

# Image Quality and Current Techniques for Dose Optimization in Abdominal CT: What Every Radiologist Should Know

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**Abstract** The perceived image quality in computed tomographic (CT) scanning is determined by the image noise, image contrast, spatial resolution, and artifacts. The knowledge of physical basis of these determinants and scanning parameters affecting them helps in designing CT protocols that can achieve diagnostically acceptable image quality at acceptable radiation doses. The emerging CT scanning techniques such as automated tube current modulation, optimal tube voltage, and use of iterative image re-construction have allowed considerable reduction in radiation dose without compromising the image quality. The article provides a brief description of each of these scanning techniques. As CT has remained the workhorse of medical imaging and the risks of medical radiation exposure have gained far more attention than before, radiologists need to understand these scanning techniques and incorporate them into the clinical practice.

**Keywords** Computed tomographic (CT) scanning · Automated tube current modulation (ATCM) · Image noise · Image contrast · Spatial resolution

## Introduction

Computed tomographic (CT) scanning accounts for approximately 50 % of all medical radiation exposure [1]. Besides the obvious importance of judicious use of clinical

imaging, there are many technologic advances available and being implemented in present day CT examination to limit the radiation burden. With multi-detector CT (MDCT) being readily available and often a first choice imaging modality for a number of clinical scenarios, the knowledge of CT dose-reduction techniques should no longer be just limited to the CT technologists or the medical physicists. The practicing radiologists and the clinicians should have a basic understanding of these techniques and their underlying principles. This review aims at providing an overview of the factors determining the CT image quality and various dose-reduction approaches provided by modern day CT scanners. A brief description of the underlying physical basis and practical utility of CT dose-reduction methods has been presented here, without going into the mathematical details.

## Image Quality Parameters

The process of CT image formation can be divided into three steps—scanning, image reconstruction, and display. There are various parameters which influence each of these steps as described in Fig. 1. The images thus obtained are subjected to assessment by the reporting radiologist who evaluates the image quality and extracts the diagnostic information out of these. The main parameters governing the perceived quality of CT images are (a) noise (b) image contrast (c) spatial resolution, and (d) the image artifacts.

## Image Noise

When a phantom containing only water is imaged, it is observed that the CT numbers fluctuate around zero and are never uniform. This randomness in the CT attenuation

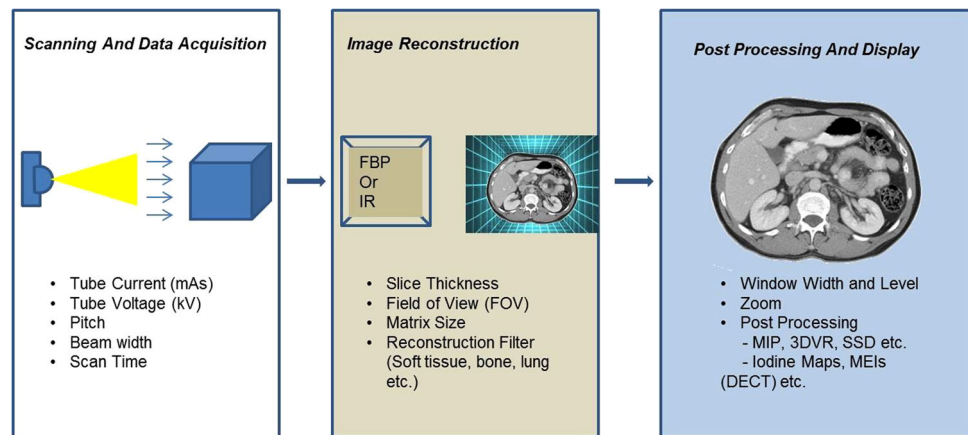
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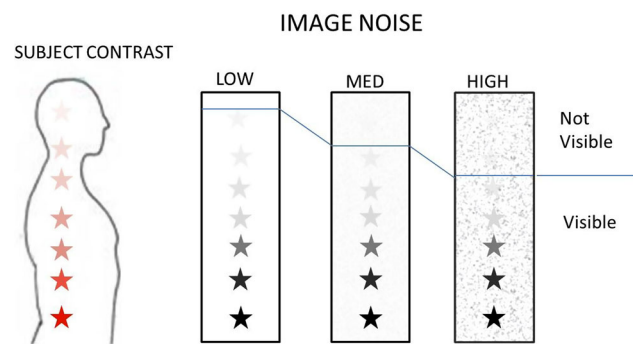
**Fig. 1** Steps involved in CT image formation. Various factors affecting each of these steps are enumerated in lower half of the boxes. *FBP* filtered back-projection, *IR* iterative reconstruction



values of otherwise homogenous material appears as graininess on CT images (Fig. 2). Analogous to the radiography, it depends on the number of X-ray photons contributing to each detector measurement. The insufficiency of photons arriving at the detectors after penetrating the body limits the accuracy to which each picture point can be calculated within the matrix and presents as picture grain or noise [2]. A statistical ROI function available on most CT scanners measures the average and standard deviation (SD) of the CT numbers belonging to the enclosed pixels. The SD relates to the image noise [2, 3]. The X-ray tube amperage and scan time (mAs), slice thickness, and the tube voltage affect the image noise by altering the number of detected X-ray photons. The image noise is also affected by the reconstruction filter. Smooth filters reduce noise and the sharp filters enhance it [3].

