CT Radiation Dose Optimization and Estimation: an Update for Radiologists

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In keeping with the increasing utilization of CT examinations, the greater concern about radiation hazards from examinations has been addressed. In this regard, CT radiation dose optimization has been given a great deal of attention by radiologists, referring physicians, technologists, and physicists. Dose-saving strategies are continuously evolving in terms of imaging techniques as well as dose management. Consequently, regular updates of this issue are necessary especially for radiologists who play a pivotal role in this activity. This review article will provide an update on how we can optimize CT dose in order to maximize the benefit-to-risk ratio of this clinically useful diagnostic imaging method.

Index terms: CT radiation dose; Low dose CT; Radiation dose reduction; MDCT

INTRODUCTION

Since the introduction of the multi-section CT, the speed and z-axis coverage of CT scanning have dramatically increased. As a result, the clinical utility has considerably increased in our practice not only in general applications, but also in newer applications such as cardiac CT (1, 2) and dual energy CT (3). As CT utilization increases, the concern about radiation hazards from CT also increases (4). In fact, the worldwide average annual per-capita effective dose from medical procedures has approximately doubled in the past 10-15 years (5). A study (5) has also found an uneven distribution of medical radiation exposure, which is greater in highly developed countries. For example, the 2006 United States data showed that medical imaging contributed to approximately half (3.0 mSv) of the total radiation dose (5.6 mSv) (5, 6). The greatest contributor to medical radiation exposure is CT. In the United States, the number of CT scans is increasing by approximately 10% per year (5, 6). In South Korea, the increasing rate is even steeper, approximately 11-31% per year (7).

In conjunction with the increasing concerns about potential CT radiation hazards, various CT dose-saving strategies have been developed (8, 9). Thus, the benefit-risk ratio of CT examinations can be maximized with optimized CT imaging techniques using these strategies. Although there are several uncertainties in quantifying lifetime risks from CT examinations, per-capita cumulative CT radiation dose should be minimized particularly in the younger population because they have unequivocally higher radiosensitivity and longer life expectancy than the older population. In this article, currently available CT dose-saving strategies will be reviewed, which will ultimately facilitate our rational use of CT.
CT DOSE PARAMETERS

For dose optimization, understanding CT dose parameters including tube potential, tube current, pitch, weighted CT dose index (CTDIw), volume CT dose index (CTDIcon), dose-length product (DLP), and effective dose is a prerequisite. The definition of each parameter and its effects on radiation dose are summarized in Table 1. The ultimate goal of dose optimization is to minimize radiation dose for obtaining diagnostic quality of CT images. Therefore, we need to determine how to obtain diagnostic quality of CT images for dose optimization. The four fundamental elements determining CT image quality consist of image noise, image contrast, spatial resolution, and artifacts (Table 2). In

**Table 1. CT Radiation Dose Parameters**

| Dose Parameter                  | Definition                                                                 | Effects on Radiation Dose                                    | Unit  |
|--------------------------------|---------------------------------------------------------------------------|----------------------------------------------------------------|-------|
| Tube potential or voltage      | X-ray beam energy                                                         | Proportional to square of tube voltage change                | kV    |
| Product of tube current and time | Photon fluence; number of photons in defined exposure time               | Directly proportional to radiation dose                       | mAs   |
| Pitch                          | Ratio of table feed per gantry rotation to nominal width of beam collimation | Inversely proportional to radiation dose                      | -     |
| CTDIw                          | Average radiation dose in scan volume measured in standard CT phantoms; 1/3 CTDI center + 2/3 CTDI periphery | Directly proportional to radiation dose in unit volume, influenced by pitch factor | mGy   |
| CTDIcon                        | CTDIw/pitch                                                               | Directly proportional to radiation dose in unit volume, irrespective of pitch factor | mGy   |
| Dose-length product            | CTDIcon x scan length (cm)                                                | Directly proportional to total scanned radiation dose        | mGy·cm|
| Effective dose                 | Overall risk-related radiation exposure; Σ Wt (tissue weighting factor) x Ht (tissue equivalent dose) | Directly proportional to total scanned radiation dose and overall risk of irradiated tissue | mSv   |

