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Reducing Radiation Dose in Body CT: A Primer on Dose Metrics and Key CT Technical Parameters

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OBJECTIVE. The purpose of this article is to describe how to interpret radiation dosimetric data available on a body CT dose report and to explain the effects of key operator-chosen CT parameters on patient radiation dose.

CONCLUSION. To apply dose reduction strategies in body CT, radiologists must understand the information contained in the CT dose report and know the effects of key CT technical parameters on patient radiation exposure.

The benefits of CT to the practice of medicine are indisputable but concerns regarding increased cancer risk from CT continue to escalate. Of the 67 million CT examinations performed in the United States in 2006, more than 21 million were abdominal and pelvic CT examinations—over 30% of all CT studies. Although CT accounts for only about 17% of imaging examinations, CT is reportedly responsible for almost half of the collective effective dose from medical procedures in the United States [1]. According to risk projection models, in a few decades 1.5–2% of all cancers in the United States may be attributable to the use of CT [2]. Cancer risk from CT may no longer be theoretic because a recent study reports, for the first time, a direct increase in cancer rates related to radiation exposure from CT [3].

Radiologists should adhere to both the principle of ALARA (as low as reasonably achievable—referring to radiation dose) and the principle of AHARA (as high as reasonably achievable—referring to benefit) [4]. The objective is to achieve diagnostic-quality images addressing the clinical concern in the most dose-efficient manner. Although equipment manufacturers continue to make progress toward automating CT dose optimization, radiologists must accept the primary responsibility for minimizing radiation dose to patients from CT. To apply dose reduction strategies in body CT, radiologists must know how to interpret radiation dosimetric data available on a CT dose report and understand the effects of key CT technical parameters on patient radiation exposure. The purpose of this article is to provide this necessary background information.

CT Dose Report

To understand and make optimal use of dose reduction strategies in CT, radiologists should first be familiar with the concepts of exposure, absorbed dose, and effective dose. The term “exposure” refers to the intensity of an x-ray beam. It is a measure of the ability of an x-ray beam to ionize air. It is best understood as the number of x-ray photons passing through a given area and is usually measured using an ionization chamber and an electrometer. The absorbed radiation dose is the amount of energy absorbed per unit mass at a specific point and is measured in grays (1 Gy = 1 J/kg). The effective dose is a weighted average of the doses to all exposed organs taking into account varying organ radiosensitivities of different organs. The effective dose is the uniform whole-body dose that results in the same stochastic risk as the absorbed dose from any given nonuniform exposure such as a CT examination of the abdomen. The effective dose is measured in sieverts (Sv).

Using the measurement of effective dose, the risks of exposure to ionizing radiation from different diagnostic procedures, such as abdominal radiography (0.6 mSv) and abdominal CT (10 mSv), can be directly compared [5]. However, one should be cautious in the application of the concept of effective dose to medical imaging. Effective dose was primarily designed for use in radiation protection to assess risk for a population in general epidemiologic terms. Although effective dose can be used to

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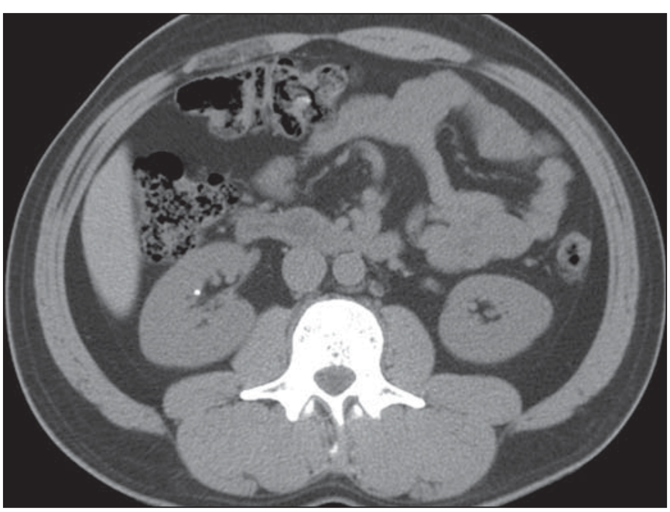
compare the stochastic risk of radiation exposure from diagnostic imaging for an idealized reference patient model, it should not be used to predict absolute risk for an actual individual patient [6, 7]. Even for the reference patient, the relative uncertainty in the estimated value of effective dose for medical exposure has been estimated to be approximately ± 40% [8].