### Image Contrast

The image contrast in CT is decided by the subject contrast and display contrast. The display contrast depends only on the window level and width selected by the user and hence is arbitrary. Henceforth for the purpose of discussion the CT image contrast refers only to the subject contrast. The ability of CT scanning to detect structures that offer only slight differences in signal compared with the surroundings or so called “low contrast resolution” distinguishes the CT modality from other forms of non-tomographic radiography [4]. The subject contrast in CT is determined by differential X-ray attenuation of different types of tissue resulting in differences in the intensity reaching the detectors. The high peak kilo-voltage and relatively high beam filtration makes Compton scattering as the chief interaction of the X-ray photons responsible for generation of CT image contrast. The Compton scatter interaction depends on the differences in tissue electron density or



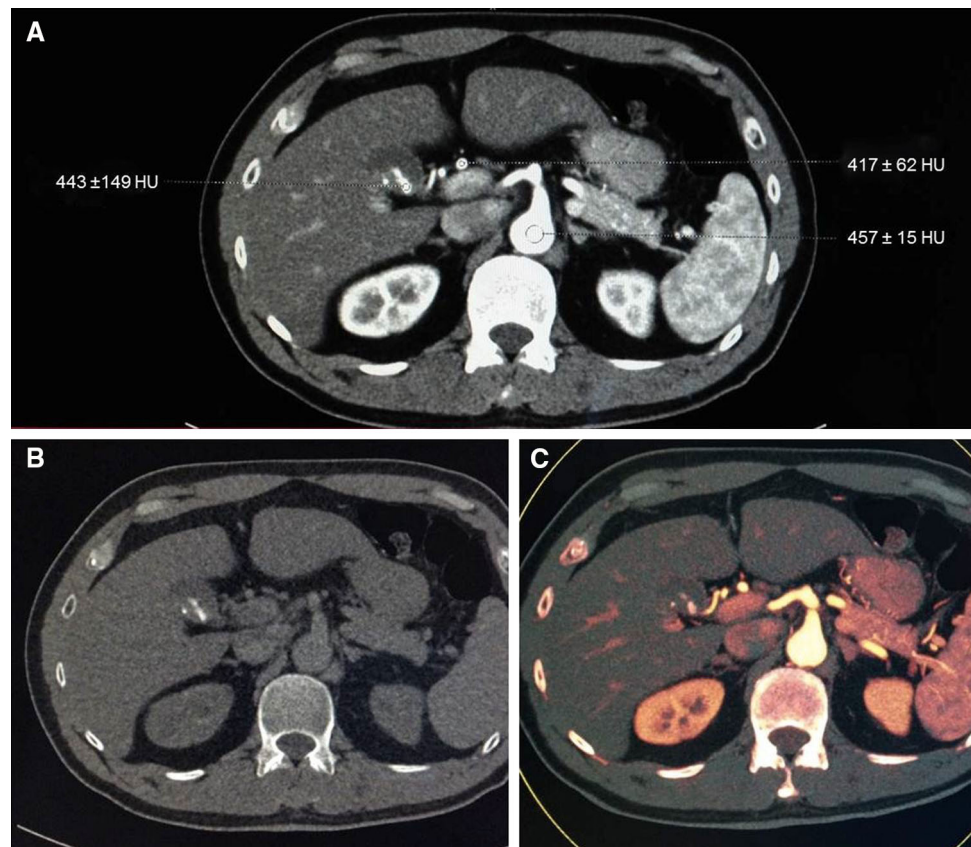
**Fig. 2** Drawing illustrates the effect of increased image noise resulting in increased graininess and inability to delineate structures which differ slightly in signal intensity from the surroundings (low contrast resolution)

simply the physical density of the tissues [5]. Thus, the differences in physical densities of different soft tissues primarily decide the image contrast in single energy CT scanning. The contrast scale or the CT numbers (Hounsfield units) are linear attenuation coefficients of the given voxel in relation to that of water. The contrast scale of a scanner should routinely be tested with various CT phantoms that contain materials (water, fat, soft tissue, and bone) designed to provide specific CT numbers [3, 4].

### Image Contrast in Dual-Energy CT (DECT)

Besides Compton scatter, another mechanism of X-ray attenuation important to diagnostic imaging is the photoelectric effect. In this phenomenon the photon interacts with a tightly bound inner-shell (K-shell) electron and ejects it out. Unlike Compton Effect, the photon is completely absorbed and there are no scattered radiations [6]. The photoelectric absorption highly depends on the atomic number of the material and its occurrence radically

**Fig. 3** Material-specific differentiation with dual-energy CT (DECT). Axial 5-mm-thick reconstructed images from a CT study shows similar attenuation of the arteries and calcified gallstones (a) on 120 kV optimal contrast images. The virtual unenhanced (b) and iodine map overlay (c) images differentiates the calcified gallstones from iodine in the arteries



