Optimal Tube Potential for Radiation Dose Reduction in Pediatric CT: Principles, Clinical Implementations, and Pitfalls[1]

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LEARNING OBJECTIVES

After reading this article and taking the test, the reader will be able to:

- Describe the basic principles of the use of optimal tube potential for reducing radiation dose in pediatric CT examinations.
- Discuss how optimal tube potential is determined for pediatric CT examinations.
- Identify the common pitfalls associated with the use of a lower tube potential in pediatric CT.

TEACHING POINTS

See last page

In addition to existing strategies for reducing radiation dose in computed tomographic (CT) examinations, such as the use of automatic exposure control, use of the optimal tube potential also may help improve image quality or reduce radiation dose in pediatric CT examinations. The main benefit of the use of a lower tube potential is that it provides improved contrast enhancement, a characteristic that may compensate for the increase in noise that often occurs at lower tube potentials and that may allow radiation dose to be substantially reduced. However, selecting an appropriate tube potential and determining how much to reduce radiation dose depend on the patient's size and the diagnostic task being performed. The power limits of the CT scanner and the desired scanning speed also must be considered. The use of a lower tube potential and the amount by which to reduce radiation dose must be carefully evaluated for each type of examination to achieve an optimal tradeoff between contrast, noise, artifacts, and scanning speed.

Abbreviations: AEC = automatic exposure control, CNR = contrast-to-noise ratio, $CTDI_{vol}$ = volume CT dose index

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Introduction

The use of computed tomography (CT) has increased substantially over the past 30 years because of its important role in depicting and staging disease (1). However, concerns have recently arisen about the potential risk for cancer induction that results from the radiation used in CT (2–4). Although the existence of such a risk posed by the amount of radiation typically delivered in diagnostic CT examinations remains controversial, the consensus is that patients should not receive more radiation than is necessary, a belief that is consistent with the concept of "as low as reasonably achievable" (ALARA) (5,6). Minimizing radiation doses delivered during pediatric CT examinations is particularly important because the radiation risk for children is two to three times greater than the risk for adults (7). This is because children are more sensitive to radiation than adults are, and they have a much longer life expectancy (7).

A common method to reduce radiation dose is to adjust the x-ray tube current by using weight- or size-based technique charts (8–10). Another important technique is the use of automatic exposure control (AEC), which automatically adapts the tube current in both angular and longitudinal directions according to the patient's size to maintain predefined image noise or image quality characteristics (11–14). Although this widely used approach may help achieve a dose reduction of 40%–50% without sacrificing image quality, it may be less effective in pediatric patients (12,13,15,16).

Use of optimal tube potential is another important technique for reducing radiation dose in pediatric CT examinations. Many researchers have studied the use of low-tube-potential CT with the purpose of improving image quality or reducing radiation dose, particularly among pediatric patients (17–34). On the basis of these studies, the use of a lower tube potential appears to be appropriate because of improved enhancement of iodinated contrast material and no substantial increase in image noise. However, selecting an appropriate tube potential and determining how much to reduce the radiation dose is not a straightforward task. Scanning speed, motion artifacts, patient size, and diagnostic task must be considered and carefully evaluated before the patient examination.

In this article, we summarize existing techniques for reducing radiation dose in pediatric CT examinations, provide a tutorial for optimizing tube potential in pediatric CT examinations, and describe how to implement a technique chart for tube potential and tube current settings for pediatric body CT examinations. We also discuss special considerations and common pitfalls associated with the use of lower tube potentials for pediatric imaging.

Currently Available Techniques for Reducing Dose in Pediatric CT

Before performing a pediatric CT examination, it is essential to fully evaluate the risks and benefits of the requested examination. CT is justified only if the benefit clearly exceeds the potential risks associated with exposure to radiation (35). Alternative imaging modalities that use less or no radiation should be considered if they are able to fulfill the clinical imaging task (36).

When performing a CT examination, every effort must be made to optimize the scanning protocol so that a minimum radiation dose is used to obtain the necessary diagnostic information. Previously, considerable attention was paid to inappropriately used adult techniques in children and small adults (37,38). Currently, it is common practice in the CT community to adapt the dose level to the patient's size, which is also a requirement for American College of Radiology (ACR) accreditation of pediatric CT facilities (39,40). Scanning techniques that depend on patient size include one or more of the following elements: size-dependent beam-shaping filters, manual tube current technique charts, AEC, and optimal tube potential (41).

Teaching Point

Patient Size–Dependent Beam-Shaping Filter

Beam-shaping filters (eg, bowtie filters) are designed to reduce the intensity of incident x-rays toward the periphery of the body in the axial plane, resulting in stronger-intensity x-rays in the

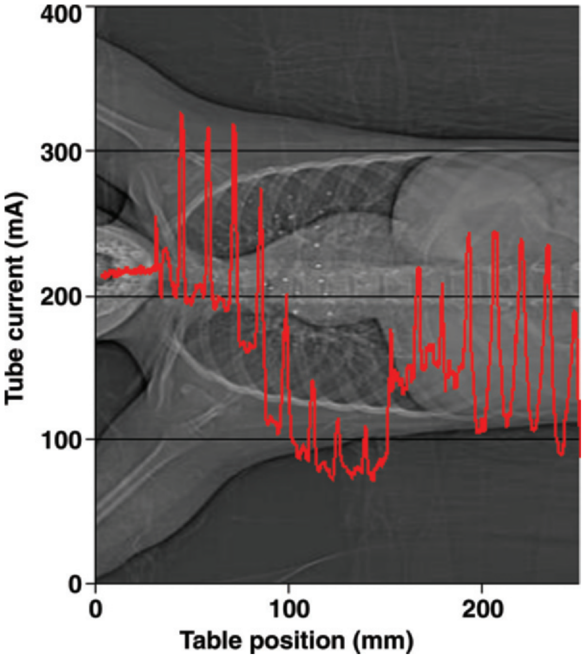


Figure 1. Use of AEC to reduce radiation dose in a pediatric patient. Graph superimposed on a chest CT radiograph shows the tube current, which was automatically modulated according to the attenuation level at each projection view angle. The projection view angles correspond to a particular table position, a result of the continuous translation of the patient table that occurs during helical scanning.

center of the body and weaker-intensity x-rays in the peripheral regions, a characteristic that corresponds to the attenuation levels in an oval-shaped human body. Beam-shaping filters substantially reduce the delivered radiation dose, especially to the patient's skin. However, because of their smaller size, children require a beam-shaping filter specifically designed for a smaller body; those that are designed for adults are ineffective in children (42). Because of the geometric dependencies of the beam-shaping filter, patient centering in the scan field of view is critical.

