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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 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.

1. Introduction

Computed tomography (CT), as a preeminent diagnostic imaging modality, has been known to contribute a significant portion of the overall radiation dose received by patients from a diagnostic procedure (ICRP-102 2007, Mettler 2007, Shrimpton and Edyvean 1998, Brenner *et al* 2001, 2007, NEXT 2006, Brenner and Georgsson 2005, Brenner 2004, Brenner and Elliston 2004, Huda and Vance 2007, McNitt-Gray 2003, AAPM 2008). The advent of multi-detector CT (MDCT) has fueled a dramatic increase in use of CT, and the radiation

exposures associated with various CT scan protocols have been studied extensively (Brenner *et al* 2001, 2007, NEXT 2006, Brenner and Georgsson 2005, Brenner 2004, Brenner and Elliston 2004, Huda and Vance 2007, McNitt-Gray 2003). There is a consensus from the medical physics community that an increase in MDCT utilization must be accompanied by a heightened understanding of radiation dose and risk (AAPM 2008). The American College of Radiology has issued a strong recommendation on tracking and reporting patient-specific exposures and resulting organ doses (Amis *et al* 2007).

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 Boisselle *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 (Boisselle *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. Winer-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 (CTDI_c), CTDI peripheral dose values (CTDI_p) 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.

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, ImPACT 2004)): 16×0.625 mm (8×1.25 mm) and 16×1.25 mm (8×2.5 mm), as well as 2×0.625 mm and 1×5 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 x-ray spectrum for a CT scanner, including xcomp5r, TASMIP, Srs-78, Specgen T, Specgen B and xraytbc (Meyer *et al* 2004, Siewerdsen *et al* 2004). In this research the software xcomp5r was used to generate the x-ray source spectrum (Nowotny and Höfer 1985). For a given tube voltage kVp, there are several additional variables that impact the x-ray spectrum. In xcomp5r, these variables include the anode angle, the distance from the source to the user-defined location where the spectrum is recorded and the flat filter (absorbers) thickness and material definition.

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

$$(CF)_{E,NT} = \frac{(CTDI_{100, \text{air, measured per 100 mAs}})_{E,NT}}{(CTDI_{100, \text{air, simulated per particle}})_{E,NT}} \quad (1)$$

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

$$D_{\text{absolute}} = D_{\text{simulated}} \times CF \times N, \quad (2)$$

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

$$D_{\text{total}} = D_{\text{absolute}} \times K \quad (3)$$

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° 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, $4 \times 5 \text{ mm}^2$ 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 $3\text{ mm} \times 3\text{ mm} \times 3\text{ mm}$, 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 $16 \times 1.25\text{ mm}$ 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 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 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%.

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 anatomically 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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References