Note.— CTDIcon = volume CT dose index, CTDIw = weighted CT dose index

**Table 2. CT Image Quality Parameters**

| Image Quality Parameter          | Definition                                                                 | Relationships with Radiation Dose Parameters | Calculation Method                                      |
|---------------------------------|---------------------------------------------------------------------------|---------------------------------------------|--------------------------------------------------------|
| Image noise                     | Random variation of CT numbers                                            | Inversely proportional to square root of radiation dose; inversely proportional to tube voltage; inversely proportional to fourth power of spatial resolution; influenced by section thickness and reconstruction algorithm | Standard deviation of measured CT numbers               |
| Contrast-to-noise resolution    | Ability to distinguish between different CT numbers                       | Not significantly changed at different tube voltages in most of materials except for some with high atomic numbers such as iodine (increased at low tube voltage); low contrast resolution greatly influenced by image noise level | (S_A – S_B) / σ, where S_A and S_B are CT attenuation measured in structures A and B and σ is measured image noise |
| Spatial resolution              | Ability to distinguish small details of object                            | Inversely proportional to focal spot size, detector collimation; influenced by reconstruction algorithm | Voxel size; line-pairs per centimeter; point or line spread function; modulation transfer function |
| Artifacts                       | Unwanted structures degrading CT image quality                            | Photon starvation artifacts – inversely proportional to radiation dose; physiologic motion artifacts – inversely proportional to temporal resolution; other artifacts – no direct relationships with radiation dose parameters | -                                                |
principle, radiation dose is inversely proportional to the square of the image noise. Image contrast is significantly augmented by the use of a contrast agent and is influenced by tube potential in some materials with high atomic numbers such as iodine due to the different photoelectric interactions (Fig. 1). The required image quality of CT differs somewhat among different diagnostic tasks. Consequently, the required CT radiation dose is also fairly diverse and thus should be tailored according to clinical indications. For example, very low dose CT can be used for the identification of high-contrast lesions, such as urinary stones, colonic polyps (virtual colonography), or lung nodules.

**STRATEGIES FOR CT DOSE OPTIMIZATION**

To be compliant with the so-called “as low as reasonably achievable (ALARA)” principle, it is imperative to justify CT examinations beforehand. In this respect, radiologists should play an important advisory role in this decision with referring clinicians. When equal or greater diagnostic yields are expected, CT should be replaced by alternative imaging modalities with no or less ionizing radiation, such as sonography, magnetic resonance (MR) imaging, or radionuclide voiding cystography. On the other hand, radiologists should make every effort to reduce the radiation dose of CT examinations while maintaining diagnostic quality when CT is indicated (8-10). For example, minimizing the scan range of CT examinations as required is a straightforward way to achieve this goal. For multi-phase CT protocols, the number of repeated scanings should be minimized and precontrast scanning should be used only when diagnostic information on precontrast CT images is not obtainable from postcontrast scanning. Because of the substantial radiation dose of perfusion CT, its clinical indication and imaging protocol should be carefully determined (11). In the following sections, other useful CT dose-saving strategies are described with recent updates. A checklist for CT dose optimization is described in Table 3.

**Body Size-Adapted CT Protocols**

Body size-adapted CT protocol is a fundamental part of CT dose optimization because the minimal radiation dose required for diagnostic image quality would be varied even at the same diagnostic task depending on body size and habitus. Then, the optimal tube voltage and current should be determined for the adapted radiation dose. One of the common misconceptions is that lowering the tube voltage at the same tube current is a good strategy for low dose CT. Actually, a higher tube current should be used at a lower tube potential to compensate for the increased image noise (12, 13). Another misconception is that tube current modulation can automatically adapt the CT radiation dose to different body sizes and anatomic regions and therefore allows a constant image quality. However, this is erroneous and a user-defined target image quality considering patient size and diagnostic task should be determined for each anatomic region and each tube voltage.

