

Radiation Dose in Computed Tomography of the Heart

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Currently, computed tomographic (CT) imaging of the heart is mainly used for the quantification of coronary artery calcification as an indirect measure of coronary plaque burden^{1,2} and, less frequently, for minimally invasive coronary angiography.³ CT imaging of the heart and coronary arteries without unsharpness due to motion artifact first became possible with the introduction of electron beam computed tomography (EBCT) in 1983.⁴ More recently, so-called multislice spiral computed tomographic (MSCT) scanners with gantry rotation speeds fast enough to produce diagnostic images of the heart under certain conditions have become widely available.⁵ As a consequence, cardiac CT imaging, most often performed for the purpose of calcium scoring,² is increasingly applied to the general public. In many centers, patients have access to such studies without physician referral. This has created concerns for public health because of the radiation dose associated with CT imaging.⁶⁻⁸

Many clinicians and researchers working with patients with cardiovascular diseases may yet be unfamiliar with the radiation doses that are received during various cardiac CT imaging protocols and how they differ between the various scanner types that are currently used. To further complicate matters, radiation dose estimates can be expressed in various ways. For these reasons, the doses reported in previous publications on cardiac CT have varied widely, and it is not always clear what parameters were being reported.^{3,9-11}

The purpose of this article is to discuss the current concepts of radiation dose measurement and estimation in CT imaging and to provide comparative estimates for radiation doses received during cardiac examinations with use of EBCT or MSCT. This information may be helpful to physicians who perform calcium scoring, counsel patients contemplating cardiac calcium scoring, or are considering referring their patients for such studies.

Scanner Types

EBCT scanners acquire 1 scan at a time, using a technique termed "prospective ECG-triggering," in which radiation is produced only during a predetermined instant of the cardiac cycle.⁴ With current technology, MSCT scanners can acquire several (currently 2, 4, 8, or 16) parallel scans simultaneously, using either prospective ECG-triggering or another technique termed "retrospective gating." When retrospective gating is used, radiation is produced and image data are

acquired continuously throughout the cardiac cycle, although images are usually reconstructed only during ventricular diastole.⁵ The continuous production of radiation throughout the cardiac cycle causes the radiation dose of retrospectively gated studies to be higher than that of prospectively triggered studies if the degree of image noise is to be equivalent in both types of studies.

As with conventional x-ray equipment, image noise in CT images decreases and image quality increases with the number of photons received by the detector array. The number of photons received by the detector array is dependent on the number of photons produced by the x-ray tube and the degree of photon attenuation by the patient's body. In addition, smaller scan widths result in higher image noise because of the lower number of photons used in the image formation. The number of photons produced by the x-ray tube increases with the x-ray tube voltage, which is expressed in peak kilovolts (kVp), and with the product of the tube current multiplied by the exposure time, expressed in milliamperere seconds (mAs). To maintain the degree of image noise for large patients or narrow scan widths, the number of photons produced by the tube must be increased. This results in an increase of the radiation dose received by the patient.¹²

With current EBCT technology, these scanner settings are limited to 130 kVp and 65 mAs, respectively, for the 100-ms exposure time used for cardiac imaging.¹³ Therefore, adjustments of the scanner settings to compensate for increased attenuation, eg, in patients with large body size, are not possible. In contrast, MSCT scanners available from various manufacturers allow modification of a wider range of scan parameters to adjust for patients' body size to avoid an increase of image noise due to increased photon attenuation. Therefore, the variability of the MSCT scanner radiation doses reported in the literature may in part be due to differences in these scanner settings.

To directly compare radiation doses delivered by EBCT and the various MSCT scanners, it is important to give specific information about the MSCT scan parameter settings and the resultant image noise.¹² In addition, it is essential to specify the type of dose parameter being compared.

Fundamental Concepts in Radiation Dose

It is important to distinguish "radiation exposure," which relates to the quantity of ionization events in air that are

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(*Circulation*. 2003;107:917-922.)

