Estimation of adaptive computed tomography dose index based on body weight in pediatric patients

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Abstract

Purpose. The aim of this study was to create an adaptive computed tomography dose index (by volume) (CTDIvol) for pediatric patients that would be fitted to a patient’s particular body weight and to compare the adaptive CTDIvol with the CTDIvol displayed on the screen of the CT console.

Materials and methods. CT images of 60 children whose body weights were known were available for calculating the total amount of modified CT numbers as an attenuation area. The attenuation area values of four differently sized acryl phantoms were also calculated. The dose measurements of all four phantoms were carried out. We combined the results of the abdominal CT and phantom experiments. The weight-based complementary ratio for adaptive CTDIvol was calculated, and the result was applied to an example of pediatric abdominal CT protocol.

Results. The adaptive CTDIvol was always larger than the displayed CTDIvol with both small and large fields of view (FOV). The adaptive CTDIvol was 2.2 times larger than the displayed CTDIvol in the maximum value with the large FOV and 1.2 times larger with the small FOV.

Conclusion. We radiologists must be the child’s advocate and protect children from the deleterious effects of any of our technologies, including a lower indication of CTDIvol on the screen of a CT console.

Key words CT · Radiation dose · Pediatric · CTDI

Introduction

In 1981 the United States Food and Drug administration (FDA) introduced the computed tomography dose index (CTDI) as a physical dose quantity to characterize the radiation exposure delivered by CT scanners. Since the introduction of CTDI, it has commonly been used as the most specific dose quantity for CT. In recent years, dose information, including the local dose index expressed as the CTDI (required by International Electrotechnical Commission standards in many countries) and the total dose index expressed as the dose length product (DLP), has been automatically displayed on CT consoles. This automatic display is quite useful in the daily practice of CT examinations because we can compare it with diagnostic reference levels established in the CTDI and DLP that are appropriate to the peculiar condition of exposure in pediatric CT. Also, we can estimate the effective dose (E in mSv) derived from the values of DLP for a CT examination using appropriately normalized coefficients.

Weight-based radiation dose reduction is widely known to be the simplest and most reliable method for putting into practice the “as low as reasonably achievable” (ALARA) concept. It is based on optimization for patients of various sizes, from newborn infants to large adults. On the other hand, CTDI is not adjusted for patient size. It is measured using only two representative (16 cm and 32 cm diameter) CT dosimetry phantoms in the regions of the adult head and trunk, respectively.

Our hypothesis is that the CTDIvol displayed on the CT scanner appears to be a rough approximation; it may
be a low estimate and may not be correct for children and small adults. Few reports have focused on the relation between body weight and the CTDI.\(^5\) The purpose of this study was to create a revised CTDI\(_{vol}\) for pediatric patients that would be fitted to a patient’s particular body weight and to create a coefficient that can calculate an adaptive CTDI\(_{vol}\) from the displayed CTDI\(_{vol}\).

**Materials and methods**

The institutional review board approved the review of medical records. Informed consent for retrospective analysis of CT was not needed.

To create a revised CTDI\(_{vol}\) that is adaptable to the patient’s body weight, we had to determine the sizes of phantoms that approximately equaled a variety of human body sizes. First, retrospective evaluation using a single section of a pediatric abdominal CT scan was carried out for 60 patients whose body weights were already known at the CT examination. The total amount of modified CT numbers was obtained for these particular images. Second, the total amounts of modified CT numbers for four different-sized phantoms were calculated, and dose measurement of the phantoms was carried out to assess particular nCTDI\(_{vol}\). We integrated the results of the retrospective analysis and the phantom experiment in one equation to establish the ideal CTDI\(_{vol}\) that was adapted to the patient’s body weight.

**Retrospective analysis of pediatric abdominal CT**

We retrospectively reviewed 60 noncontrast pediatric abdominal CT scans obtained between May 2002 and March 2003. All CT examinations were performed for some clinical purpose, as follows: neoplasm and its follow-up (\(n = 26\)), hepatobiliary and pancreatic problems (\(n = 14\)), abdominal inflammation (\(n = 10\)), intestinal abnormality (\(n = 1\)), and other (\(n = 9\)). There were 23 girls and 37 boys, whose ages ranged from 13 days to 16 years (mean 6.38 ± 4.8 years). Their body weights ranged from 3.5 to 70 kg (mean 22.4 ± 14.6 kg).

All CT scans were obtained on an eight-data acquisition system MDCT scanner (LightSpeed Ultra; General Electric Medical Systems, Milwaukee, WI, USA). The scanning parameters were 120 kV, 2.5 mm detector configuration, 0.5 s gantry cycle, 1.63:1.00 helical pitch, and 5 mm reconstructed slice thickness. A published weight-based guideline for a single helical CT scan was obtained for the tube current setting.\(^5\)

Differing amounts of bone, fat, water, and soft tissue components contributed to the CT image. The sum of the pixel values was converted to CT images. Each pixel in a CT image represents the attenuation of the X-ray beam. A pixel value is computed with reference to the linear attenuation coefficients of water.

\[
\text{pixel value} = \text{CT number (HU)} = \frac{\mu_{\text{pixel}} - \mu_{\text{water}}}{\mu_{\text{water}}} \times 1000
\]

The attenuation area is defined as follows:

\[
\text{attenuation area} = \sum \{(\text{pixel value} + 1000) \\
\times [\text{pixel size (mm)}]^2\}
\]

The attenuation area was calculated at the scan level of the porta heptasis in a particular patient because this slice included the largest part of the liver parenchyma (Fig. 1). The attenuation area was calculated using a personal computer with MATLAB software (version 6.5, release 13; The MathWorks, Natick, MA, USA). The relation of the attenuation area in a particular abdominal CT and body weight was evaluated by the regression method.

**Dose measurements of phantoms**

Four different-sized polymethylmethacrylate (PMMA) CT phantoms (8, 16, 24, and 32 cm) in the shape of cylinders were used for the dose measurements. We calculated the total modified CT numbers per single slice in each size of phantom as an attenuation area using the same methodology described above for abdominal CT to correlate axial human body size with PMMA phantom diameter.

The phantom was always positioned at the isocenter of the gantry. A pencil-beam ion chamber (model 10X5–3CT; Radcal, Monrovia, CA, USA) was used together with a radiation dose monitor (radiation monitor controller model 9015; Radcal). Weighted CTDI was obtained as follows: CTDI\(_w\) = \(\frac{1}{3}\) central CTDI + \(\frac{1}{3}\) peripheral CTDI. Normalized CTDI volume (corrected for pitch factor; 1 and mGy/100 mAs, nCTDI\(_{vol}\)) was calculated from the value of CTDI\(_w\). The scanning parameters were 120 kV, 100 mAs, 20 mm beam width, and 1 s per gantry cycle with a single axial slice technique. Data acquisition was performed in both small and large field of view (FOV) modes, respectively. The relation of the attenuation area and nCTDI\(_{vol}\) was analyzed by a regression method with statistical software.

**Creation of complementary ratio for revised CTDI**

We combined two results (retrospective analysis of abdominal CT and phantom experiment) in one equation to establish the relation between body weight and cylinder diameter through the medium of the value of the