

# Quality Control and Optimization of Computed Tomography Dose Index Volume (CTDI<sub>vol</sub>) of LightSpeed RT16 Xtra CT Scanner

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## Abstract

Dose estimation and quality control in computed tomography (CT) scanners are useful in controlling the dose of radiation given to patients while tests are carried out. The study was performed in a 16-slice Computed Tomography (CT) system of LightSpeed RT16 Xtra CT scanner. Quality control was done using a vendor-provided QA Phantom, and the six aspects of image quality were measured. For CT dosimetry, Computed Tomography Dose index volume (CTDI<sub>vol</sub>) was performed using Computed Tomography Dose Index (CTDI) Phantom. CTDI Phantom consists of three parts: Pediatric Head, Adult Head, and Adult Body Phantom. A 10 cm long pencil ion chamber DCT-10 was used to measure the dose at different positions inside the CTDI Phantom. Data were collected using MagicMax Universal software. For dose estimation of the CTDI<sub>vol</sub> Report of AAPM Task Group, 96 and 111 formalisms were used. For Pediatric Head, Adult Head, and Adult Body Phantom the measured CIDI<sub>vol</sub> was 61.04 mGy, 48.11 mGy, and 18.08 mGy respectively. The study has shown deviations of 7%, 15%, and 19% between estimated and console-displayed doses for Pediatric Head, Adult Head, and Adult Body scan techniques respectively. The six aspects of image quality measured by QA Phantom were found to be compatible with the specifications of the machine and CTDI<sub>vol</sub> measured by CTDI Phantom were found within a tolerance limit of  $\pm 20\%$ . Hence, the QC and dosimetry of the mentioned machine are within the limit.

## Keywords

Quality Control, CTDI<sub>vol</sub>, LightSpeed RT<sup>16</sup> Xtra CT Scanner, Phantom

## 1. Introduction

For developing countries, the availability of CT machines is limited. For that reason, these CT machines are used to diagnose large numbers of patients in hospitals. In CT, the radiation beam comes from various angles and the dose of CT radiation received by the patient is high [1]. The International Commission on Radiological Protection (ICRP) estimated that the threshold for Occupationally exposed workers is 20 mSv a year, averaged over defined periods of 5 years with no single year >50 mSv and for the public is one mSv a year, higher values are permitted if the average over 5 years is not above one mSv a year [2]. At present, Square Hospitals Limited is one of the few hospitals in Bangladesh that has a LightSpeed RT16 Xtra CT scanner. CT scans can be performed on the head, chest, abdomen, and pelvis, among other areas of the body, using this scanner.

Quality control is defined as the operational methods and activities made use of to satisfy standards of quality. To ensure that the CT scanner provides consistent image quality, a Quality Assurance (QA) program is implemented, which involves measuring six aspects of image quality with a Quality Assurance (QA) Phantom. Contrast scale, high contrast spatial resolution, low contrast detectability, noise and uniformity, slice thickness, and laser light accuracy are all measured. The QA Phantom contains three sections, each corresponding to a single plane. The scan locations can be set manually using the computer interface of the scanner and named according to their distance in millimeters. The section's scan locations were resolution block, contrast membrane, and water bath. We have investigated six aspects of image quality, including contrast scale, high contrast spatial resolution, low contrast detectability, noise and uniformity, slice thickness, and laser light accuracy. The capacity to discern subtle differences in an image's intensity is known as contrast resolution. The metric is employed in medical imaging to assess the level of acquired image quality. The capacity to identify minutely spaced lines or holes whose signals significantly deviate from the background is known as high-contrast resolution. The capacity to identify and distinguish objects with only minute variations in signal strength is known as low-contrast resolution. Image noise, which is often a component of electrical noise, is the random changing of brightness or color information in images. The image sensor and circuitry of a digital camera or scanner may produce it. Slice Thickness is a crucial factor in deciding how much anatomical detail is recorded in the picture.

