Although there has been increasing recognition of the importance of reducing radiation dose when performing multidetector CT examinations, the increasing complexity of CT scanner technology, as well as confusion about the importance of many different CT scan parameters, has served as an impediment to radiologists seeking to create lower dose protocols. The authors seek to guide radiologists through the manipulation of 8 fundamental CT scan parameters that can be altered or optimized to reduce patient radiation dose, including detector configuration, tube current, tube potential, reconstruction algorithm, patient positioning, scan range, reconstructed slice thickness, and pitch. Although there is always an inherent trade-off between image quality or noise and patient radiation dose, in many cases, a reasoned manipulation of these 8 parameters can allow the safer imaging of patients (with lower dose) while preserving diagnostic image quality.

Key Words: Radiation dose, automated tube current modulation, low kVp, pitch, iterative reconstruction

INTRODUCTION
Over the past several years, there has been increasing recognition of the importance of reducing radiation dose in radiologic studies, particularly with regard to multidetector CT. The growth in the volume of multidetector CT studies performed in the United States, the increasing use of CT in susceptible populations (including children), and growing concerns on the part of the general public with regard to radiation exposure have provided an impetus for performing these studies with the least possible radiation dose [1]. It is now believed that as many as 0.4% of all malignancies in the United States can be traced back to radiation from CT studies performed between 1991 and 1996 and that radiation from CT studies currently being performed may ultimately account for 1.5% to 2% of all cancers in the future [2]. Unfortunately, despite this growing awareness among radiologists, the introduction of each new generation of CT scanners has resulted in scanning protocols that are increasingly complex and difficult to manipulate for a radiologist whose primary expertise lies in clinical imaging rather than medical physics. When facing increasing criticism regarding CT dose, the best way to tackle the issue is to understand all the factors and parameters that can affect radiation dose and image quality and examine how these can be altered to reduce dose.

In this review, we seek to provide radiologists with a simple approach to the manipulation of 8 CT parameters that can be changed by the radiologist, while designing or altering scan protocols, to reduce patient radiation dose: (1) detector configuration, (2) tube current, (3) tube potential (kVp), (4) reconstruction algorithm, (5) patient positioning, (6) scan range, (7) reconstructed slice thickness, and (8) pitch.

DETECTOR CONFIGURATION
Detector configuration is a term encompassing the number of data channels being used in the z axis and the “effective detector thickness” of each data channel. For example, a detector configuration of 64 × 0.5 mm would suggest the use of 64 data channels in the z-axis, each of which has an effective thickness of 0.5 mm. Notably, the effective detector thickness represents the smallest possible reconstructed slice thickness: if a study is acquired with an effective detector thickness of 0.5 mm, images cannot be reconstructed to any smaller interval (such as 0.25 mm). Both the number of channels used and the effective detector thickness can be varied depending on how many channels in a detector array are used, which channels in a detector array are used, and the manner in which different channels are combined [3]. Different scanners from different manufacturers have different included...
detector channels, and the detector configurations available on different equipment can vary widely [4].

The product of the detector configuration (the number of data channels multiplied by the effective detector row thickness) is equivalent to the beam collimation. The physical manifestation of the beam collimation is the x-ray beam size as defined at the gantry isocenter [5]. In the example given previously, with a detector configuration of $64 \times 0.5$ mm, the beam collimation would be $32$ mm. This has a critical impact on patient dose because the x-ray beam is actually slightly wider than the beam collimation as a result of a penumbra effect (also referred to as “overbeaming”), resulting in a small amount of “wasted” radiation dose with each rotation [6]. As a result, if a smaller beam collimation ($4 \times 0.5$ mm) is used, the relative wasted dose is relatively high compared with the primary beam size. On the other hand, with a larger $64 \times 0.5$ mm collimation, the wasted dose is relatively low compared with the primary beam size. However, it should be noted that with the newest CT scanners, the penumbra effect is much less apparent compared with earlier generations of spiral CT technology because the extent of the overbeaming or penumbra effect is less pronounced (relative to the beam width). Therefore, the need to obtain a smaller effective detector thickness to improve spatial resolution in the z-axis must be balanced against the increased radiation dose associated with such a detector configuration, as well as longer scan times [3,5].

