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Emerging Techniques for Dose Optimization in Abdominal CT[1]

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Recent advances in computed tomographic (CT) scanning technique such as automated tube current modulation (ATCM), optimized x-ray tube voltage, and better use of iterative image reconstruction have allowed maintenance of good CT image quality with reduced radiation dose. ATCM varies the tube current during scanning to account for differences in patient attenuation, ensuring a more homogeneous image quality, although selection of the appropriate image quality parameter is essential for achieving optimal dose reduction. Reducing the x-ray tube voltage is best suited for evaluating iodinated structures, since the effective energy of the x-ray beam will be closer to the k-edge of iodine, resulting in a higher attenuation for the iodine. The optimal kilovoltage for a CT study should be chosen on the basis of imaging task and patient habitus. The aim of iterative image reconstruction is to identify factors that contribute to noise on CT images with use of statistical models of noise (statistical iterative reconstruction) and selective removal of noise to improve image quality. The degree of noise suppression achieved with statistical iterative reconstruction can be customized to minimize the effect of altered image quality on CT images. Unlike with statistical iterative reconstruction, model-based iterative reconstruction algorithms model both the statistical noise and the physical acquisition process, allowing CT to be performed with further reduction in radiation dose without an increase in image noise or loss of spatial resolution. Understanding these recently developed scanning techniques is essential for optimization of imaging protocols designed to achieve the desired image quality with a reduced dose.

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Abbreviations: ASIR = adaptive statistical iterative reconstruction, ATCM = automated tube current modulation, $CTDI_{vol}$ = CT dose index volume, FBP = filtered back projection, MBIR = model-based iterative reconstruction, SAFIRE = sinogram-affirmed iterative reconstruction

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See discussion on this article by Mahesh (pp 17–18).

ONLINE-ONLY SA-CME LEARNING OBJECTIVES

After completing this journal-based SA-CME activity, participants will be able to:

- Describe the use of automated tube current modulation and the appropriate image quality parameter to reduce radiation dose.
- List the advantages and practical implications of lowering tube voltage to reduce radiation dose.
- Discuss the use of iterative image reconstruction to develop scanning protocols with reduced radiation dose but no reduction in image quality.

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TEACHING POINTS

See last page

Introduction

Although the true risks from medical radiation exposure have been questioned, the increasing cumulative dose in patients who require multiple imaging studies has led to renewed efforts to reduce radiation dose (1–3). With computed tomographic (CT) scans accounting for approximately one-half of all medical radiation exposure, there is a pressing need to investigate dose reduction techniques and to implement these emerging techniques in routine clinical practice (4,5). The optimization of scanning protocols requires a collaborative effort between radiologists, medical physicists, and CT technologists, and a full recognition of the potential decrease

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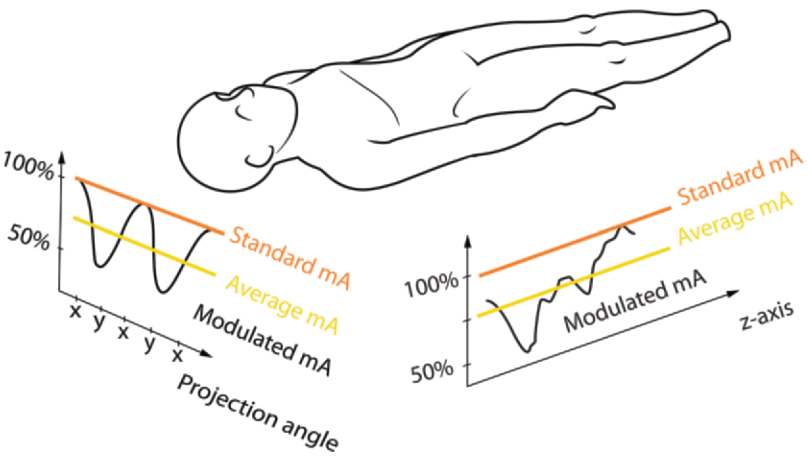


Figure 1. Drawing illustrates the modulation of tube current (in milliamperes *[mA]*) along the longitudinal (z) and horizontal (x-y) axes of the patient.

in image quality when radiation dose is reduced. Recent advances in CT scanning techniques, such as automated tube current modulation (ATCM), use of optimal tube voltage, and improved utilization of iterative image reconstruction, have allowed the reduction of CT radiation dose while maintaining diagnostic image quality. In this article, we discuss the use of these emerging techniques for the development of optimal imaging protocols aimed at achieving the desired image quality with a reduced dose.

Image Quality Parameters and CT Dose

The main influences on the perceived quality of CT images are *(a)* noise (standard deviation of CT numbers within a region of interest as well as the spatial frequencies of the noise [noise power spectrum]), *(b)* contrast-to-noise ratio (CNR) (the ability to distinguish CT number differences from background noise), *(c)* spatial resolution (the ability to resolve small objects that are adjacent to each other), and *(d)* image artifacts. The ideal CT image will have the least noise, the highest contrast and spatial resolution, and no artifacts. However, improving the image quality has conventionally been associated with increasing the radiation dose. For example, to reduce image noise by a factor of two, a fourfold increase in CT dose is required; to double the spatial resolution, an eightfold increase in CT dose is required; and to reduce artifacts from photon starvation, at least some increase in CT dose is required (6).

As mentioned earlier, a number of recent advances have allowed radiation dose reduction while maintaining diagnostic image quality. These include advances in *(a)* CT scanner hardware (higher-power x-ray source allowing better x-ray beam filtration, improved detector capability), *(b)* scanning technique (ATCM, optimized tube voltage), and image reconstruction (iterative reconstruction) (7).

Automated Tube Current Modulation

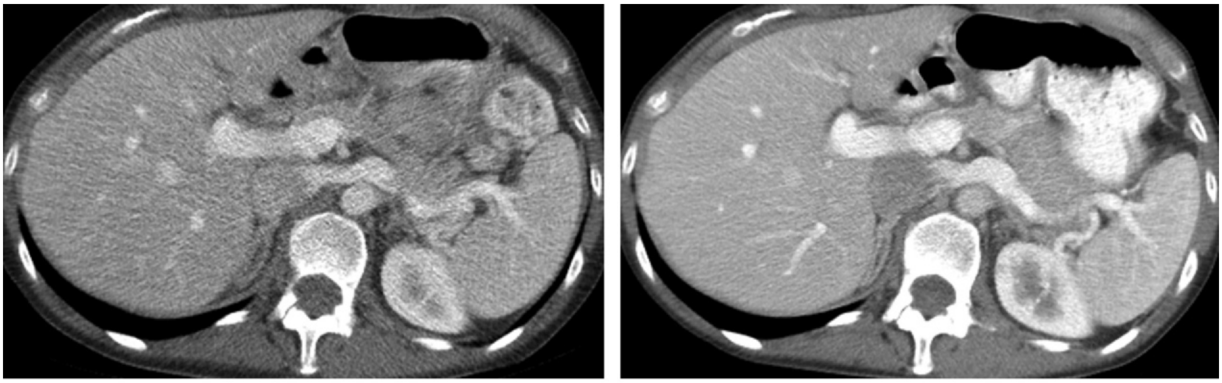
At a constant tube current (in milliamperes), image noise is influenced by patient attenuation; hence, areas on an image that represent thicker portions of the body such as the thoracic inlet and pelvis will have higher noise and reduced quality. ATCM varies the tube current during image acquisition to account for differences in patient attenuation, ensuring a more homogeneous image quality (8–10). The two components of tube current modulation (Fig 1) include *(a)* longitudinal (z-axis) modulation, in which the milliamperage is varied along the longitudinal axis of the patient such that lower-attenuation portions of the body will be imaged with lower milliamperage than will higher-attenuation portions, and *(b)* angular (x-y–axis) modulation, in which the milliamperage is varied during x-ray tube rotation between anteroposterior and lateral projections such that it is reduced in the direction of the lower-attenuation projection.

