Curr Radiol Rep (2015) 3:80
DOI 10.1007/s40134-014-0080-x

CT Radiation Exposure: An Overview

Walter Huda

Published online: 11 November 2014

Abstract Radiation used in CT examinations needs to be selected on image quality criteria, and also take into account the patient's exposure. X-ray beam quality is controlled by the X-ray tube voltage (kV), and affects the X-ray beam penetration through a given patient as well as the contrast in the resultant image. CT radiation output can be characterized by the volume CTDI ($CTDI_{vol}$), is controlled by tube output and CT pitch, and determines the random image noise (mottle). The total amount of radiation used to perform an examination is the dose length product (DLP), which is obtained by multiplying $CTDI_{vol}$ with the corresponding scan length, and is related to the corresponding patient radiation exposure. Optimal choices for CT X-ray beam characteristics (i.e., kV, $CTDI_{vol}$, and DLP) need to take into account patient physical characteristics as well as the diagnostic imaging task at hand. Because CTDI metrics only describe the radiation incident on the patient, patient organ doses need to be obtained using conversion factors that account for patient size and examination scan length. Organ doses can be converted into organ risks, which can be summed to estimate the patient carcinogenic radiation risks. A 20-year-old adult weighing 75 kg, undergoing an abdominal pelvic CT examination [$CTDI_{vol}$(L) 15 mGy & DLP 700 mGy-cm], has an estimated cancer induction risk of about 0.1 %. Practitioners need to understand radiation risks to be able to identify indicated examinations, where there is expected to be a net patient benefit, and to comply with the ALARA principle where unnecessary radiation is eliminated.

Keywords CT radiation · Exposure · Risks · ALARA principle

This article is part of Topical Collection on *Radiation Exposure and Reduction*

W. Huda (✉)
Department of Radiology and Radiological Science, Medical University of South Carolina (MUSC), Charleston, SC 29425-3230, USA
e-mail: walterhuda@hotmail.com

Introduction

There has been a dramatic increase in the medical radiation doses in the past few decades [1–3••]. In the 1980s, the average US inhabitant received an annual effective dose of about 3.6 mSv. Most of this radiation (3 mSv) was from natural background radiation, with only about 15 % being manmade. The dominant manmade source of radiation exposure was diagnostic medical imaging, which in the 1980s was primarily from radiography and fluoroscopy examinations. By 2006, however, the average American was receiving 6 mSv every year, with the medical exposure having increased to about 3 mSv. Nearly half of medical exposure was from diagnostic CT examinations, and about a quarter from Nuclear Medicine studies, the latter being overwhelmingly nuclear cardiology. In the US, medical exposures have thus increased by about 600 % in a single generation. CT imaging continues to expand both quantitatively and qualitatively, albeit at a slower rate than hitherto [4].

CT is widely recognized as the dominant source of medical radiation exposure in most developed countries, and focusing on this imaging modality is critical to ensuring that medical radiation is being used in an appropriate manner [5–7]. This paper explains the way in which CT radiation output is currently specified, and how choices made regarding CT X-ray beam characteristics are expected to impact on the resultant image quality. Optimal

selection of techniques for common CT examinations is described for both adult and pediatric patients including an outline of how patient radiation doses, and corresponding risks, can be estimated in routine clinical practice. An overview is provided of the steps to be taken that will ensure an adequate level of patient protection undergoing CT and other types of radiological imaging that involves exposure to ionizing radiations.

CT Radiation Output

Radiation Quantity

The radiation generated by a CT scanner at a given X-ray tube voltage (kV), is determined by the choice of tube current (mA) and the 360° X-ray tube rotation time (s), or the mAs [8]. Knowledge of the mAs, however, is insufficient to define the X-ray tube output as this does not account for X-ray tube characteristics. At the same mAs, CT output can easily vary by a factor of two between individual scanners from one vendor, or between different vendors [9]. In helical CT, the radiation incident on the patient also depends on the choice CT pitch, defined as the table movement per 360° X-ray tube rotation time divided by the nominal X-ray beam width (i.e., detector length x number of detectors). When pitch is reduced, as often occurs in some cardiac CT examinations, patient doses will usually increase substantially [10]. With increasing CT pitch, as can occur for dual source CT scanners, patient doses can be reduced [11].

The volume Computed Tomography Dose Index ($CTDI_{vol}$) is a universal metric that specifies how much radiation is being emitted from the CT scanner per unit scan length [12]. Figure 1 shows the important difference between mAs and $CTDI_{vol}$, with the former being a *relative* indicator of radiation intensity, analogous to the rate at which an automobile engine rotates. However, it is only the $CTDI_{vol}$ that provides an *absolute* specification of radiation intensity that is universally understood by all CT imaging practitioners, analogous to the role played by the speedometer in automobiles. $CTDI_{vol}$ is expressed in mGy, and may be measured in a small acrylic phantom (S) which has a 16 cm diameter or a larger phantom (L) with a 32 cm diameter [8]. As depicted in Fig. 2, a measured value $CTDI_{vol}$(S) equal to 10 mGy in a small phantom may be taken to correspond to a value $CTDI_{vol}$(L) of 5 mGy in a large phantom, and vice versa [13]. All CTDI metrics displayed on CT scanners currently depend only on the techniques selected by the operator (i.e., kV, mAs, and pitch), and are therefore independent of all patient characteristics. Despite the fact that the name includes the word "dose", CTDI metrics should never be interpreted as any kind of "patient dose" [14].

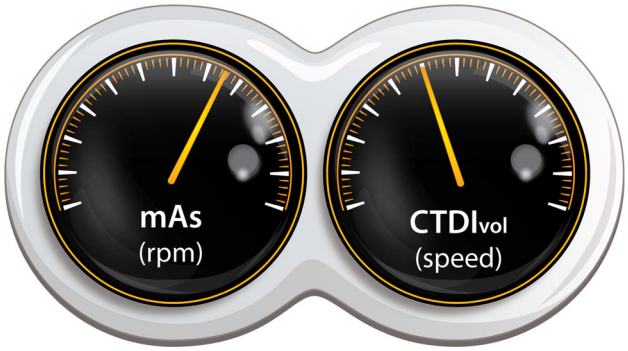


Fig. 1 Selected mAs on a CT scanner (*left*) is a *relative* indicator of the amount of radiation being used. The $CTDI_{vol}$ (*right*) value is a *universal* metric of CT output used to predict the patient dose on any CT scanner operated at any tube voltage

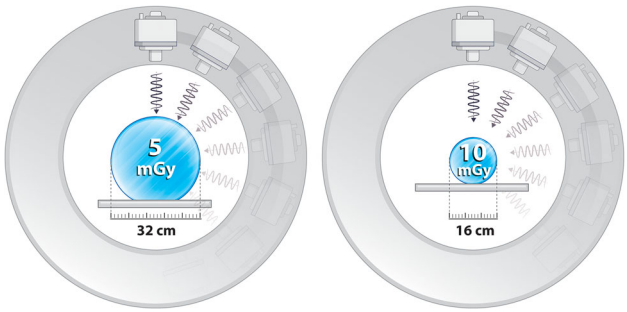


Fig. 2 For the same techniques (kV and mAs) and CT pitch value, the measured $CTDI_{vol}$ in a 32-cm (L)-diameter phantom (*left*) will be approximately one half the corresponding $CTDI_{vol}$ in a 16-cm (S)-diameter phantom (*right*)

For a given amount of attenuation in a patient, the choice of $CTDI_{vol}$ will determine how much radiation reaches the CT detectors [15]. The radiation intensity at the CT X-ray detectors is directly related to the amount of noise (mottle) in the resultant image [16], which affects detection of low-contrast lesions. Accordingly, $CTDI_{vol}$ is one of the most important protocol parameter choices that must be selected by operators in every clinical CT examination. Since CT is a quantum mottle limited medical imaging systems [17], the only technical way to reduce mottle in CT imaging is to use more radiation, which also increases patient dose. Quadrupling the number of photons used to generate an image must quadruple the patient dose, and will also halve the image noise.

Radiation Quality

Radiation quality is a measure of the penetrating power of an X-ray beam, quantified as a thickness of Aluminum (Al) that represents the Half Value Layer (HVL) of the X-ray beam. A typical CT X-ray beam has a HVL of about 7 mm

Al [18], meaning that this thickness of Aluminum attenuates an X-ray beam intensity by 50 %. The primary determinants of X-ray beam quality in CT are the X-ray beam filtration and X-ray tube voltage. CT beam qualities are higher than conventional radiography and fluoroscopy (~3 mm Al) and much higher than digital mammography (~0.4 mm Al) [19, 20]. When the HVL increases, the X-ray beam average energy increases and thereby becomes more penetrating, and vice versa. Beam quality primarily relates to patient penetration, and generally needs to be adjusted whenever patient size is changed.

X-ray beams used in CT have average photon energies of ~60 keV, and will lose about half their X-ray photons passing through ~4 cm of soft tissue [21]. Table 1 shows the nominal sizes of patient expressed as a soft tissue Half Value Layer thickness, which permits the transmitted X-ray intensities to be estimated. X-ray transmission in a newborn (3 HVL) is about 10 % (i.e., $(1/2)^3$), whereas transmission through a 40 cm patient (10 HVL) is only about 0.1 % (i.e., $(1/2)^{10}$) at the same beam quality. Attenuation in an oversized adult is therefore 100 times greater than a newborn, graphically illustrating the importance of accounting for patient size when designing rational CT imaging protocols [22]. CT imaging of a newborn patient is most likely to use the lowest X-ray tube voltages that are currently available on most CT scanners, namely 80 kV [23••]. For the largest patients, however, the highest tube voltages will be more appropriate, namely 140 kV [24].

