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Abbreviations:
DI = dose index
DLP = dose-length product

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Radiation Exposure at Chest CT: A Statement of the Fleischner Society[1]

The introduction of helical single–detector row computed tomography (CT) and, more recently, multi–detector row CT has greatly increased the clinical indications for CT. Correspondingly, CT examinations now account for greater than one-half of the radiation dose due to medical procedures in the population of North America. The level of CT radiation dose, especially in the pediatric population, is of concern to radiologists, medical physicists, government regulators, and the media. This review addresses this problem with particular reference to radiation dose in chest CT. Specifically it outlines the topics of measurement units used to quantify radiation exposure, factors affecting CT scanner dose efficiency, scanner settings that determine the administered radiation dose, and radiation dose reduction in chest CT. A table of reference dose values is provided. Given the wide variation documented in chest CT radiation exposure, the authors suggest that reference standards be promoted to minimize excessive CT radiation exposure. In addition, further research into the complex relationship between radiation exposure, image quality, and diagnostic accuracy should be encouraged in order to establish the minimum radiation dose necessary to provide adequate diagnostic information for standard clinical questions.

The introduction of helical, or spiral, computed tomography (CT) in the late 1980s revolutionized diagnostic medical imaging (1–4). Single–detector row CT scanners and, more recently, multi–detector row CT scanners markedly increased the clinical indications for CT. As a result there has been a considerable increase in the number of CT examinations performed and in the average scanned volume obtained per examination. Studies in the United States and the United Kingdom have shown an approximately twofold increase in the number of CT examinations performed between the late 1980s and the late 1990s (5–7). In a study performed at an academic institution in the United States, Mettler et al (5) found that CT accounted for 11.1% of diagnostic radiologic procedures in 1999 compared with 6.1% in 1990. The results of this study showed that CT accounted for about 67% of the total effective radiation dose from diagnostic radiology in 1999. The results of a radiation dose survey performed in the United Kingdom in 1989 showed that CT accounted for 2% of all radiologic examinations and 20% of the effective dose (6). A second survey, in 1995, showed that CT accounted for 4% of all radiologic examinations (7) and 40% of the effective dose, which is a twofold increase. In the study by Mettler et al (5), 11% of the CT examinations were performed in the pediatric population, which was a higher percentage than previously estimated.

The increase in population radiation exposure from CT, particularly in children, has been of concern to radiologists, medical physicists, government regulators, and the media (8). The suggestion that excessive radiation doses are being prescribed for CT has appropriately aroused the attention of the radiologic community (9,10). Radiologists and medical physicists must be attentive to their responsibility to maintain an appropriate balance between diagnostic image quality and radiation dose (11). It has been suggested that in the rapidly evolving field of multi–detector row CT, issues of radiation dose may have been diminished in the quest for increased image quality, diagnostic accuracy, and new imaging techniques (12).

In this communication we outline the determinants of CT radiation dose and discuss the interaction between image quality and radiation dose. Specifically we outline these topics: *(a)* measurement units used to quantify radiation exposure, *(b)* parameters that affect CT radiation dose and efficiency, and *(c)* advances in dose reduction in chest CT. A complete review of radiation dosimetry and bioeffects is beyond the scope of this article, and interested readers are referred to more complete works in these areas (12–16).

RADIATION DOSE MEASUREMENT

There are many methods currently in use for quantifying ionizing radiation (Table 1) (17). The fact that several methods exist attests to the complexity of this issue.

The simplest parameter, radiation exposure, is determined by measuring ionization in air caused by the x-ray beam. The measurement unit is coulombs per kilogram (abbreviation, C/kg). It has limited clinical value, as it does not take into account the area irradiated, the penetrating power of the radiation, or the radiation sensitivity of the irradiated organs. From radiation exposure we can calculate the skin entrance dose, which is important when examining deterministic effects such as skin erythema. Although deterministic effects are not encountered in routine CT, they are of potential concern in CT fluoroscopy (12).

