CT Radiation Dose Reduction by Modifying Primary Factors

Sarabjeet Singh, MD, Mannudeep K. Kalra, MD, James H. Thrall, MD, Mahadevappa Mahesh, MS, PhD

Given the steep trajectory of evolution in multidetector CT technology and its complexity, dose reduction efforts require knowledge of interplay between CT scanning parameters and their impact on radiation dose and image quality in CT examinations. We describe the effects of modifying primary scanning factors such as tube current, tube potential, and pitch factor (applicable to helical CT scanning) in CT.

PRIMARY SCANNING FACTORS

Tube Current

Tube current adjustment is one of the most common approaches to optimizing radiation dose [1]. Expressed in milliamperes, tube current is the number of x-ray photons produced and has a direct linear relationship with radiation dose. On the other hand, image quality expressed in terms of image noise is inversely proportional to the square root of change in the applied tube current, which implies that if tube current is halved, there is 50% reduction in radiation dose and about a 40% increase in image noise [1].

Tube current can be optimized with a manual selection of fixed tube current or modulating tube current (often called dose modulation, automatic exposure control) [2,3]. Tube current modulation techniques in multidetector CT scanners (especially 16-slice and beyond) work best for most body CT examinations and are becoming efficient even for pediatric CT examinations.

The use of fixed tube current for radiation dose optimization should imply a single–tube current acquisition of a scan series, not a single–tube current scanning of all patients or clinical indications. Patient age and size, clinical indication, and the body region being imaged should determine specific fixed tube current values.

Younger patients, particularly children, should be scanned at lower tube currents with radiation doses as low as reasonably achievable and certainly lower than for most adult patients [4]. Several investigators have described the use of lower fixed tube current settings for scanning children using body weight, length, or regional diameter. For example, Frush et al [5] described a color-coded Broselow-Luten pediatric system based on patient weight or length for reducing radiation dose for pediatric CT using different fixed tube current settings for children belonging to different color zones. Because of large size variations in children, from as small as 2 to 3 kg to >100 kg, it is important to stratify fixed tube current settings for pediatric body CT in ≥5 or 6 size categories to ensure that infants and smaller children are scanned at lower fixed tube currents compared with larger children [4,5]. For radiation dose optimization in children, the Image Gently[SM] campaign of the Alliance for Radiation Safety in Pediatric Imaging also offers an approach for optimizing the use of fixed tube current on the basis of the anteroposterior diameter of children in the chest, abdomen, and head [6].

Because head size does not vary much in adults, a single fixed tube current is used for most adult head CT scans. It then becomes imperative for users to ensure that the volume CT dose index of routine or standard head CT examinations (with 16-cm phantom size) does not exceed the reference level of 75 mGy set forth in the ACR's CT accreditation program [7,8].

On the other hand, adult body CT examinations must generally be performed with dose modulation techniques when available. With the use of fixed tube current settings for adult body CT, patient size (especially lateral dimension) should be the first guiding factor for optimizing fixed tube current. Although body weight remains the most commonly used metric to adjust fixed tube current, other studies have also described the use of transverse diameter estimated from anteroposterior localizer radiographs [3,4].

Smaller adults need lower fixed tube currents compared with average-sized adult, whereas larger than average patients need greater tube currents. In general, obese adult patients may need up to 2 times higher fixed tube current for obtaining adequate diagnostic images of the abdomen compared with an average-sized adult [9]. As stated earlier, chest CT (Figure 1) should be performed at lower fixed tube currents compared with abdominal CT (Figure 2). Although lower

