CT Dose Index and Patient Dose: They Are *Not* the Same Thing[1]

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In 1981, Shope et al (1) published "A Method for Describing the Doses Delivered by Transmission X-ray Computed Tomography." In that article, they introduced the computed tomography (CT) dose index (CTDI) as a metric to quantify the radiation output from a CT examination consisting of multiple contiguous CT scans (ie, multiple adjacent transverse rotations of the x-ray tube along the patient longitudinal axis). A new dosimetric method was required for CT because the irradiation geometry was quite different from that of other x-ray modalities in use at that time; namely, the x-ray tube irradiated only a narrow section of the anatomy while it made a full rotation around the patient and did so for multiple rotations along the length of the patient. The CTDI method sought to create an "index" to reflect the average dose to a cylindrical phantom in the central region of a series of scans. The word "index" was specifically included in CTDI's name to distinguish the quantity from the radiation dose absorbed by a patient.

This method, which has been defined in detail elsewhere (2–6), was subsequently adopted by the Center for Devices in Radiological Health of the Food and Drug Administration and its definition included in the Code of Federal Regulations (6). These regulations specify the composition, diameter, and length of two polymethylmethacrylate (ie, acrylic, Lucite) cylindrical phantoms that are to be used for CTDI measurements. To quantify the scanner output for head CT examinations, a 16-cm-diameter phantom is to be used. To quantify scanner output for body examinations, a 32-cm-diameter phantom is to be used. These are referred to as the head and body CTDI phantoms, respectively. Both phantoms are 14–15 cm long (Fig 1).

The standardization of the CTDI phantoms marked a crucial step in quantifying the radiation output of a CT scanner in a consistent and reproducibly measured fashion. This is because the primary beam emitted from the scanner (originally a relatively thin fan beam, which with current technology has expanded to cone beams of up to 16 cm width along the patient longitudinal axis) produces a substantial amount of scattered radiation when it interacts with the patient. Hence, consistent radiation output measurements required consistent phantoms.

Early estimates of dose from a CT examination did not use the CTDI methodology and measured only the dose from a single scan acquisition. Specifically, only the peak radiation dose emitted by the scanner from a single tube rotation and at a single table position was measured, and this underestimated the dose delivered to a typical adult patient by a factor of two to three. The reason for this underestimation was that the measurement neglected the "tails" of the dose distribution caused by scattered radiation produced from scans at adjacent table positions (Fig 2a) (1,7). Because most clinical examinations involve multiple scans (ie, gantry rotations) as the patient is translated through the gantry, the dose distribution to the patient is the sum of the overlapped "single-scan" dose distributions (Fig 2b). For examinations with a sufficient number of scans, the average dose over the central scan width of the imaged anatomy will reach an equilibrium value, which is referred to as the multiple scan average dose (MSAD) (Fig 2b).

In the early days of CT, direct measurement of the MSAD was a labor-intensive process. It required multiple scan acquisitions, which placed heavy loads on the x-ray tube. The long scan times necessitated use of dosimeters that could integrate dose accurately over several minutes, such as film or thermoluminescent dosimeters. Conversely, the

Figure 1

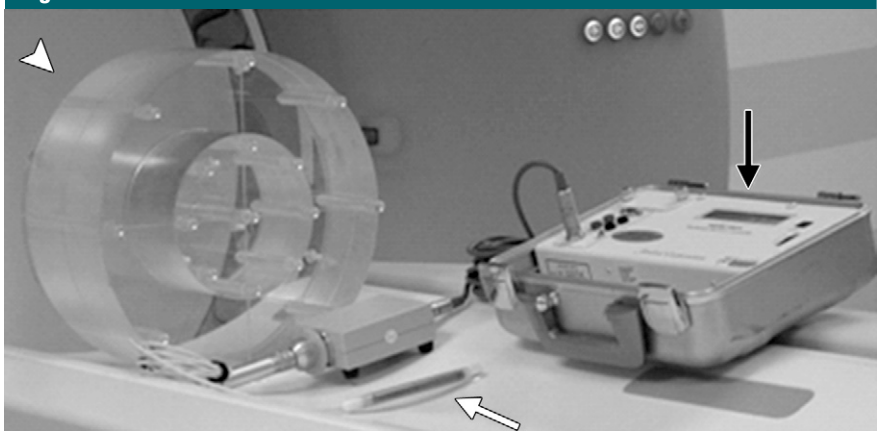

Figure 1: Equipment typically used to measure CTDI100 includes an integrating electrometer (black arrow), a 100-mm-long CTDI ionization chamber (white arrow), and a CTDI phantom made of polymethylmethacrylate (arrowhead). The phantom is placed with its long axis perpendicular to the plane of the transverse CT scan and the ion chamber placed in one of the holes through the phantom. CTDI100 is obtained by integrating the dose over 100 mm from a single transverse scan and dividing it by the nominal beam width. (Reprinted, with permission, from reference 7.)