Absorbed dose is determined by measuring the energy absorbed per unit mass within an object. The measurement unit is the gray (abbreviation, Gy). Unlike radiation exposure, the gray is dependent on the composition of the object or subject placed in the radiation beam. However, absorbed dose does not account for the differing radiation sensitivity of organs, and it cannot provide a whole-body risk estimate or be used to facilitate comparisons between examinations in different parts of the body. Equivalent dose is a modification of absorbed dose that incorporates weighting factors to account for the different biologic effect of various sources of radiation. For x rays, the radiation weighting factor is 1 and the equivalent dose has the same numerical value as absorbed dose (18).

Effective dose is a measurement that estimates the whole-body dose that would be required to produce the same stochastic risk as the partial-body dose that was actually delivered in a localized radiologic procedure. Effective dose is useful because it allows comparison to other types of radiation exposure such as whole-body radiation exposure secondary to natural background radiation. Effective dose is calculated by summing the absorbed doses to individual organs weighted for their radiation sensitivity (18). The measurement unit is the sievert (abbreviation, Sv). Effective dose has limitations because it represents the radiation detriment for the general population or the specific population of radiation workers and may not be appropriate for many patient populations (19,20). However, it is currently the best measurement available.

TABLE 1
Methods of Quantifying Ionizing Radiation

Method	Conventional Units	International System of Units, or SI
Radiation exposure	roentgens (R)	coulombs per kilogram (C/kg)
Absorbed dose	rads (rad)	grays (Gy)
Equivalent dose	rems	sieverts (Sv)[†]
Effective dose	Effective dose equivalent (Sv)*	sieverts (Sv)[‡]

Note.—Abbreviations of the units of measure are in parentheses.
* 1977 tissue-weighting factors.
† D multiplied by ICRP radiation weighting factor w_R. The w_R for x rays is 1.
‡ 1990 tissue-weighting factors.

Effective dose can be calculated for chest CT by using dose distributions precalculated for specific CT scanner geometry and beam quality (7,20–23). These precalculated distributions can be individualized for the CT technical parameters by entering specific tube current, tube voltage, scanned volume, and pitch values. It must be noted that the calculated effective dose values are for reference subjects or phantoms, not for specific patients. Once the effective dose has been calculated, risk estimates for stochastic effects can be produced by using a linear extrapolation of radiation exposure data from Japanese atomic bomb survivors (18,24,25). While the stochastic risk depends on such factors as nationality and age at exposure, the International Commission on Radiological Protection, or ICRP, has recommended the use of a conservative risk of 50 additional fatal cancers induced per million people of the general population exposed to 1 mSv of medical radiation (18). The assessment of stochastic risk is discussed in further detail in ICRP report 60 (18).

Since it is not actually possible to measure the absorbed dose inside a patient, it is not feasible to calculate the exact effective dose for each patient examination. However, it is possible to make a good estimate of the dose and thus the effective dose. In a clinical setting, it would be helpful to be able to estimate the effective dose before the CT examination. This has been made possible recently by the availability on the CT scanners of data derived from measurements made in head and body phantoms. These data are shown on the scanners as the CT dose index (DI) and dose-length product (DLP) and can be used to calculate an effective dose in a reference subject (Fig 1).

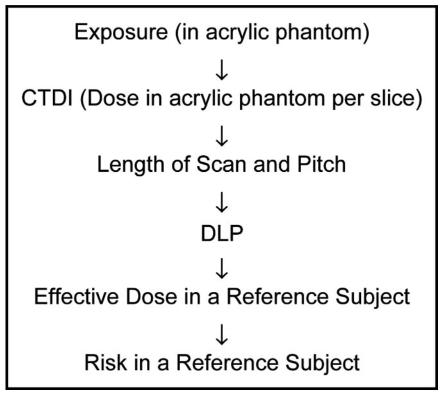

Figure 1. Diagram shows algorithm for the estimation of radiation exposure risk from CT.

