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MDCT Radiation Dose

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Introduction

Emergence of multidetector computed tomography (MDCT) scanners in radiology practice has increased the number of CT studies being performed for different clinical applications. This has raised concerns about risk of radiation-induced cancer with low-dose exposure associated with CT scanning. This chapter describes radiation dose quantities, scanning parameters that can be adjusted to optimize CT dose, and strategies for CT dose reduction.

Radiation Dose Quantities

Absorbed dose is the energy deposited in tissue/organs per unit mass measured in Gray (Gy). It is the basic quantity used for assessing the relative radiation risk to the tissue or organ. Effective dose represents a calculated quantity that accounts for the difference in radiosensitivity of different tissues. It compares relative radiation risk from different radiological procedures and is expressed in Sievert (Sv) [1, 2]. However, the principal dosimetric quantities that are displayed on the CT user interface include CT dose index volume (CTDI vol) and dose length product (DLP). These quantities can be applied to sequential or helical scanning for both single-slice or multislice scanners. The CTDI integrates the radiation dose delivered both within and beyond the scan volume. CTDI represents average absorbed dose across the field of view for contiguous CT acquisitions and takes into account regional variations in the absorbed dose. When CT scanning is performed with either gap or overlap between sequential scans (based on pitch values), the CTDI is scaled accordingly and results in the dose descriptor CTDI vol (mGy). While CTDI does not provide the dose given to any specific patient, it is a standardized index of the average dose delivered from the scan series. DLP represents the integrated dose and is equal to the average dose within the scan volume (mGy/cm).

In fact, most scanners provide CTDI vol and DLP values prior to actual patient scanning. These dose quantities can be used to compare radiation dose for different CT examinations, equipments, or imaging centers.

Strategies for Dose Reduction

There are several scanning factors that affect radiation dose associated with MDCT scanning [3]. These factors include those that can be modified by users and those that cannot be adjusted. CT scanning factors that can be adjusted to optimize radiation dose include tube potential, tube current, gantry rotation time, automatic exposure control, detector configuration, pitch, table speed, slice collimation, scan length, scan modes, scan region of interest, scanning phases, postprocessing image-based filters, metal artifact reduction (MAR) software, and shielding devices. In addition, there are several scan features that users cannot change, including scanner geometry, X-ray beam filters, prepatient tracking of X-ray tube focal spot, and projection adaptive reconstruction filters. We will focus on the scanning features that users can adjust to optimize dose.

Although there is an inverse relationship between image noise, an important component of image quality, and radiation dose, several studies have shown that diagnostic information can be achieved with substantial dose reduction [3]. Hence, all efforts to reduce dose must be preceded by evaluation of dose reduction effect on diagnostic requirements for specific indication or region of interest and for specific patient size or age.

Scanner Geometry

There is a considerable difference between geometry of single-slice CT and MDCT scanners that affects the distance between the focal spot of the X-ray tube and isocenter of the scanner. Also, it is not uncommon for large medical centers to have two or more scanner types. If all scanning parameters are kept constant, a scanner with short geometry will produce more interaction of radiation with the patient and lower image noise than a long geometry scanner. Thus, when scanning protocols are prepared for MDCT scanners, it is important to be cautious about the "transfer" of scanning parameters from one scanner type to another. Careful transfer of protocols helps in maintaining image quality with identical or reduced radiation dose depending on scanner geometry and other features (such as reconstruction algorithms) [4].

Tube Current (mA) and Tube Current-Time Product (mAs)

Tube current [milliamperage (mA)] reduction is the most frequently used method of reducing dose. Tube current-time product [milliampere second (mAs)] settings are proportional to the number of photons in the defined exposure time. There is a linear relationship between tube current and radiation dose. Thus, a 50% mAs reduction results in radiation dose reduction by half. However, mAs reduction should be performed carefully, as it leads to increase in image noise that can adversely affect diagnostic image quality.

Reduction in gantry rotation time (scan time) has been a main focus of MDCT developments toward improved temporal resolution. Use of fast scan time on MDCT implies shorter exposure time and lower radiation dose if all other parameters are kept constant. To allow a shorter exposure time, the X-ray tubes are designed to give better radiation output and improved heat capacity and heat dissipation. With development of MDCT scanners, tube cooling issues have been essentially eliminated, thus allowing a substantially higher mA with fast scan time.

