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Management of patient dose and image noise in routine pediatric CT abdominal examinations

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Abstract The aim was to propose a strategy for finding reasonable compromises between image noise and dose as a function of patient weight. Weighted CT dose index ($CTDI_w$) was measured on a multidetector-row CT unit using CTDI test objects of 16, 24 and 32 cm in diameter at 80, 100, 120 and 140 kV. These test objects were then scanned in helical mode using a wide range of tube currents and voltages with a reconstructed slice thickness of 5 mm. For each set of acquisition parameter image noise was measured and the Rose model observer was used to test two strategies for proposing a reasonable compromise between dose and low-contrast detection performance: (1) the use of a unique noise level for all test object diameters, and (2) the use of a unique dose efficacy level defined as the noise reduction per unit dose. Published data were used to define four weight classes and an acquisition protocol was proposed for each class. The protocols have been applied in clinical routine for more than one year. $CTDI_{vol}$ values of 6.7, 9.4, 15.9 and 24.5 mGy were proposed for the following weight classes: 2.5–5, 5–15, 15–30 and 30–50 kg with image noise levels in the range of 10–15 HU. The proposed method allows patient dose and image noise to be controlled in such a way that dose reduction does not impair the detection of low-contrast lesions. The proposed values correspond to high-quality images and can be reduced if only high-contrast organs are assessed.

Keywords Pediatric CT · Dose reduction · Dosimetry · Image quality · Optimization

Introduction

CT examinations represent about 2–6% of all radiological examinations performed on adults, but contribute 27–40% of the effective dose delivered in radiology [1, 2, 3, 4]. In this context, several strategies have been adopted by manufacturers to reduce patient dose [5 , 6, 7, 8], and the many papers devoted to CT dose assessment have increased radiologists' awareness of the high dose level associated with CT examinations [9, 10, 11, 12]. The use of pediatric CT is increasing quite rapidly, probably more rapidly than for adult CT, even if not used as a first-line imaging modality in many cases. It is an important diagnostic tool that certainly improves health care. However, the technical parameters chosen for an examination must be such that radiation risks are maintained as low as reasonably achievable for the diagnostic intent. For a given set of machine parameters (voltage, tube current–time product, pitch, etc.), it has been shown that effective doses for a small infant are larger than for an adult [13, 14, 15]. As a result, pediatric CT acquisition protocols need to be carefully controlled.

Several approaches have been used to propose image acquisition parameters that give a reasonable compromise between image quality and patient dose, some based on the assessment of clinical data [15, 16, 17, 18] and others on simulated data [19, 20, 21]. However, while these studies show that it is possible to reduce the

dose for pediatric CT, there is still a need to define objectively levels of image quality adequacy.

The goal of this paper is to propose a general method for setting dose levels for abdominal pediatric CT acquisitions using CTDI test objects. We consider the CT unit from a dose point of view, and also in terms of the adequacy of the acquisition parameters for detecting low-contrast lesions using a mathematical model observer.

Materials and methods

Dose characterization of the unit

The Multi-detector row CT (MDCT) used for this study was a four-row LightSpeed QX/I (GE Medical Systems, Milwaukee, Wis). Acquisitions were performed with a detector collimation of 2.5 mm, acquiring four slices per 360° tube rotation. For each tube voltage (kV), the normalized weighted CTDI, $_{n}CTDI_{w}$, (mGy mAs^{-1}), was measured using the standard PMMA CTDI test objects of 16 and 32 cm in diameter [22]. Measurements of $_{n}CTDI_{w}$ were also made in air at the center of rotation of the gantry and in a PMMA CTDI test object of 24 cm in diameter designed for this study. Doses delivered during helical acquisitions were expressed in term of volume CTDI: $CTDI_{vol}=CTDI_{w}/pitch$, where pitch is defined as the table distance traveled in one rotation of the X-ray source divided by the total collimated width of the X-ray beam [22, 23]. A subscript was added to the standard notation to indicate the test object diameter in which the measurement was expressed (e.g., a $CTDI_{vol}$ measured in a CTDI test object of 16 cm diameter is quoted as $CTDI_{vol,16}$).