The detector configuration should be determined on the basis of the type of study performed, the necessary slice thickness for multiplanar reformations (MPRs), and the need for 3-D images. For routine acquisitions, without the need for thin-section MPR images or 3-D imaging, an extremely thin effective detector thickness is unnecessary. In such cases, if thick-section 5-mm images in the axial, coronal, and sagittal planes are sufficient for radiologist review, the effective detector thickness can be widened to 1.25 or 1.5 mm (rather than 0.75 or 0.5 mm), without a marked impact on image quality (either for the axial images or MPR images). However, if thin-section MPRs or high-resolution 3-D images (with increased spatial resolution in the z-axis) are required, a thin effective detector thickness (0.5-0.75 mm) may be necessary, although there is an associated dose penalty.

**TUBE CURRENT**

Increases in tube current or the product of tube current and scan time (mAs) result in improved image quality, decreased image noise, and increased patient dose. In general, the relationship between tube current and patient dose is essentially linear, with increases in mAs resulting in a comparable percentage increase in patient dose [3]. Although tube current can be manually controlled, most operators use automated tube current modulation (also known as automated exposure control) for most applications. This computer software, which has various names depending on the specific vendor (ie, CARE Dose 4D on Siemens, Dose-Right on Phillips, Auto mA/Smart mA on GE, and SURE Exposure 3D on Toshiba), automatically increases the mAs in those parts of the body with the greatest attenuation and decreases the mAs in those parts of the body with lower attenuation. As a result, the software usually increases the mAs in the shoulder and hips (which have relatively more attenuation) and decreases the mAs in the abdomen and thorax (which have relatively less attenuation) [1,7]. In most cases, automated tube current modulation reduces patient dose and minimizes photon starvation artifacts.

Automated tube current modulation works on different principles depending on the specific vendor, although a combination of 4 different strategies is generally used: Patient size modulation varies the mAs on the basis of a global evaluation of the overall size of the patient as seen on the scout radiograph. Z-axis modulation changes the mAs constantly along the z-axis of the patient depending on the patient attenuation at each point, as determined using the scout image. Angular (x, y) modulation changes the mAs as the x-ray tube rotates $360^\circ$ around the patient to account for different attenuations in different projections of the x-ray beam [1]. Combined x, y, and z modulation adjusts the mAs in all 3 axes on the basis of the patient’s attenuation [8]. The user must define a minimum acceptable image quality, the definition of which will vary depending on the specific vendor and system. For example, GE users must specify a noise index value that is maintained as a constant on every image in a series, while Siemens users must specify an image quality reference mAs, the mAs that would be used for an average-sized patient [1].

In almost all cases, automated tube current modulation should be enabled and used by all users, although some significant caveats and exceptions must be kept in mind:

- Small or pediatric patients, when incorrectly positioned or improperly centered within the CT gantry, can result in excessively noisy images with artificially low mAs. Not only must technologists be properly taught to correctly position patients, but it may also be helpful to modify the noise index or increase the reference mAs for smaller patients (while still capping the maximum mAs to avoid inadvertently high radiation doses) [8,9].

- Many institutions do not use automated tube current modulation when performing scans of the head, particularly in pediatric patients, because of the difficulty in correctly placing the head in the isocenter of the gantry. A small deviation of the head from the isocenter can result in a significant increase in image noise.

- Patients’ arms should be out of the field of view for both scout-image and axial-image acquisition. Any discrepancy in the position of the arms between the scout and axial images can adversely affect the algorithm’s function and theoretically increase dose. If
Pediatric modulation settings should not be used for extremely large or obese patients, there may be a risk that as the pitch increases on these two vendors’ scanners (with resultant increases in noise), the scanner software will automatically increase the tube current proportionally to compensate for the increased pitch. As long as the effective mAs or mAs per slice remains constant, the noise level (and dose) will remain unchanged [11].

KvP

Reducing kVp can be an effective means of reducing the radiation dose imparted during an examination. As a general rule of thumb, the radiation dose changes with the square of kVp, and a reduction in kVp from 120 to 100 reduces radiation dose by 33%, while a further reduction to 80 kVp can reduce dose by 65% [12,13]. As a further advantage, particularly in angiographic studies, in which vascular contrast is critical, reductions in kVp result in an increased attenuation of iodine (even with a constant dose of contrast) as a result of the photoelectric effect and approaching the k-edge of iodine, potentially improving contrast-to-noise ratios. Nevertheless, unlike reductions in mAs, which have a linear and relatively predictable effect on image noise and contrast-to-noise ratios, decreases in kVp can result in nonlinear, exponential increases in image noise, often necessitating a concomitant increase in mAs to preserve image quality. Regardless, several studies evaluating the use of low kVp in a number of different protocols (angiography, venous phase studies, cardiac and coronary studies, etc) have shown that such reductions can markedly reduce radiation doses, while still preserving acceptable image quality [12,14,15].