In conventional film-screen radiography, exposure factors that are too high are visually apparent because the image is too dark. However, with CT, image quality improves (the amount of image noise decreases) as the amount of radiation exposure increases. Although noisy or grainy CT images indicate that the amount of radiation may be insufficient to yield diagnostic-quality images, one cannot determine from visual inspection of CT images that the radiation exposure is far in excess of that required to produce a diagnostic study (Fig. 1).

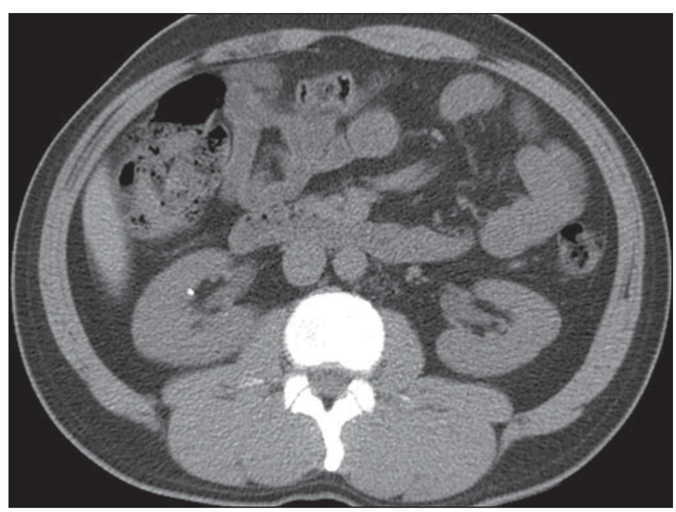
Currently, CT scanners provide dose data for each examination in a dose report as a DICOM image that can be easily stored in PACS with the anatomic CT images for the corresponding study (Fig. 2). The most important information in the dose report is the volume CT dose index ($CTDI_{vol}$) and the dose-length product (DLP). The $CTDI_{vol}$, expressed in milligrays (mGy), is a measure of the CT radiation output directed at a given patient. The $CTDI_{vol}$ is determined by the particular scanning protocol (i.e., CT parameter settings) and is standardized for body imaging on the basis of scanning an acrylic phantom 32 cm in diameter. Simplistically, $CTDI_{vol}$ can be considered the average radiation output per slice of the CT scanner and depends only on the type of scanner and acquisition parameters such as x-ray tube peak kilovoltage (kVp) and tube current–time product. It is independent of patient size and scan length [9]. The $CTDI_{vol}$ is displayed on the CT console just before the patient is scanned and can be used to alert the operator that the protocol should be modified if the $CTDI_{vol}$ is deemed excessive for that particular study [10]. The $CTDI_{vol}$ is the metric used by the American College of Radiology (ACR) for CT site accreditation. The ACR has established diagnostic reference levels for $CTDI_{vol}$ that indicate the 75th percentile at ACR-accredited facilities. For abdominal CT the ACR $CTDI_{vol}$ diagnostic reference level is 25 mGy with the pass-fail criterion at 30 mGy [11].

The DLP, expressed in milligrays × centimeters (mGy × cm), is the product of the $CTDI_{vol}$ (mGy) and scan length (cm). It represents the integrated dose over the length of the exposure and reflects the total amount of radiation incident on the patient. A change in DLP reflects changes to CT dose parameters and changes in scan length. Limiting the scan length (*z*-axis coverage) to only the anatomy of clinical interest and decreasing the number of acquisition phases for multiphase examinations are potent strategies for dose reduction. An estimate of the effective dose can be calculated from the DLP by taking the product of the DLP and a body part–specific conversion factor (k). For CT of the abdomen and pelvis, k is equal to 0.015 mSv/mGy × cm [12]. For example, if the total examination DLP on the dose report is 1000 mGy × cm, then an estimate of the effective dose would be 15 mSv because (1000 mGy × cm) × (0.015 mSv/mGy × cm) = 15 mSv. Because the effective dose is proportional to the DLP, the effective dose will reflect the combined effects of CT parameter changes and changes made to scan length.