Teaching Point

With use of ATCM, it is important to determine the desired image quality for each scanning protocol. This serves as the reference standard on the basis of which the milliamperage is modulated. The desired image quality can be defined in terms of either a predefined reference image quality or estimated image noise.

ATCM based on Reference Image Quality

CT systems that make use of reference image quality for ATCM include Quality Reference (effective tube current–time product, expressed in milliampere-seconds) (Siemens Healthcare, Forchheim, Germany), Reference Case (Philips Healthcare, Best, the Netherlands), and Image Quality Level (Toshiba Medical Systems, Tokyo, Japan) (11). The reference image quality defines the effective milliamperage required to produce the desired image quality in a “reference” patient



a. b.

Figure 2. Axial 5-mm-thick reconstructed image from a CT study performed with a noise index of 35 at 2.5-mm thickness **(a)** shows increased noise compared with the corresponding image from a prior study performed with a noise index of 22 at the same thickness **(b)**. The CT dose index volume ($CTDI_{vol}$) values were 1.88 and 4.67 mGy, respectively.

(adult patient weighing 80 kg, pediatric patient weighing 20 kg) (12). The scanner then adapts the tube current according to the patient's size and attenuation profile (size, shape, and density) to obtain images with a quality similar to that of the reference images at the selected reference image quality (eg, image quality at 30, 60, and 120 mAs). Selecting a lower reference image quality reduces the radiation dose while increasing image noise.

ATCM based on Estimated Image Noise

CT systems that make use of estimated image noise with ATCM include Noise Index (GE Healthcare, Waukesha, Wis) and Standard Deviation (Toshiba Medical Systems) (11). The tube current is modulated based on the patient's attenuation profile as calculated from the scout image, to obtain images with constant noise closer to the prescribed noise index (13). Selecting a higher noise index allows more noise on resultant images; hence, a lower tube current is used, resulting in lower radiation dose. Conversely, selecting a lower noise index results in a higher radiation dose (Fig 2). The following equation can be used to compute the dose for different noise indices:

$$\frac{Dose_2}{Dose_1} = \left[\frac{NI_1}{NI_2}\right]^2, \qquad (1)$$

where $Dose_2$ and $Dose_1$ are the doses for conditions 2 and 1, respectively, and NI_1 and NI_2 are the corresponding noise indices. For example, doubling the noise index will decrease the dose by a factor of four, so long as the minimum and maximum milliamperage settings are not limited (ie, the tube current is not restricted at either the minimum or maximum milliamperage).

Noise Index and Section Thickness

When selecting a noise index for a CT protocol, it is important to know the reconstructed section thickness at which the noise index is applied (13), since to maintain the same image noise (constant noise index), the radiation dose would have to be increased for the reconstruction of thin sections compared with thick sections. The degree of tube current modulation is calculated based on the noise index applied at the first prospectively reconstructed section thickness. The following equation can be used to compute the noise index for scans with the same dose but different section thicknesses:

$$\frac{NI_2}{NI_1} = \sqrt{\frac{t_1}{t_2}}, \qquad (2),$$

where NI_2 and NI_1 are the noise indices for conditions 2 and 1, respectively, and t_2 and t_1 are the corresponding section thicknesses. If thin (0.625-mm) sections are prospectively reconstructed that could serve for three-dimensional (3D) or multiplanar reconstruction, and thick (5-mm) sections are used for image viewing, a higher noise index should be applied to the thin sections to avoid excessively high tube current and dose (Fig 3). For example, a study performed with a noise index of 15 and 5-mm-thick reconstructed images would have approximately the same radiation dose as a study performed with a noise index of 40 and 0.625-mm-thick reconstructed images. Appropriate selection of the noise index and the reconstructed section thickness at which the noise index is applied is essential for achieving optimal image quality with reduced dose when dose-modulated tube current is used (13). The tube current modulation does not depend on the choice of prospective section thickness when a reference image quality is used (14).

Teaching Point

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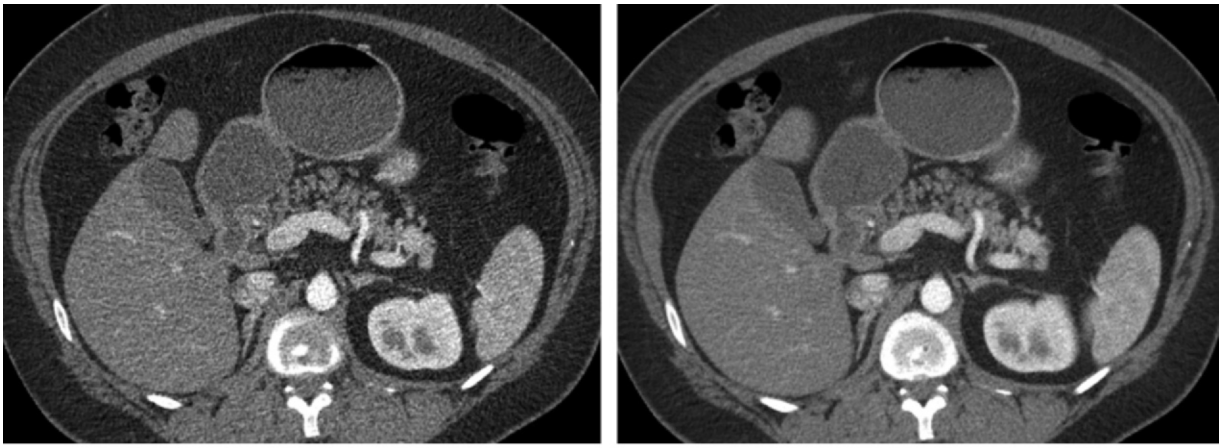
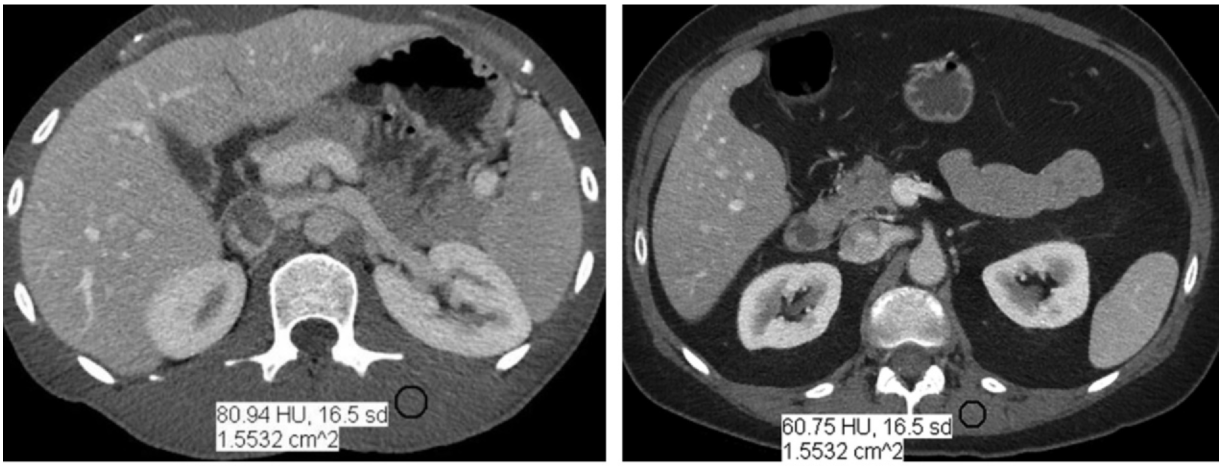


a. b.

