Published in final edited form as:

Phys Med Biol. 2017 March 07; 62(5): 1759–1776. doi:10.1088/1361-6560/aa5911.

A measurement-based generalized source model for Monte Carlo dose simulations of CT scans

Xin Ming¹, Yuanming Feng^{1,2}, Ransheng Liu^{1,2}, Chengwen Yang^{1,2}, Li Zhou³, Hezheng Zhai⁴, and Jun Deng^{5,*}

¹Department of Biomedical Engineering, Tianjin University, Tianjin, China

²Department of Radiation Oncology, Tianjin Medical University Cancer Institute & Hospital, Tianjin, China

³Center for Radiation Physics and Technology, Sichuan University West China Hospital, Chengdu, Sichuan, China

⁴Institute of Radiation Medicine Chinese Academy of Medical Sciences, Tianjin, China

⁵Department of Therapeutic Radiology, Yale University, New Haven, CT, USA

Abstract

The goal of this study is to develop a generalized source model (GSM) for accurate Monte Carlo dose simulations of CT scans based solely on the measurement data without *a priori* knowledge of scanner specifications. The proposed generalized source model consists of an extended circular source located at x-ray target level with its energy spectrum, source distribution and fluence distribution derived from a set of measurement data conveniently available in the clinic. Specifically, the central axis percent depth dose (PDD) curves measured in water and the cone output factors measured in air were used to derive the energy spectrum and the source distribution respectively with a Levenberg-Marquardt algorithm. The in-air film measurement of fan-beam dose profiles at fixed gantry was back-projected to generate the fluence distribution of the source model. A benchmarked Monte Carlo user code was used to simulate the dose distributions in water with the developed source model as beam input. The feasibility and accuracy of the proposed source model was tested on a GE LightSpeed and a Philips Brilliance Big Bore multi-detector CT (MDCT) scanners available in our clinic. In general, the Monte Carlo simulations of the PDDs in water and dose profiles along lateral and longitudinal directions agreed with the measurements within 4%/1mm for both CT scanners. The absolute dose comparison using two CTDI phantoms (16 cm and 32 cm in diameters) indicated a better than 5% agreement between the Monte Carlo-simulated and the ion chamber-measured doses at a variety of locations for the two scanners. Overall, this study demonstrated that a generalized source model can be constructed based only on a set of measurement data and used for accurate Monte Carlo dose simulations of patients' CT scans, which would facilitate patient-specific CT organ dose estimation and cancer risk management in the diagnostic and therapeutic radiology.

*Corresponding author: Jun Deng, Ph.D., Department of Therapeutic Radiology, Yale University, 15 York Street, LL508-Smilow, New Haven, CT 06510-3221, USA, Tel: 203-200-2013, Fax: 203-200-2054, jun.deng@yale.edu.

Keywords

CT; organ dose; source modeling; Monte Carlo simulation; patient-specific dosimetry

1. Introduction

Driven largely by the technological advances as well as a fee-for-service healthcare model, the use of computed tomography (CT) in cancer diagnosis and therapy has increased dramatically in the past thirty years. It is estimated that more than 78.7 million CT scans were performed in the United States alone in 2015, a 26-fold increase as compared to 3 million CT scans in 1980 (IMV, 2015). While radiation dose from a single CT scan to individual patient may decrease as a result of dose reduction efforts, the over-usage of CT scans in the clinic may lead to an increase in radiation exposure to the patient population, which may become a serious public health issue due to increased secondary cancer risk, particularly to the children (McNitt-Gray, 2002; Brenner and Hall, 2007; Valentin, 2007; Berrington de Gonzalez *et al.*, 2009; Pearce *et al.*, 2012; Mathews *et al.*, 2013; Leuraud *et al.*, 2015). Therefore, it is very important to accurately estimate the radiation doses to the individual patient from CT scans in order to mitigate the potential risk in cancer diagnosis and radiotherapy.

As in vivo organ dose measurement is not feasible, Monte Carlo (MC) simulation coupled to some computational anthropomorphic phantoms has been used as a more practical way to assess the radiation dose from CT scans. There have been numerous studies along this line comparing Monte Carlo simulation results to experimental measurements with various phantoms (Jarry *et al.*, 2003; DeMarco *et al.*, 2005; Staton *et al.*, 2006; DeMarco *et al.*, 2007; Deak *et al.*, 2008; Gu *et al.*, 2009; Ding *et al.*, 2010; Li *et al.*, 2011; Kim *et al.*, 2012; Lee *et al.*, 2012). Specifically, a full or energy spectrum-based Monte Carlo simulation was employed to compute the organ dose from CTs in both adult and pediatric patients, and even fetus, and to investigate the patient-specific cancer risk.

The goal of this study is to build a generalized source model for accurate Monte Carlo dose simulations of CT scans based only on the measurement data without *a priori* knowledge of scanner configurations from the manufacturers. With automatic beam commissioning, there will be no fine-tuning involved which usually takes a long time and requires Monte Carlo expertise. The required measurement data can be conveniently obtained in the clinic on any CT scanner, avoiding the need for any proprietary specifications from the manufacturers. With the developed source model as beam input for Monte Carlo dose engine, it will be possible to obtain the 3D dose distribution of any organ-at-risk (OAR) in individual patient from CT scans accurately and efficiently. The long-term goal of this research is to create a systematic and accurate method for patient-specific organ dose estimation of individual patients undergoing CT scans from different manufacturers.

2. Materials and Methods

2.1 Two CT scanners

In this study, two commercial CT scanners available in our clinic were used as test platforms: a GE LightSpeed (LS) 16-slice MDCT (GE Medical Systems, Milwaukee, WI) and a Philips Brilliance Big Bore (BB) 16-slice MDCT (Philips Medical Systems, Best, Netherlands). The proprietary information such as the geometry and materials of x-ray tube, inherent filters and collimators were not provided by the manufacturers. The GE LS 16-slice MDCT is a third-generation MDCT with 16 rows of 1.25 mm wide detectors. It offers the users six options of x-ray beam collimation: 16×1.25 mm (8×2.5 mm), 16×0.625 mm (8×1.25 mm), 2×0.625 mm and 1×5 mm. The scanner is capable of axial as well as helical scans with multiple pitch options. The system supports nominal x-ray energies of 80, 100, 120 and 140 kVp, equipped with the head and body bowtie filters. The Philips 16-slice MDCT provides three common collimation settings: 16×0.75 mm, 16×1.5 mm and 8×3 mm. The kilovolt peaks of x-ray tube used in this study were 80, 90, 120 and 140 kVp. The major characteristics of the two CT scanners have been listed in Table 1.