Although the numeric values of the $CTDI_{vol}$ and DLP are critical for dose management, calculations of effective dose from DLP are only crude estimates. For any given exposure factors, the patient dose depends on patient size: The larger the patient, the smaller the patient dose. For body CT of a large patient, the radiation is attenuated by the thicker torso, resulting in decreased exposure of radiosensitive tissues. In addition to patient size, the conversion factor (k) also varies depending on the International Commission on Radiological Protection (ICRP) weighting factors for radiosensitive organs, anatomic region irradiated, x-ray tube voltage, and patient age. For example, the DLP-to–effective dose conversion factor in a newborn infant undergoing a CT examination is approximately five times higher than that for an average-sized adult [13]. Thus, if technical factors are not adjusted properly on the basis of size for children or small adults, these patients can easily be subjected to excessive CT radiation exposures.



A



B

Fig. 1—35-year-old man with history of nephrolithiasis.
A and **B**, Axial 5-mm-thick unenhanced CT images at level of kidneys from two separate studies performed 2 years apart. Earlier study (**A**) was obtained at tube current–time product of 400 mAs and later study (**B**) was obtained at 80 mAs but both studies were obtained using tube voltage of 120 kVp. Both studies are of diagnostic quality and both show small calculus in right kidney. Although **B** is noisier (grainier) than **A**, it is not apparent from visual inspection of images that radiation dose to patient from first study (**A**) is five times greater than dose from latter study (**B**).

Exam Description: CT/Abdomen W T W/O Con

Dose Report

Series	Type	Scan Range (mm)	CTDIvol (mGy)	DLP (mGy-cm)	Phantom cm
1	Scout	–	–	–	–
2	Helical	I5.250–I145.250	2.95	51.53	Body 32
200	Axial	I5.250–I5.250	6.02	6.03	Body 32
3	Helical	I5.250–I145.250	16.91	286.55	Body 32
3	Helical	I5.250–I265.250	16.86	488.11	Body 32
4	Helical	I5.250–I145.250	16.90	286.39	Body 32
		Total Exam DLP:		1118.61	

Fig. 2—Example of CT dose report provided by CT scanner as DICOM-compliant image that is sent to PACS. Study depicted is multiphase study of liver. Series 1 represents CT topograms; Series 2, unenhanced phase; Series 200, bolus tracking sequence for scan delay; first Series 3, arterial phase; second Series 3, portal venous phase; and Series 4, delayed phase. Scan range column indicates beginning and ending table positions for each scan phase. Volume CT dose index (CTDIvol) and dose-length product (DLP) are key dosimetric data as explained in text. Diameter of phantom used for dose calibration is indicated in last column. No dose data is listed for CT topograms because radiation exposure is very low compared to conventional scanning phases. W T W/O Con = with and without contrast material, mGy-cm = mGy × cm.

CT Technical Parameters

For management of CT radiation dose, radiologists should be cognizant of the effects of various scanning parameters on radiation dose. These factors include the x-ray tube current expressed in milliamperes (mA), tube current–time product expressed in milliampere-seconds (mAs), x-ray tube peak voltage (kVp), x-ray tube rotation time (exposure time), helical pitch, reconstructed slice thickness, image noise, automatic exposure control (AEC), and noise-reducing image reconstruction algorithms (Table 1).

The tube current–time product (mAs) is a measure of the photon flux and is the product of the tube current (in milliamperes) and exposure time (in seconds). The exposure time is determined by the x-ray tube rotation time. Radiation dose is directly proportional to the mAs if all other factors are held constant. Thus, if the mAs for a particular study is reduced by 50%, the radiation dose is also reduced by 50%. If all other factors are held constant, radiation dose can be reduced by lowering the tube current (mA) or decreasing the x-ray tube rotation time (increasing the x-ray tube rotation speed). However, image noise is proportional to $1 / \sqrt{(\text{mAs})}$. For example, halving the tube current–time product from 400 to 200 mAs produces a 50% reduction in dose but results in a 41% increase in image noise [5]. It should be noted that to maintain a sufficient photon flux (mAs), modern scanners with faster rotation times may require higher mA settings than older units. For example, 800 mA at a rotation time of 0.5 second results in 400 mAs, which is equivalent to 400 mA at a rotation time of 1.0 second.