0091-2182/11/$36.00 • DOI 10.1016/j.jacr.2011.02.001

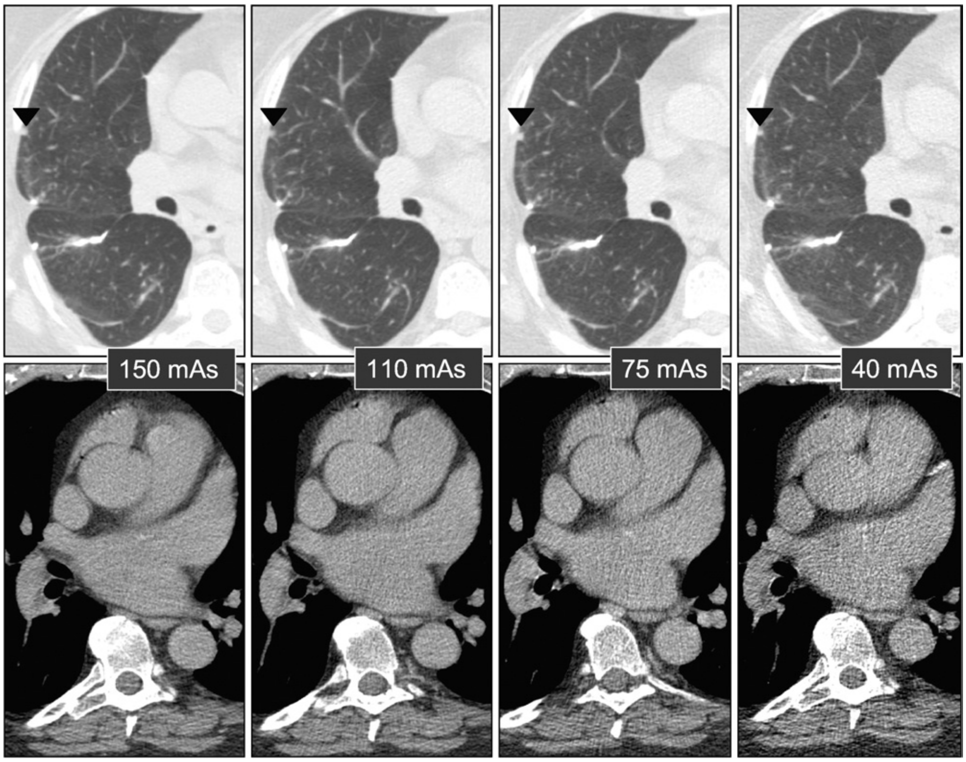

Fig 1. Transverse chest CT images acquired at tube current–time product levels of 150, 110, 75, and 40 mAs, with the remaining scan parameters held constant. Subpleural nodule (black arrow) is seen at all 4 radiation dose levels. Although there was an increase in image noise with decrease in radiation dose, the images were acceptable for diagnostic interpretation at all dose levels.

fixed tube current results in increased noise, which is more appreciable in the mediastinum, prior studies report that tube current can be reduced down to 15 to 50 mA without affecting the detection of lung or mediastinal abnormalities [10].

The process of radiation dose optimization with fixed tube current selection must begin with making age-specific and size-specific protocols but must continue onward, tailoring radiation doses according to specific clinical indications in each body region [4]. For chest CT, fixed tube current settings for individual patient size groups should be adjusted to reduce radiation doses for the evaluation of lung nodules, lung cancer screening, air trapping or expiratory phase imaging, quantification of emphysema, and follow-up imaging. These indications allow users to reduce radiation dose further for chest CT examinations. For example, Naidich et al [10] showed that fixed tube currents as low as 10 to 20 mA at 120 kVp are sufficient for detection of lung nodules and architecture.

Other important indications for low fixed tube current settings are beneficial in screening CT examinations for evaluation of coronary artery calcification, lung cancer, and colonic polyps.

Tube Potential

Tube potential, represented as peak kilovoltage, indicates the energy of the x-ray photons. Change in radiation dose is approximately proportional to the square of the change in applied tube potential. A smaller change in tube potential, for example, from 140 to 120 kVp, leads to a larger decrease in radiation dose by about 35% to 40% [2]. With a decrease in tube potential, image noise increases, as does image contrast, particularly between water and positive contrast media. For high-contrast situations or regions, such as CT angiography or chest CT, lower tube potential does not necessitate a concomitant increase in tube current. In certain low-contrast situations, such as liver lesion evaluation, and in larger patients, some increase in tube current is beneficial in reducing noise while enhancing contrast.

Most infants can be scanned at 80 kVp, with the dual advantage of achieving not only lower radiation dose but also improved image contrast [4]. Weight-based reduction of radiation dose with the use of

Fig 2. Transverse abdominal CT images acquired at tube current–time product levels of 200, 150, 100, and 50 mAs, with the remaining scan parameters held constant. Left renal cyst (white arrow) measuring 2 cm is seen at all 4 radiation dose levels. However, conspicuity of small liver vessels (black arrows) is compromised at 50 mAs because of increased image noise compared with images acquired at 100 to 200 mAs.

lower tube potential and tube current for pediatric chest and abdominal CT has been reported in prior publications [5,9]. In particular, lower tube potential (100-kVp) scanning is being increasingly used to reduce radiation dose during CT angiography in thin patients. At Massachusetts General Hospital, most CT angiographic examinations in children are performed at 80 to 100 kVp to reduce radiation dose while improving image contrast because of the higher sensitivity of lower tube potentials for iodinated contrast media.

In adult patients, lower tube potential can be used in several CT protocols for reducing radiation dose while maintaining diagnostic information. Most important, a lower tube potential of 80 kVp should be used for stroke perfusion CT protocol (dynamic CT scanning) to increase the sensitivity of CT for detection of iodinated contrast media and also to reduce cumulative skin dose. The use of 80 and 100 kVp has been reported for CT pulmonary angiographic evaluation for pulmonary embolism. For CT pulmonary angiography, Sigal-Cinqualbre et al [11] reported the use of 80 kVp for patients weighing <60 kg and 100 kVp for those weighing 60 to 75 kg, with 40% to 50% decreases in associated radiation dose.

The use of lower tube potential has also been recommended for nonobese patients undergoing coronary CT angiography, with 80 kVp reserved for patients weighing <60 kg and 100 kVp for subjects weighing <85 to 90 kg (body mass index < 30 kg/m^2) [11,12]. Coronary CT angiography performed at 80 or 100 kVp for these subjects can result in 50% to 80% dose reductions compared with 120 kVp.