- AAPM. The measurement, reporting, and management of radiation dose in CT. Report of AAPM Task Group 23 of the Diagnostic Imaging Council CT Committee. 2008. Available at <http://www.aapm.org/pubs/reports/rpt96.pdf>
- Amis ES, et al. American College of Radiology white paper on radiation dose in medicine. *J Am Coll Radiol*. 2007; 4:272–84. [PubMed: 17467608]
- Angel E, et al. Radiation dose to the fetus for pregnant patients undergoing multidetector CT imaging: Monte Carlo simulations estimating fetal dose for a range of gestational age and patient size. *Radiology*. 2008; 249:220–7. [PubMed: 18796678]
- Ay MR, Zaidi H. Development and validation of MCNP4 C-based Monte Carlo simulator for fan- and cone-beam x-ray CT. *Phys Med Biol*. 2005; 50:4863–85. [PubMed: 16204878]
- Boiselle PM, Reddy S, Villas PH, Liu A, Seibyl JP. Pulmonary embolus in pregnant patients: survey of ventilation-perfusion imaging policies and practices. *Radiology*. 1998; 207:201–6. [PubMed: 9530317]
- Brenner DJ. Radiation risks potentially associated with low-dose CT screening of adult smokers for lung cancer. *Radiology*. 2004; 231:440. [PubMed: 15128988]
- Brenner DJ, Elliston C, Hall E, Berdon W. Estimated risks of radiation-induced fatal cancer from pediatric CT. *Am J Roentgenol*. 2001; 176:289–96. [PubMed: 11159059]
- Brenner DJ, Elliston CD. Estimated radiation risks potentially associated with full-body CT screening. *Radiology*. 2004; 232:735–8. [PubMed: 15273333]
- Brenner DJ, Georgsson MA. Mass screening with CT colonography: should the radiation exposure be of concern? *Gastroenterology*. 2005; 129:328–37. [PubMed: 16012958]
- Brenner DJ, Hall EJ, Phil D. Computed tomography—an increasing source of radiation exposure. *N Engl J Med*. 2007; 357:2277–84. [PubMed: 18046031]
- Caracappa, PF. Thesis Nuclear Engineering and Engineering Physics. Rensselaer Polytechnic Institute; Troy, NY: 2001. Investigation of patient dose from CT examination using the VIP-Man model.
- Castro MA, Shipp TD, Castro EE, Ouzounian J, Rao P. The use of helical computed tomography in pregnancy for the diagnosis of acute appendicitis. *Am J Obstet Gynecol*. 2001; 184:954–7. [PubMed: 11303204]
- Chen MY, Zagoria RJ, Saunders HS, Dyer RB. Trends in the use of unenhanced helical CT for acute urinary colic. *Am J Roentgenol*. 1999; 173:1447–50. [PubMed: 10584780]
- Damilakis J, Perisinakis K, Voloudaki A, Gourtsoyiannis N. Estimation of fetal radiation dose from computed tomography scanning in late pregnancy: depth-dose data from routine examinations. *Invest Radiol*. 2000; 35:527–33. [PubMed: 10981996]
- Deak P. Validation of a Monte Carlo tool for patient-specific dose simulations in multi-slice computed tomography. *Eur Radiol*. 2008; 18:759–72. [PubMed: 18066555]
- DeMarco JJ, Cagnon CH, Cody DD, Stevens DM, McCollough CH, O'Daniel J, McNitt-Gray MF. A Monte Carlo based method to estimate radiation dose from multidetector CT (MDCT): cylindrical and anthropomorphic phantoms. *Phys Med Biol*. 2005; 50:3989–4004. [PubMed: 16177525]
- Felmlee JP, Gray JE, Leetzow ML, Price JC. Estimated fetal radiation dose from multislice CT studies. *Am J Roentgenol*. 1990; 154:185–90. [PubMed: 2104708]
- Forsted DH, Kalbhen CL. CT of pregnant women for urinary tract calculi, pulmonary thromboembolism, and acute appendicitis. *Am J Roentgenol*. 2001; 178:1285–6. [PubMed: 11959748]
- George XD, Dennis MD, Charles WC. Characteristics of kilovoltage x-ray beams used for cone-beam computed tomography in radiation therapy. *Phys Med Biol*. 2007; 52:1595–615. [PubMed: 17327651]
- Gu J, Bednarz B, Xu XG, Jiang SB. Assessment of patient organ doses and effective doses using the VIP-MAN adult male phantom for selected cone-beam CT imaging procedures during image guided radiation therapy. *Radiat Prot Dosim*. 2008; 131:431–43.
- Huda W, Vance A. Patient doses in adult and pediatric CT. *Am J Roentgenol*. 2007; 188:540–6. [PubMed: 17242266]

