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Demonstration of Dose and Scatter Reductions for Interior Computed Tomography

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Abstract

With continuing developments in computed tomography (CT) technology and its increasing use of CT imaging, the ionizing radiation dose from CT is becoming a major public concern particularly for high-dose applications such as cardiac imaging. We recently proposed a novel interior tomography approach for x-ray dose reduction that is very different from all the previously proposed methods. Our method only uses the projection data for the rays passing through the desired region of interest. This method not only reduces x-ray dose but scatter as well. In this paper, we quantify the reduction in the amount of x-ray dose and scattered radiation that could be achieved using this method. Results indicate that interior tomography may reduce the x-ray dose by 18% to 58% and scatter to the detectors by 19% to 59% as the FOV is reduced from 50 to 8.6 cm.

Keywords

computed tomography (CT); cardiac CT; interior tomography; Farmer chamber; pencil chamber; x-ray dose reduction; x-ray scatter reduction; CTDI

Tremendous progress in computed tomography (CT) technology such as the introduction of multi-slice CT and subsecond rotation times have increased the clinical uses of CT. The number of CT scans per year in the United States has increased from 3 million in 1980 to 62

million in 2006.^{1,2} Computed tomography is among the largest source of radiation exposure in radiology. It has been estimated that although CT studies constitute only 4% of all radiological procedures, they account for 40% of the radiation dose delivered.³ Cancer risk associated with CT is now becoming a growing concern. It has been estimated that 1.5% to 2% of all cancer cases may be associated with CT.¹ Cancer risk is especially a concern for high-dose applications such as cardiac CT.⁴ Computed tomography technology is rapidly evolving, and multi-detector CT scanners with up to 320 slices have now become available. Cardiac CT imaging is also progressing considerably, with technologies such as controlled cardiac CT^{5,6} and dynamic volumetric cardiac CT^{7,8} and scanning trajectories such as saddle curve⁹ and composite circling.¹⁰ Hence, dramatic progress in CT technology, especially with multi-detector CT, is foreseeable in the near future.¹¹ As a result, future CT imaging procedures are likely to improve the diagnosis of cardiovascular disease, including coronary artery disease, risk factor assessment, and other cardiovascular diseases such as congenital heart disease in infants and children. Therefore, it is necessary to find ways for effective dose reduction, especially for high-dose applications like cardiac CT.

Most of the proposed dose-reduction strategies have focused on modulation of scan parameters, modification of scanner geometry, and use of prospective gating.^{12,13} Although these methods can reduce dose, significant additional dose reduction can be achieved by developing acquisition and corresponding reconstruction methods that require less raw projection data. The first landmark work along this direction is the well-known fan-beam half-scan formula.¹⁴ Since then, several imaging conditions have been proposed that have enabled further reduction in the amount of required projection data.^{15–17} However, these conditions do not apply to the case where the x-ray beam is entirely limited to the region of interest (ROI) to be reconstructed. Such a condition could reduce the dose significantly for interior imaging applications such as cardiac CT. Last year, our group¹⁸ proposed an improved condition in this direction. As illustrated in Figure 1, we proved that the image on an interior line segment can be exactly reconstructed as long as it is within the scan field of view and x-ray attenuation coefficients are known for some portion of the segment. This line segment can be completely within the object, and the projection data necessary for reconstruction are fully truncated. The interior tomography method was also independently proposed by Kudo et al.^{19,20} Figure 2 compares the difference between the data acquisition configurations of conventional CT and interior tomography. The filtered back-projection (FBP) algorithm used in current CT scanners requires the object to be completely within x-ray beam in all views, but interior tomography only needs the ROI to be within the x-ray beam in all views. Owing to reduced x-ray scan field of view (SFOV), the interior tomography method not only decreases the x-ray dose but also the x-ray scatter.

Because our interior tomography approach is for interior imaging and uses localized projection data, the external markers on the human body cannot serve as known regions for reconstruction. We recently tested the feasibility of using blood intensity in aorta and/or bone intensity in vertebra as the known information for cardiac imaging applications and obtained very promising results.²¹ Figure 3 compares an FBP reconstruction image using nontruncated projection data with an interior tomography reconstruction image using only projection data for the rays passing through the ROI. Details of the reconstruction method can be found in Bharkhada et al.²¹ The known information for blood can be obtained by scanning a blood sample. Alternatively, the known information can be obtained from existing previous CT scans or from low-dose CT scans obtained immediately before the interior tomography scanning session. We believe that our interior tomography method can significantly reduce x-ray dose and scatter for interior CT imaging applications such as cardiac and dental.