X-ray beam quality also affects the contrast of a given lesion in all X-ray imaging modalities, including CT [25, 26]. As the average photon energy increases, achieved by increasing the X-ray tube voltage and/or the beam filtration, lesion contrast decreases. This is illustrated by the calculated Hounsfield Unit (HU) values in Table 2, where the HU is the contrast of a particular material relative to that of water. Data in Table 2 illustrate how quantitative changes in lesion contrast depend on the atomic number of the material of interest. When a lesion has an atomic number that is close to that of tissue or water, namely 7.5, the change in contrast is modest, whereas the change in HU with photon energy for Iodine is colossal. When Iodine is imaged, its high atomic number (53) requires the use of low tube voltages (i.e., 80 or 100 kV) to maximize image contrast. Angiograms performed at higher voltages (140 kV), whose average photon energies would be well above the k-shell energy of Iodine (33 keV), will have very poor contrast because of lower attenuation by iodine [25].

Table 1 Patient sizes expressed in terms of CT Soft Tissue Half Value Layers and the corresponding transmitted X-ray beam intensities

Patient	Weight (kg)	Nominal size (cm)	Soft tissue HVL values	Nominal transmitted fraction[a]
Newborn	3	12	3	1/8
5-year old	20	16	4	1/16
Adult	75	24	6	1/64
Oversized adult	>120	40	10	1/1024

[a] Constant beam quality

Table 2 Hounsfield Unit values as a function of CT X-ray tube voltages (kV) [Relative HU values normalized to unity at 120 kV]

Tube voltage (kV)	80	100	120	140
Nominal average photon energy (keV)	40	50	60	80
Fat	−152	−111	−89	−69
	[1.70]	[1.25]	[1.00]	[0.77]
Brain	47	43	39	37
	[1.20]	[1.08]	[1.00]	[0.93]
Soft tissue	62	58	54	52
	[1.14]	[1.06]	[1.00]	[0.96]
Cortical bone	3760	2590	1940	1330
	[1.94]	[1.34]	[1.00]	[0.69]
Calcium	9,570	5,960	3,950	2,090
	[2.42]	[1.51]	[1.00]	[0.53]
Iodine	405,000	267,000	180,000	93,200
	[2.24]	[1.48]	[1.00]	[0.52]

Total Radiation

It is "impossible" to obtain any meaningful patient dose estimate without taking the scan length into account [27, 28]. In a body CT scan, for example, the liver dose will increase approximately proportional to scan length over the organ itself, and continue to increase when the scan extends beyond the organ because of scatter from adjacent tissues [27]. Use of $CTDI_{vol}$ data, and the corresponding scan length in a specified region, can be used to obtain estimates of size-specific organ doses when appropriate correction factors are incorporated for patient size [29]. Size-specific organ doses can be used to estimate individual organ risk factors, which in CT generally relate to the stochastic risk of carcinogenesis [30•]. Individual organ cancer induction risks can be summed to obtain the resultant patient risk of carcinogenesis.

The total amount of radiation used in performing a CT scan is obtained by multiplication of the radiation intensity ($CTDI_{vol}$) and the corresponding scan length (cm) [8, 12]. When $CTDI_{vol}$ is multiplied by the scan length, one obtains the Dose Length Product (DLP) expressed in mGy-cm. A

head CT scan performed at a $CTDI_{vol}$(S) of 60 mGy and a scan length of 20 cm will result in a DLP(S) of 1,200 mGy-cm. An abdominal CT scan performed at a $CTDI_{vol}$(L) of 15 mGy and a scan length of 30 cm will result in a DLP(L) of 450 mGy-cm. The total patient radiation risk is directly related to the total amount of radiation used to perform the CT examination. This increase in risk occurs because individual organs will absorb more radiation, and because more organs will be exposed in longer scans. Doubling the scan length, for example, will double the total energy deposited in a given patient, which will most likely double the corresponding stochastic risk.

Selecting CT Techniques

Adult Techniques

Selection of CT techniques needs to take into account the size of the patient [22, 31], as graphically illustrated by the data shown in Table 1. In general, small patients enable use of the lowest X-ray tube voltages, whereas the largest patients require the use of the highest available voltages to obtain adequate penetration. Selection of techniques must also account for the diagnostic imaging task. One important finding from imaging science is that images do not possess an intrinsic "image quality", per se [32••]. The image shown in Fig. 3 would be deemed to be very poor for detection of a subtle low-contrast lesion in the liver, but perfectly adequate for measuring the aorta diameter. This example graphically illustrates why it is impossible to determine whether the image is "good" or "bad" until the reason why the image was obtained (diagnostic task) has been provided.

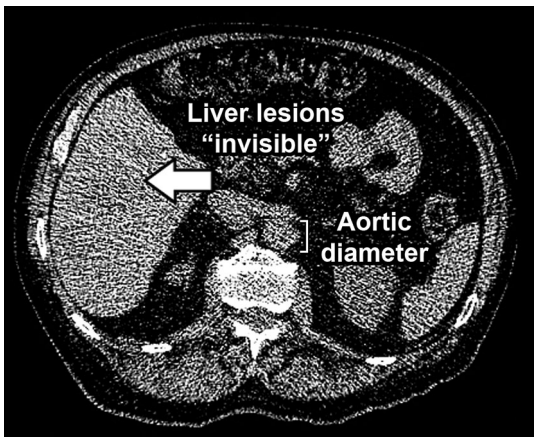


Fig. 3 This noisy abdominal CT image would be deemed to be "very poor" for detecting low-contrast (soft tissue) liver lesions, but satisfactory for measuring the aorta diameter. Image quality is thus *always* task dependent, and not an intrinsic characteristic of any image

Table 3 shows a summary of the current CT protocols for adult patients undergoing head, chest, and abdomen/pelvis CT scans at the Medical University of South Carolina. Head CT scans use 120 kV to reduce beam-hardening artifacts. When iodinated contrast media are used in chest CT scanning (Pulmonary Embolism study), the tube voltage is reduced to improve the visibility of iodinated contrast media. The "liver" scan is a generic term where the diagnostic task refers to soft tissue lesion in a soft tissue organ, and the selected $CTDI_{vol}$ is increased to reduce image mottle that is important when detecting low-contrast lesions. The protocols listed in Table 3 are based on the use of Filtered Back Projection (FBP) image reconstruction algorithms, and the use of Iterative Reconstruction (IR) technique might permit additional dose reductions [33].

Changes in patient size require the CT radiation intensity to be adjusted, which is of particular importance in body CT. Smaller patients require less radiation, and larger patients require more radiation. In CT, Automatic Exposure Control (AEC) systems are available on all modern CT scanners to adjust the X-ray tube output when patient size changes [34•]. The technique parameter that is generally changed on current CT scanners is the tube current (mA), not the tube voltage or the X-ray tube rotation time. Table 4 summarizes the commercially available AEC systems in routine clinical use today, which all adjust the X-ray tube output by changing the tube current. Tube voltage (kV) and the X-ray tube rotation time (s) are generally modified manually by the operator. The X-ray tube voltage affects both X-ray transmission through the patient, as well as X-ray tube output which is proportional to $kV^{2.6}$ [9]. Accordingly, whenever the CT X-ray tube voltage is changed, it is essential that the corresponding intensity (i.e., mAs) is also adjusted to ensure the appropriate X-ray intensity at the CT detectors.

Pediatric Techniques

Optimization of CT scanning is of particular importance when imaging pediatric patients [35–37]. Infants and children are generally recognized as being more radiosensitive than adults, and also are more likely to live long

Table 3 Protocols for normal-sized adults undergoing CT examinations in the Radiology department at MUSC

Body region	Scan type	Tube voltage (kV)	$CTDI_{vol}$ mGy (phantom size)
Head	Routine	120	60 (S)
Chest	Non-contrast	120	9 (L)
	With I contrast	100	9 (L)
Abdomen/Pelvis	Routine	120	14 (L)
	"Liver"	120	18 (L)

Table 4 Automatic exposure control systems on commercial systems

Vendor	AEC systems	Comments
GE	Auto mA and Smart mA	Attempts to control the amount of noise in the CT images using a "Noise Index" value that is selectable by the operator
Philips	Z- or D-DOM	Z-axis thickness is regulated by Z-DOM, and angular thickness is regulated by D-DOM
Siemens	CareDose4D	Users select a Reference mAs value for standard adult (75 kg); five semi-empirical strengths (very weak to very strong)
Toshiba	Sure exposure	Single scanogram modulates along the z-axis, whereas two scanograms enable angular modulation (3D)

enough to permit any radiation detriment to be expressed [38, 39]. In addition, it is important to note that dose conversion factors in younger patients are markedly higher than for adults [40, 41]. For the same incident radiation intensity in head CT examinations, infant effective doses are likely to be three to four times higher than for adults. For these reasons, understanding radiation doses in pediatric CT imaging and optimizing techniques that help eliminate unnecessary radiation doses are of particular importance [22, 25, 42].

A recent publication by Yu et al. [23••] has described in detail how CT examinations in pediatric body imaging can be optimized, using semi-empirical software tools that simulate the effects of changes in the X-ray intensity ($CTDI_{vol}$). For infants under 10 kg, they recommended using 80 kV with $CTDI_{vol}$(L) of about 2 mGy. For infants ranging between 10 and 20 kg, the X-ray tube voltage was increased to 100 kV, and the value of $CTDI_{vol}$(L) was increased to about 4 mGy. For children weighing more than 20 kg (i.e., 5 years and above), the X-ray tube voltage was further increased to 120 kV, and the value of $CTDI_{vol}$(L) ranged between 4 and 6 mGy. Importantly, the critical issue in these examinations was not the radiation deposited in the patient, but the X-ray beam quality (i.e., contrast) as well as the radiation beam quantity is transmitted to the CT detectors (i.e., noise).

