | Beam Shaping filter | Beam collimation | Z-FFS |
|---------------------|--------------|---------------------|------------------|-------|
| single-source head  | 120 kV(A)    | A: narrow           | 18 mm            | On    |
| single-source body  | 120 kV(A)    | A: standard         | 18 mm            | On    |
| dual-source head    | 120 kV(A+B)  | A, B: standard      | 38.4 mm          | On    |

Figure 4. Cross-sectional view of simulation code for dosimetry purposes.

Figure 5. Geometry of surface source.
It should be noted that for the phantom images, the simulations ran separately for each X-ray source-detector system, therefore no contributions from any scattered radiation from the first X-ray source to the second detector were calculated. This was necessary in order to keep the required computation time for the simulation to acceptable levels.

Since it is known that reducing the mAs is expected to increase the noise (measured standard deviation) by \(1/\sqrt{mAs}\), single axial scans of 30 cm-diameter water phantom were simulated for single source energy 120 kV using the 3 mm beam collimation with standard bowtie filter. In order to examine the impact of number of particles on the final image, scans were performed with \(2 \times 10^10\), \(4 \times 10^{10}\), \(8 \times 10^{10}\) and \(16 \times 10^{10}\) particles. An image using a slice thickness of 3 mm was reconstructed.

Evaluation of image quality and validation of the simulation code were performed using a series of phantoms. Physical image quality parameters (image contrast, spatial resolution and noise) were measured in the simulated images of these tests objects and the results were compared with the expected ones. Axial scans were performed at appropriate locations on each phantom and transaxial slices were reconstructed. Both X-ray sources at 100 kVp and 140 kVp were used, without z-FFS. The total number of photons was \(1 \times 10^{10}\). An image using a slice thickness of 10 mm was reconstructed. In all simulations, the statistical uncertainty was less than 2%.

\section*{Image reconstruction}

In computed tomography systems, the most widely used reconstruction method is the filtered back-projection. For this study, a fast and powerful filtered back-projection algorithm for non-helical fan-beam CT setup was implemented. First, a rebinning of the geometry of the fan beam lines into parallel lines was performed using appropriate interpolation. Figure 6 shows the fan-beam and parallel-beam geometry. The relations between the different coordinates are:

\[ \theta = \beta + \gamma \Rightarrow \beta = \theta - \gamma \]  \hspace{1cm} \text{Eq. 3} \\
\[ \rho = D \sin(\gamma) \Rightarrow \gamma = \sin^{-1}(\rho / D) \]  \hspace{1cm} \text{Eq. 4}

Then, the reconstruction algorithm performs a convolution of the parallel projection data, \(P(\rho)\), with a ramp filter, \(H(\omega)\), according to:

\[ Q_\omega = F^{-1}\{F[P(\rho)] \cdot H(\omega)\} \]  \hspace{1cm} \text{Eq. 5}

where \(F\) and \(F^{-1}\) denote the Fourier transform and the inverse Fourier transform respectively [23]. Then back-projection is applied, which means smearing of the filtered projection data over the image plane according to:

\[ f(x, y) = \int_0^\pi Q_\omega(x \cos \theta + y \sin \theta) d\theta \]  \hspace{1cm} \text{Eq. 6}

The input parameters required by the algorithm are the simulated projections, the source to isocenter distance, the detector element size at isocenter, the angular step between projections and the matrix size. The output is an image matrix representing the values of the attenuation coefficients of the imaged object.

Figure 6. a) Geometry of fan-angle beam and b) geometry of parallel-beam.
The following steps were followed for the reconstruction of the images of all test phantoms: First, the data were filtered in the spatial domain. Next, the filtered data were back-projected. During reconstruction, additional filtering was utilized to eliminate ring artifacts in each projection. This filter was applied to projection data before the back-projection algorithm. The method adopted for processing the projection data acquired with a dual-source computed tomography (DSCT) imaging system, comprised of the following steps: a) acquisition of two separate projection data sets, one from each x-ray tube b) insertion of the raw data structure into MATLAB for reconstruction and c) creation of the final image by appropriately weighing the two reconstructed images using the following relation:

\[ f = w \times f_{\text{low}} + (1 - w) \times f_{\text{high}} \]

where \( w \) is the weighting factor, \( f \) denotes the CT value in the mixed image, and \( f_{\text{low}} \) and \( f_{\text{high}} \) are the CT values of the low and high kV image, respectively [13].