Figure 3. **(a)** Axial 0.625-mm-thick reconstructed image from a CT study performed with a noise index of 30 at 0.625-mm thickness demonstrates significant noise. **(b)** Axial 5-mm-thick reconstructed image from the same study shows reduced noise and improved image quality.



a. b.

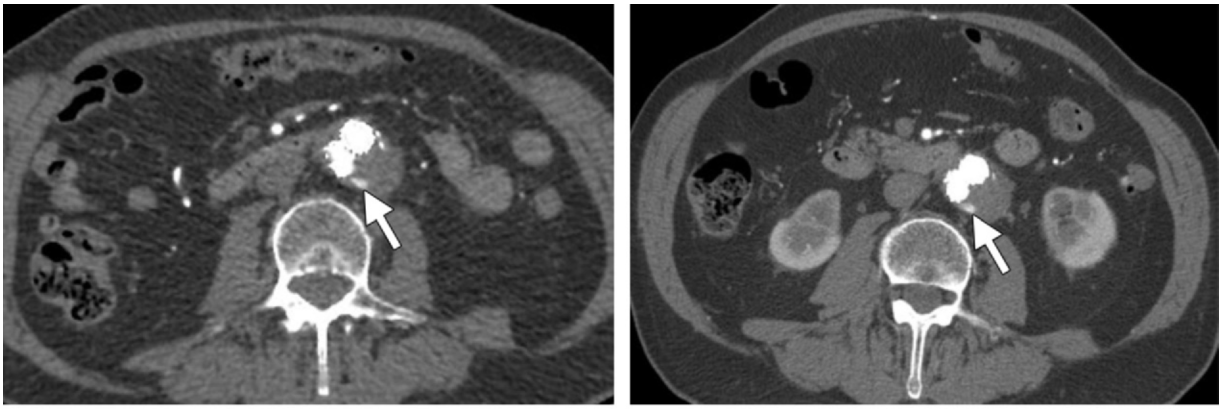
Figure 4. **(a)** Axial 5-mm-thick reconstructed image from a CT study that was performed with a noise index of 30 at 0.625-mm thickness. **(b)** Axial 5-mm-thick reconstructed image from a CT study performed with the same noise index in a different patient with more intraabdominal fat shows improved image quality compared with **a**.

ATCM Modification for Patient Habitus

At a constant noise level, perceived image quality increases with increasing patient diameter; hence, images obtained in thin patients are usually perceived to be less acceptable than those obtained in larger patients, despite similar measured image noise (Fig 4). This phenomenon is due to increased intraabdominal fat in larger patients, which provides inherent contrast around organs and allows better toleration of image noise. ATCM that makes use of reference image quality operates to maintain a constant image quality, and as such should produce images with less noise in smaller patients and more noise in larger patients (14). With ATCM that makes use of a noise index, a constant image noise is maintained regardless of patient diameter, which can reduce image quality in thin patients (14). To achieve acceptable image quality in thinner patients, scans with noise less than that seen in average-sized patients can be performed with either of the following modifications: *(a)* using a lower noise index for thin patients compared with average-sized patients, or *(b)* limiting the minimum milliamperage of higher-level automated milliamperage selection so that the milliamperage does not become too low. Conversely, in obese patients, higher noise is tolerated for acceptable image quality, so that a higher noise index or limiting the maximum milliamperage of automated milliamperage selection can be used to reduce the radiation dose.

Modification for CT Protocol based on Required CNR

For CT scanning protocols used for the evaluation of soft tissues (eg, detection of liver metastasis),



a. b.

Figure 5. CT angiographic images (0.625-mm thickness) obtained with noise indices of 60 **(a)** and 30 **(b)** clearly depict a high-contrast endoleak (arrow). $CTDI_{vol}$ values were 2.76 and 14.24 mGy, respectively.

which require differentiation of small CT number differences from background noise (ie, superior low-contrast detectability), no modification of the reference image quality or noise index from optimized standard protocols is suggested, since it could compromise diagnostic accuracy (15). However, for CT protocols used for the evaluation of large contrast differences (eg, CT angiography), increased image noise does not usually affect diagnostic accuracy because there is a large contrast difference between evaluated structures, so that lower reference image quality or a higher noise index can be used, thereby allowing a reduction in radiation dose (Fig 5) (12,16). Protocol-specific optimization of reference image quality or noise index based on the structures being evaluated allows optimal radiation dose reduction.

Patient Centering with Use of ATCM

Proper centering of the patient in the field of view when obtaining the scout image is essential for good estimation of the patient's attenuation profile and, hence, of proper modulation of milliamperage to achieve uniform image noise (17). Proper centering also allows the higher-intensity x-rays passing through the isocenter of the bowtie filter to pass through the thicker central portion of the patient's body, thereby reducing image noise (Fig 6).

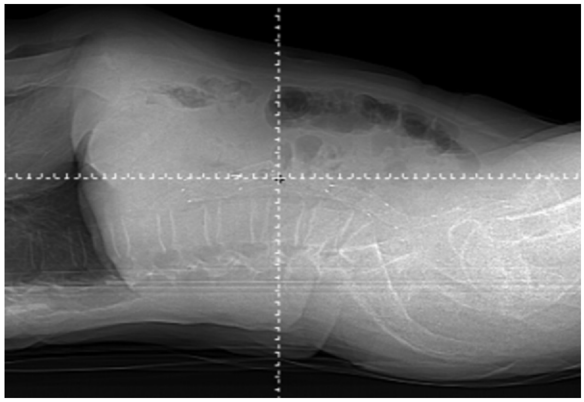


Figure 6. CT scout image with a grid overlay helps confirm that the patient is properly centered.

Optimal Tube Voltage

Tube Voltage and Radiation Dose

An increase in x-ray tube voltage (in kilovolts) increases the tube output and the effective energy of the x-ray beam, resulting in improved penetrating power of the x-ray beam and reduced image noise, albeit at the expense of increased radiation dose. Unlike tube current, which has a linear relationship with radiation dose, tube voltage has an exponential effect on radiation dose. At a constant milliamperage, effective dose increases by approximately 50% if the kilovoltage is increased from 120 kV to 140 kV and decreases by approximately 65% if the kilovoltage is reduced from 120 kV to 80 kV (10,18,19).