Despite improved temporal resolution with MDCT, radiation dose can be higher than single-slice CT due to an increase in overall mAs. Increase in mAs with MDCT may be explained on the basis of increasing applications for thinner slices, which require higher mAs to maintain similar noise. Also, improved temporal resolution of MDCT allows multiphase acquisition protocols with high spatial resolution, which are associated with higher dose. Furthermore, in contradiction to single slice CT, considerably higher mAs values (about 800 mA at half-second rotation) can be used with MDCT scanners for single- or multiphase CT studies.

Unfortunately, there is no data to limit increasing mAs with decreasing reconstructed slice thickness. Multi-institutional studies are needed to determine the optimum section thickness for various clinical applications and to define maximum possible tolerable image noise for these applications. It is important to remember that MDCT allows volumetric data acquisition, which can be used to retrospectively reconstruct thinner sections, albeit with higher noise. To avoid higher image noise, some institutions acquire thin sections in a prospective manner with smaller pitch and slower table speed. In such circumstances, a relatively higher noise may be acceptable, as thinner sections have higher spatial resolution and less partial volume artifacts. However, in regions with low inherent contrast, such as the abdomen, a small increase in noise can affect conspicuity of small, low-contrast liver lesions.

Several studies have recommended mAs reduction for MDCT scanning [5, 6]. In certain applications and patients, mAs can be reduced due to small patient size (children and small adults), high inherent contrast where diagnostic quality of MDCT is not affected by higher noise (CT colonography, kidney stone CT, routine chest CT, pelvic CT, maxillofacial CT, bony skeleton CT), and applications where lower resolution is acceptable (CT perfusion). Adaptation schemes for adjusting mAs and kilovoltage peak (kVp) to age, weight, or size of children has been evaluated for dose reduction with MDCT scanners [7]. As dose requirements in chest CT are much smaller than for the abdomen because of low X-ray absorption in the lungs, chest CT scanning can be performed at lower tube current than abdominal examinations. For chest MDCT [8], acceptable image quality with 50% reduction in tube current has been reported. Use of reduced tube current (20–80% dose reduction) has also been reported to be as effective as standard-dose scans performed at higher tube current for evaluation of acute lung injury, follow-up of malignant lymphoma and extrapulmonary primary tumors, lung cancer screening, coronary artery calcium screening, pulmonary nodule, benign asbestos-related pleural-based plaques, benign diseases in young patients, and guided lung biopsy [3]. Likewise, tube current reduction (20–80% dose reduction) has also been reported for routine head, paranasal sinuses, neck, abdomen, and pelvis MDCT examinations [3].

Automatic Exposure Control (AEC)

Automatic exposure control (AEC) techniques enable automatic adjustment of tube current to the size and attenuation of the body part being scanned to obtain constant CT examination image quality with lower radiation dose. Available automatic exposure control (AEC) techniques include angular modulation (in the X/Y plane), Z-axis modulation, and combined modulation techniques (both X/Y and z-axis). Several studies have shown 20–70% dose reduction with the use of these AEC techniques, depending on body region and patient size. Dose reduction with AEC has been reported in several MDCT applications, including neck, chest, abdomen, pelvis, and extremities. Furthermore, recent studies have reported substantially higher dose reductions with combined dose modulation techniques compared with angular dose modulation for MDCT of the neck, chest, abdomen, and pelvis [9–12].

Compared with the fixed tube current, AEC techniques can help in homogenizing MDCT scanning protocols for different scanners and applications. This is possible by selection of required or desired image quality (for example, noise index with Auto mA, GE Healthcare, and reference effective mAs with CARE Dose 4D, Siemens Medical Solutions) with AEC techniques instead of fixed-current values. Some vendors allow the user to control the extent of dose reduction to avoid an excessive drop in mA. Users can select desired image quality for AEC based on study indication irrespective of patient size. In fact, initial AEC studies have reported that selection of fixed tube current for scanning was associated with higher dose for small patients and lower dose for large patients [9, 13]. For clinical applications (such as CT angiography, kidney stone CT) that can tolerate higher image noise, the user can select a lower image quality requirement with AEC to achieve further dose reduction.