For water phantom, projections are smoothed using a local average of the k-nearest neighbors, resulting in decreased noise. The value of \( k \) controls the smoothness. Image noise was evaluated as a standard deviation of Hounsfield unit values in a circular region of interest (ROI), positioned at the water phantom center. Results were expressed as standard deviation (SD) of CT numbers. For low-contrast phantom, the mean CT-numbers of water and PMMA were calculated in a ROI of 12-pixels in diameter.

**Results**

**X-ray spectrum calculations**

The corresponding output photon energy spectra, which were simulated according to the technical characteristics, are shown in Figure 7. The energy spectra provided the probability of specific energy values for the MCNP code. The number of photons relates to the center of each energy interval (1 keV).

The mean x-ray spectrum energies for spectra simulated and spectra obtained from Siemens website [15], for 80, 100, 120, 140 and 140 Sn kV (with an additional 0.4-mm tin filter), are compared in Table 3.

**CTDI dose calculations**

Initially, the calculation was made free-in-air for a single tube potential 120 kVp and a beam collimation 18 mm for both (narrow and standard) beam shaping filters. Table 4 displays the reported values by ImPACT (Imaging Performance Assessment of CT scanners) and the calculated values of the CTDI free-in-air. Table 4, also, summarizes conversions factors obtained by MCNP code and used in order to convert the tally F6:p results of the MC simulation given in units of MeV/g/source particle to absorbed dose in units of mGy/100mAs.

Then, the center and peripheral CTDI\(_{100}\) calculations for the head and body phantoms were obtained under the same conditions as in the free-in-air calculation. All simulation results were normalized to 100 mAs using the conversion factor, which was calculated above. Table 5 presents the reported values by ImPACT (Imaging Performance Assessment of CT scanners) and the results simulated in MCNP code for both beam shaping filters.

![Figure 7. Simulated spectra based on Monte Carlo technique for 80 kV, 100kV, 120 kV and 140kV with 0.9-mm titanium and 3-mm aluminum filter and for 140 kV with 0.9-mm titanium, 3-mm aluminum filter and extra 0.4-mm tin filter.](image)

**Table 3. Mean energies (keV) for simulated spectra versus spectra obtained from Siemens website [15].**

| U (kV) | Siemens | This Study |
|-------|---------|-----------|
| 80    | 52.04   | 52.8      |
| 100   | 58.59   | 59.9      |
| 120   | 63.92   | 65.6      |
| 140   | 68.2    | 70.6      |
| 140Sn | 86.72   | 86.1      |

**Table 4. Conversion factors from MeV/g/particle to mGy/100mAs, obtained from measurements and simulations for 18 mm collimation and for the single tube potential of 120 kV.**

| Beam-shaping filter | ImPACT CTDI\(_{100}\) in air (mGy/100 mAs) | Simulated CTDI\(_{100}\) in air (MeV/g/particle) | Conversion factor (mGy/g/particle / 100mAs/Mev) |
|---------------------|------------------------------------------|-----------------------------------------------|-----------------------------------------------|
| standard            | 20.7                                    | 1.85x10\(^4\)                               | 1.12x10\(^5\)                                 |
| narrow              | 19.4                                    | 1.86x10\(^4\)                               | 1.04x10\(^5\)                                 |

**Table 5. CTDI\(_{100}\) calculations (mGy) for head and body phantoms, obtained from ImPACT reported values and simulations for 18 mm collimation and for the single tube potential of 120 kV.**

| phantom | position | ImPACT | simulated | difference (%) |
|---------|----------|--------|-----------|----------------|
| body    | CTDI\(_{100}\)\(_{rc}\) | 4.8    | 4.69      | -2.29          |
|         | CTDI\(_{100}\)\(_{rcg}\) | 8.5    | 8.90      | 4.70           |
| head    | CTDI\(_{100}\)\(_{rc}\) | 14.0   | 14.01     | 0.07           |
|         | CTDI\(_{100}\)\(_{rcg}\) | 13.3   | 13.52     | 1.65           |
Then, the calculation was made using dual-source dual-energy mode (both tubes at 120kV) and beam collimation 38.4 mm. Table 6 displays the experimental values obtained by the measurement of CTDI free-in-air and those simulated in MCNP code. Table 6, also, shows the conversion factor obtained by MCNP code and used in order to convert the tally F6:p results of the MC simulation given in units of MeV/g/source particle to absorbed dose in units of mGy/100mAs for dual-source dual-energy mode.

