ORIGINAL RESEARCH

Dose Assessment in Head CT Scans: Phantom-Based Protocol Optimization

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ABSTRACT
Computed Tomography (CT) scans result in higher radiation dose deposition compared to conventional radiology exams. These tests significantly contribute to both individual and collective radiation exposure, making them a global public health concern. Therefore, optimizing imaging protocols is essential to reduce radiation doses while preserving diagnostic image quality. The development of phantoms plays a crucial role in evaluating and refining different acquisition protocols to achieve this balance.To ensure accurate representation, phantoms must exhibit X-ray absorption characteristics similar to those of the human head. In this study, two cylindrical polymethylmethacrylate (PMMA) head phantoms were tested. One was a standard CT head phantom with a 16 cm diameter, while the other, a newly developed smaller phantom, had a 12 cm diameter. Both phantoms measured 15 cm in length.CT scans were conducted using a GE LightSpeed VCT scanner with 64 channels, employing various acquisition protocols. The central slice of each phantom was irradiated multiple times, and a pencil ionization chamber was used to measure the CT air kerma index in PMMA (Ck,PMMA,100) and the CT dose index (CTDI). Based on these measurements, the weighted and volumetric CT dose index values (CTDIw and CTDIvol) were determined for 10 cm scan lengths in helical mode.
Scans were performed at different voltage levels (80, 100, and 120 kV) and varying tube current-time products (mAs). Using routine head scan protocols, the absorbed dose (CTDIvol) ranged from 39.22 to 49.67 mGy. However, optimized protocols resulted in absorbed doses between 20.89 and 31.93 mGy, achieving a reduction of up to 57.94% in the smaller 12 cm phantom.
Keywords: Computed tomography, Phantom, Dosimetry

INTRODUCTION

Computed Tomography (CT) is one of the most used exams for radiologic diagnostic in medicine. It is a very fast test that can produce high quality images. However, the increasing demand for CT had a considerable impact on doses provided to patients and on the exposure of the population as whole, being a public health concern worldwide.[1,2] According to UNSCEAR report the use of CT contributed with 62% of the collective dose from diagnostic radiological tests[3]. Many factors collaborated to the increased demand for CT scans, including the constant technological evolution of the equipment associated to greater availability and a relative tendency to decrease exam costs.[4,5]

Patients undergoing CT scans can range from neonates to oversized adults. However, radiation doses in CT are generally measured in cylindrical PMMA phantoms, that represent a standard adult patient. These phantoms are designed to simulate a head, 16 cm in diameter, and a body, 32 cm.[6,7]

It is difficult to obtain reliable quantitative values of patient doses from any measurements performed in these standard phantoms, because patients have sizes and body compositions that can differ markedly from the phantoms, such as pediatric and obese patients. The development of phantoms allows testing different acquisition protocols.[8,9] For this, the phantoms must have an X-ray beam absorption characteristic similar to the represented patient.

The increasing demand for CT scans in pediatric patients is mainly due to the high rates of traumatic injuries from car accidents, falls on bicycles, blunt trauma, traumatic brain injury, as well as a significant increase in the incidence of childhood neoplasms, being the CT images used in the diagnostic process. Therefore, acquisition protocols should be used that determine the reduction of the radiation dose without compromising the diagnostic quality.[10,11] The risks of stochastic effects increase in children due to the tissue radiosensitivity allied to the long-life expectancy. The dose deposited in a pediatric patient is directly related to the energy that

was retained during the process of exposure to ionizing radiation.[12,13]
This study utilized two CT head phantoms—a standard head phantom and a smaller-volume phantom—to analyze dose distribution and determine the computed tomography dose index (CTDI). Additionally, various acquisition protocols were tested by adjusting the X-ray tube voltage (80, 100, and 120 kV) and tube current-time product (mAs).