Low Tube Voltage Scanning

Routine abdominal CT studies are conventionally performed at 120 kV, which is optimal for soft-tissue imaging. However, when iodinated contrast material is administered and evaluation of iodinated structures is the primary task (as at CT angiography), lowering the kilovoltage from 120 kV to 100 kV or even 80 kV is desirable because the effective energy of the x-ray beam will be closer to the k-edge of iodine, resulting in a higher attenuation for the iodine (18,20,21). This increases both image contrast and the CNR, despite the increase in image noise associated with lower-kilovolt scanning (Fig 7). The magnitude of the increase in image noise with a reduction in kilovoltage is higher for larger patients than for

Teaching Point

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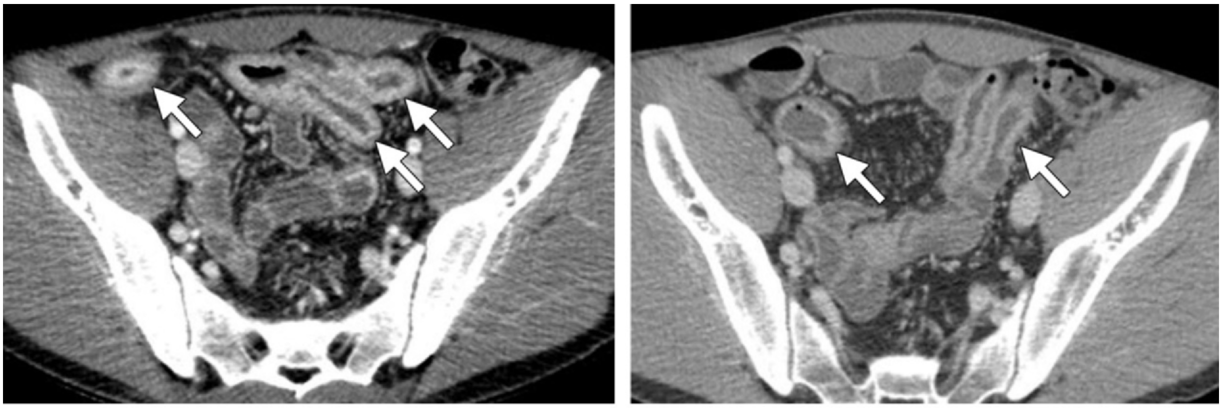


a. b.

Figure 7. CT images obtained at 80 kV **(a)** and 120 kV **(b)** in a patient with Crohn disease show that, despite the increased noise, the 80-kV image demonstrates a better CNR and allows clearer visualization of the enhancing bowel wall (arrows).

smaller patients due to reduced penetration of the x-ray beam in larger patients (22). This would negate the potential advantage of increased contrast achieved by reducing kilovoltage in larger patients, leading to an overall reduced CNR (23). Hence, lowering the kilovoltage from 120 kV to 100 kV or 80 kV is ideally suited for medium-sized and small patients when iodinated structures are being evaluated (23). Iterative reconstruction used in conjunction with low-kilovolt scanning can offset the increased noise associated with a reduction in kilovoltage and improve image quality.

Kilovoltage, Milliamperage, and Noise Index: Interrelationships and Impact on Dose and Image Quality

When the kilovoltage is reduced, the milliamperage needs to be increased to compensate for the reduced energy of the x-ray beam to achieve adequate patient penetration. For a given CT protocol, reducing the kilovoltage without increasing the milliamperage could produce images with excessive noise and reduced quality (19). When an image quality metric such as noise index is used for tube current modulation, lowering the kilovoltage at a fixed noise index could paradoxically lead to an increase in radiation dose for the scan rather than a reduction in dose. At a fixed noise index used to obtain images at a constant noise level, the automated dose modulation algorithm increases the tube current to compensate for the increased noise resulting from lowering the kilovoltage, resulting in a net increase in the radiation dose. Thus, when lowering kilovoltage to achieve dose reduction and an improved CNR, image noise cannot be held constant, and a higher noise index should be selected compared with that used for 120-kV scanning protocols (23).

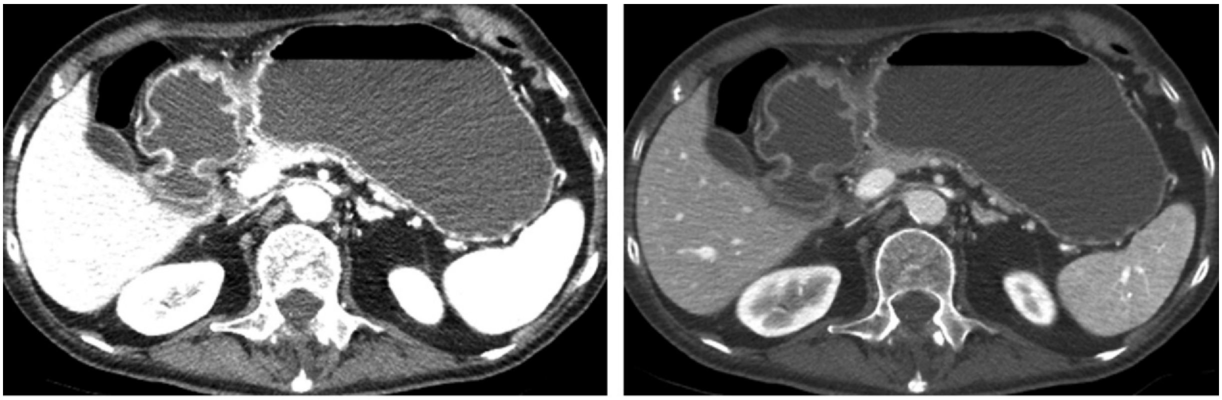
Automatic Kilovoltage Selection

Automatic kilovoltage selection has recently become available for clinical use (Care kV, Siemens Healthcare). With this tool, the scanner calculates the patient's attenuation profile from the scout image, and the user can select the contrast gain setting and acceptable image noise required for the diagnostic task (detection of high-contrast structures [eg, renal calculi at CT angiography] versus low-contrast structures [liver lesions]). On the basis of the patient's attenuation profile and the required imaging task, the scanner generates patient- and task-specific milliampere-second curves for varying kilovoltage levels (80, 100, 120, and 140 kV) that would generate the desired image quality for the body region being scanned. The kilovoltage level at which the greatest radiation dose reduction can be achieved can then be chosen to maximize dose reduction while maintaining image quality (23,24).

Practical Aspects of Low-Kilovolt Scanning

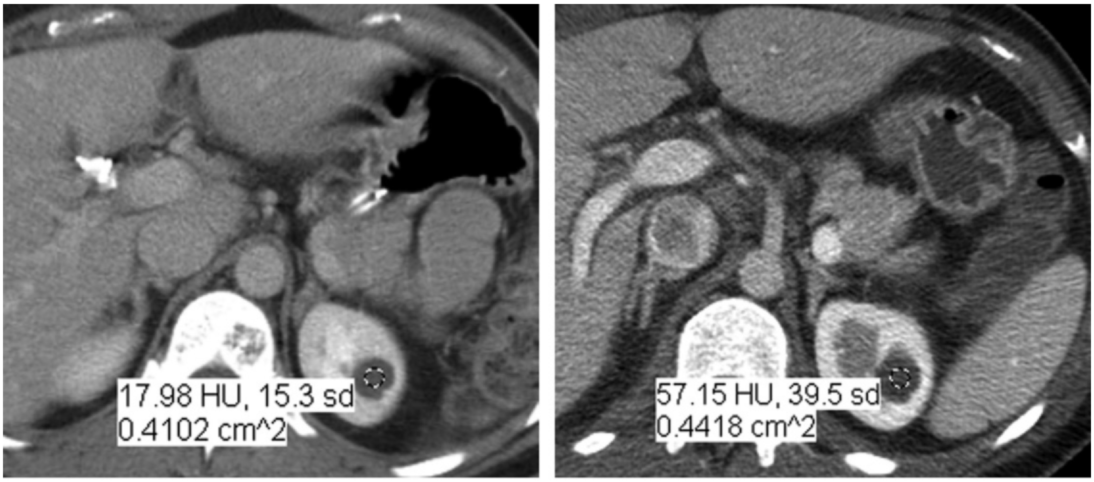
Image Viewing.—Because of higher image contrast at lower kilovoltage, the window width and level need to be increased during viewing to maintain an imaging appearance similar to that of 120-kV images. Viewing images at a wider window width also reduces perceived image noise, with resultant perceived improvement of image quality (Fig 8) (21).