2.2 Monte Carlo source modeling

Monte Carlo method has proven to be the most accurate dose calculation algorithm so far for radiotherapy treatment planning (Nahum, 1988; Rogers and Bielajew, 1990; Mackie, 1990; Andreo, 1991; Mohan, 1997; Ma *et al.*, 1999; Arnfield *et al.*, 2000; Miften *et al.*, 2001). Generally speaking, there are two approaches to obtain the phase space of clinical beams for Monte Carlo dose calculations: a phase space approach and a source model approach (Rogers *et al.*, 1995; Ma *et al.*, 1997; Ma, 1998; Deng *et al.*, 2000). The phase space approach relies on explicitly simulating the transport of the particles through the treatment machine using the Monte Carlo technique, potentially accounting for all physical processes of clinical significance. However, Monte Carlo expertise is generally required to perform accurate simulations. Also, the limited number of particles in a phase space file produces an inherent systematic statistical uncertainty in the final dose distributions. In the past few years, we have developed various multiple-source models to represent the photon beams from Varian linear accelerator and have shown that a well-designed multiple-source model can be used to reproduce the dose distributions in a water phantom simulated with phase space as well as the measured dose distributions in water to within 3%/1mm for both MV photon beams (Deng *et al.*, 2000; Deng *et al.*, 2003; Deng *et al.*, 2004), and kV photon beams emanated from the onboard imager (OBI) system installed on a Varian Trilogy linear accelerator (Deng *et al.*, 2012a; Deng *et al.*, 2012b; Zhang *et al.*, 2012).

In this study, we propose to build a generalized source model to represent the kV photon beams emanated from the CT x-ray source. This is based on our previous experience with kV photon beams in the clinic and observations from Monte Carlo simulations that almost all of the photons contributing to the patient doses are from the x-ray target, and only a few percent of total photons are scattered by the collimating blades and bowtie filter. Hence, the proposed source model consists of an extended circular source located at the x-ray target level, with its characteristics defined by the energy spectrum, source distribution and fluence distribution. Both the energy spectrum and source distribution will be derived at the x-ray

target level, while the fluence distribution will be reconstructed at the source model plane located at 40 cm upstream above the isocenter. An in-house C++ code has been developed to perform an automatic beam commissioning to generate the source model based on a set of measurement data.

2.2.1 Required measurement data for source modeling—For a given tube voltage kVp/filtration of a given CT scanner, a minimum set of measurement data will be required to generate the source model, which includes (1) the central axis PDD distribution in water; (2) the dose profiles along lateral and longitudinal directions at isocenter level; and (3) in-air beam output measured at isocenter through a series of cylindrical cones.

In this study, as shown in Figure 1, all the measurements were performed at a fixed gantry with the x-ray tube stationary at 12 o'clock position and shooting downward. The central axis PDD was measured in a 30 cm × 30 cm × 30 cm PTW MP3 water tank (PTW, Freiburg, Germany), with depths from the surface (isocenter plane) to 20 cm in water. A PTW Semiflex 31010 ionization chamber with an active volume of 0.125 cc was used for charge collection at various depths in water (Figure 1a). The x-ray beam output was measured with a PTW Farmer chamber (0.6 cc, TW30013) at the isocenter along the central axis, 20 cm down below the lower surface of cones (Figure 1b). The cones used in this study were five stainless steel cones of various diameters (i.e., 6, 7.5, 10, 12.5 and 15 mm) made by Brainlab (Brainlab AG, Feldkirchen, Germany). Each cone was 10.5 cm in height, thick enough to block all of the x-ray transmission through the cone.

The EBT3 Gafchromic films (International Specialty Products, Wayne, NJ) of 8' × 10' were fixed and exposed at the isocenter of CT scanner with 2 cm of Styrofoam underneath to reduce the backscatter from the CT table. To study the impact of 2-cm Styrofoam on the backscatter from carbon fiber CT table, we have further conducted a series of Monte Carlo simulations with a 140-kVp energy spectrum from the source and scored the phase space downstream at various interfaces between air, 2-cm Styrofoam and 2-cm carbon fiber table. Specifically, we compared the particles scored at the same level with and without the 2-cm Styrofoam. Our MC simulations indicated that the backscatter of CT X-rays from carbon fiber CT table was less than 1% and the 2-cm Styrofoam further reduces the backscatter down to a negligible level. Two pieces of films were combined together to cover the elongated lateral profiles of fan-beam along x-axis (Figure 1c). The EBT3 films were first scanned with an EPSON GT-15000 flatbed scanner (EPSON America, Long Beach, CA) to extract the reflective images with the resolution of 300 dpi. Then a Matlab code was used to convert the pixel values of these images into the dose profiles at the CT isocenter.

2.2.2 Energy spectrum—The energy spectrum of x-ray photon beams is important for treatment planning dose calculation as well as diagnostic imaging analysis, allowing calculation of the half value layer, air kerma, mean energy and other properties. Unfortunately, it is often not available from the manufactures and is difficult to measure due to high fluence of photon beams in the clinic. Several groups have proposed some semi-empirical or empirical models to generate the energy spectrum of tungsten target as a function of x-ray tube parameters (Boone and Seibert, 1997; Meyer *et al.*, 2004; Siewerdsen *et al.*, 2004; Poludniowski *et al.*, 2009). However, additional proprietary information such as

the anode angle and the filter material and thickness are often required. In this study, we proposed to derive the x-ray energy spectrum based on the measured central-axis percent depth dose (PDD) distribution in water.