Manual Tube Current Technique Chart

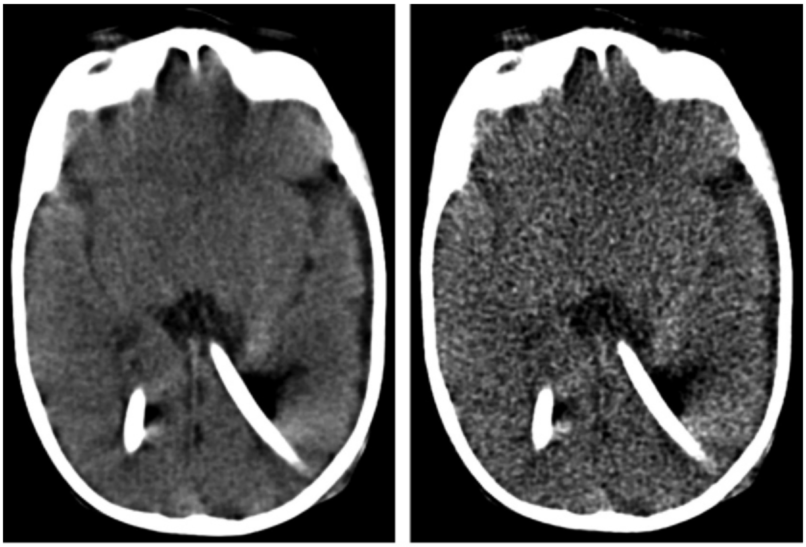
Because of their smaller size, children attenuate the x-ray beam much less than adults do. Thus, children require less radiation to achieve sufficient diagnostic image quality. Manual adjustment of the tube current on the basis of patient size is the most straightforward way to reduce radiation dose in children. Reducing scanner output from that required for an adult to a level suitable for a child depends on the diagnostic task being performed. For body CT in infants, the tube current setting may be reduced from the adult setting by a factor of 4–5 (43). For head CT in infants, a tube current reduction factor of 2–2.5 is appropriate (43).

Automatic Exposure Control

With AEC, the tube current is automatically modulated according to the attenuation level (ie, the size) of the patient (Fig 1). For smaller patients, the tube current is automatically decreased to adapt to the lower attenuation level. Conversely, the tube current is appropriately increased for larger patients. The use of AEC is an efficient way to tailor radiation dose to achieve a target image quality. The paradigm for defining target image quality varies among the different CT manufacturers. For instance, GE uses the term *noise index*, Siemens uses the term *quality reference mAs*, Philips uses the term *reference image*, and Toshiba uses the term *standard deviation*.

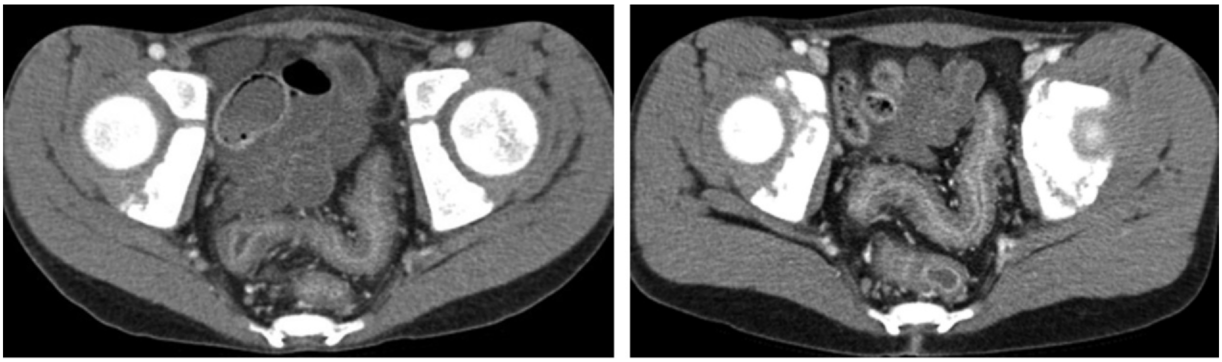
Although the use of AEC is an efficient way to achieve predetermined and consistent image quality, the operator remains responsible for selecting an appropriate target image quality for each diagnostic task and patient. Typically, pediatric studies need lower noise and thinner sections than are used for adult examinations. Use of a higher-than-necessary target image quality will result in an unnecessarily high radiation dose (41).

Characterization of solid organs and depiction of low-contrast lesions generally requires lower noise levels and, thus, higher radiation doses. Conversely, higher noise levels and lower radiation doses may be tolerated for evaluation of high-contrast structures. Evaluating previous CT images also may help determine if the use of a lower radiation dose is acceptable (44). In Figure 2, an 80% dose reduction was acceptable for a shunt follow-up examination of a 15-month-old girl.



a. b.

Figure 2. Eighty-percent dose reduction in a 15-month-old girl. **(a)** CT image acquired with the original scanning technique (volume CT dose index [$CTDI_{vol}$], 24.2 mGy) for a shunt follow-up examination. **(b)** CT image, reconstructed after simulation of image data acquired at one-fifth of the original dose, shows that the 80% dose reduction is diagnostically acceptable for the purpose of assessing the size of the ventricle.



a. b.