- Hurwitz LM, Yoshizumi T, Reiman RE, Philip CG, Erik KP, Donald PF, Greta T, Giao N, Lottie B. Radiation dose to the fetus from body MDCT during early gestation. *Am J Roentgenol*. 2006; 186:871–6. [PubMed: 16498123]
- ICRP-89. Basic anatomical and physiological data for use in radiological protection: reference values. *Ann ICRP*. 2002; 32:1–277.
- ICRP-102. Managing patient dose in multi-detector computed tomography (MDCT). *Ann ICRP*. 2007; 37:1–80.
- ICRP-103. The 2007 recommendations of the International Commission on Radiological Protection. *Ann ICRP*. 2007; 37:1–332.
- ImPACT. GE LightSpeed16 CT scanner technical evaluation imaging performance assessment of CT scanners group. 2004. Available at <http://www.impactscan.org/reports/MHRA04015.htm>
- Jaffe TA, Yoshizumi TT, Toncheva GI, Nguyen G, Hurwitz LM, Nelson RC. Early first-trimester fetal radiation dose estimation in 16-MDCT without and with automated tube current modulation. *Am J Roentgenol*. 2008; 190:860–4. [PubMed: 18356429]
- Jarry G, DeMarco JJ, McNitt-Gray MF. Monte Carlo dose verification of a commercial CT scanner with applications for patient specific dosimetry. *Med Phys*. 2002; 29:1344.
- Khurshed A, Hillier MC, Shrimpton PC, Wall BF. Influence of patient age on normalized effective doses calculated for CT examinations. *Br J Radiol*. 2002; 75:819–30. [PubMed: 12381691]
- McNitt-Gray MF. Radiation issues in computed tomography screening. *Semin Roentgenol*. 2003; 38:87–99. [PubMed: 12698594]
- Mettler, FA. Magnitude of radiation uses and doses in the United States: NCRP Scientific Committee 6-2 analysis of medical exposures. *Advances in Radiation Protection in Medicine*; 16 April; 2007. Presentation
- Meyer P, Buffard E, Mertz L, Kennel C, Constantinesco A, Siffert P. Evaluation of the use of six diagnostic x-ray spectra computer codes. *Br J Radiol*. 2004; 77:224–30. [PubMed: 15020364]
- NEXT. What's NEXT? Nationwide evaluation of x-ray trends: 2000 computed tomography. Conf. Radiation Control Program Directors; Department of Health and Human Services; 2006. (CRCPD publication no NEXT 2000CT-T)
- Nowotny VR, Höfer A. Ein programm für die Berechnung von diagnostischen. Röntgenspektren *Fortschr Röntgenstr*. 1985; 6:685–9.
- Paulson EK, Kalady MF, Pappas TN. Clinical practice: suspected appendicitis. *N Engl J Med*. 2003; 348:236–42. [PubMed: 12529465]
- Pelowitz, DB. MCNPXTM User's Manual Version 2.5.0. 2005.
- Ratnapalan S, Bona N, Chandra K, Koren G. Physicians' perceptions of teratogenic risk associated with radiography and CT during early pregnancy. *Am J Roentgenol*. 2004; 182:1107–9. [PubMed: 15100102]
- Remy-Jardin M, Remy J. Spiral CT angiography of the pulmonary circulation. *Radiology*. 1999; 212:615–36. [PubMed: 10478224]
- Schuster ME, Fishman JE, Copeland JF, Hatabu H, Boiselle PM. Pulmonary embolism in pregnant patients: a survey of practices and policies for CT pulmonary angiography. *Am J Roentgenol*. 2003; 181:1495–8. [PubMed: 14627562]
- Shi CY, Xu XG. Development of a 30-week-pregnant female tomographic model from CT-images for Monte Carlo organ dose calculations. *Med Phys*. 2004; 31:2491–7. [PubMed: 15487729]
- Shrimpton PC, Edyvean S. CT scanner dosimetry. *Br J Radiol*. 1998; 71:1–3. [PubMed: 9534691]
- Siewerdsen JH, Waese AM, Moseley DJ, Richard S, Jaffray DA. Spektr: a computational tool for x-ray spectral analysis and imaging system optimization. *Med Phys*. 2004; 31:3057–67. [PubMed: 15587659]
- Staton RJ, Lee C, Lee C, Williams MD, Hintenlang DE, Arreola MM, Williams JL, Bolch WE. Organ and effective doses in newborn patients during helical multislice computed tomography examination. *Phys Med Biol*. 2006; 51:5151–66. [PubMed: 17019030]
- Stovall M, Blackwell CR, Cundiff J, Novack DH, Palta JR, Wagner LK, Webster EW, Shalek RJ. Fetal dose from radiotherapy with photon beams: report of AAPM Radiation Therapy Committee Task Group No. 36. *Med Phys*. 1995; 22:63–82. [PubMed: 7715571]

- Taranenko, V.; Zhang, J.; Zhang, D.; Xu, XG.; Shi, C. Preliminary external dosimetry data from a new set of mother/fetus models. 2007; American Nuclear Society Ann. Meeting; 2007. p. 448
- Thomton FJ, Paulson EK, Yoshizumi TT, Frush DP, Nelson RC. Single versus multi-detector row CT: comparison of radiation doses and dose profiles. *Acad Radiol.* 2003; 10:379–85. [PubMed: 12678177]
- Toth TL, Cesmeli E, Ikhlef A, Horiuchi T. Image quality and dose optimization using novel x-ray source filters tailored to patient size. *Proc SPIE.* 2005; 5745:283–91.
- Winer-Muram HT, Boone JM, Brown HL, Jennings SG, Mabie WC, Lombardo GT. Pulmonary embolism in pregnant patients: fetal radiation dose with helical CT. *Radiology.* 2002; 224:487–92. [PubMed: 12147847]
- Xu XG, Taranenko V, Zhang J, Shi C. A boundary-representation method for designing whole-body radiation dosimetry models: pregnant females at the ends of three gestational periods—RPI-P3, -P6 and -P9. *Phys Med Biol.* 2007; 52:7023–44. [PubMed: 18029991]
- Yoshizumi TT, Nelson RC. Radiation issues in multidetector row helical CT. *Crit Rev Comput Tomogr.* 2003; 44:95–117. [PubMed: 12757315]
- Zhang J, Na Y, Xu XG. Development of whole-body phantoms representing an average adult male and female using surface-geometry methods. *Med Phys.* 2008; 35:2875.

