

# CT Dosimetry: Comparison of Measurement Techniques and Devices<sup>1</sup>

## TEACHING POINTS

See last page

*John A. Bauhs, PhD • Thomas J. Vrieze, RT(R) • Andrew N. Primak, PhD • Michael R. Bruesewitz, RT(R) • Cynthia H. McCollough, PhD*

In x-ray computed tomography (CT), the most common parameter used to estimate and minimize patient dose is the CT dose index (CTDI). The CTDI is a volume-averaged measure that is used in situations where the table is incremented in conjunction with the tube rotation. Variants of the CTDI correct for averaging across the field of view and for adjacent beam overlaps or gaps. CTDI is usually measured with a pencil-shaped ionization chamber, although methods have been developed that use alternative detectors, including an optically stimulated luminescence probe and a solid-state real-time dosimeter. Because the CTDI represents an averaged dose to a homogeneous cylindrical phantom, the measurements are only an approximation of the patient dose. Furthermore, dose from interventional or perfusion CT, in which the table remains stationary between multiple scans, is best evaluated with point dose measurements made with small detectors. CTDI and point dose values are nearly the same for measurement of surface dose from spiral CT. However, for measurement of surface dose from perfusion CT, the dose is overestimated by a factor of two or more with CTDI values in comparison with point dose values. Both CTDI and point dose measurement are valuable for evaluating CT scanner output and estimating patient dose.

©RSNA, 2008

**Abbreviations:** CTDI = CT dose index, FDA = Food and Drug Administration, IEC = International Electrotechnical Commission, MSAD = multiple scan average dose, OSL = optically stimulated luminescence, TLD = thermoluminescent dosimeter

**RadioGraphics 2008; 28:245-253 • Published online 10.1148/rg.281075024 • Content Codes:** **CT** **PH**

<sup>1</sup>From the Department of Radiology, Mayo Clinic, 200 First St SW, Rochester, MN 55905. Presented as an education exhibit at the 2006 RSNA Annual Meeting. Received February 16, 2007; revision requested April 11 and received June 7; accepted June 11. J.A.B. receives research support from RTI Electronics; A.N.P. receives research support from Siemens; C.H.M. receives research support from Siemens and RTI Electronics; all other authors have no financial relationships to disclose. **Address correspondence to** C.H.M. (e-mail: [mccollough.cynthia@mayo.edu](mailto:mccollough.cynthia@mayo.edu)).

©RSNA, 2008

## Introduction

Radiation dose in x-ray computed tomography (CT) is typically measured by using a simple cylindrical phantom and expressed as a volume-averaged CT dose index (CTDI). Understanding the methods for quantifying radiation dose is important for users of this technology. As compared to screen-film radiography, CT delivers considerably more dose to the patient (Table 1). When a screen-film radiograph is exposed to too much radiation, the film is overexposed and gives a visual indication of excessive patient dose. Conversely, CT images never look overexposed, as CT provides an image of normalized tissue attenuation values. Also, image noise is reduced if radiation is increased; consequently, a CT image looks fine (often better) if excessive radiation is used. **Thus, without dose measurements, the CT user lacks the visual cues needed to appropriately adjust the tube current–time product or tube potential in order to avoid excessive patient dose.** (In computed and digital radiography, the situation is similar to CT; overexposure will reduce image noise and can happen without the operator being aware, as radiologists do not often complain about image noise being too low.)

In general, clinical CT examinations involve exposures from multiple rotations of the x-ray source, such that the dose to the irradiated volume is the accumulated dose from the adjacent scans. One term used to describe this accumulated dose is the multiple scan average dose (MSAD), which as its name implies, is the dose from a multiple scan examination, averaged over one scan interval in the central portion of the multiple-scan dose profile. The MSAD was measured by using film or thermoluminescent dosimeters (TLDs), either of which required substantial time to make the measurement due to the tedious nature of calibration, handling, and reading. Measurement of the MSAD involved making multiple CT scans of a cylindrical phantom as it was stepped through the x-ray scan plane.

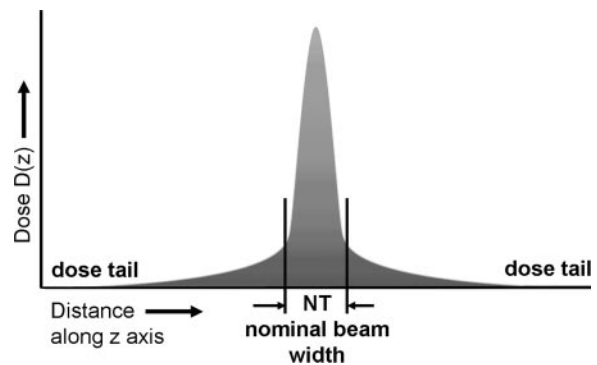
In 1981, Shope et al (2) introduced the concept of CTDI, defining it as the integral of the single scan radiation dose profile along the z axis, normalized to the thickness of the imaged section (“slice thickness”). They showed that with corrections for scan spacing, the CTDI can estimate MSAD in a standardized, convenient manner. Thus, the more easily measured ionization chamber–based CTDI became the predominant method of measuring radiation dose from a CT system.