The CT DI measurement is made in a plastic phantom designed to simulate either a head (16-cm diameter) or a body (32-cm diameter). To account for imperfect collimation of the x-ray beam and radiation scattered from neighboring scan sections, CT DI measurements are integrated over a 100-mm length. Measurements are made in the center of the phantom and close to the periphery by using an ionization chamber. Beam attenuation causes the dose inside the phantom to change with depth. Therefore, a weighted CT DI is calculated by adding together one-third of the central values and two-thirds of the peripheral values. The DLP depends on the weighted CT DI, the pitch, and the length of the scan. The manufacturer loads tables within the scanner software that provide the weighted CT DI value for the scan parameters specified. The start and end locations of the helical CT study are selected

TABLE 2
Normalized Effective Dose Coefficients

Body Part	Effective Dose Coefficients ($mSv \cdot mGy^{-1} \cdot cm^{-1}$)
Head	0.0021
Neck	0.0048
Chest	0.014
Abdomen	0.012
Pelvis	0.016

Note.—Reprinted, with permission, from reference 26.

from the localizing image taken before the study and are used to calculate the length of the study. A DLP for that specific CT examination can be calculated by using these values. Table 2 provides general conversion coefficients from DLP to effective dose for typical regions of the body (26). Once the effective dose is calculated, the risk of cancer induction can be estimated by using the population average risk of 50 induced cancers per millisieverts of effective dose per million people exposed (18).

Conversion of effective dose or DLP calculations to radiation risk can be performed for the pediatric population as it can for the adult population (27–29). It must be noted that there is a pronounced age effect on the risk of cancer from ionizing radiation (30,31). Children may be an order of magnitude more sensitive than adults to the risk of cancer induction from the same effective dose of ionizing radiation (Fig 2). This arises from the fact that they have more time to express the cancer and the fact that they have more rapidly dividing cells than adults have.

CT RADIATION EXPOSURE

The relatively high radiation dose associated with CT results from two properties of the technique (32). First, unlike analogue film radiography, in which the image acquisition and display are both reliant on film, CT is a digital technique in which image acquisition and display can be independently manipulated. Therefore, when CT radiation dose is excessive the image does not become too dark (as it does in film radiography) but instead improves because of decreased image noise (33). Second, visualization of image noise is enhanced by the ability to map the entire gray scale onto selected segments of the CT number scale. As a result, image degradation due to quantum noise (mottle) is easily visible and may interfere with image analysis. As a result of these two effects and the overwhelming drive of radiologists for the very best image quality to support high levels of diagnostic accuracy, CT images are often obtained with high radiation exposure to the patient, which may not be recognized by the radiologist.

In the early 1990s, concern was raised regarding radiation dose in chest CT (34–36). Patient radiation-dose surveys showed (6,7,37–40) wide variations in radiation exposure between different sites and equipment. These CT data demonstrated that greater consideration needed to be given to optimizing chest CT exposures. However, the current data concerning CT radiation dose indicate that insufficient progress has been made.

SCANNER RADIATION EFFICIENCY

Several physical aspects of CT scanners result in wasted radiation dose. These include the shielding effect of the collimator between the patient and the detector, imperfect collimation of the x-ray beam, and movement of the x-ray focal spot. The sum of all these effects is measured by the geometric efficiency of the CT scanner.

Single–detector row CT scanners, with their wide single–detector row configuration, have higher geometric efficiency than multi–detector row scanners. The decreased geometric efficiency of multi–detector row CT scanners arises from three major factors: the gaps between detector elements in the array, the effect of focal spot penumbra, and the motion of the focal spot. Since the focal spot of the x-ray tube is not a point, the collimator cannot perfectly collimate the beam. Therefore the edge of the beam, or penumbra, has spatially varying x-ray intensity. In single–detector row helical CT scanners this portion of the beam can be detected and used in the reconstruction process. With multi–detector row CT scanners, however, use of the penumbra would result in different readings from detectors in this region compared with those in the central, or umbra, portion of the beam. Therefore the active detectors in multi–detector row CT scanners measure only the umbra of the x-ray beam. Radiation in the penumbra falls on inactive detectors and is discarded, although it contributes to patient radiation dose. In addition, thermal and mechanical stresses within the x-ray tube cause the focal spot to move. As a result, the x-ray beam wanders slightly across the detector array during CT data acquisition. Widening the x-ray beam to compensate for the penumbra and focal spot motion leads to a decrease in the geometric efficiency and an increase in the radiation dose. Because the penumbra is a fixed size, its effect is greatest on four-section CT scanners operating with thin-section collimation. The effect is progressively less severe with eight-, 16-, and 32-section multi–detector row CT scanners. Manufacturers have devised beam-tracking systems to stabilize the position of the x-ray beam and thereby minimize the radiation-wasting effect of focal spot motion (41).