Figure 1.

Flowchart for development, validation and application of the MDCT models using Monte Carlo simulations. The left side of the vertical dashed line indicates the work or data provided by the literature or previous research. All the measured data in this research were those reported by DeMarco *et al* (2005). The patient phantoms were developed by Xu *et al* (2007). The right side of the vertical dashed line indicates the simulations and calculations in this study.

Figure 2.

The geometry of CT scanner simulations. In this study it is assumed that the photons are emitted from a point at the location of the anode of the x-ray tube. The software Xcomp5r was used to generate the energy spectrum for the emitted photons. The flat filter, the bowtie filter and the beam shape were all considered. The CTDI phantom was used to validate the modeled MDCT scanner.

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.

Figure 4.

Representation of the 16 x-ray sources and bowtie filters used to simulate the single axial CT scan and the measurements of the body CTDI phantom in the center and periphery.

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.

Figure 6.

Results showing the comparison of our Monte Carlo results with MOSFET measurement and simulation data reported by DeMarco *et al* (2005) for a single axial scan surface air-kerma distribution profile along the z -direction involving a body phantom scanned with 120 kVp x-rays and $4 \times 5 \text{ mm}^2$ beam collimation. For each set among the three sets of surface doses, the surface doses are normalized to the maximum one in this set. Our Monte Carlo results include percent relative errors ranging from 2% to 8% that are shown as error bars. For simulation results by DeMarco *et al* (2005), the percent relative errors are less than 2% for all data points and are not shown. The MOSFET measurement data by DeMarco *et al* (2005) are presented with a y -axis relative error based upon the reproducibility results, and an x -axis uncertainty for a positioning error of $\pm 0.1 \text{ cm}$.

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).

Figure 8.

Comparison of doses to the 3, 6, and 9 month fetuses from chest scans of pregnant patients.

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.

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

Table 2

Comparison of measured and simulated CTDI₁₀₀ 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.

kVp	Position	Measured CTDI ₁₀₀ (mGy/100 mAs)	Simulated CTDI ₁₀₀ (mGy/100 mAs)	% Difference
80	Center	1.34	1.30	-2.80
	Peripheral	3.45	3.43	-0.60
100	Center	2.97	2.86	-3.77
	Peripheral	6.66	6.79	1.89
120	Center	5.12	4.98	-2.78
	Peripheral	10.48	10.97	4.69
140	Center	7.65	7.52	-1.76
	Peripheral	15.01	15.87	5.72

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.

