Radiation Dose at Cardiac Computed Tomography
Facts and Fiction

Walter Huda, PhD, W. Taylor Rowlett, MD, and U. Joseph Schoepf, MD

Abstract: Cardiac computed tomography (CT) dosimetry makes use of two radiation parameters: a volume CT dose index (CTDI) and a dose length product (DLP). The volume CTDI quantifies the intensity of the radiation used to perform CT examinations, whereas DLP quantifies the amount of radiation used. CTDI metrics can be converted into patient dose metrics by using dose/CTDI conversion factors. In cardiac CT imaging, these need to take into account the x-ray tube voltage, scan length, and scan region, as well as patient size. Organ doses to patients in cardiac CT can be converted into cancer risks when patient demographic factors are taken into account. A risk analysis of patients undergoing cardiac CT angiography at our institution showed that a majority (62%) were males, with a median age of approximately 60 years and a median weight of approximately 90 kg. The median DLP was approximately 1100 mGy cm, corresponding to an effective dose of approximately 29 mSv in normal-sized patients. The average patient lifetime risk for a radiation-induced cancer was estimated to be 0.12%, with 85% of it attributed to lung cancer. Patients with an age and weight at the 10th percentile, who also received a DLP at the 90th percentile, would have cancer risk estimates approximately double the average value. Radiation risks are required to determine whether examinations are indicated, defined as examinations in which individual patient benefit exceeds corresponding patient risk. Understanding radiation risks in cardiac CT encourages operators to use the least amount of radiation to achieve satisfactory diagnostic performance.

Key Words: CT dosimetry, computed tomography dose index, dose length product, effective dose, radiation risk

(*J Thorac Imaging* 2010;25:204–212)

In 2006, the average radiation dose from diagnostic medical imaging in the United States was estimated to be approximately 3 mSv per year, an increase of approximately 600% from the corresponding medical dose in the 1980s.[1] The average US inhabitant also receives approximately 1 mSv per year from ubiquitous background radiation (ie, terrestrial radioactivity, cosmic radiation, and primordial radionuclides) and approximately 2 mSv per year from domestic radon exposure. About half of the medical radiation dose is a result of the use of computed tomography (CT) imaging, whose use has grown substantially with nearly 70 million examinations reported in 2006.[1] Increased CT utilization has occurred because of dramatic improvements in technical performance and expanded clinical applications, including cardiac CT.[2] Major technological advances of CT over the last decade include the introduction of multidetector CT, dual-source CT scanners, and increased x-ray tube rotation speeds.[3,4] With dual-source CT systems, it is now possible to scan a complete adult chest with electrocardiogram synchronization in less than 1 second (> 40 cm/s) and achieve a temporal resolution of 75 ms for an individual CT image.[5] It is likely that technical advances in CT will continue to occur and that clinical applications will expand in the foreseeable future. There is little doubt that most patients benefit from the diagnostic information obtained from the clinical use of this imaging modality.[5,6]

Marked improvements in diagnostic imaging performance of CT have been accompanied by increased concern regarding higher radiation doses and corresponding patient risks.[7–9] Radiation doses in CT are markedly higher than in conventional radiography. A chest CT, for example, has an effective dose (E) of approximately 5 mSv, which is equivalent to approximately 100 chest radiographic examinations because the latter have effective doses of approximately 0.05 mSv.[10] Radiation dose quantities in CT are currently expressed in CT dose index (CTDI).[11] Converting CTDI dose indices into meaningful dose metrics (ie, organ and effective dose) requires explicit consideration of both technical and patient factors.[12] Converting organ doses into radiation risk data also requires considerable care and must take into account patient demographics.[13,14] Quantifying patient detriment needs to take into account the life expectancies of the exposed population and the lengthy latent period associated with radiation-induced cancers.[15]

The purpose of this paper is four-fold as follows: (1) to explain the dose parameters [CTDI, dose length product (DLP)] currently used in CT dosimetry; (2) to show how these radiation quantities can be converted into organ dose and effective dose, taking into account technical factors (kV, scan length, and specific scan region) and patient characteristics; (3) to quantify radiation risks in a cohort of patients undergoing cardiac CT using retrospectively gated imaging; and (4) to describe how cardiac CT patients should be protected by ensuring only indicated examinations are performed that use no more radiation than is required to obtain the necessary diagnostic information.

CT DOSIMETRY PARAMETERS

CTDI

The CTDI was introduced into clinical practice in the early days of CT.[16] CTDIs are measured in cylindrical acrylic phantoms for one single rotation of the x-ray tube through 360 degrees.[11] About half the x-ray beam energy is

From the Department of Radiology and Radiological Sciences, Medical University of South Carolina, Charleston, SC.

U. Joseph Schoepf is a medical consultant for and receives research support from Bayer-Schering, Bracco, General Electric, Medrad, and Siemens.

Walter Huda receives research support from Siemens.

Reprints: W. Taylor Rowlett, MD, Department of Radiology and Radiological Sciences, Medical University of South Carolina, 96 Jonathan Lucas Street, MSC 323, Charleston, SC 29425-3230 (e-mail: rowlettw@musc.edu).

deposited in the directly irradiated region of the phantom, and the remaining is deposited in scatter tails. Although CTDIs are measured in a single x-ray tube rotation, they quantitatively predict the radiation dose that would be obtained from a series of contiguous rotations of the x-ray tube. A measured CTDI in the acrylic phantom also predicts the dose from a helical scan performed with a pitch of 1.[17]

CTDIs are normally measured using 100 mm long, pencil-shaped ionization chambers that capture (integrate) the radiation dose profiles along the long axis of the patient.[18] As the integration length of the ionization chamber is 100 mm, this CTDI metric is sometimes referred to as $CTDI_{100}$.[19] In cardiac CT, the phantom used to measure CTDI is 32 cm in diameter, and measurements are made at both the center ($CTDI_c$) and the periphery ($CTDI_p$). The quantity that is measured by the pencil chamber ionization chamber is air kerma, which quantifies the x-ray beam intensity and is measured in mGy. Current practice defines a weighted $CTDI_w$, as

$$CTDI_w = 1/3CTDI_c + 2/3CTDI_p$$

which yields the average intensity of the x-ray beam when the averaging is done over all the phantom radii.[20]

Table 1 shows typical values of $CTDI_w$ for a range of CT scanners operated over the range of x-ray tube voltages currently encountered on commercial CT scanners.[21] For a given CT scanner, the value of $CTDI_w$ is primarily determined by the choice of x-ray tube voltage (kV), x-ray tube current (mA), and x-ray tube rotation time (s). Of great importance is the fact that the mAs and x-ray tube voltage (kV) do not predict the amount of radiation that is delivered to the patient. CTDI values shown in Table 1 differ by up to a factor of 2 for a given kV because of differences in x-ray tube design as well as differences in x-ray tube filtration and/or beam-shaping filters.[3,4] Accordingly, CT protocols should never be expressed in terms of the selected mAs, but in terms of the actual amount of radiation that is used (ie, CTDI).