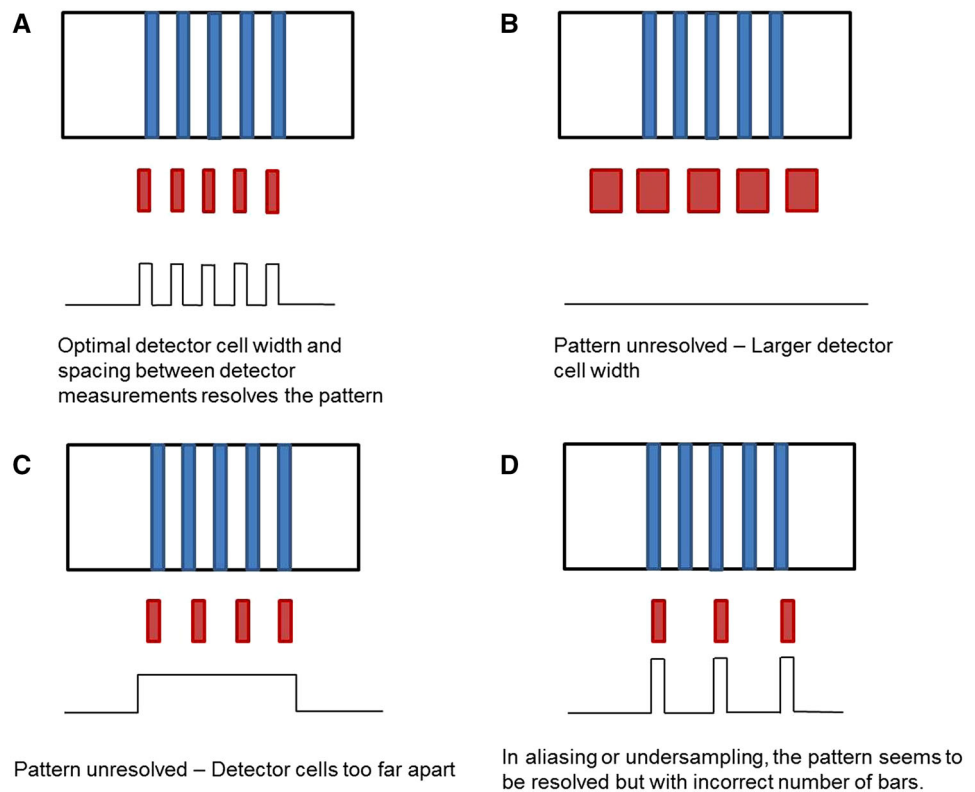
increases when the photon energy approaches the K-shell binding energy [6, 7•]. This so called K-edge effect is not important for the major elemental components of the human body i.e., carbon, oxygen, hydrogen, and nitrogen; owing to every low K-shell binding energy (ranging from 0.01 to 0.053 keV). However, the diagnostic range tube voltage has the potential to exploit the material-specific attenuation properties of elements with higher K-edges such as calcium (4 keV) and iodine (32 keV) [7•, 8].

The CT numbers are arbitrary units of linear X-ray attenuation in relation to the water. This shortcoming in distinguishing different materials is somewhat addressed by the DECT where the materials are more objectively discriminated based on CT number ratio or the dual-energy index (DEI) [8]. The DEI is defined as the ratio of the CT number of a given material on the low-energy image to the CT number of the same material on the high-energy image. The DECT techniques thus better discriminate materials of similar physical densities and are utilized to characterize uric acid crystals in patients with gout, iron deposition in hemosiderosis and different kinds of urinary stones, besides, differentiating calcium from the iodine [9•] (Fig. 3).

### Spatial Resolution

Spatial resolution is the ability to distinguish small, closely spaced objects and is determined by the acquisition geometry of the CT scanner, the reconstruction algorithm and the reconstructed slice thickness [4]. The fundamental concept underlying spatial resolution is sampling, which in CT imaging primarily depends upon the size of the detector cells and the spacing of detector measurements used to reconstruct the image (Fig. 4). The closely placed detector measurements and smaller detector cell width result in higher spatial resolution. The general rule, known as the Nyquist criterion, states that resolving  $N$  line-pairs per centimeter requires measuring at least  $2N$  samples per centimeter. Undersampling leads to misregistration by the computer of information relating to sharp edges and small objects, known as aliasing [10]. It manifests in the image as evenly spaced lines which can obscure finer details and limit the spatial resolution. It can be minimized by increasing the number of projections per rotation using slower rotation speed or using flying focal spot. Other factors that affect the spatial resolution are focal spot size, matrix size, and reconstruction filter [3].

**Fig. 4** Illustration of the effect of detector cell size and spacing on spatial resolution. Spatial resolution i.e., the ability to distinguish closely spaced objects, is often tested using phantoms with *bars* and spaces pattern (*line pairs*)



## Artifacts

The artifacts in CT imaging can be grouped into patient based and technical artifacts. The commonly seen patient-based artifacts are due to patients' movements and indwelling metallic implants. The technical artifacts are encountered secondary to the physical properties of the X-ray photon, limitations of the reconstruction algorithms or the defects in detector cells. Some of the commonly encountered technical artifacts are beam hardening, photon starvation, and partial volume averaging.

## Beam Hardening

As the beam passes through an object, the lower energy photons are absorbed faster than the higher energy photons and X-ray beam hardens comprising more and more of higher energy photons. The hardened beam therefore is more intensely perceived by the detectors than would be expected from the original X-ray beam [10]. This process results in the altered attenuation profile of the object which leads to two kinds of artifacts: the cupping artifacts and the appearance of dark bands or streaks [4, 10]. The ways to minimize beam hardening include—filtration, calibration correction, and correction software [10]. In dual-energy

imaging, creation of virtual monochromatic images from the image dataset can be utilized to reduce beam hardening (Fig. 5).

## Photon Starvation

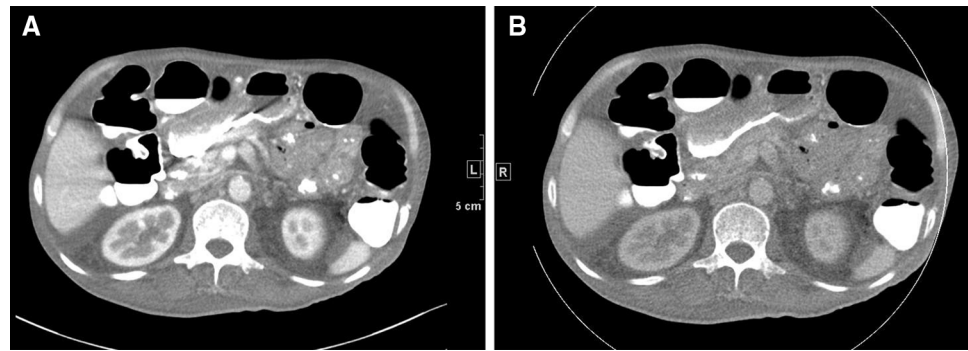
This happens when an insufficient number of photons reach the detectors after passing through very high attenuation structures in same projection (e.g., pelvis with bilateral hip prostheses). It results in markedly increased image noise and horizontal streaks in the images. Photon starvation can be minimized using automated tube current modulation (ATCM) (described in later sections) techniques during scanning [10].

