Estimation of Radiation Dose in CT Based on Projection Data

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Published online: 18 February 2016
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Abstract Managing and optimizing radiation dose has become a core problem for the CT community. As a fundamental step for dose optimization, accurate and computationally efficient dose estimates are crucial. The purpose of this study was to devise a computationally efficient projection-based dose metric. The absorbed energy and object mass were individually modeled using the projection data. The absorbed energy was estimated using the difference between intensity of the primary photon and the exit photon. The mass was estimated using the volume under the attenuation profile. The feasibility of the approach was evaluated across phantoms with a broad size range, various kVp settings, and two bowtie filters, using a simulation tool, the Computer Assisted Tomography SIMulator (CATSIM) software. The accuracy of projection-based dose estimation was validated against Monte Carlo (MC) simulations. The relationship between projection-based dose metric and MC dose estimate was evaluated using regression models. The projection-based dose metric showed a strong correlation with Monte Carlo dose estimates ($R^2 > 0.94$). The prediction errors for the projection-based dose metric were all below 15 %. This study demonstrated

the feasibility of computationally efficient dose estimation requiring only the projection data.

Keywords CT · Computed tomography · Patient-specific dosimetry

Introduction

In the past decade, the superior imaging attributes of computed tomography (CT) have significantly increased its clinical utilization [1, 2]. In 2011, the number of CT exams performed in the USA reached 80 million [3]. In parallel to such increase come concerns about the potential health risk associated with CT radiation exposure [4–6]. To maximize patient benefit and minimize the potential risk, it is well agreed within the CT community that the radiation dose should be closely managed and optimized [7]. As a fundamental step to achieve that, accurate and computationally efficient dose estimation is crucial [8]. Such dose estimation could provide quantitative information about the dose reduction potential of different strategies, aid in the evaluation of CT system design, ensure improved scanner performance, and build towards patient-specific dosimetry.

Currently, two types of indices are available for CT dose quantification. The first type is quality assurance dose indices (e.g., the volume CT dose index (CTDI_{vol}) and the dose-length product (DLP)). These metrics effectively characterize scanner radiation output and are widely used to perform acceptance and constancy testing [9]. However, since these indices quantify dose only in uniform phantoms, they cannot accurately reflect patient-specific radiation dose level. In attempting to quantify patient-specific dose, several studies have developed software programs that simulate the organ dose for standard anthropomorphic phantoms and scanners

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(ImPact [10], CT-Expo [11]). However, these approaches generally rely on pre-calculated organ dose and are limited to a small number of phantoms [12]. Recently, there has been a growing effort to combine Monte Carlo simulation and computational phantoms to simulate organ dose across a relatively large patient population [13–15]. Although very accurate, the Monte Carlo method is usually computationally expensive since it requires calculations of a large number of photons propagating in the object. Even if implemented based on graphics cards (GPU), it generally requires minutes for a single simulation.

In this study, we advanced a possibility of using a projection-based dose metric to achieve accurate and computationally efficient estimation of CT radiation dose. As an initiative study to validate the feasibility of this approach, we devise the projection-based dose metric using a simulation program that generates realistic projection data. The accuracy of the projection-based dose metric was further comprehensively validated against the radiation dose estimated using Monte Carlo simulation across a broad range of patient sizes, anatomy, and scanning parameters.

Materials and Methods

Approach

Absorbed dose is a measure of the energy deposited in the object by ionizing radiation per unit mass. To estimate absorbed dose, the imparted energy and object mass were explicitly modeled using projection data. The modeling process is detailed as follows.

Absorbed Energy

In this study, the intensity of the absorbed energy was estimated using the difference between the intensity of scanner output signal and energy-integrated detected signal. The intensity of scanner output signal was estimated using the calibration air scan, which is an acquisition without a patient in the scan field of view and usually performed beforehand for different kVp settings and bowtie filters. The intensity of detected signal was determined from the patient scan as schematically demonstrated in Fig. 1.