To measure the CTDI, the radiation dose from a single CT scan is collected by a 100-mm-long ionization chamber, and then the integrated dose is normalized to the nominal beam width.

**Table 1**  
Typical Effective Dose Values for Both CT and Non-CT Imaging Examinations

| Imaging Examination                  | Typical Effective Dose Values (mSv) |
|--------------------------------------|-------------------------------------|
| Non-CT studies                       |                                     |
| Hand radiography                     | <0.1                                |
| Dental bitewing radiography          | <0.1                                |
| Chest radiography                    | 0.1–0.2                             |
| Mammography                          | 0.3–0.6                             |
| Lumbar spine radiography             | 0.5–1.5                             |
| Barium enema examination             | 3–6                                 |
| Coronary angiography (diagnostic)    | 5–10                                |
| Sestamibi myocardial perfusion study | 13–16                               |
| Thallium myocardial perfusion study  | 35–40                               |
| CT studies                           |                                     |
| Head CT                              | 1–2                                 |
| Chest CT                             | 5–7                                 |
| Abdominal CT                         | 5–7                                 |
| Pelvic CT                            | 3–4                                 |
| Abdominal and pelvic CT              | 8–12                                |
| Coronary artery calcium CT study     | 1–3                                 |
| Coronary CT angiography              | 5–14                                |

Note.—These values can be compared with the average U.S. annual effective dose equivalent from all sources of  $\approx 3.6$  mSv (1).



**Figure 1.** The radiation dose profile along a line perpendicular to the plane of a single axial CT scan shows a peak where the primary beam slices through the CTDI phantom. The tails of the dose profile are caused by scattered radiation. The integral of the area under the curve is normalized to the nominal beam width  $NT$  to determine the CTDI. A  $CTDI_{100}$  value is obtained if integration limits of  $\pm 50$  mm are used.

Hence, CTDI is not a dose to a point, but rather an average dose over a volume. Although CTDI is measured by using a single scan, it can be used to estimate the average dose from multiple scans where the table is incremented between successive tube rotations (ie, the MSAD). In contrast,

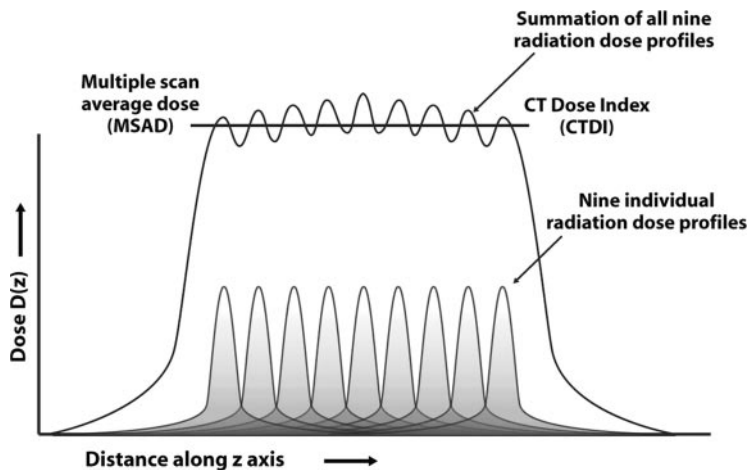
Teaching Point

Teaching Point

**Table 2**  
**Mathematical Expressions and Descriptions for MSAD and CTDI Variants**

| Dose Index                  | Mathematical Expression   | Description  |
|-----------------------------|---|--|
| Multiple scan average dose  | $MSAD = \frac{1}{I} \int_{-I/2}^{+I/2} D_{N,i}(z) dz$                       | Measured with multiple TLDs and scans; represents the average dose over one scan interval ( <i>I</i> ) in the central portion of a multiple scan ( <i>N</i> ) dose profile where the first and last scans contribute negligible dose to the central region |
| CT dose index               | $CTDI_{\infty} = \frac{1}{T} \int_{-\infty}^{+\infty} D(z) dz$              | Equivalent to MSAD if all of the scatter tails are included and the scan interval equals the nominal section thickness ( <i>I</i> = <i>T</i> ); measured with one scan by using multiple TLDs or one ionization chamber                                    |
| CTDI as defined by the FDA* | $CTDI_{FDA} = \frac{1}{NT} \int_{-7T}^{+7T} D(z) dz$                        | Measured with TLDs to obtain correct integration limits  |
| CTDI measured over 100 mm   | $CTDI_{100} = \frac{1}{NT} \int_{-50mm}^{+50mm} D(z) dz$                    | Measured with a pencil ionization chamber  |
| Weighted CTDI               | $CTDI_w = \frac{1}{3} CTDI_{100,center} + \frac{2}{3} CTDI_{100,periphery}$ | Uses CTDI <sub>100</sub> measurements  |
| Volume CTDI                 | $CTDI_{vol} = \frac{NT}{I} CTDI_w = \frac{CTDI_w}{pitch}$                   | Uses CTDI <sub>w</sub> measurements  |

\*FDA = U.S. Food and Drug Administration.