CTDI (Computed Tomography Dose Index) is an estimation of the radiation dose delivered to the patient during a CT scan. The CTDI value is computed with the help of a specialized device known as a CTDI Phantom, which is a cylinder-shaped device made of a material that mimics the density of human tissue. The CTDI Phantom is a useful tool in CT dose management because it enables radiologists and medical physicists to measure and optimize radiation dose for various imaging protocols. When choosing overall dose levels in CT, different scanner manufacturers use quite diverse approaches, which can and does cause misunderstanding in the medical environment. One method is to determine the actual noise

level in the reconstructed CT image and use that value as a reference to change the mA levels used to determine the overall dosage levels. This method results in larger mA values and radiation exposures when lower noise levels are chosen. One can set the maximum and lowest mA values as a safety measure. Setting a nominal mA level that the system utilizes as guidance for the overall dose settings is an alternative method of mA modulation; however, in this case, higher nominal mA values lead to higher mA settings and a higher radiation dose to the patient. mA modulation techniques yield good CT pictures at almost ideal radiation dose levels when applied appropriately. Consequently, it is a crucial component of contemporary CT scanners that have to be applied, particularly for body imaging. But when the CT computer is given control over something as significant as radiation dose regulation, the operators, technologists, radiologists, and medical physicists must have a solid understanding of how to operate the mA modulation technology on each CT scanner they use [3]. CTDI volume helps estimate the dose that is given to the patient, and hence, it is crucial for understanding optimal radiation dose levels.

A 2014 study sought to examine the characteristics of the most commonly used QA Phantom. The Phantoms' inter-Phantom variations in CT-number values, image uniformity, and low contrast resolution were assessed. Comparisons were made between manual image analysis and the results obtained from the automatic evaluation software QA lite. Some inter-Phantom variations were observed in the Phantoms' low contrast resolution and CT-number modules, which were interpreted as significant depending on the regulatory framework used. The homogeneous modules were discovered to be more invariant [4].

CTDI Phantoms are commonly used in CT scanner quality assurance programs. These programs include regular CTDI and other parameter measurements to ensure that scanners are operating within acceptable limits. Many studies have shown that these programs are effective in maintaining CT imaging quality and safety [5]. CTDI measurements can provide radiologists and other clinicians with useful information when making patient care decisions. CTDI measurements, for example, can help identify patients at risk of receiving an excessive radiation dose and guide decisions about whether to repeat scans or use alternative imaging techniques. Ongoing research is required to address the limitations of CTDI measurements and to improve their accuracy and usefulness in clinical practice.

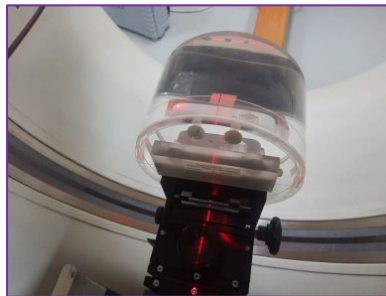
To determine CT radiation doses for various patient sizes ranging from infants to large adults, the dependence of the computed tomography dose index (CTDI) upon the size of the Phantom, the kVp selected and the scan mode employed were studied. Measurements were done on Phantom sizes ranging from 6 cm to 32 cm. The x-ray tube potential ranged from 80 to 140 kVp. The scan modes utilized for the measurements included consecutive axial scans, single-slice helical scans with variable pitch, and multislice helical scans with variable pitch. The results were consolidated into simplified equations that related the Phantom diameter and kVp to the measured CTDI. The CTDI appeared to be an exponential function of

Phantom diameter. For the same kVp and mAs, the radiation doses for smaller Phantoms are much greater than for larger sizes. A method was also given for converting axial CT dose measurements to appropriate MSAD values for helical CT scans [6].