Using such approach, the primary radiation was accurately quantified. The scattered radiation was categorized into three types: (1) absorbed scatter radiation (radiation that is absorbed in the body after one or multiple scattering events and thus contributes to patient dose), (2) detected scatter radiation (scattered radiation that reaches the detector due to geometrical reasons), and (3) undetected scatter radiation (scattered radiation that does not reach the detector). The projection-based dose metric effectively accounts for the first two types

of scattered radiation. However, it accounts the third type of scattered radiation as absorbed radiation. Thus, the projection-based dose metric tends to overestimate the absorbed dose. This systematic error was incorporated with an empirical constant.

Mass

The mass is determined by object density and volume. During the reconstruction process, the detected x-ray intensity is log-converted and back-projected to obtain the attenuation coefficient μ . Since dense materials generally yield high attenuation coefficients and light materials generally yield low attenuation coefficients, the attenuation coefficient μ can be used as a substitute for object density. Further, the volume of the scanned object can be estimated using the area under the attenuation profile and the object thickness (z dimension). Thus, mass can be estimated by multiplying the area under the attenuation profile with the object thickness (z dimension).

Dose

Using the projection-based substitutes of absorbed energy and mass, the average dose distributed within the cross section of the scanned object can be estimated in every projection as

$$\hat{D}_{\text{View}=j} = \hat{E}_{\text{View}=j} / \text{mass} \quad (1)$$

The projection-based dose metric can be further defined as the average of $\hat{D}$ across projections as follows:

$$D_{\text{Proj}} = \sum_{\text{views}} \frac{\hat{E}_{\text{View}=j}}{\text{mass}} / \text{Number of projections} \quad (2)$$

Projection-based dose metric estimated the average dose distributed within one cross section of the phantom. To validate the theory and quantify accuracy, projection-based dose metric was benchmarked against the dose results from Monte Carlo simulation. The dose value that has the same physical meaning (average dose for one cross section) was deduced using both methods. Average dose distributed within one cross section was estimated from Monte Carlo simulation as

$$D_{\text{MC}} = \frac{\sum_{\text{Cross section}} \text{Absorbed Energy}}{\text{Mass}_{\text{Cross section}}}. \quad (3)$$

Phantom Simulation Study

A simulation package was used to assess the accuracy of the technique. We used the Computer Assisted Tomography SIMulator (CATSIM) for the simulation of the air scan and

Fig. 1 **a** Different types of scattered radiation in this study: (1) absorbed scatter radiation (radiation that is absorbed in the body after one or multiple scattering events and thus contributes to patient dose), (2) detected scatter radiation (scattered radiation that reaches the detector due to geometrical reasons), and (3) undetected scatter radiation (scattered radiation that does not reach the detector). **b** The intensity of

the absorbed energy was estimated using the difference between the intensity of scanner output signal and energy-integrated detected signal as $I_{\text{absorbed}} = I_{\text{air}} - I_{\text{patient}}$. I_{air} refers to the intensity of scanner output signal. It was estimated using the calibration air scan, which is an acquisition without a patient in the scan field of view. I_{patient} refers to the intensity of the detected signal determined from the patient scan

the patient scan [16]. The CATSIM simulation tool models the x-ray spectrum, the geometry of the tube rotation, the material and geometry of the bowtie filter, the x-ray transmission and detection process. The quality of CATSIM simulations has been validated via a series of phantom measurements in terms of the x-ray spectrum, noise property (including electronic noise), and scattering profile on a commercial CT scanner (LightSpeed VCT, GE Healthcare, Waukesha, WI) [16]. The Monte Carlo simulation was performed using simulation software (CATDOSE, GE Healthcare, Waukesha, WI) [16]. A series of mathematical phantom simulation were performed to ascertain the feasibility of the projection-based dose metric across a broad range of patient sizes, patient anatomy, and scanning parameters. As a fundamental validation of the effectiveness of projection-based dose metric, simulations were performed on phantoms with simple structures. Eight cylindrical water phantoms (size range from 12 to 40 cm) and seven ellipsoidal water phantoms (size range from 10 cm × 6.5 cm to 40 cm × 26 cm) were created. In addition, helical body phantoms (HBP) were created to emulate the anatomy in the abdomen region. The phantoms contained four materials: water, bone, spine, and air, cast into four sizes ranging from 10 cm × 6.5 cm to 40 cm × 26 cm (Fig. 2). For each phantom, average dose distributed within the cross sectional area at the central slice of the phantom was estimated. The scan was simulated using a GE Lightspeed scanner at 120 kVp, 40 mm collimation, 1 pitch, and the large bowtie filter. A total of 1000 projections were generated per 360 degrees. The projection-based dose metric was benchmarked against the estimated dose from Monte Carlo simulation using linear regression.

