Computed Tomography Dosimetry: From Basic to State-of-the-art Techniques

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Abstract— **In this review, the basic and state-of-the-art techniques for evaluating radiation dose in computed tomography (CT) are described. CT dose index (CTDI) and dose-length product are indicators for measuring, comparing, and communicating the radiation output of a CT system. Although volume CTDI ($CTDI_{vol}$) is not the absorbed dose of an actual patient, the American Association of Physicists in Medicine (AAPM) has proposed conversion factors to translate CTDIvol into patient dose estimates at the center of the scanned volume to obtain size-specific dose estimates. Recently, several disadvantages of CTDI have been noted, especially for wide-beam CT. To eliminate these disadvantages, the International Electrotechnical Commission has described a modified CTDI definition that covers wide-beam CT. The AAPM has proposed measuring the accumulated dose at the scanning range midpoint to estimate the equilibrium dose instead of measuring CTDI. This review also introduces methods of obtaining the average organ dose from point measurements and Monte Carlo calculation, which are generally used for estimating the patient dose in CT.**

Keywords— **computed tomography, dosimetry, computed tomography dose index, dose-length product**

I. Introduction

Since the computed tomography (CT) scanner was first introduced for clinical use, the medical information derived from CT scans has contributed to saving many lives not only in developed countries but also in developing countries worldwide. The evolution of CT scanners has greatly enhanced their value in medical diagnosis. However, radiation doses in CT examinations have become relatively higher than those in radiological examinations [1]. The dose to an individual from one CT examination does not cause radiation-induced biological effects, but it is crucial to manage radiation dose in CT examinations appropriately.

When considering radiation dose in CT examinations, it is important to understand that the absorbed dose distribution within each patient differs from that of other X-ray examinations (e.g., radiography and fluoroscopy). This is because the X-ray beam is narrowed by passing through the collimator, and the exposure is controlled by using an X-ray tube that is rotated around the patient. Hence, specific methods must be used for evaluation of radiation doses in CT scans.

II. Computed tomography dose index

The CT dose index (CTDI) is a basic method for describing the doses delivered by CT scans [2]. CTDI is based on measuring the kinetic energy released per unit mass (kerma) of air in cylindrical polymethyl methacrylate phantoms 16 cm (for adult head and child) and 32 cm (for adult body) in diameter (Fig. 1). The index is measured from one axial CT scan and is calculated by dividing the air kerma by the product of slice thickness and the number of slices. The CTDI is defined by the following equation:

$$\mathrm{CTDI} = \frac{1}{BW}\int_{-\infty}^{+\infty} D(\mathrm{z})dz \qquad (1)$$

where BW is the nominal X-ray beam width along the z-axis, and D(z) is the dose profile along the z-axis, which consists of primary and secondary (scattered) radiation, from a single acquisition. The unit of CTDI is mGy.