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Circulation is available at <http://www.circulationaha.org>

DOI: 10.1161/01.CIR.0000048965.56529.C2

TABLE 1. Radiation Dosimetry Parameters

Variable	Parameter	Physical Equivalent	SI Unit*	Conventional Unit*
Radiation exposure	CTDI ₁₀₀	No. of ions produced in air by photons	C/kg	roentgen (R)
Absorbed radiation dose	CTDI, CTDI _w , CTDI _{vol} , MSAD	Radiation energy absorbed by patient's body	gray (Gy)	rad
Cumulative radiation dose	DLP	Total radiation energy absorbed by patient's body	mGy × cm	mrad × cm
Effective dose	E	Biological effect of radiation dose received	sievert (Sv)	rem

C indicates coulomb; CTDI, computed tomography dose index; CTDI₁₀₀, radiation exposure measured by means of an ionization chamber with a length of 100 mm; CTDI_{vol}, volume CTDI; CTDI_w, weighted average of peripheral and center values of CTDI₁₀₀; DLP, dose-length product; E, effective dose; MSAD, multiple scan average dose; and SI, Système International d'Unités.

*Conversion factors between SI and conventional units: 1 C/kg=3876 R; 1 Gy=100 rad; 1 Sv=100 rem.

produced by x-ray photons, from “radiation dose” (“absorbed radiation dose”), which describes the amount of radiation energy deposited in the patient’s body as a result of exposure. Unfortunately, the terms “exposure” and “dose” are sometimes used interchangeably. Radiation exposure is usually a measured quantity, whereas the absorbed dose is typically calculated from the exposure and from estimates of energy absorption per body mass unit (eg, kilograms of body weight). The fact that radiation dose can be expressed in many different ways makes it difficult to compare doses that have been reported for specific CT applications in the published literature.

The fundamental radiation dose parameter in CT is the computed tomography dose index (CTDI).¹⁴ The volume CTDI (CTDI_{vol}), a derivative of the CTDI, can be used to express the average dose delivered to the scan volume for a specific examination. Another important parameter is the effective dose (E), which is useful in assessing and comparing the potential biologic risk of a specific examination.¹⁵ The CTDI is measured, whereas E and other radiation dose parameters discussed below are calculated from the CTDI.

For MSCT systems, until recently, differing assumptions were made in the calculation of derived radiation dose parameters, and these were surrounded by controversy. In addition to the possible differences in scanner settings mentioned above, this is another reason why the results of MSCT dosimetry could vary by a factor of 4 or more. These definitional issues were resolved in a consensus agreement on CT dosimetry parameters by the International Electrotechnical Commission.¹⁶ The following sections will discuss the radiation dose parameters put forth in this international standard and the relationships among them. The various parameters with their units of measure are summarized in Table 1.

CTDI

The CTDI is usually measured with thermoluminescent dosimeters,¹⁷ but other measurement techniques are available. This measurement is very labor-intensive and therefore is rarely performed. The measured results represent absorbed dose, and the SI unit of measurement is the gray (Gy). The conventional unit is the rad. The CTDI value represents the integral under the radiation dose profile in the z-axis (Figure 1) of a single scan that would produce 1 tomographic image (Figure 2). The maximum of the radiation dose profile is termed the “peak dose.”

CTDI₁₀₀

The CTDI₁₀₀ is a measured parameter of radiation exposure. This measurement is more convenient than the CTDI and is the measurement of choice performed by medical physicists in the clinical setting. It is obtained with an ionization chamber (Figure 3) that integrates the radiation exposure of a single axial scan over a length of 100 mm.^{18–20} The ionization events occurring in the chamber produce a current that is proportional to the number of ionization events.

The ionization chamber is placed in a commercially available, round polymethylmethacrylate (Plexiglas) phantom of 16- or 32-cm diameter (Figure 3). The SI unit of measure is coulomb/kg (C/kg). The conventional unit is the roentgen (R). The CTDI₁₀₀ is measured in the center of the Plexiglas phantom, as well as at peripheral locations (at the 12, 3, 6, and 9 o'clock positions) to represent the spatial distribution of the radiation exposure. The measured exposure can be converted to dose as described in the next section.

CTDI_w

The CTDI_w is the weighted average of the CTDI₁₀₀ measurements at the center and the peripheral locations of the phantom. This parameter reflects the average absorbed dose over the x and y dimensions (Figure 1) of the Plexiglas phantom as an approximation of the average radiation dose to