Image noise characterization

The three CTDI test objects were imaged in helical mode with a pitch of 0.75, detector collimation of 2.5 mm (i.e., 4×2.5 mm), reconstructed slice thickness of 5 mm, and at tube voltages of 80, 100, 120 and 140 kV for a wide range of tube currents (i.e., 40–400 mA in steps of 40 mA) using a 360° tube rotation time of 0.8 s. The detector collimation and reconstructed slice thickness chosen correspond to the acquisition conditions used for routine pediatric abdominal examinations in our center. For each acquisition, the reconstructed field of view was set equal to 4 cm larger than the diameter of the imaged CTDI test object. To assess image noise, the standard deviation of CT numbers was measured in a 300-pixel region of interest (ROI) at five locations of the test object (one measurement at the center and four measurements at 1 cm from the periphery of the test object) and an average standard deviation, σ, was calculated. Measurements were performed on one image for each combination of test object and $CTDI_{vol}$ values using the software available on the CT unit.

Compromise between noise and image noise

One possible way to set reasonable dose levels is to choose an image noise level and keep it constant whatever the abdominal diameter of the patient. Another possibility is to look at the rate at which noise decreases as dose increases, as a function of the abdominal diameter of the patient. From the statistics properties of quantum noise (i.e., image noise inversely proportional to the square root of dose) it appears that at some point as dose increases we start to lose in the strides we make in noise reduction. Thus, there is a dose level above which noise reduction with dose increase is no longer justified. To determine the adequacy of the image acquisition protocol, minimum image quality requirements were set by specifying the size and contrast of the structure to be detected. Only acquisition protocols allowing the detection of a low-contrast structure (i.e., having a contrast with background equal to 10 HU whatever the tube voltage used) smaller than or equal to 10 mm were accepted. This choice leads to comparable low-contrast detection requirements as the ones tested, for example, when using standard CT test objects such as the Catphan 500 (The Phantom Laboratory, Cambridge, NY) and thus assures a good level of image quality. To establish whether the acquisition protocols fulfilled the low-contrast detection constraint proposed, the Rose model observer was applied on each combination of test object and $CTDI_{vol}$ values. This model is described by the following relationship: $SNR = CNR \times (N)^{1/2}$, where CNR is the contrast-to-noise ratio (i.e., 10 HU divided by averaged standard deviation, σ) and N the number of pixels of the low-contrast structure to be detected. For structure diameter d and pixel size dimension Δ, N is given by; $(\pi/4) \times (d/\Delta)^2$ [24] (see Table 2). A SNR value of 5.0 was systematically adopted when using this model observer since it corresponds to a situation where detection is performed with a high confidence level.

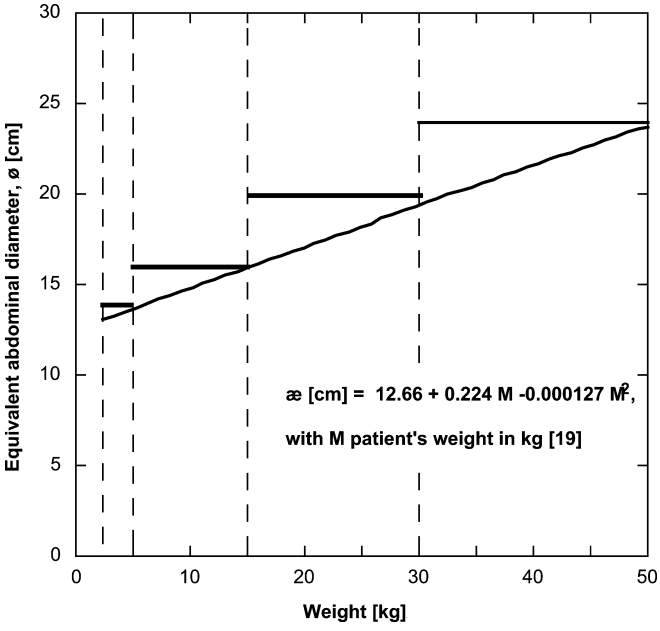

Fig. 1 Equivalent abdominal diameter as a function of patient weight according to Huda et al. [19]

Link between experimental results and clinical applications

To relate the experimental results to clinical applications, patient weights were converted to equivalent diameter of the abdominal section using the relationship proposed by Huda et al. [19] illustrated in Fig. 1. To reduce the number of acquisition protocols, the pediatric population was split into four weight classes: 2.5–5, 5–20, 20–35, and 35–50 kg, giving the following equivalent diameters: 14, 16, 20, and 24 cm.