For a more anthropomorphic validation, the projection-based dose metric was tested on a 4D extended cardiac-torso

(XCAT) phantom [17]. The scan was performed using a clinical cardiac protocol (120 kVp, pitch of 1, large bowtie filter, GE LightSpeed VCT). One view of the projection data generated by CATSIM was illustrated in Fig. 3a. The central slice of the dose map simulated by CATDOSE was shown in Fig. 3b. Similarly, the projection-based dose estimate was benchmarked against Monte Carlo simulation using linear regression.

The projection-based dose metrology was applied to a range of clinical scanning parameters, including four kVp settings (80, 100, 120, and 140 kVp) and two types of bowtie filters (small bowtie and large bowtie filter, GE LightSpeed VCT). Four HBP phantoms and four water phantoms were simulated in each setting. The similar linear regression analysis was performed on the simulation results.

Results

Figure 4 illustrates the relationship between projection-based dose estimate (D_{proj}) and Monte Carlo dose estimate (D_{MC})

Fig. 2 Helical body phantom used in this study. The phantoms contained four materials: water, bone, spine, and air, cast into four sizes ranging from 10 cm × 6.5 cm to 40 cm × 26 cm

Fig. 3 Simulation results of the XCAT phantom. (a) One view of the projection data generated by CATSIM. (b) Dose map generated by CATDOSE

for cylindrical water phantoms, elliptical water phantoms, and helical body phantoms, respectively. Strong linear relationship was found for all three types of phantoms with R^2 greater than 0.95 ($p < 0.001$). The average differences between D_{proj} and D_{MC} were 9.1, 9.3, and 9.8 % for cylindrical water phantoms, elliptical water phantoms, and helical body phantoms, respectively.

Figure 5 shows the corresponding relationship between projection-based dose estimate (D_{proj}) and Monte Carlo dose estimate (D_{MC}) for the XCAT phantom. The water phantom

results were used as the basis for regression model. The difference between D_{proj} and D_{MC} was 8.9 % for the XCAT phantom.

Figure 6a illustrates the linear dependencies between projection-based dose estimate (D_{proj}) and Monte Carlo dose estimate (D_{MC}) for different kVp settings for four HBP phantoms (size range from 10 cm × 6.5 cm to 40 cm × 26 cm). The average percentage difference between D_{proj} and D_{MC} was 15.2 % across four kVp settings with R^2 equal to 0.94. The differences were large for 80 and 140 kVp, mainly due to the

Fig. 4 Linear regression models of projection-based dose metric and Monte Carlo dose estimate for cylindrical water phantoms (a), elliptical water phantoms (b), and HBP phantoms (c) ($p < 0.001$)

Fig. 5 Linear regression model of projection-based dose metric and Monte Carlo dose estimate for XCAT phantoms ($p < 0.001$)

effect of undetected scattered radiation as detailed in the discussion.

Figure 6b illustrates the linear relationship between projection-based dose estimate (D_{proj}) and Monte Carlo dose estimate (D_{MC}) for two types of bowtie filters. The average percentage difference between D_{proj} and D_{MC} was 10.3 % for large bowtie filter and 8.9 % for small bowtie filter at 120 kV.

Discussion

Quantifying CT radiation dose is essential for the evaluation and optimization of CT systems, for the implementation of dose-saving strategies, and for improved scanner performance. Current Monte Carlo based methods fall short of meeting the practical requirement of estimating radiation dose in a

computationally efficient manner. In this study, we developed a projection-based dose estimation method and further validated the accuracy against Monte Carlo simulation across a broad range of sizes, anatomy and scanning parameters. Such method enables one to quantify the cost-effectiveness of different dose reduction techniques quickly. If incorporated as a scanner built-in function, it may further provide a more accurate estimation of the dose-saving potential for different strategies. Such information may aid in the design and modification of CT design.