$CTDI_{vol}$ and DLP

One radiation parameter currently provided in most commercial CT scanners is the volume CTDI, expressed as $CTDI_{vol}$, which was developed to explicitly account for the pitch in helical CT.[22] $CTDI_{vol}$ is defined as $CTDI_w$ divided by the CT pitch, where the pitch is the ratio of the distance moved by the patient table per 360-degree rotation of the x-ray tube divided by the nominal width of the x-ray beam.[23] A table movement of 40 mm per 360-degree rotation of the x-ray tube for a CT scanner with a 40 mm wide x-ray beam corresponds to a pitch of 1. If the table only moved 20 mm per 360-degree rotation of the x-ray tube, the pitch would be reduced to 0.5, and doses (ie, $CTDI_{vol}$) would increase because parts of the phantom would be irradiated more than once (ie, overlap of doses). Moving the table 60 mm for a 360-degree rotation of the x-ray tube results in an increased pitch of 1.5, and doses (ie, $CTDI_{vol}$) would be reduced because parts of the phantom would not be directly irradiated, resulting in gaps in the radiation profiles.

Once the x-ray tube voltage (kV), tube current/rotation time (mAs), and CT pitch are specified, the $CTDI_{vol}$ value is "fixed" and is independent of the patient being scanned or of the length of the patient that is scanned. Selection of a short scan length of 10 cm, or a longer scan length of 100-cm, would result in exactly the same $CTDI_{vol}$. However, the amount of radiation received by a patient having a 100-cm scan will be about 10 times greater than what the patient receives in a 10-cm scan, assuming that all other factors are kept constant. For this reason, all CT scanners also provide the DLP, which is the product of $CTDI_{vol}$ and the scan length.[11,17]

DLP is measured in mGy cm and quantifies the total amount of radiation used to perform a given CT scan. In helical CT scanning, there can be ambiguity about the precise definition of the scan length because of over-ranging required for the interpolation of projection data. A robust alternative for determining DLP is to multiply $CTDI_{vol}$ (mGy) by the product of the beam width (cm) and the number of 360-degree rotations of the x-ray tube.

TABLE 1. Representative Values of $CTDI_w$ (μGy/mAs) for Four Commercial CT Scanners Operated at X-ray Tube Voltages Ranging From 80 to 140 kV

CT Vendor/Model	80 kV	100 kV	120 kV	140 kV
Definition AS (Siemens)	22	44	76	113
VCT (GE)	34	62	95	133
Aquilion 16 (Toshiba)	44	79	121	N/A
Brilliance 16 (Philips)	N/A	N/A	71	102

VCT indicates volume computed tomography.

CTDI and DLP Usage

$CTDI_{vol}$ is a measure of the intensity of radiation being used and quantifies the amount of radiation being delivered per unit distance (cm) of the patient. $CTDI_{vol}$ is the correct parameter to ensure that the intensity of radiation being used is appropriate. For example, the American College of Radiology CT accreditation program has a reference $CTDI_{vol}$ value for a routine abdominal CT scan in an average-sized adult (70 kg) of 25 mGy.[24] Any facility that uses more than this amount of radiation in performing this type of scan will be notified that they are exceeding the normal-sized adult abdomen reference dose, and an investigation into the need for such a high value would be warranted. In addition, if $CTDI_{vol}$ for a routine abdominal CT examination exceeds 30 mGy, the scanner will fail to receive American College of Radiology CT accreditation. It is important to note that the appropriate $CTDI_{vol}$ depends on the diagnostic task.[25] Screening of asymptomatic patients for lung cancer would use less radiation than a routine chest CT examination.[26] The appropriate $CTDI_{vol}$ also depends on patient size and would be increased for larger patients and reduced for pediatric patients.[27,28]

DLP explicitly takes into account radiation intensity and scan length. A DLP value is not a measure of how much radiation a patient receives, as DLP does not take into account any characteristics of the patient. DLP simply quantifies the total amount of radiation used. DLP data allow different facilities the ability to compare the amount of radiation used when performing similar examinations on a specified group of patients (eg, between 65 and 75 kg).[29,30]

Table 2 shows pitch and DLP data for typical coronary CT angiography examinations.[31,32] The DLP data in Table 2 can be used to derive $CTDI_{vol}$ values assuming a cardiac CT scan length of approximately 16 cm. It is evident that the pitch value has a critical effect on the amount of radiation used to perform the examination.[33] $CTDI_{vol}$ and

TABLE 2. Typical Pitch and DLP Values in Coronary Cardiac CT Angiography

Type of Scan	Pitch	Typical DLP (mGy cm)
Retrospective gating	~0.2	1200
Step and shoot	~1.0	300
Dual source	~3.5	75-150

DLP measure the amount of radiation used to perform an examination and should not be mistaken for any kind of "patient dose" per se. To obtain patient doses, it is necessary to use conversion factors that permit $CTDI_{vol}$ and DLP data to be "translated" into patient doses.

DOSE CONVERSION FACTORS

Rationale

CTDI data can be used to derive patient doses by using appropriate dose/CTDI conversion factors.[12,30,34] Multiplication of a generic dose/CTDI conversion factor by the appropriate CTDI will generate the corresponding patient "dose." Understanding how dose/CTDI conversion factors are generated helps clarify when it is legitimate to use such factors to estimate patient doses and helps the medical imaging practitioners better understand when their use would be inappropriate and could result in erroneous patient doses.

Consider modeling a cardiac CT examination using a mathematical phantom of the type depicted in Figure 1 or making measurements in an anthropomorphic phantom.[35,36] In each case, it is possible to obtain an accurate and comprehensive assessment of the absorbed doses to all radiosensitive organs and tissues of interest for radiologic assessment purposes as well as to determine $CTDI_{vol}$ and DLP for this examination. Patient doses, expressed as either organ doses or effective doses, are directly proportional to the amount of radiation used to perform the examination (ie, $CTDI_{vol}$ and DLP). Doubling $CTDI_{vol}$, for example, would double the DLP, organ dose, and effective dose. Measured or computed ratios of organ dose/$CTDI_{vol}$ or effective dose/DLP (E/DLP) obtained in this manner are examples of CT dose conversion factors.