To start with, an EGSnrc/DOSXYZnrc Monte Carlo code has been employed to simulate the central axis depth dose distributions in water for a series of mono-energetic x-ray photon beams, with energy ranging from 11 keV to 150 keV. No PDDs were simulated for energy lower than 10 keV due to cutoff energy of 10 keV in Monte Carlo simulation. Specifically, a fan beam with 56 cm lateral span and 2 cm width along longitudinal direction was defined to mimic the fan beams observed in the clinical CT scanners. The measured PDD (PDD_{mea}) in water was considered as the weighted sum of the mono-energetic PDDs (PDD_{ikeV} , $i = \{1, 2, \dots, E_{max}\}$) in water, with the weighting factors (w_i , $i = \{1, 2, \dots, E_{max}\}$) being the energy spectrum specific to the x-ray photon beams. To solve this nonlinear multi-dimensional fitting problem, we applied a Levenberg-Marquardt algorithm to minimize an objective function as defined in Equation 1 (Press *et al.*, 1992). The minimization of the χ^2 value for the fitting between the measured PDD and the fitted PDD would yield the energy spectrum. Based on our previous Monte Carlo study, an empirical function (Equation 2) with three variables (a , b , c) was proposed to represent the energy spectrum of x-ray photon beams in CT scanners.

$$\begin{aligned} \text{objective function} &= PDD_{mea} - \sum_{E_{max}}^i PDD_{ikeV} * w_i; \\ \text{Energy Spectrum} &= \{w_1, w_2, \dots, w_i, \dots, w_{E_{max}}\}. \end{aligned} \quad 1$$

$$\text{Energy Spectrum} = \exp\left(a + b * i + c * \frac{\ln i}{i^2}\right), i = \{1, 2, \dots, E_{max}\}. \quad 2$$

The parameter i denotes the discrete level of mono-energetic x-rays from 11 keV to the maximum keV, while w_i indicates the weighting factor for the corresponding mono-energy. In order to further improve the accuracy of the derived energy spectrum, the characteristic radiation peaks of the tungsten target were also considered in the fitting with the corresponding reference information obtained from the literature (Johns and Cunningham, 1983).

2.2.3 Source distribution—As the scatter photons make negligible contribution to the patient doses as compared to the primary photons, we only considered the primary photons and appropriated them with a modified Gaussian distribution without considering the heel effects in CT X-rays. As shown in Figure 2a, the beam output measured at the isocenter with an ion chamber was proportional to the area integral over the source distribution extended by the solid angle defined by a particular size of cone. If the source distribution along radial direction was represented by a modified Gaussian function (Equation 3) with two parameters (A , B), the contribution from the source would be proportional to the area integral of this equation with respect to the off-axis distance.

$$\begin{aligned} \text{Objective function} &= \text{Out Factor}_{\text{mea}} - \int \text{Source Distribution}(x) \\ \text{Source Distribution} &= \exp\left(-\frac{(x-A)^2}{B^2}\right) \end{aligned} \quad 3$$

In the Equation 3, x represents the distance from the center to the edge in the source distribution. In this study, we defined the output factors as the ratios of the signal detected by ionization chamber at the isocenter with the cone of a certain size to the signal of open beam, which could be represented as the ratios of the area integral of source distribution with respect to a certain off-axis distance to the total area integral of source distribution. Therefore, with output factors being a function of two fitting parameters (A , B) of the modified Gaussian function, the Levenberg-Marquardt algorithm has been employed to do the nonlinear least-squares fitting to determine the primary photon source distribution.

2.2.4 Fluence distribution—The fluence distribution of photons was determined through the film measurement of dose profile at the isocenter. As shown in Figure 1c, two films were combined together to measure the dose profiles along lateral and longitudinal directions of the fan-beam from CT scanners. The obtained dose profiles were then back-projected onto a hypothetical fluence distribution plane, located at 40 cm above the isocenter level (Figure 2b). We intentionally set the fluence distribution plane outside the CT scanner and between the target and CT isocenter, leaving enough space for the large-sized patients or immobilization devices. The obtained fluence distribution will be sampled by Monte Carlo algorithm to transport millions of particles from the source to the patient anatomy with proper probability.

2.3 Monte Carlo simulations with source models

With the required data measured on the GE and Philips MDCT scanners as input, we derived a source model for each scanner with our automatic beam commissioning code. Each source model consists of a detailed numerical description of the energy spectrum, source distribution, fluence distribution, location and shape of each source for a particular CT head. An EGS4/BEAM Monte Carlo code, MCSIM, was employed to reconstruct the photon beams from the generated source model and calculate the dose distributions in water as well as two CTDI phantoms (Nelson *et al.*, 1985; Rogers *et al.*, 1995; Ma *et al.*, 2002). In all Monte Carlo simulations, the energy cutoffs for electrons (ECUT) and photons (PCUT) and the energy thresholds for δ -ray production (AE) and bremsstrahlung production (AP) were set as ECUT = AE = 521 keV and PCUT = AP = 10 keV, respectively. The voxel size ranged from 0.25 cm $\times$ 0.25 cm $\times$ 0.25 cm in dose profile simulations to larger steps along depth direction in depth dose simulations. The calculation time of each Monte Carlo simulation was about 4 hours on a single CPU workstation in order to achieve a statistical uncertainty (1σ) of less than 2%. The benchmark results of EGS4/MCSIM have been reported previously (Li *et al.*, 2000).

In simulating dose distribution in the CTDI phantom with MCSIM, a series of 12 co-planar fields around the gantry rotation axis with an interval of 30° were simulated to mimic the axial mode of CT acquisition. In some earlier works, Gu *et al.* (Gu *et al.*, 2009) pointed out

that 16 X-ray projections along the CT scan trajectory would be enough to ensure the source modeling accuracy. Khursheed *et al.* (Khursheed *et al.*, 2002) also found out that it is sufficient to apply 18 sources to approximate the continuous axial movement of the X-ray source, without significantly affecting the calculated organ doses. We also tested 16 sources, 360 sources and 15 degrees shifting of starting angle for Monte Carlo simulations and our results indicated less than 2.9% dose differences compared to 12-sources adopted in this method. After each 360° gantry rotation, the phantom isocenter moved by a certain distance, equivalent to the table movement. In helical scan mode, the pitch value was considered in computing the incremental table movement (i.e., isocenter movement) between two consecutive gantry fields. The protocol-specific settings for LightSpeed and Brilliance Big Bore scanners are listed in Table 1.