**Figure 2.** The radiation dose profiles from nine adjacent axial CT scans along a line perpendicular to the axial scans, when summed, produce the MSAD profile. The value of MSAD is averaged over one scan interval in the central portion of the profile. In contrast to the CTDI, which can be used to estimate MSAD, the MSAD is measured with many TLDs and multiple CT scans.

for scans such as perfusion CT where the table is not incremented during the examination, the CTDI is a poor descriptor of dose, and a point dose measurement is more appropriate.

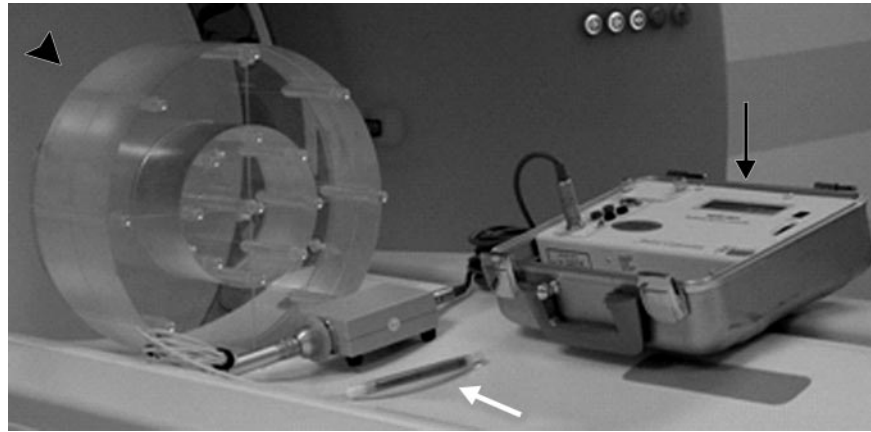
We review basic principles of CTDI, differentiate CTDI from point dose measurements, and describe measurement techniques for both. Specific topics discussed are CTDI concepts, alternative methods of measuring CTDI, limitations of CTDI, and point dose concepts.

### CTDI Concepts

Radiation dose profile from a single axial CT scan extends beyond the limits of the collimated scan

width, due to penumbral and scatter effects (Fig 1). These dose profile regions outside the nominal scan width are known as the tails of the dose distribution. The tails play a significant role in total dose accumulated from a series of scans along the longitudinal axis (Fig 2). CTDI represents the integral under the radiation dose profile from a single axial scan, including the dose tails, divided by the nominal beam width *NT*.

Definitions for various CT dose metrics are given in Table 2, where *D(z)* equals the radiation



**Figure 3.** Equipment typically used to measure  $CTDI_{100}$  includes an integrating electrometer (black arrow), a 100-mm-long CTDI ionization chamber (white arrow), and a CTDI phantom made of polymethylmethacrylate (arrowhead). The phantom is placed with its long axis perpendicular to the plane of the axial CT scan, and the ionization chamber is placed in one of the holes through the phantom. The  $CTDI_{100}$  is obtained by integrating the dose from a single axial scan and dividing by the nominal beam width.

dose profile along the  $z$  axis,  $N$  equals the number of tomographic sections imaged simultaneously in a single rotation of the x-ray tube, and  $T$  equals the width of one tomographic section along the longitudinal  $z$  axis. In multi-detector row CT, several detector elements may be grouped together to form one tomographic section. In single-detector row CT, where  $N$  equals 1, the  $z$ -axis collimation ( $T$ ) is the nominal scan width.

The equivalence of the MSAD and the CTDI requires that all contributions from the tails of the radiation dose profile be included in the CTDI dose measurement and that the scan interval equal the total nominal section width. The integration limits required to attain this equivalence depend on the nominal beam width  $NT$  and the length of the scattering medium. To standardize CTDI measurements, the FDA introduced the integration limits of  $\pm 7T$ . The CTDI measurement made by using these limits is referred to as the  $CTDI_{FDA}$ . To obtain a measurement over these exact limits, a linear array of TLDs was typically used. The FDA also specified two standard phantoms for the scattering media for CTDI measurements. The two phantoms consist of polymethylmethacrylate cylinders, both of 14 cm length: one for head examinations with a diameter of 16 cm and a second for body examinations with a diameter of 32 cm. These phantoms are referred to as, respectively, the head and body CTDI phantoms.