A complete description of any imaging protocol has to take into account not only the $CTDI_{vol}$, but the X-ray beam quality [23••, 25, 31]. Selection of optimal techniques requires explicit consideration of the contrast to noise (CNR) ratio of any given lesion, which is affected by X-ray beam quality. Increased tube voltage, for example, will reduce noise because more radiation reaches the CT X-ray detectors, as well as reducing image contrast for all types of lesion as shown by the data in Table 2. For this reason, it is essential that CT imaging practitioners address the issue of image quality, which is even more important than the issue of patient dosimetry. When image quality is sub-optimal, real harm can be inflicted on patients resulting from reduced diagnostic performance [43]. Data shown in Fig. 4 show the energy fluence transmitted through varying patient thickness values for the range of tube voltages currently encountered in CT imaging [31]. These data can be combined with CT output data [9] and lesion contrast values (Table 2) to enable an assessment of CNR for a specified imaging task. It is more appropriate to compare imaging protocols in terms of the image quality than the energy that is deposited in a patient in a radiological examination [32••].

Reference Doses

The American College of Radiology (ACR) runs a CT Accreditation program. To be accredited, facilities need to meet minimum performance standards for image quality metrics such as spatial resolution and detection of low-contrast lesions [44]. In addition, the amount of radiation used to perform routine CT examinations should not exceed specified Reference Doses, normally set at the 75th percentile of a large scale survey of doses over a large and representative survey of facilities. Current ACR CT Accreditation Reference Doses are listed in Table 5 [45]. When $CTDI_{vol}$ values exceed the Reference Doses, a facility would be notified in writing that their doses exceed the recommended Reference Doses. The final column in Table 5 shows the values that are upper $CTDI_{vol}$ limits currently deemed to be acceptable by the ACR, and exceeding these upper $CTDI_{vol}$ limits would currently "fail" the accreditation application.

Data in Table 5 refer to the US, and it is of interest to compare these values with comparable reference doses in Europe [46]. For head CT scans, the European Commission (EC) recommends $CTDI_{vol}$(S) values of 60 mGy. Values of $CTDI_{vol}$(S) recommended in the UK for Multi Slice CT scanners are 65 mGy for the cerebellum, but 100 mGy when imaging the posterior fossa. Overall, currently, techniques ($CTDI_{vol}$) for head CT scans are generally very similar throughout the world. For body CT, the EC Reference Doses [i.e., $CTDI_{vol}$(L)] are 25 mGy similar to the values currently used in the US. In the UK, however, the current Reference Dose [i.e., $CTDI_{vol}$(L)] for body CT is much lower at 14 mGy. Accordingly, the MUSC abdomen protocol shown in Table 3 would be deemed to be a relatively low in the US, but relatively high in the UK.

The study of Yu et al. used a value of $CTDI_{vol}$(L) of 4 mGy when scanning a 5-year-old child (20 kg) [23••]. Unfortunately, this value is incommensurate with the ACR CT Reference Dose for this sized patient (i.e.,

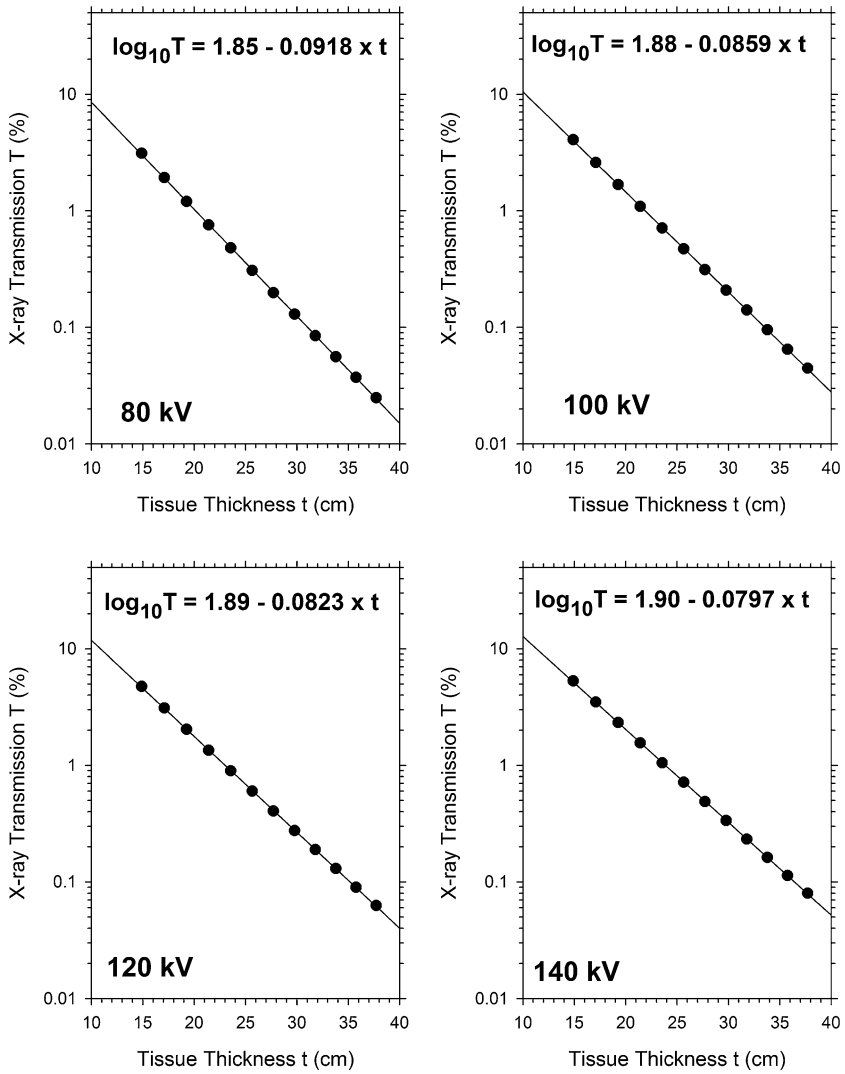


Fig. 4 Transmission of CT X-ray energy fluence through different tissue thickness at four X-ray tube voltages commonly used clinically (80, 100, 120, and 140 kV). The equation in each figure is a least-squares fit to a *straight line* of the shown data

Table 5 ACR CT accreditation reference doses for $CTDI_{vol}$

Patient	Body region	Reference dose (mGy)	"Failing dose" (mGy)
Adult	Head	75 (S)	>80
	Abdomen	25 (L)	>30
Pediatric	Abdomen	20 (S)	>25

$CTDI_{vol}$(S) 20 mGy). The reason for this is that the vendor has provided the CT output in a large phantom, whereas the ACR specifies the CT output in a small phantom. Currently, one vendor specifies the CT output in pediatric body CT using the small phantom (GE), two vendors use the large phantom (Siemens & Philips), and one vendor (Toshiba) recently changed from using a phantom based on patient size (Field of View) to one based on anatomy (i.e., body vs head) [13]. A $CTDI_{vol}$(L) of 4 mGy corresponds to a $CTDI_{vol}$(S) of 8 mGy [13] (Fig. 2), so that the optimized Mayo clinic protocol deposits only about 40 % of energy (dose) in a dosimetry phantom or patient compared to the ACR Reference value. Information on beam quality (kV) is required to assess image quality in terms of lesion contrast (Table 2) and image noise computed using X-ray transmission data (Fig. 4) as well as changes in CT output with changes in tube voltage (i.e., $kV^{2.6}$) [9].

Interpreting Radiation in CT

Computing (Embryo) Doses

Absorbed doses to any organ (e.g., embryo) can be calculated by taking into account the intensity of the radiation used in a CT examination (i.e., $CTDI_{vol}$), and three additional factors (F1 through F3) which are depicted graphically in Fig. 5 [47••].

CT scanner design (F1) The first factor that has to be taken into account is the characteristics of the design of a given CT scanner [9]. As shown in Fig. 5a, most current scanners operated at a $CTDI_{vol}$(L) of 10 mGy would likely deliver a dose of 14 mGy to an organ such as the liver in a normal-sized adult (75 kg) for a "long" scan.
Scan Length (F2) Scan length is a key determinant of any organ dose [27], where a scan from one patient extremity upto the middle of an organ would be expected to deliver about half the dose that would have been received by this organ during a "whole body" scan (Fig. 5b).
Patient Size (F3) At constant CT techniques (kV & mAs), increasing the patient size reduces patient doses, and vice versa [48, 49]. Figure 5c shows that for a $CTDI_{vol}$(L) of 10 mGy, the liver dose in a large patient (100 kg) is reduced to 9 mGy and Fig. 5d shows that the liver dose in a small (45 kg) patient is increased to 18 mGy [30].

To illustrate how these terms would be used in a clinical example, consider a female patient who undergoes an abdominal/pelvic CT examination [$CTDI_{vol}$(L) 15 mGy & DLP(L) 700 mGy-cm] when she is four weeks pregnant. For a typical CT scanner, F1 will be equal to 1.4 in a standard-sized patient. The scan length of 47 cm, obtained by dividing the DLP by the $CTDI_{vol}$, is relatively "long" so that F2 will be very close to unity. Assuming that the patient has an AP dimension of 17 cm, much less than the 23 cm of an average sized adult, F3 will be approximately 1.2. Multiplication of $CTDI_{vol}$(L) by these three factors (i.e., F1, F2, and F3) results in an embryo dose of 25 mGy.