MATERIALS AND METHODS

The experiment was carried out using a GE LightSpeed VCT CT scanner with 64 channels. To conduct this study, experimental measurements were obtained using two head phantoms, both made of polymethyl methacrylate (PMMA). These phantoms were designed and constructed by the research team at the Center for Research in Biomedical Engineering (CENEB) at the Federal Center for Technological Education of Minas Gerais (CEFET-MG). They represent the head of a standard adult and a pediatric patient.
The adult head phantom is a cylindrical model with a diameter of 16 cm and a length of 15 cm, serving as the reference standard for dose measurements in head CT scans. The pediatric head phantom, also cylindrical, has a smaller diameter of 12 cm while maintaining the same length of 15 cm, simulating the head size of a pediatric patient.
Both phantoms are designed with five openings to accommodate dosimeters—phantoms.
one central opening and four peripheral openings positioned 90° apart. Each opening measures 1.27 cm in diameter and extends the full 15 cm length of the phantom. The peripheral openings are located 1 cm from the phantom’s outer edge. Figure 1 provides an illustration detailing the dimensions of the adult and pediatric PMMA

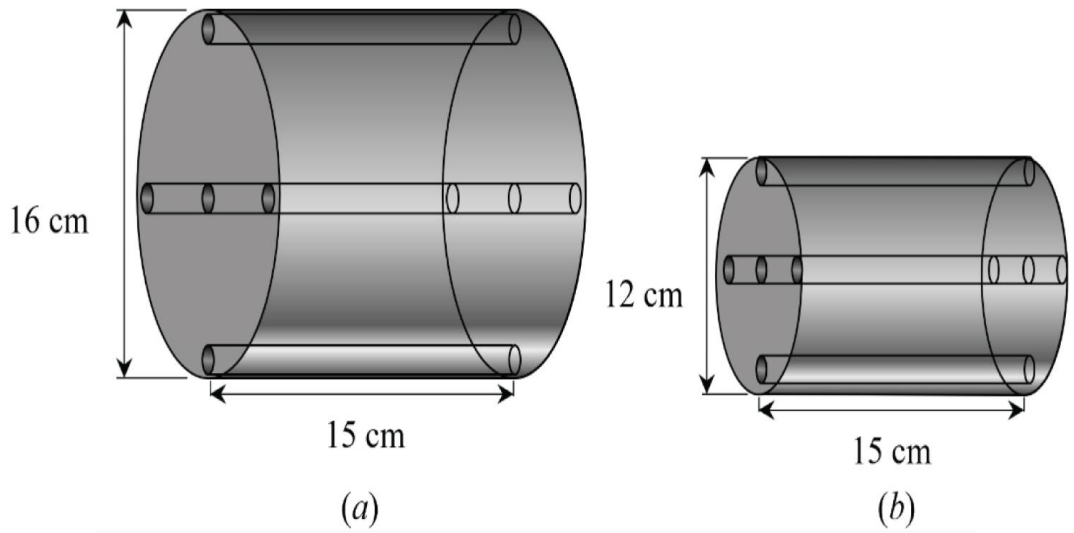

Figure 1: Head phantoms dimensions: (a) Adult standard and (b) Pediatric.

The standard adult head phantom serves as the reference for dose measurements in head CT scans. As a result, all head CT scans conducted on a specific scanner include a report that provides an estimated patient absorbed dose (CTDI), calculated based on the scan of this phantom. Figure 2 presents an image of these phantoms positioned at the isocenter of the CT scanner's gantry.

Figure 2: PMMA head phantom images: (a) Adult standard and (b) Pediatric.

Dose measurements were conducted by placing the head phantom at the isocenter of the CT scanner's gantry and aligning its openings with reference positions at 3, 6, 9, and 12 o'clock, using the scanner's laser guidance system. The phantom's openings were initially filled with PMMA rods, which were sequentially removed to allow precise placement of the pencil ionization chamber for dose assessment in all five regions.

A RADCAL ACCU-GOLD model 10X6-3CT pencil ionization chamber was used to measure the CT air kerma in PMMA (Ck, PMMA,100) at each opening of both phantoms. Initially, a scout scan was performed to verify proper phantom alignment and determine the exact position of the central slice. The central slice was then irradiated multiple times.

For each chamber placement, five measurements were taken, resulting in a minimum of 25 measurements per protocol for each phantom. During central slice irradiations, the remaining openings were filled with PMMA rods to maintain structural consistency. Based on these measurements, the weighted and volumetric CT Dose Index values (CTDIw and CTDIvol) were calculated for 10 cm scans of the central region of the head phantoms in helical mode.