To convert Monte Carlo simulations into absolute doses, absorbed doses were first measured at isocenter of a 5-cm diameter acrylic ball phantom per TG-61 protocol with an EXRADIN A12 ionization chamber (Standard Imaging Inc., Middleton, WI) for the various CT scan protocols listed in Table 1 (Ma *et al.*, 2001). Monte Carlo simulation was then conducted to a chamber volume inside the ball phantom with our proposed source model. The ratio of the measured dose to the Monte Carlo simulation result yielded a conversion factor, specific to the corresponding scan protocol and the source model. With the conversion factors, the Monte Carlo simulation results could be converted into absolute doses in water and in the CTDI phantom.

2.4 Validation of Monte Carlo simulations

To evaluate the feasibility and accuracy of the proposed source model, the absolute doses in two CTDI phantoms were measured and compared with Monte Carlo simulations for a variety of protocols on both CT scanners. For the GE LightSpeed scanner, the absorbed doses at 10 locations in a CT dose nested phantom (Model 76-424-4156ST, Supertech, Elkhart, IN) were measured with an EXRADIN A12 ionization chamber (Figure 3a). The diameter of peripheral cylinder is 32 cm. For the Philips Brilliance Big Bore scanner, a standard head CTDI dosimetry phantom with a diameter of 16 cm was used to measure the CT doses at 4 locations with an EXRADIN A12 ionization chamber (Figure 3b). The two phantoms have also been scanned into treatment planning system and converted into digital phantoms for Monte Carlo simulations with MCSIM. The MCSIM simulation results of these two phantoms were first converted into absolute doses following the aforementioned procedure, and then compared with the measurement data for validation.

3. Results

3.1 Energy spectra of photon beams in CT scanners

The results of all fitting parameters in Equations 2 and 3 were shown in Table 2. The derived energy spectra from the measured PDDs in water as proposed in this study were shown in Figure 4. As a reference, two publically available codes: Spektr (Siewerdsen *et al.*, 2004) and SpekCalc (Poludniowski *et al.*, 2009) were used to generate the energy spectra and compute the mean energy for the corresponding GE and Philips CT scanners in this study. Specifically, the Boone/Fewell tube was chosen in Spektr with a HVL of 7 mm aluminum.

In SpekCalc, the tube angle was set at 12 degrees with a HVL of 7 mm aluminum. The overall shapes of the derived energy spectra were similar to those generated with Spektr and SpekCalc. The mean energies of our derived spectra were 0.6% to 10% higher than those determined with Spektr and SpekCalc for 80 to 140 kVp photon beams. These discrepancies may be caused by some specific settings in Spektr and SpekCalc related to the manufacturers' specifications of x-ray tubes, which are not available to us. McMillan *et al.* have also reported that their measurement-derived spectra for cone-beam CT were more than 10% harder than the spectra extracted with Spektr (McMillan *et al.*, 2013).

3.2 Central axis depth dose in water

The central axis depth dose comparison in water between the measurement results and the Monte Carlo simulated results based on our developed source models are shown in Figure 5 for GE LightSpeed 16-slice MDCT and in Figure 6 for Philips Brilliance Big Bore 16-slice MDCT. Overall, the agreement between the measurements and the Monte Carlo simulations is clinically acceptable for all the photon beams being investigated, with less than 5% discrepancy from the surface to 5 cm in water and less than 1% discrepancy from 5 cm to 20 cm depth. The larger discrepancies at the shallow depths were most likely caused by the large uncertainty in the measurements of PDDs in water with ionization chamber for kilo-voltage x-ray beams. The removal of lower energy scatter photons in our source modeling can also contribute to the slightly harder energy spectra shown in Figure 4 and higher surface dose observed in Figure 5 and 6.

3.3 Dose profiles along lateral and longitudinal directions

The lateral and longitudinal dose profiles at isocenter plane obtained from the Monte Carlo simulations were compared with the radiochromic film measurements. As shown in Figure 7 and Figure 8, the agreement in lateral dose profiles are excellent for both GE and Philips CT scanners with less than 4% discrepancies between the Monte Carlo simulations and film measurements for a range of 40 cm along x-axis. The 2-cm longitudinal dose profile comparisons are clinically acceptable with better than 4% agreement between the Monte Carlo simulations and film measurements (Figures 9 & 10). In penumbra region, the distance-to-agreement (DTA) between the Monte Carlo simulations and the film measurements are less than 1 mm. The lower sampling resolution (pixel size 1.5 mm × 1.5 mm) in the film measurements and the film heterogeneity may lead to some discrepancies observed in high dose region. The removal of scatter photons in our source modeling may also lead to slightly sharper penumbra in the longitudinal dose profiles as compared to the film measurements.

3.4 Absolute dose comparison in two CTDI phantoms

With a series of source models generated for the various protocols of both GE and Philips scanners, the absolute doses were compared in the 32-cm CTDI phantom for the GE scanner and in a 16-cm CTDI phantom for the Philips scanner. Considering the large uncertainties in the measurements and Monte Carlo simulations for the points proximal to the couch, only the point doses located at 12, 3 and 9 o'clock positions as shown in Figure 3 were compared. Shown in Table 3 and Table 4, the agreement between the Monte Carlo simulations and the ion chamber measurements for the three site-specific scan protocols available on the two

scanners is within 5% for all considered data points. Note that we obtained this absolute dose agreement with our source model derived from a set of measurement data without *a priori* knowledge of scanner specifications, and our results were comparable to those reported by other groups who carried out full Monte Carlo simulations of x-ray tubes with manufacturers' specifications incorporated (Jarry *et al.*, 2003; DeMarco *et al.*, 2005; Gu *et al.*, 2009; Lee *et al.*, 2011; Lee *et al.*, 2012; Kim *et al.*, 2012).