It is critical to note that dose/CTDI conversion factors are applicable only for examinations performed in a similar manner to the one that was used to obtain the conversion factor.[37] Factors such as the x-ray tube voltage, scan region and scan length as well as patient characteristics must be similar to the examination used to acquire the dose conversion factor. Use of conversion factors that do not take into account differences in either x-ray techniques or patient size characteristics is inappropriate. Most current estimates of patient effective dose in the scientific literature are erroneous because of failure to use E/DLP conversion factors that explicitly take into account patient characteristics (ie, size) and CT technique factors (ie, kV).[37,38]

Dose/CTDI Conversion Factors

Table 3 shows ratios of (organ dose)/($CTDI_{vol}$) obtained using a popular CT dosimetry spread sheet (ImPACT) for cardiac and chest CT scans performed at 120 kV in a 70-kg adult patient. When a normal-sized adult undergoes a 16-cm cardiac CT examination at 120 kV, the heart dose is approximately 40% higher than the $CTDI_{vol}$ displayed on the operator's console. If this patient underwent a complete 35-cm chest CT examination, the heart dose would be 60% higher than the $CTDI_{vol}$ displayed on the console. However, if the scan region or range differs markedly from those specified in Table 3, then these conversion factors are inappropriate and should not be used. Chest CT conversion factors generally cannot be used for heart CT imaging and vice versa. In addition, the (organ dose)/($CTDI_{vol}$) conversion factors listed in Table 3 depend on the x-ray tube voltage used to perform the CT examination (see below).

Patient size is an important factor that always needs to be taken into account in CT dosimetry.[34,39,40] A recent study of breast doses in 30 women showed that for constant CT scanner techniques, the average glandular doses ranged

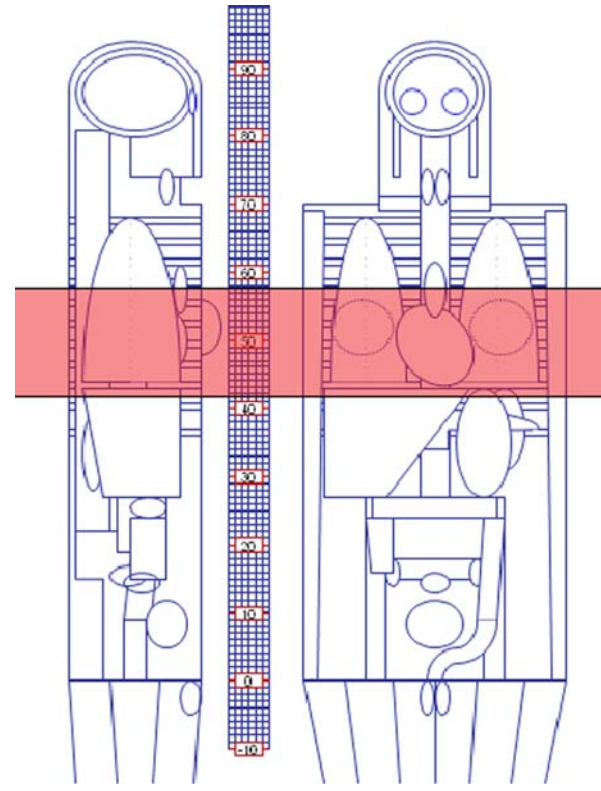

FIGURE 1. Mathematical phantom used by the ImPACT spread sheet to compute organ doses in CT. The shaded region depicts a 16-cm long heart scan that ranges from $z=42$ up to $z=58$. (Color figure available at www.thoracicimaging.com)

TABLE 3. Representative Organ Doses in Cardiac and Chest CT Scans [Organ Dose (mGy) per Unit $CTDI_{vol}$ (mGy)] Performed at 120 kV

Organ	Organ Sensitivity	16-cm Heart Scan	35-cm Chest Scan
Breast	High	~1.1	1.2
Lung	High	1.1	1.6
Red bone marrow	High	0.2	0.5
Liver	Moderate	0.3	0.9
Esophagus	Moderate	0.9	1.8
Thyroid	Moderate	< 0.1	0.3
Heart	Low	1.4	1.6

from approximately 30 mGy in the smallest patients to about half this value for the largest patients.[41] Figure 2 shows how the radiation doses vary with patient weight (kg) for adults undergoing chest CT examinations at 120 kV and when the amount of radiation used ($CTDI_{vol}$ and DLP) is kept constant. Data shown in Figure 2 were generated by modeling patients as uniform cylinders of water and calculating the change in water phantom dose as a function of water phantom diameter for a constant CT x-ray tube output at 120 kV.[42,43] The data shown in Figure 2 illustrate quantitatively how reducing patient size will increase organ doses and vice versa.

Table 4 shows published values of E/DLP for chest CT scans performed at 120 kV as a function of patient age (ie, size).[12,34] Data in Table 4 take into account the most recent tissue-weighting factors provided by the International Commission on Radiological Protection (ICRP) in Publication 103, in which the relative radiosensitivity of the breast was increased from 0.05 to 0.12.[13] These data show the importance of taking into account patient characteristics, as the E/DLP conversion factors increase by nearly a factor of 5 when a newborn is scanned relative to a 70-kg adult. Plotting the E/DLP conversion factor as a function of patient weight yields the graph depicted in Figure 3, which permits interpolation (or extrapolation) of E/DLP conversion factors for chest CT examinations that are applicable to any patient size (kg).

Factors That Affect Dose Conversion Factors

The selection of a set of technical parameters (kV, mAs, pitch, scan length, etc) defines the amount of radiation that is directed at the patient and is independent of any dosimetry phantom size. Data in Table 4 allow computation of an effective dose in chest CT from DLP data, irrespective of whether a small or large phantom was used to quantify the CT output. As expected, patient dose is independent of the arbitrary choice of phantom size that is selected to quantify CT radiation output.[17,21] This is because the use of a smaller phantom size simply doubles the CTDI dose and halves the corresponding E/DLP conversion factor. Practitioners simply need to ensure that the conversion factor used is the same as the phantom size used to quantify the CT output. Contrary to common belief by many in the imaging community, it is irrelevant which sized phantom is used to quantify the CT scanner output, and the radiation used to perform scans in adult or in pediatric patients can be quantified using CTDI measured in either small or large phantoms.