As an example of how the projection-based dose metric can be used to quantify dose reduction, consider the selection of protocols for an abdominopelvic CT scan. Two protocols were considered: a standard protocol (protocol A, 120 kVp, 250 mAs) and a modified protocol (80 kVp, 540 mAs). The dose metric was applied for two hypothetical patients with sizes 16 and 24 cm. Two helical body phantoms were modeled in the CATSIM software to estimate projection-based dose metric. The estimated doses for protocol A were 26 and 15 mGy, respectively. The doses for protocol B were 20 and 11 mGy, respectively. A similar approach can be extended to quantify the dose reduction potential of the bowtie filter designs and mA modulation techniques. However, it should be noted that the use of dose metrics for quantifying dose reduction potential is not limited to projection-based dose estimates. For example, size-specific dose estimates (SSDE) can be used in a similar way. In fact, a preliminary comparison between projection-based dose metric and SSDE shows that the two metrics offer comparable accuracy in dose estimation under fixed tube current scans. However, most notably, projection-based dose estimation technique can offer advantages in dose estimation under tube current modulation. Since projection-based dose metric estimates the absorbed dose on each

Fig. 6 Linear regression models of projection-based dose metric and Monte Carlo dose estimate across four kVp settings (a) and two types of bowtie filters (b) ($p < 0.001$). The simulation was performed using four

water phantoms and four HBP phantoms with size range from 10 to 40 cm under each scanning setting

projection plane, the technology can theoretically accounts for both the axial and longitudinal tube current modulation. Other dose estimates such as $CTDI_{vol}$ and SSDE may not provide such advantages.

It should be noted that the correlation coefficient between D_{proj} and D_{MC} is slightly larger than 1, which indicates that the projection-based dose metric generally produces higher dose estimation than Monte Carlo method. Such discrepancy is due to the fact the projection-based dose metric overestimates the absorbed energy, mainly the undetected scatter radiation. The slope of the regression indicated the percentage of the undetected scatter radiation of the total absorbed dose. On average, the undetected scatter radiation was about 13 % to the total absorbed dose ($y = 1.13x$). This constant can be regarded as an average number across all kVp settings, phantom sizes, and phantom types. However, the amount of undetected scatter radiation varied across phantoms and scanning parameters. For large phantoms, the undetected scatter radiation is smaller compared with small phantoms. Thus, the slope for low mGy/mAs (large phantoms) is lower than that for high mGy/mAs (small phantoms). However, it should be noted that the maximum error for projection-based dose metrics is less than 15 % across phantoms with a broad size range. This essentially indicated that the projection-based dose metric provided reasonably accurate dose estimation for clinical patients under common protocols.

Our study has several limitations; first, the projection-based dose metric estimates the average dose distributed within the cross section of the patient. However, it does not provide detailed organ dose estimates. Secondly, as discussed above, projection-based dose metric overestimates the absorbed dose. A further study is needed to better incorporate the undetected scattering effect. Thirdly, the implementation of such method in clinical system requires one to extract the projection data from CT scanners, which might be difficult due to the proprietary concerns. Besides, in actual clinical settings, the daily calibration air scans varied significantly mainly due to the random and thermal variation of the detector gains. Future studies should model such impact and explore strategies to implement the metric in actual clinical settings. Fourthly, in this paper, we mainly focused on axial scans and helical scans with pitch of 1. The effect of pitch on projection-based dose metric will be included in our future study. Lastly, other factors, such as beam hardening effect [18, 19] and patient miscentering [20] may also have an impact on patient dose. These will be the topics for future studies. Finally, the study was performed using simulation software; further study will be needed to validate the technique against physical measurements. However, taking into considerations noted above, the projection-based dose metrology provides reasonably accurate results compared with Monte Carlo dose estimates. Consistent linear relationship was found for phantoms with different sizes and materials, kVp settings, bowtie filters, and

in a realistic phantom. This study demonstrated the feasibility of computationally efficient dose estimation requiring only the projection data.