However, the  $CTDI_{FDA}$  is measured over a length that is dependent on the nominal beam

width. In order to standardize the measurement length, the  $CTDI_{100}$  was defined. The  $CTDI_{100}$ , like the  $CTDI_{FDA}$ , requires integration of the radiation dose profile from a single axial scan over specific integration limits. In the case of  $CTDI_{100}$ , the integration limits are  $\pm 50$  mm centered over the dose profile, which corresponds to the 100-mm length of the commercially available pencil ionization chamber.  $CTDI_{100}$  is acquired by using a 100-mm-long, 3 cm<sup>3</sup> active volume CT pencil ionization chamber and the two standard CTDI acrylic phantoms (Fig 3).

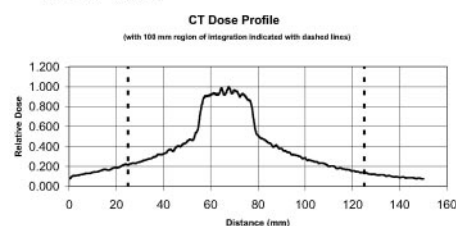
The CTDI may vary across the field of view. For the body phantom, the CTDI is typically a factor of two higher at the surface than at the phantom center. The weighted CTDI ( $CTDI_w$ ) was defined to provide an estimate of the CTDI averaged across the field of view. The  $CTDI_w$  is defined as the sum of one-third the  $CTDI_{100}$  measured at the center of the phantom and two-thirds the  $CTDI_{100}$  measured at the periphery of the phantom. According to standard 60601-2-44 of the International Electrotechnical Commission (IEC), an international CT safety standard,  $CTDI_w$  must use the  $CTDI_{100}$  and is expressed in terms of dose to air (3). (Compliance with these standards is not legally required in the United States, but they have nevertheless become a de facto standard because they are required by law in European Union countries.) To convert from exposure to dose in air, an  $f$  factor of 0.87 rad/R is used. In SI units, where air kerma is measured, the conversion from air kerma to dose in air (both have the units of milligray) is 1.0 mGy/mGy.





a.

| Customer Information      |                  | Landauer Scanner Information |             |
|---------------------------|------------------|------------------------------|-------------|
| Customer Facility:        | Mayo Clinic      | Process Number:              | CT181196A   |
| Account Number:           | 2182             | Serial Number:               | A0802370    |
| Customer Contact:         | Glenn M Storchio | Date Received:               | 8/11/06     |
| Date Exposed:             |                  | Date Scanned:                |             |
| kVp:                      | 120              | Diagnose Scanner:            | 1           |
| mAs:                      | 100              | Slit Width (mm):             | 0.2         |
| Phantom:                  | Head             | Sample Increment (mm):       | 0.65        |
| Position in Phantom:      | is               | Smoothing Type:              | 20 pt ave   |
| Nominal Slice Width (mm): | 13.2             | Customer Comments:           | small focus |
| Number of Slices:         | 1                |                              |             |
| Table Increment (mm):     | 0                |                              |             |
| Total Table Travel (mm):  | 0                |                              |             |
| Total Scan Length (mm):   | 13.2             |                              |             |
| Scan Mode:                | Axial            |                              |             |
| Scanner Manufacturer:     | Siemens          |                              |             |
| Scanner Model:            | Sensation 64     |                              |             |



|                               | Nominal | Observed |
|-------------------------------|---------|----------|
| Total Integrated dose (mrad): | 44853   |          |
| Slice Thickness (mm):         | 19.2    | 25.1     |
| Number of Slices:             | 1       | 1        |

|  |       |       |
|--|-------|-------|
| CTDI <sub>100</sub> = (44853) / (19.2 * 1) = | mrad  | mGy   |
|  | 2,336 | 23.36 |

b.

**Figure 4.** (a) An OSL dosimeter (arrow) is inserted into a head CTDI phantom to measure the CTDI<sub>100</sub>. The probe is exposed to a single axial scan, then returned to the manufacturer for reading with a precision scanning laser. (b) The resultant radiation dose profile information is returned to the user in both print and digital formats.

All CTDI measurements made with a pencil ionization chamber are performed with a stationary patient table. To represent dose for a specific scan protocol, it is essential to take into account any gaps or overlaps between the radiation dose profiles from consecutive rotations of the x-ray source. For this purpose, the volume CTDI<sub>w</sub> (CTDI<sub>vol</sub>) was defined. In spiral CT, the ratio of the table travel distance per rotation (*I*) to the nominal beam width (*NT*) is referred to as pitch (pitch = *I*/*NT*). The CTDI<sub>vol</sub> is defined as the CTDI<sub>w</sub> divided by pitch. CTDI<sub>vol</sub> is conceptually similar to the MSAD but is standardized with respect to the integration limits (±50 mm) and the factor used to convert the exposure or air kerma measurement to dose in air. CTDI<sub>vol</sub> represents the dose within the scan volume from a particular scan protocol for a standardized phantom. It is a measure of scanner output and not

patient dose. IEC standard 60601-2-44 requires that the CTDI<sub>vol</sub> be displayed on the console of modern CT scanners before the scan is initiated (3).