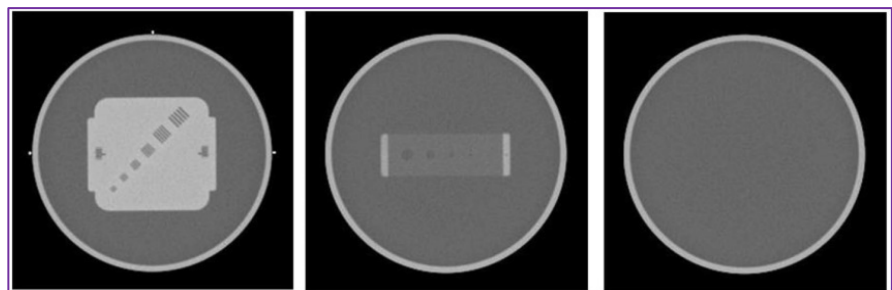
## 2. Materials and Methods

### 2.1. Quality Assurance Phantom

The QA Phantom shown in **Figure 1** and **Figure 2** consists of three sections, each corresponding to a single scan plane. Section 1 assesses image quality in four ways: contrast scale, high contrast resolution, slice thickness, and laser light accuracy. Section 2 assesses low contrast detectability. Section 3 assesses noise and consistency [7]. All the scans were performed according to the vendor-provided interface and input parameters, which were mentioned in a tabular format.



**Figure 1.** Quality assurance phantom.



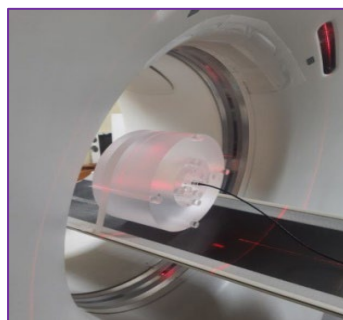
**Figure 2.** Console-displayed images of the three sections of QA Phantom.

The water and Plexiglas CT values in the Phantom, provide as a reference point for tracking the system contrast scale over time. The resolution block has bars that measure 1.6 mm, 1.3 mm, 1.0 mm, 0.8 mm, 0.6 mm, and 0.5 mm. The standard deviation of the pixel values was assessed using the standard methodology in a single or multiple bar pattern to provide a quantitative tool for assessing changes in system resolution. Hounsfield units (HU) were used to note the CT values. The Phantom portion's doped polystyrene membrane is pierced with holes measuring 10.0 mm, 7.5 mm, 5.0 mm, 3.0 mm, and 1.0 mm, and it is immersed in water. CT numbers were recorded and the ROI was positioned at several points. Noise limits the resolution of low contrast and obscures anatomical structures similar to the

surrounding tissue. Measured X-ray energy, reconstruction techniques, and electrical, mechanical, and mathematical variances in the electronic outputs all contribute to noise. The uniformity difference between the center ROI and the average of the edge ROIS should be  $0 \pm 3$ , and the center ROI's standard deviation should be equal to  $3.2 \pm 0.3$ . The slice thickness was determined by counting the visible holes in the displayed image using the specified Window level and width.

## 2.2. CTDI Phantom

A LightSpeed RT<sup>16</sup> Xtra CT scanner with CTDI Phantom placed on the patient bed is shown in **Figure 3**. The CTDI Phantom is made of a poly methyl-methacrylate (PMMA) compound that, depending on its radii, is a prototype of an adult head and body. The CTDI index can be used to estimate the dose of patients during CT scans. This is accomplished by placing a pencil ionization chamber at the center and four positions of the periphery of the CTDI Phantom and calculating the dose inside. The CTDI value is then calculated using the radiation dose measured by the Phantom's dosimeters. The CTDI value is expressed in milliGray (mGy) units and is used to estimate the radiation dose received by a patient during a CT scan. The dose measured by the Ionization chamber shown in **Figure 4** is converted using the analog-to-digital converter device shown in **Figure 5** into a machine-readable output. The device's software is known as MagicMaX software version 2.1.2 which displays the computer reading of the dose measured by the pencil ionization chamber.



**Figure 3.** CT machine with a CTDI phantom.



**Figure 4.** Pencil ionisation chamber.



**Figure 5.** MagicMaX analog-to-digital converter.

Nominal beam width refers to the beam width as reported by the scanner. With a nominal x-ray beam width of  $NT$ , the scanner is rotated once in a circular (axial or sequential) style. The primary and scattered radiation is measured over a 100-mm length, and the center of the x-ray beam is at  $z = 0$ . The measurement of the dose distribution,  $D(z)$ , along the  $z$ -axis, is described by the  $CTDI_{100}$  and is defined as [8].

$$CTDI_{100} = \frac{1}{NT} \int_{-50 \text{ mm}}^{+50 \text{ mm}} D(z) dz \quad (1)$$