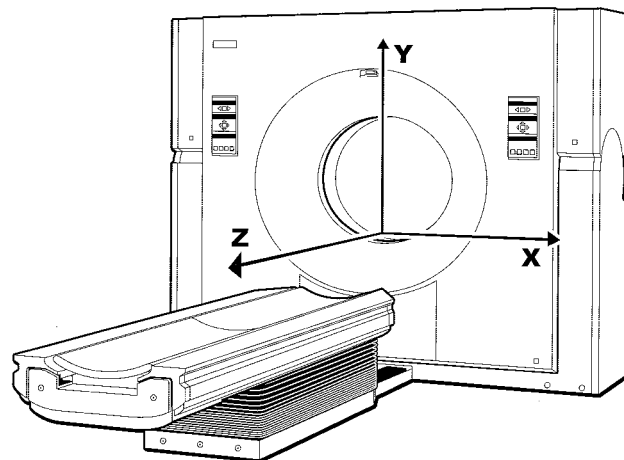


Figure 1. Coordinate system used for computed tomographic scanning. The x- and y-axes are oriented “transaxially,” that is, horizontally and vertically within the plane of a scan. Successive parallel scans are arranged “axially” along the z-axis. The “width” of an individual scan is determined along the z-axis.

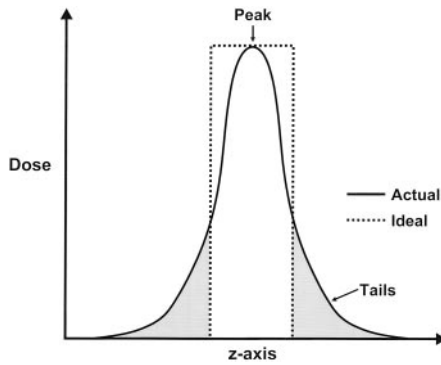


Figure 2. Distribution of radiation dose in a single CT scan. In an ideal situation, radiation dose would drop sharply at the edges of a scan. In an actual situation, however, the radiation dose profile broadens gradually at the edges of a scan with tails at either side. These tails are due to both x-ray beam divergence and to internal radiation scatter by the body tissues. The maximum of the radiation dose profile represents the “peak dose.”

a cross section of a patient’s body. The CTDI_w is calculated using Equation 1:

$$(1) \quad \text{CTDI}_w = [2/3 \text{CTDI}_{100}(\text{Periphery}) + 1/3 \text{CTDI}_{100}(\text{Center})] \times f$$

The term *f* reflects the difference between the absorption of radiation in air and the absorption in another media. It is used to convert radiation exposure, expressed in C/kg, into absorbed dose, expressed in Gy (the SI unit for CTDI_w). For the calculation of CTDI_w, the appropriate value for *f* is 33.7 Gy × C⁻¹ × kg⁻¹ (in conventional units, 0.87 rad/R).²¹

Multiple Scan Average Dose

The multiple scan average dose (MSAD) is the average radiation dose over the central scan of a CT study consisting of multiple parallel scans¹⁴ (Figure 4). The MSAD correctly describes the average patient dose only if the scan protocol uses more than just a few parallel scans.¹⁴ Like the CTDI, the MSAD requires thermoluminescent dosimeters for measurement^{17,22} and is rarely performed. Typically, the MSAD is

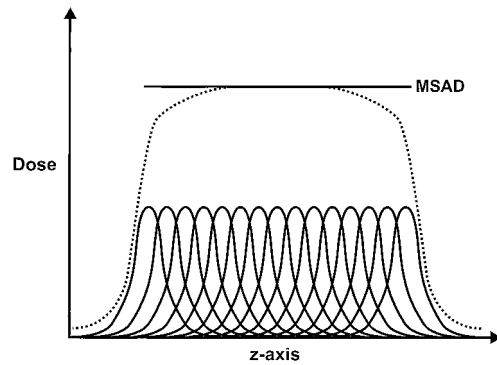


Figure 4. Typical relationship of the distributions of radiation dose of multiple parallel scans. Depending on the relationship of scan width and patient table advance between successive scans, an accumulation of dose due to scan overlap may occur. The value of the multiple scan average dose (MSAD) is determined by the extent of slice overlap and is valid only if the scan protocol uses more than just a few scans.

higher than the peak of the radiation dose profile of a single scan (Figure 2) by a factor of 2 to 3. The SI unit of measurement for the MSAD is the Gy.

The numeric value of the MSAD is directly related to the spatial separation of successive scans. This spatial separation is dependent on the advance of the patient table during a spiral CT examination. The table advance is quantified as a dimensionless variable termed “pitch.” Since the advent of MSCT, several definitions of pitch have been used because of controversy as to whether the table advance should be expressed relative to the width of a single scan or relative to the combined width of all simultaneously acquired scans. This is one of the reasons for the high degree of variance among radiation dose values reported in the literature for cardiac MSCT.