Figure 3. Improved contrast enhancement at a lower tube potential in an 11-year-old boy. Two CT enterographic examinations were performed in a 4-month interval, one at 120 kV with a $CTDI_{vol}$ of 5.18 mGy **(a)** and the other at 100 kV with a $CTDI_{vol}$ of 3.98 mGy **(b)**, following a 50-second delay after contrast material injection. The 100 kV image shows improved contrast enhancement and visualization of mural stratification. (Reprinted, with permission, from reference 41.)

Principle of Optimal Tube Potential Techniques for Reducing Dose in Pediatric CT

Contrast

The use of a lower tube potential to reduce radiation dose in pediatric patients has been actively investigated (18–20,24,30). Most CT examinations involve the use of iodinated contrast material. The increase in iodine attenuation at lower tube potentials provides more iodine signal and improves the conspicuity of hyper- or hypovascular structures (33,34). Figure 3 shows CT images obtained at 100 kV and 120 kV in an 11-year-old boy. Because of the increased enhancement of iodine at 100 kV, mural stratification is more visible in the 100-kV image.

Teaching Point

The increased enhancement of iodine on CT images obtained at a lower tube potential is fundamentally due to the increased linear attenuation coefficient of iodine relative to that of water (Fig 4) (45). The energy dependence of the linear attenuation coefficients for iodine and water differ, primarily due to the photoelectric interaction, which is not only inversely propor-

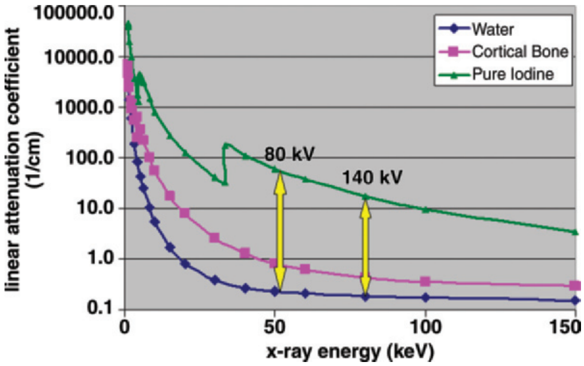


Figure 4. Graph shows the linear attenuation coefficients of iodine, cortical bone, and water, which are plotted as a function of x-ray energy. Note the difference between the linear attenuation coefficients of iodine (top curve) and water (bottom curve) at the mean energy of a typical 80-kV (long arrow) and 140-kV (short arrow) x-ray beam.

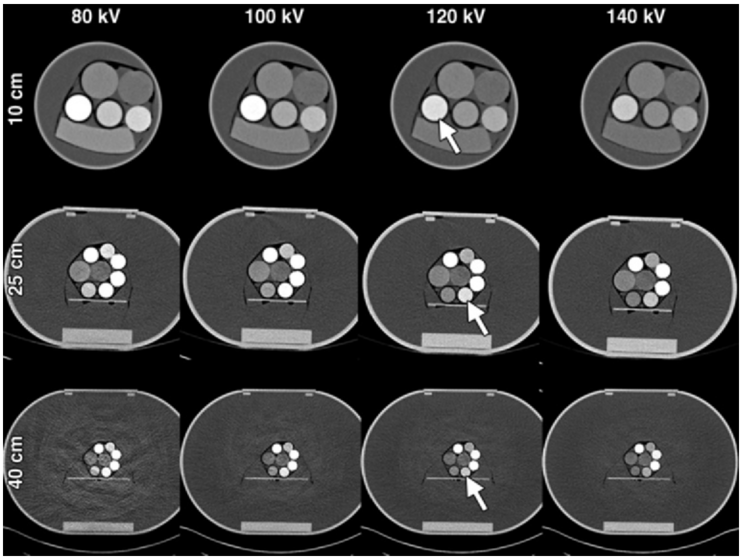
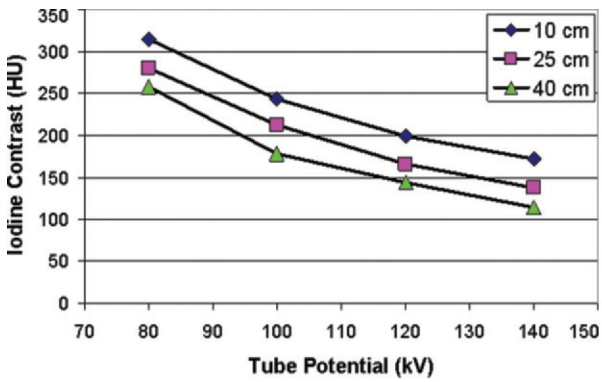


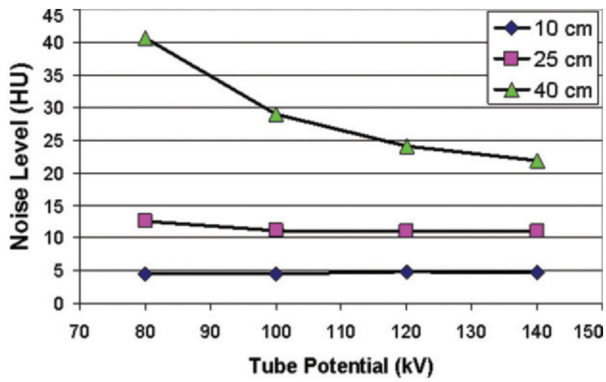
Figure 5. Comparison of CT images obtained at four different tube potentials. CT images obtained at 80 kV, 100 kV, 120 kV, and 140 kV show water phantoms with lateral widths of 10 cm, 25 cm, and 40 cm, which represent typical attenuation levels of a newborn, adolescent, and adult, respectively. For each phantom, the prescribed $CTDI_{vol}$ (10 cm, 3.9 mGy; 25 cm, 6.6 mGy; 40 cm, 21.4 mGy) was matched for all tube potentials. Several different types of contrast material—including iodine with a concentration of 6.9 mg/mL (arrows) in 120 kV images—were placed inside the water bath to allow the level of contrast enhancement to be measured. The contrast inserts in the 10-cm phantom are a subset of those used in the 25- and 40-cm phantoms.

tional to the cube of the x-ray photon energy, but also approximately proportional to the cube of the atomic number. Because iodine has a much higher atomic number (Z = 53) than water (effective Z = 7.4), the linear attenuation coefficient of iodine increases much more dramatically than that of water as the x-ray energy decreases. The CT number for iodine is the scaled (by 1000) relative difference of the linear attenuation coefficients of iodine and water at a given energy; thus, the contrast expressed in the CT number is higher at lower tube potentials.