TABLE 4. Representative E/DLP (μSv/mGy cm) Conversion Factors for Use in Chest CT Imaging

Patient Characteristics		Phantom in Which DLP Is Measured	
Age (y)	Weight (kg)	32-cm Diameter	16-cm Diameter
Newborn	3.5	95	47
1	10	63	32
5	19	44	22
10	33	32	16
Adult	70	21	10

Figure 4 shows how the E/DLP conversion factor (4-cm wide beam/120 kV) varies with the anatomic position (z) and the anthropomorphic phantom (Fig. 1) in which effective doses are measured.[12] The maximum E/DLP value (37 μSv/mGy cm) is located in the region of the breast ($z \sim 56$ cm), which is a particularly sensitive region. The E/DLP conversion factor at the top of the lung ($z \sim 68$ cm), however, is only approximately 7 μSv/mGy cm, which is more than a factor of 5 lower than the E/DLP in the vicinity of the breast. These data clearly show that a single conversion factor cannot be used for all examinations in the chest region. The most appropriate E/DLP conversion factor for a 70-kg adult undergoing a cardiac CT examination ($z \sim 42$ to 58 cm) at 120 kV is 26 μSv/mGy cm, which is much greater than the value (21 μSv/mGy cm) listed in Table 4 for chest CT examinations. Reports of reduced "effective doses" could arise as a result of the use of different E/DLP conversion factors because of different scan lengths or different tissue-weighting factors used to obtain patient effective doses. Accordingly, any valid intercomparison between different scanners or protocols should be based on $CTDI_{vol}$ and DLP data and not on the computed effective doses.[44]

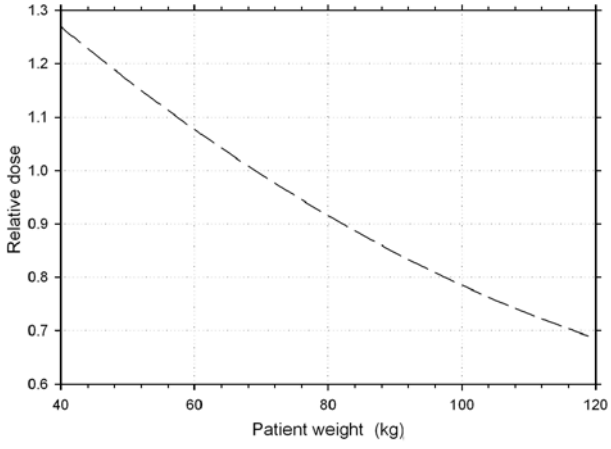

FIGURE 2. Changes in dose to organs in the chest as a function of patient weight at a constant incident radiation intensity ($CTDI_{vol}$ and DLP). Doses have been normalized to unity for a 70-kg adult.

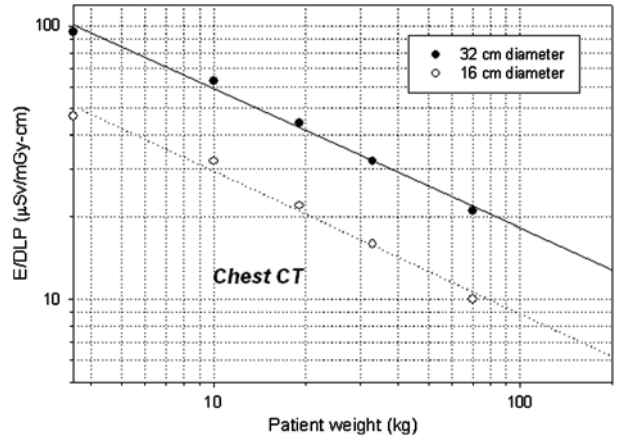

FIGURE 3. E/DLP conversion factors for chest CT examinations as a function of patient weight, obtained using ICRP 103 weighting factors. Closed circles are for DLP in 32-cm diameter acrylic phantoms, and open circles are for DLP in 16-cm diameter acrylic phantoms.

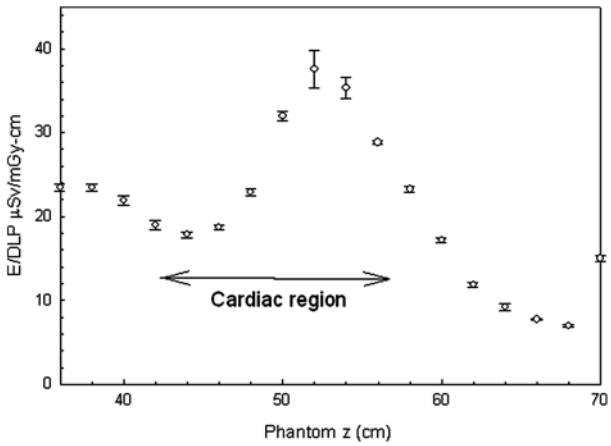

FIGURE 4. Values of E/DLP for 4-cm scan lengths in the chest region as a function of the long patient axis location (*z*) shown in Figure 1, obtained using ICRP 103 weighting factors.

A recent study showed that E/DLP conversion factors were independent of x-ray tube voltage (kV) for head CT examinations but were significantly influenced by the choice of kV for body examinations.[12] On average, increasing the x-ray tube voltage from 80 to 140 kV increased body E/DLP factors by approximately 4% per 10 kV increase in x-ray tube voltage. The likely reason why higher kV results in increased E/DLP factors is the increase in the x-ray beam penetration, which would increase the relative dose to radiosensitive organs within the body. Increasing the x-ray tube voltage from 80 to 140 kV increases the E/DLP for body scans by approximately 25%, and neglecting this factor could result in sizeable errors in estimating patient effective doses.

PATIENT DOSES AT CARDIAC CT

Organ Doses

The absorbed dose in any organ may be obtained by dividing the total energy absorbed (joules) by the organ mass (kg).[10] Absorbed doses are expressed in gray (Gy) or mGy, where 1 Gy is 1000 mGy. Organ doses are primarily used to quantify the radiation risk associated with the deposition of ionizing radiation within the organ or tissue. One type of radiation risk is known as deterministic and includes factors such as skin burns and epilation.[13] Deterministic effects, however, have a threshold dose that is of the order of 2000 mGy (2 Gy), below which such effects do not occur.[45] The most important radiation risks in diagnostic radiology are the stochastic processes of carcinogenesis and the induction of genetic effects.[13,14,46] It is generally accepted that the radiation risk of carcinogenesis is directly proportional to the absorbed dose and that there is no threshold dose. Genetic risks require the exposure of the gonads and are generally negligible in cardiac imaging.