4. Discussion

Source modeling has been widely used in modern radiation therapy treatment planning dose calculation in patient anatomy for MV treatment systems such as linear accelerators (Chetty *et al.*, 2000; Deng *et al.*, 2000), Tomotherapy (Yuan *et al.*, 2015) and CyberKnife (Deng *et al.*, 2003; Deng *et al.*, 2004), as well as kV imaging system such as kVCBCT (Spezi *et al.*, 2011; McMillan *et al.*, 2013).

For CT X-rays, Turner *et al.* previously suggested a noninvasive approach to derive energy spectrum and filtration description based on the measured half-value-layer (HVL) of X-ray photon beams specific to the CT scanner (Turner *et al.*, 2009). However, up to 200 CT scans would be measured for four tube voltages and two bowtie filters. Boone applied a real-time dosimeter near the edge of the scanner's field of view (FOV) to measure the kerma rate versus time in CT scans (Boone, 2009). McKenney *et al.* and Whiting *et al.* applied this method to characterize bowtie profiles with different sizes of radiation meters (McKenney *et al.*, 2011; Whiting *et al.*, 2014). Randazzo and Tambasco further developed a rapid noninvasive method to construct virtual equivalent source models by measuring the spatially varying HVLs with concentrically cylindrical Al filters (Randazzo and Tambasco, 2015). Recently, Li *et al.* proposed a noninvasive technique for directly measuring the CT bowtie filter attenuation with a linear array x-ray detector (Li *et al.* 2015).

In this work, we proposed a new method to construct a generalized source model based solely on measurement data to facilitate imaging dose calculation in patient anatomy as a result of CT scans. Without *a priori* knowledge of scanner specifications, nor depending on numerous measurements of CT X-ray beams with advanced equipment, our approach represents a different attempt to develop a full source model (i.e., energy spectrum, fluence distribution, source distribution etc.) for CT dosimetry based on a limited set of data, striking a balance between MC dose simulation accuracy and clinical practicality.

The assessment of imaging doses from CT scans has been a hot topic in the literature as well as among the general public due to the concern of the increased cancer risk from radiation exposure (Dougeni *et al.*, 2012). Compared to all the early works performing either full Monte Carlo simulations (Kim *et al.*, 2012) or energy spectrum-based Monte Carlo simulations for the CT dose estimation (Jarry *et al.*, 2003; DeMarco *et al.*, 2005; DeMarco *et al.*, 2007; Gu *et al.*, 2009; Ding *et al.*, 2010; Lee *et al.*, 2011; Lee *et al.*, 2012), we have shown that the developed generalized source model is accurate in calculating CT doses in the phantoms and in water for a variety of site-specific scan protocols at two widely-used CT scanners.

One of the major advantages of our approach is that the developed source model is purely measurement-based and does not depend on the proprietary specifications from the CT manufacturers. In addition, the required measurement data for source modeling (as defined in section 2.2.1) can be easily obtained in the clinic via ion chamber measurement and film dosimetry. With measured data as input, the source model can be generated with our automatic beam commission procedure, which greatly reduces the time and efforts spent in fine-tuning the Monte Carlo simulations. The validation results between the measurements and the Monte Carlo simulations with our source models have confirmed that our source models can accurately replicate the dose distribution from the radiation beams of CT scanners in the phantoms and in water.

However, one limitation of the source modeling approach is that the accuracy of the generated source modeling largely depends on the quality and accuracy of the measurement data, which in our case consists of the PDD in water, the lateral and longitudinal dose profiles at isocenter planes and cone output factors at isocenter. The measurement inaccuracy will be directly reflected in the source model parameters such as energy spectrum, fluence profile and source distribution, which would subsequently affect the 3D dose distributions in not only phantoms but also patient anatomy. Hence, appropriate and accurate measurements would be highly desired to ensure the model accuracy and consistency among the same type of CT scanner.

Moving forward, we plan to carry out a systematic study with our developed source models on the CT doses to all the clinically relevant organs-at-risk in the diagnostic and therapeutic radiology for a full spectrum of patients (infants to obese) as a result of 3D or 4D CT scans. The inclusion of concomitant doses from CT scans as well as kVCBCT scans into the total treatment planning process could help the clinicians make informed decision for each individual patient and reduce the potential cancer risk associated with imaging doses.

5. Conclusions

We have proposed a novel method to generate a generalized source model based solely on the measurement data for Monte Carlo CT dosimetry simulations. The Monte Carlo simulations with the developed source models have been validated with a variety of dose measurements in water and at two CTDI phantoms. Our work indicates that it is feasible to derive beam characteristics from a set of measurement data without *a priori* knowledge of x-ray tube configurations, and to replicate the dose distributions from CT x-ray beams with a generalized source model. The developed source models can be used to facilitate patient-specific CT dose estimations and cancer risk management in the clinic.

References

- Andreo P. Monte Carlo techniques in medical radiation physics. *Phys Med Biol.* 1991; 36:861–920. [PubMed: 1886926]
- Arnfield MR, Siantar CH, Siebers J, Garmon P, Cox L, Mohan R. The impact of electron transport on the accuracy of computed dose. *Med Phys.* 2000; 27:1266–74. [PubMed: 10902555]
- Berrington de Gonzalez A, Mahesh M, Kim KP, Bhargavan M, Lewis R, Mettler F, Land C. Projected cancer risks from computed tomographic scans performed in the United States in 2007. *Arch Intern Med.* 2009; 169:2071–7. [PubMed: 20008689]