## Partial Averaging

Partial volume artifact occurs when tissues of widely different attenuation coefficient are part of the same CT voxel resulting in beam attenuation proportional to the average value of these tissues [10]. Another form of partial averaging artifact results when a high attenuation object protrudes into the path of an incident beam in one projection (right to left) but not so in the opposite projection (left to right). This inconsistency in the measurement of



**Fig. 5 a** Axial 3-mm-thick reconstructed image from a CT study demonstrates beam-hardening streak artifacts from dense oral and rectal contrast. **b** Axial 3-mm-thick virtual monochromatic (at 140 kV) reconstructed image from same study displayed at same window level and width shows disappearance of streak artifacts and improved image quality



attenuation profile between the views results in different shades of gray appearing in the image [10]. It also results in a small high density object appearing as a larger lower density object, for example, when looking at cortical bone in thick CT slices [4]. Hence, these artifacts can best be avoided using a thin acquisition section width [4, 10].

### Dose Reduction Techniques

The recent advances in CT scanner hardware as well as introduction of newer scanning techniques have allowed radiation dose reduction without compromising the diagnostic quality of the examination [11•]. The review focuses on the later arm of dose-reduction strategy i.e., the advances in scanning techniques such as ATCM, use of optimal tube voltage, and improved utilization of iterative image reconstruction. The advances in CT scanner hardware (such as higher-power X-ray source allowing better X-ray beam filtration, optimal beam filtration, and improved detector capability) are not discussed here as these are not directly controlled by the operators and as such their knowledge is of little practical utility to the practicing radiologists.

#### Automated Tube Current Modulation (ATCM)

The tube current and tube potential determine photon flux and beam energy, which in turn decide the image quality and radiation dose. Reduction in tube current results in increased image noise whereas undue increase leads to increased radiation dose [12•]. Modulation of tube current is essential to strike a balance between the image noise and radiation exposure, the conflicting determinants of diagnostic acceptability. ATCM techniques are aimed at adjusting tube current in an effort to maintain constant image quality at the lowest dose [12•, 13]. These techniques are

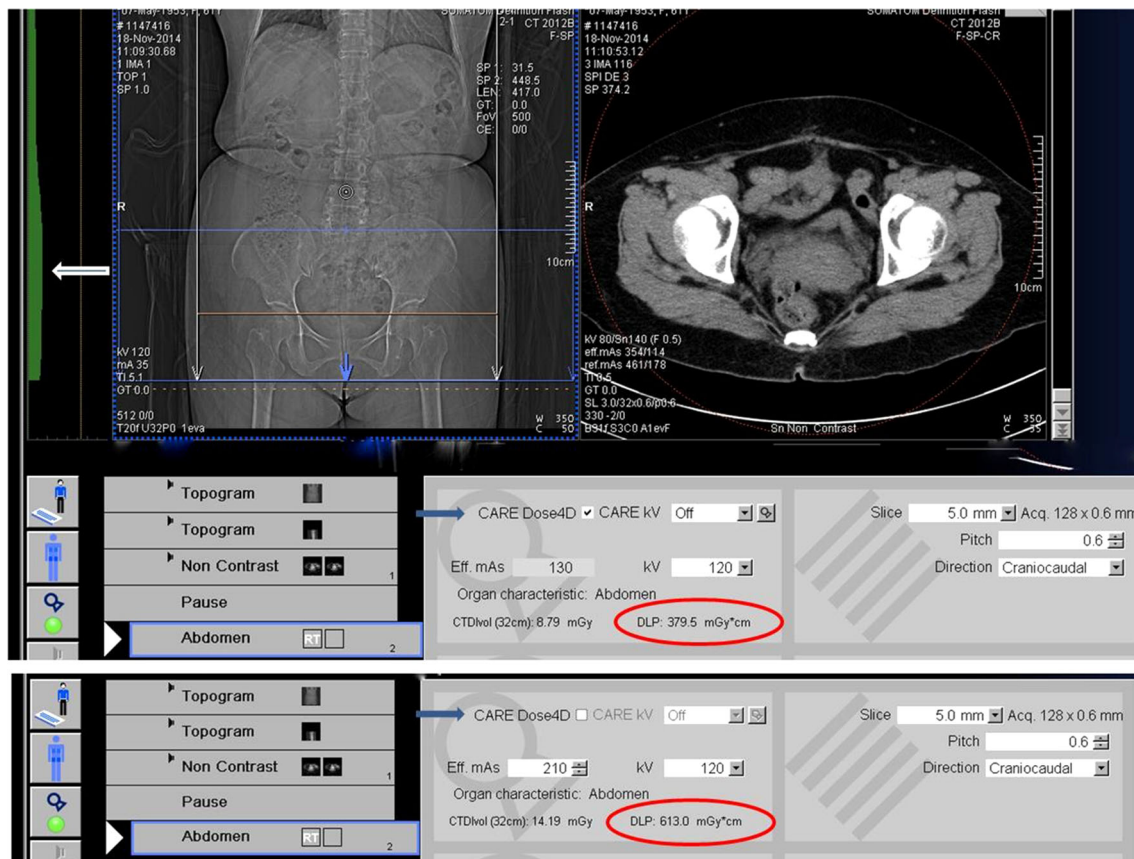
based on the fact that image noise depends on the X-ray quantum mottle in the transmitted beam projections. ATCM aims to alter the tube current on the basis of real-time regional body anatomy to maintain constant image noise with improved dose efficiency [13, 14]. The additional benefits provided by these techniques are reduction in tube loading (heating) and minimization of streak artifacts caused by photon starvation [12•]. Two distinct techniques for ATCM are angular ( $x$ - $y$ ) modulation and  $z$ -axis modulation.