Both the center ( $CTDI_{100}$ , center) and the periphery ( $CTDI_{100}$ , periphery) are measured using the  $CTDI_{100}$ . A reliable estimate of the average dose to the Phantom (at the central CT slice along  $z$ ) can be obtained by combining the center and periphery readings using a 1/3 and 2/3 weighting scheme. This yields the weighted CTDI,  $CTDI_w$  is [3].

$$CTDI_w = \frac{1}{3} CTDI_{\text{central}} + \frac{2}{3} CTDI_{\text{peripheral}} \quad (2)$$

The pitch is calculated by dividing the nominal beam width ( $NT$ ) (in mm) by the table translation distance (mm) during a complete 360-degree rotation of the gantry. This relationship means that the  $CTDI_w$  is transformed using to get the volume CTDI ( $CTDI_{\text{vol}}$ ).

$$CTDI_{\text{vol}} = \frac{CTDI_w}{\text{pitch factor}} \quad [9] \quad (3)$$

where,  $N$  = number of slices in a single axial scan

$T$  = width of one slice (mm)

Pitch factor = 1 (for axial scan)

### 3. Results and Discussion

#### 3.1. Quality Assurance (QA) Test

The temperature and humidity of the CT Simulation Room were 22.5°C and 50% respectively during the Quality Assurance (QA) test.

### 3.1.1. Contrast Scale

**Table 1** tabulates the CT values of water and Plexiglas in the Phantom, which serve as the benchmark for monitoring the system contrast scale over time. The average CT number of water is 4.72 and that of Plexiglass is 132.495. The difference between the two is 127.775, which is within  $120 \pm 12$  which means the contrast scale test is within the reference limit.

**Table 1.** Contrast scale test data.

Position	CT number	Standard deviation
Overwater 12 o'clock (0, 8)	3.54	3.14
Overwater 3 o'clock (8, 0)	5.14	3.26
Overwater 6 o'clock (0, -8)	5.30	3.12
Overwater 9 o'clock (-8, 0)	4.90	2.87
Over plexiglass above line pattern (-2, 2)	132.68	3.89
Over plexiglass below line pattern (2, -2)	132.31	3.84

### 3.1.2. High Contrast Spatial Resolution

Images that were reconstructed using the Bone method are shown in **Figure 6** and show all five 0.6 mm bars. The CT values were noted and tabulated in **Table 2** when the box ROI was placed over the bar pattern and scaled to fit within it. For a ROI in the 1.6 mm bar pattern, the standard deviation is within  $37 \pm 4$  for the common algorithm. Consequently, the high contrast spatial resolution test met the requirements.



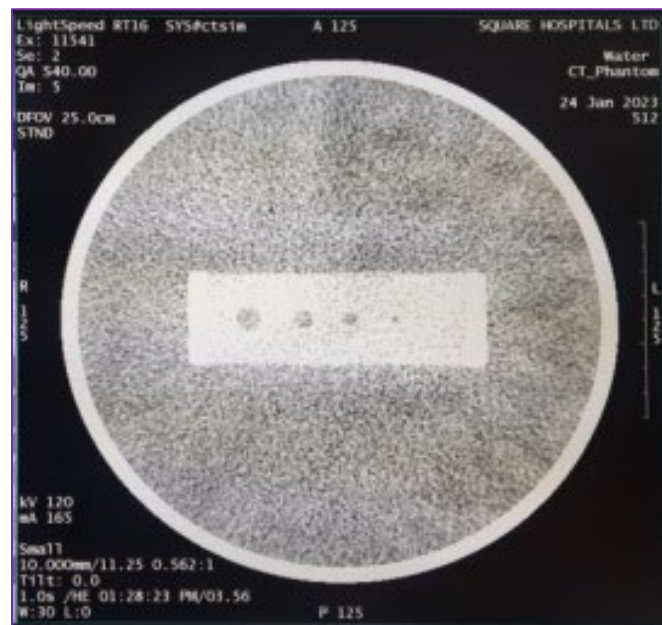
**Figure 6.** Resolution block section with bars.