Hounsfield Unit Measurements.—Iodinated structures have higher Hounsfield unit values at lower kilovoltage, a fact that could lead to increased incidence of pseudoenhancement within renal or hepatic cysts surrounded by enhancing parenchyma, with simple cysts having an attenuation higher



a. b.

Figure 8. Abdominal CT images obtained at 80 kV viewed at a typical window width and level of 400/40 **(a)** and an optimized window width and level of 950/150 **(b)** show how the use of increased window width and level when viewing an image obtained at low-kilovolt scanning reduces perceived image noise, with resultant perceived improvement of image quality.



a. b.

Figure 9. Pseudoenhancement in a renal cyst. **(a)** CT image obtained at 120 kV shows a lesion in the kidney with a measured attenuation of about 18 HU. **(b)** On a subsequent CT image obtained at 80 kV, the lesion has an attenuation of nearly 60 HU. The lesion was confirmed to be a simple cyst at MR imaging.

than the conventional thresholds for nonenhancing lesions (Fig 9) (25). Lesion characterization based on Hounsfield units standardized to 120-kV scans would not be applicable.

Beam-hardening Artifacts.—The reduced penetration of the x-ray beam at lower kilovoltage can cause increased incidence of beam-hardening artifacts, especially adjacent to high-attenuation structures such as bone and metal hardware.

Intravenous Contrast Material Reduction.—Given equivalent radiation doses, the CNR for scans performed at 80 kV is more than double that for scans performed at 120 kV. In older patients, for whom the risk of radiation-induced cancer is minimal and the risk of contrast material–induced nephropathy is higher, the amount of iodinated contrast material administered can be reduced by scanning at a lower kilovoltage without radiation dose reduction (26).

Higher-Kilovolt Scanning.—In extremely large patients, increasing the kilovoltage from 120 kV to 140 kV is preferable because it allows better penetrating power of the x-ray beam, resulting in reduced image noise and improved image quality (19).

Iterative Image Reconstruction

Conventional noise reduction filters applied after scanning reduce image noise but also reduce spatial resolution and contrast. With the development of iterative image reconstruction, it is now possible to selectively identify and reduce image noise while maintaining image contrast and resolution.

Teaching Point

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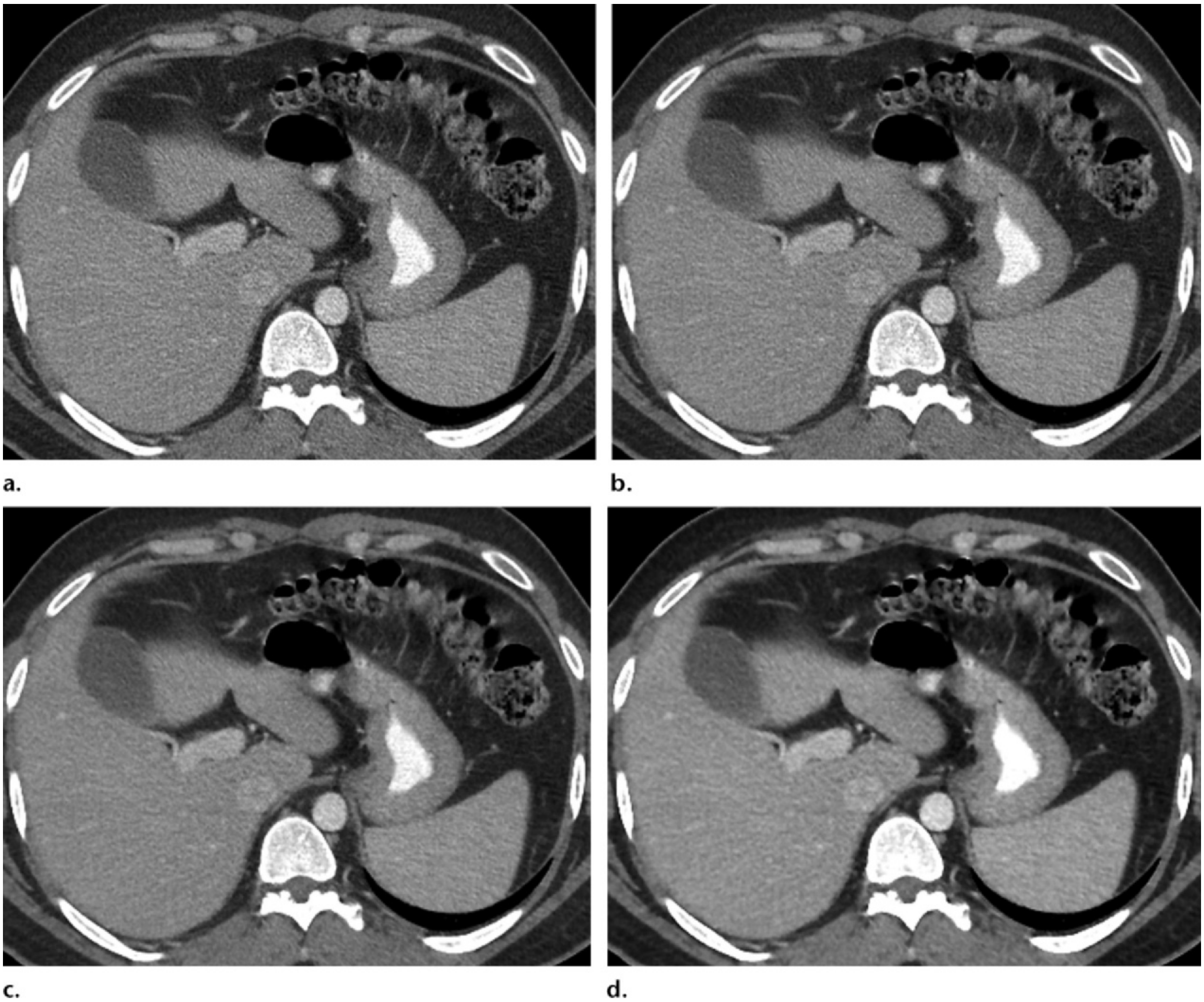

Figure 10. Axial CT images (0.625-mm thickness) reconstructed with filtered back projection (FBP) reconstruction **(a)**, 30% ASIR **(b)**, 70% ASIR **(c)**, and 100% ASIR **(d)**.

This allows either *(a)* improvement in image quality without increasing the radiation dose, or *(b)* maintenance of image quality at a lower radiation dose compared with traditional FBP image reconstruction (27,28). Iterative image reconstruction broadly consists of the following steps (Fig 10).

1. After reconstruction of the initial CT image, the image data are forward projected to generate a simulated projection.
2. The simulated projection is compared with the original measured projection to identify differences between the two, which contribute to image noise.
3. The calculated difference is applied to the simulated projection to correct for inconsistencies, and a new CT image is reconstructed.
4. Steps 1–3 are repeated multiple times, with the difference between the simulated and measured projections decreasing with each subsequent iteration.
5. The iterative process is usually discontinued when a predefined image quality criterion is met or the difference between the simulated and measured projections is minimal.

The iterative process can be performed in the image domain, the raw data domain, or both. With use of a statistical model of noise, variations in projection data contributing to image noise are identified and removed to generate images with reduced noise. All of the major CT vendors have clinically available iterative reconstruction techniques, which work using different algorithms (Table).