#### Angular Modulation

The angular-modulation technique was introduced as early as 1994 when a dose reduction of 8.2 % was achieved in abdominal CT examination [15]. In this technique, the tube current or 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 (usually the anteroposterior direction). The attenuation measured for the modulation can be obtained in two ways—(i) online, real time from the immediate previous rotation of the X-ray tube around the patient (CARE dose 4D, Siemens Medical Solutions; DoseRight Dose Modulation, Philips Medical Systems; SureExposure, Toshiba Medical Systems) and the tube current varies as square root of the measured attenuation [16, 17•] (ii) from the anteroposterior and lateral scanograms and in this form the tube current varies sinusoidally between the limits defined by the asymmetry of regional anatomy (SmartMA, GE Healthcare) [17•].

#### Z-axis or Longitudinal Modulation

The  $z$ -axis modulation (AutomA, GE Medical Systems; Real E.C., Toshiba Medical, Tokyo, Japan) technique functions somewhat differently than angular modulation to



**Fig. 6** Screenshots from a CT scanner operating console show dose reduction of 38 % when a study was performed using automated tube current modulation (CareDose 4D). The *green vertical bar* (*white*

*arrow*) at the *upper corner* of image demonstrates variation in tube current (mAs) along the Z-axis of the patient (Color figure online)

maintain a pre-selected quantum noise level in the image data [13, 18]. It computes the tube current needed to obtain images with a selected noise level from the patient's localizer radiograph projection data. The projection data from a single localizer radiograph is used to determine the density, size, and shape information of the patient [17]. The tube current is then varied along the longitudinal axis of the patient such that lower attenuation portions of the body are imaged with lower milliamperage than the higher-attenuation portions [17]. Scanning the abdomen with Z-axis modulation can reduce the dose by 34.1–44.9 % with no increase in noise [19].

### Combined Tube Current Modulation

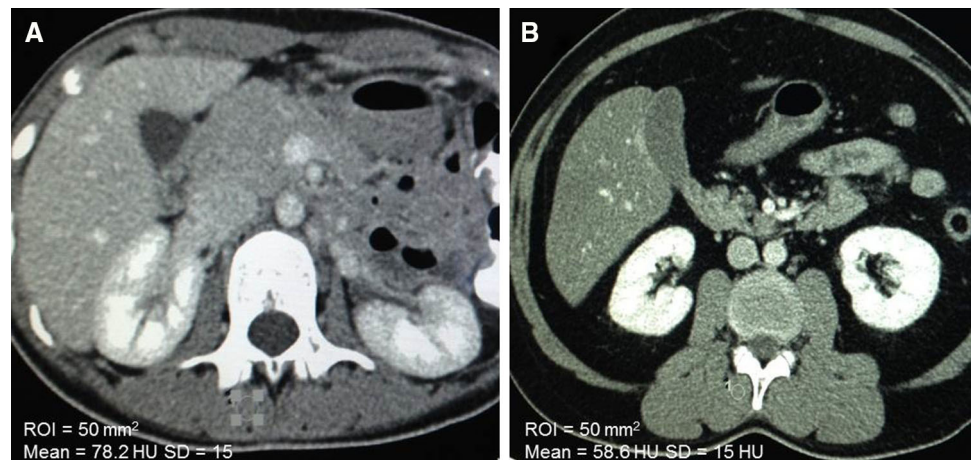
Combined tube current modulation system combines the previous two methods to vary the mA in a more comprehensive way, taking into account the patient's attenuation in all three dimensions as well as the desired level of image quality [18]. The commonly available combined modulation systems are Care Dose 4D (Siemens, Siemens Medical Solutions) and AutomA 3D (GE Healthcare) [20]. Care

Dose 4D makes use of the effective milliamperage [(mAs x rotation time)/pitch factor] to compensate for the helical pitch [21]. In this technique, besides varying according to the patient's size and shape, real-time angular dose modulation measures the actual attenuation in the patient during the scan and adjusts tube current accordingly for different body regions and different angles during rotation of the gantry [21, 22] (Fig. 6). The image quality is defined by the operator-selected image quality reference mAs and the adaptation strengths (weak, average, or strong). Using the projection data from the scanogram, the algorithm compares the actual patient sections to a "standard-sized" patient and together with the pre-selected adaptation strength (average by default) controls the modulation of tube current [21, 22]. In a phantom study, a dose reduction of up to 44 % was achieved using this technique without compromising the image quality [20].

### ATCM Modification

The perceived image quality is better in larger patients than the thin patients despite having same measure of image

**Fig. 7** Axial 3-mm-thick reconstructed images from two different patients with same level of image noise (SD = 15 HU) show that the perceived image quality is better in presence of higher amounts of intra-abdominal fat



noise (Fig. 7). This is owing to presence of higher amounts of intra-abdominal fat which provides inherent contrast around organs and compensates for the noise levels [higher contrast-to-noise ratio (CNR)] [11••, 23]. To achieve either a low-noise examination or a low-dose examination on a scanner with ATCM, there are always complex trade-offs among noise index (NI), slice thickness, and radiation dose [23]. ATCM techniques that are based on NI (GE healthcare) where in a constant image noise is maintained regardless of patient diameter suffer from reduced image quality in thin patients. Diagnostically acceptable image quality in thinner patients, compared to average-sized patients can be achieved with either using a lower NI for thin patients or limiting the minimum milliamperage so that the milliamperage does not become too low. In obese patients, higher noise is tolerated for acceptable image quality, so that a higher NI or limiting the maximum milliamperage can be used to reduce the radiation dose [22].