Scattered radiation is formed by the interaction of the primary beam with the body of the patient. Scattered radiation exits the body in all directions, and if detected it reduces contrast and may generate artifacts. In plain chest radiography 50%–90% of film darkening (depending on the technique chosen) is due to scattered radiation, which contributes to the low soft-tissue contrast of this technique (42). The extensive collimation at CT reverses this ratio, and 90% of detected x rays are primary image photons. This partially accounts for the improved soft-tissue contrast at CT.

Because of scatter and imperfect collimation, the radiation intensity profile does not fall to zero at the edge of the nominal section width. It has been shown by using a single-section CT scanner that contiguous sections generate a peak radiation dose approximately 50% greater than that of a single CT section (10-mm collimation, 10-mm table increment, measured at the surface of a 15-cm head phantom) (43). The increase in radiation dose associated with multiple, adjacent CT sections has been measured and is characterized by the multiple-scan average dose, or MSAD, parameter (44,45). Helical CT scanning with a pitch of 1 results in a dose distribution that is essentially equivalent to that of contiguous single–detector row CT imaging (46). Overlapping sections or helical CT scanning with a pitch of less than 1 can result in even higher doses if techniques are not adjusted. Radiation dose can be reduced if gaps are introduced between scanned sections (47,48). However, diagnostic information can be lost by using section gaps because only a portion of the chest is imaged. For this reason, gaps between sections are practical only when diffuse processes such as interstitial lung disease are imaged (49).

CT detectors vary in their efficiency. Ideally a detector should count all incident beam x-ray photons. Depending on the technology used, however, detectors will record only 60% (high-pressure xenon detectors) to 95% (solid-state detectors) of the incident x-ray photons. Most current detectors are solid state. The accuracy of conversion of the absorbed x-ray signal into an electrical signal is known as the conversion efficiency. The overall dose efficiency of the scanner is the product of the geometric efficiency, the quantum detection efficiency, and the conversion efficiency (50). The overall dose efficiency can vary substantially between scanners. Noise is also introduced by the electronics of the data acquisition system of the scanner. The sum of quantum noise and electronic noise results in differences in image quality between scanners at the same radiation dose.

CT is similar to other radiologic techniques in that the primary x-ray beam is filtered to eliminate low-energy photons, which would be preferentially absorbed relative to high-energy photons and contribute to radiation dose. With CT, additional spatially varying filtration is often placed in the primary x-ray beam. These filters reduce *(a)* the necessary dynamic range of the detector system in the periphery of the detector array and *(b)* the radiation dose for larger fields of view. They are often referred to as bow tie filters because of their shape, and they create variations in entrance radiation exposure depending on both the size of the object and its position in the field of view. For some CT scanners, multiple filters of varying shapes are moved into place based on the specified field of view of the scan. In other scanners these filters are permanently positioned. These filters substantially reduce radiation dose of CT scans in adult patients, but they are less effective in pediatric patients.

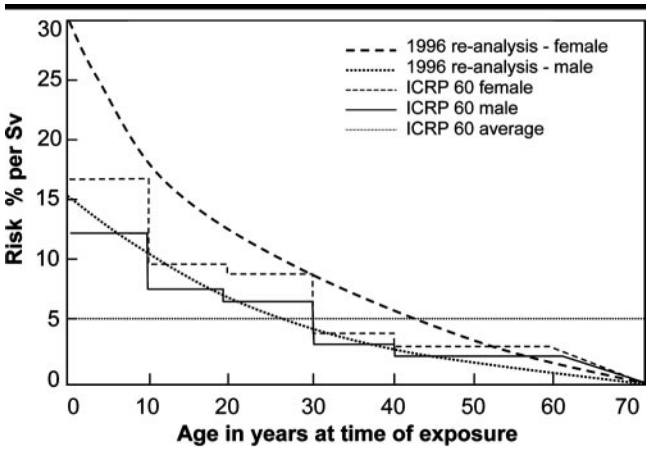

Figure 2. Graph compares lifetime mortality risk from cancer per sievert to age at the time of exposure. *ICRP* = International Commission on Radiological Protection. (Adapted and reprinted, with permission, from reference 70.)