The $CTDI_w$ and $CTDI_{vol}$ were calculated according to the Eq. 1 and 2:[13,14]

CTDIw = 13· (CTDI100, central + 23·CTDI100, per) (1)

CTDIvol = CTDIwpitch (2)

where, $CTDI_{100, central}$ is the dose index value found at the central position and CTDI100, per is the average dose index value at the peripheral positions of the head phantom. The scans were performed using different voltage values (80, 100 and 120 kV) and charge (mA.s). In order to obtain the CT Dose Index (CTDI) values from the air kerma values the measurements were adjusted using a conversion factor (F_c) air/PMMA. The F_c used are 1.0418, 1.0324, and 1.0106 for the X-ray beam generated with 120, 100 and 80 kV, respectively.[14-16]

The protocol for irradiating the central slice of the phantom in axial mode was performed using the following parameters: a tube current of 100 mA, a charge of 100 mAs, a tube rotation time of 1 second, a beam thickness of 10 mm, and three different voltage settings (120, 100, and 80 kV).

Additionally, helical scans covering a 10 cm length in the central region of the head phantoms were conducted to determine the optimal tube current using the scanner's automatic exposure control (Auto mA) at different voltage levels. Typically, during the initial slices of a scan, the system adjusts the tube current (mA), stabilizing once the first few slices are irradiated, as the phantom maintains a uniform size throughout.

After defining a reference current value based on the stabilized current in the central slice, additional scans were performed with fixed current settings lower than those suggested by automatic exposure control. For each tested current value, image noise in the central slice was measured to determine the optimal current setting for each phantom and voltage level.

The scanning protocols for the central region of the phantom were configured using pitch values as close as possible to 1, as supported by the CT system. Table 1 presents the standard CT head scanning protocol used in routine examinations, irrespective of the patient's size or age.

Table 1: Routine protocol of CT head scan.

X-ray Tube Voltage (kV)	**Tube Current (mA)**	Exposure **(mAs)**	**Rotation time (s)**	**Slice Thickness (mm)**	**Table Pitch Ratio**	**Image Reconstruction Thickness (mm)**
120	200	100	0.5	40	0.984	1.25

In order to validate the quality of the CT images, a noise analysis of the central slice image was performed in each helical CT scan, aiming at maintaining the diagnostic quality of the images. The noise value had its maximum acceptable limit of 1%, considering that the phantom is homogeneous.[15-17]

The limitation of noise when using a homogeneous material directly affects the quality of diagnostic images of the human body. Therefore, as a control parameter for evaluating new protocols, a noise threshold of 1% in the central slice image was established to ensure the diagnostic accuracy of patient imaging.

Four regions of interest (ROIs) were selected and analyzed in the image. Noise (N) was determined as a percentage by calculating the standard deviation relative to the mean Hounsfield Unit (HU) value, using Equation 3.

N%=(HU+1000SDHU)×100 (3)

Where:

- N%N\%N% = Noise percentage
- SDHU\text{SD}_{HU}SDHU = Standard deviation of Hounsfield Unit (HU) values
- HU¯\overline{HU}HU = Mean Hounsfield Unit value

This formula expresses noise as a percentage by normalizing the standard deviation relative to the adjusted mean HU value.

RESULTS

Dose measurements

Table 2 presents the average values and standard deviations of both point-specific and weighted air kerma in PMMA (Ck,100,PMMA and Cw), as well as the absorbed dose (CTDIw). These values were derived from Ck,100,PMMA measurements taken at five positions within the phantoms. The data was obtained using the defined parameters for central slice irradiation (10 mm) with a fixed charge of 100 mAs.

Table 2: Values of Ck,100, PMMA, Cw and CTDIw in mGy standard deviation for head phantoms.