introduction of CTDI by Shope et al (1) provided a much more practical method with which to estimate the MSAD and hence quantify the radiation output of a CT system. First, although the CTDI could be measured by using only a single rotation of the x-ray tube, it represented the dose from a series of scan acquisitions. Second, it facilitated the use of ionization chambers, making measurements faster and easier to acquire. Because the x-ray beam from a CT scanner was too narrow to completely cover the sensitive volume of existing ionization chambers, a 100-mm-long pencil ionization chamber was developed and the partial irradiation effect corrected on the basis of chamber length and nominal beam width (8).

The CTDI technique uses this long ionization chamber to integrate the primary and scattered radiation delivered with a single scan (ie, one gantry rotation) and normalizes it to the nominal beam width. This normalization cleverly incorporated a scanner's dose efficiency. That is, if the radiation dose profile from a CT system was unnecessarily wide (ie, the primary beam was wider than the imaged section width), the CTDI would be higher than that from a system with a more narrow beam that better matched the width of the imaged section (9,10). In addition, Shope and colleagues (1) demonstrated that the CTDI could be easily scaled to reflect the common situation when the radiation beams were not contiguous (ie, when there were gaps or overlaps between consecutive rotations of the x-ray tube). Thus, CTDI-based metrics became the reference standard for measuring, comparing, and communicating the radiation output of a CT system (3).

In recent years, however, the strengths and weaknesses of the CTDI have been debated (11–15). Criticisms of the CTDI are based on two primary arguments: *(a)* the 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, and *(b)* 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. This means that the average dose (eg, MSAD) that would occur in the much longer "typical-sized" adult torso is underestimated with CTDI measurements in the 14-cm-long body CTDI phantom; the underestimation owing to the use of this phantom can be as much as 40% (11,13). Another important limitation of the CTDI concept is that it is not applicable for CT exposures where the patient remains stationary throughout the scan. Whether from wide cone-beam systems that image a large volume without table increment or CT perfusion examinations, the CTDI value presented on the scanner console is an overestimate of both the average dose within the scan volume and the dose to the skin (16–19).

These criticisms, however, are based on the belief that CTDI should estimate the patient dose, as opposed to quantifying the radiation output of CT systems. In fact, because patients and the wide range of clinical applications and scan protocols used to scan them vary so dramatically, there is no single phantom that can be used to accurately estimate the dose to all patients. Any dose metric designed to estimate patient dose for a "typical" adult will underestimate the actual absorbed dose for a pediatric patient or overestimate the actual absorbed dose for an obese patient.

Instead, because the volume CTDI ($CTDI_{vol}$) (3,5,20) is displayed on the scanner console before the initiation of a scan (to allow the operator to confirm that the proper scanner output is programmed) and recorded as part of the patient's examination information, many users incorrectly assume that it is the dose to that particular patient. The CTDI values are included in either a screen-captured "patient dose report" or a structured Digital Imaging and Communications in Medicine dose report, which reinforces the incorrect belief that CTDI is a measure of patient dose. In fact, the actual dose to any given patient is directly dependent on the size and shape of the patient (19,21–24). The $CTDI_{vol}$ is a standardized measure of the radiation output of a CT system, measured in a cylindrical acrylic phantom, that enables users to gauge the amount of emitted radiation and compare the radiation output between different scan protocols or scanners. Complex calculations are required to map scanner output to patient dose, taking into account the patient's size, irradiated organs, body composition, and scan range (19,21–25).