Statistical Iterative Reconstruction

All statistical iterative reconstruction algorithms achieve significant noise reduction and have been reported to allow dose reductions in the range of 30%–50% while maintaining image quality (21,29–36). With ASIR (GE Healthcare), SAFIRE (Siemens Healthcare), and iDose (Philips Healthcare) (4), the degree of noise reduction on the final image can be customized with user-selected levels. For example, a 100% ASIR image

Types of Iterative Image Reconstruction Techniques

Technique	Vendor	Data Used for Identifying Noise and Iteration Process	Adjustable Levels of Iteration
Statistical iterative reconstruction			
Adaptive statistical iterative reconstruction (ASIR)	GE Healthcare	Imaging and projection data	0%–100% (increments of 10%)
Image reconstruction in image space	Siemens Healthcare	Imaging data	None
Sinogram-affirmed iterative reconstruction (SAFIRE)	Siemens Healthcare	Imaging and projection data	Strengths 1–5
Adaptive iterative dose reconstruction 3D	Toshiba Medical Systems	Imaging and projection data	Levels 1–3
iDose	Philips Healthcare	Imaging and projection data	Levels 1–7
Full iterative reconstruction			
Model-based iterative reconstruction (MBIR)	GE Healthcare	Imaging and projection data, model of CT system optics	None
Iterative model reconstruction	Philips Healthcare	Imaging and projection data, model of CT system optics	Levels 1–3

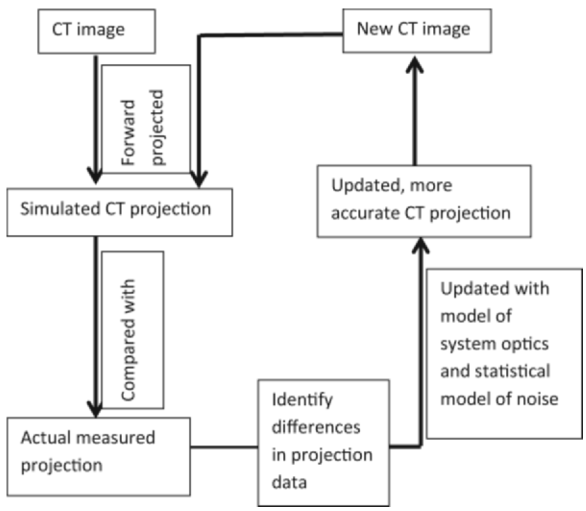

Figure 11. Diagram of the steps of iterative reconstruction. With each successive iteration, the difference between the simulated and measured projection is reduced. The iterative process is usually discontinued when the predefined image quality criterion is met.

represents the maximum noise reduction achievable with the statistical model used for ASIR. The degree of ASIR applied ranges between 0% and 100% (in increments of 10%) and can be selected, with varying blending of the FBP image and the 100% ASIR image (Fig 11).

In addition to demonstrating reduced noise, images reconstructed with iterative reconstruction techniques can have an altered texture with a smoothed appearance. The degree of smoothing increases with the degree of noise suppression used, which can lead to reduced sharpness of organ margins, a smoothed appearance of solid organs, and reduced perception of small objects. For routine abdominal CT studies in which the dose is reduced by 30%–40% (relative to earlier FBP protocols), a moderate level of iteration such as 30%–50% ASIR, SAFIRE strength 3 (S3), or iDose level 3–4 is considered optimal for reducing noise without affecting image quality (Fig 12) (29,30).

Teaching Point

Image quality depends not only on the level of iterative reconstruction used, but also on the inherent image noise prior to the application of iterative reconstruction. Use of a higher level of iteration on scans with low image noise accentuates image smoothness but does not do so on scans with high image noise (37). Hence, when radiation dose is further reduced for extremely low-dose protocols (such as renal stone CT), a higher level of iteration is needed to offset the associated increase in image noise. For example, a higher ASIR of 70%–80% can be used to enhance noise reduction with extremely low-dose protocols (38). The optimal level of iterative reconstruction depends on the degree of radiation dose reduction used for scanning and the desired spatial resolution and differs among scanning protocols.

Full Iterative Reconstruction

Image reconstruction using FBP assumes that x-rays originate from a point source, interact at a point within the image voxel, and are detected at the central point of a detector cell. This allows representation of the attenuation coefficient along the path of a CT projection as a pencil beam, which simplifies analysis of CT findings and allows image reconstruction in real

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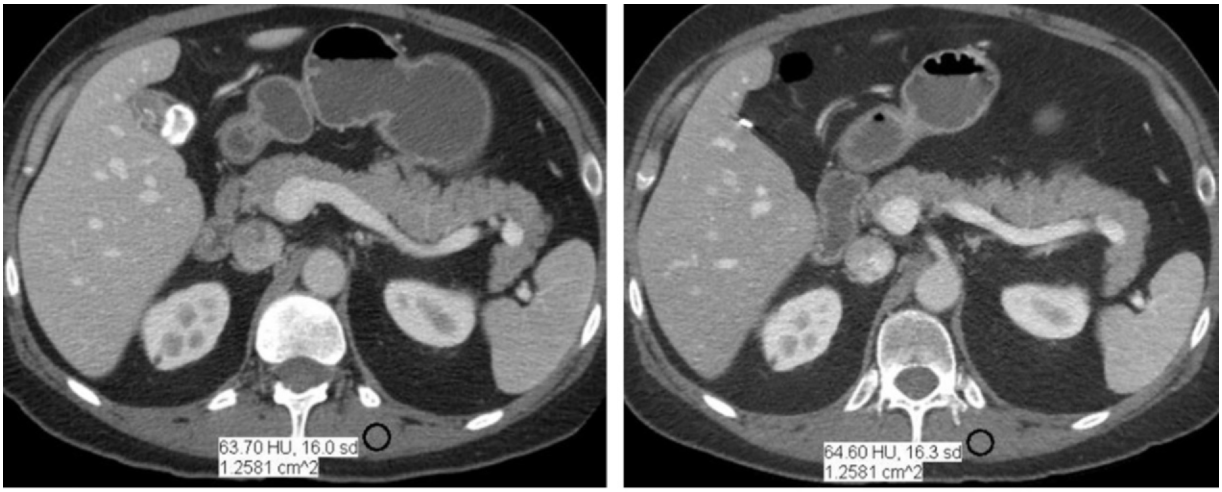

a. b.

Figure 12. CT images obtained at a fixed tube current with FBP reconstruction **(a)** and with ATCM at a noise index of 30 at 0.625-mm thickness with 30% ASIR reconstruction **(b)** show similar image noise and quality. $CTDI_{vol}$ values were 21.3 and 12.3 mGy, respectively.

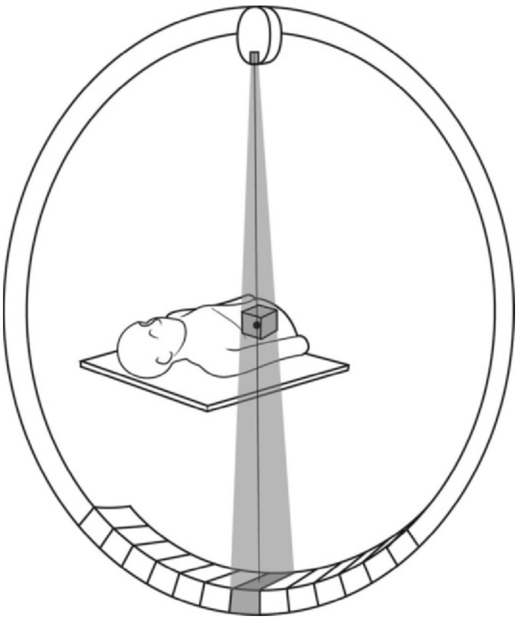

Figure 13. Drawing illustrates CT projection acquisition. The entire projection is represented as multiple pencil beams for FBP image reconstruction, one of which is shown as a single line. MBIR allows more comprehensive 3D modeling of x-ray generation from the focal spot, interaction with the patient, and capture at the detectors (shaded area).

time as the data are acquired (Fig 13). However, these assumptions reduce spatial resolution and contribute to image noise. The significantly increased image noise and beam-hardening artifacts limit the use of FBP image reconstruction with low-dose scans. MBIR (GE Healthcare) incorporates a detailed model of the CT scanner geometry as a two-dimensional configuration of the focal spot, 3D image voxels to represent patient attenuation, and two-dimensional interactions with detectors instead of assuming a point source, attenuation, and detection. This information is incorporated into the iterative reconstruction process, in addition to the statistical modeling of image noise used for ASIR (39). This allows improved noise suppression compared with ASIR, and, because corrections are applied to projection data prior to FBP image reconstruction, spatial resolution is improved and image artifacts (such as beam hardening) are reduced (Fig 14) (39). Use of MBIR can potentially reduce radiation dose at abdominal CT by 60%–70% while maintaining the CNR and spatial resolution (Figs 15, 16) (40,41).

