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The development, validation and application of a multi-detector CT (MDCT) scanner model for assessing organ doses to the pregnant patient and the fetus using Monte Carlo simulations

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Abstract
The latest multiple-detector technologies have further increased the popularity of x-ray CT as a diagnostic imaging modality. There is a continuing need to assess the potential radiation risk associated with such rapidly evolving multi-detector CT (MDCT) modalities and scanning protocols. This need can be met by the use of CT source models that are integrated with patient computational phantoms for organ dose calculations. Based on this purpose, this work developed and validated an MDCT scanner using the Monte Carlo method, and meanwhile the pregnant patient phantoms were integrated into the MDCT scanner model for assessment of the dose to the fetus as well as doses to the organs or tissues of the pregnant patient phantom. A Monte Carlo code, MCNPX, was used to simulate the x-ray source including the energy spectrum, filter and scan trajectory. Detailed CT scanner components were specified using an iterative trial-and-error procedure for a GE LightSpeed CT scanner. The scanner model was validated by comparing simulated results against measured CTDI values and dose profiles reported in the literature. The source movement along the helical trajectory was simulated using the pitch of 0.9375 and 1.375, respectively. The validated scanner model was then integrated with phantoms of a pregnant patient in three different gestational periods to calculate organ doses. It was found that the dose to the fetus of the 3 month pregnant patient phantom was 0.13 mGy/100 mAs and 0.57 mGy/100 mAs from the chest and kidney scan, respectively. For the chest scan of the 6 month patient phantom and the 9 month patient phantom, the fetal doses were 0.21 mGy/100 mAs and 0.26 mGy/100 mAs, respectively. The paper also discusses how these fetal dose values can be used to evaluate imaging procedures and to assess risk

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using recommendations of the report from AAPM Task Group 36. This work demonstrates the ability of modeling and validating an MDCT scanner by the Monte Carlo method, as well as assessing fetal and organ doses by combining the MDCT scanner model and the pregnant patient phantom.

(Some figures in this article are in colour only in the electronic version)

1. Introduction


Among various patient populations, children and pregnant females are particularly important due to their relatively greater radiosensitivity. CT scans of the maternal body are often used to identify suspected pulmonary embolus (Remy-Jardin and Remy 1999), appendicitis (Castro et al 2001, Paulson et al 2003), renal colic (Chen et al 1999) or trauma during pregnancy in some situations. The majority of pregnant patient CT examinations are performed in areas away from the uterus, so the fetus is not directly exposed to radiation. However, low levels of radiation from scattering still may carry a small risk of harm to a developing fetus. Accidental scans of the female patient whose pregnancy was unknown have also been reported (Shi and Xu 2004). When administering CT scans of pregnant patients, the dose to the fetus, which is much more radiosensitive than the mother, must be considered. Recent reports by Boiselle et al (1998), Ratnapalan et al (2004) and Schuster et al (2003) regarding physician standards of practice for pregnant patient CT imaging provided conflicting opinions on the proper use and associated risks of CT imaging techniques (Boiselle et al 1998, Ratnapalan et al 2004, Schuster et al 2003). Clearly, dosimetry tools that accurately quantify the dose to the mother and fetus would help improve planning and administration of CT scans of the maternal body. Recently, Hurwitz et al reported radiation dose to the fetus from the body MDCT scan during early gestation using physical measurements (Hurwitz et al 2006). Based on this research, ICRP Publication 102 has summarized the typical scanning protocols and radiation doses for scanning used in a CT center for imaging pregnant patients (ICRP-102 2007). Angel et al reported radiation dose to the fetus for pregnant patients undergoing MDCT imaging using Monte Carlo simulations (Angel et al 2008). However, these data are still limited because fetal dose and organ doses of the pregnant patients vary according to various patient phantoms, CT scanner model or CT machine, scan protocols, etc. There is a need to conduct further study on doses to pregnant patients and their fetuses involving MDCT scanners, contemporary clinical CT protocols and different patient phantoms.
Fetal radiation doses from maternal body CT scans have been measured for conventional axial CT scanners and single-detector helical CT scanners (Winer-Muram et al. 2002, Damilakis et al. 2000, Felmlee et al. 1990). Major design differences exist between single-detector CT and MDCT machines, and care must be exercised in developing scanning protocols to minimize the fetal dose (Yoshizumi and Nelson 2003). Radiation doses from MDCT protocols are typically greater than those from single-detector CT due to the design of MDCT protocols (Thomton et al. 2003). Generally, radiation doses to a pregnant patient and fetus can be determined by two different techniques: direct measurement or Monte Carlo simulation. In direct measurement several dosimeters are placed in cavity locations in an anatomically simplified physical phantom and exposed by being scanned in a CT with scan protocol (Hurwitz et al. 2006). Because of limitation induced by scarcity of specific physical phantoms or unavailability of clinical CT modality, sometimes the direct measurement of organ doses is problematic, even impossible. In contrast, Monte Carlo simulations predict radiation dose for any scan protocol using anatomically realistic computational pregnant patient phantoms such as those that have recently become available from RPI and elsewhere (Angel et al. 2008, Xu et al. 2007). Different pregnant patient phantoms may give rise to different doses to the fetuses and organs.