Table 3 shows typical organ doses in mGy per mGy $CTDI_{vol}$ in a cardiac CT scan and for a whole chest CT scan. The data in Table 3 are only applicable for a normal-sized (ie, 70-kg) adult patient when imaged with a CT scanner operated at 120 kV. Note that for all organs that are completely irradiated in the chest CT scan (eg, lung, breast, esophagus, and heart), the respective organ doses are markedly higher than the corresponding $CTDI_{vol}$ value. The absorbed dose to the heart in a chest CT, for example, is 60% higher than $CTDI_{vol}$. Organs close to the directly irradiated region, but that primarily receive scattered radiation, have lower doses.[47] The liver dose in a cardiac CT scan, and the thyroid dose in a chest CT scan, both have doses that are only one-third of $CTDI_{vol}$. Organs located further from the directly irradiated region, such as the thyroid on a cardiac CT examination, generally receive very low radiation. In addition, the dose to the embryo or fetus in any cardiac CT examination performed on an expectant mother will be very low and is therefore deemed to be negligible.[48,49]

Organ doses depend on CT technique factors including x-ray tube voltage, x-ray tube current/rotation time, and CT pitch.[23,50] Organ doses are generally increased by increasing kV and mAs and by reducing the pitch. X-ray tube voltage (kV) is the most important technique factor that affects patient dose.[51] Reducing the x-ray tube voltage when all other factors are kept constant would likely result in the largest reduction in patient dose.

Effective Doses

Patients undergoing most diagnostic radiologic examinations have more than one organ that is exposed to radiation. The effective dose is computed by taking into account the dose to each organ and that organ's relative radiosensitivity. Effective doses are measured in sievert (Sv) or millisievert (mSv), with 1 Sv being equal to 1000 mSv. If a patient has an effective dose of 10 mSv for a cardiac CT examination, the patient radiation risk is comparable with the risk of a uniform whole body dose of 10 mGy.[13] One important reason for the popularity of the effective dose in the medical imaging community is that this parameter accounts for all exposed organs in the patient and is directly related to the patient stochastic risk (ie, carcinogenesis and genetic effects).[52,53] In addition, use of the effective dose permits a direct comparison of the radiation received in one radiologic examination with that received by a similar patient undergoing any other radiologic examination. Effective doses to a 70-kg adult undergoing coronary CT angiography would be approximately 31 mSv for a retrospectively gated study, approximately 8 mSv for a "step/shoot" protocol, and approximately 2 to 4 mSv for a dual-source scanner based on techniques listed in Table 2 combined with an E/DLP of 26 μSv/mGy cm. For comparison purposes, typical effective doses for CT examinations include 2 mSv for head, 3 mSv for neck, 7 mSv for chest, 8 mSv for abdomen, and 6 mSv for pelvic imaging.[31] Adult effective doses for body examinations are higher than for head examinations because the most radiosensitive organs are located in the body.[13]

Effective doses associated with cardiac CT may also be compared with natural background exposure. Each year, the average US inhabitant receives approximately 1 mSv from terrestrial radioactivity, cosmic radiation, and primordial radionuclides such as potassium-40.[1] In addition, the average effective dose from domestic radon exposure in the United States is approximately 2 mSv per year. Effective doses may also be compared with current US regulatory dose limits for radiation workers (50 mSv/y) and for members of the public (1 mSv/y).[23,50] Effective doses for the most highly exposed radiation workers (ie, interventional neuroradiologists) were estimated to be 0.007 mSv per procedure, or 5 mSv for 700 procedures.[54] These comparisons help place medical radiation exposures into an appropriate context and help nontechnical personnel

understand how much radiation a cardiac CT patient actually receives.

It is important to note that effective doses are not a risk quantity per se.[13] Although any effective dose can be converted into a radiation risk estimate, this must be performed with great care. To estimate the risk associated with any effective dose, proper account is taken of those organs/tissues that have been exposed and the demographics (ie, age and sex) of the exposed population.[13,14,46] For example, an effective dose of 100 mSv for a newborn male, when delivered as a uniform whole body dose, is associated with a risk of cancer of approximately 2.6%, whereas the same effective dose for a newborn female would be nearly twice as high. In contrast, an effective dose of 100 mSv for a 60-year-old patient from uniform whole-body exposure would have an average radiation risk of only approximately 0.5%.[14]

RADIATION RISKS

Risk Estimates

The principal radiation concern for patients undergoing cardiac CT imaging is the induction of cancer. Important scientific bodies that summarize our knowledge of the carcinogenic risks of ionizing radiation include (a) the ICRP; (b) the United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR); and (c) the US National Academy of Sciences Committee on the Biological Effects of Ionizing Radiation (BEIR). All three scientific bodies agree that the most sensitive organs and tissues are the breasts, lungs, stomach, red bone marrow, and the colon. Moderately sensitive organs include the thyroid, liver, bladder, and esophagus. For radiation protection purposes, ICRP/UNSCEAR/BEIR all agree that it is prudent to assume that there is no threshold radiation dose below which the carcinogenic radiation risk should be taken to be nonexistent.

Radiosensitive organs and tissues that receive the highest radiation doses in cardiac CT are the lungs and the breast.[55] Age and sex are very important factors when estimating the radiation risks of any exposure to ionizing radiation. Figure 5 shows how the risk of radiation-induced breast cancer, and corresponding cancer mortality, varies with age in females and indicates that the breast radiation risk varies by a factor of over 100 between a newborn and an elderly individual. Figure 6 shows how the risk of radiation-induced lung cancer varies with age and sex. For a constant radiation dose, the risk in females is more than twice the risk in males.

FIGURE 5. Radiation-induced breast cancer incidence and mortality per 100,000 women, each exposed to an average glandular breast dose of 100 mGy.