Various patient size parameters have been used for body size-adapted CT protocols for both children and adults (13, 14). Among the parameters, body weight or body mass index has been traditionally applied to body size-adapted CT protocols due to its easy applicability (13, 14). Many investigators however have found that cross-sectional dimensions are better adapted to body habitus than the traditional parameters, i.e. body weight or body mass index (15-19). An attenuation-based parameter using a scout has been used in other studies (20, 21). The image noise
measured on calcium scoring images (22) or a timing bolus image (23) was used to obtain uniform image quality of cardiac CT. Nevertheless, such parameters other than body weight or body mass index have not been commonly used for body size-adapted CT protocols mainly because of a difficulty in clinical implementation. Recently, a practical pediatric chest CT protocol based on cross-sectional area and mean attenuation of the body was developed and provided less noise variation irrespective of body habitus (24). In a study (25), CTDIvol was used to represent an optimal radiation dose for an individual, and other factors determining image noise such as section thickness, image reconstruction algorithm, and tube voltage were also taken into consideration to establish the optimal radiation dose. A CT protocol derived from a best-fit equation is generally preferred to a classic CT dose table or chart.

### Tube Current Modulation

Tube current modulation greatly contributes to CT dose optimization by reducing the CT dose according to body size, shape, and attenuation without degrading image quality. The tube current may be adjusted in the x-y plane (angular mode), the z-axis, or a combination of both. CT dose reduction achieved by tube current modulation has been reported to be up to 26-50% in children and adults (25-27). We need to understand the principles of different tube current modulation techniques for their proper use (28). In addition, a patient should be positioned at the CT isocenter to avoid faulty modulation of the tube current. Other factors influencing tube current modulation include tube voltage, maximum tube current, scan speed, and scan direction (29-31). Indeed, the fact that the tube current is frequently saturated to its maximum level at a lower tube voltage, faster scan speed, or a combination of both has seldom been recognized (30, 31). With thicker or denser body parts, the tube current saturation occurs earlier. Since image quality is subject to deterioration resulting from the increased image noise at regions scanned at a saturated tube current, the use of a higher tube voltage, slower scan speed (either slower gantry rotation time or a lower pitch), or a combination of both should be considered as a remedy for this potential pitfall. As previously mentioned, the target image quality index for tube current modulation, which differs according to patient size, each anatomic
region, individual diagnostic task, and tube voltage, should be set currently by an operator. In this regard, we should be aware that tube current modulation is not truly automatic. The experience on how we can determine this target image quality index for tube current modulation is very limited (24, 32, 33).

**Optimal Tube Voltage**

Optimal tube voltage should be determined for patient size and each type of CT examination to achieve an optimal tradeoff between contrast, noise, artifacts, and scanning speed (13, 34, 35). Concrete knowledge on CT physics and the diagnostic purposes of CT examination is mandatory for this task. The importance of optimal tube voltage has been recently emphasized for CT dose optimization in order to maximize the clinical benefits of CT examination at a low radiation dose and in order to determine the most dose-efficient tube voltage. Based on CT physics, the iodine contrast, image noise, and iodine contrast-to-noise ratio (CNR) show different behaviors at different tube voltages and different phantom sizes: increasing iodine contrast at lower tube voltages, that is, decreasing for larger phantoms; almost identical noise level for a 10-cm phantom and a dramatic increase in noise level for a 40-cm phantom; markedly increasing iodine CNR at lower tube voltages for a 10-cm phantom; and minimally increasing iodine CNR at lower tube voltages for a 40-cm phantom (34). In regard to the diagnostic task, radiologists should determine the degree of importance of iodine CNR or image noise for a particular type of CT examination. In general, iodine CNR is more important in contrast-enhanced examinations, while image noise is more important in precontrast examinations or in detecting low-contrast lesions and is less important in detecting high-contrast lesions. In contrast to the benefits of lower tube voltages to contrast-enhanced CT, the potential benefits of higher tube voltages to CT exams requiring lower image noise have not been thoroughly investigated. For instance, image noise should be sufficiently low to increase low-contrast detectability. The reduction of image noise can be achieved not only by using an adaptive noise reduction filter (36) or sliding-thin-slab averaging algorithm (37), but also by simply using a higher tube voltage. Likewise, a higher noise at lower tube voltages may adversely affect the assessment of ground-glass opacity in the lungs (38). Higher tube voltages are also commonly used in an unenhanced brain CT requiring lower noise in assessing low-contrast intracranial structures (13, 39, 40). Furthermore, a higher tube voltage may produce less severe artifacts from metallic objects or thick bones such as the skull base than a lower tube voltage. A general strategy for selecting optimal tube voltage at different phantom sizes and different noise constraints (reflecting different diagnostic tasks) was recently proposed (41, 42).