Using Monte Carlo methods, several research groups have simulated the CT sources and assessed doses to adult, pediatric and pregnant patients (Winer-Muram et al. 2002, Khursheed et al. 2002, Deak 2008, DeMarco et al. 2005, Staton et al. 2006). For example, Caracappa (2001) modeled a CT scanner for effective dose calculation using the VIP-Man phantom. Winner-Muram et al. (2002) modeled a helical CT for fetal radiation dose calculation using CTDI data for several different kVp values for validation. Similarly, DeMarco et al. (2005) modeled the multidetector CT (MDCT) and validated it with CTDI values for several kVp values. DeMarco et al. (2005) also used dose profiles for the single axial scan, contiguous axial scan and two helical scans using the CTDI body phantom, as well as the dose profiles for the contiguous axial scan and two helical scans using anthropomorphic phantoms for further validation. However, an insufficient level of detail on the CT parameters was reported in these papers and it has been difficult for others to repeat or to compare results. The first systematic validation of a Monte Carlo model of the MDCT for patient-specific dose simulations was reported by Deak (2008), who aimed to extend an existing voxel-based Monte Carlo tool by accounting for arbitrary scanners and scan protocols. The current study in this paper was motivated by overcoming the limitations imposed by the absence of detailed data and steps of the MDCT scanner model development and validation. Also, this study provides results for a typical example of fetal dose from particular CT examinations, based on the provided energy spectrum in this paper and other scan parameters. Different from the two studies in which Hurwitz et al. (2006) and Angel et al. (2008) provided fetal doses specifically for the scans with the fetus partially or entirely contained within the x-ray beam, this study, being complementary, focused on scans (chest and kidney) that do not have the fetus exposed directly in the x-ray beam. The methodology employed in this research can be used to facilitate the study of MDCT dosimetry involving anatomically realistic computational patient phantoms.

This paper describes the development, validation and application of a Monte Carlo model of a GE LightSpeed 16-MDCT scanner (ImPACT 2004). For validation, CTDI values for several different kVp values as well as dose profiles were used to compare and fine-tune the parameters in the MDCT scanner model. Using the validated MDCT scanner model in combination with a set of pregnant patient phantoms representing 3, 6 and 9 month gestational periods (Xu et al. 2007), radiation doses to the mother and fetus were calculated for selected imaging procedures. Finally, discussion on the fetal dose and radiation risk is presented.
2. Methods and materials

This section systematically describes the steps of an MDCT modeling and validation method. Application of this validated MDCT model to assess the organ doses as well as fetal doses of the pregnant patient phantoms is covered in detail.

Figure 1 illustrates the general workflow of the MDCT modeling, validation and application procedure in this study. It starts with determining MDCT source parameters including the x-ray source energy spectrum, internal filter, external filter (bowtie filter), beam shape, etc. These parts together can be regarded as the MDCT scanner model. Once these parts as well as the CTDI body phantom have been defined, the Monte Carlo simulations obtain the CTDI center dose values (CTDIc), CTDI peripheral dose values (CTDIp) and the dose profiles along the surface of the CTDI body phantom. Then the reference values from the literature or the physical measurements are used to compare with the simulated values in order to validate the modeled MDCT scanner and its movement. When the scanner is validated, the CTDI phantom is replaced by the patient phantom and the Monte Carlo simulations are performed to assess the doses to the patient. In the sections below, we explain each of these steps in detail.
The development, validation and application of a MDCT scanner model

2.1. CT scanner model

All simulations and dose measurements were performed for a GE LightSpeed 16-MDCT. This section describes several important components and parameters of the MDCT scanner used to construct the Monte Carlo model.

2.1.1. CT scanner.

The 16-MDCT scanner modeled here (a LightSpeed 16, General Electric Healthcare Corporation, Waukesha, WI) is a third-generation MDCT, with 16 rows of 0.625 mm wide detectors as well as eight rows of 1.25 mm detectors. It offers the user the following x-ray beam collimation options (given in the format of $N \times T$, where $N$ represents the number of data channels and $T$ represents the nominal width of each data channel; each data channel may represent at least one detector row (NEXT 2006, ImPACT2004)): 16 $\times$ 0.625 mm (8 $\times$ 1.25 mm) and 16 $\times$ 1.25 mm (8 $\times$ 2.5 mm), as well as 2 $\times$ 0.625 mm and 1 $\times$ 5 mm modes. The scanner can operate in both axial and helical modes. For the helical scan, the pitch can be selected as 0.625, 0.875, 1.35 or 1.675 in the 8-channel mode or 0.5625, 0.9375, 1.375 or 1.75 in the 16-channel mode (Khursheed et al 2002). The system supports nominal x-ray energies of 80, 100, 120 and 140 kVp. X-ray beam shaping filtration including both head and body bowtie filters is equipped and used to compensate for the variation in body thickness across the transverse sections of the body, improving the image quality and reducing the dose to the peripheral region of the body. The distance from the focal spot to the isocenter (SID) is 54 cm, and the distance from the focal spot to the detector (SDD) is 95 cm. The fan beam is collimated in the $x$–$y$ plane to a fan angle of 55°.

2.1.2. Source geometry and movement. The x-rays in a CT scanner used for imaging are produced from the interaction of an electron beam in a converting medium known as a target. While direct sampling from the initial electron beam has been done, a more practical approach is to sample from the x-ray source spectrum produced in the target. An x-ray source model was used in this study in order to improve the efficiency of the Monte Carlo simulations. The source model assumes that the photons are emitted from a point at the location of the x-ray tube anode as shown in figure 2. Several tools have been developed to accurately generate the
Figure 3. Photon energy spectrum (after a 2.5 mm aluminum filter) used in this research for the MDCT source modeling. This energy spectrum was generated by Xcomp5r using 120 kVp, 12.0° anode angle, 2.5 mm aluminum flat filter thickness, and 10 cm distance from the point source with 1 keV energy bins in the data output.

CT scanner manufacturers typically share only the tube voltage for specific scan protocols. Information on the anode angle, filter shape and material as well as other details is often needed for research purposes. The parameters used in this study for a tube voltage of 120 kVp include 12.0° of anode angle, 2.50 mm aluminum flat filter thickness and a spectrum distance of 10.0 cm where the spectrum was recorded. The mean photon energy for the spectrum (after the flat filter) was 55.1 keV (see figure 3).