The relationship of dose to kVp is more complicated. There is an exponential relationship of kVp to radiation dose: Radiation dose is proportional to kVp raised to an exponential power of more than 2, a power ranging from 2.5 to 3.1 depending on patient size [14]. For a typical abdominal CT phantom, decreasing the peak kilovoltage from 140 to 120 kVp reduces the dose by 28–40% and decreasing to 80 kVp reduces the dose by about 65% [9, 14]. However, tube voltage changes are limited because users can select from only several preset peak kilovoltage settings—typically 80, 100, 120, and 140 kVp. Thus, precise adjustment of radiation dose is not possible solely through manipulation of the peak kilovoltage setting [15].

Decreasing the kVp setting will increase image noise as a result of the reduced photon flux and photon energy. To compensate for the increased noise, one can increase the mAs setting; increasing the mAs setting does offset some, but usually not all, of the dose reduction from lowering the kVp. This strategy (low-tube-voltage, high-tube-current CT technique) is efficient for decreasing radiation dose while maintaining image quality in small and average-sized patients [16]. For large patients, lowering the kVp results in increased noise that may not be overcome by increasing the tube current. In large patients a higher tube voltage may be the most dose-efficient strategy because there is better penetration into the organs of interest [17].

TABLE 1: CT Technical Parameters and Effects on Radiation Dose

CT Technical Parameter	Definition	Effect of CT Technical Parameter on Radiation Dose
X-ray tube current (mA)		Radiation dose is directly proportional to x-ray tube current
X-ray tube rotation time (s) = exposure time	Time for one complete rotation of the CT gantry	Radiation dose is directly proportional to x-ray tube rotation time and is inversely proportional to rotation speed
X-ray tube current–time product (mAs)	Product of the x-ray tube current (mA) and exposure time (s)	Radiation dose is directly proportional to x-ray tube current–time product
X-ray tube peak kilovoltage (kVp)		Radiation dose is proportional to kVp raised to an exponential power ranging from 2.5 to 3.1 depending on patient size
Pitch	Table distance traveled in one 360° gantry rotation divided by the total collimated width of the x-ray beam	Radiation dose is inversely proportional to pitch
Reconstructed slice thickness		Radiation dose is not directly affected by reconstructed slice thickness if all other factors are unchanged. However, to maintain the same image noise levels with varying slice thickness, patient dose must vary in inverse proportion to the reconstructed slice thickness. For example, 2.5-mm slice thickness will require about twice the dose of 5-mm slice thickness
NI		Radiation dose is inversely proportional to square root of NI. Decreasing NI by 5% increases dose by 10.8%, whereas increasing NI by 5% decreases dose by 9.3%

Note—NI = noise index.

An additional benefit of lowering the kVp setting for imaging small and average-sized patients is a potential improvement in the diagnostic quality for certain studies. Attenuation produced by iodine-based contrast material increases up to 100% if tube voltage is decreased from 140 to 80 kVp because of the higher probability of photoelectric interactions at the lower tube voltage. Despite the increased noise at the lower kVp settings, the increased attenuation of enhancing arterial structures against the surrounding tissues produces a higher contrast-to-noise ratio, which is especially advantageous for interpretation of CT angiographic studies. Because of its increased sensitivity for detection of contrast enhancement, a low-tube-voltage, high-tube-current CT technique may also improve the conspicuity of small hypervascular tumors of the liver and pancreas [16].

Dual-energy CT is a recent innovation that involves acquisition of datasets at two different kV energy settings at the same anatomic location to provide material-specific information based on the differences in attenuation between the two energies. The closer the energy level of the x-ray beam is to the K edge of a particular substance, the greater the attenuation of the beam. Dual-energy CT can be performed by simultaneously applying two x-ray tubes at different kV and mA settings with dual-source CT or with single-source CT using fast kV switching. With a dual-source system, dual-energy scanning is reported to be dose-equivalent to conventional single-energy scanning because the tube current in each x-ray tube can be adjusted independently to optimize image quality and minimize dose. The radiation dose of dual-energy scanning using a single-source system with rapid kV switching is reported to be approximately 20% higher than conventional CT because the tube current setting cannot be changed as the kV is switched on the single x-ray tube [18]. Dual-energy CT can be used to efficiently remove overlying bony structures from datasets, better characterize liver and renal masses, determine kidney stone composition, and improve delineation of blood vessels on CT angiographic studies. The two energies most frequently used with current systems are 80 and 140 kVp. Because the K edge of iodine (33.2 keV) is closer to 80 kVp than it is to 140 kVp, the attenuation of contrast-enhanced structures such as organs and blood vessels is significantly higher at 80 kVp. With postprocessing techniques the iodine content can be identified and subtracted from a dual-source contrast-enhanced acquisition to yield virtual unenhanced images. This capability can obviate an unenhanced CT scan in a multiphasic CT protocol resulting in a reduction of total radiation dose [19].