It is now widely accepted that low-kVp protocols (100, 80, and even 70 kVp) are most useful in thin, nonobese patients, particularly in vascular or angiographic studies, in which improvements in the attenuation or conspicuity of iodine can be helpful in terms of image contrast-to-noise ratios (Fig. 2) [16]. Most scanners offer no tools to correctly select an appropriate kVp, and the decision to decrease kVp must be made at the discretion of the radiologist, taking into account the type of study being performed (ie, arterial or venous phase acquisition), the portions of the body being imaged, and a global assessment of patient body habitus. Although some sites have attempted to use algorithms that base the decision on a patient’s body mass index, this may often be unsatisfactory because it is the distribution of the patient’s body fat that has the ultimate impact on image quality. Data regarding the correct kVp selection in any given patient are still evolving, and each practice must determine its comfort level with low-kVp techniques by slowly instituting such changes over time. Although the Society of Cardiovascular Computed Tomography has published recommendations regarding kVp levels in patients undergoing cardiac CT, including a suggested kVp

In addition, users should be aware of the terms effective mAs (used by Siemens) and mAs per slice (used by Philips), which are relevant to any discussion of tube current. These two essentially synonymous terms, defined as tube current × rotation time/pitch, are used to convey the concept that as the pitch increases on these two vendors’ scanners (with resultant increases in noise), the scanner software will automatically increase the tube current proportionally to compensate for the increased pitch. As long as the effective mAs or mAs per slice remains constant, the noise level (and dose) will remain unchanged [11].

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of 100 in patients weighing ≤90 kg, no similar guidelines have been published for conventional body and neurologic imaging applications [17].

Notably, the increase in image noise associated with lower kVp techniques must be balanced by an increase in the tube current. The reference mAs (on Siemens scanners) or noise index (on GE scanners) must be appropriately altered to increase tube current, but the tube current should not be raised too much because the dose savings from a lower kVp would be canceled out. Instead, the tube current must be mildly increased, preserving image quality (with perhaps minimal degradation), while still ensuring some dose savings [16].

**RECONSTRUCTION ALGORITHM**

Traditionally, CT reconstruction algorithms have used filtered back projection (FBP). The mathematical algorithms underlying FBP assume a “nonexact” relationship between the projection data acquired at the CT scanner and the data displayed in the final image. As a result, FBP removes very little noise from the final image, is limited in the use of data with lower contrast-to-noise ratios, and generally requires studies to be performed with larger radiation doses to be of diagnostic quality [1,18].

As a result, each of the major CT vendors now offers its own proprietary iterative reconstruction algorithms as a replacement for FBP. Iterative reconstruction, now increasingly feasible because of better algorithms and improvements in hardware, involves the use of computationally intensive iterative “correction loops,” which repeatedly compare the reconstructed data with an “ideal” image, until an acceptable final image is created [1]. Accordingly, these iterative techniques allow the reconstruction of images with improved image quality, even in cases in which the imparted radiation dose and contrast-to-noise ratios are low. A number of studies using each of the different vendors’ software packages have concluded that iterative reconstruction is a viable option in terms of both reducing radiation dose and improving image quality [19-22]. For example, a study by Kaza et al [23] found that iterative reconstruction techniques could lower radiation dose by up to 30% in CT enterographic studies, while still maintaining acceptable image quality.

Overall, when iterative reconstruction is available on a group’s CT scanner, there is little downside to its implementation over more traditional FBP techniques in terms of either diagnostic quality or radiation dose. However, regardless of which vendor’s iterative reconstruction algorithm is used, there is no simple way to determine the extent to which dose can subsequently be reduced. In general, once iterative reconstruction is instituted, dose should gradually be decreased over time (via changes in the acceptable noise index or reference mAs on the system’s automatic exposure control software) until reaching a point at which the radiologist no longer feels comfortable with the image quality. Importantly, the images acquired with iterative reconstruction can have a smoothed, “artificial” appearance compared to FBP images, necessitating some acclimatization on the part of the interpreting radiologist.

