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Abstract. The bowtie filter is an essential element of computed tomography scanners. Implementation of this filter in a Monte Carlo dosimetry platform can be based on Turner's method, which describes how to measure the filter thickness and relate the x-ray beam as a function of bowtie angle to the central beam. In that application, the beam hardening is accounted for by means of weighting factors that are associated to the photons according to their position (fan angle) and energy. We assessed an alternative approximation in which the photon spectrum is given a fan angle-dependent scaling factor. The aim of our investigation was to evaluate the effects on dose accuracy estimation when using the gold standard bowtie filter method versus a beam scaling approximation method. In particular, we wanted to assess the percentage dose differences between the two methods for several water thicknesses representative for different patients of different body mass index. The largest percentage differences were found for the thickest part of the bowtie filter and increased with patient size. © 2014 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.1.3.033507](https://doi.org/10.1117/1.JMI.1.3.033507)]

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1 Introduction

There is an increasing concern about radiation dose associated with computed tomography (CT) examinations due to the relatively high patient exposure for each examination and the number of examinations being performed. Therefore, there have been many approaches to estimate the dose delivered to patients during CT examinations, from using the *k*-factors for effective dose assessment,¹ to more recent approaches such as the size specific dose estimates as proposed by the American Association of Physicists in Medicine (AAPM) Task Group 204.² However, these approaches do not provide organ dosimetry, which is needed to assess the probability of cancer risk of individuals.³ For that purpose, Monte Carlo simulations tuned for patient specific dosimetry are considered as the gold standards.

In order to develop a complete Monte Carlo simulation of a CT scanner, the source and filtration should be known. Today, this is usually manufacturer proprietary data. Back in 2003, Jarry et al.⁴ described how to model a CT source where the radiation transport in the bowtie filter does not need to be fully simulated; instead, they proposed to give weighting factors to the photons according to the path length that they have traveled inside the bowtie filter. Later on, Turner et al.⁵ described how to practically implement this idea of a source model for a multi-detector CT for Monte Carlo simulations. Through a series of CT measurements, angle- and energy-specific weighting factors

were obtained and applied to the so-called central beam. This method will be shortly described later and is considered as the gold standard approach.

Prior to implementation of the same approach in a new Monte Carlo platform for patient specific CT dosimetry, we aimed at better understanding the effect of the bowtie filter on dose accuracy estimation when using the gold standard bowtie filter method as described by Jarry et al.⁴ and Turner et al.,⁵ versus the use of a simpler method that accounts only partially for the bowtie filter effects. This simpler method uses fan angle-specific weighting factors instead of factors for each energy bin of the spectrum and each fan angle. To estimate the differences between the two methods, dose in air was calculated after attenuation of the spectra with water thicknesses equivalent to different patient sizes. In addition, we calculated the differences in organ doses estimated by running the Monte Carlo simulation for a chest CT of the female International commission on radiological protection (ICRP) phantom.

2 Method

2.1 Characterization of the Source and Monte Carlo Implementation

Following the method proposed by Turner et al.,⁵ characterization of the source of a Siemens Definition Flash CT scanner (Erlangen, Germany) was performed. Turner's methodology is divided into two parts: (1) finding an equivalent energy

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spectrum (EES) in the center of the beam and (2) finding an equivalent filter description for the different fan angles.

Initially, the first and second half value layers in the center of the beam (HVL1 and HVL2) were measured with a DCT10 pencil beam ionization chamber and Barracuda electrometer (RTI Electronics, Anaheim, CA) and foils of aluminum (Al) of 1 and 2-mm thicknesses (Goodfellow, 99% purity). The measurements were performed in the service mode at 120 kVp so that the tube was located at a fixed position. We characterized the large bowtie filter, which is typically used for body and dual energy scans.

Subsequently, to find the equivalent filtration of the bowtie filter at each angle, the dose profile across the fan beam was measured with the same ionization chamber positioned at increasing distances from the isocenter.⁵ The measurements were performed with increments of 1 cm, which is equivalent to 0.96 deg.