### Modification Based on Intent of Study

The CT examination studies which intend to differentiate soft tissue structures of similar attenuation (such as detection of liver metastasis) require better image quality and hence least possible background noise [24]. The reference image quality or NI thus should not be altered in such cases. Whereas, for the evaluation of large contrast differences (e.g., detection of stones in CT KUB), increased image noise does not usually affect the diagnostic accuracy. The large contrast difference between evaluated structures (such as the calcific stone and renal parenchyma), allows tolerance for higher background noise, thereby allowing a reduction in radiation dose [11••].

### Optimized Tube Voltage

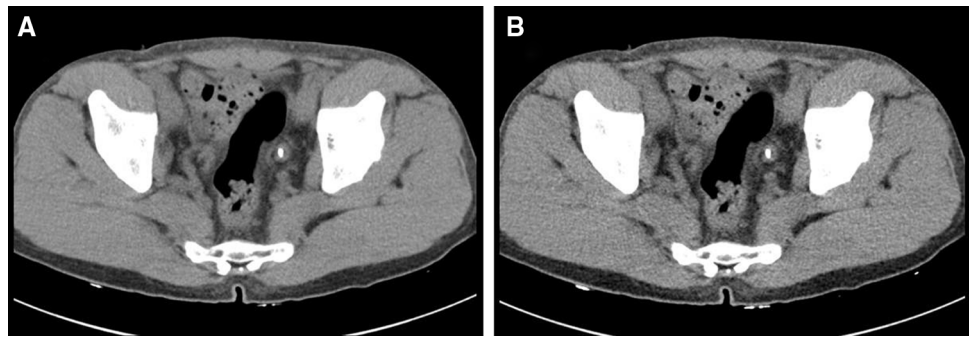
The CT imaging at lower kV (80 or 100 kV) results in greater iodine-related attenuation difference than similar images at 120 kV [25]. The potential benefits of lower kV scanning at contrast-enhanced CT include reduced radiation dose with similar CNR, preservation or improved conspicuity of lesion and reduction in contrast load [11••, 23]. Although being routinely integrated in pediatric abdominal CT scanning protocols, lower kV scanning is challenging in adult patients because of the increased noise and susceptibility to beam-hardening artifacts [11••].

### Low KV Scanning

When evaluation of iodinated structures (CT angiography) is the primary task of imaging, lowering the kilo-voltage from 120 to 80 kV is desirable because the effective energy of the X-ray beam will be closer to the k-edge of iodine (33 kV), resulting in a higher attenuation for the iodine [26]. This results in increase in image contrast and the CNR, which compensates well for increased image noise associated with lower kV scanning. This approach has additional advantage of ability to reduce the iodine burden and hence risk of contrast nephropathy [11••, 27]. Noda et al. demonstrated that the contrast burden can be reduced by 33 % with 80-kV setting while maintaining the image quality and detectability of malignant pancreatic tumors [28]. Low kV settings and decreasing the amount of contrast usage is particularly beneficial to the older patients, for whom the risk of radiation-induced cancer is minimal but the risk of contrast material-induced nephropathy is higher [11••]. During evaluation of images obtained at lower kV, the window width and level need to be increased



**Fig. 8** Axial 5-mm-thick images from a dual-energy CT examination reconstructed at 120 kV (a) and 80 kV (b) demonstrates that despite an increase in image noise at lower kV, the conspicuity of stone is better due to greater differences in the attenuation of stone and surrounding soft tissue (contrast-to-noise ratio) at 80 kV



while viewing to maintain imaging appearance similar to that of 120-kV images. This also reduces the perceived image noise, and improves the perceived image quality [11••]. Besides CT angiography, low kV scanning is helpful in imaging renal stones where large inherent contrast difference between the abnormality (calculus) and surrounding tissue compensates well for the increased image noise and same diagnostic information can be obtained at much lower radiation dose (Fig. 8).

#### Automatic kV Selection

The automatic kV selection tools (such as Care kV, Siemens Healthcare) makes use of the attenuation profile of the patient generated from the scanogram dataset, and the user-specified examination type to determine the optimal kV [29]. The tool calculates patient-specific mAs curves for all kV levels based on the given scan range, patient anatomy, and user-selected contrast behavior (tissue of interest for contrast-enhanced scans) necessary to obtain the pre specified image quality [29, 30] (Fig. 9). The estimated dose is then calculated based on these kV-specific mAs curves. If the lowest possible kV setting is not possible due to other imaging parameters (such as pitch settings, tube current limits, or scan range), the next best kV setting is suggested by the tool [30]. The average dose reduction achieved by adapting Care kV in abdominal CT examinations is around 20 % [30].

#### Image Reconstruction

The most widely used CT image reconstruction algorithm is filtered back-projection (FBP), which is fast, robust and allows adequate image reconstruction for most situations of day to day imaging in non-obese patients with routine-level radiation dose [27, 31•]. Image reconstruction using FBP assumes that X-rays originate from a point source, interact