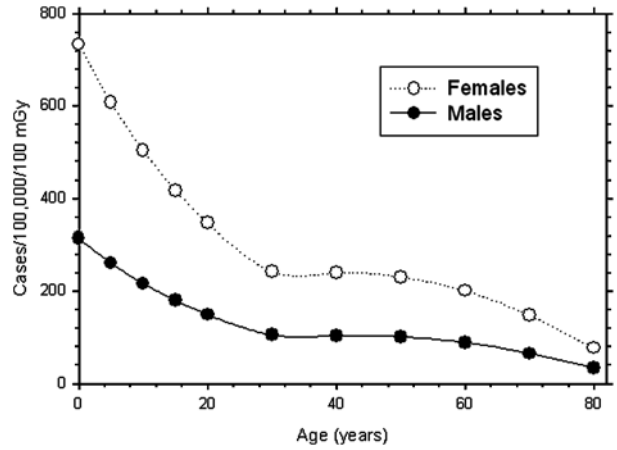

FIGURE 6. Radiation-induced lung cancer incidence for males and females per 100,000 individuals, each exposed to an average lung dose of 100 mGy.

A uniform whole body dose of 1 Gy is equivalent to an effective dose of 1 Sv, and Table 5 shows how the incidence and mortality of radiation-induced cancer vary with sex and age. These radiation risk factors apply only to uniform whole-body irradiation, and use of such conversion factors for an effective dose for any radiologic examination is incorrect. A cardiac CT examination delivers high doses to the lungs and breasts, but deposits little energy in the head and pelvis. As a cardiac CT examination is not the same as uniform whole-body irradiation, conversion of effective doses into risks using data in Table 5 is problematic and could result in serious errors. The most appropriate way to estimate radiation risks in cardiac CT is to obtain organ doses that may be converted into corresponding organ radiation risks.[13]

Cardiac CT Risks

Estimating the risk to patients undergoing diagnostic imaging examinations requires that the amount of radiation used to perform the examination is taken directly from the CT scanner for each patient. For a cardiac CT examination, these factors are $CTDI_{vol}$ and DLP. Patient-specific CTDI, together with imaging characteristics including x-ray tube

TABLE 5. Nominal Radiation Risk Coefficients for Radiation-induced Cancer (% per Sv of Uniform Whole-Body Irradiation) and the Corresponding Fatality Fraction (Fatal Cancers/Induced Cancers)

Patient Age (y)	Male Cancer Incidence, %/Sv (Fatality Fraction)	Female Cancer Incidence, %/Sv (Fatality Fraction)
20	9.8 (0.52)	16.5 (0.46)
40	6.5 (0.58)	8.9 (0.57)
60	4.9 (0.65)	5.9 (0.70)

voltage and CT pitch, permit the estimation of organ doses in a standard-sized patient. Organ doses in a standard-sized patient need to be corrected for the size of the patient population being scanned, using adjustment factors of the type depicted in Figures 2 and 3. Patient size-corrected organ factors can then be converted into the risk of cancer incidence (and mortality), provided that patient age and sex are explicitly taken into account.

At the Medical University of South Carolina, a preliminary risk analysis has been carried out for patients undergoing cardiac CT angiography. A majority of the 104 consecutive patients who were studied were males (62%), with a median age of 59 years and a median weight of 92 kg. All scans were performed at 120 kV with median DLP of 1100 mGy cm, which may be taken to correspond to an effective dose of approximately 29 mSv. In normal-sized patients, the organs receiving the highest radiation doses were the lung (76 mGy) and the female breasts (92 mGy). The average patient lifetime risk for radiation-induced cancer was estimated to be 0.12%, of which 90% would be fatal. Approximately 85% of the estimated radiation risk arose from lung irradiation. Patients with an age and weight at the 10th percentile, and who also received a DLP at the 90th percentile, would have a cancer risk estimate that was approximately double the average value.[55]

It is important to note that there is a controversy regarding radiation risks associated with diagnostic medical imaging procedures.[56] Major leading scientific bodies and many research groups recommend the use of a linear no-threshold model for radiation-induced cancer, with a no-threshold dose below which the radiation risk would be zero.[57,58] Most of the reliable epidemiologic evidence of radiation risks has been generated in effective doses of greater than approximately 100 mSv. Some critics believe that current radiation risks are being overestimated and result in unnecessary concern.[59] Other critics include those who consider exposure to low levels of radiation to be beneficial (hormesis)[60,61] and those who consider the current estimated radiation risks at low doses to be too low.[62] To put these controversies into perspective, it is noteworthy that there is no credible direct evidence of radiation-induced cancer from CT scanning over the past 30 years. The absence of a scientific consensus regarding radiation risks in cardiac CT strongly suggests that these radiation risks, if any, are most likely "low."[26,63]

In the United States today, approximately 38,000 people die in automobile accidents each year, which corresponds to a risk of approximately 0.13 per 1000 inhabitants.[64] A diagnostic radiologic examination could carry a similar numerical risk, but two such risks cannot be directly equated. If a typical 50-year-old individual is killed in an automobile accident, he/she would lose about 30 years of normal life expectancy. The harm to any patient having a radiologic examination with a risk of 0.13 per 1000 patients would need to explicitly take into account both the life expectancy of the patient and the latent period for radiation-induced cancers.[65] Life expectancy of cardiac CT patients is likely to be lower than the life expectancy for the general population, and radiation-induced cancers have latent periods that are measured in decades. The harm (detriment) from the radiologic risk of 0.13 per 1000 will thus likely be much less than the harm from a risk of 0.13 per 1000 of dying in any automobile accident. Accordingly, "patient detriment" is the quantity that would need to be determined, rather than a numerical value of "patient risk" per se. Patient detriment is likely to become an increasingly active and important research field in medical imaging.

Protecting Patients From Harm

To minimize any patient risks, radiation protection philosophy invokes the principle of "justification."[13,29] Knowledge of patient risks permits clinicians to balance such risks against the corresponding benefits.[66] A "justified" (indicated) examination is one in which patient benefit is judged to be greater than patient harm. Identifying indicated examinations requires a professional judgment to be made by physicians who clearly understand the radiation dosimetry and diagnostic performance issues in cardiac CT imaging.[67,68]

A second and equally important radiation protection principle is the one known as "optimization." Understanding radiation risks in cardiac CT requires practitioners to minimize patient doses/risks without compromising important diagnostic information. An "optimized" CT protocol is one that uses no more radiation than is required to extract the required diagnostic information. In practice, this requires those responsible for performing the cardiac CT examination to keep patient exposures as low as reasonably achievable.[69,70] Application of "as low as reasonably achievable" requires input from individuals knowledgeable about how the CT techniques affect patient dose and image quality and can thereby help maximize diagnostic performance relative to radiation dose.[29]

CONCLUSIONS

Currently, $CTDI_{vol}$ and DLP are the two parameters used by CT scanner technology to provide an assessment of a patient's radiation exposure. By using appropriate conversion factors, these quantities can be translated into effective doses and organ doses, provided the technical factors (kV, scan length, and region) and patient characteristics (patient weight) are taken into account. Once effective and organ doses are known, a patient's overall risk of experiencing a carcinogenic event from that given study can be estimated. Recent estimates in cardiac CT from our practice are approximately 0.12% for radiation-induced cancer incidence. To minimize such risk, it is imperative that clinicians ensure that only indicated examinations are performed, using the lowest doses that are reasonably achievable.