The EES was then obtained with the SPEKTR⁶ tools in MATLAB® based on the measurements of HVL1, HVL2, and the peak tube voltage (part 1). Next, the EES was used to determine the thickness of the bowtie as a function of fan angle (part 2) as follows: we determined the thicknesses of Al that would attenuate the EES such that the associated exposure would match the measured one for the same angle.

In this way, knowing the position, direction, and the energy of each photon in a Monte Carlo framework, it is possible to retrieve the proper weighting factor.

Before implementing these weighting factors in our Monte Carlo framework, we wanted to evaluate the effects on dose accuracy estimation when using the gold standard bowtie filter method versus a partial method. With the term “partial” we mean that weighting factors would be dependent on the angle but not on the energy. For this purpose, we have investigated a new set of weighting factors that would be applied to the EES. By means of the SPEKTR tools, we calculated the equivalent thickness of teflon and Al that would lead to the same exposure as measured. Then, we calculated for each angle a single weighting factor identical to an energy averaged attenuation factor induced by the obtained amount of teflon or Al [see Eq. (1)]. As the thickness is correlated to a specific fan angle, these weighting factors were called “global angle specific weighing factors,” which is, in effect, just scaling the beam intensity.

The global angle-specific weighting factors were calculated for each of the equivalent thicknesses of the bowtie filter of Al and teflon as shown in Eq. (1):

$$\frac{1}{e^{-\mu x}} = \frac{\int \psi(E) e^{-\mu(E)x} dE}{\int \psi(E) dE}, \quad (1)$$

where $\psi(E)$ is the energy fluence, μ is the linear attenuation coefficient, and x is the thickness of the penetrated material. For each thickness, this expression represents the energy averaged air kerma fraction of the beam.

Advantages of the simpler method are straightforward: instead of having a large lookup table to account for a complete simulation of the bowtie filter as a function of the energy and the thickness, a simple two-column table with the angle and the corresponding global angle-specific weighting factor was produced to account for a partial beam hardening. We chose the global angle-specific weighting factors calculated with Al thicknesses.

2.2 Dose in Air (Incident Dose)

As the EES attenuated by a global angle-specific weighting factors [Eq. (1)] is not the same as the EES being hardened by angle- and energy-specific weighting factors (gold standard method), a calculation of dose in air without any attenuation in the beam (incident dose) was first performed to compare the two methods. The dose in air was calculated for each fan angle (or bowtie thickness). To accomplish this, an in-house tool based on Boone's paper⁷ was previously used to implement the weighting factors in the Monte Carlo framework. The calculation of the dose was performed for one projection.

The percentage difference between the doses in air calculated with the two methods was assessed, using the gold standard method as a reference.

2.3 Dose in Air after Attenuation with Different Thicknesses

To quantify the difference in dose that this approach will have when used for patients, we calculated the dose in air for the two methods as described in Sec. 2.2; specifically, we estimated the dose in air after passing through different thicknesses of water, all equivalent to specific patient sizes.

From a database of patients who were scanned with a thoracic-abdominal CT protocol, a female and a male for each BMI (body mass index) category (underweight, normal, overweight, and obese) were chosen. For these patients, the water equivalent thicknesses of the lateral (LAT) and anterior-posterior (AP) diameters were estimated for each line of the topogram using a Siemens research tool (proprietary information). As an example, Fig. 1 shows the water thickness of the LAT and AP diameters at each position of the topogram for two patients of different BMI: underweight and obese. The maximum values for the LAT and AP diameters were chosen to explore the differences in dose. Although we are aware that the bowtie filter is designed to be thickest where the path through the patient is shorter,⁸ we decided to study the extreme cases, by performing our calculations with the maximum water equivalent thicknesses. The calculation of the dose was separately performed for the AP and LAT diameters.