**Scan Modes**

Several CT scan modes are available for clinical CT examinations, including spiral scanning with or without ECG synchronization, sequential scanning with or without ECG synchronization, and dual energy spiral scanning. Dose issues specific to each scan mode are described in the following section.

**Overbeaming and Overranging**

Overbeaming is the waste dose beyond the edge of the detector rows of a multi-section CT (Fig. 2). The magnitude of overbeaming is inversely proportional to the number of detector rows. Therefore, its contribution to unnecessary radiation exposure to patients has reduced with modern multi-section CT systems. To acquire spiral scanning, a CT system needs at least half a rotation beyond the planned scan length in order to reconstruct the first and the last images (Fig. 2). This unnecessary
radiation exposure outside the planned scan length is called as overranging. Overranging is proportional to beam collimation, reconstructed slice width, and pitch, while it is irrespective of a planned scan length (43). The contribution of overranging to the total CT dose is therefore considerably higher for CT examinations with shorter scan ranges such as pediatric CT and cardiac CT (44). Fortunately, adaptive section collimation technology was recently developed to eliminate overranging during spiral scanning (45) (Fig. 2). This useful collimation technology is currently available only in some of the CT models.

**Retrospectively ECG-Gated Spiral Scanning**

Cardiac CT with retrospective ECG gating was renowned for delivering higher radiation exposure to patients. The radiation dose of cardiac CT is now comparable to or even lower than that of chest CT due to innovative dose-reducing techniques (46). The relatively higher radiation dose, approximately 15 mSv on average, of retrospectively ECG-gated spiral scanning used for cardiac or coronary CT angiography is mainly attributed to the low pitch factor. For example, 0.2. ECG-based tube current modulation, in which 20% or 4% of the nominal value is used outside a target cardiac phase, can reduce the radiation dose of retrospectively ECG-gated spiral scanning to approximately 10 mSv or 6 mSv, respectively (2, 47). The use of heart rate-dependent pitch values can additionally reduce the radiation dose of cardiac CT at higher heart rates (48).

**Prospectively ECG-Triggered Sequential Scanning**

Prospectively ECG-triggered sequential (or step-and-shoot) scanning can further reduce the radiation dose of cardiac CT in the range of 1-4 mSv (2, 46, 49). This sequential scanning can be used in other body regions. However, potential pitfalls related to this scan mode, including prolonged scan time and stair-step artifacts due to different contrast enhancement or motion misregistration, should be carefully considered prior to the examination (2).

**High-Pitch Dual Source Spiral Scanning**

High-pitch dual source spiral scanning with or without ECG triggering is the most recent advance in CT imaging techniques. With this scan mode, pitch can be increased to 3.0-3.4, which results in a substantial reduction of radiation dose of cardiac CT, by approximately 1 mSv (46, 50). This high-pitch spiral scanning considerably decreases not only motion artifacts but also the requirement of sedation in free-breathing patients. Hence, the scan mode is regarded as very useful in pediatric patients and uncooperative adult patients (51). However, it should be noted that overranging increases with this scan mode due to a combination of high pitch and longer collimation. Moreover, adaptive section collimation technology to protect this overranging is not available for this scan mode.

**Dual Energy Scanning**

Dual energy CT has expanded clinical applications of CT examinations, along with cardiac CT. Dual energy scanning can be performed with either two X-ray sources, kVp switching of one X-ray source, or dual-layer (“sandwich”) detectors (3, 52, 53). Each method has different radiation dose profiles. Dual energy scanning using a dual-source CT system is almost dose-equivalent to single energy scanning (3, 54-59). In contrast, the radiation dose of dual energy scanning using a single-source system with rapid kVp switching is currently higher (e.g. 8 mSv for dual energy chest CT) than that of dual-source dual energy scanning or single energy scanning (52). A recent study (60) showed that the use of additional tin filtration in the high-energy X-ray beam of a dual-source CT system provided several benefits for dual energy CT applications, including a similar or lower radiation dose compared with the conventional single energy CT, increased dual-energy contrast, and improved image quality of dual-energy material-specific (e.g. virtual noncontrast) images. Moreover, the virtual noncontrast imaging of dual energy CT has a potential to reduce the radiation dose by omitting precontrast scanning (61).