Helical pitch is defined as the table distance traveled in one 360° gantry rotation divided by the total collimated width of the x-ray beam. Increasing the pitch moves the table more rapidly through the gantry and reduces the radiation dose by shortening the scanning time. If all other parameters are held constant, radiation dose is inversely proportional to pitch. Doubling the pitch halves the dose. Disadvantages of increasing the pitch are increased image noise and increased effective section thickness. We should note that MDCT scanners using AEC will compensate for the increase in pitch by increasing the tube current (mA) to maintain a constant noise level. For these reasons, although increasing pitch is effective for shortening the scanning time, it is not a practical strategy for radiation dose reduction with single-source MDCT scanners [15].

When performing an MDCT study, one selects the acquired slice thickness (detector collimation or effective detector-row thickness) but can reconstruct the images at a different slice thickness depending on the requirements of the particular case. The acquired slice thickness determines the minimum reconstructed slice thickness because the reconstructed slice thickness is usually a multiple of the acquired slice thickness. Thus, if the acquired slice thickness is 0.625 mm, the reconstructed slice thickness can be 0.625, 1.25, 2.5, or 5.0 mm.

Another parameter, detector configuration, refers to the number of data channels used (number of slices acquired per rotation) and the effective detector-row thickness of each data channel. The product of the number of data channels and the effective detector-row thickness determines the x-ray beam collimation. A 64-slice CT scanner with 0.625-mm detector-rows produces a total beam collimation of 4.0 cm (64 × 0.625 mm). A 16-slice CT scanner in a configuration of 16 × 0.625 mm detector-rows produces a total beam collimation of 1.0 cm. Each time the CT gantry rotates, some radiation at the cranial and caudal edges of the beam falls beyond the detectors (the penumbra of the beam) making the incident x-ray beam about 2 mm wider than the selected beam collimation or detector configuration. The penumbra is not used for image formation and produces excess radiation dose called "overbeaming." For a given acquired slice thickness, a CT scanner with a higher number of detector-rows produces a wider beam and requires fewer rotations to cover a given scan length at a constant pitch resulting in less overbeaming. Therefore, for acquiring thinly collimated slices, 64-slice scanners are more dose-efficient with regard to overbeaming than 16-slice scanners [20].

Although reconstructed slice thickness with MDCT does not directly affect radiation dose, if all other factors are held constant, thinner reconstructed slices produce noisier images because fewer photons contribute to image formation. Noise is proportional to $1 / \sqrt{T}$, where T is the reconstructed or nominal section thickness. Therefore, a 2.5-mm-thick section will have 1.4 times more noise than a 5-mm section. Depending on the slice thickness required for a diagnostic study, the mAs, kVp, or both may need to be increased to offset the increased noise from the thinner sections [5]. Image noise is approximately inversely proportional to the square root of the radiation dose. The radiation dose must change in inverse proportion to the slice thickness to maintain constant image noise for varying reconstructed slice thicknesses. For example, a 2.5-mm-thick section will require about twice as much radiation as a 5-mm-thick slice to maintain the same noise level [21].

AEC and automatic tube current modulation are applications in which CT tube current (mA) is adjusted during scanning to the minimum level necessary to obtain a constant preselected image quality based on the size and density of the patient section being scanned. The adjustments are based on patient thickness and density calculated from the scanned projection radiograph (topogram or CT scout image). The adjustments can be made along the craniocaudal axis of the patient (z-axis modulation) accounting for varying patient thickness as the table moves through the gantry and in the axial direction (angular modulation) to account for varying patient thickness at different angles of gantry rotation. The goal of AEC is to maintain the selected image quality level at all anatomic locations using the minimum required radiation exposure. This technology can reduce radiation dose by 20–44% when the appropriate image quality setting is chosen [10].