- Boone JM, Seibert JA. An accurate method for computer-generating tungsten anode x-ray spectra from 30 to 140 kV. *Med Phys.* 1997; 24:1661–70. [PubMed: 9394272]
- Boone JM. Method for evaluating bow tie filter angle-dependent attenuation in CT: Theory and simulation results. *Med Phys.* 2009; 37:40–8.
- Brenner DJ, Hall EJ. Computed tomography--an increasing source of radiation exposure. *N Engl J Med.* 2007; 357:2277–84. [PubMed: 18046031]
- Chetty I, DeMarco JJ, Solberg TD. A virtual source model for Monte Carlo modeling of arbitrary intensity distributions. *Med Phys.* 2000; 27:166–72. [PubMed: 10659754]
- Deak P, van Straten M, Shrimpton PC, Zankl M, Kalender WA. 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]
- DeMarco JJ, Cagnon CH, Cody DD, Stevens DM, McCollough CH, Zankl M, Angel E, McNitt-Gray MF. Estimating radiation doses from multidetector CT using Monte Carlo simulations: effects of different size voxelized patient models on magnitudes of organ and effective dose. *Phys Med Biol.* 2007; 52:2583–97. [PubMed: 17440254]
- Deng J, Chen Z, Roberts KB, Nath R. Kilovoltage imaging doses in the radiotherapy of pediatric cancer patients. *Int J Radiat Oncol Biol Phys.* 2012a; 82:1680–8. [PubMed: 21477943]
- Deng J, Chen Z, Yu JB, Roberts KB, Peschel RE, Nath R. Testicular doses in image-guided radiotherapy of prostate cancer. *Int J Radiat Oncol Biol Phys.* 2012b; 82:e39–47. [PubMed: 21489702]
- Deng J, Guerrero T, Ma CM, Nath R. Modelling 6 MV photon beams of a stereotactic radiosurgery system for Monte Carlo treatment planning. *Phys Med Biol.* 2004; 49:1689–704. [PubMed: 15152924]
- Deng J, Jiang SB, Kapur A, Li J, Pawlicki T, Ma CM. Photon beam characterization and modelling for Monte Carlo treatment planning. *Phys Med Biol.* 2000; 45:411–27. [PubMed: 10701512]
- Deng J, Ma CM, Hai J, Nath R. Commissioning 6 MV photon beams of a stereotactic radiosurgery system for Monte Carlo treatment planning. *Med Phys.* 2003; 30:3124–34. [PubMed: 14713079]
- Ding A, Gu J, Trofimov AV, Xu XG. Monte Carlo calculation of imaging doses from diagnostic multidetector CT and kilovoltage cone-beam CT as part of prostate cancer treatment plans. *Med Phys.* 2010; 37:6199–204. [PubMed: 21302776]
- Dougeni E, Faulkner K, Panayiotakis G. A review of patient dose and optimisation methods in adult and paediatric CT scanning. *Eur J Radiol.* 2012; 81:e665–83. [PubMed: 21684099]
- Gu J, Bednarz B, Caracappa PF, Xu XG. 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. *Phys Med Biol.* 2009; 54:2699–717. [PubMed: 19351983]
- IMV. CT Market Summary Report. Des Plaines, IL: IMV Medical Information Division; 2015.
- Jarry G, DeMarco JJ, Beifuss U, Cagnon CH, McNitt-Gray MF. A Monte Carlo-based method to estimate radiation dose from spiral CT: from phantom testing to patient-specific models. *Phys Med Biol.* 2003; 48:2645–63. [PubMed: 12974580]
- Johns, HE., Cunningham, JR. *Physics of Radiology*, Fourth Edition. Springfield, IL, USA: Charles C Thomas; 1983.
- Kim S, Song H, Movsas B, Chetty IJ. Characteristics of x-ray beams in two commercial multidetector computed tomography simulators: Monte Carlo simulations. *Med Phys.* 2012; 39:320–9. [PubMed: 22225302]
- Lee C, Kim KP, Long D, Fisher R, Tien C, Simon SL, Bouville A, Bolch WE. Organ doses for reference adult male and female undergoing computed tomography estimated by Monte Carlo simulations. *Med Phys.* 2011; 38:1196–206. [PubMed: 21520832]
- Lee C, Kim KP, Long DJ, Bolch WE. Organ doses for reference pediatric and adolescent patients undergoing computed tomography estimated by Monte Carlo simulation. *Med Phys.* 2012; 39:2129–46. [PubMed: 22482634]