Despite several encouraging reports, there are several limitations to AEC techniques. Notably, selection of desired image quality for AEC must be performed carefully, as selection of higher quality can lead to higher radiation exposure. Further studies are necessary to define reference image quality requirements for different clinical applications of MDCT so that AEC techniques can then be applied in a more scientific manner.

Tube Potential (kVp)

Tube potential determines the energy of incident X-ray beams. Variation in the tube potential causes a substantial change in CT dose as well as image noise and contrast. Reduction in kVp leads to dose reduction and an increase in image noise as well as image contrast. Most MDCT examinations are performed at either 120 or 140 kVp, with little change to lower values. Some recent reports suggest substantial dose reduction with use of low kVp (80–100) for CT angiography studies in the head, chest, and abdomen [14–16]. In the abdomen, compared with 120 kVp, use of a 100-kVp protocol resulted in about a 37% dose reduction for MDCT angiography of the abdominal aorta and iliac arteries [16]. Likewise, substantial dose reduction (30–56%) has also been reported in cerebral and pulmonary MDCT angiography studies with use of lower tube potential (80 kVp). Use of lower kVp (80–100) for dose reduction has also been recommended for chest and abdominal MDCT in newborns and infants [7]. As reduction in kVp can result in substantial increase in image noise, image quality maybe impaired if the patient is too large or the tube current is not appropriately increased to compensate for the lower tube voltage. Thus, for very large patients, higher tube voltage is generally more appropriate.

Scanning Modes

Overranging of the X-ray beam with helical scanning on MDCT scanners leads to some amount of unused radiation extending beyond the beginning and ending of the region of interest. Due to this phenomenon, to achieve greater dose efficiency, efforts must be directed toward use of a single helical acquisition rather than multiple helical scans in the absence of overriding clinical considerations such as breathing movements. Use of multiple contiguous helical acquisitions should be restricted with modern high-speed MDCT scanners. However, this may be unavoidable in multiregion MDCT studies such as simultaneous neck and chest CT (position of the arm) or simultaneous chest and abdomen CT (differential delay time for contrast enhancement).

Scan Length

With rapid improvement in temporal resolution of MDCT, there is a tendency to increase the scan length (including regions beyond the actual area of interest in the neck, chest, abdomen, or pelvis). This increases the effective radiation dose to the patient [3]. Proliferation of whole-body screening CT studies must be restricted. It is also essential to draw the attention of requesting physicians to dose consequences of increasing scan length and to establish guidelines to restrict the examination to what is absolutely essential. In this regard, technologists or monitoring radiologists should restrict acquisition of any "extra images" beyond the actual area of interest [17].

Scan Collimation, Table Speed, and Pitch

These factors are interlinked to each other as well as to the detector configuration used for MDCT scanning. For helical CT scanners, pitch was defined as the ratio of table feed per gantry rotation to the nominal width of the X-ray beam. With MDCT, two terminologies were introduced for pitch: slice or volume pitch (ratio of table feed per gantry rotation to the nominal slice width) and beam pitch (ratio of table feed per gantry rotation to the beam width or effective detector thickness). The latter is preferred over the former terminology. A beam pitch of 1.0:1 implies acquisition without overlap or gap, a beam pitch of less than 1.0:1 implies an overlapping acquisition, and a beam pitch of greater than 1.0:1 facilitates an interspersed acquisition. An increase in the pitch decreases the duration of exposure received by the anatomical part being scanned. Faster table speeds for a given collimation result in higher pitch, shorter exposure time, and lower dose. A narrow collimation with slow table travel speed results in lower pitch, longer exposure time, and higher dose.