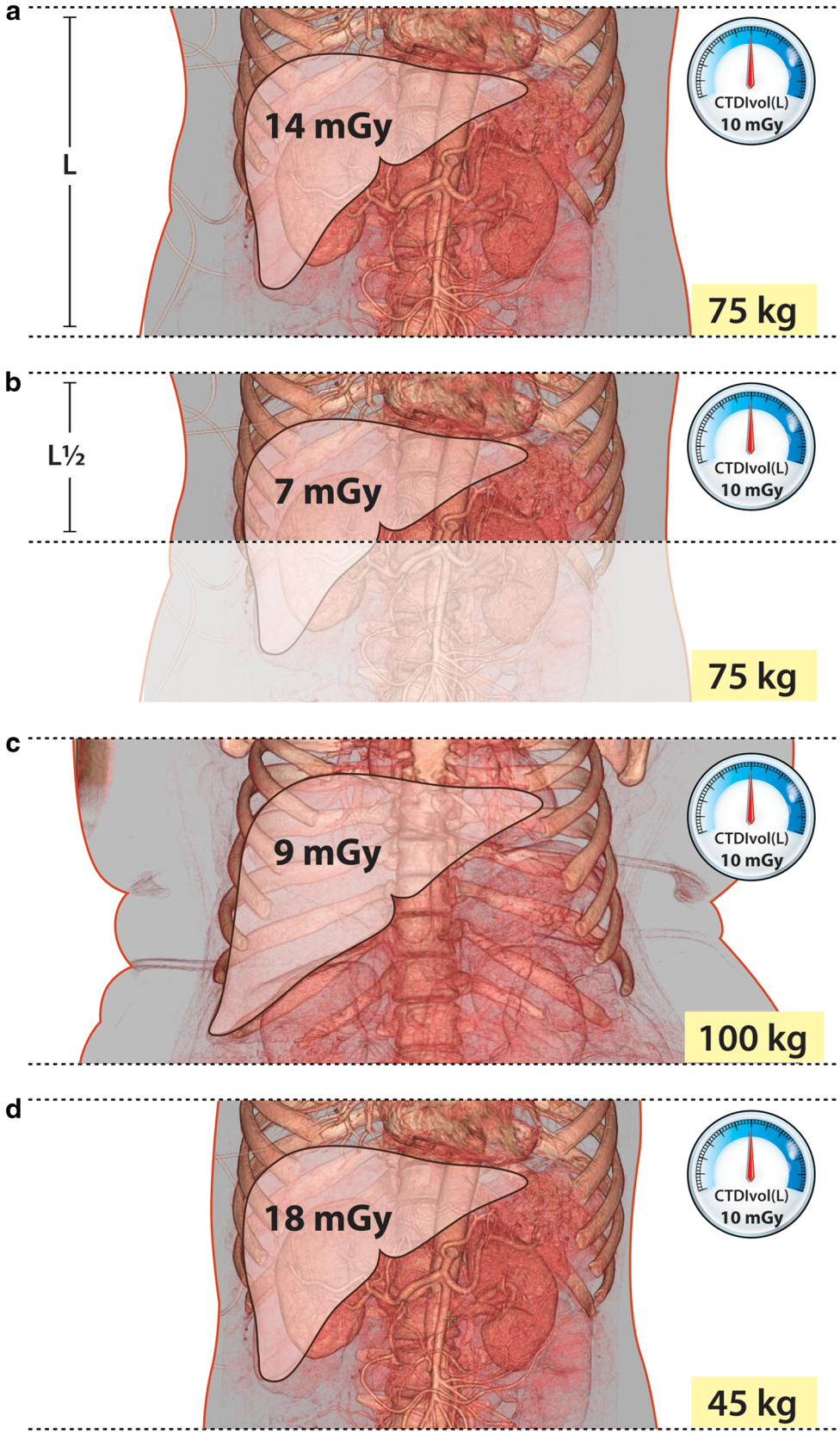
At this radiation dose to the embryo of 25 mGy, there would be zero risk of deterministic effects such as embryonic death, congenital malformation, or mental retardation [50]. The stochastic risk of childhood cancer would be taken to be up to about 1 in 500 (0.2 %), comparable to the background incidence of childhood cancer in the absence of any radiation exposure of 1 in 500 [51, 52]. For prospective CT examinations, these risk estimates would be used when performing a "risk/benefit" analysis to decide whether a given examination is deemed to be worthwhile (see below). For any retrospective assessment, no consideration of any possible medical intervention would be deemed necessary because this embryo dose is below 100 mGy, and any radiation risks would therefore be deemed to be low in comparison with the normal risks of pregnancy [50–52].

Organ Doses and Risks

For any CT examination, it is possible to obtain dose estimates of all radiosensitive organs in standard-sized patients either by experimental measurement or by using Monte Carlo techniques [53–55]. Correction factors can be applied to generate exam-specific organ doses that account for factors such as patient size, age, scanned region, and scan length [56–59]. Consider an abdominal/pelvic CT examination performed using a $CTDI_{vol}$ of 15 mGy and a DLP of 700 mGy-cm. In a normal-sized adult (75 kg), absorbed doses to organs that are directly (& fully) irradiated can be readily estimated by using publicly available software, and shown to be about 20 mGy [60]. Doses to organs that only partially irradiated will be markedly lower [27], and organs that are only subjected to scattered radiation will generally receive very low doses [61].

To convert organ doses into a corresponding radiation risk requires that patient demographics be taken into account [38, 62]. For a North American population, it is customary to use the age- and sex-specific organ risk estimates provided by the Committee on the Biological Effects of Ionizing Radiation (BEIR) [63]. For a 20-year-old male (75 kg) undergoing an abdomen/pelvis examination (DLP 700 mGy-cm), the total cancer incidence risk is estimated at about 1 in 1,000 (0.1 %), with a comparable risk value for a similar-sized 20-year-old female patient [30•]. Two organs dominate this risk, the colon (35 %) and the bladder (25 %).

Generating a long list of organ doses is of likely to be of very limited practical value for most imaging practitioners [64]. The risk to an individual patient can be estimated, and the individual responsible for ordering and/or performing the examination would be expected to have an understanding of the magnitude of these risks. However, it is unlikely to be helpful or practical to operators and/or patients to describe the radiation exposure in terms of a quantitative risk number [65]. The medical imaging community clearly needs to understand and communicate the nominal amount of radiation that patients receive in radiological examinations. As described in the next section, the effective dose (E), used by the International Commission on Radiological Protection (ICRP) for radiation protection purposes, can also be used in medical radiology [64•, 66•]. Employing the effective dose provides imaging practitioners a quantitative indication of the amount of radiation used in any type of radiological examination and is a ubiquitous educational tool in medical imaging.


a
L
14 mGy
CTDIvol(L)
10 mGy
75 kg
b
L½
7 mGy
CTDIvol(L)
10 mGy
75 kg
c
9 mGy
CTDIvol(L)
10 mGy
100 kg
d
18 mGy
CTDIvol(L)
10 mGy
45 kg

◀**Fig. 5** **a** In a normal-sized patient, F1 is the ratio of organ dose for "long" scans to the corresponding $CTDI_{vol}(L)$. In this example, use of a $CTDI_{vol}(L)$ of 10 mGy in a abdomen CT scan results in a liver dose of 14 mGy. **b** F2 takes into account the scan length where this example shows that reducing the scan length to one half of an abdominal CT scan is expected to reduce the liver dose to approximately 7 mGy. This can be compared to (**a**) which shows that a full length abdominal CT scan results in a liver dose of 14 mGy. **c** Example of F3, where an abdominal CT scan performed using $CTDI_{vol}(L)$ of 10 mGy results in a liver dose of 9 mGy in a large (100 kg) patient compared to 14 mGy in a standard-sized patient (**a**). **d** Another example of F3, where an abdominal CT scan performed using $CTDI_{vol}(L)$ of 10 mGy results in a liver dose of 18 mGy in a small (45 kg) patient, compared to 14 mGy in a standard-sized patient (**a**)

Effective Doses

Figure 6 shows the idea underlying the effective dose concept whereby any non-uniform deposition of energy in a patient undergoing any radiological examination such as a chest CT scan (Fig. 6a) can be equated to a corresponding uniform pattern of energy deposition (Fig. 6b). For a given patient, the patient detriment for the non-uniform and uniform patterns of energy deposition would be taken to be (very) approximately "similar". The effective dose is thus a practical tool to deal with non-uniform energy deposition in radiological examination, which permits any examination in a given patient to be compared with any other type of radiological examination in the same patient. Table 6 provides the organ weighting factors that are currently recommended for use by the ICRP in Publication 103 for computing the effective dose [67]. The remainder weighting factor is to be applied to remainder tissues consisting of the adrenals, extrathoracic region, gall bladder, heart, kidneys, lymphatic nodes, muscle, oral mucosa, pancreas, prostate (males only), small intestines, spleen, thymus, and uterus/cervix (females only). It is important to note that the ICRP explicitly states that any such effective dose estimates are not used to estimate patient risks [67].

The effective dose for a normal-sized adult undergoing an abdominal/pelvic CT scan (DLP 700 mGy-cm) is about 10 mSv [68, 69]. This effective dose can be directly compared to other examples of radiological examinations as shown in Table 7. An abdominal/pelvic CT scan with an effective dose of 10 mSv is between "moderate" and "high" dose examinations. Effective doses are primarily determined for the purposes of offering patients some quantitative idea of the amount of radiation a patient receives. As explained in the prior section, patient risks should always be obtained directly from organ doses and by taking patient demographics into account.

