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RESEARCH
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Image quality and radiation dose assessment for the clinically applied 16-slice CT scanner using PMMA phantom and quality assurance phantom

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Abstract

The goal of this study was to assess the radiation dose delivered accuracy and the acceptable image quality parameter from the 16-slice Philips CT scanner in order to evaluate the current methodology for quality controls of computed tomography systems. The measured volume computed tomography dose index was 101 mGy and the displayed volume computed tomography dose index was 10.2 mGy and the dose length product is 680 mGy*cm for head scan, and the measured volume computed tomography dose index was 6.1 mGy and the displayed volume computed tomography dose index was 6.3 mGy and the dose length product was 587 mGy*cm for body scan, respectively. The image quality parameters were 4.4 HU, 2.3 HU, and 10 HU for uniformity, contrast-to-noise ratio, and computed tomography number accuracy (CT number accuracy) for quality assurance phantom, respectively. We propose that discussions and collaboration regarding patient and particular equipment adjustments be coordinated among radiologists, medical imaging technologists, and medical physicists. This can increase image quality, reduce absorbed radiation, and improve hospital medical care.

Keywords Computed tomography scanner, Image quality parameter, CT dose index, Quality assurance phantom, PMMA phantom, Medical radiation dose

Background

Computed tomography (CT) scanner is the largest source of medical radiation exposure, while modern computed tomography scanner can produce images with surprisingly low doses and may be the lower-dose alternative to radiographs of the urinary tract, lumbar spine, and pelvis [1]. Computed tomography utilizes ionizing radiation (X-ray), and the radiation dose received by the patient from the multi-slice CT scanner is much greater than that of any other diagnostic X-ray modality. Ionizing radiation use in medical imaging is increasing rapidly, mostly because of the greater use of X-ray computed tomography (CT) scanner [2]. Currently, over half of the ionizing radiation exposure to humans is carried on by medical treatments [3]. In clinical practice, multi-slice CT scanners (2, 16, 32, 64, and 120-slice scanners) produced better image quality but delivered a higher radiation dose than single-slice scanners [4, 5]. The most valuable diagnostic technique used by radiologists today to find disease inside the human body is computed tomography, which promotes a larger dose deposition than traditional radiology exams. It is important for both diagnostic and therapeutic purposes and is one of the most common radiological examinations undertaken worldwide [6, 7]. Diagnostic X-rays contribute to nearly 50% of the total annual collective effective dose of

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radiation from man-made and natural sources to the general population in western countries. Many publications in the fields of radiation protection and science deal with CT scanning dosimeters. Dosimeter testing and image quality testing are two crucial components of evaluating the functionality of a CT scanner after installation and maintenance.

Evaluation of image quality in a computed tomography is important to ensure that diagnostic questions are correctly answered while keeping the radiation dose to the patient as low as is reasonably possible. It is a component of quality control of a medical X-ray device. Image quality and radiation dose are affected by the detector system, the output from the X-ray tube, and the image reconstruction techniques, among other factors. A quality assurance test for CT is necessary to ensure that the system is fulfilling its technical specification [8]. The quality assurance program included the careful selection of CT scanner technical parameters that control patient exposure and measurement of physical CT dose parameters and image quality parameters. The purposes of estimating CT image quality are whether the image is correctly positioned at the adequate image quality with the appropriate radiation dose. To obtain the radiation dose delivered by a CT scanner, as well as to determine an image quality parameter, we used a radiation-dose-measuring technique and an image-processing software package in this study. To address the problem under consideration, we estimate the acceptance level of the image quality parameter and the radiation dose under various conditions of CT scanner parameters, such as X-ray tube current, tube voltage, exposure time, and pitch. We also get the solution. Our objective was to increase image quality and reduce radiation exposure to the patient using a quality assurance phantom and a PMMA CT phantom. This is the first study carried out at S. Chari Hospital, Trento, Italy. Because every previous study was conducted in Europe and Italy, we recommended that image quality tests and radiation dose measurements for CT scans be carried out in other Italian and African countries. These measurements are necessary for any CT scanner to be accepted after installation and during maintenance in order to verify that the image parameters and CT dose index are acceptable. The rest of this paper is organized as follows: Section 2 provides the materials and methods of radiation dose measurement and estimation of image quality parameter of 16-slice CT scanner using dose-measuring technique and image software. In Sect. 3, we present the result and discussion of the study. Finally, Sect. 4 deals with the summery and conclusion.

Material and methods

Using a quality assurance phantom, and a PMMA phantom for standard methods and techniques, a CT scanner was used to measure CT dose descriptors, such as volumetric computed tomography (CTDIvol) and dose length product (DLP), and image quality parameters, such as uniformity, CT number accuracy, and image noise.