Clinical application

The resulting protocols were applied in routine use for one year (507 standard abdominal examinations). For ethical reasons, ac-

Table 1 Fit parameters of the relationships between the $_nCTDI_w$ parameter and the diameter of the CTDI test object used to perform the measurements: $_nCTDI_w = a\ test\ object\ Ø + b$

Tube voltage (kV)	a (mGy mAs^{-1} cm^{-1})	b (mGy mAs^{-1})	Regression coefficient
80	–0.00223	0.103	0.998
100	–0.00365	0.182	0.999
120	–0.00555	0.282	0.999
140	–0.00789	0.401	0.999

quisition protocols while varying dose levels were not tested. The use of the model observer was considered sufficient to assure that image noise were sufficiently low to provide images of good diagnostic quality. All cases were reviewed and evaluated with consensus by three senior radiologists (FG, DL, PS). The adequacy of image quality was determined by asking the radiologists to assess the perceived level of mottle in the CT images and to assign each examination to one of two categories:

– Inadequate for diagnosis
– Diagnostically adequate

Results

For each tube voltage, a linear relationship could be established between the $_nCTDI_w$ and the test object diameter with a regression coefficient higher than 0.990 (see Table 1). The measured $CTDI_{vol}$ with the test objects of 16 and 32 cm in diameter were systematically in good agreement with the values indicated by the unit (differences within ±10%).

For each image acquisition condition, the mean CT number measured in each of the four ROIs was within the averaged standard deviation, σ. The relationship between the averaged standard deviation values, σ, and the volume CTDI, $CTDI_{vol,Ø}$, for the three test objects are presented in Fig. 2. These data shows that a low noise level can hardly be obtained when imaging large test objects. Figure 3 shows the derivatives, E_f, with respect to dose, of the data presented in Fig. 2. The parameter E_f represents the noise reduction per mGy as a function of the $CTDI_{vol}$ and is a mean to evaluated the dose efficacy in reducing image noise. This graph shows that in the low dose range a small increase in dose allows a drastic noise reduction, whereas in the high dose range dose efficacy becomes very low.

The diameter of a structure with a contrast of 10 HU with background which can be detected with a high confidence level according to the Rose model used in this study was then calculated. The results obtained for each test object are shown in Fig. 4. For each test object the dose efficacy levels (data presented in Fig. 3) of 0.25, 0.5 and 1.0 have been added on the graph. As expected, large differences in dose are needed to maintain the low-contrast detection performance as the test object diameter is varied from 16 cm to 32 cm.

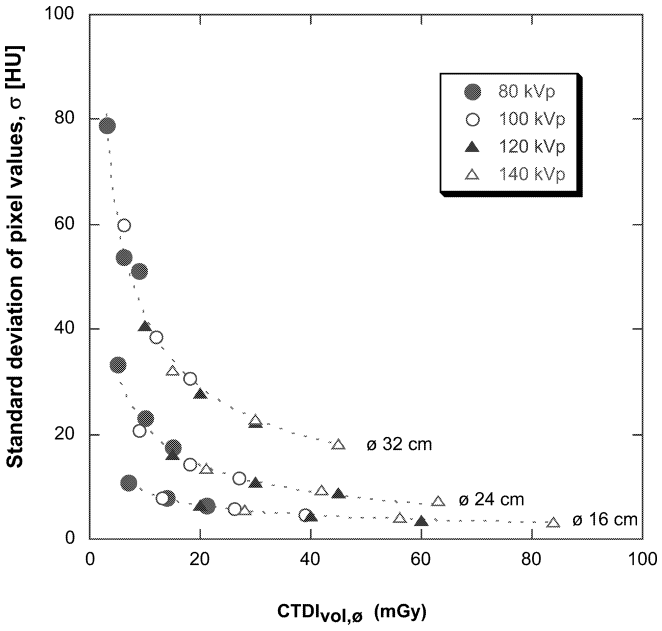

Fig. 2 Averaged standard deviation (i.e., image noise) variations as a function of $CTDI_{vol}$ in test objects of 16, 24 and 32 cm diameter