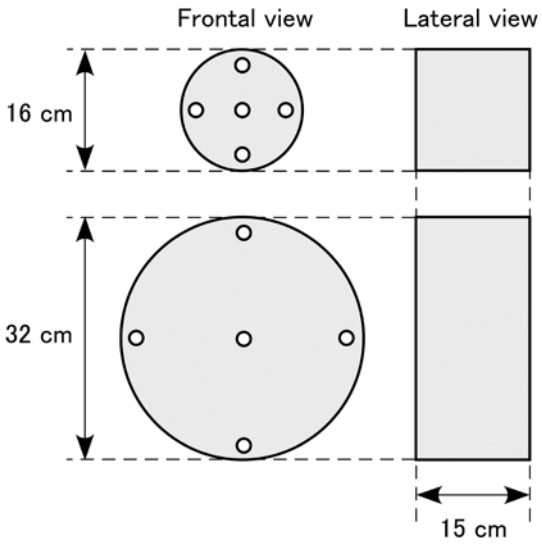


Fig. 1 Schematic of the polymethyl methacrylate phantoms

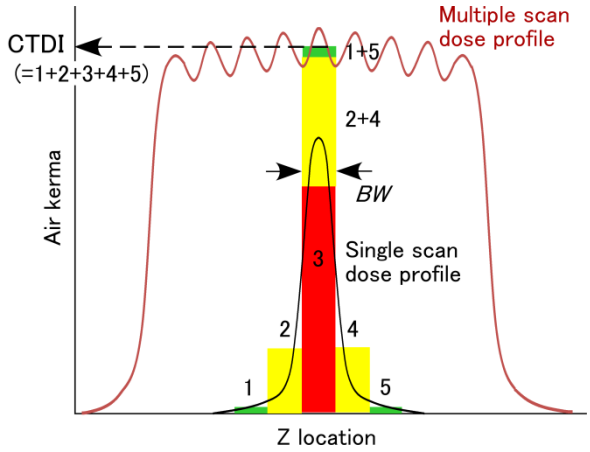


Fig. 2 Meaning of CTDI. When the radiation dose from a single scan equals the sum of areas 1 to 5, CTDI represents the sum of areas 1 to 5 divided by *BW*.

The index is measured by using a single acquisition, but it can be used to estimate the average dose from multiple acquisitions when the table is incremented during acquisitions. If all of the scatter tails are measured and the table increment equals *BW*, the result represents the average value in the central portion, which has the length of *BW*, of the multiple scan dose profile (Fig. 2).

A pencil-type ionization chamber that has a 100-mm active length is inserted in the phantom's holes to measure the index. However, the chamber can only measure the primary dose and scatter tails within a 100-mm length along the z-axis [3]. The index, which is called $CTDI_{100}$, is defined by the following equation:

$$CTDI_{100} = \frac{1}{BW}\int_{-50mm}^{+50mm} D(z)dz \tag{2}$$

Air kerma between the central and peripheral regions of the phantom are different in CT scans. To take this difference into consideration, the weighted CTDI ($CTDI_w$) is defined by the following equation:

$$CTDI_w = \frac{1}{3}\cdot CTDI_{100,c} + \frac{2}{3}\cdot CTDI_{100,p} \tag{3}$$

where $CTDI_{100,c}$ is $CTDI_{100}$ at the center of the phantom and $CTDI_{100,p}$ is the average of the $CTDI_{100}$ at four points along the periphery of the phantom. In other words, $CTDI_w$ represents the average air kerma over the in-plane direction [4].

To represent the dose for a consecutive CT scan, it is essential to take pitch, gaps, or overlaps into consideration. $CTDI_{vol}$ is defined by the following equations.

1. In the case of sequential acquisitions:

$$CTDI_{vol} = \frac{BW}{I}\cdot CTDI_w \tag{4}$$

2. In the case of helical acquisitions:

$$CTDI_{vol} = \frac{1}{p}\cdot CTDI_w \tag{5}$$

where *I* is the table increment between each acquisition, and *p* is the pitch factor (= table feed per rotation/nominal X-ray beam width along z-axis). From these equations, the local air kerma for a specific CT protocol can be obtained. $CTDI_{vol}$ is the most familiar dose parameter because it is regulated to be displayed on the console of CT scanners [2].

As a dose descriptor in CT, the multiple scan average dose (MSAD) is also used. The air kerma for a certain part of the cylindrical polymethyl methacrylate phantom (Fig. 1) with multiple acquisitions is measured by using a small dosimeter, such as a thermoluminescent dosimeter (TLD) or radiophotoluminescent glass dosimeter (RPLD) [2]. Theoretically, the MSAD and CTDI are equivalent dose values because MSAD equals the dose value integrated over the dose profile for one rotation, which is equal to the CTDI. In the early days of CT, direct measurement of the MSAD was generally performed, but it required multiple scan acquisitions, which placed heavy loads on the X-ray tube [5].

One should know that the $CTDI_{vol}$ is not the absorbed dose of an actual patient, but CTDI is an indicator for measuring, comparing, and communicating the radiation output of a CT system [6]. For estimating patient doses from $CTDI_{vol}$, the American Association of Physicists in Medicine (AAPM) has released conversion factors to translate $CTDI_{vol}$ into patient dose estimates at the center of the scanned volume, which are described in section V [7].