Measurement Position	Phantom					
	Adult standard (16 cm)			Pediatric Phantom (12 cm)		
	120 kV	100 kV	80 kV	120 kV	100 kV	80 kV
Central	17.52 ± 0.03*	11.28 ± 0.09	5.94 ± 0.01	22.99 ± 0.07	14.84 ± 0.06	8.02 ± 0.04
3 o'clock	19.48 ± 0.02	12.94 ± 0.22	7.10 ± 0.06	23.84 ± 0.24	15.57 ± 0.14	8.61 ± 0.10
6 o'clock	18.17 ± 0.09	11.98 ± 0.11	5.84 ± 0.02	21.98 ± 0.18	14.24 ± 0.15	7.72 ± 0.08
9 o'clock	19.04 ± 0.04	12.83 ± 0.08	6.94 ± 0.06	23.67 ± 0.22	15.46 ± 0.08	8.61 ± 0.09
12 o'clock	21.39 ± 0.38	14.10 ± 0.08	8.93 ± 0.10	25.26 ± 0.54	17.29 ± 0.22	9.87 ± 0.13
Weighted Air KermaC_w(mGy)	18.85 ± 0.10	12.40 ± 0.11	6.69 ± 0.04	23.46 ± 0.39	15.37 ± 0.12	8.47 ± 0.08
Weighted CT Dose Index $CTDI_w$(mGy)	19.64 ± 0.10	12.80 ± 0.12	6.76 ± 0.04	24.44 ± 0.40	15.87 ± 0.12	8.56 ± 0.08

*Standard deviation

The protocol utilizing a voltage of 120 kV resulted in the highest absorbed dose, recorded at position 12, with values of 21.39 mGy for the adult phantom and 25.26 mGy for the pediatric phantom. Conversely, the lowest dose values were observed at position 6 for both phantoms at 80 kV, measuring 5.84 mGy in the adult phantom and 7.72 mGy in the pediatric phantom. The similarity in dose values at positions 3 and 9 suggests proper alignment of the object within the gantry isocenter.

Upon analyzing the obtained measurements, the pediatric phantom consistently exhibited higher dose values. This is because, while the irradiation parameters for the central slice remained the same, the pediatric phantom's smaller cross-sectional area led to greater dose deposition. Additionally, scans performed at 80 kV consistently resulted in the lowest dose values, as the lower average beam energy reduced radiation absorption. In contrast, the 120 kV setting produced the highest dose deposition due to its higher beam energy.

Optimized CT scan protocols

Table 3 presents the absorbed dose values (CTDIvol) along with their standard deviations for both phantoms when scanned using routine and optimized protocols. These measurements were obtained at different voltage levels while applying an optimized charge to the X-ray tube during scans of the central region.

In the optimized protocols, the charge value (mAs) was adjusted to ensure that the noise in the central slice remained below 1%, without compromising image quality. All other parameters—pitch, tube rotation time, beam thickness, and image reconstruction—were kept consistent with those used in the routine protocol (as outlined in Table 1).

Table 3: Routine and optimized protocols.

Phantom Size	Protocol Type	Voltage (kV)	Tube Charge (mAs)	$CTDI_{VOL}$ (mGy) ±SD
Adult (16 cm)	Standard	120	200	39.92 ± 0.21*
	Optimized 1	120	160	31.93 ± 0.16
	Optimized 2	100	240	31.22 ± 0.28
	Optimized 3	80	420	28.87 ± 0.18
Pediatric (12 cm)	Standard	120	200	49.67 ± 0.82
	Optimized 4	120	100	24.83 ± 0.41
	Optimized 5	100	144	23.23 ± 0.18
	Optimized 6	80	240	20.89 ± 0.20

*Standard deviation

During the evaluation of new scanning protocols, the pitch value was maintained at 0.984, consistent with the routine protocol, as it was the closest available setting to 1 on the CT scanner.

For the adult standard head phantom (16 cm in diameter), the absorbed dose across the tested protocols ranged from 28.87 to 31.93 mGy. The lowest dose was recorded using the Optimized Protocol 3 (Opt. 3), which applied a voltage of 80 kV and a tube charge of 420 mAs. The noise level in the central slice image for this scan was 0.978%, meeting the established diagnostic quality criteria.

Implementing Opt. 3 resulted in a 27.68% reduction in absorbed dose, lowering it from 39.92 mGy to 28.87 mGy.

For the pediatric head phantom (12 cm in diameter), the lowest absorbed dose was 20.89 mGy, recorded under Optimized Protocol 6 (Opt. 6) with 80 kV and 240 mAs. The noise level in the central slice image for this scan was 0.929%, ensuring that image quality remained within diagnostic standards. The Opt. 6 protocol led to a 57.94% reduction in absorbed dose, decreasing it from 49.67 mGy to 20.89 mGy. Furthermore, the absorbed dose in the routine pediatric phantom scan was 19.63% higher than that of the adult phantom, highlighting the increased radiation deposition in smaller anatomical structures.