During a CT scan, the x-ray source moves along a single axial, contiguous axial or helical axial trajectory. In this study, a total of 16 x-ray sources situated along a single axial scan trajectory were sampled uniformly for the assessment of CTDI dose values, as shown in figure 4. In order to assure that this x-ray source was modeled sufficiently, a set of dose calculations were also performed using 24 and 32 sources along the CT scan trajectory. In comparison, CTDI dose values from all three source definitions were within 5% of measured CTDI dose values. Similar results were reported by Khursheed et al (2002), in which both 18 sources and 72 sources were tested, and results showed that 18 sources were sufficient to approximate the continuous axial movement of the source, without significantly affecting the calculated organ doses. Therefore, in this study the 16-source definition as shown in figure 4 was used. For contiguous axial and helical axial scans, x-ray sources were situated along the corresponding contiguous axial or helical axial trajectory to achieve the simulation of the source motion, and each rotation of the motion trajectory was filled by 16 sources.
2.1.3. Bowtie filter. Two types of bowtie filters are used for CT scans: head bowtie filters and body bowtie filters. Since the real shape of the bowtie filter is relatively complicated (Toth et al. 2005), a simplified geometry involving a rectangle and an elliptical cutout was used for this study (see figure 2). An initial shape for the elliptical cutout is assumed, and then adjusted through trial-and-error until the average difference between measured and calculated CTDI values (over all kVp values) is less than 5%. The bowtie filter is typically made of Teflon, PMMA or aluminum (Ay and Zaidi 2005, George et al. 2007). For this study, aluminum was selected as the bowtie filter material for all simulations.

2.1.4. Beam shape. A fan beam is generally used for CT diagnostic procedures. To define the specific beam shape, the position of the focus point of the CT x-ray source, the fan angle and the beam width need to be determined. In the MCNPX (MCNP eXtended Version 2.5.0) code used in this study, there are at least three ways to define the specific shape of the fan beam. One way is to use many discrete point sources, each consisting of many individual pencil beams which simulate the fan-shaped beam (Khursheed et al. 2002). The second way is to use the collimators or jaws to shape the cone beam into the specific fan beam (Gu et al. 2008). The third way is to use the cookie-cutter function provided by MCNPX (Pelowitz 2005) to simulate only those photons emitted in the direction of interest. In this research the cookie-cutter method was used involving a spherical surface source and a suitable cookie-cutter cell (cuboid cell was defined in this research) as illustrated in figure 2.

2.2. Monte Carlo method

All simulations were performed using the MCNPX Monte Carlo code (Pelowitz 2005). MCNPX is a general purpose Monte Carlo radiation transport code that tracks all particles at all energies necessary for these simulations. The MCNPX package provides geometry modeling based on a combinatorial system using planes, cylinders, cones and spheres. In this study, the photon physics mode with the default energy cut-off was used. The photon transport
model creates electrons but assumes that they travel in the direction of the primary photon and that the electron energy is deposited at the photon interaction site to satisfy the condition of charged particle equilibrium (CPE) (DeMarco et al 2005), which is a valid assumption in the energy range of diagnostic x-rays. Under the conditions of CPE, the collision kerma is equal to absorbed dose and is recorded using the type 6 (F6:p) tally of the MCNPX. In all simulations, the number of histories was selected to achieve relative errors less than 5% in most organs (or detectors), and less than 10% for organs with very small volumes or located at large distances from the primary beam.

The F6:p tally results in MCNPX are normalized per source history. In order to determine the absorbed dose from each CT scan procedure, the tally values in units of MeV/gram/source particle were converted to absorbed dose in units of mGy/100 mAs by a conversion factor (CF). An in-air normalization method that is based on pencil-chamber exposure readings for a single axial scan was taken at the center of the CT gantry. The CFs used in this study have been described in a previous paper (Jarry et al 2002), as a function of both beam energy $E$ and beam collimation $NT$. In the current study, it is modified and defined as

$$\text{(CF)}_{E,NT} = \frac{(CTDI_{100, \text{air, measured per 100 mAs}})_{E,NT}}{(CTDI_{100, \text{air, simulated per particle}})_{E,NT}}$$

(1)

where $(CTDI_{100, \text{air, measured per 100 mAs}})_{E,NT}$ in units of mGy/100 mAs is measured by the ion chamber in air at the scanner isocenter, and $(CTDI_{100, \text{air, simulated per particle}})_{E,NT}$ in units of MeV/gram/particle is obtained by simulating the ion chamber under the same scan protocol. The absorbed dose to the organ or tissue in units of mGy/100 mAs is

$$D_{\text{absolute}} = D_{\text{simulated}} \times \text{CF} \times N,$$

(2)

and $N$ is the number of x-ray tube rotations during this CT scan. If the clinical tube current (mA) and exposure time (s) per rotation are known, the final absorbed dose in units of mGy is

$$D_{\text{total}} = D_{\text{absolute}} \times K,$$

(3)

where $K$ is the ratio of mAs per rotation to 100 mAs. In this study, we reported the organ dose in units of mGy/100 mAs based on equation (2).

2.3. CTDI phantom

The function of both the standard body CTDI dosimetry phantom and the standard head CTDI dosimetry phantom is to perform measurements of doses in the CT scan. The body CTDI phantom is 15 cm in length with a diameter of 32 cm. It is made of polymethylmethacrylate (PMMA) with a density of 1.19 g cm$^{-3}$. It has five sockets that can hold PMMA inserts or pencil ion chambers. One socket is at the center and the four others are 1 cm below the phantom surface, each 90$^\circ$ apart from its neighbor. Every socket is 1.37 cm in diameter and 15 cm in length, which provides enough space for a standard ion chamber. In this research, only the body CTDI phantom was modeled since only chest and kidney scans were considered. The ion chamber was modeled as a set of three concentric cylinders with a length of 10 cm (Deak 2008). The active volume consisted of an air-filled inner cylinder with a diameter of 6.7 mm. The second cylinder had a diameter of 10.2 mm. The space between the first and second cylinders represents the chamber wall which was simulated as C552 air-equivalent plastic with a density of 1.76 g cm$^{-3}$. The third cylinder had a diameter of 13.7 mm. The space between the second and third cylinders represents the ion chamber build-up cap, simulated as polyacetal plastic with a density of 1.43 g cm$^{-3}$. For all in-air simulations only the model of the ion chamber was used.
2.4. Surface dose profiles on the CTDI phantom