Potential limitations of MBIR include prolonged image reconstruction time and altered image texture. MBIR is computationally intensive, which significantly increases image reconstruction time compared with that for FBP or iterative image reconstruction in the image domain or projection data domain (up to 1 hour for abdominopelvic CT), possibly affecting workflow (40). Images reconstructed with MBIR have a noticeably altered appearance compared with FBP images. In addition to reducing image noise, the spatial frequencies of the noise (noise power spectrum), which determines the quality of image noise, is also altered, with a shift toward lower spatial frequencies when MBIR is used, resulting in a smoothed image appearance and a decrease in perceived image quality (42). Familiarity with the appearance of images reconstructed with MBIR with regular-dose protocols is recommended before using this technique with low-dose protocols.

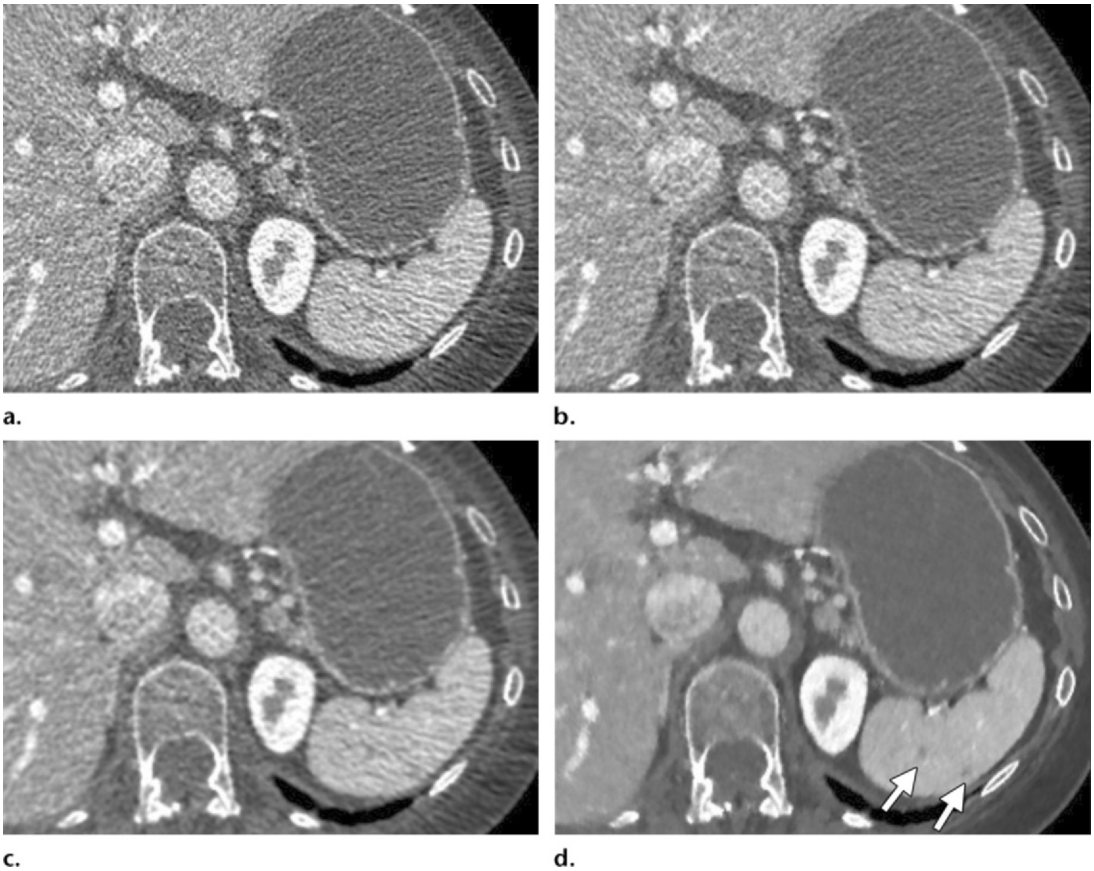

Figure 14. Axial 0.625-mm-thick images from a low-dose CT study ($CTDI_{vol}$ = 2.5 mGy, effective dose = 1.8 mSv) reconstructed with FBP **(a)**, 50% ASIR **(b)**, 100% ASIR **(c)**, and MBIR **(d)** show a progressive decrease in image noise. Two small hypoattenuating splenic lesions are seen on the MBIR image (arrows in **d**) that are not easily seen with FBP reconstruction or ASIR.

CT Dose Reduction and Maintenance of Diagnostic Accuracy

The application of low-dose CT protocols in clinical practice needs to be tailored to the specific diagnostic task required so that diagnostic accuracy is maintained. The diagnostic accuracy when evaluating objects with high contrast (such as renal calculi at CT angiography) is only minimally affected with use of low-dose protocols, whereas it is more adversely affected when evaluating objects with low contrast (such as hepatic metastasis). This holds true even when iterative image reconstruction is applied, such that even though the measured image noise would remain constant with low-dose protocols, lesion detectability would be reduced with use of aggressive dose reduction techniques (32). The optimal dose reduction achieved with iterative image reconstruction depends on the imaging task and should not be based solely on the goal of reducing image noise.

Conclusion

Several recent advances in CT scanning techniques have allowed reduction of radiation dose at CT. Radiologists need to understand the latest dose optimization strategies and should incorporate them into clinical practice by collaborating with physicists and CT technologists. Use of ATCM should be standard practice for CT scanning, with proper modification of the image quality parameter based on the required task. Modifying CT protocols to optimize the tube voltage based on the diagnostic task and patient habitus allows further reduction in dose and improves the visualization of iodinated structures at lower kilovoltage. Iterative image reconstruction, which allows images to be obtained at a reduced radiation dose without an increase in image noise, should also be incorporated into standard practice when available. In so doing, serious consideration should be given to the altered image quality compared with FBP to allow selection of the appropriate level of iteration for the protocol.