This controversy was resolved in the international consensus agreement on CT dosimetry parameters.¹⁶ Pitch is now defined as the distance of patient table advance in the z direction (mm) per 1 rotation of the gantry divided by the total nominal scan width. For MSCT systems, the total nominal scan width includes all simultaneously acquired scans and corresponds to the distance (mm) in the z direction covered by all detector rows that are active during a scan (Figure 5). If the patient table advance during 1 gantry rotation is less than the total nominal scan width (ie, pitch <1), scan overlap occurs. Scan overlap increases as pitch decreases.

The MSAD for non-spiral scans can be estimated from the CTDI using Equation 2:

$$(2) \quad \text{MSAD} = \frac{N \times T}{I} \times \text{CTDI}$$

where *N* is the number of scans, *T* is the nominal scan width (mm), and *I* is the distance between scans (mm). For MSCT systems, *N* × *T* is the total nominal scan width, and *I* corresponds to the patient table advance during 1 gantry rotation. Therefore, given the definition of pitch above, the MSAD for spiral scans can be expressed as

$$(3) \quad \text{MSAD} = \frac{1}{\text{Pitch}} \times \text{CTDI}$$

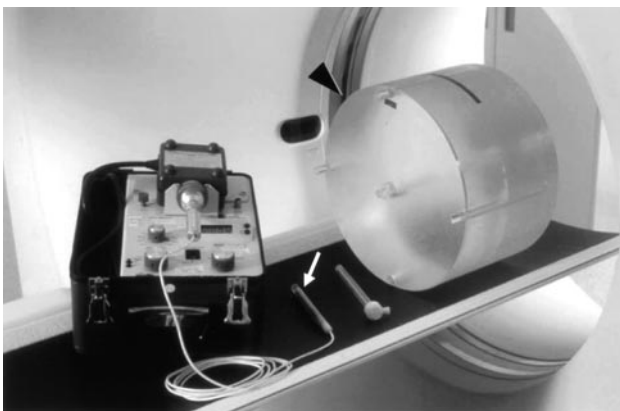


Figure 3. Instrumentation for measurement of radiation exposure in computed tomography. The white arrow indicates the ionization chamber; black arrowhead, Plexiglas phantom.

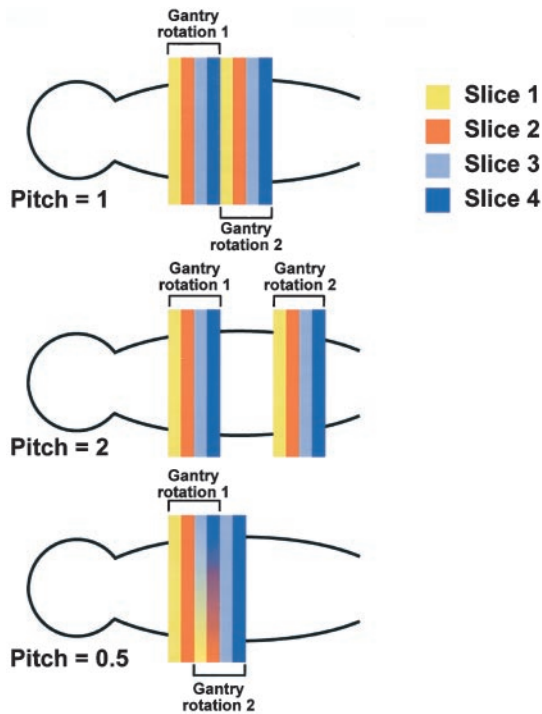


Figure 5. Examples of different degrees of pitch during a CT examination, relative to total nominal scan width in an MSCT scanner acquiring 4 slices simultaneously per gantry rotation. Upper panel, Pitch=1 (no scan overlap). Center panel, Pitch=2 (gap between the scans acquired during 2 successive gantry rotations). Lower panel, Pitch=0.5 (overlap by 50% of total nominal scan width between 2 successive gantry rotations).

From this equation, it follows that if pitch is equal to 1, the MSAD is equal to the CTDI. Therefore, a correct and consistent definition of pitch is crucial to ensure comparability of MSAD calculations among different CT imaging systems and protocols. The MSAD increases with a decrease in pitch.