It can also be helpful to compare patient effective doses (Table 7) with US natural background radiation ($E \sim 3$ mSv/year), or with US regulatory dose limits to

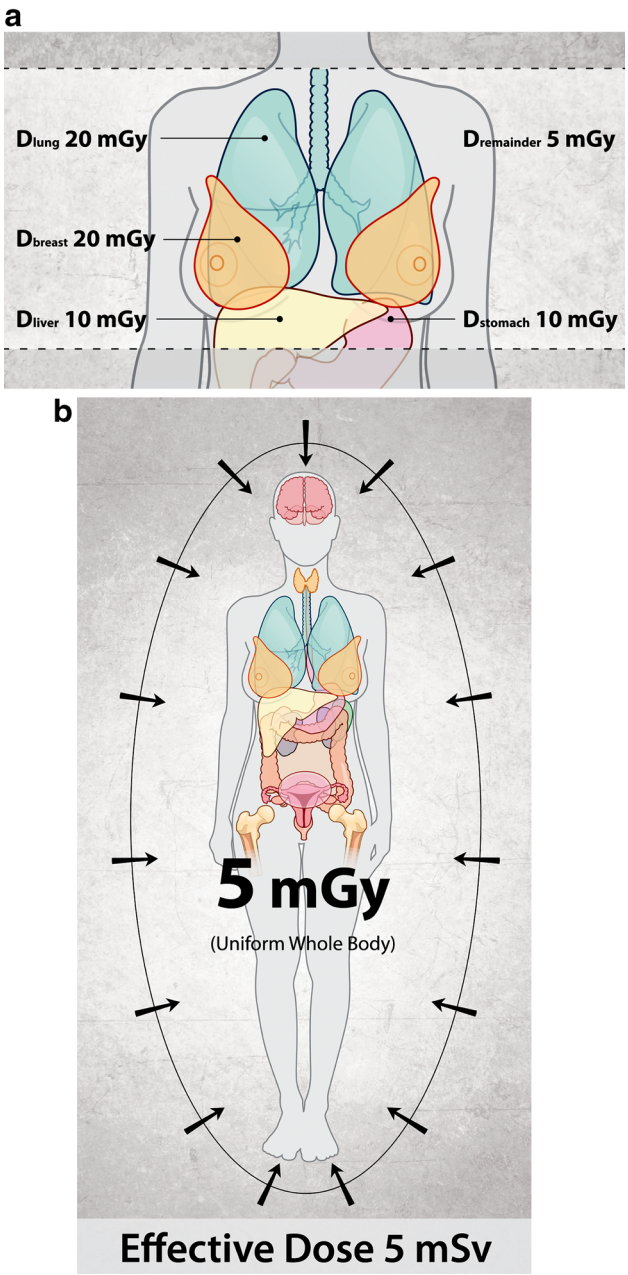


Fig. 6 A graphical illustration of the concept of Effective Dose (E) to deal with non-uniform exposures routinely encountered in Radiological Imaging. In **a**, there is a non-uniform pattern of dose in a patient who has undergone a chest CT examination. In **b** a uniform whole-body dose of 5 mGy has the same detriment in this patient as the chest CT examination, so the effective dose is 5 mSv

individuals who are occupationally exposed to radiation ($E = 50$ mSv/year), or who are members of the public ($E = 1$ mSv/year) [19, 20]. Natural background, which are also expressed in terms of effective dose, can be directly compared to the medical effective dose data shown in

Table 6 Organ weighting factors recommended by the ICRP in Publication 103

Tissue	Weighting factor
Red bone marrow; colon; lung; stomach; breast; remainder[a]	0.12
Gonads	0.08
Bladder; esophagus; liver; thyroid	0.04
Bone surface; brain; salivary glands; skin	0.01

[a] See text for explanation

Table 7 Representative values of patient effective dose in Radiological Imaging

Dose descriptor	Effective dose (mSv)	Radiological examinations
Very low	<0.1	PA chest X-ray; extremity X-ray; DEXA study
Low	0.1–1	Abdomen X-ray; extremity CT; T/L spine X-ray
Moderate	1–10	Chest CT; small bowel series; ^{99m}Tc NM study
High	>10	PET/CT; chest/abdomen/pelvis CT; coronary PCTA

Table 6. It is helpful to consider the effective dose as a kind of currency that provides an indication of "how much" radiation an individual receives. The effective dose is helfpul when explaining how much radiation patients receive in radiological examinations [1, 64•, 66•]. This dose metric is arguably the best dose metric currently available to the medical imaging community, illustrated by a recent NIH funding opportunity entitled *Decreasing Patient Radiation Dose from CT Imaging: Achieving Sub-mSv Studies* [70].

Protecting Patients

Risk Models

For radiation protection practice, there are good reasons for adopting the Linear No-Threshold model, which is depicted graphically in Fig. 7. In the absence of any radiation exposure, it is currently estimated by the American Cancer Society (ACS) that about 40 % of Americans will get cancer in their lifetime [71], and depicted as the background cancer incidence in Fig. 7. A uniform whole-body dose of 100 mGy, corresponding to an effective dose of 100 mSv, is currently estimated to add another 1 % to this background cancer risk. The issue of radiation-induced carcinogenesis remains problematical [63, 67, 72], and the lack of a scientific consensus is elegantly illustrated by the concurrent publication, in the same journal, of radically different perspectives on this topic [73, 74]. Even when radiation risks are accepted as existing, specific quantitative risk estimates can differ by substantial amounts depending on the models used and the exposed populations [63, 75, 76].

As depicted in Fig. 8, the imaging community has to act in the presence of considerable scientific uncertainty. As a

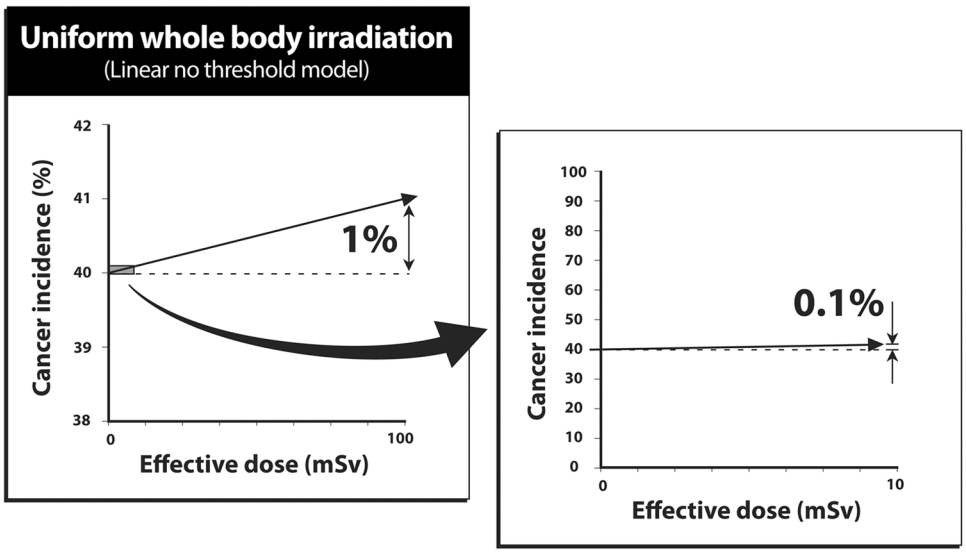


Fig. 7 No-Threshold Model illustrating that there is no threshold dose below which the risk is zero. The additional risk estimate of 0.1 % for a 10 mSv effective dose to an average aged individual can be compared to the background incidence of cancer in the US (40 %), graphically illustrating that current radiation risk estimates are "low"

result, two kinds of error may occur when making a decision regarding the existence of radiation risks in medical imaging. The first error is to assume that risks are real, and subsequently discover they do not exist (Fig. 9a). The second error is to assume that such risks are non-existent, and subsequently discover these risks are in fact real (Fig. 9b). The precautionary principle recommends that we act on the assumption that the risks are real. Assuming that radiation risks are non-existent, and later being proved wrong, would most likely be unacceptable to the medical imaging community.

Adopting the LNT model for radiation protection practice is currently recommended by the most important scientific bodies that review the science of radiation risks. These leading organizations are the International Commission on Radiological Protection (ICRP), the United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR), and the National Academy of Sciences (NAS) committee in the US known as Biological Effects of Ionizing Radiations (BEIR) [63, 67, 72]. All these bodies recommend that for radiation protection purposes, we need to act on the assumption that radiation risks in medical imaging actually exist [77]. It is important to note that this is a judgment call as to how we should proceed, and not a definitive statement regarding the issue of whether such risks are known to truly exist. Any assessment of radiation risks must always take into account the very large uncertainties in current radiation risk estimates [78–80].