In Figure 5, CT images of three water phantoms that were scanned with all available tube potentials are seen. The lateral widths of the three phantoms are 10 cm, 25 cm, and 40 cm, representing typical attenuation levels for a newborn, an adolescent, and an adult, respectively. The adult-sized phantom was included to provide a reference. For each phantom, the prescribed volume CT dose index ($CTDI_{vol}$) was matched for all tube potentials: at 10 cm, 3.9 mGy was used; at 25 cm, 6.6 mGy; and at 40 cm, 21.4 mGy. Several different types of contrast material, one of which contained iodine with a concentration of 6.9 mg/mL, were added to the water to allow measurement of contrast. In Figure 6a, the contrast of iodine is plotted as a function of tube potential. On average, the contrast of iodine at 80 kV was 70% and 100% higher than that at 120 kV and 140 kV, respectively, and the contrast of iodine at 100 kV was 25% and 50% higher than that at 120 kV and 140 kV, respectively. Image contrast also decreased as phantom size increased, a result of beam hardening effects.



a.



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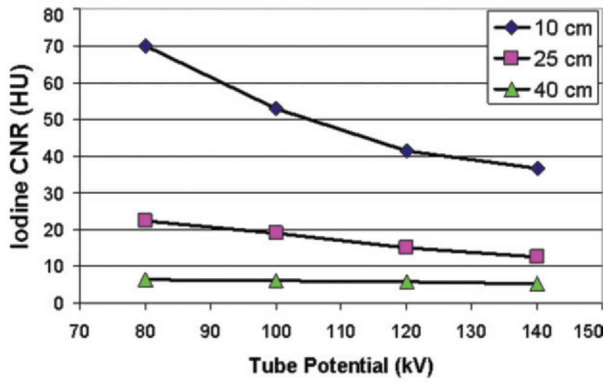


c.

Figure 6. Graphs show the changes in iodine contrast **(a)**, noise level **(b)**, and iodine contrast-to-noise ratio (CNR) **(c)**, which are plotted as a function of tube potential, for 10-, 25-, and 40-cm phantoms, which represent a newborn, adolescent, and adult, respectively. For each phantom, the scanner radiation output was kept constant for all tube potentials.

Noise

Another important factor that affects image quality is noise. In Figure 6b, noise level is plotted as a function of tube potential by using the same data that were used in Figure 6a. The $CTDI_{vol}$ was matched for all tube potentials. For the 10-cm phantom, the noise level was almost identical at each tube potential. For the 25-cm phantom, noise level slightly increased at 80 kV. For the 40-cm phantom (which represents a typical adult size), noise level was dramatically increased at lower tube potentials. In addition, significant photon-starvation artifacts were seen in the images of the adult-sized phantom obtained at 80 kV, a finding due to the decreased penetrating capability of the lower-energy photons (46).

Contrast-to-Noise Ratio

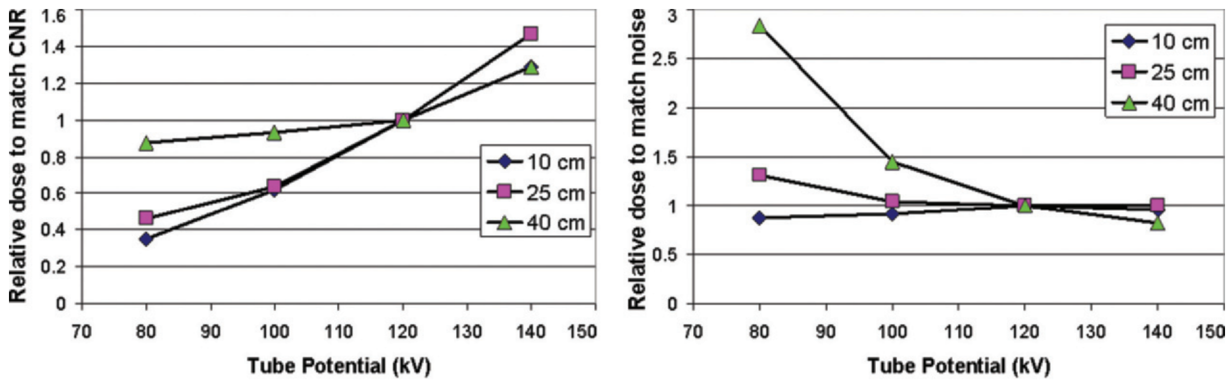
On the basis of the resultant contrast and noise levels, one can see that among pediatric patients, the use of low tube potential settings provides better iodine contrast without increasing image noise, given the same radiation dose at each tube potential (18). Iodine CNR typically is used to represent the combined effect of iodine contrast enhancement and image noise, both of which are important image quality metrics. In Figure 6c, iodine CNR is plotted as a function of tube potential for all three phantoms; the images of the 10- and 25-cm phantoms obtained at a lower tube potential had substantially improved CNR. A slight increase of iodine CNR for the 40-cm phantom was also seen, but this finding has no clinical value because of the increase in noise and image artifacts.

Reducing Radiation Dose When CNR Is to Be Matched

Because of the increased iodine CNR at lower tube potentials, radiation dose may be reduced to achieve similar or improved iodine CNR relative to the more commonly used 120 kV. In Figure 7a, the relative dose required at each tube potential to achieve the same iodine CNR as that obtained at 120 kV is seen. With the 10-cm phantom, the required dose at 80 kV is only 35% of the amount required at 120 kV, and at 100 kV, the required dose is 62% of the amount required at 120 kV. With the 25-cm phantom, the doses are 46% of the 120-kV dose at 80 kV and 63% of the 120-kV dose at 100 kV.