In Table 7, the center and peripheral CTDI\textsubscript{100} calculations for the head phantom were obtained under the same conditions as in the free-in-air measurement are shown. All simulation results were normalized to 100 mAs using the conversion factor, which was calculated above. Table 7 presents the experimental results and the results simulated in MCNP code for narrow beam shaping filter.

**Validation of simulation results using image quality**

**Image noise**

Figure 8 shows the simulated cylindrical water-filled phantom profiles, using MCNP5 based CT simulator. Simulated profiles were divided by the corresponding blank scan and were normalized at the central detector element.

Table 8 shows the results of standard deviation (SD in circular ROI) of CT numbers in a simulated reconstructed image from the water phantom, obtained with the traditional FBP reconstruction algorithm (3 mm slice thickness) at single tube potential 120 kV, when the simulation took into account a quantity of 2x10\textsuperscript{10}, 4x10\textsuperscript{10}, 8x10\textsuperscript{10} and 16x10\textsuperscript{10} photons per projection. The image noise was inversely correlated to the number of particles.

Table 6. Conversion factors from MeV/g.particle to mGy/100mAs, obtained from measurements and simulations for the dual-source and 38.4 mm beam collimation.

|                | dual-source dual-energy (kV) | measured CTDI\textsubscript{100} in air (mGy/100 mAs) | Simulated CTDI\textsubscript{100} in air (MeV/g.particle) | Conversion factor (mGy·g·particle / 100mAs.MeV) |
|----------------|-------------------------------|--------------------------------------------------------|----------------------------------------------------------|-------------------------------------------------|
|                | 120                           | 10.96                                                  | 2.34x10\textsuperscript{7}                                | 4.67x10\textsuperscript{7}                       |

Table 7. CTDI\textsubscript{100} calculations for head phantom, obtained from measurements and simulations for the dual-source mode and 38.4 mm beam collimation with z-FFS.

|                | Dual-source (kV) | position | Measured CTDI\textsubscript{100} (mGy) | Simulated CTDI\textsubscript{100} (mGy) | difference (%) |
|----------------|-----------------|----------|---------------------------------------|---------------------------------------|---------------|
|                | 120             | CTDI\textsubscript{100} | 8.096 | 8.020 | -0.94 |
|                | 120             | CTDI\textsubscript{100P} | 8.941 | 8.599 | -3.82 |

Table 8. Image noise (SD in circular ROI) calculated in simulated reconstructed image from the water phantom, obtained with the traditional FBP reconstruction algorithm (3 mm slice thickness) at single tube potential 120 kV.

| Number of photons per projection | 2x10\textsuperscript{10} | 4x10\textsuperscript{10} | 8x10\textsuperscript{10} | 16x10\textsuperscript{10} |
|---------------------------------|---------------------------|---------------------------|--------------------------|--------------------------|
| SD (HU)                         | 38.67                     | 30.16                     | 20.36                    | 14.81                    |
Table 9. Simulated CT numbers of water and PMMA in the low contrast phantom.

| Material          | 12 pixel diameter ROI | 20 mm-semicylindrical block | 2 mm-semicylindrical block |
|-------------------|-----------------------|-------------------------------|----------------------------|
| water cylinder (1.5 cm) | -2.78 | -0.83 | |
| PMMA              | 108.16               | 12.41                         |                            |

Spatial resolution

Figure 9c shows the simulated image obtained from the high-contrast resolution phantom using a filtered backprojection algorithm and a slice thickness of 10 mm. X-ray sources were at 100 kVp and 140 kVp, without z-FFS. The total number of photons was $1 \times 10^{10}$. In this image, the 3th group of air-holes (1.2 mm in diameter) was marginally discernible.

CT number linearity

Figure 9d shows the simulated image obtained from the CT number linearity phantom using a filtered backprojection algorithm and a slice thickness of 10 mm. X-ray sources were at 100 kVp and 140 kVp, without z-FFS. The total number of photons was $1 \times 10^{10}$. The simulated image is the mixed image of the low 100 kV image (50%) and the high 100kV image, which is equivalent to a 120 kV image. Table 10 summarizes the calculated CT numbers of the four different materials in the simulated image and the nominal values obtained from NIST [24].