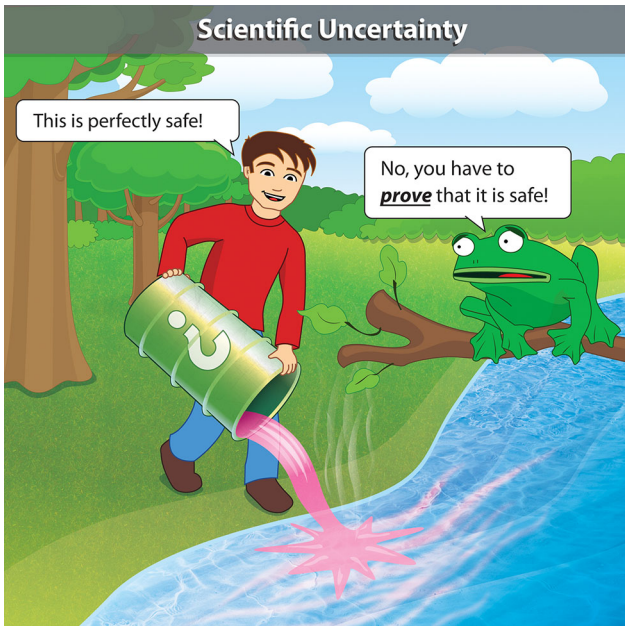


Fig. 8 Precautionary principle illustrating the existence of scientific uncertainty

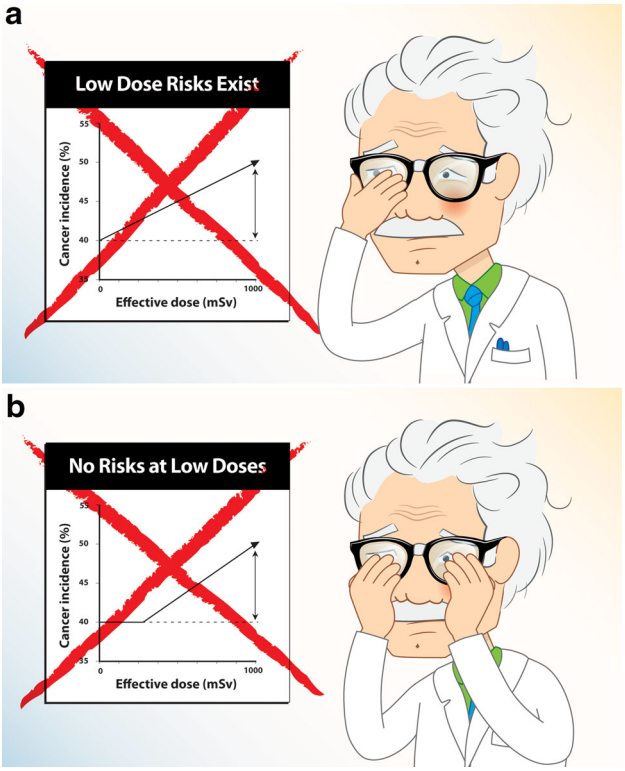


Fig. 9 **a** Illustrates the assumption of the existence of risks at low doses turns out to be erroneous whereas **b** shows the assumption of no risks at low doses turns out to be erroneous. In the presence of scientific uncertainty, it is better that risks are assumed to be real, rather than assumed to be non-existent, which can be summed up in the phrase *Better be Safe than Sorry*

Risk Data

There is now increasing evidence supporting the use of the LNT model in radiation protection practice. A paper by Brenner et al. showed that small but statistically significant risks were observed in a group of A-bomb survivors exposed to relatively low doses (34 mSv) [81•]. Radiation workers in the UK were enrolled in a National Registry of Radiation Workers (NRRW) in the late 1970s to study radiation-related effects from occupational exposures. A recent and third analysis of the NRRW cohort showed risks of leukemia and solid tumors that are consistent with the current risks of studies on A-bomb survivors [82]. These comparisons pertain to leukemia, solid tumors, as well as solid tumors without lung cancer to eliminate the possible confounding effects of smoking in radiation workers. Two important aspects of these exposures were that there were protracted over many decades, and predominantly within the regulatory limits in force in the UK over this time period.

There is also increasing evidence of radiation risks at the low doses encountered in the practice of radiology. Doll and

Wakeford concluded that radiation doses of the order of 10 mGy received by the fetus in utero produce a consequent increase in the risk of childhood cancer [83•]. Children who underwent CT scans in the UK have shown clear evidence of an increase in the incidence of leukemia as well as brain tumors [84]. A recent Australian study showed an increased cancer incidence with increasing number of pediatric CT scans, where the average effective dose was estimated to be as low as 4.5 mSv per scan [85]. There are an additional number of ongoing studies of children who underwent CT examinations throughout the world, and their findings are expected to improve our understanding of this important topic.

Radiologists, who are ultimately responsible for the radiation received by their patients, clearly must be knowledgeable about the magnitude of current radiation risk estimates. Figure 10 shows cancer incidence risks for a uniform whole-body equivalent dose of 10 mSv based on the current BEIR VII risk estimates [63]. The data in Fig. 10 show quantitatively how radiation risks depend on the amount of radiation used, as well as the age and sex of exposed individuals. In males, the age-averaged cancer induction risk was about 0.10 % per 10 mSv, and this average value occurred in individuals aged 20. In females, the age-averaged cancer induction risk was about 0.17 % per 10 mSv, and this average value also occurred in individuals aged 20. Compared to a 20-year old, the cancer induction risk in newborns is approximately a factor 3 higher, and in 70-year olds is approximately a factor of three lower. It is worth repeating that uncertainties in these risk estimates are generally taken to be factors of two to three in both directions [73, 74, 86]. Additional information on these our understanding of radiation risk uncertainties can be obtained from the scientific literature [78–80] .

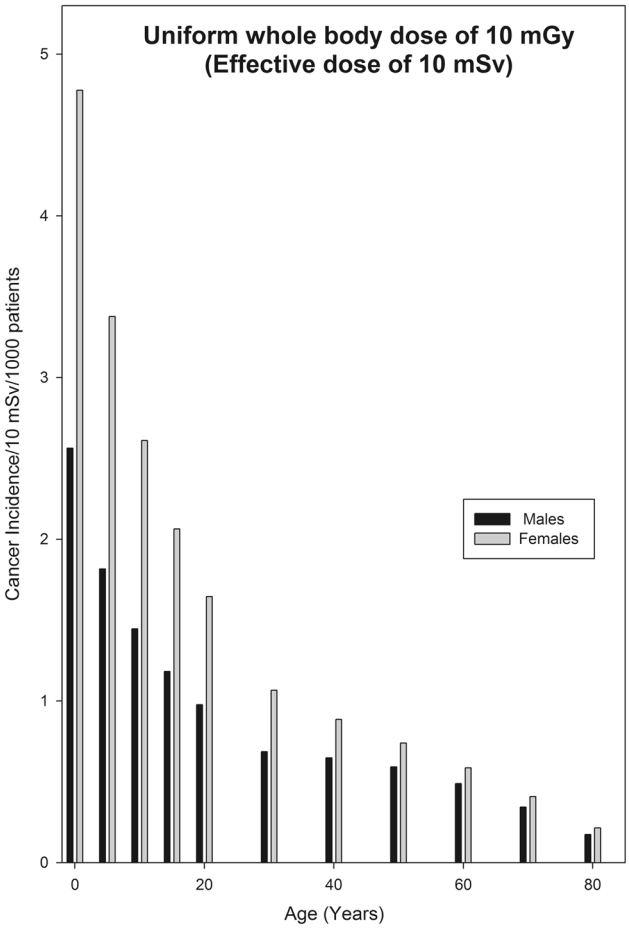


Fig. 10 BEIR VII risk data for cancer induction from uniform body doses of 10 mGy, equivalent to an effective dose of 10 mSv

Protection Practice

When patients require a quantitative understanding of the radiation associated with radiological examinations, it is helpful to use the effective dose (i.e., radiation currency). A given radiological examination can be compared with other types of examination such as chest X-rays, so a chest CT scan is equivalent to about 100 chest X-ray examinations (PA + Lateral). It can also be very helpful to compare patient effective doses with natural background radiation as well as regulatory dose limits for both radiation workers and members of the public. A single PA chest X-ray, with an effective dose of 0.01 mSv, exposes a patient to as much radiation as s/he receives in 1 day from natural background. A chest CT, with an effective dose of 5 mSv, corresponds to about 10 % of the current regulatory annual dose limit to radiation workers in the US.

When it is assumed that radiation risks "exist," as depicted by the data in Fig. 10, patient exposures in radiological examinations require justification [87]. This means that patient benefits need to exceed all examination risks as indicated graphically in Fig. 11. In the United Kingdom, there is a regulatory requirement to formally justify all exposures in all radiological examinations [88]. Radiologists are trained to identify which are appropriate examinations, what kind of diagnostic information might be obtained, as well as the corresponding patient doses and risks. Such individuals therefore have the knowledge to decide whether to proceed with a given examination for any clinical indication [89, 90].

Any assumption of the existence of radiation risks also requires that all unnecessary radiation exposure be eliminated. This aspect of radiation protection is the ALARA principle, meaning As Low As Reasonably Achievable [91•, 92]. Dose reduction is only appropriate when diagnostic information from radiological examinations is not compromised. For any radiological examination, the primary focus should not be on existence of any radiation risks but whether the planned examinations are worthwhile. This requires a mastery of what benefit diagnostic tests

Fig. 11 Any radiological examination requires the patient benefit from the diagnostic information to exceed any risks, including those of radiation exposure. This requires for practitioners to understand the magnitude of (any) radiation risks, including the associated uncertainties, and to understand patient benefits of acquired diagnostic information

bring to the patient, what is known about risks, and also uncertainties in our knowledge of radiation risks.

It is especially important that any communication of radiation risks to patients should always include explicit consideration of the corresponding benefits to the patient. Patients are likely helped by being informed that an exam is worthwhile, and the benefits are believed to outweigh any risk. This is akin to driving an automobile to a Radiology department, where the well-known risks of dying in automobile accident (30,000 fatalities in the US each year) do not stop a patient from undertaking a "worthwhile" activity just because there may be a risk with this activity. It is essential to recognize that there is virtually no activity that is entirely risk free–free. Focusing on radiation risk, without explicit consideration of the corresponding patient benefit(s), is unlikely to be helpful for ensuring appropriate use of X-rays in medical imaging [93, 94]. Patients generally benefit substantially from diagnostic data that radiological examinations provide, and the role of the imaging community role is to ensure that only indicated examinations are performed, and all unnecessary radiation is eliminated (i.e., ALARA) [95].