This relationship between exposure and pitch does not apply to scanners that utilize effective mAs and maintain a constant value of effective mAs irrespective of pitch value so that dose does not vary when pitch is changed. Many MDCT scanners automatically recommend or make the appropriate tube current adjustment to maintain a constant image noise when pitch is changed. Pitch has relatively lower effect on image quality of MDCT scanners than on single-slice CT scanners. A higher pitch is generally more dose efficient but tends to cause helical artifacts, degradation of section sensitivity profile (slice broadening), and decrease in spatial resolution [18]. Alterations in pitch can have varying effects on image quality in different situations. For instance, in CT colonoscopy and CT angiography studies, image quality and reconstruction artifacts are less affected by the pitch value than by beam collimation, so that a higher pitch with narrow beam collimation may be preferable to reduce radiation dose [19, 20]. However, in situations such as imaging of metastatic liver lesions or pancreatic lesions, which generally require thin collimation, an increased pitch may affect detectability, as lesions may be missed due to degradation of section sensitivity profile [21].

Generally, thicker beam collimation in MDCT results in more dose-efficient examinations, as overbeaming constitutes a smaller proportion of the detected X-ray beam. However, a wider collimation can limit the thinnest reconstructed sections. Conversely, although thin-beam collimation increases overbeaming X-rays, it allows reconstruction of thinner sections. Hence, beam collimation and pitch must be carefully selected to address specific clinical requirements. For instance, a wider collimation and pitch greater than 1:1 are usually sufficient for CT angiography studies and screening CT examinations, such as CT colonography and renal calculus.

Shielding

Shielding devices can be used to protect radiosensitive organs such as the breast, eye lenses, and gonads in pediatric patients and young adults, as these structures frequently lie in the beam pathways. With lead shields, thyroid and breast radiation doses can be reduced by an average of 45 % and 76 %, respectively, in patients undergoing routine head CT [22]. The use of a shield for radioprotection of eye lenses in paranasal sinus CT has also been found to be a suitable and effective means of reducing skin radiation by 40%. Recently, thinly layered bismuth-impregnated radioprotective latex shields have been found to reduce surface dose to breast, thyroid, and lens when they lie in the area of interest. However, use of gonadal shielding during CT examinations is controversial. A testis capsule (shield) can reduce the absorbed dose to the testes in abdominal CT whereas the lead apron is not appropriate for dose reduction to the ovaries (due to their inconstant position).

Metal Artifact Reduction (MAR) Algorithm

CT image quality can be affected by streak or starburst artifacts from metallic implants, such as joint replacement prosthesis, dental implants, or surgical clips. Often, either a second series of images are acquired (for face or neck CT in case of dental implants) to reduce loss of information from these artifacts or tube current is increased in an attempt to reduce these artifacts. Linear interpolation of reprojected metal traces and multidimensional adaptive filtering of raw data have been developed to reduce starburst artifacts from metallic implants' high attenuation objects [23, 24]. The MAR algorithms reduce starburst artifacts from metallic implants and are expected to be released for clinical applications in near future.

Noise-Reduction Filters (NRFs)

Noise-reduction filters (NRFs) have been developed to reduce noise in images obtained with reduced radiation dose or in thin images. Several approaches have been used to reduce noise in scan data sets comprising linear low-pass filter, nonlinear smoothing, and nonlinear three-dimensional filters.

NRFs are based on the principle that any image consists of a set of structural pixels representative of structures of interest and a set of nonstructural pixels representative of nonstructural regions. The NRFs perform isotropic filtering of nonstructured regions with a low-pass filter and directional filtering of the structured regions with a smoothing filter, operating parallel to the edges and with an enhancing filter operating perpendicular to the edges. Two-dimensional (2-D), nonlinear NRFs reduce noise in low-radiation-dose CT images but adversely affect the image contrast and sharpness [25–27]. A recent report documented a three-dimensional (3-D) NRF technique (3-D optimized reconstruction algorithm, or 3-D ORA, Siemens Medical Solutions) that generalizes the 2-D, nonlinear smoothing technique in all three directions (in X-, Y- and Z-axes) in order to avoid loss of image contrast and sharpness [28]. These NRFs may improve image noise without affecting contrast and lesion conspicuity in low-dose CT [28].

Conclusions

Several recent surveys have shown considerable variation in radiation dose with MDCT, which can lead to a higher radiation exposure from MDCT [29, 30]. Appropriate selection of scanning protocols and newer dose reduction techniques can help in radiation dose optimization.

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