References

1. McCollough CH, Guimaraes L, Fletcher JG: In defense of body CT. *AJR Am J Roentgenol* 193:28–39, 2009
2. Berdahl CT, Vermeulen MJ, Larson DB, Schull MJ: Emergency department computed tomography utilization in the United States and Canada. *Ann Emerg Med* 2013
3. IMV: CT Market Outlook Report, 2012. IMV Medical Information Division, Des Plaines, 2012
4. Brenner DJ, Hall EJ: Computed tomography—an increasing source of radiation exposure. *N Engl J Med* 357:2277–2284, 2007
5. Pearce MS, Salotti JA, Little MP, McHugh K, Lee C, Kim KP, et al: Radiation exposure from CT scans in childhood and subsequent risk of leukaemia and brain tumours: a retrospective cohort study. *Lancet* 380:499–505, 2012
6. FDA: Draft guidance for industry and food and drug administration staff: pediatric information for x-ray imaging device premarket notifications, in guidance documents (medical devices and radiation-emitting products) 2012. U.S. Department of Health and Human Services: Rockville
7. McCollough CH, Chen GH, Kalender W, Leng S, Samei E, Taguchi K, et al: Achieving routine submillisievert CT scanning: report from the summit on management of radiation dose in CT. *Radiology* 264:567–580, 2012
8. Boone JM, Hendee WR, McNitt-Gray MF, Seltzer SE: Radiation exposure from CT scans: how to close our knowledge gaps, monitor and safeguard exposure—proceedings and recommendations of the Radiation Dose Summit, sponsored by NIBIB, February 24–25, 2011. *Radiology* 265:544–554, 2012
9. McCollough CH, Primak AN, Braun N, Kofler J, Yu L, Christner J: Strategies for reducing radiation dose in CT. *Radiol Clin N Am* 47: 27–40, 2009
10. ImPACT's CT Dosimetry Tool: CT dosimetry version (available at URL: <http://www.impactscan.org/ctdosimetry.htm>)
11. CT-expo software programs for CT dose calculation (available URL: <http://www.mh-hannover.de/1604.html>)
12. Kalender WA: Dose in x-ray computed tomography. *Phys Med Biol* 59:R129–R150, 2014
13. Tian X, Li X, Segars WP, Paulson EK, Frush DP, Samei E: Pediatric chest and abdominopelvic CT: organ dose estimation based on 42 patient models. *Radiology* 270:535–547, 2014
14. Li X, Samei E, Segars WP, Sturgeon GM, Colsher JG, Frush DP: Patient-specific radiation dose and cancer risk for pediatric chest CT. *Radiology* 259:862–874, 2011
15. Geyer AM, O'Reilly S, Lee C, Long DJ, Bolch WE: The UF/NCI family of hybrid computational phantoms representing the current US population of male and female children, adolescents, and adults-application to CT dosimetry. *Phys Med Biol* 59:5225–5242, 2014
16. De Man B, Basu S, Chandra N, Dunham B, Edic P, Iatrou M, McOlash S, Sainath P, Shaughnessy C, Tower B, Williams E: "CatSim: a new computer assisted tomography simulation environment". *Proc. SPIE 6510, Medical Imaging 2007: Physics of Medical Imaging*

17. Segars WP, Mahesh M, Beck TJ, Frey EC, Tsui BM: Realistic CT simulation using the 4D XCAT phantom. *Med Phys* 35:3800–3808, 2008
18. Ay MR, Mehranian A, Maleki A, Ghadiri H, Ghafarian P, Zaidi H: Experimental assessment of the influence of beam hardening filters on image quality and patient dose in volumetric 64-slice X-ray CT scanners. *Phys Med* 29:249–260, 2013
19. Habibzadeh MA, Ay MR, Asl AR, Ghadiri H, Zaidi H: Impact of miscentering on patient dose and image noise in x-ray CT imaging: phantom and clinical studies. *Phys Med* 28:191–199, 2012
20. Tsalafoutas IA, Georgolopoulou P, Abatzoglou I: CT dosimetry considerations in non typical conditions: the effect of scan field of view and table height selection. *Phys Med* 28:83–90, 2012