Most routine abdominal and pelvic CT examinations in adults are performed at 120 kVp. In the abdomen and pelvis, the use of lower tube potential has been described for aortoiliac CT angiography, for which the use of 100 kVp can result in about a 35% dose reduction compared with 120 kVp, without affecting diagnostic quality [12]. Advances in x-ray tube technology are beginning to allow automatic tube potential modulation, which will enhance further radiation dose reduction.

Pitch

Pitch is defined as the ratio of the table travel during an x-ray tube rotation to the x-ray beam width [2]. Pitch < 1 implies an overlapping x-ray beam, whereas pitch > 1 allows nonoverlapping scanning with gaps between the radiation beams. In certain CT scanners, the change in radiation dose with alteration in pitch is offset by a corresponding change in tube current to maintain a constant image noise. For example, when pitch is increased, such scanners increase the tube current as well, and this is expressed as "effective tube current–time product" [13]. Thus, the selection of pitch must be guided by the requirement of scanning speed or time and thinner reconstructed slice thickness. To enable shorter scan duration, for example, in CT angiography or in uncooperative or breathless patients, a higher nonoverlapping pitch may be preferred. For routine abdominal imaging, a pitch of 1 is normally set, although a pitch > 1 is feasible. For most cardiac CT scans (retrospective electrocardiographically gated coronary CT angiography), low pitch factors (0.2-0.4) are often used to avoid any type of data gaps and motion artifacts, which translates to higher dose. Certain multidetector CT scanners, in particular the dual-source CT scanners, allow very high nonoverlapping pitch (pitch factors up to 3.0 to 3.6) for faster scanning while reducing radiation dose, particularly for cardiac CT angiography (generally in patients with regular heart rates of <60-65 beats/min) and in children [14].

CONCLUSION

Understanding of the primary scanning factors, such as tube current and tube potential, can help build scanning protocols that allow radiologists to obtain necessary diagnostic information while reducing radiation doses to as low as reasonably achievable.

REFERENCES

1. Kalra MK, Maher MM, Toth TL, et al. Strategies for CT radiation dose optimization. Radiology 2004;230:619-28.
2. Mahesh M. MDCT physics: the basics—technology, image quality and radiation dose. Philadelphia, Pa: Lippincott Williams & Wilkins; 2009.
3. McCollough CH, Primak AN, Braun N, Kofler J, Yu L, Christner J. Strategies for reducing radiation dose in CT. Radiol Clin North Am 2009;47:27-40.
4. Singh S, Kalra MK, Moore MA, et al. Dose reduction and compliance with pediatric CT protocols adapted to patient size, clinical indication, and number of prior studies. Radiology 2009;252:200-8.
5. Frush DP, Soden B, Frush KS, Lowry C. Improved pediatric multidetector body CT using a size-based color-coded format. AJR Am J Roentgenol 2002;178:721-6.
6. Alliance for Radiation Safety in Pediatric Imaging. Image Gently. Available at: http://www.pedrad.org/associations/5364/ig/. Accessed January 25, 2011.
7. Smith AB, Dillon WP, Lau BC, et al. Radiation dose reduction strategy for CT protocols: successful implementation in neuroradiology section. Radiology 2008;247:499-506.
8. American College of Radiology. ACR CT Accreditation Program requirements. Available at: http://www.acr.org/accreditation/computed/ct_reqs.aspx. Accessed January 25, 2011.
9. Uppot RN, Sahani DV, Hahn PF, Gervais D, Mueller PR. Impact of obesity on medical imaging and image-guided intervention. AJR Am J Roentgenol 2007;188:433-40.
10. Naidich DP, Marshall CH, Gribbin C, Arams RS, McCauley DI. Low-dose CT of the lungs: preliminary observations. Radiology 1990;175:729-31.
11. Sigal-Cinqualbre AB, Hennequin R, Abada HT, Chen X, Paul JF. Low-kilovoltage multi-detector row chest CT in adults: feasibility and effect on image quality and iodine dose. Radiology 2004;231:169-74.
12. Schindera ST, Graca P, Patak MA, et al. Thoracoabdominal-aortoiliac multidetector-row CT angiography at 80 and 100 kVp: assessment of image quality and radiation dose. Invest Radiol 2009;44:650-5.
13. Mahesh M, Scatarige JC, Cooper J, Fishman EK. Dose and pitch relationship for a particular multislice CT scanner. AJR Am J Roentgenol 2001;177:1273-5.
14. Achenbach S, Marwan M, Schepis T, et al. High-pitch spiral acquisition: a new scan mode for coronary CT angiography. J Cardiovasc Comput Tomogr 2009;3:117-21.

Sarabjeet Singh, MD, Mannudeep K. Kalra, MD, and James H. Thrall, MD, are from Massachusetts General Hospital, Boston, Massachusetts. Mahadevappa Mahesh, MS, PhD, is from The Russell H. Morgan Department of Radiology and Radiological Science, Johns Hopkins University, Baltimore, Maryland.

Mahadevappa Mahesh, MS, PhD, Johns Hopkins University, The Russell H. Morgan Department of Radiology and Radiological Science, 601 N Caroline Street, Baltimore, MD 21287-0856; e-mail: mmahesh@jhmi.edu.