Teaching Point

Because the 25-cm phantom is near the upper limit for the size of a typical pediatric patient, it appears that radiation dose may be substantially reduced by using a lower tube potential if the



a. b.

Figure 7. **(a)** Graph shows the relative radiation dose required at each tube potential to obtain the same iodine CNR for all three phantoms. For the 10-cm phantom, 35% of the 120-kV dose is required at 80 kV to achieve the same iodine CNR as at 120 kV, and 62% of the 120-kV dose is required at 100 kV. For the 25-cm phantom, 46% of the 120-kV dose is required at 80 kV, and 63% of the 120-kV dose is required at 100 kV. **(b)** Graph shows the relative radiation dose required at each tube potential to obtain the same noise level for all three phantoms. For the 10-cm phantom, 88% of the 120-kV dose is required at 80 kV to achieve the same noise level as at 120 kV, and 92% of the 120-kV dose is required at 100 kV. For the 25-cm phantom, 129% of the 120-kV dose is required at 80 kV, and 101% of the 120-kV dose is required at 100 kV.

goal is to match CNR. However, this scenario may not apply to all clinical tasks. For example, when matching iodine CNR for each tube potential, the contrast at 80 kV for the 10-cm phantom is about 70% higher than it is at 120 kV. Therefore, if the iodine CNR were matched, the resulting noise at 80 kV would also be 70% higher than it is at 120 kV. For some diagnostic tasks such as evaluation of relatively large vessels at CT angiography, the increased contrast of iodine may sufficiently compensate for the dramatically increased noise. However, for characterization of organs or structures that do not demonstrate iodine uptake (ie, unenhanced imaging), there is less of a benefit, and reducing radiation dose by matching the CNR of iodine is not appropriate (32). To select the most dose-efficient tube potential for pediatric patients, noise must be considered independently of iodine CNR.

Reducing Radiation Dose When Noise Is to Be Matched

On the basis of the noise levels measured in images obtained with equivalent radiation doses, the relative dose that is required at each tube potential to achieve an equivalent noise level may be estimated. Figure 7b clearly demonstrates that if the noise level in an image obtained at 120 kV is to be matched, the potential for dose reduction at lower tube potentials is limited or nonexistent. For the 10-cm phantom, radiation dose is reduced by 12% at 80 kV and by 8% at 100 kV compared with the dose at 120 kV. For the 25-cm phantom, a 29% dose increase is required at 80 kV to match the noise level at 120 kV. As a reference, the 40-cm phantom (which represents adult-sized patients) required a 183% dose increase at 80 kV to compensate for the increased noise.

Reducing Radiation Dose When Both CNR and Noise Are Incorporated

Teaching Point

According to the results described in the previous section, selecting the most dose-efficient tube potential and estimating how much radiation dose is reduced in pediatric CT depend on the diagnostic task being performed. When the task involves evaluating only vessels or structures that demonstrate substantial iodine contrast material enhancement, the use of iodine CNR may be appropriate. If the diagnostic task involves evaluating structures that do not enhance, matching noise levels is more appropriate, and the ability to reduce radiation dose is limited at lower tube potentials. Many diagnostic tasks, such as routine contrast-enhanced abdominopelvic examinations, fall somewhere between these two scenarios. In these cases, a quality index that incorporates both iodine CNR and noise level is an attractive alternative for determining the most dose-efficient tube potential (47). A general strategy, in which a noise constraint is applied when matching the iodine CNR, has recently been developed (48). By adjusting the noise constraint parameter, the maximally increased noise level at a lower tube potential relative to that at a higher, reference tube potential may be adjusted on the basis of the image quality requirements of different diagnostic tasks.

Scanning Speed and Tube Current Limit

Scanning speed and tube current limit are two other important factors that must be considered before a lower tube potential is used in pediatric examinations. CT systems limit the tube current and, consequently, radiation dose that may be delivered to patients. Therefore, a tradeoff typically exists between the scanning speed and the maximum achievable radiation dose, and the use of a lower tube potential usually leads to tighter restriction of radiation dose output. For example, in pediatric body mode, the $CTDI_{vol}$ per 100 effective mAs (ie, tube current–rotation time product divided by pitch) on a dual-source 64-section scanner (Definition; Siemens Medical Solutions, Forchheim, Germany) is 1.47 mGy at 80 kV, 3.33 mGy at 100 kV, and 5.74 mGy at 120 kV. To achieve a given amount of radiation output, a decrease in helical pitch is often required, resulting in a slower scanning speed. For example, assume that a CT examination of a child who weighs 25 kg requires a minimum of 6 mGy to generate an acceptable noise level in images of the chest, abdomen, and pelvis. On a Definition CT scanner, the corresponding effective mAs is 408 at 80 kV, 180 at 100 kV, and 105 at 120 kV. For a 30-cm scan range and a rotation time of 0.33 seconds, the shortest options for scan time at 80 kV use the maximum tube current of 500 mA and a helical pitch of 0.4, yielding a scan time of 12.9 seconds. At 100 kV, a helical pitch of 0.9 may be used, resulting in a scan time of 5.7 seconds. At 120 kV, a pitch of 1.4 can be used, resulting in a scan time of only 3.7 seconds. If the patient's condition does not allow for a scan time of longer than 4 seconds, neither 80 kV nor 100 kV is an acceptable option, even though superior iodine contrast enhancement is achievable at these tube potentials.