A simple analogy is the following: The operation of a car's engine is reflected by a tachometer. It reports revolutions per

Figure 2

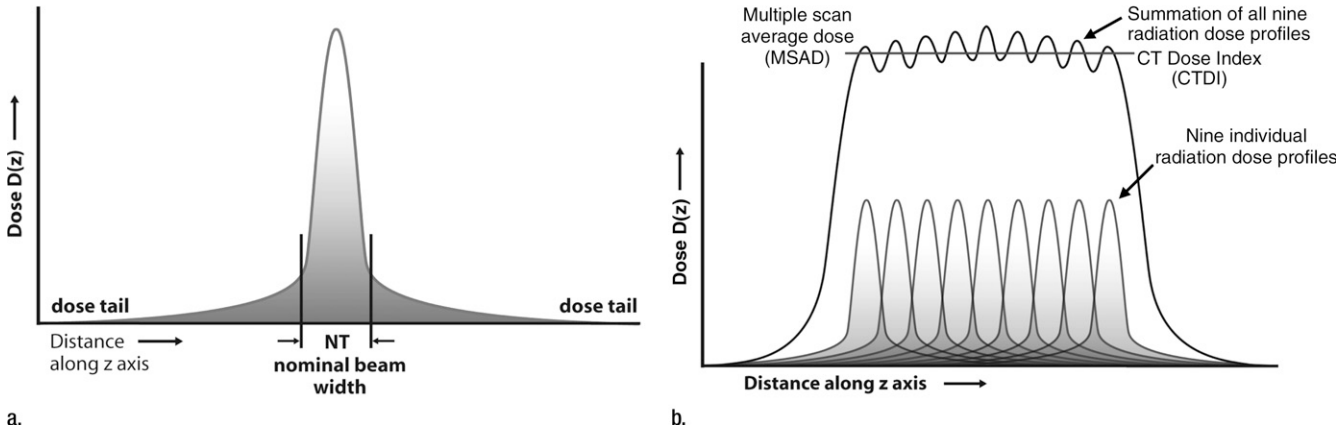

Figure 2: **(a)** Radiation dose profile along a line perpendicular to the scan plane shows a peak dose level at the center of the primary beam and long dose tails caused by scattered radiation. *NT* = nominal beam width. **(b)** The radiation dose profiles from nine adjacent transverse CT scans along a line perpendicular to the transverse scans, when summed, produce the MSAD profile. The value of MSAD is the average value of this profile over one scan interval in the central portion of the profile. (Reprinted, with permission, from reference 7.)

minute of the engine's crankshaft. Although the literal translation for the term "tachometer" from the Greek is "speed measurer," it does not, in fact, measure the speed (velocity) of a car. The speedometer, which reports the distance (miles or kilometers) that the car will travel in 1 hour at its current velocity, must be calibrated for the specific tire diameter on the vehicle. By changing the size of the tires from those for which the vehicle was calibrated, one will reduce the accuracy of the speedometer. Increasing the tire's diameter allows a greater distance to be traveled per revolution of the crankshaft, and, thus, a higher speed is achieved for a given number of revolutions per minute. Likewise, for a given CT scanner output (ie, tachometer, engine output), changing the diameter of the patient (ie, tire diameter) will change the dose absorbed by that patient (vehicle velocity).

Although the need to take patient size into account when estimating patient dose has been well established, the widespread misinterpretation of CTDI as a measure of patient dose continues.

Equipped with accurate knowledge of scanner output and estimates of patient size (eg, from the CT radiograph), scan region (eg, thorax or abdomen), and scan length, estimates of patient-size–specific dose may be determined with an accuracy of approximately 10% (19,22–27). Thus, as long as scanner output continues to be measured and reported by using a standardized, highly reproducible, and pragmatic measurement technique, such as the $CTDI_{vol}$ method, patient dose can be accurately estimated. It is imperative, however, that the community be aware that the CTDI is not patient dose.

$CTDI_{vol}$ provides a very useful way to compare the doses delivered by various scan protocols or to achieve a specific level of image quality for a specific size patient. With use of technique charts and diagnostic reference levels, $CTDI_{vol}$ can be used to prescribe the right dose for a specific patient size and diagnostic task. However, $CTDI_{vol}$ cannot be used as a surrogate for patient dose, either in epidemiologic assessments of potential late effects or for potential deterministic effects (eg, skin injury) (17,18). Neither $CTDI_{vol}$ nor its derivative, dose-length product (DLP, which is the product of $CTDI_{vol}$ and the irradiated scan length), should be used to estimate effective dose or potential cancer risk for any individual patient. The published "*k* factors" used to convert DLP to effective dose all assume a standard-sized patient (28–31). The "standard" patient used for adult *k* factors is relatively thin by today's standards (nominal body mass of 70 kg) (28–32). Similarly, the *k* factors for newborns and 1-, 5-, 10-, and 15-year-old children refer to a generic child of that age, even though the dimensions assigned to an age do not always correlate well with individual patient sizes (33). For both children and adults, the idealized patient model is a hermaphrodite; that is, it has the sexual organs of both sexes. Thus, the patient models used to estimate dose by using DLP do not represent a real patient.