- Leuraud K, Richardson DB, Cardis E, Daniels RD, Gillies M, O'Hagan JA, Hamra GB, Haylock R, Laurier D, Moissonnier M, Schubauer-Berigan MK, Thierry-Chef I, Kesminiene A. Ionising radiation and risk of death from leukaemia and lymphoma in radiation-monitored workers (INWORKS): an international cohort study. *Lancet Haematol*. 2015; 2:e276–81. [PubMed: 26436129]
- Li JS, Pawlicki T, Deng J, Jiang SB, Mok E, Ma CM. Validation of a Monte Carlo dose calculation tool for radiotherapy treatment planning. *Phys Med Biol*. 2000; 45:2969–85. [PubMed: 11049183]
- Li X, Samei E, Segars WP, Sturgeon GM, Colsher JG, Toncheva G, Yoshizumi TT, Frush DP. Patient-specific radiation dose and cancer risk estimation in CT: part II. Application to patients. *Med Phys*. 2011; 38:408–19. [PubMed: 21361209]
- Li X, Shi JQ, Zhang D, Singh S, Padole A, Otrakji A, Kalra MK, Xu XG, Liu B. A new technique to characterize CT scanner bow-tie filter attenuation and applications in human cadaver dosimetry simulations. *Med Phys*. 2015; 42:6274–82. [PubMed: 26520720]
- Ma CM. Characterization of computer simulated radiotherapy beams for Monte Carlo treatment planning. *Radiation Physics and Chemistry*. 1998; 53:329–44.
- Ma CM, Coffey CW, DeWerd LA, Liu C, Nath R, Seltzer SM, Seuntjens JP. American Association of Physicists in M. AAPM protocol for 40–300 kV x-ray beam dosimetry in radiotherapy and radiobiology. *Med Phys*. 2001; 28:868–93. [PubMed: 11439485]
- Ma CM, Faddegon BA, Rogers DW, Mackie TR. Accurate characterization of Monte Carlo calculated electron beams for radiotherapy. *Med Phys*. 1997; 24:401–16. [PubMed: 9089592]
- Ma CM, Li JS, Pawlicki T, Jiang SB, Deng J, Lee MC, Koumrian T, Luxton M, Brain S. A Monte Carlo dose calculation tool for radiotherapy treatment planning. *Phys Med Biol*. 2002; 47:1671–89. [PubMed: 12069086]
- Ma CM, Mok E, Kapur A, Pawlicki T, Findley D, Brain S, Forster K, Boyer AL. Clinical implementation of a Monte Carlo treatment planning system. *Med Phys*. 1999; 26:2133–43. [PubMed: 10535630]
- Mackie, TR. The dosimetry of ionizing radiation Volume III. Kase, KR, Bjarnagard, BE., Attix, FH., editors. New York: Academic; 1990. p. 541–620.
- Mathews JD, Forsythe AV, Brady Z, Butler MW, Goergen SK, Byrnes GB, Giles GG, Wallace AB, Anderson PR, Guiver TA, McGale P, Cain TM, Dowty JG, Bickerstaffe AC, Darby SC. Cancer risk in 680,000 people exposed to computed tomography scans in childhood or adolescence: data linkage study of 11 million Australians. *BMJ*. 2013; 346:f2360. [PubMed: 23694687]
- McKenney SE, Nosrati A, Gelskey D, Yang K, Huang SY, Chen L, Boone JM. Experimental validation of a method characterizing bow tie filters in CT scanners using a real-time dose probe. *Med Phys*. 2011; 38:1406–15. [PubMed: 21520852]
- McMillan K, McNitt-Gray M, Ruan D. Development and validation of a measurement-based source model for kilovoltage cone-beam CT Monte Carlo dosimetry simulations. *Med Phys*. 2013; 40:111907. [PubMed: 24320440]
- McNitt-Gray MF. AAPM/RSNA Physics Tutorial for Residents: Topics in CT. Radiation dose in CT. *Radiographics*. 2002; 22:1541–53. [PubMed: 12432127]
- Meyer P, Buffard E, Mertz L, Kennel C, Constantinesco A, Siffert P. Evaluation of the use of six diagnostic X-ray spectra computer codes. *British Journal of Radiology*. 2004; 77:224–30. [PubMed: 15020364]
- Miften M, Wiesmeyer M, Kapur A, Ma CM. Comparison of RTP dose distributions in heterogeneous phantoms with the BEAM Monte Carlo simulation system. *J Appl Clin Med Phys*. 2001; 2:21–31. [PubMed: 11674835]
- Mohan, R. Why Monte Carlo?. 12th International Conference on the Use of Computers in Radiation Therapy; Salt Lake City, USA. 1997. p. 16–8.
- Nahum, AE. Monte Carlo Transport of Electrons and Photons. Jenkins, TM, Nelson, WR., Rindi, A., editors. New York: Plenum Press; 1988. p. 3–20.
- Nelson, W., Hirayama, H., Rogers, DW. The EGS4 code system. Stanford linear accelerator code SLAC-265. Stanford, CA: 1985.
- Pearce MS, Salotti JA, Little MP, McHugh K, Lee C, Kim KP, Howe NL, Ronckers CM, Rajaraman P, Sir Craft AW, Parker L, Berrington de Gonzalez A. Radiation exposure from CT scans in

- childhood and subsequent risk of leukaemia and brain tumours: a retrospective cohort study. *Lancet*. 2012; 380:499–505. [PubMed: 22681860]
- Poludniowski G, Landry G, DeBlois F, Evans PM, Verhaegen F. SpekCalc: a program to calculate photon spectra from tungsten anode x-ray tubes. *Phys Med Biol*. 2009; 54:N433–8. [PubMed: 19724100]
- Press, WH., Teukolsky, SA., Vetterling, WT., Flannery, BP. *Numerical Recipes in C. The Art of Scientific Computing*. 2. Cambridge University; 1992.
- Randazzo M, Tambasco M. A rapid noninvasive characterization of CT x-ray sources. *Med Phys*. 2015; 42:3960–8. [PubMed: 26133596]
- Rogers, DW., Bielajew, AF. *The dosimetry of ionizing radiation Volume III*. Kase, KR.Bjarnsgard, BE., Attix, FH., editors. New York: Academic; 1990. p. 427-539.
- Rogers DW, Faddegon BA, Ding GX, Ma CM, We J, Mackie TR. BEAM: a Monte Carlo code to simulate radiotherapy treatment units. *Med Phys*. 1995; 22:503–24. [PubMed: 7643786]
- Siewerdsen JH, Waese AM, Moseley DJ, Richard S, Jaffray DA. Spektr: A computational tool for x-ray spectral analysis and imaging system optimization. *Medical Physics*. 2004; 31:3057–67. [PubMed: 15587659]
- Spezi E, Volken W, Frei D, Fix MK. A virtual source model for kilo-voltage cone beam CT: source characteristics and model validation. *Med Phys*. 2011; 38:5254–63. [PubMed: 21978069]
- 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]
- Turner AC, Zhang D, Kim HJ, DeMarco JJ, Cagnon CH, Angel E, Cody DD, Stevens DM, Primak AN, McCollough CH, McNitt-Gray MF. A method to generate equivalent energy spectra and filtration models based on measurement for multidetector CT Monte Carlo dosimetry simulations. *Med Phys*. 2009; 36:2154–64. [PubMed: 19610304]
- Valentin J. Managing patient dose in multi-detector computed tomography (MDCT). ICRP Publication 102. *Ann ICRP*. 2007; 37:1–79. iii.
- Whiting BR, Evans JD, Dohatcu AC, Williamson JF, Polite DG. Measurement of bow tie profiles in CT scanners using a real-time dosimeter. *Med Phys*. 2014; 41:101915. [PubMed: 25281967]
- Yuan J, Rong Y, Chen Q. A virtual source model for Monte Carlo simulation of helical tomotherapy. *J Appl Clin Med Phys*. 2015; 16:4992. [PubMed: 25679157]
- Zhang Y, Yan Y, Nath R, Bao S, Deng J. Personalized assessment of kV cone beam computed tomography doses in image-guided radiotherapy of pediatric cancer patients. *Int J Radiat Oncol Biol Phys*. 2012; 83:1649–54. [PubMed: 22285667]

Figure 1. Experimental setup for required data measurements: (a) PDD distribution measurement, (b) the cone output factors measurement, and (c) dose profile measurement.

Figure 2. Schematic diagrams showing the source model and the measurements of (a) cone output factors and (b) dose profiles at isocenter.