For the MDCT scanner operating in the helical scan mode, the spatially varying dose characteristics can be used as a further validation for the model we have created for the scanner. To this end, a total of 20 MOSFET detectors were simulated in MCNPX along the longitudinal (z) axis to calculate the dose profiles. The arrangement of the 20 MOSFET detectors in the simulations follows the arrangement described by DeMarco et al. (2005) who provided both the measured and calculated dose profile data that were used in this study for comparison. Simulations were performed using an identical set of scan parameters reported by Demarco et al. (2005) (i.e., 120 kVp, 4 × 5 mm² beam collimation, body bowtie filter) with the model of the body CTDI phantom to evaluate the dose profiles. The air kerma was scored along the surface of the body CTDI phantom. For the dose profile, the calculated data presented in this paper, as well as the calculated and measured data provided by DeMarco et al. (2005), were normalized to the maximum dose, respectively.

2.5. RPI pregnant patient phantoms and dose calculations

A set of realistic computational phantoms of a pregnant patient at the end of three gestational periods of 3, 6 and 9 months—called RPI-P3, RPI-P6 and RPI-P9 phantoms—were previously developed at RPI (Xu et al. 2007). Different from the pregnant patient phantoms only based on CT images and with a limited number of organs defined in Angel et al.’s (2008) study, in RPI-P3, RPI-P6 and RPI-P9 phantoms, organ volumes and masses were carefully adjusted to agree with reference values recommended in the ICRP Publication 89 (ICRP-89 2002). A total of 35 organs and tissues were included. The particular emphasis was placed on developing a realistic representation of fetus that consists of skeleton (except for the 3 month phantom), brain and soft tissue. Organs were voxelized with a resolution of 3 mm × 3 mm × 3 mm, and each phantom contained a total of about 25 million voxels (Taranenko et al. 2007). The RPI-P phantoms, illustrated in figure 5, were finally implemented into MCNPX for dose calculations.

A series of simulations was performed to calculate organ and fetal doses to the RPI-P3, RPI-P6 and RPI-P9 patient phantoms from a standard chest scan. An additional simulation was performed to calculate organ and fetal doses to the RPI-P3 patient phantom from a kidney scan. In this research the kidney scan was only applied to RPI-P3 to assure that the fetus is out of the field of view (FOV) of the CT scan. A pitch of 1.375 and a beam collimation of 16 × 1.25 mm were used. The scan lengths are shown in figure 5. For all scans involving the pregnant patient phantoms (the chest and kidney scans), the arms of the phantoms were removed to represent a patient whose arms are raised above the shoulders outside the chest scanning region.

3. Results and discussion

3.1. Validation of the MDCT scanner model

In this section, the results of the validation for the MDCT scanner model are presented. First, comparisons between calculated and measured CTDI values are presented and analyzed. Subsequently, calculated dose profiles on the surface of a CTDI phantom are compared with measured and calculated dose profiles reported in the literature (DeMarco et al. 2005).

3.1.1. Measured versus simulated values of CTDI phantoms. In order to validate the MDCT scanner model developed in this research, the body CTDI phantom was modeled and CTDI
values were calculated and compared with the measured values. The CTDI values were determined by taking the product of the normalized CTDI dose calculated by MCNPX and the kVp-dependent conversion factor. The kVp-dependent conversion factors were calculated from single axial scans simulated free-in-air at the scanner isocenter for tube potentials of 80, 100, 120 and 140 kVp using the 20 mm beam collimation with the body bowtie filter. For each simulation, the result was the Monte Carlo calculated air kerma (in units of MeV/gram per source particle) that occurred in the simulated ion chamber. The conversion factors for each kVp value are listed in table 1. Using these conversion factors, the simulated CTDI values (air-kerma values at the center and at the 12:00 peripheral position of the CTDI phantom) were calculated and are shown in table 2. The Monte Carlo simulated CTDI values at the center showed good agreement with measured values. The largest percent difference was 5.72% for the peripheral CTDI dose of the 120 kVp scan, and the smallest percent difference was 0.60% for the peripheral CTDI dose of the 80 kVp scan. Tables 1 and 2 used various kVp for scanner validation; however, all calculations of patient organ dose were performed using the 120 kVp scan protocol.

3.1.2. Dose profiles. In addition to CTDI values, the dose profiles on the surface of the CTDI phantom were calculated for further validation of the MDCT scanner modeling. The comparison between our simulated results with those experimental data reported by DeMarco et al (2005) for MOSFET detector measurements from a single axial scan of the CTDI body phantom is provided in figure 6. The error bars in the curve of MOSFET measurements indicate the uncertainty in the axial direction (z-axis) from MOSFET placement and in the dose direction due to MOSFET output variability (DeMarco et al 2005). The simulated dose profile values from DeMarco et al (2005) are also included in figure 6. The simulation

![Figure 5. RPI-P3, RPI-P6 and RPI-P9 phantoms at the end of three gestational periods of 3, 6 and 9 months, respectively. The anatomic coverage of the chest scan ranged from just above the clavicle bone down to below the diaphragm (around the middle of the liver). The anatomic coverage of the kidney scan was performed only for the RPI-P3 phantom and ranged from above the middle of the liver to below the bottom of the left kidney.](image)
The development, validation and application of a MDCT scanner model

Table 1. Conversion factors for converting the Monte Carlo simulation results from MeV/gram per source particle to mGy/100 mAs for the LightSpeed MDCT scanner. The percent relative error for the MCNPX results is less than 5%. The measurements (DeMarco et al 2005) and MCNPX simulations were performed in air using the body bowtie filter.