USER-SPECIFIED SCAN PARAMETERS

Reduction in radiation dose results in increased image noise and decreased image quality. Studies assessing the subjective evaluation of chest CT scans have demonstrated that radiologists consistently gave higher image-quality scores to images obtained with a higher radiation dose (51,52). Image noise can be measured by placing a region of interest (>100 pixels) in an area of uniform density in the body (eg, the thoracic aorta) (32). The SD of the pixel values represents image noise. It is noted that the choice of reconstruction algorithm influences image noise, and higher noise is obtained by using high-spatial-frequency reconstruction algorithms (eg, bone or lung algorithms) rather than low-spatial-frequency algorithms (eg, standard, soft-tissue algorithm). High-spatial-frequency reconstruction algorithms are most commonly used when one is searching for fine structures within bones or lung tissue. The increase in image noise associated with the high-spatial-frequency algorithm is not a problem in these applications, because of the high radiographic contrast of these tissues.

Radiation dose and image noise can be modified by adjusting the tube current, scan time, and tube voltage. In practice, the tube current is usually adjusted to change the radiation dose and image noise. In most CT scanners, the tube current is adjustable in steps from 20 mA to approximately 400 mA. The radiation dose can also be linearly affected by scan time, but the time is usually minimized in imaging of the chest to reduce the effect of patient motion. Increasing the tube voltage increases the output of the x-ray tube. If the tube current and scan times are not changed, increasing the tube voltage will increase the radiation dose to the patient. Changes in tube voltage also affect CT tissue attenuation values, which can change tissue contrast in a complex fashion. In practice, tube voltage is not commonly adjusted between patients when chest CT is performed. It is noted that the radiation exposure delivered at a given tube voltage and current setting will vary greatly between CT scanners of different models and manufacturers because of differences in scanner geometry (x-ray tube-to-patient separation) and x-ray tube filtration.

Helical CT scanners introduced a new parameter: pitch. For single–detector row helical CT scanners, pitch is defined as the table travel per 360° x-ray tube rotation divided by the beam collimation (53). In many cases the table feed (eg, 5 mm per x-ray tube rotation) and beam collimation (eg, 5 mm) are identical, and the resultant pitch is 1. This yields one helical turn per section thickness and a radiation exposure equal to that of contiguous transverse sections. However, the table can be made to feed more rapidly (eg, 10 mm per x-ray tube rotation) without changing the beam collimation (5 mm). This results in a pitch of 2. Examinations with pitch values greater than 1 cover larger volumes in shorter times, which provides either reduced motion artifact or thinner sections. Scans obtained with elevated pitch have lower image quality because the section profile is broadened. However, the radiation dose delivered by the examination is decreased by the value of the pitch (eg, one-half of the radiation exposure for a pitch of 2) if the tube voltage and current are kept constant. It should be noted that in many multi–detector row scanners, the tube current is automatically increased to compensate for increased noise at higher pitch values, which may cancel out the radiation dose reduction. In some cases, such as when helical CT is used in the detection of pulmonary embolism, it has been shown that improved image contrast can be obtained with reduced radiation dose by using thinner sections at pitch values of 1.5–2 (3).

One manufacturer of multi–detector row CT scanners has redefined pitch as the table travel divided by the detector aperture (54). This definition elevates the value of the pitch by the number of de-

TABLE 3
Proposed Reference Dose Values for Routine CT Examinations on the Basis of Absorbed Dose to Air

	Reference Dose Value	
Examination	$CTDI_w$ (mGy)	DLP (mGy · cm)
Routine head*	60	1,050
Face and sinuses*	35	360
Vertebral trauma†	70	460
Routine chest†	30	650
High-resolution CT of lung†	35	280
Routine abdomen†	35	780
Liver and spleen†	35	900
Routine pelvis†	35	570
Osseous pelvis†	25	520

Note.—Reprinted, with permission, from reference 60. $CTDI_w$ = weighted CT DI.
* Data relate to head phantom (polymethylmethacrylate, 16-cm diameter).
† Data relate to body phantom (polymethylmethacrylate, 32-cm diameter).

tector rows. With the standard definition, for example, the acquisition of four 1.25-mm sections at a table speed of 10 mm per second in a 0.5-second scanner results in a pitch of 1. However, by using the same parameters but using the detector aperture as the denominator, the pitch value increases to 4. We agree with others (55) that this new definition of pitch does not demonstrate clearly the relationship between radiation dose and x-ray beam overlap found with the original definition. For this reason we believe that the original definition of pitch is preferable. We also note that the original definition of pitch is being adopted by an international standards committee on CT terminology (56).