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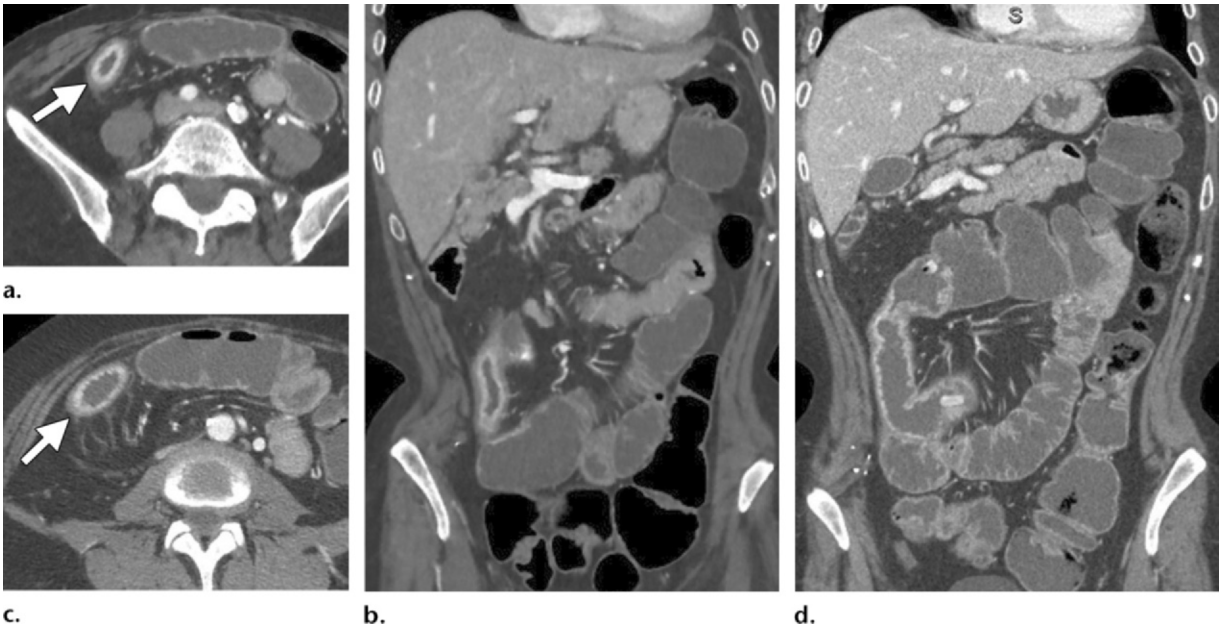

Figure 15. **(a, b)** Axial 2.5-mm-thick **(a)** and coronal 2-mm-thick **(b)** MBIR images from low-dose CT enterography performed in a patient with Crohn disease at 80 kV with a noise index of 60 at 0.625-mm thickness ($CTDI_{vol}$ value = 1.9 mGy) show wall thickening and enhancement involving the neoterminal ileum (arrow in **a**). **(c, d)** Corresponding 30% ASIR images from an earlier standard-dose CT enterographic study performed in the same patient at 120 kV with a noise index of 30 at 0.625-mm thickness ($CTDI_{vol}$ value = 7.9 mGy) show similar findings (arrow in **c**).

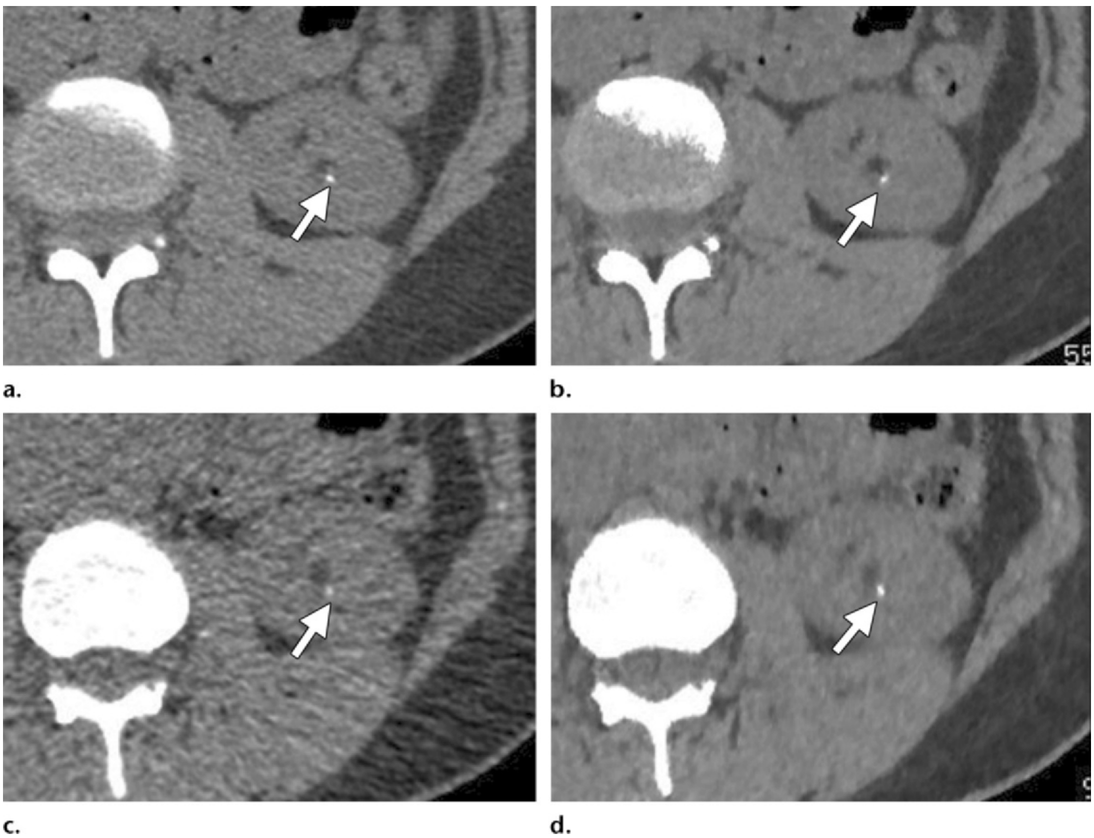

Figure 16. **(a, b)** Axial 1.25-mm-thick images from a CT study performed at 120 kV and 80 mA that were reconstructed with ASIR **(a)** and MBIR **(b)** show a 2-mm calculus in the left kidney (arrow). $CTDI_{vol}$ = 3.57 mGy. **(c, d)** On axial 1.25-mm-thick images from a repeat low-dose CT study performed at 120 kV and 20 mA, the calculus is not well seen on the image reconstructed with ASIR due to increased image noise (arrow in **c**), but it is easily seen on the image reconstructed with MBIR due to better noise reduction (arrow in **d**). $CTDI_{vol}$ = 0.9 mGy.

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Emerging Techniques for Dose Optimization in Abdominal CT

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Page 5
With use of ATCM, it is important to determine the desired image quality for each scanning protocol. This serves as the reference standard on the basis of which the milliamperage is modulated. The desired image quality can be defined in terms of either a predefined reference image quality or estimated image noise.

Page 6
Appropriate selection of the noise index and the reconstructed section thickness at which the noise index is applied is essential for achieving optimal image quality with reduced dose when dose-modulated tube current is used. The tube current modulation does not depend on the choice of prospective section thickness when a reference image quality is used.

Page 8
However, when iodinated contrast material is administered and evaluation of iodinated structures is the primary task (as at CT angiography), lowering the kilovoltage from 120 kV to 100 kV or even 80 kV is desirable because the effective energy of the x-ray beam will be closer to the k-edge of iodine, resulting in a higher attenuation for the iodine. This increases both image contrast and the CNR, despite the increase in image noise associated with lower-kilovolt scanning.

Page 10
With the development of iterative image reconstruction, it is now possible to selectively identify and reduce image noise while maintaining image contrast and resolution.

Page 12
For routine abdominal CT studies in which the dose is reduced by 30%–40% (relative to earlier FBP protocols), a moderate level of iteration such as 30%–50% ASIR, SAFIRE strength 3 (S3), or iDose level 3–4 is considered optimal for reducing noise without affecting image quality.