<table>
<thead>
<tr>
<th>kVp</th>
<th>Beam collimation (mm)</th>
<th>Measured CTDI_{100} in air (mGy/100 mAs)</th>
<th>Simulated CTDI_{100} in air (MeV/gram/particle)</th>
<th>Conversion factor (mGy gram/particle/100 mAs/MeV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>80</td>
<td>4 × 5</td>
<td>8.10</td>
<td>7.15 × 10^{-6}</td>
<td>1.13 × 10^{6}</td>
</tr>
<tr>
<td>100</td>
<td>4 × 5</td>
<td>14.70</td>
<td>8.34 × 10^{-6}</td>
<td>1.76 × 10^{6}</td>
</tr>
<tr>
<td>120</td>
<td>4 × 5</td>
<td>22.71</td>
<td>9.33 × 10^{-6}</td>
<td>2.44 × 10^{6}</td>
</tr>
<tr>
<td>140</td>
<td>4 × 5</td>
<td>31.93</td>
<td>1.03 × 10^{-5}</td>
<td>3.11 × 10^{6}</td>
</tr>
</tbody>
</table>

Table 2. Comparison of measured and simulated CTDI_{100} results from a single axial scan in the CTDI body phantom using the LightSpeed scanner operated at each kVp settings, all scans using 100 mAs. Measured dose data were released by DeMarco et al (2005). All scans used a 20 mm beam collimation and a body bowtie filter. The duration of the scan was 1 s.

<table>
<thead>
<tr>
<th>kVp</th>
<th>Position</th>
<th>Measured CTDI_{100} (mGy/100 mAs)</th>
<th>Simulated CTDI_{100} (mGy/100 mAs)</th>
<th>% Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>80</td>
<td>Center</td>
<td>1.34</td>
<td>1.30</td>
<td>−2.80</td>
</tr>
<tr>
<td></td>
<td>Peripheral</td>
<td>3.45</td>
<td>3.43</td>
<td>−0.60</td>
</tr>
<tr>
<td>100</td>
<td>Center</td>
<td>2.97</td>
<td>2.86</td>
<td>−3.77</td>
</tr>
<tr>
<td></td>
<td>Peripheral</td>
<td>6.66</td>
<td>6.79</td>
<td>1.89</td>
</tr>
<tr>
<td>120</td>
<td>Center</td>
<td>5.12</td>
<td>4.98</td>
<td>−2.78</td>
</tr>
<tr>
<td></td>
<td>Peripheral</td>
<td>10.48</td>
<td>10.97</td>
<td>4.69</td>
</tr>
<tr>
<td>140</td>
<td>Center</td>
<td>7.65</td>
<td>7.52</td>
<td>−1.76</td>
</tr>
<tr>
<td></td>
<td>Peripheral</td>
<td>15.01</td>
<td>15.87</td>
<td>5.72</td>
</tr>
</tbody>
</table>

Values presented from this study are in good agreement with both the measured values and the simulation values reported by DeMarco et al (2005), considering the uncertainties associated with the measurements and simulations, but systematically underestimate the measured dose in the plateau region of the profile curve. Several explanations contribute to this discrepancy. One is the limits of the approximation of MOSFET detectors in MCNPX. Second, the differences in the curves in figure 6 near the penumbra may be attributed to the assumption that photons were emitted from a point source and subsequently collimated before reaching the phantom. This point source approximation could help explain the differences in width of the penumbra region between the calculated and measured dose profiles. In reality the CT source has a finite spot size with scatter contributions from the collimation assembly and tube housing, which would increase the width of the geometric penumbra as shown in the measurement data provided in figure 6. Also, the use of the cookie-cutter function to model the 20 mm nominal beam width may affect the shape of the curve shown in figure 6 (Pelowitz 2005).

Results of the contiguous axial scan simulation for the CTDI body phantom are illustrated in figure 7(a). Also included in the figure are measured and calculated data reported in the literature (DeMarco et al 2005). All three sets of data demonstrate the spatial variation of dose at the surface of the phantom with approximately the same degree of variation from peak to trough with the same period of variation. It should be noted that the results from the Monte Carlo simulations are much smoother since a very large number of sample points was used in comparison with the measurements involving only 20 MOSFET detectors. Results
from the helical scans of a pitch of 1.375 and a pitch of 0.9375 are shown in figures 7(b) and (c), respectively. Again, results are compared to the measured and calculated data from the literature (DeMarco et al 2005). Overall, the calculated results are in good agreement with the measured values, especially with those provided by DeMarco et al (2005), under helical scan conditions.

Overall, these results show very good agreements between the measured and simulated values for the single axial scan, the contiguous axial scan and the helical scans providing confidence in the MDCT dose calculation using RPI pregnant patient phantoms.

3.2. Absorbed dose to mother and fetus using pregnant patient phantoms

Following the validation of the MDCT source modeling, this section describes the calculated doses to the mother and fetus from maternal body chest and kidney CT scans. In all the simulations involving the pregnant patient phantoms, 10 million initial photons were sampled to guarantee statistical uncertainties less than 5% for the absorbed doses to the organs in the FOV.