Radiation dose is described as the amount of energy deposited per unit mass, say, in a patient's body, as a result of ionizing radiation exposure. The SI unit of radiation dose is the gray, and $1 \text{ Gy} = 1 \text{ J/kg}$ (energy per mass). The commonly used radiation dose parameter in CT is the CT dose index (CTDI). In practice, CTDI_{100} (in Gy or mGy) is the measured parameter, using a pencil ionization chamber to integrate the radiation exposure of a single axial scan for a length of 100 mm.²² Recently, a novel and more practical method has been proposed and validated,^{23,24} which uses a 0.6-cm^3 Farmer chamber (small volume ionization chamber) to measure CTDI_L , the accumulated dose for any scan length including CTDI_{100} . The Farmer ionization chamber is placed in a standard polymethyl methacrylate (PMMA) phantom of either 16 or 32 cm in diameter. CTDI_{100} is measured on the central and peripheral axes in the phantom. CTDI_w is the weighted average of the CTDI_{100} measurements at the center and the peripheral locations of the phantom given as

$$\text{CTDI}_w = \frac{2}{3} \text{CTDI}_{100}(\text{periphery}) + \frac{1}{3} \text{CTDI}_{100}(\text{center}).$$

CTDI_w reflects the mean absorbed dose over the central plane ($z = 0$) of the phantom at the center of a 10-cm scan length for a pitch $P = 1$. Computed tomographic dose index volume (CTDI_{vol}) is likewise the mean dose on the central x-y plane at $z = 0$ for a pitch P and is derived from CTDI_w as $\text{CTDI}_{\text{vol}} = P^{-1} \text{CTDI}_w$.²² The expected dose reduction, using interior tomography, is determined from dose measurements using the Farmer chamber.

In the diagnostic x-ray energy range, the predominant source of scattered photons and electrons is through the Compton effect. In this interaction, a photon interacts with a loosely bound or free electron within the attenuating medium. This results in release of a recoil Compton electron and deflection of the incident photon from its original path with reduced energy.²⁵ The scattered photons add to the signal along different x-ray paths and present themselves as noise in the reconstructed image. Because interior tomography requires reduced SFOV, the scattered photons that would have come from the regions outside the reduced SFOV but within the nominal SFOV would be absent, thus reducing the contributions of scattered radiation. It is well known that scatter is reduced with reduction in radiation field size. The expected reduction in scattered radiation is determined from scattered radiation measurements using the diagnostic pancake-style ionization chamber.

The main goal of this article was to quantify the x-ray dose and scatter radiation reduction achieved as a result of using our interior tomography method.

METHODS

The CTDI_{100} Acquisition Method Using a Pencil Chamber

Using a pencil chamber of active length ℓ , a single axial rotation is made about its center with the phantom and table held stationary, measuring the integral of the single rotation axial dose profile over $(-\ell/2, \ell/2)$. For the determination of CTDI_{100} , the active length is $\ell = 100 \text{ mm}$. The charge q collected is given by

$$q = \frac{1}{N_k \ell} \int_{-\ell/2}^{\ell/2} f(z) dz$$

where N_k is the chamber calibration factor in mGy/nC (as given by an accredited dosimetry calibration laboratory) and

$$\text{CTDI}_L = \frac{1}{N \times T} \int_{-L/2}^{L/2} f(z) dz$$

where $N \times T$ is the actual detector length. The chamber is vented, and pressure and temperature corrections are also applied.