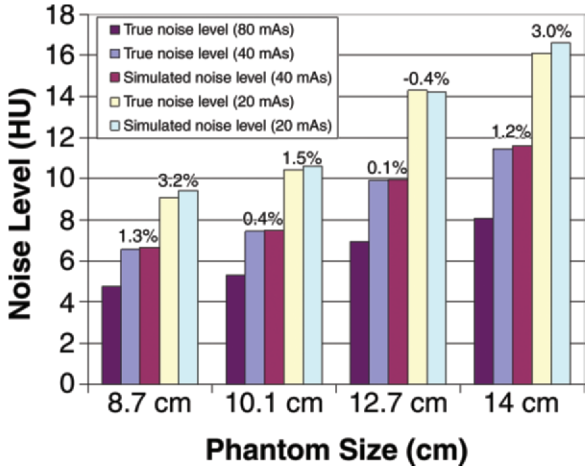


Figure 8. Graph shows the effectiveness of the noise insertion tool on four acrylic cylindrical phantoms (8.7 cm, 10.1 cm, 12.7 cm, and 14 cm). Two simulated dose levels (50% and 25% of the original doses) were included, and the noise levels in the simulated images were compared with those in the images acquired directly from the CT scanner.

Implementing Technique Charts for Tube Potential and Tube Current Settings

Selecting the optimal tube potential for pediatric patients is not a trivial matter and is specific to the imaging task. Ideally, a chart of appropriate tube potential and tube current values for the different patient size ranges would be constructed for each type of examination. One approach for the development of such a chart is to use different tube potentials for each patient size group and gradually decrease the tube current until the image quality approaches the minimum acceptable limit. This method requires that lower-dose images be obtained in a large number of patients, a process that is tedious and that may lead to images that are diagnostically compromised. A more elegant approach is to use a noise insertion tool to simulate images obtained with reduced dose levels by using existing examinations performed with "standard dose" levels. A range of simulated dose levels may be generated, and the diagnostic quality comparisons may be made across images of a single patient by removing patient-specific variables.

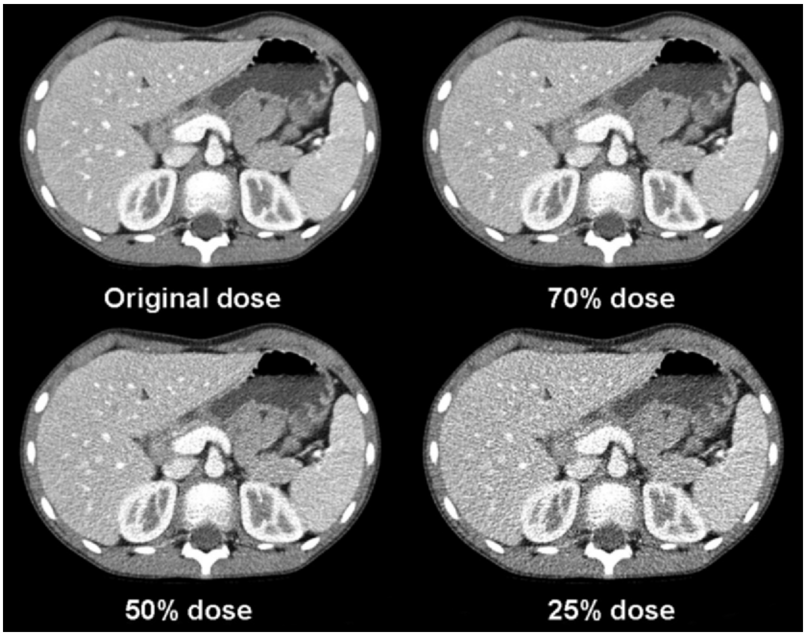


Figure 9. Comparison of simulated lower-dose images obtained from a routine-dose CT examination. CT images obtained with the original dose, 70% of the original dose, 50% of the original dose, and 25% of the original dose were compared to determine the lowest acceptable dose level for each type of CT examination.

This method enables radiologists to determine the lowest acceptable dose level without compromising the diagnostic quality of a CT examination and has been widely used for optimizing CT scanning protocols (49–53).

The following sections describe how a technique chart for tube potential and tube current settings was developed for pediatric chest and abdominopelvic examinations by using a noise insertion tool. A typical 120-kV scanning protocol was used as a starting point, and the following three steps were employed to generate an optimized weight-based chart.

Step 1: Determine the Lowest Acceptable Dose Level at 120 kV

In our original protocols for pediatric chest and abdominopelvic examinations, we used 120 kV with a quality reference mAs of 60 mAs for chest imaging and 70 mAs for abdominopelvic imaging. The AEC software (CareDose 4D; Siemens Medical Solutions, Forchheim, Germany), which modulates tube current on the basis of patient size, was enabled. To determine the lowest acceptable dose level, we simulated images by using 25%, 50%, 70%, and 90% of the original dose level used for 12 pediatric CT examinations with the 120-kV protocols. The simulation tool was used to insert noise into the raw CT data and create lower-dose images on the basis of a realistic noise model (54). The simulation tool was validated with a series of cylindrical acrylic phantoms before clinical data were used. As seen in Figure 8, the differences (recorded as percentages) between the simulated noise level and those of the actual low-dose images were lower than 3.2% for all tested phantom sizes. Figure 9 shows one example of simulated lower-dose images obtained from a CT examination in which the standard dose was administered. Two pediatric radiologists evaluated the quality of these images to determine the lowest acceptable dose for each type of examination and determined that both chest and abdominopelvic images obtained with 70% of the original dose were acceptable. Therefore, we reduced the quality reference mAs from 60 mAs to 40 mAs in the chest protocol and from 70 mAs to 50 mAs in the abdominopelvic protocol.

Table 1
Weight-based Technique Chart for Tube Potential and Tube Current Settings for Routine Pediatric Chest CT Examinations

Weight (kg)	Tube Potential (kV)	QRM*	AEC†	Pitch	Rotation Time (sec)	Kernel	Section/ Interval Thickness (mm)	Collimation (mm)	$CTDI_{vol}$ (mGy)‡
<10	80	150	On	1.2	0.33	B40f	3/3	64 × 0.6	2.1 ± 0.2
10–20	100	70	On	1.4	0.33	B40f	3/3	64 × 0.6	3.5 ± 0.3
20–45	120	40	On	1.4	0.33	B40f	3/3	64 × 0.6	5.2 ± 1.2

*QRM = Quality reference mAs, a named term on a Siemens scanner.
†We used CareDose 4D (Siemens).
‡$CTDI_{vol}$ values are based on a 32-cm CTDI phantom (55) and are presented as mean ± standard deviation.