For the CT scanners used at our institution, all of which are manufactured by GE Healthcare, the diagnostic image quality for each protocol is set by entering a "noise index" (NI) along with a range of acceptable tube current settings—that is, minimum mA (labeled "mA min") and maximum mA ("mA max")—to prevent under- or overexposure of the patient. The NI value is defined as follows [21]:

> …approximately equal to the standard deviation in the central region of the image when a uniform (20-cm water) phantom is scanned and reconstructed using the standard reconstruction algorithm.

Thus, the selected NI value represents the expected SD of the attenuation values in an image of a uniform region. One can also gauge the amount of noise on a clinical CT study by measuring the SD of the attenuation values within a circular region-of-interest (ROI) caliper placed on an anatomic structure of homogeneous attenuation such as the liver or distended urinary bladder on an unenhanced study. If AEC is functioning properly, the measured SD will be approximately equal to the operator-selected NI for the CT study. A higher NI setting results in a lower radiation dose at the cost of a noisier image. As with image noise in general, the NI is inversely proportional to the square root of the dose. Decreasing the NI by 5% increases the dose by 10.8%, whereas increasing the NI by 5% decreases the dose by 9.3% [21].

The operator sets the NI based on the desired diagnostic image quality for a given reconstructed slice thickness for a given protocol. The prespecified NI, then, can become the primary determinant of radiation dose. The CT system will automatically modify the tube current to maintain the selected NI even if other parameters such as table speed (pitch) or gantry rotation time are changed. It is important to recognize that if the NI is set lower than necessary, overexposure of the patient occurs. Placing a limitation on the maximum tube current can mitigate this possibility to a certain extent if set properly. For dose reduction with AEC, it is critical to properly match the required image quality to the reformatted slice thickness of the CT protocol, which is usually determined by the clinical indication [15]. Erroneously assigning the desired NI for a 5-mm slice thickness to a slice thickness of 2.5 mm would needlessly double the radiation dose to the patient.

Although NI is the input parameter used by the GE Healthcare AEC system (Auto mA 3D) to define image quality, other manufacturers of CT systems use different terms and methods. The Toshiba AEC system (Sure Exposure 3D) is similar to that of the GE system in that both are based on a selected noise reference value. These AEC systems were designed to maintain uniformity of image quality in different anatomic regions of the same patient [22]. With the Toshiba AEC system, the user selects a desired SD of pixel values to specify the image quality and the tube current can be modulated within a range of minimum and maximum tube current settings. A higher value for SD results in noisier images. Although the SD is similar to the NI setting on a GE Healthcare system, there is one important difference: With NI, the SD is referenced to a "standard reconstruction algorithm" and is not affected by the reconstruction algorithm used for a particular case. The SD setting on a Toshiba system is applied to the first reconstruction algorithm used during the application of AEC. Thus, a smooth reconstruction algorithm would result in less radiation exposure than a noisier edge-enhancing algorithm [23].

The Philips Healthcare (DoseRight) and Siemens Healthcare (CARE Dose 4D) AEC systems are similar in that both are based on a selected reference for the determination of acceptable image noise. With Philips Healthcare's CT systems, the user chooses a reference image from a prior satisfactory patient examination to specify the desired image quality to match. Previously stored image data from a prior acceptable examination (including the raw CT projection data and the CT projection radiograph) are used as the reference data for comparison with the CT projection radiograph and other data from the patient being scanned. On the basis of the type of CT examination (i.e., protocol requirements for the anatomic regions studied), the software modulates the tube current to the lowest possible levels to achieve the same noise levels as the reference case. Siemens uses the concept of a "reference effective mAs" input value to set desired image quality with AEC. The effective mAs is used to compensate for the helical pitch used during the application of a given tube current and is defined as *mAs* / pitch, where *mA* is the tube current and *s* is the gantry rotation time in seconds. For each CT protocol, the user selects the effective mAs typically used for CT in a patient of average size (70–80 kg for adults and 20 kg for pediatric patients). The system software assesses the size of the cross section of the patient being scanned and modifies the tube current relative to the reference effective mAs. The tube current variations are determined from the CT projection radiographs and fine-tuned by an online feedback system. Rather than keeping noise constant for all patients, the noise target (SD of CT attenuation values in a uniform region) is varied on the basis of the size of the patient using an empiric algorithm. The system functions on the principle that different-sized patients require different levels of image noise to maintain diagnostic image quality. Compared with the average-sized patient, lower noise levels are selected when imaging small patients to improve delineation of anatomy and higher noise levels are accepted in large patients to avoid excessively high exposure factors. The user can also adjust the degree to which tube current is adjusted for patient size by choosing "weak," "average," or "strong" compensation settings [23–25].