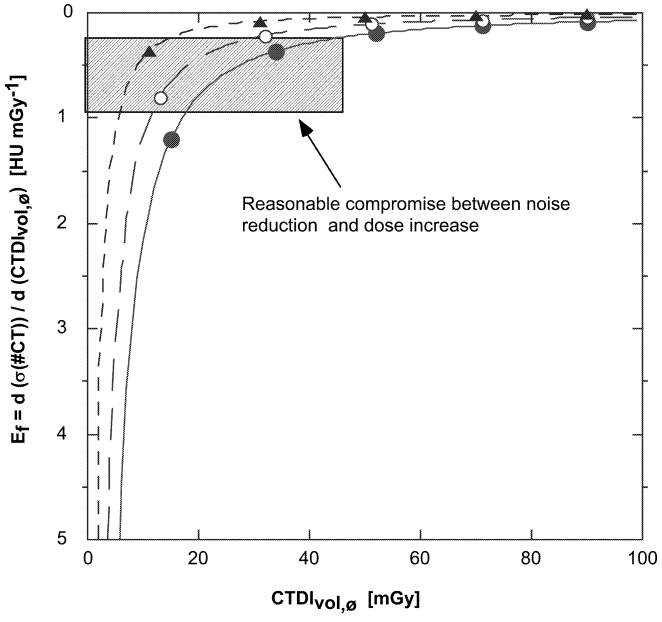

Fig. 3 Dose efficacy, E_f, as a function of $CTDI_{vol,Ø}$ for 16, 24 and 32 cm diameter test objects. As expected there exists a range of $CTDI_{vol,Ø}$ values where a modest increase in dose drastically reduces image noise. It can also be shown also that in the high $CTDI_{vol,Ø}$ range, the dose efficacy becomes very low. The hatch-marked region represents the position along the noise versus dose curve where a reasonable compromise between image quality and patient dose is expected

Table 2 Smallest diameters, *d*, of a structure having a contrast with background of 10 HU, detectable according to the Rose criteria as a function of the dose efficacy E_f

E_f (HU mGy^{-1})	Test object ∅ (cm)	$CTDI_{vol,\varnothing}$ (mGy)	σ (HU)	Δ (mm)	d (mm)
1.0	16	5.9	11.8	0.391	2.6
1.0	24	11.7	19.8	0.547	6.1
1.0	32	17.5	31.2	0.703	12.4
0.5	16	9.4	9.4	0.391	2.1
0.5	24	18.5	15.0	0.547	4.6
0.5	32	27.8	24.0	0.703	9.5
0.25	16	14.8	7.5	0.391	1.7
0.25	24	29.4	11.2	0.547	3.5
0.25	32	44.2	18.2	0.703	7.2

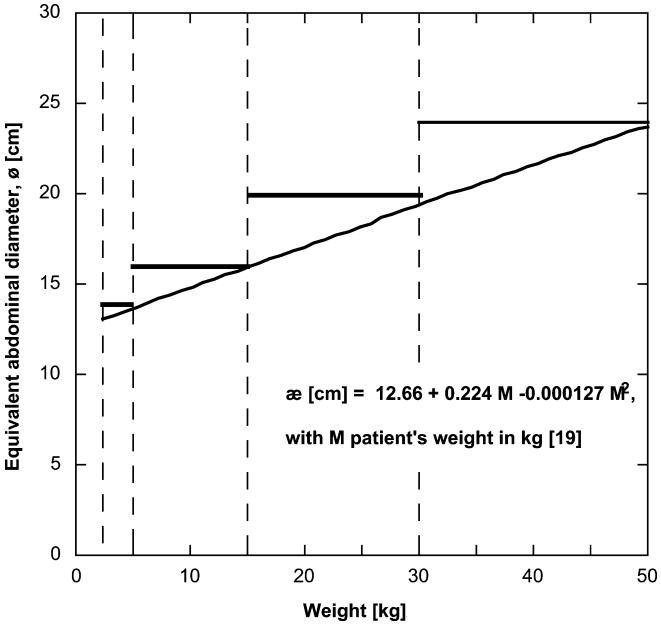

Fig. 4 Diameter of the smallest structure, *d* (having a contrast with a background of 10 HU), detectable according to the Rose criteria as a function of $CTDI_{vol}$ for the three test objects involved in the study. The diameter, *d*, of the structure detectable with a high confidence level is also indicated for each test object at dose efficacy levels: E_f=1.0, 0.5 and 0.25

First option: use of a fixed image noise level

Using the data from Fig. 4 it is possible to choose an image quality level in term of low-contrast detection and to keep it constant, whatever the size of the CTDI test object. For example, the detection of a 10-mm structure requires a $CTDI_{vol,16}$ of 0.4 mGy, a $CTDI_{vol,24}$ of 4.9 mGy, and a $CTDI_{vol,32}$ of 25.5 mGy in 16-, 24- and 32-cm-diameter test objects, respectively. This represents a factor of almost 64 in dose between the 16- and 32-cm-diameter test objects. If the size of the structure to be detected is decreased by a factor of 2 (i.e., *d*=5 mm), the dose is increased by factors in the range of 3.1–4.0 ($CTDI_{vol,16}$ of 1.6 mGy, $CTDI_{vol,24}$ of 16.3 mGy and $CTDI_{vol,32}$ of 79.5 mGy). Thus, significant increases in dose are required with increasing diameter if a constant low-contrast detection performance requirement is applied to all test object diameters. Consequently, this strategy seems inadequate when dealing with large variations in test object or patient diameter.