**Table 2.** High contrast spatial resolution data.

Bar size	CT number	Standard deviation
1.6 mm	72.91	38.30
1.3 mm	71.33	29.36
1.0 mm	69.38	20.39
0.8 mm	72.32	10.37
0.6 mm	70.64	4.67
0.5 mm	71.05	3.80

### 3.1.3. Low Contrast Detectability

The doped polystyrene membrane in this Phantom portion, which is seen in **Figure 7**, is submerged in water and punctured with holes of the following sizes: 10.0 mm, 7.5 mm, 5.0 mm, 3.0 mm, and 1.0 mm. The ROI was placed at different positions and CT numbers were recorded which is given in **Table 3**.

**Figure 7.** Contrast membrane section with holes.**Table 3.** Low contrast detectability data.

Position of Region of Interest (ROA)	CT number	Standard deviation
polystyrene, just above the holes	14.10	3.70
water section, above the membrane	5.42	3.30
polystyrene, just below the holes	13.80	3.44
water section, below the membrane	4.34	3.14

### 3.1.4. Noise and Uniformity

**Table 4** records the CT number and displays it. The standard deviation of the

center ROI is within  $3.2 \pm 0.3$  and the uniformity difference between the center ROI and the average of the edge ROIS is within  $0 \pm 3$ . Both conditions fulfill the criteria.

**Table 4.** Noise and uniformity data.

Position	CT number	Standard deviation
Center of image (0, 0)	1.28	3.01
12 o'clock (0, 8)	1.40	3.50
3 o'clock (8, 0)	1.24	2.51
6 o'clock (0, -8)	2.04	2.69
9 o'clock (-8, 0)	1.50	2.70

### 3.1.5. Slice Thickness Test

Two equally gray holes in the image count as one millimeter of slice thickness, while the black lines in the image indicate a whole millimeter of the slice thickness. **Table 5** contains the determined values. Slice thickness does not vary by more than  $\pm 1$  mm from the expected value when evaluated according to the instructions. So the slice thickness test criterion is fulfilled.

**Table 5.** Slice thickness data.

Slice thickness	Window width/level	Number of visible lines
2.5	250/-25	2.25
3.75	250/+25	3.75
5.0	250/+50	5.5

### 3.1.6. Laser Light Accuracy

Four sticky white papers have been put on both sides wall for two vertical lasers and the other two have been put on the CT gantry head, patient table, and floor for ceiling laser. The laser is lightened up, and a line is drawn based on the laser light width with a black pen as a reference. By the visual inspection, the laser light was within the black line.

## 3.2. CT Dosimetry Using CTDI Phantom

CTDI<sub>100</sub> was calculated using the collected data with a temperature of 22.5°C and pressure of 1013.4 hPa. The data were collected using the pencil ionization chamber and analyzed using the MagicMax Universal software at different locations of the CTDI Phantom for Pediatric Head, Adult Head, and Adult Body combinations and calculated using Equation (1) and are tabulated in **Tables 6-8** given below:

The CTDI<sub>100</sub> of the Adult Body Phantom was measured using Equation (1) at different tube voltage with current constant at 250 mA. The graph in **Figure 8** implies

**Table 6.** CTDI<sub>100</sub> measurement for pediatric head Phantom in mGy.

Position of ionization chamber	CTDI <sub>100</sub> (mGy)			Average CTDI <sub>100</sub> (mGy)	Average CTDI <sub>100</sub> (mGy)
Centre	0.780	0.778	0.750	0.769	0.769
Periphery (12 o'clock)	0.850	0.846	0.836	0.844	0.761 (Average of all periphery)
Periphery (3 o'clock)	0.768	0.762	0.752	0.761	
Periphery (6 o'clock)	0.696	0.696	0.680	0.691	
Periphery (9 o'clock)	0.754	0.754	0.742	0.750	

**Table 7.** CTDI<sub>100</sub> measurement for adult head Phantom in mGy.

Position of ionization chamber	CTDI <sub>100</sub> (mGy)			Average CTDI <sub>100</sub> (mGy)	Average CTDI <sub>100</sub> (mGy)
Centre	0.579	0.577	0.597	0.584	0.584
Periphery (12 o'clock)	0.626	0.636	0.626	0.629	0.610 (Average of all periphery)
Periphery (3 o'clock)	0.615	0.626	0.615	0.618	
Periphery (6 o'clock)	0.585	0.583	0.583	0.583	
Periphery (9 o'clock)	0.605	0.617	0.609	0.610	