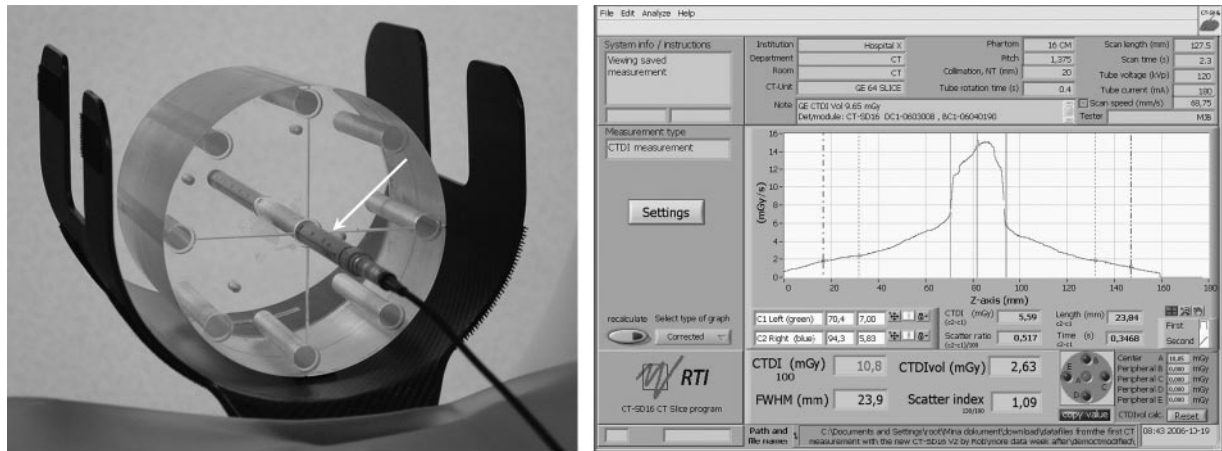
### Alternative Methods of Measuring CTDI

Although CTDI is normally measured by integrating the dose from a single CT scan using a 100-mm-long ionization chamber, it can also be obtained by making many point dose measurements along the z axis to determine the dose profile *D*(*z*) and then integrating under the dose profile curve to obtain the integrated dose. Whether obtained with ionization chamber or with point dose measurements, the integrated dose is divided by the nominal beam width to obtain the CTDI. The linear array of point dose measurements can be made by using TLDs, film, metal oxide semiconductor field-effect transistors, optically stimulated luminescence (OSL), or solid-state (diode) detectors. The limits of integration will determine how much dose from the scatter tails is included. The linear density of the measured data points along the z axis will determine the accuracy of the CTDI measurement. Two systems for making point dose measurements are described herein.

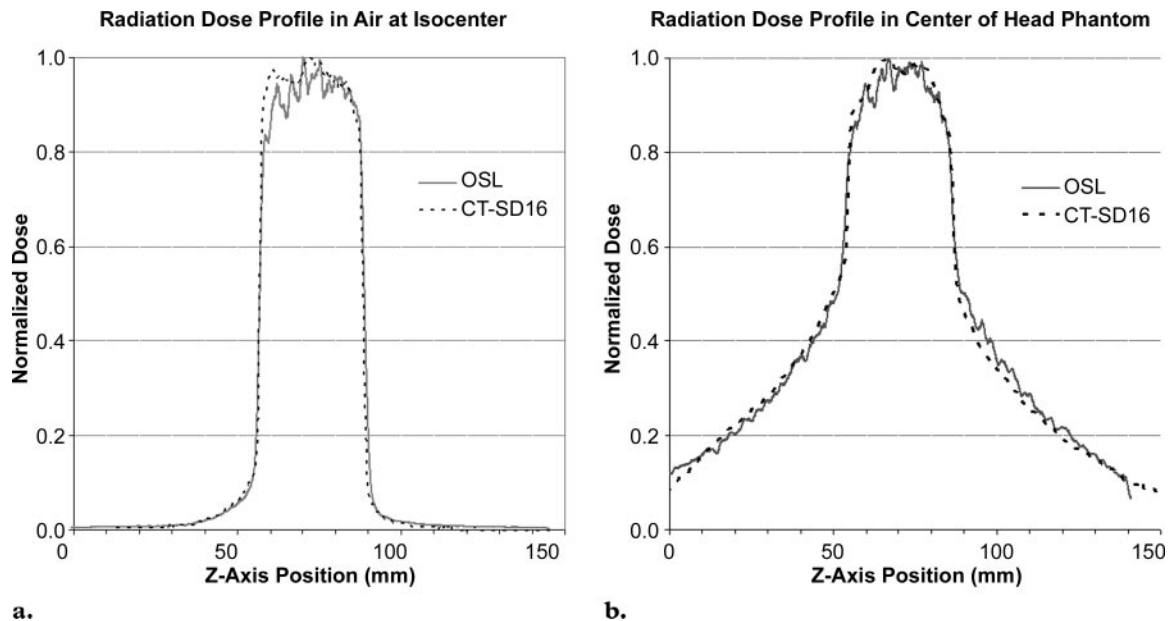
An OSL dosimeter can be used to measure the dose profile and determine the CTDI<sub>100</sub>. For example, Landauer (Glenwood, Ill) makes an OSL CTDI probe. The probe consists of an OSL-coated film substrate placed within a light-tight plastic cylinder. The probe can be inserted into a CTDI phantom, similar to the CT pencil ionization chamber. A single exposure is made with the patient table in a stationary position. After exposure, the probe is returned to the manufacturer for reading with a precision scanning laser. The result is a digital radiation dose profile that can be integrated to determine the CTDI<sub>100</sub>. A CTDI phantom and OSL dosimeter are shown in Figure 4a with the dosimetry results shown in Figure 4b.