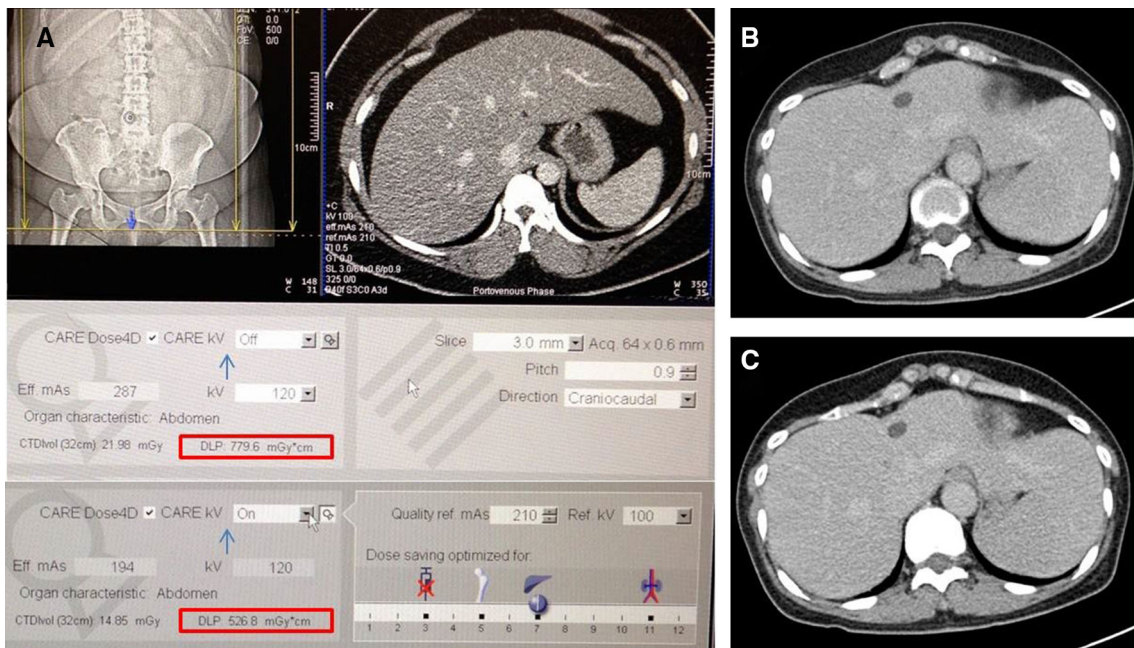
at a point within the image voxel, and are detected at the central point of a detector cell [11••]. The acquired projection data are assumed to be free of noise, first filtered to enhance or diminish certain image characteristics (e.g., edge enhancement or smoothing) and subsequently projected back to reconstruct the imaged volume [32]. In low-dose situations, these underlying assumptions render images obtained with FBP to high noise, streak artifacts, and poor low-contrast detectability [27, 31•].

#### Iterative Reconstruction

Iterative reconstruction (IR) is an alternative image reconstruction method that allows imaging at lower radiation doses with similar noise levels and image quality compared to routine-dose FBP [31•]. This reconstruction method incorporates a physical model that can accurately characterize the data acquisition process (noise, beam hardening, scatter, etc.). This ability of IR methods allows for improved image quality, particularly with low-dose CT scanning where insufficient data during image acquisition is more pronounced than routine CT [18]. In non-mathematical terms, the iterative image reconstruction consists of the following steps [11••, 33•]—(i) forward projection of image data to generate simulated or model projection (ii) comparison of measured projection to a model projection and identification of differences between the two (iii) correction of image noise based on the differences measured (iv) repetition of this IR loop multiple times with decrease in differences between model and measured projections with each iteration (v) termination of the IR loop when predefined image quality criteria are met (Fig. 10). The drawbacks of IRs include blotchy appearance of the images and longer computational time [31•].

The iterative process can be performed in the raw data domain alone, in the image domain alone or in both domains [31•, 33•]. There are two kinds of IR processes—statistical and full. The statistical methods incorporate

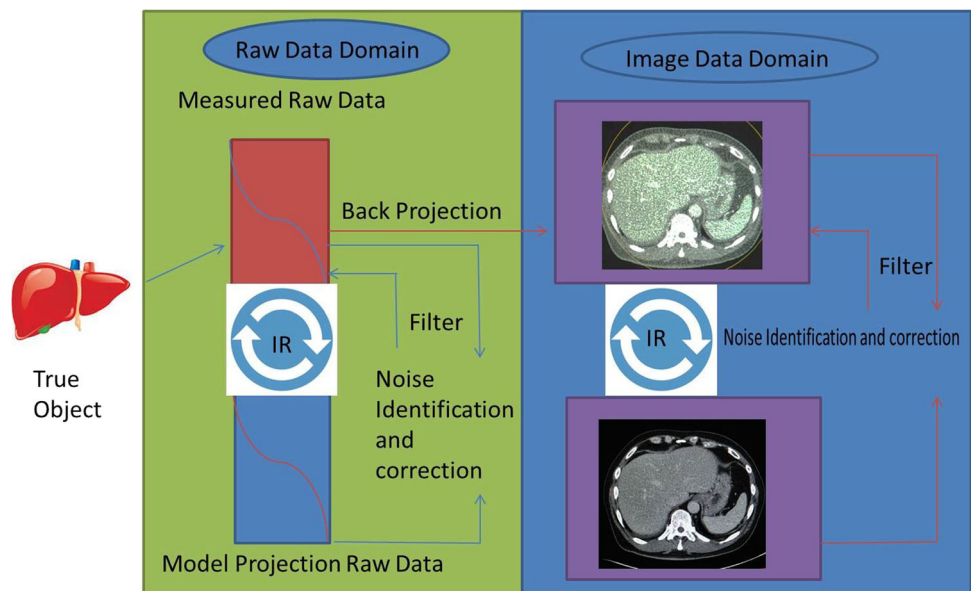




**Fig. 9** a The screenshots from a workstation during planning of a CT study demonstrates about 32 % reduction in radiation dose upon utilization of automatic kV selection (Care kV). Axial 5-mm-thick

images obtained from another CT study shows similar image quality with **c** and without **b** application of Care kV. A dose reduction of 14 % was achieved in this study with application of Care kV

**Fig. 10** Schematic representation of the iterative reconstruction (IR) process. It involves repeated alteration of the acquired or measured imaging information (raw data, image, or both) upon comparison with information generated based on modeling (full or statistical). The iteration loop stops when the desired image quality is achieved



counting statistics of the detected photons into the reconstruction process and assume a fixed statistical distribution of the photons [33]. Full or ‘model based’ iterative processes try to model the acquisition process as accurately as possible. Besides the detected photons (registered by the

detector pixel), the modeling also takes into account the photons scattered outside the detector area or absorbed by the object [33]. The full iterative processes incorporate configuration of the focal spot, attenuation profile of the patient and two-dimensional interactions with detectors