Second option: use of a fixed dose efficacy

An alternative method for finding a reasonable compromise between patient dose and image noise is to choose a dose efficacy level, E_f, and to estimate the $CTDI_{vol}$ that corresponds to that level. From Figs. 3 and 4 it appears that an E_f value in the range of 0.25–1.0 HU mGy^{-1} offers a good compromise between low-contrast detection and dose. The standard deviation and size of the smallest lesion detectable in this range of dose efficacy are summarized in Table 2. From these data it can be seen that a 10-mm low-contrast structure can be detected in all of the test objects with a dose efficacy value of 0.5 HU mGy^{-1}. This value was chosen to define the acquisition protocols to be used in routine clinical practice.

Clinical use

Using the diameter of children as a function of age and making interpolation of the $CTDI_{vol,\varnothing}$ and image noise, σ, with a dose efficacy of 0.5, the following $CTDI_{vol,\varnothing}$ values to be applied in clinical routine were found:

- 2.5–5 kg→equivalent ∅ of 14 cm→$CTDI_{vol,14}$= 7.1 mGy (σ=8.6 HU)
- 5–20 kg→equivalent ∅ of 16 cm→$CTDI_{vol,16}$= 9.4 mGy (σ=9.4 HU)
- 20 and 35 kg→equivalent ∅ of 20 cm→$CTDI_{vol,20}$= 14.0 mGy (σ=11.5 HU)
- 35 and 50 kg→equivalent ∅ of 24 cm→$CTDI_{vol,24}$= 18.6 mGy (σ=15.0 HU)

From the data in Table 1, these $CTDI_{vol,\varnothing}$ values can be converted into a $CTDI_{vol}$ measured in a standard CTDI test object (16 or 32 cm) For each weight class,

Table 3 CT acquisition parameters for routine abdominal CT as a function of patient's weight

Parameter	2.5–5 kg	5–15 kg	15–30 kg	30–50 kg
360° gantry rotation time (s)	0.8	0.8	0.8	0.8
Pitch	0.75	0.75	0.75	0.75
Detector collimation (mm)	2.5	2.5	2.5	2.5
Reconstructed slice thickness (mm)	5	5	5	5
Proposed $CTDI_{vol,ø}$ (mGy)	$CTDI_{w,ø14}$=7.1	$CTDI_{w,ø16}$=9.4	$CTDI_{w,ø20}$=14.0	$CTDI_{w,24}$=18.6
Corresponding $CTDI_{vol,16}$ (mGy)	6.7	9.4	15.9	24.5
Calculated (kV/mA)	120/32	120/45	120/78	120/120
Proposed (kV/mA)	80/90	100/70	120/80	120/120

Table 4 Comparison of the $CTDI_{vol}$ values obtained in this study with data available in the literature

Proposed $CTDI_{vol}$ values	2.5–5 kg	5–15 kg	15–30 kg	30–50 kg
$CTDI_{vol}$ (mGy) (this study)	6.7	9.4	15.9	24.5
Ref $CTDI_w$ (mGy) [15][a]	17	20–23	23–28	34–41
Ref $CTDI_w$ (mGy) [25]	20	20–25	25–30	–
Age classes	<2 years	2–6 years	6–14 years	14–18 years
Ref $CTDI_w$ (mGy) [27]	5.6	12	14	23.5

[a] The $_{eff}CTDI_{w,16}$ was calculated using a $_nCTDI_{w,16}$ of 0.277 mGy mAs^{-1} (value measured at 140 kV on our unit), a pitch factor of 0.75 and a 360° gantry rotation time of 0.8 s.

$CTDI_{vol,16}$ considered to give acceptable compromises between dose and image noise are given in Table 3 together with tube voltage and tube current possible combinations. From the data in Table 4 it can be observed that the $CTDI_{vol}$ for routine examinations using the proposed protocols is one half to one third of some of the values reported in the literature [25, 26] but are in good agreement with values proposed by Greess et al. [27].