CTDI₁₀₀ Using Farmer Chamber

With the small chamber method, the accumulated dose at $z = 0$ is measured by integrating the current from the ionization chamber located at a fixed point in the phantom at the midpoint ($z = 0$) of the scanned length $L = vt_0$, whereas the phantom and chamber are translated by the couch through the beam plane at velocity v during the helical acquisition of total time t_0 . Using the Farmer chamber guarantees that the scan length is always identical to the integration length L of the single rotation dose profile $f(z)$. For the determination of CTDI₁₀₀, this scan length must be 100 mm. Then, $\text{CTDI}_{100} = pD_L(0)$, where $D_L(0)$ is the dose measured by the Farmer chamber and p is an arbitrary acquisition pitch.²⁴

Experimental Validation of Farmer Chamber

We measured the CTDI₁₀₀ at the center and the periphery of the PMMA body phantom 32 cm in diameter. The phantom length is 25 cm, which is greater than the standard phantom of 14 cm length. The additional length provides better accuracy in the measurement of CTDI₁₀₀. A 10-cm-long Victoreen 500-200 pencil ionization chamber (Victoreen Instruments, Mödling, Austria) and a NE 2505/3A (0.6 cm³) Farmer chamber (Nuclear Enterprises Ltd, Reading, England) are used to make the measurement. In a GE Lightspeed 8-slice CT scanner, a single axial rotation is used for the measurements with the pencil chamber, and a helical series is used for the Farmer chamber. The measurements are made at the same kVp and mA for both the chambers. Details for the scan protocols are presented in the results.

Interior Tomography Dose Measurements

The same PMMA cylindrical body phantom and the Farmer chamber NE 2505/3A (0.6 cm³) are used for interior tomography dose measurements. Helical trajectory is used for a GE Lightspeed 16-slice scanner with a large (50 cm) SFOV. To create small SFOVs and thus simulate the interior tomography method, collimator lead plates 0.0625 inch thick are placed in front of the standard xy CT collimator to produce various openings (field sizes). This results in small SFOVs while the large SFOV is still set on the CT scanner. Thus, the same bowtie filter is used for small or large SFOV x-ray beams. The measurements for each SFOV are made at 3 different voltages: 80, 100, and 120 kVp. To measure the SFOV size, we exposed Kodak X-omat V x-ray film (Kodak) at the isocenter using a stationary gantry. The SFOV was determined from the measured width on the blackened film.

Interior Tomography Scatter Measurements

Scatter measurement is also carried out using the PMMA body phantom. The measurements are made for the same kVp, mA, and SFOVs as used for dose estimation on the GE Lightspeed 16-slice scanner. However, these measurements are made using a large-volume (15 cm³) diagnostic pancake-style ionization chamber (Fluke Biomedical 96035B). This chamber is used because it has greater sensitive volume and produces more current/charge that is suitable for low-dose measurements. In addition, the pencil chamber is too long for scatter measurements. To measure the scattered radiation reaching the detectors, the pancake chamber was placed just outside the primary beam approximately 180 degrees from the x-ray source, slightly off the detector z -position less than 3 cm. A stationary x-ray beam was

used. The charge q collected is multiplied by the chamber calibration factor (milligray per nanocoulomb) to obtain the scattered radiation dose in milligray.

RESULTS

To demonstrate that the CTDI_{100} measurements made with the Farmer chamber are the same as those obtained with the commonly used pencil chamber, CTDI_{100} was measured at both the peripheral and the central axes of the phantom with both ionization chambers, using CT irradiation parameters of 120 kVp, 300 mA, rotation $\tau = 1$ seconds, and $N \times T = 8 \times 1.25 = 10$ mm. We also calculated CTDI_{vol} from these measurements. An axial scan was used for measurements with the pencil chamber. A helical scan with pitch $p = 0.875$, table velocity

$$v = \frac{p \times N \times T}{\tau} = 8.75 \text{ mm/s}$$

for a total exposure time of $t_0 = 11.5$ seconds, and an acquisition scan length $L = vt_0 = 100.625$ mm were used for measurements with the Farmer chamber. The accredited dosimetry calibration laboratory calibration factor for the pencil chamber is 2.735 mGy/nC, and the calibration factor for the Farmer chamber is 41.69 mGy/nC. Table 1 compares the measured CTDI_{100} and CTDI_{vol} values using the pencil chamber and the Farmer chamber. It is easy to see that values for CTDI_{100} and CTDI_{vol} measured with both chambers agree within 1%.