**Noise-Reducing Image Reconstruction Algorithms**

The use of noise-reducing image reconstruction algorithms may have a potential to reduce the CT radiation dose. However, conventional noise-reduction filters decrease image noise but simultaneously decrease lesion contrast and conspicuity (62) (Fig. 3). This trade-off actually limits the dose-saving potential of these noise-reduction filters. Recently, decoupling between image noise and spatial resolution has been performed in noise-reducing image reconstruction algorithms using iterative reconstruction (63, 64). Consequently, a 40-50% dose reduction can be achieved by means of iterative reconstruction algorithms without degrading image quality. Iterative reconstruction algorithms are continuously improving in terms of image...
CT DOSE ESTIMATION

The actual risks of radiation exposure from low-dose diagnostic imaging are considerably uncertain (6). However, many investigators believe that low levels of ionizing radiation in the range of 5-125 mSv have a very small but statistically significant increase in cancer risk. Several factors predominantly influencing cancer risk from radiation exposure should be carefully considered in establishing CT protocols. These include genetic susceptibility, age at exposure, and sex. The estimation of CT dose helps to provide some direction in terms of CT dose optimization. The estimation itself also may increase the awareness of the necessity of CT dose optimization. For dose estimation, the effective dose is generally used because the concept can indicate the amount of whole-body average irradiation resulting from the partial-body irradiation of diagnostic imaging and can be used to compare radiation doses between different procedures. Two methods are used to calculate the dose estimates: one method is the Monte Carlo simulation by using standard mathematical phantoms, for which CT dosimetry calculators such as ImPACT (UK National Health Service CT Evaluation Centre, London, England) and CT Expo (Medizinische Hochschule, Hannover, Germany) are commercially available; the other method is a DLP-based method in which CT dose estimates are calculated by multiplying a DLP value and an appropriate conversion factor. The former also can provide estimates of the lifetime attributable risk of cancer. Tissue weighting factors were recently updated in the International Commission on Radiological Protection (ICRP) publication 103. Major changes involve the gonads (0.20 → 0.08) and breast (0.05 → 0.12), which culminates in a decrease in dose estimates of pelvic CT and abdominopelvic CT and an increase in chest CT and cardiac CT. Accordingly, conversion factors for the DLP-based method should be updated (69) (Table 4). We should take notice that the effective dose of the same CT examination is subject to being changed according to the conversion factors used for calculating DLP-based CT dosimetry. Therefore, it is highly recommended that CTDIvol and DLP values be described along with CT dose estimates.

Data on cumulative CT dose estimates can be used to further tailor and eventually improve CT dose optimization strategies or guidelines. This type of investigation conducted in patients who have CT examinations at a young age will be of a great value, as pointed out in the report on the Biologic Effects of Ionizing Radiation (BEIR) VII.
In fact, such studies based on actual individual CT dosimetry are ongoing in South Korea and other countries. Diagnostic reference levels (DRLs) of various types of CT examinations, defined as the third quartile of the collected dose data, offer good reference values in CT dose optimization. This activity identifies high dose practices and encourages the imagers to determine the causes and solutions to using a relatively higher dose, which results in modification of such practices to lower the dose. The DRLs have shown a tendency to gradually decrease due to recent developments in dose-reducing techniques and increased awareness of radiation dose issues, but they still show wide variations (71-75). The DRLs of head, chest, and abdominal CT examinations in adults from different countries are described in Table 5.

Regarding multiple CT scans without table movement,
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![Graph showing difference in radiation dose between peak dose and CT dose index](modified from reference [60]). Dose from CT scanning without table increment is overestimated by factor of two or more with CT dose index (CTDI) values in comparison with point dose values. such as perfusion CT, conventional CTDI was found to overestimate the dose by a factor of two or more, compared with point dose values (76) (Fig. 4).

**CONCLUSION**

In this review article, the essential requirements and latest updates of CT dose optimization for radiologists are described. Because strategies for CT dose optimization and estimation of radiation risks are constantly evolving and being updated, educational efforts including this review article should also be continuous and regularly updated. From this study, radiologists will undoubtedly be ready for exploring the clinical benefits of CT.

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