In the past, the tube current of CT scanners was uniform at all angles around the patient. However, the chest is an elliptical object that has higher attenuation from left to right than from anterior to posterior. Manufacturers have introduced programs that alter the tube current, which increases radiation dose laterally and decreases it in the thinner anteroposterior direction. This has been shown to decrease radiation dose (57–59) with minimal effect on image quality. In the future we believe CT scanners will adjust dose automatically during the scanning process to compensate for the size and density of the body section being scanned, which will result in a signal-to-noise ratio that is adequate for diagnosis but is not excessive.

Repeated scanning of the same region (eg, unenhanced and contrast material enhanced) increases the radiation dose in a linear fashion. Therefore, if unenhanced CT is routinely performed prior to the contrast-enhanced CT, the radiation dose is doubled. This effect can be markedly reduced if the unenhanced CT is a high-resolution study (eg, 1-mm collimation at 10-mm spacing), for which the radiation dose is 10% of that of contiguous conventional CT or helical CT with a pitch of 1.

The tube current and voltage settings are usually set according to local experience and practice. Radiation dose surveys have noted wide variation in these settings between institutions (6,7,37–40). To decrease this variation and protect the public from inadvertent overexposure, the European communities have published suggested reference dose values (60) for many CT examinations (Table 3). These reference dose values were obtained by surveying a large number of institutions in Europe and adopting the 75th percentile of responses as the reference dose values. These values serve as a guide to acceptable practice in Europe.

DOSE REDUCTION IN CHEST CT

The concept of reduced tube current for conventional 10-mm collimation chest CT was introduced in 1990 by Naidich et al (61), who demonstrated acceptable image quality for assessment of lung parenchyma with low tube current settings (20 mAs). While these images were adequate for assessing lung parenchyma, they had a considerable increase in noise, which resulted in marked degradation of image quality with mediastinal windows. For this reason the authors noted that such low-dose techniques were most suited for assessment of children and possibly for screening patients at high risk for lung cancer. These recommendations have been implemented and further studied in lung cancer screening programs (62–64).

Similar dose reduction strategies have been applied to thin-section CT (also known as high-resolution CT) of the chest, in which no significant difference in lung parenchymal structures was seen between low-dose (40 mAs) and high-dose (400 mAs) thin-section CT images (65). Although differences were not statistically significant, changes in ground-glass opacity were difficult to assess on low-dose images because of the increased image noise. Therefore, it was recommended that 200 mAs should be used for initial thin-section CT and lower doses (ie, 40–100 mAs) should be used for follow-up CT examinations.

The radiation dose associated with thin-section CT has been controversial. DiMarco and Briones (34) quoted the high value of 120–140 mGy, which had been reported in an early article on thin-section CT (66). This dose estimate was measured by using contiguous 1.5-mm sections, 510 mAs, and CT DI methods in a head-sized (16-cm) CT quality control phantom. The CT DI measurement was designed to facilitate dose comparisons between CT scanners and was not appropriate for use in describing the relative dose of thin-section CT. As previously noted, the effective dose is a better measure of radiation dose because it takes into account the significant reduction in radiation risk associated with the noncontiguous sections used in thin-section CT. With 10-mm intersection gaps, the effective dose of thin-section CT is 10% that of either conventional contiguous CT or helical CT with a pitch of 1. The effective dose of thin-section CT is reduced to approximately 5% with the use of 20-mm intersection gaps. As noted previously, low-dose, thin-section CT can also be performed in selected patients. It has been shown that three low-dose, thin-section CT sections provide an effective dose comparable to that of posteroanterior chest radiography (0.05 mSv), with no significant loss of diagnostic accuracy in interstitial lung disease ($P > .25$) (49).