To quantify the dose reduction achieved using the interior tomography method, we next measured CTDI_{100} values at the central and peripheral axes of the phantom with the Farmer chamber using a helical trajectory. CTDI_{vol} values were calculated from these values at various kVp and for 5 different SFOVs, 1 large (50 cm) SFOV, and 4 small SFOVs (8.6, 13.4, 16.8, and 20.6 cm) using the extra lead plate collimation to define the 4 different small SFOV apertures. These SFOVs were derived from the fan beam widths measured as increased film density from 4 sets of x-ray films at the isocenter. The measured CTDI_{100} and CTDI_{vol} values along with the percentage dose reduction for the small SFOVs as compared to the 50 cm SFOV are presented in Tables 2–4. From these tables, it can be seen that as the SFOV decreases, the percentage dose reduction increases. In addition, the percentage dose reduction is greater at the periphery than at the center as one would expect. Although the actual dose increases with increasing kVp, no significant differences are observed in percentage dose reduction for the same small SFOV. In other words, percentage dose reduction is a function of size of the SFOV but seems to be independent of the kVp.

To quantify the scatter reduction at the detectors achieved using the interior tomography method, we measured scattered radiation as previously described in the methods at the same SFOVs, kVp values, and mA. The calibration factor of the diagnostic pancake ionization chamber used is 0.1682 mGy/nC. The measured scattered radiation along with the percentage reduction for the small SFOVs as compared to the 50 cm SFOV are presented in Table 5. From this table, we can see that as the SFOV decreases, the percentage of scatter reduction increases. Although the actual scattered radiation increases with increasing kVp, no significant differences are observed in percentage reduction in scattered radiation for the same small SFOV. In other words, percentage reduction in scattered radiation is a function of size of the SFOV, but it seems to be independent of the kVp.

DISCUSSION

Progress in CT technology has led to an enormous increase in the use of CT for noninvasive imaging. Popular high-dose CT applications such as cardiac imaging, while providing

clinically meaningful data, have become a growing concern owing to the ionizing nature of x-rays and their potential risks to cause cancer. Several directions have been proposed to reduce CT x-ray dose. We believe that our recently proposed interior tomography method, which uses only the localized projection data to reconstruct the ROI, has a significant potential to reduce CT x-ray dose. As an added advantage, it can also reduce scattered radiation, which could provide improved reconstructed image quality. The reduction in x-ray dose and scattered radiation that could be achieved using our interior tomography method was quantified by measuring x-ray dose and scattered radiation for several SFOVs and by comparing them to the measurements for a large reference SFOV.

Our results indicate that as the x-ray beam SFOV is decreased, the dose at the center and the periphery also decrease, thus leading to an overall dose reduction. As expected, the dose reduction is greater at the periphery than at the center. Similarly, as the SFOV was decreased, the amount of scattered radiation also decreased. Although the actual dose and scattered radiation increase with the kVp, the percentage reduction in dose and scattered radiation seems to be a function of SFOV only and independent of kVp.

The measured dose and scattered radiation for a few different SFOVs give us an estimate of the x-ray dose and scatter radiation reduction potential of our interior tomography method. In addition, our measurements are limited to a circular SFOV that is adequate for cardiac imaging owing to the shape and location of the heart. For other interior applications, the SFOV could potentially be different in different views. A model for predicting x-ray dose and scattered radiation reduction using interior tomography is considered beyond the scope of this paper but will provide additional insights and capability to predict dose and scattered radiation reduction for any desired SFOV.

In conclusion, our interior tomography method has a great potential for x-ray dose reduction. The results indicate that it can reduce x-ray dose in the range of 18% to 58% and scatter in the range of 19% to 59%. Thus, we have demonstrated that it gives a significant dose reduction for interior applications such as cardiac CT and thus helps to make the use of CT safer. At the same time, reduced scatter radiation could provide improved reconstructed image quality.

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FIGURE 1.

Exact interior reconstruction settings. A, One known region adjacent to the reconstructible region. B, Two known regions adjacent to the reconstructible region, where $f(x)$ is the image and $g(x)$ the Hilbert transform.

FIGURE 2.

Difference between conventional CT and interior tomography. A, Conventional CT with the x-ray beam containing the object support. B, Interior tomography with the x-ray beam only covering the ROI, where 1 is for an object support, 2 for an ROI, and 3 for a known subregion within the ROI.