REFERENCES

1. National Council on Radiation Protection and Measurements Report No. 160. *Ionizing Radiation Exposure of the Population of the United States*. Bethesda, MD: National Council on Radiation Protection and Measurements; 2009.
2. McCollough CH, Guimarães L, Fletcher JG. In defense of body CT. *AJR Am J Roentgenol*. 2009;193:28–39.
3. Kalender WA. *Computed Tomography. Fundamentals, System Technology, Image Quality, Applications*. New York, NY: Wiley; 2005.
4. Hsieh J. *Computed Tomography: Principles, Design, Artifacts, and Recent Advances*. Bellingham, WA: SPIE; 2003:230.
5. Bastarrika G, Thilo C, Headden GF, et al. Cardiac CT in the assessment of acute chest pain in the emergency department. *AJR Am J Roentgenol*. 2009;193:397–409.
6. Rubinstein R, Halon DA, Gaspar T, et al. Usefulness of 64-slice multidetector computed tomography in diagnostic triage of patients with chest pain and negative or nondiagnostic exercise treadmill test result. *Am J Cardiol*. 2007;99:925–929.

7. Linton OW, Mettler FA Jr. National conference on dose reduction in CT, with an emphasis on pediatric patients. *AJR Am J Roentgenol*. 2003;181:321–329.
8. The U.S. Food and Drug Administration. Radiation-Emitting Products Computed Tomography (CT). Available at: http://www.fda.gov/cdrh/ct/risks.html. Published 2002. Last updated July 27, 2009. [Accessed October 17, 2009].
9. Semelka RC, Armao DM, Elias J Jr, et al. Imaging strategies to reduce the risk of radiation in CT studies, including selective substitution with MRI. *J Magn Reson Imaging*. 2007;25:900–909.
10. Huda W. Radiation dosimetry in diagnostic radiology. *AJR Am J Roentgenol*. 1997;169:1487–1488.
11. McNitt-Gray MF. AAPM/RSNA Physics tutorial for residents: topics in CT. radiation dose in CT. *Radiographics*. 2002;22:1541–1553.
12. Huda W, Ogden KM, Khorasani MR. Converting CT dose length product (DLP) to effective dose. *Radiology*. 2008;248:995–1003.
13. International Commission on Radiological Protection Publication 103. The 2007 Recommendations of the ICRP. *Ann ICRP*. 2007;37:2–4.
14. National Research Council Committee to Assess Health Risks from Exposures to Low Level of Ionizing Radiation NRC. *Health Risks From Exposure to Low Levels of Ionizing Radiation: BEIR VII, Phase 1*. Washington, DC: National Academies Press; 2005.
15. Hall E. *Radiobiology for the Radiologist*. Philadelphia, PA: Lippincott Williams & Wilkins; 2006.
16. 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.
17. Galanski M, Hidajat N, Maier W, et al. *Radiation Exposure in Computed Tomography*. 4th ed. Hamburg, Germany: COCIR; 2000.
18. Suzuki A, Suzuki MN. Use of a pencil-shaped ionization chamber for measurement of exposure resulting from a computed tomography scan. *Med Phys*. 1978;5:536–539.
19. Boone JM. The trouble with CTD100. *Med Phys*. 2007;34:1364–1371.
20. Leitz W, Axelsson B, Szendro G. Computed tomography dose assessment—a practical approach. *Rad Prot Dosimetry*. 1995;57:377–380.
21. ImPACT. impactscan.org. Available at: http://www.impactscan.org/ctdosimetry.htm. Published June 27, 2006. [Accessed October 6, 2009]
22. IEC. Medical Electrical Equipment—Part 2-44: Particular Requirements For the Safety of x-Ray Equipment for Computed Tomography. International Electrotechnical Commission Standard 60601-2-44 Ed 2 Amendment 1. Geneva: IEC; 2003.
23. Bushberg JT, Seibert JA, Leidholdt EM Jr, et al. *Essential Physics of Medical Imaging*. 2nd ed. Philadelphia, PA: Lippincott Williams & Wilkins; 2001:500.
24. McCollough CH, Bruesewitz MR, McNitt-Gray MF, et al. The phantom portion of the American College of Radiology (ACR) computed tomography (CT) accreditation program: practical tips, artifact examples, and pitfalls to avoid. *Med Phys*. 2004;31:2423–2442.
25. International Commission on Radiological Units and Measurements (ICRU) Report 54. Medical Imaging—The Assessment of Image Quality; 1996.
26. Brenner DJ. Radiation risks potentially associated with low-dose CT screening of adult smokers for lung cancer. *Radiology*. 2004;231:440–445.
27. The Alliance for Radiation Safety in Pediatric Imaging. Image Gently. Available at: http://www.pedrad.org/associations/5364/ig/. Published January 22, 2008. [Accessed October 6, 2009]
28. Ogden K, Huda W, Scalzetti EM, et al. Patient size and x-ray transmission in body CT. *Health Phys*. 2004;86:397–405.
29. International Commission on Radiological Protection (ICRP) Publication 87. Managing patient dose in computed tomography. *Ann ICRP*. 2000;30:7.
30. Bongartz G, Golding SJ, Jurik GA, et al. European guidelines on quality criteria for computed tomography. *EUR*. 1999;16262. ISBN 92-828-7478-8.
31. Mettler FA Jr, Huda W, Yoshizumi TT, et al. Effective doses in radiology and diagnostic nuclear medicine: a catalog. *Radiology*. 2008;248:254–263.
32. Stolzmann P, Leschka S, Scheffel H, et al. Dual-source CT in step-and-shoot mode: noninvasive coronary angiography with low radiation dose. *Radiology*. 2008;249:71–80.
33. Hsieh J, Londt J, Vass M, et al. Step-and-shoot data acquisition and reconstruction for cardiac x-ray computed tomography. *Med Phys*. 2006;33:4236–4248.
34. Shrimpton PC, Hillier MC, Lewis MA, et al. National survey of doses from CT in the UK: 2003. *Br J Radiol*. 2006;79:968–980.
35. Lee C, Williams JL, Bolch WE. The UF series of tomographic computational phantoms of pediatric patients. *Med Phys*. 2005;32:3537–3548.
36. Huda W, Sandison GA. The use of the effective dose equivalent, HE, as a risk parameter in computed tomography. *Br J Radiol*. 1986;59:1236–1238.
37. Huda W. Computing effective doses from dose-length product in CT. *Radiology*. 2008;248:321.
38. Hurwitz LM, Reiman RE, Yoshizumi TT, et al. Radiation dose from contemporary cardiothoracic multidetector CT protocols with an anthropomorphic female phantom: implications for cancer induction. *Radiology*. 2007;245:742–750.
39. International Commission on Radiological Protection Publication 89. Basic anatomical and physiological data for use in radiological protection: reference values. *Ann ICRP*. 2002;32:3–4.
40. Huda W, Vance A. Patient radiation doses from adult and pediatric CT. *AJR Am J Roentgenol*. 2007;188:540–546.
41. Angel E, Yaghmai N, Jude CM, et al. Monte Carlo simulations to assess the effects of tube current modulation on breast dose for multidetector CT. *Phys Med Biol*. 2009;54:497–512.
42. Huda W, Atherton JV, Ware DE, et al. An approach for the estimation of effective radiation dose at CT in pediatric patients. *Radiology*. 1997;203:417–422.
43. Huda W, Scalzetti EM, Roskopf M. Effective doses to patients undergoing thoracic computed tomography examinations. *Med Phys*. 2000;27:838–844.
44. Hausleiter J, Meyer T, Hermann F, et al. Estimated radiation dose associated with cardiac CT angiography. *JAMA*. 2009;301:500–507.
45. Wagner LK, Eifel PJ, Geise RA. Potential biological effects of following high x-ray dose interventional procedures. *J Vasc Inter Radiol*. 1994;5:71–84.
46. UNSCEAR 2000. *Sources and Effects of Ionizing Radiation. 2000 Report to the General Assembly with Annexes*. New York, NY: United Nations; 2000:654.
47. Boone JM, Cooper VN III, Nemzek WR, et al. Monte Carlo assessment of computed tomography dose to tissue adjacent to the scanned volume. *Med Phys*. 2000;27:2393–2407.
48. Huda W. When a pregnant patient has a suspected pulmonary embolism, what are the typical embryo doses from a chest CT and a ventilation/perfusion study? *Pediatr Radiol*. 2005;35:452–453.
49. Wagner LK, Lester RG, Saldana LR. *Exposure of the Pregnant Patient to Diagnostic Radiations*. 2nd ed. Madison, WI: Medical Physics; 1997.
50. Wolbarst AB. *Physics of Radiology*. 2nd ed. Madison, WI: Medical Physics; 2000:376.
51. McCollough CH. Automatic exposure control in CT: are we done yet? *Radiology*. 2005;237:755–756.
52. McCollough CH, Schueler BA. Calculation of effective dose. *Med Phys*. 2000;27:828–837.
53. Huda W. Medical Radiation Dosimetry. RSNA categorical course in diagnostic radiology physics: from invisible to visible—the science and practice of x-ray imaging and radiation dose optimization. *RSNA*. 2006:29–39.