The relationship between radiation exposure and image quality with both mediastinal and lung windows has been evaluated on conventional 10-mm collimation chest CT images (51) on a single model of CT scanner. Although findings of this study showed a consistent increase in mean image quality with higher radiation exposure, they did not show a significant difference in the detection of mediastinal or lung parenchymal abnormalities from 20 to 400 mAs. The authors concluded that with the CT scanner model they used, adequate image quality could be consistently obtained in average-sized patients by using tube currents of 100–200 mAs. This study was limited by the small number of patients ($n = 30$), the specific CT scanner factors (geometry, filtration, tube voltage), and the experimental design, which limited low-dose sections to two levels that often were not those with clinically relevant findings. The authors noted that to evaluate further the effect of reduced radiation dose on diagnostic accuracy in chest CT, comparison of complete chest CT studies at a variety of radiation exposures in a large number of patients would be required. However, they noted that such a study could not be performed in patients because of the unacceptable radiation dose that would result from multiple CT examinations at differing radiation exposures. Additionally, the variable effect of motion artifacts on repeated scanning would make comparison difficult.

A practical method for evaluating the effect of reduced radiation dose on image quality is computer simulation (67). The technique consists of obtaining a diagnostic scan with standard dose and then modifying the raw scan data by adding Gaussian-distributed random noise to simulate the increased noise associated with reduced radiation exposure. The raw scan data are then reconstructed by using the same field of view and reconstruction algorithm as the high-dose reference scan. In a validation trial, experienced chest radiologists were unable to distinguish simulated reduced-dose CT images from real reduced-dose CT images (67). Computer simulation allows investigators to determine the effect of dose reduction on diagnostic accuracy without exposing patients to radiation unnecessarily. In addition, the simulated images are in exact registration with the original images, eliminating artifacts due to volume averaging or motion. This technique has been used recently to evaluate the diagnostic effect of radiation dose reduction in pediatric abdominal CT (68).

Finally, it should be noted that although CT is a modality with relatively high radiation dose, in some cases CT has replaced modalities with higher radiation exposures such as pulmonary angiography or bronchography (Table 4).

TABLE 4
Comparison of Effective Doses

Procedure	Effective Dose (mSv)
Posteroanterior chest radiograph	0.05*
Conventional CT	7.0†
Spiral CT pitch 1	7.0†
Spiral CT pitch 2	3.5†
High-resolution CT with 10-mm intersection gap	0.7†
High-resolution CT with 20-mm intersection gap	0.35†
Thin-section low-dose high-resolution CT	0.02‡
Conventional pulmonary angiography	9.0§
Digital pulmonary angiography	6.0§
Conventional bronchography	3.0‖
Annual natural background radiation	2.5*

Note.—Reprinted, with permission, from reference 69.
* Source.—Reference 70.
† Source.—Reference 71.
‡ Source.—Reference 49.
§ Calculated with data from reference 72, assuming pulmonary angiography with 5 minutes of fluoroscopy and the equivalent of 30 posteroanterior and 30 lateral views.
‖ Bronchography performed with the assumption of 2 minutes of fluoroscopy and six posteroanterior and six lateral views.

CONCLUSION

The introduction of helical and multi–detector row CT scanners has resulted in an increase in the number of indications for and diagnostic accuracy of CT. However, the current level of radiation exposure from CT is high, particularly in the pediatric population. Radiation dose surveys have indicated that there is large variation in the technical factors employed by physicians, and there is a resultant large variation in the radiation dose to patients. Reference dose values for chest CT have been developed and published. Radiologic societies should be encouraged to consider the benefits that adoption of such standards would have on medical practice. Further research into the complex relationship between radiation exposure, image noise, and diagnostic accuracy should be encouraged to establish scientifically the minimum radiation doses that provide adequate diagnostic information for standard clinical questions. Once these minimum levels of image quality are determined and validated, automatic exposure controls for CT scanners should be developed to ensure that all patients undergo CT with techniques that conform to the ALARA (As Low As Reasonably Achievable) principle. Radiologists must take the lead in promoting these measures for patient protection. Since children have an increased radiation sensitivity, initial dose reduction efforts should be focused on the pediatric population. The Society for Pediatric Radiology has recently made recommendations in this regard (73–85). Finally, it is noted that the complexity of CT requires a close collaboration between radiologists and medical physicists to successfully reduce radiation dose while maintaining diagnostic accuracy.

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