Recently, some disadvantages of CTDI have been pointed out [8-12]. First, a 100-mm-long pencil ionization chamber used to collect the dose may not be sufficiently long to measure all of the tails of the scattered dose distribution. Second, the phantoms used for CTDI measurements are shorter than an adult torso and so do not produce as much scattered radiation as would occur in a typical adult. To address these limitations, the International Electrotechnical Commission (IEC) has described a modified CTDI definition in Amendment 1 of the third edition of report 60601-2-44. The definition of $CTDI_{100}$ [equation (2)] is retained for a nominal X-ray beam width along the z-axis of ≤40 mm; when the width is >40 mm, the $CTDI_{100}$ is defined as follows (Fig. 3):

$$CTDI_{100} = \frac{1}{BW_{Ref}}\int_{-50mm}^{+50mm} D(z)dz \cdot \frac{CTDI_{Air,BW}}{CTDI_{Air,Ref}} \tag{6}$$

where BW_{Ref} is the reference nominal X-ray beam width along the z-axis, which is at or near 20 mm, $CTDI_{Air,BW}$ and $CTDI_{Air,Ref}$ are the CTDI in air for the desired and reference nominal X-ray beam widths along the z-axis, respectively. CTDI in air is defined by the following equation:

$$\mathrm{CTDI}_{\mathrm{Air}} = \frac{1}{BW}\int_{-L/2}^{+L/2} D(z)dz \tag{7}$$

where L is the air kerma integration length, which is set to the desired nominal X-ray beam width along the z-axis plus 40 mm, with a minimum total length of 100 mm. When L is >100 mm, a pencil-type ionization chamber that has an appropriate active length is prepared, or a pencil-type ionization chamber that has a 100-mm active length is used by performing a two- or three-step measurement (Fig. 4). This methodology has been adopted by the International Atomic Energy Agency in its Human Health Report 5 [13].

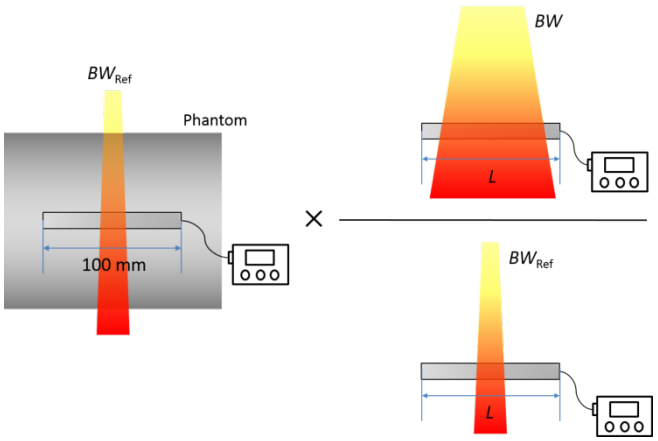


Fig. 3 Modified CTDI definition for an X-ray nominal beam width along the z-axis of >40 mm

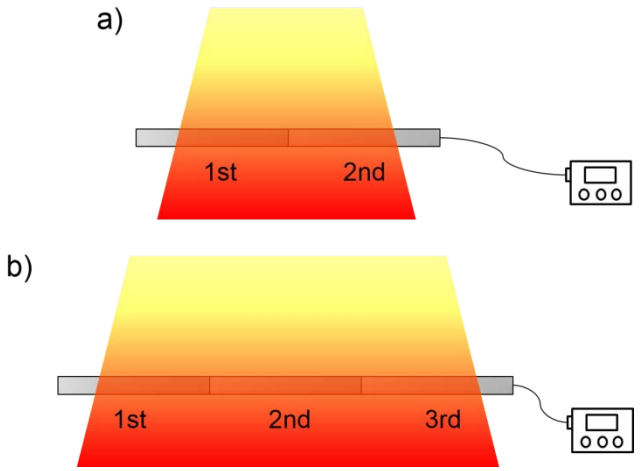


Fig. 4 Two- and three-step measurement process to measure CTDI by using the modified CTDI methodology

III. Dose-length product

The dose-length product (DLP) represents the total dose over a whole scan and is defined by the following equations.
1. In the case of sequential acquisitions:

$$\mathrm{DLP} = \mathrm{CTDI}_{\mathrm{vol}} \cdot \Delta d \cdot N \tag{8}$$

2. In the case of helical acquisitions:

$$\mathrm{DLP} = \mathrm{CTDI}_{\mathrm{vol}} \cdot L \tag{9}$$

3. In cases in which the table is not incremented during acquisitions:

$$\mathrm{DLP} = \mathrm{CTDI}_{\mathrm{vol}} \cdot n \cdot L \tag{10}$$

where Δd is the table increment per rotation, N is the number of acquisitions, L is the scanning length, and n is the number of slices generated from one sequential acquisition. The unit of DLP is mGy·cm. DLP is also used as an indicator of radiation output of a CT system, but the patient effective dose can be estimated from DLP by using the following equation:

$$\mathrm{E} = k_{\mathrm{E}} \cdot \mathrm{DLP} \tag{11}$$

where k_{E} is the conversion factor (mSv·mGy^{-1}·cm^{-1}) that depends on patient age and scanning regions [14].