Table 3 summarizes the organ and fetal dose values that are normalized per 100 mAs for the chest and kidney scans of the RPI-P3, RPI-P6 and RPI-P9 phantoms. As described previously, the organ doses to the pregnant patient phantoms (3, 6 and 9 month) from helical scans were assessed for a tube potential of 120 kVp and a pitch of 1.375. Doses from both chest and kidney scans were calculated to the pregnant patient with a 3 month fetus, while only chest scans of the pregnant patient phantoms with 6 and 9 month fetuses were considered due to safety concerns from kidney scans for these stages of gestation. It should also be
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Figure 7. Results showing comparison of our Monte Carlo results with MOSFET measurements and simulation data extracted from DeMarco et al (2005) along the longitudinal axis of the body CTDI phantom resulting from (a) a contiguous axial scan, (b) a helical scan with pitch equal to 1.375 and (c) a helical scan with pitch equal to 0.9375. The doses in each figure are normalized to the maximum one for each set of dose values. The Monte Carlo results are based upon a percent relative error ranging from 2% to 8% for all data points and are shown using error bars. For the simulation results reported by DeMarco et al (2005), the percent relative errors are less than 2% for all data points and are not shown for clarity. The MOSFET measurements are presented with a y-axis relative error based upon the reproducibility results provided in DeMarco et al (2005) and an x-axis uncertainty based upon an approximate positioning error of ± 0.1 cm to its adjacent neighbor(s) (DeMarco et al 2005).

noted that unlike the RPI-P6 and RPI-P9 phantoms the fetal skeleton is not defined in the RPI-P3 phantom since skeletal development takes place during later stages of pregnancy. All MCNPX dose values were converted to absorbed dose in units of mGy/100 mAs by the corresponding CFs shown in table 1. For the chest scan of the RPI-P3, P6 and P9 phantoms, the thymus received the largest doses of 9.91 mGy/100 mAs, 9.72 mGy/100 mAs and 9.63 mGy/100 mAs, respectively. The lungs, heart walls and breasts also received large doses from all three scans. For the kidney scan of the RPI-P3 phantom, the kidney received the largest dose of 8.24 mGy/100 mAs. The gallbladder wall, pancreas and spleen also received large doses. As expected, for both scan procedures, the organs in the FOV received higher doses. For most organs the relative statistical uncertainties in the calculated dose did not exceed 5%, except for the extremely small eye lens which had uncertainties ranging from 11.7% to 21.8%.
Table 3. Calculation results showing doses to organs of the RPI-P3, P6 and P9 phantoms. For the RPI-P3 phantom, both the chest and kidney scans were applied. For the RPI-P6 and P9 phantoms, only the chest scans were applied to the two phantoms. The values in parentheses indicate the relative uncertainties associated with the corresponding Monte Carlo simulations.

<table>
<thead>
<tr>
<th>Organ dose (mGy/100 mAs)</th>
<th>RPI-P3 MDCT scan</th>
<th>RPI-P6 MDCT scan</th>
<th>RPI-P9 MDCT scan</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Chest</td>
<td>Kidney</td>
<td>Chest</td>
</tr>
<tr>
<td>Brain</td>
<td>0.21 (0.01)</td>
<td>0.05 (0.02)</td>
<td>0.21 (0.01)</td>
</tr>
<tr>
<td>Eyeballs</td>
<td>0.31 (0.04)</td>
<td>0.08 (0.07)</td>
<td>0.34 (0.04)</td>
</tr>
<tr>
<td>Eye lens</td>
<td>0.45 (0.12)</td>
<td>0.11 (0.22)</td>
<td>0.41 (0.12)</td>
</tr>
<tr>
<td>Trachea</td>
<td>2.05 (0.02)</td>
<td>0.14 (0.05)</td>
<td>2.00 (0.02)</td>
</tr>
<tr>
<td>Tracheus</td>
<td>5.62 (0.01)</td>
<td>0.18 (0.04)</td>
<td>5.36 (0.01)</td>
</tr>
<tr>
<td>Tracheus</td>
<td>9.91 (0.01)</td>
<td>0.25 (0.04)</td>
<td>9.72 (0.01)</td>
</tr>
<tr>
<td>Lungs</td>
<td>9.21 (0.00)</td>
<td>0.71 (0.00)</td>
<td>8.61 (0.00)</td>
</tr>
<tr>
<td>Heart wall</td>
<td>9.36 (0.00)</td>
<td>0.51 (0.01)</td>
<td>8.99 (0.00)</td>
</tr>
<tr>
<td>Esophagus wall</td>
<td>5.85 (0.01)</td>
<td>0.32 (0.02)</td>
<td>5.38 (0.01)</td>
</tr>
<tr>
<td>Breasts</td>
<td>8.77 (0.00)</td>
<td>0.54 (0.01)</td>
<td>9.13 (0.00)</td>
</tr>
<tr>
<td>Stomach wall</td>
<td>4.01 (0.01)</td>
<td>4.15 (0.00)</td>
<td>3.94 (0.01)</td>
</tr>
<tr>
<td>Liver</td>
<td>5.52 (0.00)</td>
<td>3.49 (0.00)</td>
<td>5.51 (0.00)</td>
</tr>
<tr>
<td>Gallbladder wall</td>
<td>1.88 (0.02)</td>
<td>6.94 (0.01)</td>
<td>1.88 (0.02)</td>
</tr>
<tr>
<td>Pancreas</td>
<td>0.95 (0.01)</td>
<td>6.81 (0.00)</td>
<td>0.95 (0.01)</td>
</tr>
<tr>
<td>Spleen</td>
<td>2.51 (0.01)</td>
<td>7.07 (0.00)</td>
<td>2.52 (0.01)</td>
</tr>
<tr>
<td>Kidneys</td>
<td>0.89 (0.01)</td>
<td>8.24 (0.00)</td>
<td>0.88 (0.01)</td>
</tr>
<tr>
<td>Adrenals</td>
<td>1.83 (0.02)</td>
<td>5.79 (0.01)</td>
<td>1.76 (0.02)</td>
</tr>
<tr>
<td>SI wall and contents</td>
<td>0.32 (0.01)</td>
<td>3.06 (0.00)</td>
<td>0.49 (0.01)</td>
</tr>
<tr>
<td>LI wall</td>
<td>0.44 (0.01)</td>
<td>4.26 (0.00)</td>
<td>0.62 (0.01)</td>
</tr>
<tr>
<td>LI contents</td>
<td>0.40 (0.01)</td>
<td>3.88 (0.00)</td>
<td>0.53 (0.01)</td>
</tr>
<tr>
<td>Ovaries</td>
<td>0.14 (0.07)</td>
<td>0.73 (0.03)</td>
<td>0.13 (0.08)</td>
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<tr>
<td>Bladder wall</td>
<td>0.08 (0.05)</td>
<td>0.26 (0.02)</td>
<td>0.06 (0.06)</td>
</tr>
<tr>
<td>Uterine wall</td>
<td>0.22 (0.02)</td>
<td>1.23 (0.01)</td>
<td>0.40 (0.01)</td>
</tr>
<tr>
<td>Uterine contents</td>
<td>0.17 (0.02)</td>
<td>0.86 (0.01)</td>
<td>0.33 (0.01)</td>
</tr>
<tr>
<td>Placenta</td>
<td>0.35 (0.03)</td>
<td>2.36 (0.01)</td>
<td>0.91 (0.01)</td>
</tr>
<tr>
<td>Skeleton</td>
<td>4.45 (0.00)</td>
<td>1.92 (0.00)</td>
<td>4.20 (0.00)</td>
</tr>
<tr>
<td>Skin</td>
<td>1.37 (0.00)</td>
<td>0.84 (0.00)</td>
<td>1.36 (0.00)</td>
</tr>
<tr>
<td>Remainder</td>
<td>1.61 (0.00)</td>
<td>1.39 (0.00)</td>
<td>1.55 (0.00)</td>
</tr>
<tr>
<td>Fetal soft tissue</td>
<td>0.13 (0.04)</td>
<td>0.58 (0.02)</td>
<td>0.18 (0.02)</td>
</tr>
<tr>
<td>Fetal skeleton</td>
<td>NA*</td>
<td>NA*</td>
<td>0.67 (0.02)</td>
</tr>
<tr>
<td>Fetal brain</td>
<td>0.12 (0.09)</td>
<td>0.50 (0.04)</td>
<td>0.11 (0.05)</td>
</tr>
<tr>
<td>Fetus total</td>
<td>0.13 (0.04)</td>
<td>0.57 (0.02)</td>
<td>0.21 (0.02)</td>
</tr>
</tbody>
</table>