**PATIENT POSITIONING**

In a study by Li et al [24] in 2006, 95% of patients in the series were found to be improperly centered within the scanner gantry by the CT technologist. Although this may seem to be of only minimal importance, improper positioning can have a significant impact on both image noise and patient surface dose. Habibzadeh et al [25] found improper centering of phantoms in the scanner gantry to be associated with an average surface dose increase of 23% and an image noise increase of 7%. Abnormal positioning in both the horizontal and vertical axes can have a similar impact, as a study by Kaasalainen et al [26] using child-sized phantoms found an increase in the breast surface dose of 16% and in the thyroid gland of 24% for phantoms improperly positioned in the vertical axis. Just as important, small variations in patient positioning can have a marked effect on image noise, as vertical displacement of a patient by just 6 cm can result in an increase in noise of 22%. Needless to say, this can have a dramatic secondary effect on radiation dose if automatic exposure control is used because radiation doses can be markedly increased by the software to compensate for this increased noise [26-28]. Although patient positioning must be carefully monitored in all patients, extra attention should be paid to this issue in pediatric patients and patients with smaller body habitus because small errors in positioning such patients are more likely to have a significant impact on both noise and dose. Moreover, although correct patient centering is...
an important factor regardless of the type of scanner being used, it is particularly important when using dual-source CT scanners. Given the scanner geometry of dual-source mode (with the second source typically having a smaller field of view), the dose penalty from centering the patient away from the scanner isocenter can be greater than with conventional single-source acquisitions.

The primary cause for this association between improper patient centering and increased patient radiation dose is the use of bowtie filters in modern CT scanners. These filters normally compensate for patient attenuation during tube rotation, resulting in increased x-ray intensity to the thickest parts of the patient (i.e., the center of the patient) and decreased intensity to the thinnest parts of a patient (i.e., the patient surface). The bowtie filter functions under the assumption that the patient is correctly centered within the gantry [26]. In optimal circumstances, the bowtie filter can reduce patient dose by up to 50% [24]. However, when the patient is improperly positioned, the mathematical assumptions underlying this filter break down, and doses to the surface of the body increase.

In general, incorrect positioning in the horizontal axis should not be a problem in the vast majority of patients. Virtually every recent scanner has a laser guidance system that should allow the patient to be correctly placed at the isocenter of the scanner, and all technologists should be trained to correctly use this guidance mechanism. However, correct positioning in the vertical axis can be more tricky: in patients who are markedly scoliotic or unable to lie completely flat in the gantry, portions of the patient, by necessity, will inevitably be positioned outside of the isocenter with vertical displacement. In addition, technologists must be very careful when placing pillows under the patient, as excessive elevation of the patient’s head with pillows may cause portions of the patient’s upper body and head to lie above the scanner’s isocenter. Unfortunately, an error in positioning may not be immediately apparent to either the radiologist or the technician because there may not be a dramatic change in the volume CT dose index: increased surface dose to portions of the body above the isocenter may be balanced by decreased surface dose to portions of the body below the isocenter, and the volume CT dose index may seem unchanged, although portions of the body have received markedly increased skin dose. As a result, this error must be prevented through active training of technologists, who must be instructed not only to correctly position the patients but also to turn off automatic tube current modulation in any case in which there is difficulty in properly positioning the patient at the isocenter of the scanner.

**SCAN RANGE**

For many CT applications, significant reductions in scan range may be neither possible nor desirable. Nevertheless, the scan range should be reduced to the needed minimum for any examination, particularly when imaging structures such as the heart, for which an increased scan range is unnecessary (Fig. 3). Although cardiac studies are certainly the most obvious applications in which a limited scan range can be used, a small scan range may be possible in many traditional body imaging applications as well: Patel et al [29] found that using a small scan range (from the top of the aortic arch to the bottom of the heart) in CT pulmonary angiographic studies can allow diagnosis of pulmonary embolism without any loss of sensitivity but with a reduction in radiation dose of 48%.

Unfortunately, even if the scan range is kept to the absolute minimum, helical CT requires the acquisition of raw data on either end of the scan length, outside of the intended scan range, to acquire sufficient projection data. This can particularly be a problem with higher pitches and can result in additional dose as a result of wasted dose at the margins of the scan range. However, this effect can be mitigated with the use of dynamic z-collimation. Unlike a fixed collimator, which is fully open throughout the scan range, the dynamic z-collimator opens only that portion of the collimator entering the scan range (while at the beginning of the scan) and closes the proximal edge of the collimator while leaving the scan range, thus minimizing dose outside the intended scan range and resulting in a more rectangular dose profile. Dose savings with the use of dynamic z-collimation can be incremental, although the exact extent of savings will depend on the CT pitch [30,31].

**RECONSTRUCTED SLICE THICKNESS**

As the reconstructed slice thickness decreases, the number of photons within each voxel also decreases, resulting in increased image noise. To maintain constant noise levels within an image with a smaller slice thickness, the radiation dose must be consequently increased [3]. With the introduction of automatic tube current modulation, the interaction between the user-specified level of acceptable noise, reconstructed slice thickness, and radiation dose can be quite complex and will undoubtedly vary depending on the CT scanner and the specific automatic tube current modulation software being used [32].
general, the larger the reconstructed slice thickness used, the lower the patient dose. It is worth repeating, however, that the reconstructed slice thickness cannot be smaller than the acquired slice thickness.