The concept of effective dose was introduced by the International Commission on Radiological Protection in 1977 [15] and revised in 1991 and 2007 [16,17]. Tissue weighting factors, which are used for calculating the effective dose, have also been revised according to the latest findings with regard to the radiation effect for each organ or tissue.

IV. Equilibrium dose method

A previous study showed that the dose at the center of the scan range may increase with longer phantoms and scan lengths, and asymptotically approaches the equilibrium dose for large scan lengths [8,18,19]. The relationship between equilibrium dose and CTDI is shown by the following equation:

$$D_{\mathrm{eq}} = \frac{CTDI}{p} \tag{12}$$

where D_{eq} represents the equilibrium dose for a large scanning length.

The AAPM has released report 111, “Comprehensive Methodology for the Evaluation of Radiation Dose in X-Ray Computed Tomography” [18]. In this report, the AAPM has proposed measuring the accumulated dose at the midpoint of the scanning range, which is defined in the equation below, to estimate the equilibrium dose instead of measuring the CTDI.

$$D_L(0) = \frac{1}{a}\int_{-L/2}^{L/2} f(z)dz \tag{13}$$

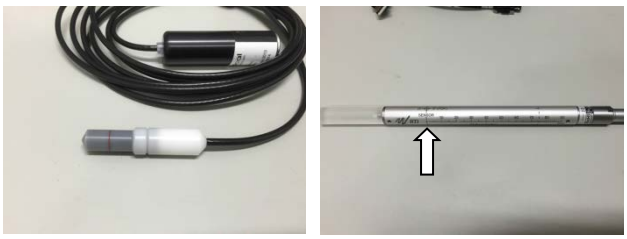
where $D_L(0)$ is the accumulated dose at the midpoint of the scanning range, a is the scan interval, and $f(z)$ is the full dose profile.

For estimating the equilibrium dose from the accumulated dose, the equilibrium function $H(L)$,

$$H(L) = \frac{D_L(0)}{D_{eq}} \tag{14}$$

is required theoretically. The report stated that the equilibrium dose method needs phantoms that are sufficiently long. For example, a water-filled, 30-cm diameter, and 50-cm long phantom is designed to be transported empty, and once placed on the table, it can be filled with water. The AAPM-International Commission on Radiation Units and Measurements CT phantom comprises high-density polyethylene and is 30-cm in diameter and 60-cm long. The phantom is designed to be modular with three different sections. The cylindrical polymethyl methacrylate (PMMA) phantoms, which are used for measuring CTDI, can also be used to be assembled contiguously for requisite lengths. A previous paper has showed that the phantom length that is required for the radiation dose profile measurement should be at least 75 cm (five PMMA phantoms) with the maximum beam width of 160 mm [20].

For measuring $D_L(0)$, a thimble ionization chamber with an active length of 20–35 mm for charge collection and a nominal collection volume of at least 0.6 cm^3 should be used (Fig. 5a). A small solid-state detector can also be used for this purpose [20,21] (Fig. 5b).



a) b)

Fig. 5 Examples of small dosimeters for measuring accumulated dose at the midpoint of the scanning range: a) a Farmer-type 0.6-cm^3 ionization chamber (10X6-0.6CT; Radcal, Monrovia, CA, USA), b) a small solid-state detector (placed 3 cm from the end of the probe [white arrow]. CT Dose Profiler; RTI Group AB, Mölndal, Sweden)

V. Size-specific dose estimates

Although $CTDI_{vol}$ is not the absorbed dose of an actual patient, the AAPM released report 204, "Size-Specific Dose Estimates in Pediatric and Adult Body CT Examinations" [22], which provided conversion factors as a function of geometric patient size to translate $CTDI_{vol}$ to patient dose estimates at the center of the scanned volume, which was named size-specific dose estimates (SSDE).