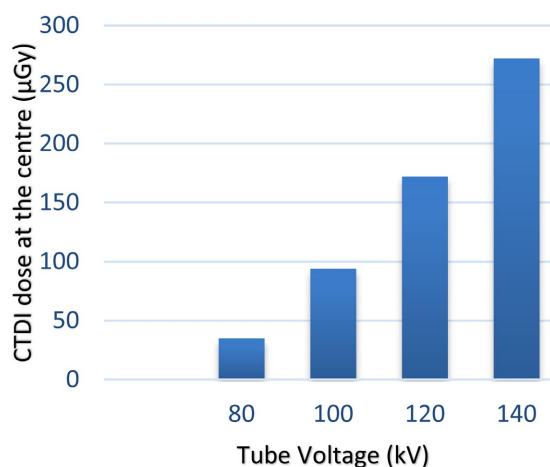
**Table 8.** CTDI<sub>100</sub> measurement for Adult Body Phantom in mGy.

Position of ionization chamber	CTDI <sub>100</sub> (mGy)			Average CTDI <sub>100</sub> (mGy)	Average CTDI <sub>100</sub> (mGy)
Centre	0.173	0.172	0.181	0.175	0.175
Periphery (12 o'clock)	0.253	0.253	0.263	0.256	0.251 (Average of all periphery)
Periphery (3 o'clock)	0.242	0.242	0.253	0.246	
Periphery (6 o'clock)	0.223	0.219	0.242	0.228	
Periphery (9 o'clock)	0.285	0.283	0.283	0.275	

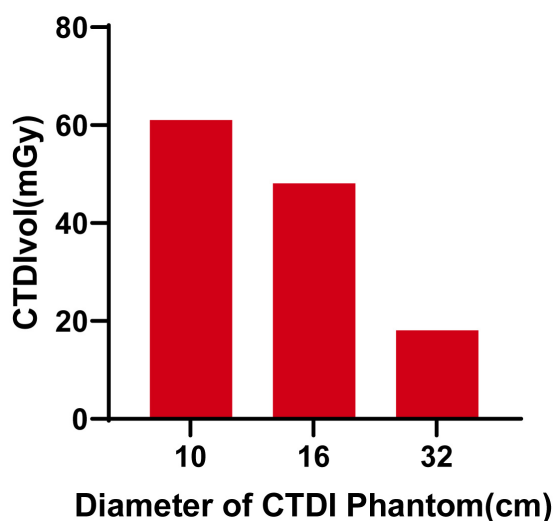
that the value of CTDI<sub>100</sub> increases when the tube voltage is increased with a fixed tube current.

The CTDI<sub>100</sub> of the Pediatric Head, Adult Head, and Adult Body were measured. The CTDI<sub>w</sub> of Pediatric Head, Adult Head, and Adult Body were measured from CTDI<sub>100</sub> by using Equation (2). Finally, the CTDI<sub>vol</sub> was calculated using Equation (3). A graph is plotted by plotting the different values of CTDI<sub>vol</sub> for different diameters of the Phantom to find the relation between CTDI<sub>vol</sub> and with Phantom diameter. The resulting graph in **Figure 9** confirms the linear reduction of CTDI<sub>vol</sub> due to an increase in diameter.