Table 2
Weight-based Technique Chart for Tube Potential and Tube Current Settings for Routine Pediatric Abdominopelvic CT Examinations

Weight (kg)	Tube Potential (kV)	QRM*	AEC†	Pitch	Rotation Time (sec)	Kernel	Section/ Interval Thickness (mm)	Collimation (mm)	$CTDI_{vol}$ (mGy)‡
<10	80	190	On	1.1	0.33	B40f	3/3	64 × 0.6	2.2 ± 0.3
10–20	100	90	On	1.4	0.33	B40f	3/3	64 × 0.6	3.8 ± 0.4
20–45	120	50	On	1.4	0.33	B40f	3/3	64 × 0.6	5.1 ± 0.7

*QRM = Quality reference mAs, a named term on a Siemens scanner.
†We used CareDose 4D (Siemens).
‡$CTDI_{vol}$ values are based on a 32-cm CTDI phantom (55) and are presented as mean ± standard deviation.

Step 2: Create a Weight-based Technique Chart for Tube Potential and Tube Current Settings

The second step is to create a weight-based chart to establish a noise-matched technique at a lower tube potential. The lowest acceptable tube current values at 120 kV—which were determined on the basis of our phantom measurements in step 1—were converted to noise-matched tube current values at 80 kV and 100 kV. For the abdominopelvic examination, the corresponding quality reference mAs values were 190 mAs at 80 kV and 90 mAs at 100 kV. For the chest protocol, the corresponding quality reference mAs values were 150 mAs at 80 kV and 70 mAs at 100 kV. These values were determined by scanning phantoms with a variety of sizes and weights at each tube potential and determining which tube current values yielded the same noise values at lower tube potentials as at 120 kV.

As was previously described, scanning speed and CT system limits must be considered when determining whether the use of a lower tube potential is appropriate. To deliver a sufficient radiation dose, scanning speed may need to be compromised and, thus, may not be acceptable. After determining maximum acceptable scanning times, two weight-based technique charts—one for a routine chest examination and the other for a routine abdominopelvic examination—were empirically developed (Tables 1, 2).

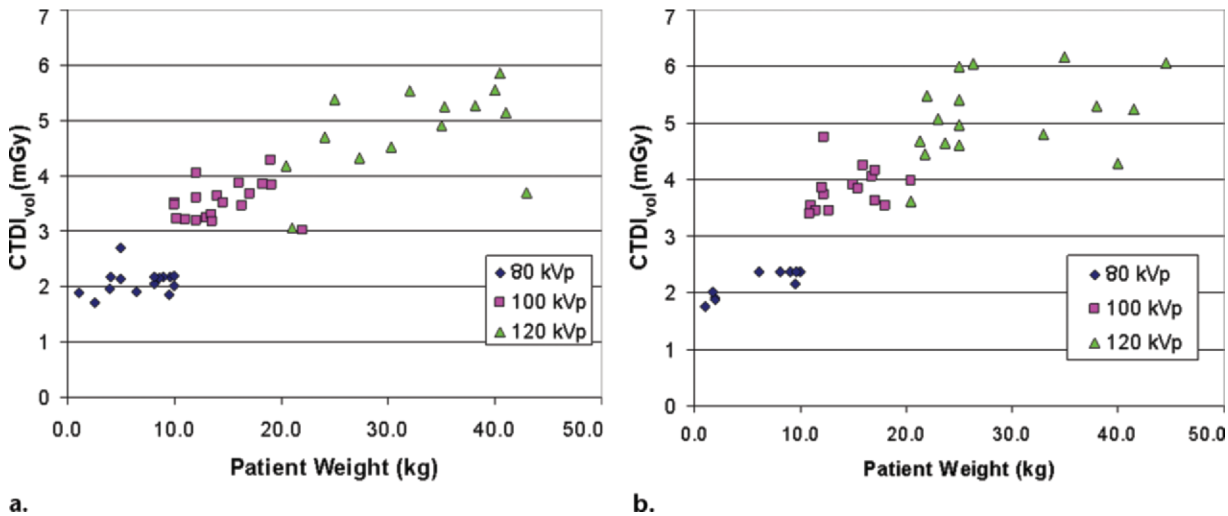


Figure 10. Graph shows the distribution of $CTDI_{vol}$ as a function of patient weight on the basis of the technique charts for pediatric chest **(a)** and abdominopelvic **(b)** CT examinations. The $CTDI_{vol}$ value is based on a 32-cm CTDI phantom.

Step 3: Refine the Techniques at 80 kV and 100 kV

The technique charts created in step 2 were developed on the basis of a noise-matching scheme. The benefit of using a lower tube potential is that it provides better contrast enhancement of iodine, which may allow radiation dose to be further reduced. In this step, the goal is to evaluate if and how much the radiation dose may be further reduced by using a lower tube potential.

Thirty-three pediatric body CT examinations were performed with the technique charts described in step 2: *(a)* 17 examinations were performed at 80 kV, and 16 were performed at 100 kV; *(b)* 12 examinations included only chest images, 18 included only abdominopelvic images, and three included both chest and abdominopelvic images; and *(c)* 27 examinations were obtained with contrast enhancement, and six were obtained without. Similar to step 1, we simulated images obtained with 25%, 50%, 70%, and 90% of the original radiation dose level. Two pediatric radiologists evaluated the quality of these images to determine the lowest acceptable dose levels for images obtained with 80 kV or 100 kV. Results of this analysis indicated that, for the protocol developed in step 2, the lowest acceptable dose levels for images obtained with 80 kV are 90% for chest CT and 100% for abdominopelvic CT, and for images obtained with 100 kV, the lowest acceptable dose levels are 70% for chest CT and 100% for abdominopelvic CT. Thus, it is not possible to achieve further dose reduction for abdominopelvic examinations, and only marginal dose reduction is possible for chest examinations.