In this report, four different measurements of torso thickness are used to represent patient size: the anteroposterior dimension (AP), the lateral dimension (LAT), the sum of the dimensions (AP + LAT), and the effective diameter (square root of the product of AP and LAT). For example, Mueller et al. [23] showed that SSDE estimates the rectal absorbed dose reasonably during CT colonography.

However, X-ray attenuation is the fundamental physical parameter that affects absorption of X-rays and thus, is more relevant than geometric patient size. Hence, the AAPM released report 220, "Use of water equivalent diameter for calculating patient size and SSDE in CT" [24], which provided conversion factors as a function of X-ray attenuation to calculate SSDE for all patients, with little or no user intervention.

Previous work has proposed the concepts of a water-equivalent area and diameter [25-29]. The water-equivalent area can be represented in terms of CT numbers, as shown in the following equation:

$$A_w = \sum \frac{\mu(x,y)}{\mu_{water}} \cdot A_{pixel} = \sum \left(\frac{CT(x,y)}{1000} + 1 \right) \cdot A_{pixel} \tag{15}$$

where A_w is the water-equivalent area, A_{pixel} is the area of a pixel in the CT image, μ is the linear attenuation coefficient, and $CT(x,y)$ is the CT number of a voxel. The water-equivalent diameter is shown by the following equation:

$$D_w = 2\sqrt{A_w / \pi} \tag{16}$$

where D_w is the water-equivalent diameter. A previous study showed that using the water equivalent diameter from one image in the center of the scan range and the mean $CTDI_{vol}$ from the entire scan provided a sufficiently accurate method for calculating the mean SSDE for CT examinations of the torso in adults [30].

Estimating the water-equivalent diameter can only be performed by using reconstructed CT images. Although two studies have shown that patient attenuation can be estimated by using CT localizer radiography [26,27], the CT localizer radiograph-based method for estimating the water-equivalent diameter is not recommended because it requires calibration of the CT localizer radiograph pixel values in terms of water attenuation.

VI. Physical measurements

Obtaining the organ dose from point measurements is another effective method in CT dosimetry. Sectioned and drilled phantoms, such as the Alderson RANDO phantom [31-33] and ATOM phantom [34-37], are used (Fig. 6). These phantoms accept small dosimeters, such as TLD [31,33], RPLD [32], metal-oxide-semiconductor field-effect transistor dosimeters [35,36,38], semiconductor detectors

[21,37], photo diode dosimeters [39,40], and optically stimulated luminescence (OSL) dosimeters [41].

The energy dependency of these small dosimeters within the energy range generally used in CT is relatively high; hence, they must be calibrated with the effective energy used. One of the methods for calibration is to compare dose values with those of an ionizing chamber by using a diagnostic X-ray system. The chamber and small dosimeters are placed adjacent to each other at the same distance from the X-ray focus in an irradiated field. Radiographic or radiochromic film may also be used instead of small dosimeters. The film is placed between any two contiguous sections, which are then sealed with black tape to prevent any exposure of the film to light.

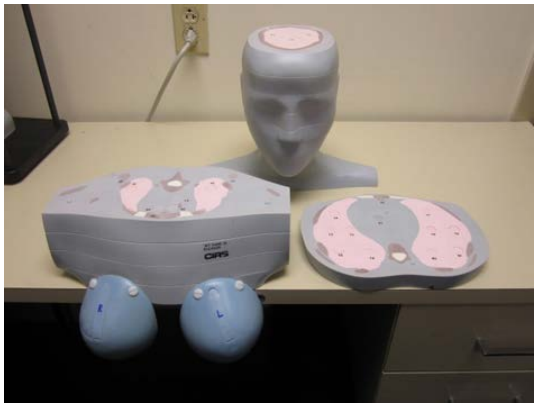


Fig. 6 An example of an anthropomorphic phantom (ATOM model 702; CIRS, Norfolk, VA, USA); it has holes for inserting small dosimeters

When TLD, RPLD, or OSL are used as small dosimeters, they should be initialized beforehand by heating (TLD and RPLD) or irradiating visible light (OSL). After initializing, the initial dose values should be read. Then, they are placed at the drilled holes that are located corresponding to targeted tissues and organs. Thereafter, the phantom is placed on the CT table and scanned. If possible, the scan should be performed multiple times by using separate sets of small dosimeters to reduce uncertainty and random error.