CTDI_{vol}

The volume CTDI (CTDI_{vol}) is a new radiation dose parameter agreed on by the International Electrotechnical Commission.¹⁶ It is based on the same concept as the MSAD, but unlike the MSAD, it is derived from the CTDI_w. This is advantageous because the CTDI_w is calculated from data that are easily measured (CTDI₁₀₀). The CTDI_{vol} averages radiation dose over x, y, and z directions (CTDI_w represents the average exposure in the x-y plane only). The CTDI_{vol} for single-slice scanners is defined as:

$$(4) \quad CTDI_{vol} = \frac{N \times T}{I} \times CTDI_w$$

Similar to the conversion of the MSAD for non-spiral scanners (Equation 2) to the MSAD for spiral scanners (Equation 3), the CTDI_{vol} for MSCT can be defined as:

$$(5) \quad CTDI_{vol} = \frac{1}{Pitch} \times CTDI_w$$

The CTDI_{vol} is now the preferred expression of radiation dose in CT dosimetry. As with the MSAD, the correct and

consistent definition of pitch is crucial for accurate and comparable determinations of the CTDI_{vol}.

Most current CT scanners can display the value for the CTDI_{vol} on the operator's console. This allows the clinician to compare the radiation doses that patients receive from different imaging protocols. Similar to the CTDI_w, the CTDI_{vol} is expressed in SI units of Gy.

However, both the CTDI_{vol} and the MSAD have an important disadvantage as measures of radiation dose: Their numeric value depends only on the spatial distribution of individual scans and is unrelated to the total number of successive scans in a CT examination (ie, total scan length).

Dose-Length Product

The dose-length product (DLP)²¹ is an indicator of the integrated radiation dose of an entire CT examination. The DLP incorporates the number of scans and the scan width (ie, total scan length). The definition of DLP is:

$$(6) \quad DLP = CTDI_{vol} \times \text{Scan Length}$$

Therefore, DLP increases with an increase in total scan length or with variables that affect the CTDI_w (eg, tube voltage or tube current) or the CTDI_{vol} (eg, pitch). Because scan length is expressed in centimeters, the SI unit for DLP is milligray (mGy) × cm. The display of the DLP value on the console of the CT scanner is required by law in many countries in Europe.

Although the DLP reflects most closely the radiation dose for a specific CT examination, its numeric value is affected by variances in patient anatomy (eg, the value of DLP is higher for taller patients only because of their greater height). Therefore, the CTDI_{vol} is more useful in designing CT imaging protocols and comparing radiation doses among different protocols.

Effective Dose

The effective dose reflects the nonuniform radiation absorption of partial body exposures relative to a whole body radiation dose and allows comparisons of risk among different CT examination protocols. The SI unit of measure is the sievert (Sv) or millisievert (mSv). The conventional unit is the rem. The effective dose is calculated from information about dose to individual organs and the relative radiation risk assigned to each organ.^{15,23} A technique called Monte Carlo simulation²⁴ is used to determine specific organ doses by simulating the absorption and scattering of x-ray photons in various tissues by using a mathematical model of the human body. Appropriate organ risk-weighting factors have been published by the International Commission on Radiological Protection.¹⁵ However, several data sets and calculation methods exist for determining the effective dose. Therefore, results of the calculation of the effective dose can vary, depending on which method is used, but they are generally in good agreement.²³

A reasonable approximation of the effective dose can be obtained using the equation²¹

$$(7) \quad E = k \times DLP$$

TABLE 2. Effective Dose of Selected Radiological Examinations

Examination	Effective Dose, mSv
Head CT	1–2
Chest CT	5–7
Abdomen and pelvis CT	8–11
Diagnostic coronary angiogram	3–10
PA and lateral chest x-ray	0.04–0.06
Average annual background radiation in the United States	3.6

PA indicates posteroanterior.

where E is the effective dose and *k* is a conversion factor (unit, mSv × mGy⁻¹ × cm⁻¹) that varies dependent on the body region that is imaged (*k* values have been published for the head, neck, thorax, abdomen, and pelvis).²¹ Typical effective dose values for frequently performed radiological examinations are listed in Table 2.

Radiation Dose in CT Examinations of the Heart

The CTDI_{vol} and effective dose values for 3 CT scanning systems that are currently used for cardiac imaging are given in Table 3. These data represent the radiation doses of typical protocols for coronary calcium scanning and CT coronary angiography. The data in Table 3 demonstrate that, in MSCT scanners that acquire 4 scans simultaneously, the radiation dose for a prospectively triggered calcium-scoring examination is approximately 25% of the radiation dose used for a

retrospectively gated calcium-scoring examination. With triggered acquisition, the radiation doses of EBCT and MSCT calcium-scoring studies are approximately equivalent.

New Technical Advances

The first generation of MSCT scanners can acquire up to 4 scans simultaneously. Very recently, a new generation of MSCT scanners has become commercially available that can simultaneously acquire up to 16 scans with a width under 1 mm.