We have then moved from this patient model expressed in water equivalent thickness for LAT and AP diameters to a water thickness equivalent to the effective diameter (ED) as described in the AAPM 204.² To have a more realistic estimate of the effect of the bowtie modeling on patient dose, we calculated the dose for half of the ED for all bowtie thicknesses (or fan angles). We hypothesized that this would be at the typical depth of critical organs. Moreover, to consider all the possible drawbacks from the simple beam intensity scaling approach, the water equivalent diameters were considered at isocenter and the maximum fan angle corresponding to each diameter was calculated to estimate the maximal difference in dose for the patients. As we chose one patient from each BMI group, the differences in doses were calculated for different extreme fan angles, corresponding to the patient sizes. The percentage difference in dose in air for the two bowtie modeling approaches and for the different water thicknesses was calculated.

2.4 Monte Carlo Framework

A Monte Carlo simulation framework for patient specific dosimetry in CT is being modified from a cone beam CT simulation

Fig. 1 Water thickness equivalence of the lateral (LAT) (black) and anterior-posterior (AP) (white) diameters at each position of the topogram for two patients of different BMI: an (a) underweight patient and (b) an obese patient.

framework developed in our group.^{9,10} The EGSnrc Monte Carlo code system (version 4-2.4.0) was used to simulate a CT scan from a phantom. However, instead of directly simulating the bowtie filter, we used weighting factors derived from measurements to describe the effect of bowtie filtration. The simulation starts with the input of a spectrum, which was experimentally obtained with the SPEKTR tool (Sec. 2.1). The weighting factors are then applied to the spectrum, implemented by a lookup table with a bilinear sampling to have energy and spatial distribution. The spatial distribution of simulated x-rays is confined within the collimation.

The CT projections are simulated in a step and shoot pattern with the source moving around a stationary three-dimensional 3-D voxel model in discrete angles. To reproduce a helical scan, the photons are spatially transformed twice, first around the phantom to simulate the rotation movement of the x-ray tube, and second in the longitudinal direction of the phantom in order to simulate the table translation. The pitch is implicitly accounted for within the scan length, the total number of rotations, and the distance between the focal spot and the isocenter.

In simulating the photon-electron transport within the phantom, the Bound Compton scatter, Rayleigh scatter, electron impact ionization, and atomic relaxations were included. For the photon cross sections, photon cross section database was used, while the National Institute of Standards and Technology data were used for the Bremsstrahlung cross sections. The simulation was speeded up by cutting off the photon and electron transport at energy thresholds of 10 and 520 keV, respectively.

The energy deposition resulting from each photon interaction was registered on the fly using tags associated to a group of voxels and the kerma approximation was made to the secondary electrons. The dose statistical uncertainty was estimated during the simulation using the so-called “history-by-history” approach as proposed by Walters et al.¹¹ More detailed information of the simulation framework can be found in Ref. 9.

2.4.1 Effects on dose accuracy between the two methods in the Monte Carlo framework

The Monte Carlo calibration factor was acquired via the simulation of an ionization chamber located free in air at the isocenter

and a subsequent simulation of one rotation of the x-ray tube. The calibration factor was calculated as the dose per particle history divided by the measured exposure per mAs. Two calibration factors were calculated, one for the gold standard method and another for the “partial” method. Practically, the only difference in the main code was the implementation of the bowtie filter lookup table containing the angle- and energy-based weighting factors versus the global angle-specific weighting factors.

To verify the bowtie weighting factors, we modeled the two standard polymethyl methacrylate (PMMA) phantoms in simulation and compared the calculated dose estimates with the measured doses. The diameters of the phantoms are 16 and 32 cm, respectively, and the doses at the central and peripheral regions were considered. In the simulation, only the bowtie implementation was different, one with the gold standard method and other with the beam scaling method (partial method). All other parameters were kept exactly the same.