Conclusions

The CT radiation incident on the patient should be specified by the $CTDI_{vol}$, Dose Length Product, and the beam quality controlled by the X-ray tube voltage (kV). $CTDI_{vol}$ is related to the image mottle, and Dose Length Product is a measure of the total amount of radiation used to perform the radiological examination. Beam quality will influence the penetration through the patient, as well as the resultant image contrast. Operators should select CT radiation quantities by taking account patient physical characteristics and the imaging task at hand. Knowledge of the radiation incident on a patient enables determination of organ and effective doses, as well as the corresponding radiation risks. To protect patients, all exposures need to be justified by a net patient benefit. Radiation exposure not needed for a satisfactory diagnostic performance should be eliminated (ALARA).

Acknowledgments The author would like to express his gratitude to R. Brad Abrahams DO for help with the artwork, and G. Don Frey PhD, Kent M Ogden PhD, and Sameer V. Tipnis PhD for useful discussions.

Compliance with Ethical Guidelines

Conflict of Interest Dr. Walter Huda is president of Hudas Physics in Medicine.

Human and Animal Rights and Informed Consent This article does not contain any studies with human or animal subjects performed by any of the authors.

References

Papers of particular interest, published recently, have been highlighted as:
- • Of importance
- •• Of major importance

1. Mettler FA Jr, Bhargavan M, Faulkner K, et al. Radiologic and nuclear medicine studies in the United States and worldwide: frequency, radiation dose, and comparison with other radiation sources: 1950–2007. Radiology. 2009;253:520–31.
2. Mettler FA Jr, Thomadsen BR, Bhargavan M, et al. Medical radiation exposure in the U.S. in 2006: preliminary results. Health Phys. 2008;95:502–7.
3. ••NCRP. Ionizing radiation exposure of the population of the United States. NCRP Report 160. National Council on Radiation Protection and Measurements. 2009. *Definitive report describing natural background and man made sources of exposure to the population of the United States in 2006.*
4. Lang K, Huang H, Lee DW, Federico V, Menzin J. National trends in advanced outpatient diagnostic imaging utilization: an analysis of the medical expenditure panel survey, 2000–2009. BMC Med Imaging. 2013;13:40.
5. Boone JM, Hendee WR, McNitt-Gray MF, Seltzer SE. Radiation exposure from CT scans: how to close our knowledge gaps, monitor and safeguard exposure–proceedings and recommendations of the Radiation Dose Summit, sponsored by NIBIB, February 24–25, 2011. Radiology. 2012;265:544–54.

6. Hricak H, Brenner DJ, Adelstein SJ, et al. Managing radiation use in medical imaging: a multifaceted challenge. Radiology. 2011;258: 889–905.
7. McCollough CH, Guimaraes L, Fletcher JG. In defense of body CT. AJR Am J Roentgenol. 2009;193:28–39.
8. McNitt-Gray MF. AAPM/RSNA Physics Tutorial for Residents: Topics in CT. Radiation dose in CT. Radiographics. 2002;22: 1541–53.
9. Elojeimy S, Tipnis S, Huda W. Relationship between radiographic techniques (kilovolt and milliampere-second) and CTDI(VOL). Radiat Prot Dosimetry. 2010;141:43–9.
10. Hausleiter J, Meyer T, Hermann F, et al. Estimated radiation dose associated with cardiac CT angiography. JAMA. 2009;301: 500–7.
11. Amacker NA, Mader C, Alkadhi H, Leschka S, Frauenfelder T. Routine chest and abdominal high-pitch CT: an alternative low dose protocol with preserved image quality. Eur J Radiol. 2012;81:e392–7.
12. Huda W, Mettler FA. Volume CT dose index and dose-length product displayed during CT: what good are they? Radiology. 2011;258:236–42.
13. Huda W, Sterzik A, Tipnis S. X-ray beam filtration, dosimetry phantom size and CT patient dose conversion factors. Phys Med Biol. 2010;55:551–61.
14. McCollough CH, Leng S, Yu L, Cody DD, Boone JM, McNitt-Gray MF. CT dose index and patient dose: they are not the same thing. Radiology. 2011;259:311–6.
15. Ogden K, Huda W, Scalzetti EM, Roskopf ML. Patient size and X-ray transmission in body CT. Health Phys. 2004;86:397–405.
16. Primak AN, McCollough CH, Bruesewitz MR, Zhang J, Fletcher JG. Relationship between noise, dose, and pitch in cardiac multidetector row CT. Radiographics. 2006;26:1785–94.
17. Ende JF, Huda W, Ros PR, Litwiller AL. Image mottle in abdominal CT. Invest Radiol. 1999;34:282–6.
18. Kruger RL, McCollough CH, Zink FE. Measurement of half-value layer in X-ray CT: a comparison of two noninvasive techniques. Med Phys. 2000;27:1915–9.
19. Bushberg JT, Seibert JA, Leidholdt EM, Boone JM. The essential physics of medical imaging. 3rd ed. Lippincott: Williams & Wilkins; 2012.
20. Huda W. Review of radiologic physics. 3rd ed. Philadelphia: Lippincott Williams & Wilkins; 2010.
21. Hubbell JH. Photon Cross Sections, Attenuation Coefficients, and Energy Absorption Coefficients from 10 keV to 100 GeV. National Standards Reference Data Service (NSRDS): National Bureau Standards (NBS) 1969; NSRDS-NBS 29:70.
22. Boone JM, Geraghty EM, Seibert JA, Wootton-Gorges SL. Dose reduction in pediatric CT: a rational approach. Radiology. 2003;228:352–60.
23. ••Yu L, Bruesewitz MR, Thomas KB, Fletcher JG, Kofler JM, McCollough CH. Optimal tube potential for radiation dose reduction in pediatric CT: principles, clinical implementations, and pitfalls. Radiographics. 2011;31:835–48. *Excellent overview of how to optimize pediatric CT protocols, and the results obtained at the Mayo Clinic (Rochester, MN) resulting from this exercise.*
24. Modica MJ, Kanal KM, Gunn ML. The obese emergency patient: imaging challenges and solutions. Radiographics. 2011;31:811–23.
25. Huda W, Lieberman KA, Chang J, Roskopf ML. Patient size and X-ray technique factors in head computed tomography examinations. II. Image quality. Medical physics. 2004;31:595–601.
26. Huda W, Ravenel JG, Scalzetti EM. How do radiographic techniques affect image quality and patient doses in CT? Semin Ultrasound CT MR. 2002;23:411–22.
27. He W, Marzolf SA, Tipnis S, Huda W. Scan region and organ doses in computed tomography. Radiat Prot Dosim. 2012;148: 444–51.
28. Huda W. CT dose metrics. Radiology. 2013;267:964–5.
29. Huda W, Sterzik A, Tipnis S, Schoepf UJ. Organ doses to adult patients for chest CT. Med Phys. 2010;37:842–7.
30. •Huda W, He W. Estimating cancer risks to adults undergoing body CT examinations. Radiat Prot Dosim. 2012;150:168–79. *A practical guide of how to estimate adult organ doses, and the corresponding cancer risks, in body CT. This approach accounts of all key factors, including type of CT scan, patient and sex, as well as patient size.*
31. Huda W, Scalzetti EM, Levin G. Technique factors and image quality as functions of patient weight at abdominal CT. Radiology. 2000;217:430–5.
32. ••ICRU. Report 54 Medical Imaging: The Assessment of Image Quality. Bethesda MD, 1996. *The defitive text on Image Quality in Medical Imaging.*
33. Smith EA, Dillman JR, Goodsitt MM, Christodoulou EG, Keshavarzi N, Strouse PJ. Model-based iterative reconstruction: effect on patient radiation dose and image quality in pediatric body CT. Radiology. 2014;270:526–34.
34. •McCollough CH. Automatic exposure control in CT: are we done yet? Radiology. 2005;237:755–6. *Good guide to all of the important issues associated with Automatic Exposure Control (AEC) in CT imaging.*
35. Arch ME, Frush DP. Pediatric body MDCT: a 5-year follow-up survey of scanning parameters used by pediatric radiologists. AJR Am J Roentgenol. 2008;191:611–7.
36. Chodick G, Ronckers C, Ron E, Shalev V. The utilization of pediatric computed tomography in a large Israeli Health Maintenance Organization. Pediatr Radiol. 2006;36:485–90.
37. Paterson A, Frush DP, Donnelly LF. Helical CT of the body: are settings adjusted for pediatric patients? AJR Am J Roentgenol. 2001;176:297–301.
38. Brenner D, Elliston C, Hall E, Berdon W. Estimated risks of radiation-induced fatal cancer from pediatric CT. AJR Am J Roentgenol. 2001;176:289–96.
39. Brody AS, Frush DP, Huda W, Brent RL. Radiation risk to children from computed tomography. Pediatrics. 2007;120:677–82.
40. Huda W, Vance A. Patient radiation doses from adult and pediatric CT. AJR Am J Roentgenol. 2007;188:540–6.
41. Khursheed A, Hillier MC, Shrimpton PC, Wall BF. Influence of patient age on normalized effective doses calculated for CT examinations. Br J Radiol. 2002;75:819–30.
42. Huda W, Lieberman KA, Chang J, Roskopf ML. Patient size and X-ray technique factors in head computed tomography examinations. I. Radiation doses. Med Phys. 2004;31:588–94.
43. Schiff GD, Kim S, Abrams R, et al. Diagnosing diagnosis errors: lessons from a Multi-institutional Collaborative Project. In: Henriksen K, Battles JB, Marks ES, Lewin DI, editors. Advances in patient safety: from research to implementation (Volume 2: Concepts and Methodology). Rockville: Agency for Healthcare Research & Quality; 2005.
44. 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: 2423–42.
45. McCollough C, Branham T, Herlihy V, et al. Diagnostic reference levels from the ACR CT Accreditation Program. J Am Coll. Radiol. 2011;8:795–803.
46. Martin CJ, Huda W. Intercomparison of patient CTDI surveys in three countries. Radiat Prot Dosimetry. 2013;153:431–40.
47. ••Huda W, Randazzo W, Tipnis S, Frey GD, Mah E. Embryo dose estimates in body CT. AJR Am J Roentgenol. 2010;194:874–80. *A practical hands on guide for estimating embryo doses in CT that accounts for the amount of radiation used, the type of CT scanner, scan length and patient size.*