Examples of the inappropriate use of $CTDI_{vol}$, DLP, and effective dose include the widely publicized reports of large variations in "doses" from CT examinations (34). The problem with such reports, however, is the lack of correction for patient size. For example, in Monte Carlo simulations of absorbed patient dose that take into account patient size, it has been shown that the effective dose increases much more slowly than does the $CTDI_{vol}$ or DLP. To achieve similar image quality, the scanner output ($CTDI_{vol}$) should be increased by about a factor of two as patient size changes from a typical adult abdomen (lateral dimension, 35–40 cm) to an obese adult abdomen (lateral width, 45–50 cm) (35–37). Even though the scanner output increases by a factor of two, the dose to many of the radiosensitive internal organs used in the calculation of effective dose does not

Figure 3

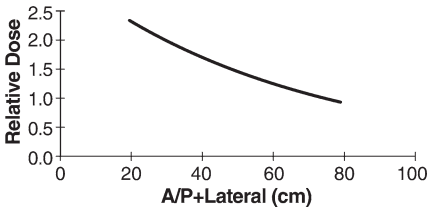

Figure 3: Graph shows relative dose (mean patient dose per 1 mGy of scanner output, $CTDI_{vol}$) for an abdominal CT scan and different patient sizes (here represented by the sum of anteroposterior [*A/P*] and lateral dimensions). Over the range of patient sizes from a newborn to a large adult, relative dose is exponentially related to patient size. For a patient with an anteroposterior dimension of 30 cm and a lateral dimension of 40 cm, the anteroposterior + lateral value would be 70 cm and the mean patient dose in the center of the scan range would be approximately equivalent to the $CTDI_{vol}$ value reported on the console. For a neonate having both anteroposterior and lateral dimensions of 10 cm, the anteroposterior + lateral value would be 20 cm and the mean patient dose in the center of an abdomen scan would be about 2.3 times the displayed CTDI value, for body $CTDI_{vol}$ measurements made by using a 32-cm phantom. $CTDI_{vol}$ measurements made on the basis of 16-cm phantoms would require a different scale factor.

increase by the same amount owing to the attenuation of the additional adipose tissue. Rather, the factor of two increase in the $CTDI_{vol}$, combined with the larger patient size, results in a net increase in effective dose of only approximately 20%–30% (23,24,38).

An important implication of the need to take patient size into account, both when estimating patient dose and when prescribing the correct scanner output settings, is that considerable variation in CTDI-based dose metrics can, and should, be expected. Facilities that adjust their CT technique appropriately for patient size, whether with use of manual technique charts or automatic exposure control (35,39–56), will prescribe a wide range of scanner output (CTDI) values. This is a good outcome, reflecting the facility's conscientiousness in "right-sizing" the dose settings on the basis of specific patient body habitus. Furthermore, variability in the image quality criteria for various diagnostic tasks and clinical applications introduces variability in the scanner output settings that one should prescribe, even for patients of the same size. For example, scanner output should vary markedly between CT colonography and CT enterography, even for the same patient. Thus, radiation management in CT requires choosing the correct settings for scanner output, not only for patient size but also for the imaging task.

In conclusion, it is imperative that measures of the radiation output of a CT system can be easily and practically measured in a consistent and robust fashion. $CTDI_{vol}$ meets these criteria. It is defined both in the United States and in international regulatory communities, and the equipment used to measure it is ubiquitously available worldwide. The $CTDI_{vol}$ tells the medical physicist precisely how the machine was operated, and it can be used, in conjunction with information regarding patient size and the scanned anatomy, to estimate patient dose (19,23–25). Dose estimates can be for organ dose, skin dose, or a mean dose in the center of the scan volume (17,19,23–25,57). The CTDI values are not, however, patient dose estimates. CTDI is the tachometer in a CT scanner—not the speedometer. Estimates of individual patient risk, and epidemiologic studies assessing potential late effects, must use patient size–specific dose estimates—they cannot use only scanner output ($CTDI_{vol}$ or DLP). Rather, use of the known exponential relationship between patient size and absorbed dose will allow patient size–specific dose estimates to be made from scanner output values (19,23,24) (Fig 3).

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