Discussion

Monte Carlo techniques have proved to be a powerful tool for the simulation of the construction and the performance of CT scanners as well as for dose assessment in clinical procedures. The used number of photons in all the simulations corresponded to a very low mAs value, which had an important impact on our results. Time limitations imposed by the MC method don’t permit the usage of a larger number.

In the present study, Somatom Definition CT scanner was simulated taking into consideration both structural and functional characteristics. Unavoidably, a few approximations regarding the geometry of the bowtie filter and the detectors were included. An equivalent source model was developed for the simulation of the z-FFS.

A careful CTDI validation was performed and the simulation results demonstrate good agreement with the expected data, as reported in the literature and the actual measurements. More specifically, the small discrepancy observed in CTDI$_{100}$ values for the body phantom can be explained by apparent (unavoidable) differences between the exact technical specifications of the scanner and those used in the simulation code. This discrepancy in the body phantom is smaller at the center than at the periphery, which is close to -2.25%. In previously published works that performed Monte Carlo simulations for CTDI estimation, the discrepancies between measured and simulated values varied between -2.6 to 8.6% [5,19- 21].

Furthermore, additional proof of the accuracy of the simulation code could be the agreement between the simulated CT numbers and the nominal values (0 HU for water). We found a mean value of -8.78 HU and an SD 17.19 HU. The high value of image noise is due to the small number of photons used for the simulation ($1 \times 10^{10}$ for each projection angle) in order to keep the simulation run-time at acceptable levels.

The dependence of image noise on the number of particles, as illustrated in Table 8, agrees with ~1/$\sqrt{2}$ expected reduction. Doubling the number of particles results in the reduction of image noise by a factor 1/$\sqrt{2}$. 
Only one portion of the low-contrast phantom, with the 2 mm semi-cylindrical PMMA block can be used to assess low contrast resolution, using a 10 mm slice thickness and taking advantage of the partial volume effect. Using a 12-pixels in diameter ROI inside the largest (1.5 cm in diameter) water cylinder, the measured CT number was found -0.83 HU, whereas the PMMA CT number in a similar ROI was found 12.41 HU. The corresponding measurements in the other portion of the phantom yielded values of -2.78 HU for water and 108.16 HU for PMMA.

In the high-contrast resolution phantom, the 3rd set of air-filled holes (1.2 mm) was visible, which is a logical value, taking into consideration the small number of photons used for the simulation (which correspond to very few mAs) and the smoothing filters applied during the reconstruction.

The deviations of the calculated CT numbers of each material from those reported by NIST were acceptable. Furthermore, in the CT number linearity test object, some artifacts are present, which could affect the accuracy of the ROI CT number measurements. These artifacts may occur due to the simulated geometry of detectors and the small number of photons used. The CT numbers estimated in the present study are very close to those reported by Gulliksrud et al [25] who performed measurements of the CATPHAN phantom. The Teflon CT number in the present simulation (976 HU) is also in agreement with the calculated value of 970 HU reported by Sharma et al [26] using one single source at 130 kV. Our calculated CT number of PMMA is 10% (12 HU) lower than the nominal value of 120 HU. CT number of air (-967) is well within the acceptable range from -960 to -994 HU, for CT scanners of different manufacturers [26].

There were, however, some limitations to this study, arising primarily from the lack of the exact technical specifications of the DSCT scanner, including the newer iterative reconstruction algorithms (SAFIRE) available on Siemens CT scanners. Image reconstruction using filtered backprojection is generally inferior to iterative reconstruction, which can offer lower image noise and better low contrast resolution. Another limitation of our filtered backprojection reconstruction is that it cannot be used with helical scan protocols. Also, the influence of cross scatter has not been considered for the DSCT system evaluated in this study, because the simulations of each X-ray source-detector system for investigation of image quality parameters run at separates files. So, the scattered radiation produced from tubes was not evaluated, using MC code.

### Conclusion

This study presents a method for modeling a DSCT with z-FFS. The reported results validate the modeling and MCNP code. Therefore, the present simulation could be extended to include more CT scanning protocols and computational anthropomorphic phantoms, in order to provide patient dosimetric information with reasonable accuracy. Work in progress includes Monte Carlo simulations with more image quality phantoms, which would further validate the code and could be used for the development of educational e-learning tools for medical physicists and trainee radiologists.

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