A solid-state, real-time dosimeter can also be used to generate the dose profile and measure the CTDI. RTI Electronics (Gothenburg, Sweden) makes a solid-state detector (model CT-SD16) that can be used for this application. The probe is shaped like a CT pencil ionization chamber and consists of a thin-walled aluminum tube filled with plastic and containing two submillimeter-thick solid-state detectors located 16 cm apart. One detector is near the tip of the probe, and the second is near the middle. The probe fits in the CTDI phantom in the same holes used for the CT pencil ionization chamber. It is inserted

Teaching Point



**Figure 5.** (a) A solid-state real-time dosimeter (arrow) is inserted into a head CTDI phantom to measure the  $CTDI_{100}$ . The probe and phantom are passed through the x-ray beam by using the motion of the patient table in a spiral acquisition mode; a real-time electrometer records the dose delivered to a submillimeter-thick dosimeter in the center of the probe. (b) The CT dose data are collected by a computer and used to generate a dose profile, which is integrated and normalized to the nominal beam width, thus giving the  $CTDI_{100}$ . If central axis and peripheral dose data are collected from consecutive scans, the software will automatically calculate  $CTDI_{vol}$ .

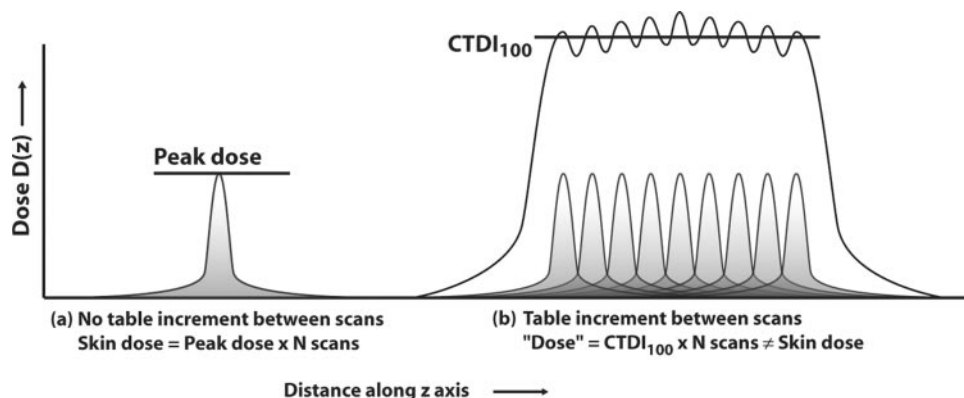


**Figure 6.** There is good agreement between dose profiles generated by the OSL dosimeter shown in Figure 4 (solid line) and the solid-state detector shown in Figure 5 (dashed line). The graphs show radiation dose profiles for probes placed in air at the isocenter of the CT scanner (a) and probes placed in the center of a CTDI head phantom (b).

through the phantom so that the tip detector is in air outside the phantom and the second detector is centered within the phantom. This first detector is used to trigger a pulsed sequence of dose measurements made by the second detector.

The solid-state dosimeter is coupled to a high-sampling-rate electrometer, which records the

dose measured in very small (millisecond to microsecond) time intervals. These essentially real-time dose data are recorded as a function of time from the start of the measurement. Instead of making measurements by using a stationary phantom and a CT exposure for only one tube rotation, the probe and phantom are passed through the x-ray beam by using the motion of the patient table in the spiral scan mode. A computer collects



**Figure 7.** Left: In perfusion or interventional CT (where there is no table movement), the peak skin dose is the relevant dose parameter for deterministic skin effects. The peak skin dose equals the peak dose from one scan times the number of scans. Right: If the peripheral  $CTDI_{100}$  is used as a surrogate for peak dose (as prescribed by IEC standard 60601-2-44), the skin dose will be overestimated by up to a factor of two. Because the definition of CTDI assumes table incrementation, CTDI is not a valid parameter for scans performed without table movement.

dose data as a function of time. The table travel speed is used to convert time into z-axis position, resulting in the dose profile  $D(z)$ . By integrating the dose profile over  $\pm 50$  mm and dividing by the nominal beam width, the  $CTDI_{100}$  is calculated. The setup of a phantom and the solid-state detector probe is shown in Figure 5a with the dosimetry results shown in Figure 5b.