* RPI-P3 does not contain fetal skeleton.

In addition to organ doses to the mother, total fetal doses and individual fetal organ doses normalized per 100 mAs from the chest and kidney scans are also provided in Table 3. In order to assess the absolute fetal doses from a typical scan, the normalized fetal dose values were multiplied by assumed exposures of 304 mAs and 80 mAs for the chest scans and kidney scans, respectively (Hurwitz et al. 2006). Actually the kidney scan in Hurwitz et al.’s study is looking for calcified stones in the renal and urinary systems. Hence, 80 mAs is used for that specific scan and is not typical of a kidney scan or an abdominal scan (200 mAs would be more typical). Besides, 140 kVp was used in Hurwitz et al.’s study, so the practical mAs may also be accordingly adjusted to match other different kVp scan protocols.
The absolute doses to different fetal organs from the chest scan in the RPI-P3, P6 and P9 phantoms are provided in figure 8. Also included in figure 8 are the total-body fetal doses. As can be seen from the figure, the absorbed dose to the fetal soft tissue increases with increasing gestational stage of the pregnant patient. This is due to the increasing size of the fetus and the subsequent decrease in the distance between the field edge and the nearest fetal point. Conversely, the absorbed dose to the fetal skeleton and brain decreased with increasing gestational stage. The decrease in these doses is due to increased shielding from soft tissue to these organs, which are mostly located in regions further away from the field edge (see figure 5). The dose to the fetal total is the mass weighted average of the doses to the fetal soft tissue, fetal skeleton and fetal brain. The fetal total doses for the chest scan of the RPI-P3, P6 and P9 pregnant patient phantoms were 0.13 mGy/100 mAs, 0.21 mGy/100 mAs and 0.26 mGy/100 mAs, respectively.

Hurwitz et al reported maximum measured doses to 3 month old fetus to be 0.6 mGy and 11.7 mGy for the chest and kidney scans, respectively (Hurwitz et al 2006). Angel et al simulated the radiation dose to the fetus resulting from an abdominal and pelvic CT examination for 24 pregnant patient phantoms with gestational age ranging from less than 5 weeks to 36 weeks, and the fetal dose ranged 7.4–14.3 mGy/100 mAs (Angel et al 2008) when using helical scan pitch 1, equivalent to 5.38–10.4 mGy/100 mAs when using helical scan pitch 1.375. In our study, the fetal dose of the RPI-P3 phantom from the kidney scan was 0.57 mGy/100 mAs. If 300 mAs is used for the kidney scan, the total fetal dose is 1.71 mGy, which is much smaller than the fetal dose reported by Hurwitz et al (2006) or Angel et al (2008). The reason for the large difference between RPI-P3 fetal dose and other reported fetal doses mainly lies in scan protocols (i.e., kVp and mAs settings, scan area, scan length, pitch). For the simulation of the kidney scan in our study, CT examinations were performed in areas away from the uterus, so the fetus was not directly exposed to radiation. For the kidney scan by Hurwitz et al (2006), and the abdominal and pelvic examination by Angel et al (2008), the fetus was covered by the CT scan, exposed directly to radiation, which resulted in the large dose to the fetus. For confirmation, the abdominal and pelvic CT examinations were also simulated for RPI-P3 in our study to test the consistency of the fetal dose value with the dose values reported in Angel et al’s (2008) study, without changing other...
scan parameters in the protocol used for the RPI-P3 kidney scan. The fetal dose for RPI-P3 abdominal and pelvic scans is 6.90 mGy/100 mAs, which agrees with Angel et al’s results (dose ranged 5.38–10.4 mGy/100 mAs, pitch 1.375).

