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 and, less frequently, for minimally invasive coronary angiography. 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. 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. As a consequence, cardiac CT diagnostic images of the heart under certain conditions have become widely available. As a consequence, cardiac CT imaging, most often performed for the purpose of calcium scoring, 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.

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.

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. 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. 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 milliamperes 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.

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. 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. 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...
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). 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. 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. 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, 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.”

### CTDI vol

The CTDI vol 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. 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 the coulomb/kg (C/kg). The conventional unit is the roentgen (R). The CTDI vol 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 CTDI vol 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
a cross section of a patient’s body. The CTDI<sub>el</sub> is calculated using Equation 1:

\[
\text{CTDIEL} = \left[ \frac{2}{3} \text{CTDI}_{100} \text{(Periphery)} + \frac{1}{3} \text{CTDI}_{100} \text{(Center)} \right] \times f
\]

The term \( f \) reflects the difference between the absorption of radiation in air and the absorption in another medium. It is used to convert radiation exposure, expressed in C/kg, into absorbed dose, expressed in Gy (the SI unit for CTDI<sub>el</sub>). For the calculation of CTDI<sub>el</sub>, the appropriate value for \( f \) is 33.7 Gy C<sup>-1</sup> kg<sup>-1</sup> (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 measurement17,22 and is rarely performed. Typically, the MSAD is 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:

\[
\text{MSAD} = \frac{N \times T}{T} \times \text{CTDI}
\]

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, \( N \times T \) 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

\[
\text{MSAD} = \frac{1}{\text{Pitch}} \times \text{CTDI}
\]
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\textsubscript{vol}

The volume CTDI (CTDI\textsubscript{vol}) is a new radiation dose parameter agreed on by the International Electrotechnical Commission.\textsuperscript{16} It is based on the same concept as the MSAD, but unlike the MSAD, it is derived from the CTDI\textsubscript{w}. This is advantageous because the CTDI\textsubscript{vol} is calculated from data that are easily measured (CTDI\textsubscript{100}). The CTDI\textsubscript{vol} averages radiation dose over x, y, and z directions (CTDI\textsubscript{w} represents the average exposure in the x-y plane only). The CTDI\textsubscript{vol} for single-slice scanners is defined as:

\begin{equation}
\text{CTDI}_{\text{vol}} = \frac{N \times T}{I} \times \text{CTDI}_w
\end{equation}

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

\begin{equation}
\text{CTDI}_{\text{vol}} = \frac{1}{\text{Pitch}} \times \text{CTDI}_w
\end{equation}

The CTDI\textsubscript{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\textsubscript{vol}.

Most current CT scanners can display the value for the CTDI\textsubscript{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\textsubscript{w}, the CTDI\textsubscript{vol} is expressed in SI units of Gy.

However, both the CTDI\textsubscript{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)\textsuperscript{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:

\begin{equation}
\text{DLP} = \text{CTDI}_{\text{vol}} \times \text{Scan Length}
\end{equation}

Therefore, DLP increases with an increase in total scan length or with variables that affect the CTDI\textsubscript{w} (eg, tube voltage or tube current) or the CTDI\textsubscript{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\textsubscript{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.\textsuperscript{15,23} A technique called Monte Carlo simulation\textsuperscript{24} 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.\textsuperscript{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.\textsuperscript{23}

A reasonable approximation of the effective dose can be obtained using the equation\textsuperscript{21}

\begin{equation}
E = k \times \text{DLP}
\end{equation}
TABLE 2. Effective Dose of Selected Radiological Examinations

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

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). Typical effective dose values for frequently performed radiological examinations are listed in Table 2.

Radiation Dose in CT Examinations of the Heart

The CTDIvol 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-scanning 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. 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. The second approach presumes that a high photon density resulting in high-quality images with low noise is required only during ventricular diastole. 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

<table>
<thead>
<tr>
<th>Acquisition Mode</th>
<th>CAC Scoring</th>
<th>CTCA Gated</th>
<th>CTCA Gated</th>
<th>CTCA Gated</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>130</td>
<td>120</td>
<td>120</td>
<td>120</td>
</tr>
<tr>
<td>mAs</td>
<td>630</td>
<td>140</td>
<td>0.36</td>
<td>0.33</td>
</tr>
<tr>
<td>Exposure time, s</td>
<td>0.1</td>
<td>0.36</td>
<td>0.5</td>
<td>0.5</td>
</tr>
<tr>
<td>mAs*</td>
<td>63</td>
<td>50</td>
<td>150</td>
<td>50</td>
</tr>
<tr>
<td>Collimation, N × T</td>
<td>1×3</td>
<td>4×2.5</td>
<td>4×2.5</td>
<td>4×2.5</td>
</tr>
<tr>
<td>Table feed, mm</td>
<td>3</td>
<td>10</td>
<td>3.75</td>
<td>10</td>
</tr>
<tr>
<td>Reconstructed scan width, mm</td>
<td>3</td>
<td>2.5</td>
<td>3</td>
<td>2.5</td>
</tr>
<tr>
<td>Pitch</td>
<td>1</td>
<td>1</td>
<td>0.375</td>
<td>1</td>
</tr>
<tr>
<td>CTDIvol, mGy</td>
<td>3.5</td>
<td>4.7</td>
<td>17.1</td>
<td>4.7</td>
</tr>
<tr>
<td>CTDIvol, mGy</td>
<td>3.5</td>
<td>4.7</td>
<td>45.6</td>
<td>4.7</td>
</tr>
<tr>
<td>DLP, mGy × cm</td>
<td>42</td>
<td>56.4</td>
<td>547</td>
<td>56.2</td>
</tr>
<tr>
<td>Effective dose, mSv†</td>
<td>0.7</td>
<td>1.0</td>
<td>2.6</td>
<td>1.0</td>
</tr>
</tbody>
</table>

CAC indicates coronary artery calcium; CTCA, computed tomographic coronary angiography; CTDIvol, weighted computed tomography dose index; CTDIvol, 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 CTDIvol (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.

References

Key Words: Arteriosclerosis ■ Calcium ■ Imaging ■ Radiation Dose ■ Tomography
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