To investigate the effects on accuracy in organ dose between different bowtie implementations, we modeled a chest CT scan with the ICRP female phantom.¹² The ICRP female phantom has a voxel resolution of $1.775 \times 1.775 \times 4.84 \text{ mm}^3$ (sagittal $\times$ coronal $\times$ vertical) and we chose to simulate the thorax scanning protocol that is commonly used in clinical practice. Specifically, we simulated a helical chest scan made of 6.5 rotations and a table translation of 30 cm. We compared the difference in dose for the lungs and the breasts in the complete scan. The central process unit (CPU) time in both cases was registered as additional information.

3 Results and Discussion

3.1 Generation of the Spectra and Filtration for the Computed Tomography Scanner, using Both Methods

A HVL1 equal to 9.0 mm Al and HVL2 equal to 20 mm Al were estimated for the Siemens Definition Flash at 120 kVp. These results were used to find the suitable EES. A visual comparison between the spectra obtained with angle- and energy-specific factors (full beam hardening) and the simpler beam

Fig. 2 Comparison of the spectra for different fan angles: 0 deg, 3 deg, 6 deg, 9 deg, 12 deg, 16 deg, 19 deg, 23 deg, and 25 deg. In black, the beam hardened spectrum denoted by deg, and in gray the spectra being scaled by the global angle-specific weighting factors denoted by deg_w.

scaling factor is shown in Fig. 2 for nine different thicknesses of the bowtie filter. It can be seen that when the EES is scaled using the global angle-specific weighting factors, the number of photons of low energy will be larger while high energy photons will be suppressed in comparison with the hardened spectrum, as expected. The difference increases with the thickness of the bowtie filter and is maximum at the thickest part.

An average difference of 1.9%, and a maximum difference at the thickest part of the bowtie filter of 3.1% was found between the weighting factors calculated for teflon and Al (Fig. 3). Discrepancies between the calculated energy averaged air kerma fractions for teflon and Al versus the measured differences in exposure were at maximum 8.8%. This approach was considered acceptable for the comparative study between this simple beam scaling and the full simulation of the beam hardening.

Fig. 3 (a) Aluminum (Al) and teflon thickness of the bowtie filter as a function of the fan angle and (b) corresponding global angle-specific weighting factor as a function of the fan angle for Al and teflon. The measurements were performed with increments of 1 cm, which is equivalent to 0.96 deg.

Fig. 4 Percentage difference in dose in air as a function of the fan angle.

3.2 Dose in Air (Incident Dose)

The maximum difference in incident dose between the two methods was -11.6% at the thickest part of the bowtie filter (Fig. 4). An overestimation of the dose is obtained when the global angle-specific weighting factors for the EES are used.

This was expected, as the lower energy photons were preserved along the different thicknesses in the EES while the high energy photons were suppressed (see Fig. 3). The contribution of low energy photons to the incident dose is prominent.

The difference in dose increased with the thicknesses, but as the bowtie filter is designed to cover a field of view wide enough to irradiate all the detectors, the x-rays crossing the last part of the bowtie will usually not penetrate normal sized patients. An angle of 25 deg would correspond to an obese patient with a lateral diameter of 56 cm at isocenter.

3.3 Dose in Air after Attenuation of Different Patient Thicknesses

The differences in dose in air due to the two methods for different water thicknesses, as associated with the maximum AP and LAT diameters, are separately presented in Fig. 5 for female and male. The largest differences in dose as a function of the fan angle were found for the thickest part of the bowtie filter and they increase with patient size. The maximum difference was for the LAT diameter of the obese patients. The differences in dose ranged from 6.5% to 16.1% for a normal patient and from 7.7% to 19.3% for an obese patient.

As the dose was calculated after passing through the maximum LAT and AP water thicknesses, there will be an overestimation

Fig. 5 Percentage difference in dose as calculated by the proposed method and by the gold standard method for the maximum lateral and anterior-posterior diameters of selected patients as a function of the fan angle.