Materials

CT scanner

A multi-detector CT scanner (16-slice Philips Big Bore) was investigated in the present study; it is installed at the radiotherapy department of the S. Chiara Hospital, Trento, Italy.

PMMA phantom

PMMA Phantom has been designed to examine CT dose index parameters such as volume computed tomography dose index (CTDIv) and dose length product (DLP). It is made of polymethylmethacrylate (PMMA) solid acrylic material with a 16-cm-diameter head and a 32-cm-diameter body phantom, 100-mm pencil ionization chambers, and dosimeter readout systems (Fig. 1).

Quality assurance phantom

The quality assurance phantom, which is employed globally for image quality assurance testing, was specifically created for the assessment of several image quality parameters in CT. All image quality tests in this investigation were conducted using quality assurance phantoms. It is a cylindrical phantom made up of many test modalities. Slice thickness, CT number accuracy, noise,

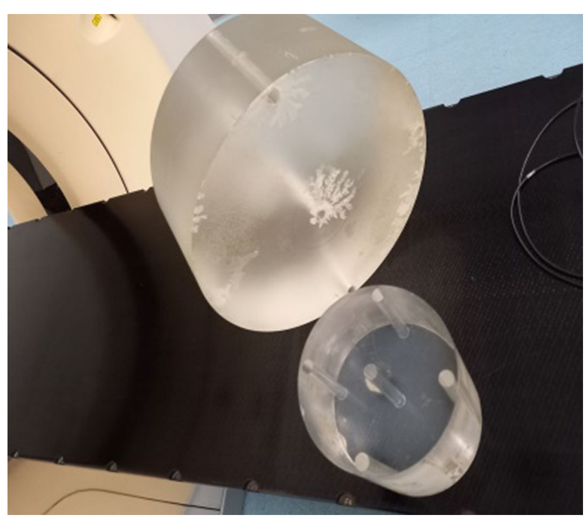

Fig. 1 Setup for CTDI and DLP determination

and uniformity were the parameters used in this study's evaluation of image quality. The American Association of Physics in Medicine recommends using a specific phantom for quality control and to evaluate the operation of the CT scanner. The quality assurance phantom consists of an independent part that can measure the required image quality parameter. It has been designed to examine a wide range of scanner parameters; these include CT number accuracy, uniformity, noise, image slice thickness, and resolution (Fig. 2).

Methods

The amount of energy imported by ionizing radiation to a small mass of material is measured as radiation dose; the most common way to measure radiation is in a pencil ionization chamber the AAPM report 96 formalism was used to estimate CT radiation dose, and the imageJ software package was used to estimate CT image quality parameters at a given scan parameter of tube voltage 120 kV, tube current 100 mAs, and collimation 24 mm. We scanned three images of the quality assurance phantom for the brain protocol and measured and recorded radiation dose for the head and body protocols from September 2021 to December 2021 while the phantom was in place. Axial or sequential CT scanning involves the rotation of the X-ray tube head around the patient or phantom with no table motion during rotation.

Data analysis

The images used for all image quality tests were generated from DICOM files and analyzed using the ImageJ software package to analyze CT phantom images to determine whether the image generated from the CT scanner is acceptable or not. Following image selection, the CT number accuracy, noise, and uniformity were analyzed using imageJ software, and data from the CT console were recorded to estimate the CT dose index, and the measured radiation dose was measured using a PMMA head and body phantom at the phantom's peripheral and central positions.

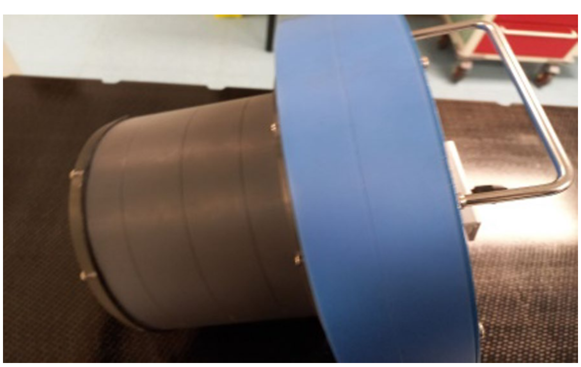

Fig. 2 Setup of CT image quality measurement

Dose measurement

Dosimeter tests for CT scanners are performed in terms of the computed tomography dose index (CTDIv) and dose length product (DLP). Based on examinations from the American Association of Physics in Medicine, this study was conducted (AAPM). It is measured in mGy and is a standardized measure of the radiation dose output of a CT scanner. The CTDI corresponds to the total energy deposited in a patient or phantom. The radiation dose of a CT scanner can be measured using PMMA head and body phantoms by a particular imaging protocol described by CTDIv and is influenced by many technical factors, including tube potential (kvp), tube current (mA), exposure time, the X-ray beam, collimation, and pitch. The CT dose index at the center and periphery of a PMMA head and body phantom was measured using a weighted computed tomography dose index (CTDIw), which can be expressed as [9].

$$\mathrm{CTDIw} = \frac{1}{3}\mathrm{CTDIc} + \frac{2}{3}\mathrm{CTDIp} \tag{1}$$

where CTDIc = CTDI at the central position of the PMMA phantom; CTDIp = the CTDI averaged over the four peripheral positions of the PMMA phantom.