FIGURE 3.

Comparison of reconstructed images for conventional CT and interior tomography. A, Original image showing the ROI. B, Image reconstructed using FBP reconstruction from nontruncated projection data set. C, Region of interest image reconstructed with interior tomography. The narrow rectangular strip within the ROI is first reconstructed assuming the blood density in the rectangle within aorta as the known information followed by the ROI. Display window is $[-400,400]$ Hounsfield unit.

TABLE 1

Comparison Between Pencil Chamber and Farmer Chamber

	CTDI ₁₀₀ at Center, mGy	CTDI ₁₀₀ at Periphery, mGy	CTDI _{vol} , mGy
Pencil	20.38	40.6	33.86
Farmer	20.42	40.02	33.49
CT Technical Manual	19.2	38.58	32.12

TABLE 2
Percentage Reduction of CTDI₁₀₀ Values and CTDI_{vol} for Small SFOVs as Compared to the 50-cm SFOV at 80 kVp

SFOV, cm	CTDI ₁₀₀ at the Center		CTDI ₁₀₀ at the Periphery		CTDI _{vol} , mGy	Percentage Reduction
	Measured, mGy	Percentage Reduction	Measured, mGy	Percentage Reduction		
8.6	3.18	35.63	4.58	61.93	4.70	57.47
13.4	3.79	23.28	6.54	45.64	6.43	41.81
16.8	4.34	12.15	8.10	32.67	7.82	29.34
20.6	4.6	6.88	9.45	21.45	8.95	19.0
50	4.94 (5.05)	0	12.03 (12.73)	0	11.05 (11.62)	0

The values in parentheses are CTDI_{vol} values calculated from the CT technical manual.

TABLE 3
Percentage Reduction of CTDI₁₀₀ Values and CTDI_{vol} for Small SFOVs as Compared to the 50-cm SFOV at 100 kVp

SFOV, cm	CTDI ₁₀₀ at the Center		CTDI ₁₀₀ at the Periphery		CTDI _{vol} , mGy	Percentage Reduction
	Measured, mGy	Percentage Reduction	Measured, mGy	Percentage Reduction		
8.6	7.56	36.79	9.21	62.97	9.9	57.87
13.4	9.68	19.06	13.47	45.84	13.95	40.64
16.8	10.41	12.96	16.69	32.89	16.68	29.02
20.6	11.2	6.35	19.71	20.75	19.28	17.96
50	11.96 (11.68)	0	24.87 (25.58)	0	23.50 (23.94)	0

The values in parentheses are CTDI_{vol} values calculated from the CT technical manual.

TABLE 4
Percentage Reduction of CTDI₁₀₀ Values and CTDI_{vol} for Small SFOVs as Compared to the 50-cm SFOV at 120 kVp

SFOV, cm	CTDI ₁₀₀ at the Center		CTDI ₁₀₀ at the Periphery		CTDI _{vol} , mGy	Percentage Reduction
	Measured, mGy	Percentage Reduction	Measured, mGy	Percentage Reduction		
8.6	13.11	36.51	14.79	63.44	16.26	57.90
13.4	16.7	19.13	21.67	46.48	22.87	40.78
16.8	18.09	12.40	26.7	33.91	27.23	29.49
20.6	19.40	6.05	31.64	21.78	31.5	18.4
50	20.65 (20.48)	0	40.45 (41.93)	0	38.62 (39.75)	0

The values in parentheses are CTDI_{vol} values calculated from the CT technical manual.

TABLE 5

Percentage Reduction in Scatter Radiation to the Detectors for Small SFOVs as Compared to the 50-cm SFOV at Approximately 180 Degrees from the X-Ray Source

SFOV, cm	Scattered Radiation, μGy					
	80 kVp		100 kVp		120 kVp	
	Measured	Percentage Reduction	Measured	Percentage Reduction	Measured	Percentage Reduction
9.4	Not measurable		0.56	57.26	1.06	58.91
13.4	0.3	43.4	0.81	40.0	1.51	41.47
16.8	0.37	30.18	0.96	28.89	1.78	31.00
20.8	0.41	22.64	1.08	20.0	2.08	19.38
50	0.53	0	1.35	0	2.58	0