Assessment of the risk associated with radiation exposure of the fetus during MDCT scans of pregnant patients is of increasing interest (ICRP-102 2007). Based on the results by Hurwitz et al (2006), ICRP Publication 102 presents a summary of results from a phantom study using the protocols for imaging pregnant patients with suspected pulmonary embolism, appendicitis and renal stones (ICRP-102 2007). In a different publication, ICRP Publication 103 suggests that absorbed doses below 100 mGy to the embryo/fetus should not be considered a reason for terminating a pregnancy (ICRP-103 2007). At embryo/fetus doses above this level, the pregnant patient should receive sufficient information to be able to make informed decisions based upon individual circumstances, including the magnitude of the estimated embryonic/fetal dose and the consequent risks of serious harm to the developing embryo/fetus and risks of cancer in later life (ICRP-103 2007). Furthermore, the AAPM Task Group 36 (TG-36) provided a general summary of the risk to the fetus as a function of radiation dose for radiotherapy (Stovall et al 1995). According to TG-36, the risk of normal tissue damage from fetal doses less than 50 mGy is negligible. Assuming a total of 300 mAs is applied to each scan, one obtains a fetal dose far below 50 mGy, and therefore the risk is negligible for a fetus of the RPI pregnant patient phantom when MDCT chest and kidney scans are performed.

Furthermore, this research is emphasized on assessment of the dose to the fetus outside the scan area, and another similar research conducted by Angel et al (2008) is focused on the assessment of dose to the fetus in the scan area. At this point the two researches can be mutually complementary in terms of fetus dosimetric study in the MDCT scan. The fetus will receive about five to nine times the dose during the abdominal and pelvic scans than that during the kidney scan, according to the doses provided before. This conclusion can be a strong indication that the dose to the fetus will sharply increase once the fetus is in the scan area of a CT procedure. Both the scan cases (fetus in the scan area studied by Angel et al and fetus outside the scan area studied in our research) can result in small fetal doses; however, all possible ways should be considered to reduce the fetal dose based on the principle of ALARA (as low as reasonably achievable). A suggestion for the clinical scan protocol can be proposed to avoid the scan involving the fetus in the scan area if not substantially compromising the diagnosis. For example, the protocol for the kidney scan of the pregnant patient should be carefully considered, ranging from above the middle of the liver to below the bottom of the left kidney without involving the fetus or a part of the fetus in the scan area (see figure 5). The fetal dose of 11.7 mGy following the scan protocol by Hurwitz et al (2006) will be reduced using this protocol to 3.42 mGy if 300 mAs is applied in this study. The use of low mAs, high pitch and a limited scan volume, if not substantially compromising the image quality, should also be considered to reduce the fetal dose (Forsted and Kalbhen 2001).

There were several limitations to this study. While the pregnant phantoms are well modeled according to ICRP reference data, they may not be typical of the general population of all pregnant patients as well as fetuses. Another limitation was that only one MDCT scanner (GE LightSpeed CT scanner) model was used in this research. It is natural that there may be substantial differences in dose performance with various manufacturers and models of CT scanners due to differences in x-ray beam filtration configuration. It is not easy to get enough information for every CT scanner manufactured by different makers and then to model these CT scanners using Monte Carlo methods; therefore, this research is an effort to model the specific CT scanner based on limited parameters and later this methodology can be extended to model other different CT scanners. This study also simulated only constant
tube current scans without considering tube current modulation methods (Jaffe et al. 2008).
All these limitations will be studied in future by applying various MDCT scanner models
to other patient phantoms, i.e., RPI adult male and female phantoms (Zhang et al. 2008),
using more scan protocols including tube current modulation. Additionally, dose and image
quality will also be involved in the further research. By testing different tube voltage and
current-time as well as x-ray source filters (Toth et al. 2005) using Monte Carlo simulations,
the dose performance can be evaluated for various CT scanners with different scan protocols.
Along with CT image quality obtained by using the noise model (Toth et al. 2005) or analyzing
simulated CT images (Ay and Zaidi 2005), the correlation between dose and image quality
will be further studied.

4. Conclusion
Since MDCT is increasingly popular, the need to assess and manage the potential exposures
and associated radiation risk can be met by using Monte Carlo models of these modalities
as well as anatomic realistic patient computational phantoms. This paper described the
development and validation of Monte Carlo models of an MDCT scanner, and the application
of these models for the calculation of absorbed doses to the mother and fetus using selected
MDCT scanning protocols.
Detailed MDCT models including energy spectrum, filters, source geometry and
movement under different scan modes were developed. The models were validated by
comparing calculated center and peripheral CTDI dose values and dose profile curves to
measurement data reported in the literature. Considering four different operating voltages
(80, 100, 120 and 140 kVp), the difference between calculated and measured center
and peripheral CTDI doses were all about or less than 5%. Also, good agreement was
achieved between calculated and measured single, contiguous and helical axial dose profile
curves.

The validated x-ray source model and the helical source movement model were then
integrated with the pregnant patient phantoms of different gestational ages to calculate organ
doses for different CT scan protocols. Results show that the risk to the fetus in the mother
during 3, 6 or 9 month gestational periods from chest MDCT scans is negligible, according
to limits established by a recent ICRP Publication 103 (ICRP-103 2007) and AAPM TG-36
(Stovall et al. 1995). Based on the comparison of the assessed fetal doses in this study and
the fetal doses reported by Hurwitz et al. (2006) and Angel et al. (2008), a suggestion for the
clinical scan protocol can be proposed to avoid the scan involving the fetus in the scan area
if not substantially compromising the diagnosis for the principle of ALARA. Such a study
demonstrates the usage of the Monte Carlo MDCT models and the patient phantoms. The
methods can be applied to adult male and pediatric patient phantoms developed in RPI and
elsewhere for the MDCT modality that is constantly evolving (Zhang et al. 2008).

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calculation.
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