54. Kemerink GJ, Frantzen MJ, Oei K, et al. Patient and occupational dose in neurointerventional procedures. *Neuroradiology*. 2002;44:522–528.
55. Schoepf U, Huda W, Abro J, et al. Realistic risks of radiation-induced cancer in a clinical patient population undergoing cardiac CT. *RSNA*. 2008.
56. International Commission on Radiological Protection Publication 99. Low-dose extrapolation of radiation cancer risk. *Ann ICRP*. 2005;35:1–142.
57. Cardis E, Vrijheid M, Blettner M, et al. Risk of cancer after low doses of ionising radiation: retrospective cohort study in 15 countries. *Br Med J*. 2005;331:77.
58. de Gonzalez A, Darby S. Risk of cancer from diagnostic x-rays: estimates for the UK and 14 other countries. *Lancet*. 2004;363:345–351.
59. Cohen BL. Cancer risk from low-level radiation. *AJR Am J Roentgenol*. 2002;179:1137–1143.
60. Cameron JR, Moulder JE. Proposition: radiation hormesis should be elevated to a position of scientific respectability. *Med Phys*. 1998;25:1407–1410.
61. Sagan LA (Editor). Special Issue on Radiation Hormesis. *Health Phys*. 1987;52:517–680.
62. Gofman JW, O'Connor E. *X-Rays Health Effects of Common Exams*. San Francisco: Sierra Club Books; 1985.
63. Brenner DJ, Doll R, Goodhead DT, et al. Cancer risks attributable to low doses of ionizing radiation: assessing what we really know. *Proc Natl Acad Sci U S A*. 2003;100:13761–13766.
64. National Highway Traffic Safety Administration. Fatality Analysis Reporting System Encyclopedia. Available at: http://www-fars.nhtsa.dot.gov/Main/index.aspx. Published July 6, 2009. [Accessed October 8, 2009].
65. Walton A, Broadbent AL. Radiation-induced second malignancies. *J Palliat Med*. 2008;11:1345–1352.
66. McCollough CH, Primak AN, Braun N, et al. Strategies for reducing radiation dose in CT. *Radiol Clin North Am*. 2009;47:27–40.
67. Huda W, Ravenel JG, Scalzetti EM. How do radiographic techniques affect image quality and patient doses in CT? *Semin Ultrasound CT MR*. 2002;23:411–422.
68. Ravenel JG, Scalzetti EM, Huda W, et al. Radiation exposure and image quality in chest CT examinations. *AJR Am J Roentgenol*. 2001;177:279–284.
69. Slovis TL. The ALARA concept in pediatric CT: myth or reality? *Radiology*. 2002;223:5–6.
70. Gerber TC, Kantor B, McCollough CH. Radiation dose and safety in cardiac computed tomography. *Cardiol Clin*. 2009;27:665–677.