Radiation Dosimetry in CT (Revised 3-25-2024) states that Percent difference between calculated CTDI<sub>vol</sub> and CTDI<sub>vol</sub> reported by scanner while this value is



**Figure 8.** Variation of dose received at the center of the CTDI Phantom.

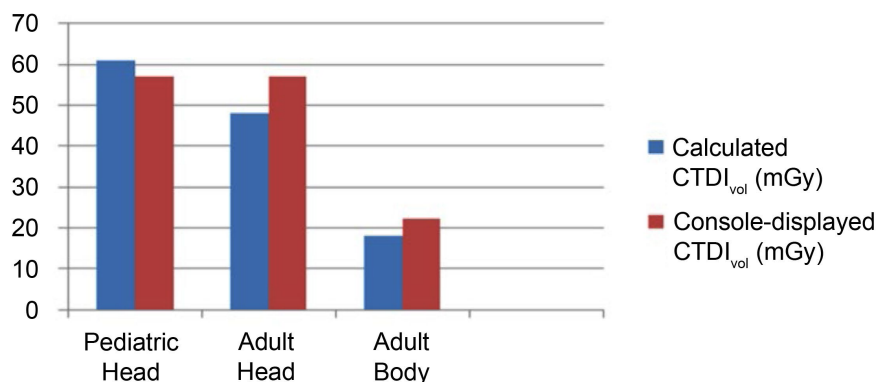


**Figure 9.** Variation of CTDI<sub>vol</sub> with the diameter of cylindrical CTDI Phantom.

not scored as a part of accreditation, the percent difference should be less than 20% [10]. The measured value and console-displayed value shown in **Table 9** have to be comparable and intimate to the reference level, which is within 20%. The comparison of CTDI<sub>vol</sub> of the CTDI Phantom with the Console Value is shown in **Figure 10**.

**Table 9.** Calculated and console-displayed values of CTDI<sub>vol</sub> for the CTDI phantom.

CTDI Phantom	Calculated CTDI <sub>vol</sub> (mGy)	Console-displayed CTDI <sub>vol</sub> (mGy)	Deviation (%)
Pediatric Head	61.04	57.02	7
Adult Head	48.11	57.02	15.62
Adult Body	18.08	22.39	19



**Figure 10.** Comparison of CTDI<sub>vol</sub> of the CTDI Phantom with the console value.

A Quality Assurance Phantom is used to determine whether or not the CT scanner produces consistent image quality. The six aspects of image quality namely contrast scale, high contrast spatial resolution, low contrast detectability, noise and uniformity, slice thickness and laser light accuracy were measured and found to be within acceptable limits.

CTDI measurements were performed in axial mode with an applied voltage of 120 kV and a current of 250 mA. Report of AAPM Task Group 96 and 111 formalism was used for dose estimation [11] [12]. The collimation of the beam was set to 1.25 mm. The CTDI<sub>100</sub> was calculated for different diameters of CTDI Phantom. It was seen that with the decreases in diameter of CTDI Phantom, the value of CTDI<sub>100</sub> increased. The central to peripheral dose ratio of Pediatric Head and Adult Head combinations were similar and found to be (C:P = 1:1) and for the Adult Body combination, it was found to be double (C:P = 1:2).

A study by Francis Hasford, B. V., in 2015 the body (pelvic) scan technique of 120 kV and 100 mAs produced a dose estimate of 20.08 mGy in the Adult body phantom, and for Adult head Phantom it was 44 mGy [9]. In the present study the CTDI<sub>vol</sub> of three Phantoms was measured. For Pediatric Head, Adult Head, and Adult Body Phantom the measured CIDI<sub>vol</sub> was 61.04 mGy, 48.11 mGy, and 18.08 mGy respectively. This study has shown deviations of 7%, 15%, and 19% between estimated and console-displayed doses for Pediatric Head, Adult Head, and Adult Body scan techniques respectively.

#### 4. Conclusions

The image quality tests that were run on the CT scanner environment using QA Phantom ensured that all of the results of image quality tests on the CT scanner met the criteria and that the image quality remained consistent and within acceptable limits. CTDI<sub>100</sub> was measured at different voltages and was found to be directly proportional to the tube voltage. CTDI<sub>vol</sub> was measured and compared to ensure that it fell within the approved tolerance range, which is 20%. In order to comply with regulations, CT systems' acceptance and consistency checks are required to include a measurement of the CT Dose Index (CTDI). In future, better

QA programs should be designed to ensure more consistent image quality for better diagnosis of diseases. The current study can be expanded by using anthropomorphic Phantom to determine whether mis-centering of the Phantom actually increases or decreases the effective dose to the Phantom's centre.

### Conflicts of Interest

The authors declare no conflicts of interest regarding the publication of this paper.

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