**PITCH**

Pitch in the multidetector, spiral CT era is defined as table travel per rotation divided by beam collimation. Pitch \( <1 \) suggests overlap between adjacent acquisitions, pitch \( >1 \) implies gaps between adjacent acquisitions, and pitch of \( 1 \) suggests that acquisitions are contiguous, with neither overlap nor gaps [7]. A smaller pitch, with increased overlap of anatomy and increased sampling at each location, results in an increased radiation dose. Alternatively, a larger pitch implies gaps in the anatomy and hence lower radiation dose. As a result, if all other parameters are unchanged, increasing pitch reduces radiation dose in a linear fashion [7,33]. Low-pitch technique is associated with less image noise, fewer artifacts, and improved signal-to-noise and contrast-to-noise ratios [34]. For routine body CT protocols, pitch \( >1 \) is generally acceptable with no compromise to image quality. However, using higher pitches (pitch \( >1.5 \)) can result in interpolation artifacts and image noise that is typically unacceptable. For most cardiac CT applications, pitch values \( <0.5 \) are routinely used, resulting in higher doses in such studies. These lower pitch values are used to overcome motion artifacts and are also stipulated by the reconstruction algorithms in use. As mentioned earlier, one major caveat to this discussion of the normal relationship between pitch and radiation dose is the fact that certain scanners use the concept of effective mAs or mAs per slice [33]. In these cases, the effect of pitch on dose is negated by proportional increases in tube current to maintain similar image noise.

Although higher pitch values have traditionally been thought to be associated with poor image quality, this is no longer true with the latest dual-source CT technology. A study by Apfeltraser et al [35] using pitch values \( >3 \) on a dual-source scanner for vascular studies suggested that image quality was more than acceptable, with radiation dose savings as high as 45% to 50%. The newest dual-source scanners offer high-pitch or FLASH mode acquisitions with pitch values \( >3 \), consequently decreasing scan times precipitously and potentially reducing radiation doses significantly. These high-pitch modes markedly reduce scan times and eliminate many motion-related artifacts, particularly those previously seen with cardiac imaging [36]. However, it should be noted that there are significant caveats that should induce some caution before using high-pitch or FLASH acquisitions on every case: Although the data suggest that FLASH acquisitions offer a significant dose reduction for vascular studies in the chest, for which retrospective or prospective gating would otherwise be used to prevent motion-related artifacts, it is unclear whether there is actually any true dose savings in nonvascular studies in the chest or abdomen. In particular, in such cases, some systems will attempt to maintain a constant effective mAs (Siemens) or mAs per slice (GE) (tube current \( \times \) rotation time/pitch) and will consequently increase the tube current to compensate for the increased pitch. In addition, in patients who are large, the system may attempt to increase the tube current but may be precluded from doing so as a result of a capped or limited mAs, and image noise may actually rise. Accordingly, in nonvascular studies in which scan speed is not as important, using a high-pitch technique may offer little dose savings and potentially worse image quality. That said, high-pitch techniques on these scanners still offer significant value in nonvascular studies in children, in whom scan speed is critical.

**CONCLUSIONS**

As both the general public and clinicians have become increasingly aware of the concerns associated with multidetector CT–related radiation dose, it has become critical for radiologists to better understand the CT scanner technology they work with. In particular, a detailed understanding of a few basic CT scan parameters is essential, and knowledge of how to manipulate these parameters to produce diagnostic images at lower doses is critical for safe imaging.

**TAKE-HOME POINTS**

- Several multidetector CT parameters can be changed to reduce radiation dose, including detector configuration, tube current, kVp, reconstruction algorithm, patient positioning, scan range, reconstructed slice thickness, and pitch.
- Although all users should generally use automated tube current modulation, users must be careful about using this software in obese patients, children, and patients with metallic hardware.
- Low-kVp protocols may be most useful in thin, non-obese patients, particularly in vascular or angiographic studies.
- Iterative reconstruction techniques allow the reconstruction of images with improved image quality, even in cases in which the imparted radiation dose and contrast-to-noise ratios are low.
- The scan range should be reduced to the needed minimum for any examination.
- Incorrect positioning of patients in the scanner gantry can be associated with significant radiation dose penalties.

**REFERENCES**