Fig. 6 Percentage difference in dose as calculated by the proposed method and by the gold standard method for the maximum effective diameter (ED) and half of the ED of selected patients as a function of the fan angle.

of the difference in dose that would be obtained in a real patient, as we considered blocks of water instead of a shape closer to the patient: the patient is usually thinner at the thickest part of the bowtie filter, as it as an elliptical shape rather than a rectangular shape. This assumption is even one of the basic premises of bowtie filter design. The present dose was calculated after passing through the maximum diameters, and it can be interpreted as the dose at the exit of the patient.

To have a better estimation of the dose, the difference in dose was also calculated for the effective diameter and for half of it. We can see from Fig. 6 that the differences are smaller than for the LAT shown in Fig. 5. For a normal patient, the differences ranged from 3.2% to 10.4%, while for an obese patient they ranged from 4.1% to 14.5%. Also in this case, we simplified the analysis by considering a rectangular block with all doses as a function of the fan angle calculated for the same thickness. The percentage difference in dose measured after a water equivalent thickness correlating with an obese patient is considerable (14.5%). Results for the effective diameter may represent the dose differences in superficial organs like the breast at the beam exit side, while our results at half of the effective diameter can be thought of as a metric relevant for inner organs.

An interesting remark is that the method of partial beam hardening underestimated the dose in air when different

water equivalent thicknesses were used to simulate the different patient sizes, while there was an overestimation for the incident dose (in air without any attenuating material in the beam). The global angle-specific weighting factors lead to an underestimation depending on the water thickness the beam has to pass through. To investigate this effect in depth, an extra calculation was done for different thicknesses of water, starting from 2 to 40 cm as is shown in Fig. 7. In this case, the difference was calculated for all the fan angles.

As can be seen in Fig. 7, for the thinnest block of water (2 cm), there is an overestimation of 8.5% and when 10 cm is reached, the effect is reversed. Another interesting fact is that the dose gradient as a function of thickness and for a given angle reduces with increasing thickness. Indeed, the same is true for the dose gradient as a function of the fan angle for a given water equivalent thickness.

3.4 Monte Carlo Framework

For the dose inside the PMMA phantom with the gold standard method, we found an agreement between measurements and simulations results within 6%. However, for the partial method, we found an agreement from -9.6% to 14.5%.

Fig. 7 Percentage difference in dose as calculated by the proposed method and by the gold standard method for different water thicknesses from 2 to 38 cm.

For the ICRP female phantom, we found a difference of 4.0% for the dose to the lungs and 1.7% for the dose to the breasts, which were underestimated by the partial modeling. With the partial method, a reduction of 12% and 6% in the CPU time was found for the simple 16-cm-diameter PMMA phantom and the more complex ICRP female phantom, respectively.

3.5 Limitations of the Study

The calculations presented for the different patient sizes were not performed with the Monte Carlo simulation framework, therefore, the scattered radiation was not taken into account. For the effects studied in this work, preliminary Monte Carlo simulations had shown that this approach was acceptable.

The required accuracy for dose determination according to the AAPM depends on the dose level and potential risk. An accuracy requirement of 7%, at an expanded uncertainty corresponding to the confidence level of 95%, is recommended for comparative risk assessments as well as for quality assurance.¹³ In this work, we found that for a complete simulation of the ICRP female phantom, the percentage difference in dose was less than 4%, which is acceptable, but the differences may increase for bigger phantoms. We concluded that the full effect of beam hardening needs to be considered, especially if it is meant to be used for patient-specific dosimetry or protocol optimization.

4 Conclusion

Incomplete modeling of the bowtie filter by implementing only a beam intensity scaling factor provoked both underestimations and overestimations of the absorbed doses as a function of the fan angle for a single projection for different water thicknesses. The difference in accuracy for organ dosimetry was smaller (4%) when a complete simulation of a chest CT scan was performed for the ICRP female phantom. However, the gold standard method of modeling the bowtie filter with approaches such as Turner's should be used to have a better accuracy for patient-specific dosimetry calculations.

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