On the basis of these results, we determined that the use of a lower tube potential (80 kV or 100 kV) does not enable substantial reduction of radiation dose, because the pediatric radiologists required that image noise be essentially matched, even when a lower tube potential was used. Thus, most of the dose reduction was achieved by optimizing the original 120-kV protocols.

Applications of the Optimized Technique Chart

By using the technique chart that was developed according to the previously described steps, we collected 89 cases from July 31, 2008 to September 22, 2009, including 31 cases that were imaged at 120 kV (14 chest and 17 abdominopelvic), 33 cases that were imaged at 100 kV (18 chest and 15 abdominopelvic), and 25 cases that were imaged at 80 kV (15 chest and 10 abdominopelvic). In all 89 cases, CT images were considered to be diagnostic. The distribution of $CTDI_{vol}$ as a function of patient weight is shown in Figure 10.

Important Considerations and Common Pitfalls

At lower tube potentials, the tube current must be appropriately adjusted relative to that used at a higher, reference tube potential (eg, 120 kV) to avoid excessive noise; a common pitfall is to keep the tube current the same and hope for the best. Because the $CTDI_{vol}$ per 100 effective mAs is substantially lower at a lower tube potential than it is at 120 kV, keeping the effective mAs the same yields a much lower radiation output at lower tube potentials and, thus, leads to substantially higher noise levels, which may not be clinically acceptable. Therefore, the tube current used at lower tube potentials should be appropriately increased according to the diagnostic task being performed; use of the same tube current as that used at 120 kV likely will compromise image quality.

Teaching Point

To reduce motion artifacts, the use of a high scanning speed is desirable in pediatric CT examinations. High scanning speed usually involves the use of a fast rotation time and a high helical pitch. However, because of the limitations of tube current inherent in CT scanners, the maximum achievable dose level (which is determined by the maximum effective mAs) may be limited, especially with the use of lower tube potential settings. Therefore, in larger children, the use of a higher tube potential may still be necessary to achieve the desired dose level and high scanning speed. It is essential to use a weight- or size-based technique chart to fully realize the benefit of a lower tube potential without substantially compromising noise and scanning speed. Well-defined guidelines regarding tube potential and tube current values for any given patient size should be available to the technologist.

Reducing dose at lower tube potentials is also highly dependent on the diagnostic task, as well as patient size. On the basis of the iodine CNR, one may conclude that the use of a lower tube potential will substantially reduce radiation dose, as is seen in Figures 6c and 7a. However, in larger patients and for some diagnostic tasks, higher noise may not be tolerated, reducing or eliminating the advantages of the use of a lower tube potential. The use of a lower tube potential must be carefully evaluated for each diagnostic task and patient size.

A constant target noise level across all patient sizes usually is not considered acceptable by radiologists. Children require a lower target noise level than adults because children have less adipose tissue between organs and tissue interface (44,56). In addition, the use of thinner sections typically is necessary to visualize smaller structures in children. A target noise level that is appropriate for adults may be too high for children. For example, a noise level of 12 HU in the liver is adequate for routine abdominal imaging in an adult, but it is unacceptably high in a 6-year-old child. This principle is true for any tube potential used in pediatric CT.

Finally, dense materials such as highly concentrated iodine contrast material and metal may introduce substantial beam hardening and streaking artifacts, which are often more severe at lower tube potentials. The more severe artifacts seen at lower tube potentials occur for two reasons. Beam hardening correction methods currently implemented in CT scanners do not take into account beam hardening that is caused by contrast material or metal. Second, because of the higher attenuation of dense materials at lower tube potentials, other data nonidealities such as scattering, partial volume, and electronic noise are more severe at lower tube potentials than at high tube potentials. Therefore, if the area to be scanned includes dense materials or the contrast of the object is too high, the image quality at lower tube potentials may be inferior to that at higher tube potentials.

Conclusions

The appropriateness of a selected tube potential and how much to reduce radiation dose depend on the patient's size and the diagnostic task being performed. They are also affected by the radiation output limits of the CT scanner and the desired scanning speed. The use of a lower tube potential should be carefully evaluated for each type of examination to achieve an optimal tradeoff among contrast, noise, artifacts, and scanning speed.

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Optimal Tube Potential for Radiation Dose Reduction in Pediatric CT: Principles, Clinical Implementations, and Pitfalls

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Currently, it is common practice in the CT community to adapt the dose level to the patient's size, which is also a requirement for American College of Radiology (ACR) accreditation of pediatric CT facilities (39,40). Scanning techniques that depend on patient size include one or more of the following elements: size-dependent beam-shaping filters, manual tube current technique charts, AEC, and optimal tube potential (41).

Page 838
The increase in iodine attenuation at lower tube potentials provides more iodine signal and improves the conspicuity of hyper- or hypovascular structures (33,34).

Page 840
Because of the increased iodine CNR at lower tube potentials, radiation dose may be reduced to achieve similar or improved iodine CNR relative to the more commonly used 120 kV.

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According to the results described in the previous section, selecting the most dose-efficient tube potential and estimating how much radiation dose is reduced in pediatric CT depend on the diagnostic task being performed. When the task involves evaluating only vessels or structures that demonstrate substantial iodine contrast material enhancement, the use of iodine CNR may be appropriate. If the diagnostic task involves evaluating structures that do not enhance, matching noise levels is more appropriate, and the ability to reduce radiation dose is limited at lower tube potentials.

Page 846
To reduce motion artifacts, the use of a high scanning speed is desirable in pediatric CT examinations. High scanning speed usually involves the use of a fast rotation time and a high helical pitch. However, because of the limitations of tube current inherent in CT scanners, the maximum achievable dose level (which is determined by the maximum effective mAs) may be limited, especially with the use of lower tube potential settings. Therefore, in larger children, the use of a higher tube potential may still be necessary to achieve the desired dose level and high scanning speed.