	Organ dose (mGy/100 mAs)					
	RPI-P3 MDCT scan		RPI-P6 MDCT scan		RPI-P9 MDCT scan	
	Chest	Kidney	Chest	Chest	Chest	Chest
Brain	0.21 (0.01)	0.05 (0.02)	0.21 (0.01)	0.21 (0.01)	0.21 (0.01)	0.21 (0.01)
Eyeballs	0.31 (0.04)	0.08 (0.07)	0.34 (0.04)	0.34 (0.04)	0.33 (0.04)	0.33 (0.04)
Eye lens	0.45 (0.12)	0.11 (0.22)	0.41 (0.12)	0.41 (0.12)	0.38 (0.12)	0.38 (0.12)
Thyroid	2.05 (0.02)	0.14 (0.05)	2.00 (0.02)	2.00 (0.02)	1.87 (0.02)	1.87 (0.02)
Trachea	5.62 (0.01)	0.18 (0.04)	5.36 (0.01)	5.36 (0.01)	5.28 (0.01)	5.28 (0.01)
Thymus	9.91 (0.01)	0.25 (0.04)	9.72 (0.01)	9.72 (0.01)	9.63 (0.01)	9.63 (0.01)
Lungs	9.21 (0.00)	0.71 (0.00)	8.61 (0.00)	8.61 (0.00)	8.29 (0.00)	8.29 (0.00)
Heart wall	9.36 (0.00)	0.51 (0.01)	8.99 (0.00)	8.99 (0.00)	8.79 (0.00)	8.79 (0.00)
Esophagus wall	5.85 (0.01)	0.32 (0.02)	5.38 (0.01)	5.38 (0.01)	5.03 (0.01)	5.03 (0.01)
Breasts	8.77 (0.00)	0.54 (0.01)	9.13 (0.00)	9.13 (0.00)	9.50 (0.00)	9.50 (0.00)
Stomach wall	4.01 (0.01)	4.15 (0.00)	3.94 (0.01)	3.94 (0.01)	3.74 (0.01)	3.74 (0.01)
Liver	5.52 (0.00)	3.49 (0.00)	5.51 (0.00)	5.51 (0.00)	5.15 (0.00)	5.15 (0.00)
Gallbladder wall	1.88 (0.02)	6.94 (0.01)	1.88 (0.02)	1.88 (0.02)	1.72 (0.02)	1.72 (0.02)
Pancreas	0.95 (0.01)	6.81 (0.00)	0.95 (0.01)	0.95 (0.01)	0.93 (0.01)	0.93 (0.01)
Spleen	2.51 (0.01)	7.07 (0.00)	2.52 (0.01)	2.52 (0.01)	2.34 (0.01)	2.34 (0.01)
Kidneys	0.89 (0.01)	8.24 (0.00)	0.88 (0.01)	0.88 (0.01)	0.84 (0.01)	0.84 (0.01)
Adrenals	1.83 (0.02)	5.79 (0.01)	1.76 (0.02)	1.76 (0.02)	1.72 (0.02)	1.72 (0.02)
SI wall and contents	0.32 (0.01)	3.06 (0.00)	0.49 (0.01)	0.49 (0.01)	0.56 (0.01)	0.56 (0.01)
LI wall	0.44 (0.01)	4.26 (0.00)	0.62 (0.01)	0.62 (0.01)	1.24 (0.01)	1.24 (0.01)
LI contents	0.40 (0.01)	3.88 (0.00)	0.53 (0.01)	0.53 (0.01)	1.14 (0.01)	1.14 (0.01)
Ovaries	0.14 (0.07)	0.73 (0.03)	0.13 (0.08)	0.13 (0.08)	0.13 (0.08)	0.13 (0.08)
Bladder wall	0.08 (0.05)	0.26 (0.02)	0.06 (0.06)	0.06 (0.06)	0.04 (0.07)	0.04 (0.07)
Uterine wall	0.22 (0.02)	1.23 (0.01)	0.40 (0.01)	0.40 (0.01)	0.57 (0.01)	0.57 (0.01)
Uterine contents	0.17 (0.02)	0.86 (0.01)	0.33 (0.01)	0.33 (0.01)	0.37 (0.01)	0.37 (0.01)
Placenta	0.35 (0.03)	2.36 (0.01)	0.91 (0.01)	0.91 (0.01)	1.04 (0.01)	1.04 (0.01)

Organ dose (mGy/100 mAs)						
	RPI-P3 MDCT scan		RPI-P6 MDCT scan		RPI-P9 MDCT scan	
	Chest	Kidney	Chest	Chest	Chest	Chest
Skeleton	4.45 (0.00)	1.92 (0.00)	4.20 (0.00)	4.02 (0.00)	4.02 (0.00)	
Skin	1.37 (0.00)	0.84 (0.00)	1.36 (0.00)	1.29 (0.00)	1.29 (0.00)	
Remainder	1.61 (0.00)	1.39 (0.00)	1.55 (0.00)	1.42 (0.00)	1.42 (0.00)	
Fetal soft tissue	0.13 (0.04)	0.58 (0.02)	0.18 (0.02)	0.23 (0.02)	0.23 (0.01)	
Fetal skeleton	NA ^a	NA ^a	0.67 (0.02)	0.62 (0.02)	0.62 (0.01)	
Fetal brain	0.12 (0.09)	0.50 (0.04)	0.11 (0.05)	0.08 (0.05)	0.08 (0.04)	
Fetus total	0.13 (0.04)	0.57 (0.02)	0.21 (0.02)	0.26 (0.02)	0.26 (0.01)	

^a RPI-P3 does not contain fetal skeleton.