Figure 6 compares dose profiles measured with the OSL and solid-state dosimeters. Agreement between measurements made with the 100-mm ionization chamber, the OSL probe, and the solid-state detector probe demonstrated the equivalence of these three techniques.

### Limitations of CTDI

CTDI is a good measure of CT scanner output for applications where the table is incremented during the scan. However, it has limitations. For instance, because CTDI is measured by using a standardized, homogeneous, cylindrical phantom, it questionably represents the dose for objects of substantially different size, shape, or attenuation, like the human body. Further, CTDI is expressed as dose to air, not dose to tissue, thus leaving CTDI a step away from tissue dosimetry. Methods exist to estimate organ doses for a variety of human or humanoid volumes; however, these typically require use of Monte Carlo simulations.

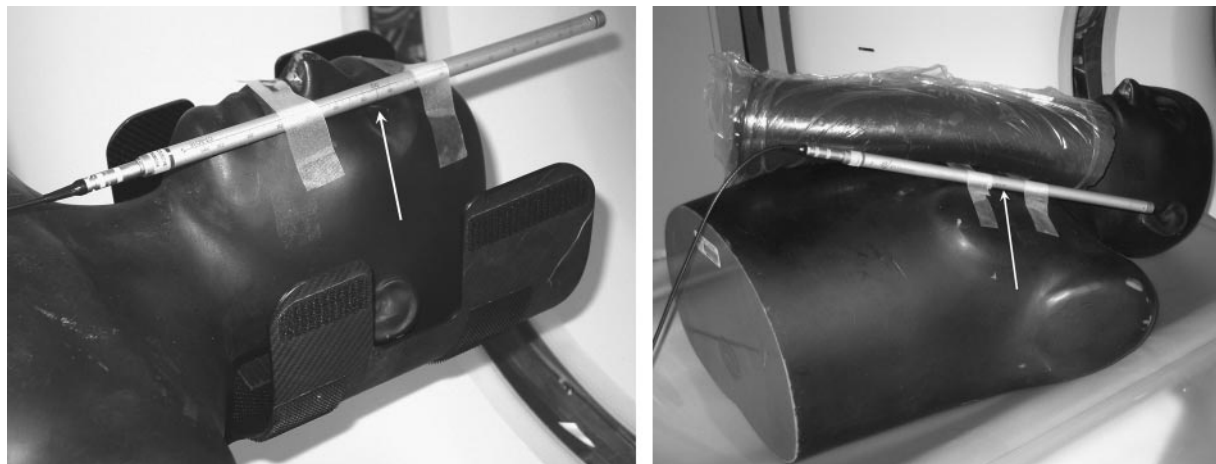
The 14-cm length of the body CTDI phantom does not provide a sufficiently long scatter path relative to the typical length of a human torso; hence, patient dose may be underestimated with CTDI (4). In addition, while the current 100-mm integration length is sufficient for measurement of the dose tails for nominal beam widths of several

centimeters, it is not sufficient for beam widths greater than 10 cm, such as the 12-cm nominal beam width for the prototype Toshiba 256-section system or C-arm-based cone-beam CT systems (5). Finally, CTDI does not indicate the dose to a specific point in the scan volume when the patient table remains stationary for multiple scans, such as for interventional or perfusion CT.

### Point Dose Concepts

To determine the dose to a point within the scan volume, a small dosimeter is required. For good spatial resolution, the dosimeter needs to be less than a few millimeters in diameter. Practical options include TLDs, OSL “dots,” metal oxide semiconductor field-effect transistors, and various types of solid-state detectors. Film is not a practical choice because its nonlinear response requires careful and time-consuming calibration steps each time the measurement is performed. Whichever dosimeter is used must be calibrated to the spectral characteristics of the CT system. Ease of use and long-term stability are important characteristics, as is the ability to place the dosimeter within the small cavities in anthropomorphic phantoms such as the Alderson Rando phantom.

Unlike CTDI, point dose measurements do not involve spatial integration or normalization to nominal beam width. For the assessment of peak skin dose from multiple exposures at the same table position, such as in perfusion CT, the contributions of the dose tails are not relevant. Figure 7 illustrates the contrast between peak dose and



a.

b.

**Figure 8.** The real-time solid-state dosimeter shown in Figure 5 was used to measure both peripheral  $CTDI_{100}$  and point dose for spiral and perfusion CT examinations of the head and thorax. The center detector of the probe (arrow) was positioned over the lens of the eye (a) and the upper thorax (b). For the chest examination, tissue-equivalent bolus material was layered over the probe. Clinical spiral and perfusion CT scans were performed at each location, and both the  $CTDI_{100}$  and point dose were measured.