Identifying the optimal image quality settings is not straightforward. There are currently no guidelines for acceptable noise levels for specific clinical indications and the perception of "acceptable" noise is subjective. However, CT protocols can be stratified with regard to noise tolerance. For body CT scans, the most noise-tolerant examinations include CT angiography, followed by CT urography and unenhanced CT for the diagnosis of urinary tract stones. Less noise-tolerant studies are CT scans for evaluation of the parenchyma of abdominal organs such as the liver, pancreas, or kidney. The least noise-tolerant studies are unenhanced CT examinations of the abdomen and pelvis for evaluation of organ parenchyma [26].

Image noise is also dependent on patient size; x-rays are attenuated by soft tissue in an exponential fashion with a half-value layer of approximately 10 cm for the abdomen [27]. Therefore, for CT to maintain a constant noise level going from thin to obese patients, the mAs setting would have to increase exponentially. Many CT systems are not capable of the very high mAs values necessary to achieve noise levels in obese patients as low as those usually obtained in average-sized or thin patients and, in any case, the radiation exposures would be unacceptably high. Fortunately, image readers have a higher tolerance for noise in large patients than in thin patients likely because of the greater amount of intraabdominal fat in the former that increases tissue contrast around the abdominal organs. Although readers require less image noise for interpreting images of smaller patients, higher noise levels are acceptable in larger patients [25]. Because it is not necessary to maintain uniform noise levels for patients of different sizes, image quality settings can be adjusted to accept higher noise levels as patient size increases.

Patient size is a critical factor in the design of body CT protocols because of its effect on noise tolerance, radiation sensitivity (smaller patients receive higher effective doses for the same amount of exposure compared with larger patients), and optimal kVp and mAs requirements. CT technical parameters (such as kVp, mAs, and noise targets) must be adjusted according to patient size. Different size measures, which include patient weight, body mass index, lateral width, and abdominal circumference and display FOV [9, 14] have been proposed for this purpose.

The use of alternative image reconstruction techniques that lessen image noise can produce a substantial reduction in radiation dose from CT. Until recently filtered back projection had been the only practical image reconstruction method. Although filtered back projection is computationally rapid, limitations of the underlying simplified assumptions and approximations regarding data acquisition (such as accounting for the effects of beam hardening and scatter) result in excessive noise and streak artifacts with low-dose CT protocols. Iterative reconstruction is a more computationally intensive reconstruction method that produces high-quality images at lower tube currents. With iterative reconstruction, an algorithm begins with an initial image of a patient, computes data that would result from projections of the initial image, compares that data to the actual projection data acquired during the CT examination, and appropriately revises the image so that the computed projection data more closely match the acquired projection data. The first generation of iterative reconstruction software, such as adaptive statistical image reconstruction (ASIR) and iterative reconstruction in image space (IRIS), functions mainly by filtering noise from the reconstructed CT images. These algorithms can be used either to decrease the image noise in large patients or to maintain diagnostically sufficient noise levels at lowered radiation exposure in average-sized patients. Using these techniques, investigators have reported dose reductions of up to 65% for abdominal CT while preserving diagnostic image quality [10, 17]. Model-based iterative reconstruction (MBIR), the most recent development, incorporates into the reconstruction process a model of the physics of the optical chain of the CT system in place of assumptions. The reconstruction algorithm models the real size of the focal spot, the attenuation of the x-rays through the patient (beam hardening and scatter), the x-ray detector geometry, and the electronic noise to develop a more accurate representation of the scanned object from the acquired data [28]. MBIR algorithms produce very low levels of image noise and have the potential for considerable reductions in radiation dose compared with conventional CT. The mean dose reduction in one preliminary investigation was 74% for abdominal imaging, but even greater dose reductions are achievable [29]. Currently the major drawback of MBIR is that the intensive computational requirements result in prolonged image reconstruction time on the order of hours for a single CT study.

Although iterative reconstruction techniques are commercially available, they require expensive upgrading of equipment. Even if one uses iterative reconstruction, CT protocols should still be optimized to further reduce patient radiation exposure. In a related article [30], we will address specific body CT protocol modifications to minimize radiation dose.

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