None of the three radiologists involved in this study considered that the amount of quantum mottle present in the images necessitated an increase in patient dose. On the contrary, images were considered of very good quality. Thus, dose levels especially for small children can be further reduced especially when the detection of low-contrast lesions is not of primary importance. Of the 507 examinations performed, only seven were considered inadequate for diagnosis. All of the rejected images had image artifacts due to metallic implants. The remaining 500 examinations were considered adequate for diagnosis.

Discussion

According to the results published recently by Cohnen et al. [28], an excellent correlation exists between effective dose and CTDI measurements. Thus results presented in Fig. 2 clearly demonstrate that a low level of image noise cannot be kept constant in a wide range of patient weight (or abdominal section diameter) without increasing significantly patient dose. The use of the low- contrast detection constraint chosen in this paper led to $CTDI_{vol}$ values ranging from 0.4 mGy for a 16 cm object to 25.5 mGy for a 32 cm object. If a $CTDI_{vol}$ of 25.5 mGy for adults seems realistic when compared to the reference dose level of 35 mGy proposed for abdominal examinations by the CEC guidelines [29], a $CTDI_{vol}$ of 0.4 mGy for children appears unrealistic and could lead to the use of the unit in a region where it is no longer quantum noise limited. For our CT unit this would require a tube voltage of 80 kV, a tube current of 18 mA, a 360° tube rotation time of 0.5 s and a pitch of 1.5. The reduction of the size of the structure to be detected by a factor of two (i.e., d=5 mm) would require $CTDI_{vol}$ values of 1.6 and 79.5 mGy for 16- and a 32-cm test objects, respectively. While the $CTDI_{vol}$ obtained for the 16-cm object seems more realistic in term of machine parameters (80 kV, 72 mA, 0.5 s and a pitch of 1.5), the parameters obtained for the 32-cm object are certainly inadequate. For our CT unit it would require a tube voltage of 140 kV, a tube current of 500 mA, a 360° tube rotation time of 0.8 s and a pitch of 0.75, delivering an effective dose of about 40 mSv for a scan length of 30 cm [29]. This clearly demonstrates that in order to maintain a reasonable patient dose, we must accept working with various noise levels, resulting in a situation where low-contrast detection will depend on patient size.

If we accept that images of large patients cannot offer the same level of low-contrast detection as images of thin patients, the problem remains on how to propose acceptable compromises. In this study, the dose efficacy concept was introduced and used at a level of 0.5 HU mGy^{-1} on the basis of results obtained from the Rose model observer. It appears that for the unit and slice thickness investigated this dose efficacy leads to an image noise level of about 10 HU for children weighing

up to 30 kg and an image noise level of 15 HU for children in the weight class 30–50 kg. According to data published by Starck et al. [30], this corresponds to images of very good quality, since images with a standard deviation of 10 HU were considered of high quality whereas images with a standard deviation of 30 HU were considered very noisy. Thus, the dose levels proposed in this paper can be reduced when image quality requirements are not particularly high.

Manufacturers are developing strategies in order to adapt tube current to the anatomy of the patient. These strategies are generally based on the assessment of the dose received by the detectors and allow the control of the amount of quantum noise present in the images. On some units the user is asked to enter the image noise level (e.g., the standard deviation value) he wants to work with. This paper provides some hints for choosing these levels. At the same time, manufacturers are proposing the display of the $CTDI_{vol}$ corresponding to the acquisition protocol chosen by the user. This quantity integrates the pitch information and corresponds to the averaged dose received in the slice. It is thus directly linked with the amount of quantum noise present in an image. However, one has to be sure that the $CDTI_{vol}$ indicated by the unit is expressed in the CTDI test object of 16 cm in diameter when dealing with pediatric acquisition, otherwise large dose underestimation would be made.

Having defined $CTDI_{vol}$ for each weight class, the tube currents were calculated from the ${}_{n}CTDI_{w}$ measured at 120 kV (see Table 3). Since no major beam hardening effect is expected when dealing with small abdominal diameters, a tube voltage of 80 kV was chosen for the first weight class (i.e., 2.5–5 kg) and a tube voltage of 100 kV was chosen for the second weight class (i.e., 5–15 kg). This allows a slight increase of the contrast-to-noise ratio of the acquisition and might allow us to further reduce patient dose without reducing the detection of lesions [21].

To conclude, we have shown that the CTDI test objects can be used to find reasonable compromises between image noise and dose as a function of patient weight, and that it is possible to work well below some of the $CTDI_{vol}$ proposed in the literature while keeping an acceptable image noise level. The “automatic exposure systems” developed by manufacturers will be of great help to control patient exposure especially when dealing with the premature or neonate population where dose efficacy is very high.

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