MSCT scan acquisition techniques have been described that are designed to reduce the radiation dose of spiral scanner by decreasing the tube current during certain positions of the x-ray source relative to the patient’s body or during predetermined portions of the R-R interval of the ECG.^{25–27} The first approach takes advantage of the fact that, because of the elliptic cross-section of the human body, the attenuation of the x-ray beam is less if it traverses the body from front to back (or vice versa) than when it traverses the body from a lateral aspect. Therefore, to generate the same degree of image noise in all projections, fewer photons need to be produced when the x-ray source is anterior or posterior to the chest than when the x-ray source is lateral to the chest.^{25,26} The second approach presumes that a high photon density resulting in high-quality images with low noise is required only during ventricular diastole.²⁷ The reasoning is that image reconstruction is usually performed during ventricular diastole because the likelihood of cardiac motion artifacts is lower during that period.

TABLE 3. Comparison of Radiation Parameters for Computed Tomography of the Heart for Quantification of Coronary Artery Calcium and for Coronary Angiography With Different Scanners

Acquisition Mode	EBCT		MSCT 1			MSCT 2		
	CAC Scoring Triggered	CTCA Triggered	CAC Scoring			CAC Scoring		
			Triggered	Gated	CTCA Gated	Triggered	Gated	CTCA Gated
kVp	130	130	120	120	120	120	120	120
mA*	630	630	140	100	300	150	150	300
Exposure time, s	0.1	0.1	0.36	0.5	0.5	0.33	0.5	0.5
mAs*	63	63	50	50	150	50	75	150
Collimation, <i>N</i> × <i>T</i>	1 × 3	1 × 3	4 × 2.5	4 × 2.5	4 × 1	4 × 2.5	4 × 2.5	4 × 1.25
Table feed, mm	3	2	10	3.75	1.5	10	3.75	1.9
Reconstructed scan width, mm	3	3	2.5	3	1.25	2.5	2.5	1.25
Pitch	1	0.66	1	0.375	0.375	1	0.375	0.375
CTDI _w , mGy	3.5	3.5	4.7	4.7	17.1	4.7	7.6	20.7
CTDI _{vol} , mGy	3.5	5.3	4.7	12.5	45.6	4.7	20.3	55.2
DLP, mGy × cm	42	63.6	56.4	150	547	56.2	243	662
Effective dose, mSv†	0.7	1.1	1.0	2.6	9.3	1.0	4.1	11.3

CAC indicates coronary artery calcium; CTCA, computed tomographic coronary angiography; CTDI_w, weighted computed tomography dose index; CTDI_{vol}, volume computed tomography dose index; DLP, dose-length product (for a scan length of 12 cm); Gated, image acquisition with retrospective ECG gating (see text); MSCT 1 and 2, multislice computed tomographic scanner 1 and 2 (from 2 different vendors); *N*, No. of scans; *T*, nominal scan width in mm; and triggered, image acquisition with prospective ECG triggering.

*mA and mAs can be increased in MSCT scanners for larger patients to avoid an increase in image noise. The mAs values for MSCT 1 and MSCT 2 were provided by the respective manufacturers and do not necessarily produce an identical level of image noise.

†Effective dose estimate, with *k* = 0.017 mSv × mGy⁻¹ × cm⁻¹. This value is averaged between male and female models (see text and reference 24).

Summary

Radiation dose estimates for CT examinations of the heart are best expressed as CTDI_{vol} (in Gy), DLP (in mGy × cm), and effective dose E (in mSv). These parameters are precisely defined and allow comparisons of the radiation doses of various CT imaging protocols. Currently, there are no federal or state regulations for acceptable radiation doses for specific CT examinations in the United States. Physicians referring for or performing cardiac CT examinations should understand the absorbed radiation doses associated with various protocols. This understanding may aid in making decisions as to whether a cardiac CT examination is indicated and which protocol addresses the clinical question at hand with the least radiation exposure. It is also important that reports of CT dosimetry in the growing body of literature on cardiac CT imaging be carefully worded and precisely defined, so that the absorbed radiation doses and potential risks of different protocols can be compared.

The calculation and comparison of radiation doses received from specific CT imaging protocols are complicated by new scan acquisition techniques designed to decrease radiation dose to the patient. Further studies will be needed to investigate the exact effects of such modifications on the parameters currently used to express radiation dose.

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KEY WORDS: arteriosclerosis ■ calcium ■ imaging ■ radiation dose ■ tomography