Figure 3. Schematic diagrams of experimental setups of two CTDI phantoms used in the dose evaluation for (a) GE CT scanner and (b) Philips CT scanner.

Figure 4.

Comparisons of energy spectra of 4 energy beams between the proposed method and the commercial software of SpekCalc and Spektr for (a) GE LightSpeed 16 MDCT and (b) Philips Brilliance Big Bore 16 MDCT. (c) and (d) are the enlarged views of (a) and (b), respectively.

Figure 5.

Comparisons of percent depth dose distributions in water between the Monte Carlo simulations and the ion chamber measurements for (a) 80, (b) 100, (c) 120 and (d) 140 kVp of GE CT scanner. The measurements were shown with black solid lines while the Monte Carlo simulations were shown as blue symbols. The error bars representing one standard deviation of the Monte Carlo simulations were less than 0.3% and hardly visible.

Figure 6.

Comparisons of percent depth dose distributions in water between the Monte Carlo simulations and the ion chamber measurements for (a) 80, (b) 90, (c) 120 and (d) 140 kVp of Philips CT scanner. The measurements were shown with black solid lines while the Monte Carlo simulations were shown as blue symbols. The error bars representing one standard deviation of the Monte Carlo simulations were less than 0.3% and hardly visible.

Figure 7.

Comparisons of lateral dose profiles between the Monte Carlo simulations and the radiochromic film measurements for (a) 80, (b) 100, (c) 120 and (d) 140 kVp of GE CT scanner. The measurements were shown with black solid lines while the Monte Carlo simulations were shown as blue symbols. The error bars representing one standard deviation of the Monte Carlo simulations were less than 0.3% and hardly visible.

Figure 8.

Comparisons of lateral dose profiles between the Monte Carlo simulations and the radiochromic film measurements for (a) 80, (b) 90, (c) 120 and (d) 140 kVp of Philips CT scanner. The measurements were shown with black solid lines while the Monte Carlo simulations were shown as blue symbols. The error bars representing one standard deviation of the Monte Carlo simulations were less than 0.3% and hardly visible.

Figure 9.

Comparisons of longitudinal dose profiles between the Monte Carlo simulations and the radiochromic film measurements for (a) 80, (b) 100, (c) 120 and (d) 140 kVp of GE CT scanner. The measurements were shown with black solid lines while the Monte Carlo simulations were shown as blue symbols. The error bars representing one standard deviation of the Monte Carlo simulations were less than 0.3% and hardly visible.

Figure 10.

Comparisons of longitudinal dose profiles between the Monte Carlo simulations and the radiochromic film measurements for (a) 80, (b) 90, (c) 120 and (d) 140 kVp of Philips CT scanner. The measurements were shown with black solid lines while the Monte Carlo simulations were shown as blue symbols. The error bars representing one standard deviation of the Monte Carlo simulations were less than 0.3% and hardly visible.

Table 1
Protocols used in this study for GE LightSpeed (LS) and Philips Brilliance Big Bore (BB) CT scanners

Protocol	Scan type	kVp	mAs	Slices	Thickness (mm)
LightSpeed					
Head & Neck	Helical (pitch = 0.937)	120	304	4	2.5
Chest	Axial	140	250	4	2.5
Prostate	Axial	140	250	4	2.5
Brilliance Big Bore					
Head	Axial	120	260	4	6
Thorax	Axial	120	250	4	3
Pelvis	Axial	120	325	4	6

Author Manuscript

Author Manuscript

Author Manuscript

Author Manuscript

Table 2

The fitting parameters in Equations 2 and 3.

GE	Equation 2			Equation 3	
	a	b	c	A	B
80kV	10.9415	-224.742	0.002695	2.08056	11.3406
100kV	6.30369	-132.861	0.002201	1.89775	11.2869
120kV	3.42992	-83.4039	0.001746	1.74723	11.2430
140kV	2.04837	-59.8876	0.001529	1.62724	11.2090
Philips					
80kV	5.98179	-139.3241	0.001543	2.06999	17.3091
90kV	2.76954	-82.50270	0.001051	2.09762	17.3040
120kV	1.07515	-46.40817	0.000739	2.10729	17.1454
140kV	1.66053	-46.89160	0.000887	2.10984	17.0691

Table 3

Dose comparison between Monte Carlo simulations and measurements on a 32-cm CTDI phantom for GE LightSpeed 16 MDCT

Label	Head			Chest			Prostate		
	Mea	Sim	%Diff.	Mea	Sim	%Diff.	Mea	Sim	%Diff.
1	1.89	1.80	-4.8	2.78	2.77	-0.4	2.90	2.76	-4.8
2	2.17	2.16	-0.5	3.26	3.19	-2.1	3.27	3.17	-3.1
3	2.06	2.05	-0.5	3.02	3.09	2.3	3.05	3.07	0.7
4	2.08	2.06	-1.0	3.13	3.06	-2.2	3.15	3.04	-3.5
5	2.59	2.65	2.3	3.90	3.82	-2.1	3.91	3.86	-1.3
6	2.41	2.51	4.1	3.60	3.72	3.3	3.61	3.69	2.2
7	2.57	2.52	-1.9	3.63	3.64	0.3	3.69	3.67	-0.5
8	3.47	3.54	2.0	5.19	5.01	-3.5	5.21	5.09	-2.3
9	3.45	3.62	4.9	4.97	4.76	-4.2	5.02	4.82	-4.0
10	3.43	3.55	3.5	4.94	4.77	-3.4	5.02	4.80	-4.4

Table 4

Dose comparison between Monte Carlo simulations and measurements on a 16-cm CTDI phantom for Philips Brilliance Big Bore 16 MDCT

Label	Head			Thorax			Pelvis		
	Mea	Sim	%Diff.	Mea	Sim	%Diff.	Mea	Sim	%Diff.
1	3.60	3.60	0.0	1.00	1.00	0.0	4.61	4.61	0.0
2	4.34	4.28	-1.4	1.20	1.19	-1.2	5.42	5.48	1.1
3	4.13	4.08	-1.1	1.10	1.13	2.3	5.16	5.23	1.3
4	4.03	4.09	1.3	1.09	1.13	3.5	5.28	5.23	-1.0