Figure 3 presents a graphical comparison of the absorbed dose values (CTDIvol) for both adult and pediatric phantoms across the different protocols listed in Table 3.

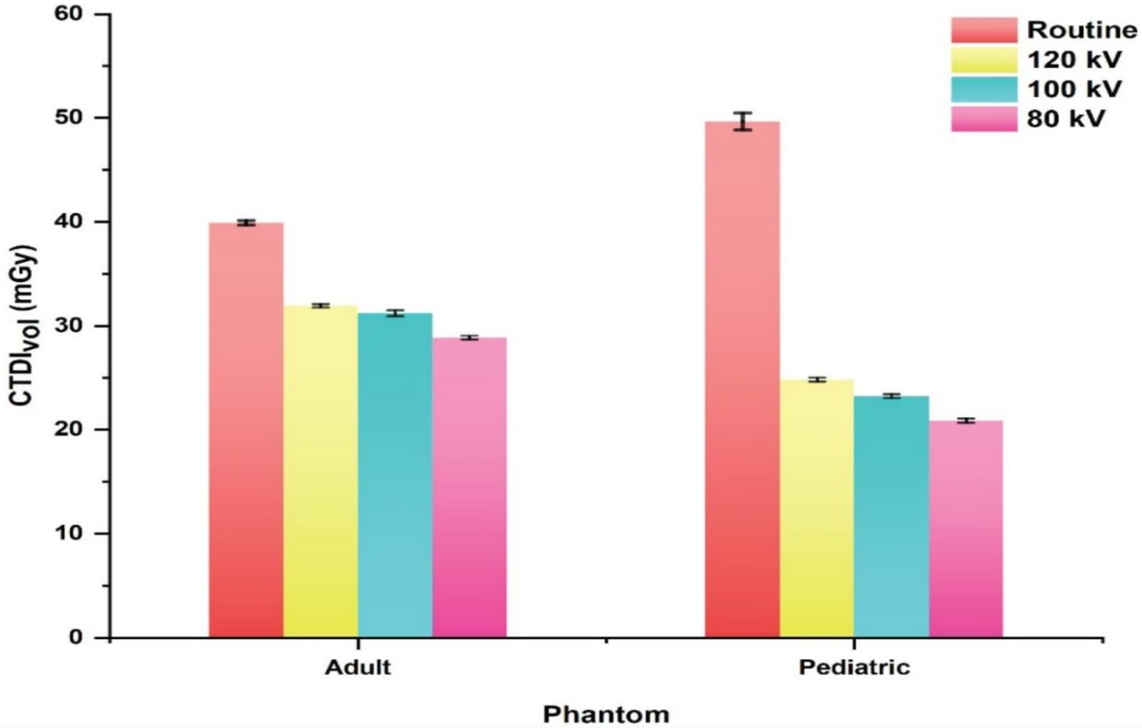

Figure 3: $CTDI_{vol}$ values for adults and pediatric head phantoms obtained with routine and optimized protocols.

Analyzing the absorbed dose values obtained from the tested protocols, it was observed that both the adult and pediatric phantoms achieved optimal CT scans at a voltage of 80 kV. The optimized protocols with the lowest absorbed dose yielded noise levels below 1%, making them a suitable alternative for reducing patient radiation exposure while preserving diagnostic image quality.

Additionally, the protocols presented in Table 3 were selected from a larger set of tested protocols, in which mAs values were adjusted iteratively until noise levels below 1% were achieved in the central slice image analysis. It is important to note that mAs and pitch values cannot be adjusted arbitrarily, as CT scanners offer only a predefined set of selectable values in their system menu for testing and implementation.

CONCLUSIONS

CT scan patients vary significantly in size, from newborns to large adults. However, radiation dose measurements are typically conducted using PMMA phantoms designed to represent a standard adult patient. This approach presents challenges in obtaining accurate quantitative dose values, as real patients may differ in size and body composition from the standard phantom. This is particularly relevant for pediatric patients, smaller adults, and individuals with larger or obese body types.

The phantoms developed in this study address these limitations by representing different patient sizes, enabling the evaluation of various acquisition protocols for head CT scans. This research provides valuable insights into dose reduction strategies for both adult and pediatric head CT scans, reinforcing the importance of implementing optimized protocols that minimize radiation exposure while preserving diagnostic image quality.

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