After scanning, the small dosimeters are removed from the phantom, and the dose values are read after adequate time has passed (for TLD) or preheating has been performed (for RPLD) to stabilize the obtained values. Examples of adequate times are 1 h for BeO and from 12 to 24 h for $CaSO_4$.

As shown in the following equation, the absorbed dose for each organ is obtained by multiplying the averaged value of the organ or tissue, the calibration factor of the small dosimeters, and the ratio of mass energy-absorption coefficients for each organ or tissue to air:

$$D_T = (M_T - M_I) \cdot k_C \cdot \frac{(\mu_{en}/\rho)_T}{(\mu_{en}/\rho)_A} \quad (17)$$

where M_T is the averaged dose value from the small dosimeters placed at locations corresponding to each organ or tissue, M_I is the averaged initial dose value, k_C is the calibration factor of the small dosimeter, $(\mu_{en}/\rho)_T$ is the mass energy-absorption coefficient for each organ or tissue, and $(\mu_{en}/\rho)_A$ is the mass energy-absorption coefficient for air.

VII. Simulation Method

Without using anthropomorphic phantoms and small dosimeters, the absorbed dose for each organ or tissue for typical clinical CT scanner models and scan protocols can be calculated on the basis of Monte Carlo (MC) simulation software. One example is the ImPACT CT Patient Dosimetry Calculator software (St. George's Hospital, London, UK) (Fig. 7) [42]. This software uses the National Radiological Protection Board MC dose data sets produced in report SR250 [43] and provides normalized organ dose data for irradiation of a mathematical (Medical Internal Radiation Dose [MIRD]) phantom.

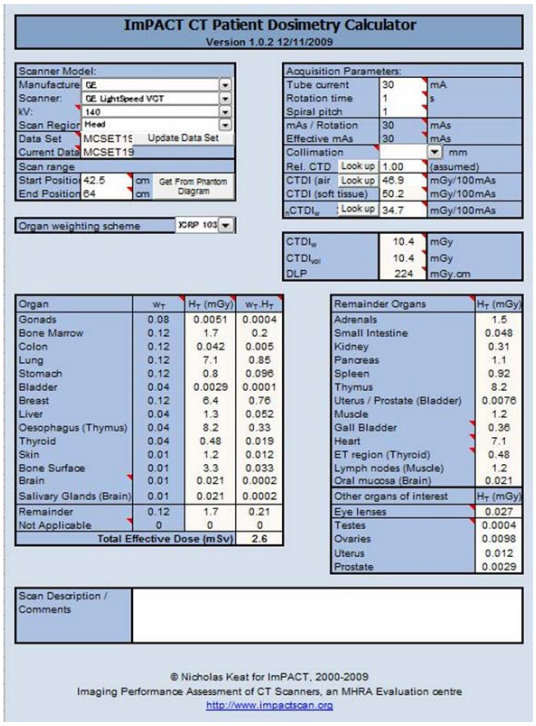


Fig. 7 ImPACT CT Patient Dosimetry Calculator [42]

There are other MC simulation programs, such as CT-Expo [44], ImPACTDose [45], and WAZA-ARIv2 [46]. CT-Expo offers dose calculation for adults, children, and infants, and takes into account overbeaming, overranging, and dose modulation (longitudinal and three-dimensional) effects. ImPACTDose offers anthropomorphic phantoms represented by 12 phantoms of both sexed and different ages (newborn, 1, 5, 10, 15 years old and adult) as well as

two voxelized human phantoms. The WAZA-ARIv2 is a web-based CT dose calculator, that can calculate organ doses of 18 body types of patients, including adults and children. The Particle and Heavy Ion Transport Code System (PHITS) [47] has been used for developing this web-based software.

In addition, there are several types of MC packages that are used for CT dosimetry, such as Monte Carlo N-Particle eXtended (MCNPX) [48-53] and Electron Gamma Shower (EGS) [54,55]. When the MC packages are used, source models of the CT scanner and human models need to be set manually for calculating absorbed doses in CT.

VIII. Conclusions

In this review, the basic and state-of-the-art techniques for evaluating radiation dose in CT are described. Understanding these techniques is necessary not only for measuring, comparing, and communicating the radiation output of a CT system but also for estimating patient dose in CT. Medical physicists should understand these techniques clearly before performing quality control in CT and optimizing patient dose and scanning protocols.

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