48. Chang KJ, Caovan DB, Grand DJ, Huda W, Mayo-Smith WW. Reducing radiation dose at CT colonography: decreasing tube voltage to 100 kVp. Radiology. 2013;266:791–800.
49. Israel GM, Cicchiello L, Brink J, Huda W. Patient size and radiation exposure in thoracic, pelvic, and abdominal CT examinations performed with automatic exposure control. AJR Am J Roentgenol. 2010;195:1342–6.
50. ACR. ACR–SPR Practice Guideline for imaging pregnant or potentially pregnant adolescents and women with ionizing radiation. Reston: Society of Nuclear Medicine; 2013.
51. Pregnancy and medical radiation. *Annals of the ICRP* 2000; 30:iii–viii, 1–43.
52. HPA. Protection of Pregnant Patients during Diagnostic Medical Exposures to Ionizing Radiation. Advice from the Health Protection Agency, The Royal College of Radiologists, and the College of Radiographers. In. RCE-9, Documents of the HPA, 2009.
53. DeMarco JJ, Cagnon CH, Cody DD, et al. A Monte Carlo based method to estimate radiation dose from multidetector CT (MDCT): cylindrical and anthropomorphic phantoms. Phys Med Biol. 2005;50:3989–4004.
54. Huda W, Ogden KM, Lavallee RL, Roskopf ML, Scalzetti EM. In-patient to isocenter KERMA ratios in CT. Med Phys. 2011;38:5362–9.
55. Huda W, Ogden KM, Lavallee RL, Roskopf ML, Scalzetti EM. KERMA ratios in pediatric CT dosimetry. Pediatr Radiol. 2012;42:527–35.
56. AAPM. Size Specific Dose Estimates (SSDE) in Pediatric and Adult Body CT examinations. *Report No. 204* 2011.
57. Huda W, Rowlett WT, Schoepf UJ. Radiation dose at cardiac computed tomography: facts and fiction. J Thorac Imaging. 2010;25:204–12.
58. Li X, Samei E, Segars WP, Sturgeon GM, Colsher JG, Frush DP. Patient-specific dose estimation for pediatric chest CT. Med Phys. 2008;35:5821–8.
59. Li X, Samei E, Segars WP, et al. Patient-specific radiation dose and cancer risk estimation in CT: part II. Application to patients. Med Phys. 2011;38:408–19.
60. http://www.impactscan.org/ctdosimetry.htm. Accessed 1 Nov 2014.
61. Boone JM, Cooper VN 3rd, Nemzek WR, McGahan JP, Seibert JA. Monte Carlo assessment of computed tomography dose to tissue adjacent to the scanned volume. Med Phys. 2000;27: 2393–407.
62. Huda W, Schoepf UJ, Abro JA, Mah E, Costello P. Radiation-related cancer risks in a clinical patient population undergoing cardiac CT. AJR Am J Roentgenol. 2011;196:W159–65.
63. Health Risks from Exposures to Low Levels of Ionizing Radiation;. BEIR VII Phase 2. National Academies Press, Washington, DC; 2006:10.
64. •Borras C, Huda W, Orton CG. Point/counterpoint. The use of effective dose for medical procedures is inappropriate. Med. Phys. 2010;37:3497–500. *Raises the important issues of using the effective dose in medical imaging (for and against).*
65. Brenner D, Huda W. Effective dose: a useful concept in diagnostic radiology. Radiat Prot Dosimetry. 2008;128:503–8.
66. •Mettler FA Jr, Huda W, Yoshizumi TT, Mahesh M. Effective doses in radiology and diagnostic nuclear medicine: a catalog. Radiology. 2008;248:254–63. *Definitive source of data on patient exposures in all areas of medical imaging including radiography, fluoroscopy, Interventional Radiology, and Nuclear Medicine.*
67. The 2007 Recommendations of the International Commission on Radiological Protection. ICRP publication 103. *Annals of the ICRP* 2007; 37:58.
68. Deak PD, Smal Y, Kalender WA. Multisection CT protocols: sex- and age-specific conversion factors used to determine effective dose from dose-length product. Radiology. 2010;257:158–66.
69. Huda W, Magill D, He W. CT effective dose per dose length product using ICRP 103 weighting factors. Med Phys. 2011;38:1261–5.
70. http://grants.nih.gov/grants/guide/pa-files/PAR-12-206.html. Accessed 1 Nov 2014.
71. http://www.cancer.org/. Accessed 1 Nov 2014.
72. Effects of Ionizing Radiation, UNSCEAR 2006 Report (Volume 1). United Nations Scientific Effects of Atomic Radiation, United Nations, New York; 2008:137.
73. Brenner DJ, Hall EJ. Cancer risks from CT scans: now we have data, what next? Radiology. 2012;265:330–1.
74. Hendee WR, O'Connor MK. Radiation risks of medical imaging: separating fact from fantasy. Radiology. 2012;264:312–21.
75. Ivanov VK, Tsyb AF, Mettler FA, Menyaylo AN, Kashcheev VV. Methodology for estimating cancer risks of diagnostic medical exposure: with an example of the risks associated with computed tomography. Health Phys. 2012;103:732–9.
76. Mobbs SF, Muirhead CR, Harrison JD. Risks from Ionizing Radiation, Radiation Protection Division of the Health Protection Agency. HPA-RPD-066 2010.
77. Huda W. Radiation risks: what is to be done? *AJR Am. J. Roentgenol.* 2015 (in press).
78. NCRP. Radiation Risk Estimates for Radiation Protection. *National Council on Radiation Protection and Measurements (NCRP)* 1993; Report 115.
79. NCRP. Uncertainties in Fatal Cancer Risk Estimates Used in Radiation Protection. *National Council on Radiation Protection and Measurements (NCRP)* 1997; Report 126.
80. NCRP. Evaluation of the Linear-Nothreshold Dose-Response Model for Ionizing Radiation. *National Council on Radiation Protection and Measurements (NCRP)* 2001; Report 136.
81. •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 USA. 2003;100:13761–6. *Written by the leading radiobiologists and epidemiologists, this is essential reading for anyone interested in the topic of radiation risks.*
82. Muirhead CR, O'Hagan JA, Haylock RG, et al. Mortality and cancer incidence following occupational radiation exposure: third analysis of the National Registry for Radiation Workers. Br J Cancer. 2009;100:206–12.
83. •Doll R, Wakeford R. Risk of childhood cancer from fetal irradiation. Br J Radiol. 1997;70:130–9. *Summarizes what is currently known about a very important study performed over 50 years ago to study the effects of radiation in pregnant patients.*
84. Pearce MS, Salotti JA, Little MP, et al. Radiation exposure from CT scans in childhood and subsequent risk of leukaemia and brain tumours: a retrospective cohort study. Lancet. 2012;380:499–505.
85. Mathews JD, Forsythe AV, Brady Z, et al. Cancer risk in 680,000 people exposed to computed tomography scans in childhood or adolescence: data linkage study of 11 million Australians. BMJ. 2013;346:f2360.
86. Brenner DJ, Hall EJ. Computed tomography–an increasing source of radiation exposure. N Engl J Med. 2007;357:2277–84.
87. International Atomic Energy Agency (IAEA). Report of a consultation on justification of patient exposures in medical imaging. Radiat Prot Dosimetry. 2009;135:137–44.
88. Malone J, Guleria R, Craven C, et al. Justification of diagnostic medical exposures: some practical issues. Report of an International Atomic Energy Agency Consultation. Br J Radiol. 2012;85:523–38.
89. Larson DB, Rader SB, Forman HP, Fenton LZ. Informing parents about CT radiation exposure in children: it's OK to tell them. AJR Am J Roentgenol. 2007;189:271–5.
90. Lee CI, Haims AH, Monico EP, Brink JA, Forman HP. Diagnostic CT scans: assessment of patient, physician, and radiologist

awareness of radiation dose and possible risks. Radiology. 2004;231:393–8.
91. •Semelka RC, Armao DM, Elias J Jr, Huda W. Imaging strategies to reduce the risk of radiation in CT studies, including selective substitution with MRI. J Magn Reson Imaging. 2007;25:900–9. *Overview of ways of reducing patient CT doses and risks.*
92. Sierzenski PR, Linton OW, Amis ES Jr, et al. Applications of justification and optimization in medical imaging: examples of clinical guidance for computed tomography use in emergency medicine. J Am Coll Radiol. 2014;11:36–44.
93. Lazarus E, Mayo-Smith WW, Mainiero MB, Spencer PK. CT in the evaluation of nontraumatic abdominal pain in pregnant women. Radiology. 2007;244:784–90.
94. Sadigh G, Kelly AM, Cronin P. Challenges, controversies, and hot topics in pulmonary embolism imaging. AJR Am J Roentgenol. 2011;196:497–515.
95. ICRP Publication 105. Radiation protection in medicine. *Annals of the ICRP* 2007; 37:1–63.