**Table 1** Various kinds of commercially available iterative IR techniques

Vendor	IR technique	IR process
Siemens	IRIS (iterative reconstruction in image space)	Statistical in image domain
	SAFIRE (sinogram-affirmed iterative reconstruction)	Statistical in both raw data and image domains
GE healthcare	ASIR (adaptive statistical iterative reconstruction)	Statistical in both raw data and image domains
	MBIR (model-based iterative reconstruction)	Full IR in both raw data and image domains
Philips	iDose	Statistical in both raw data and image domains
	Iterative model reconstruction	Full IR in both raw data and image domains
Toshiba	AIDR 3D (adaptive iterative dose reconstruction 3D)	Statistical in both raw data and image domains

instead of point source, into the modeling. It leads to better matching of the artificial image to the acquired raw data [11•, 33•].

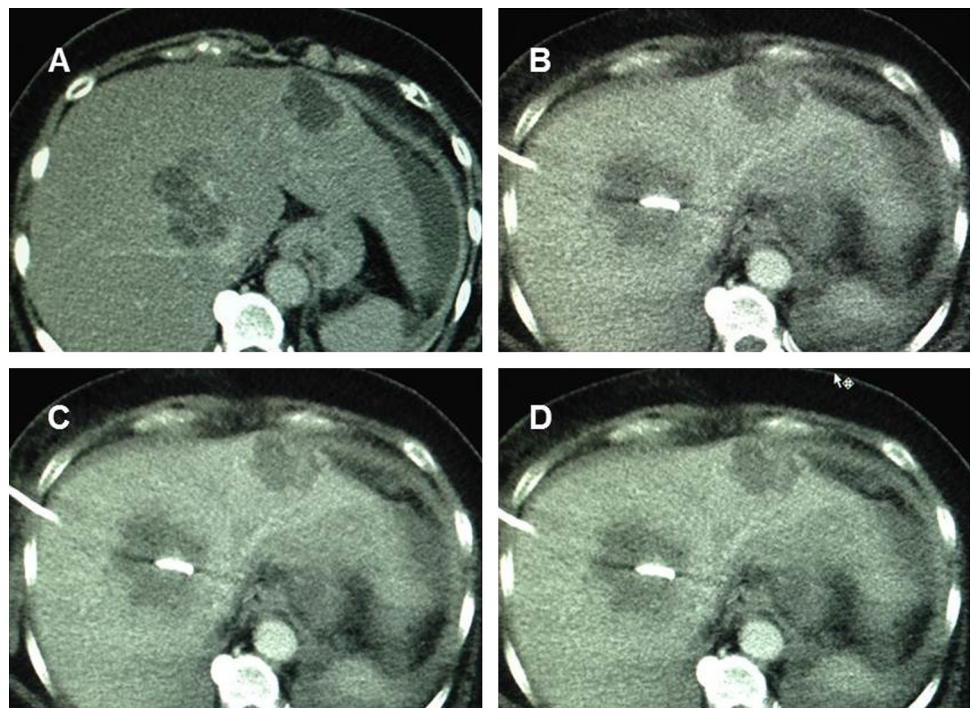
Various vendor based IR techniques are available as described in Table 1.

### SAFIRE (Sinogram-Affirmed Iterative Reconstruction)

We routinely use SAFIRE algorithm for low-dose scanning at our institution which is briefly discussed here. SAFIRE received FDA clearance in November 2011. This IR

technique operates in both the raw data and image domains with up to five strengths of noise modeling and regularization [34] (Fig. 11). The manufacturer stated optimal strength for CT examination is 3, which is further confirmed by Kim et al. in their recent study [35]. In addition, they concluded that the diagnostic performance of scanning at 30 mAs with any strength is comparable to that of scanning at 100 mAs for the diagnosis of acute appendicitis [35]. The level of noise reduction and noise texture changes depend on the user defined strength, with strength 1 being noisier and strength 5 being smoother. The strengths do not affect the number of iterations or the

**Fig. 11** SAFIRE (Sinogram-affirmed iterative reconstruction). Axial 5-mm-thick reconstructed images from a low-dose (0.57 mSv) CT obtained during image guided drainage of liver abscess, at different strengths of user defined iteration profile (**b**—level 1, **c**—level 2, **d**—level 3) demonstrate increased smoothness with increasing strength. The image quality is comparable to the earlier routine CT study which was performed at a much higher radiation dose (2.9 mSv)



reconstruction time. SAFIRE can be used with dual-energy imaging as well. It can reduce the radiation dose as well as image noise by up to 60 % [34].

## Conclusion

Knowledge of factors determining the CT image quality helps in tailoring the CT examination to the diagnostic task in hand and allows implementation of dose reduction strategies. The recent advances in CT scanning techniques have allowed considerable reduction in CT radiation doses. A practical knowledge of these techniques is essential in designing and implementation of abdominal CT scanning protocols. ATCM should be routinely used with proper modification of the image quality parameters based on the required imaging task. Optimization of kV based on the diagnostic task and patient habitus allows further reduction in dose and improves the visualization of iodinated structures (at low kV). Iterative image reconstruction allows images to be obtained at a reduced radiation dose without an increase in image noise and should also be incorporated into standard practice. Its use however should be tailored to the imaging task and not solely targeted at reducing image noise.

## Compliance with Ethics Guidelines

**Conflict of Interest** Dr. Rahul Lohan declares no potential conflicts of interest.

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

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- Of importance
- Of major importance

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