**Table 3**

**Comparison of Peripheral Point Dose Data and Peripheral  $CTDI_{100}$  Data for Spiral and Perfusion CT Studies of the Head and Thorax**

| CT Study                               | Tube Potential (kVp) | Tube Current–Time Product (mAs) | Pitch | Peripheral $CTDI_{100}$ /Pitch (mGy) | Peripheral Point Dose (mGy) |
|--|----------------------|---------------------------------|-------|--------------------------------------|-----------------------------|
| Spiral head CT                         | 120                  | 340                             | 0.8   | 52.3                                 | 44.4                        |
| Brain perfusion study                  | 80                   | 250                             | ...   | 392                                  | 202                         |
| Spiral chest CT (coronary angiography) | 120                  | 170                             | 0.2   | 73.4                                 | 70.5                        |
| Lung perfusion study                   | 100                  | 170                             | ...   | 272                                  | 140                         |

Note.—Studies were performed with a Sensation 64 scanner (Siemens Medical Solutions, Malvern, Pa). Perfusion scans were located over the CT-SD16 “D2” point dosimeter (RTI Electronics) to estimate dose to the lens of the eye (head studies) or to an implantable cardiac rhythm management device (thoracic studies) (Fig 8). For overlapping scans (pitch < 1), the peripheral  $CTDI_{100}$  value must be divided by pitch, making it analogous to  $CTDI_{vol}$ .



CTDI. Figure 8 shows our experimental setup for the measurement of point dose from spiral and perfusion scans of the head and thorax. Table 3 provides a comparison of data between peripheral point dose and peripheral CTDI<sub>100</sub> for spiral and perfusion scans of the head and thorax. These data indicate that CTDI overestimates peak dose by a factor of two for the phantom and CT system studied.

### Summary

The CTDI is a simple, standardized measure of the dose output of a CT scanner that can be used to compare different scan techniques on a single scanner or between scanners. CTDI is an averaged dose to a standard phantom from a multiple scan examination where the patient table is incremented between scans. Besides the ionization chamber measurement method, equivalent methods have been developed that use alternative dosimeters, including an OSL probe and a solid-state, real-time dosimeter. Variations of CTDI have been developed to represent the CTDI averaged over the field of view (CTDI<sub>w</sub>) and corrected for overlaps or gaps between scans (CTDI<sub>vol</sub>).

Because the CTDI is an averaged dose to a homogeneous cylindrical phantom, the measurements are only an approximation of patient dose. Another limitation is that CTDI overestimates dose for scans where the patient table is not incremented, such as in interventional and perfusion CT. For these CT applications, the CTDI can overestimate peak dose by a factor of two. Therefore, point dose measurements made with dosimeters with small active volumes are necessary to determine skin doses from interventional and perfusion examinations.

**Acknowledgment:** The OSL dosimeters used in this study were provided by the manufacturer.

### References

1. National Council on Radiation Protection and Measurements. Ionizing radiation exposure of the population of the United States. Bethesda, Md: National Council on Radiation Protection and Measurements, 1987.
2. Shope TB, Gagne RM, Johnson GC. A method for describing the doses delivered by transmission x-ray computed tomography. *Med Phys* 1981;8:488–495.
3. International Electrotechnical Commission. Medical electrical equipment, part 2-44: particular requirements for the safety of x-ray equipment for computed tomography. IEC publication no. 60601-2-44. Ed 2.1. Geneva, Switzerland: International Electrotechnical Commission, 2002.
4. Dixon RL, Ekstrand KE. A film dosimetry system for use in computed tomography. *Radiology* 1978; 127:255–258.
5. Boone JM. The trouble with CTDI100. *Med Phys* 2007;34:1364–1371.

Teaching  
Point

Teaching  
Point

## From the Archives of the AFIP Pyelonephritis: Radiologic-Pathologic Review

*William D. Craig, CDR, MC, USN, et al*

RadioGraphics 2008; 28:245-253 • Published online 10.1148/rg.281075024 • Content Codes:  CT  PH

---

### Page 256

In general, routine radiologic imaging is not required for diagnosis and treatment of uncomplicated cases in adult patients.

### Page 261

When imaging is warranted, CT is the modality of choice for evaluating acute bacterial nephritis. It provides comprehensive anatomic and physiologic information that accurately characterizes both intra- and extrarenal pathologic conditions.

### Page 263

Infection is only one cause of interstitial nephritis. The inflammation can be induced by drugs, granulomatous diseases, metabolic disorders, and immunologically mediated mechanisms.

### Page 271

The most common extrapulmonary site of tuberculosis is the urinary tract, with almost all cases resulting from hematogenous seeding. Despite the presumed route of spread from the lungs to the kidney, less than 50% of patients in whom urinary tract tuberculosis is ultimately diagnosed have abnormal results from chest radiography.

### Page 275

Because malacoplakia most commonly manifests as a mucosal mass involving the bladder or ureter, the most frequent renal finding is obstruction secondary to a lesion in the lower tract (54).