For helical CT examinations, the parameter estimating the dose in a patient slice is the CTDIvol [10]. It is defined as

$$\mathrm{CTDIvol} = \frac{\mathrm{CTDIw}}{p_{\mathrm{f}}} \tag{2}$$

where p_{f} = the pitch factor [9, 11]. CIDTvol = the volume computed tomography dose index.

$$\mathrm{DLP} = \mathrm{CTDIvol} * L \tag{3}$$

where L is the total scan length

Image quality parameter measurement

Physical parameters like CT number accuracy, uniformity, noise, special resolution, and low contrast resolution can be used to describe the image quality of a CT scanner.

According to the technique shown in Table 1, the quality assurance phantom was used to test uniformity, image noise, and CT number accuracy.

There are cylinders of various materials, such as acrylic, air, bone, water, and polyethylene, to measure the accuracy of the CT number. The primary CT number accuracy for each material allows you to choose a region of interest (ROI) location in Fig. 3.

Uniformity

Five 8 mm.[2] regions of interest (ROI) were placed in the image, one in the center and four on the periphery, at angles of 0°, 90°, 180°, and 270° clockwise, to measure uniformity. The uniformity was calculated from the standard deviation of the five different regions of interest (ROI) in the phantom image in Fig. 3. It is the mean value for every ROI divided by the mean value of the area covering the whole phantom image. The CT number for all five ROIs must be within this range (5HU) of the center ROI mean value [12]. The uniformity value was defined as Eq. (4) [8]

$$\Delta \text{HU} = \text{HUmax} - \text{HUmin} \tag{4}$$

where HUmax is the mean HU value in the ROI with the highest average value and HUmin is the mean HU value in the ROI with the lowest average value of the five ROI.

Nose

It is one of the CT scanner's performance indicators that inversely relates to both slice thickness and the square root of the dose [13]. Image signal and image noise are key parameters in image quality assessment. In an ideal case, image signal is directly linked to the detected number of photons, while image noise is the pixel's stochastic fluctuation around its mean value. The ratio of these two quantities yields the signal-to-noise ratio. The contrast-to-noise ratio (C/N) was employed to evaluate the signal level in the presence of noise in Fig. 4, and it is computed by dividing the average gray scales of a region of interest (ROI) minus the background region of interest (ROI) by the background region of interest (ROI) [9]. The measured image noise deviation from baselines established by acceptance tests should not above 10% [14].

$$\text{CNR} = \frac{x_{\text{s}-x_{\text{bg}}}}{\sigma_{\text{bg}}} \tag{5}$$

Table 1 Acquisition parameters used for measurement of uniformity, noise, and CT number accuracy

Technique	Helical (axial)
Kvp	120
mAs	150
Rotation time	1000 mAs
Pitch (helical scan)	0.98

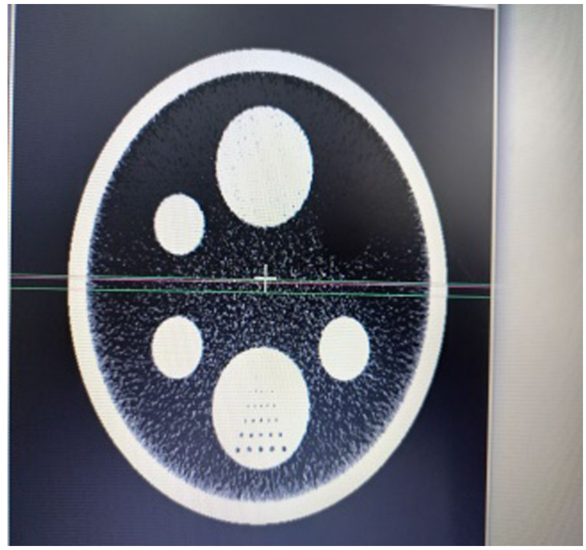

Fig. 3 The measured CT number of different materials in the quality assurance phantom brain axial protocol

where CNR = the contrast-to-noise ratio; X_{s} = the average gray scale of the interested region of interest; X_{bg} = the region of interest in the background; σ_{bg} = is the standards deviation of the background.

Resolution

High contrast (high contrast resolution), low contrast, and high contrast High contrast resolution determines the minimal size of detail that can be observed in the plane of the slice with a contrast more than or equal to 10, whereas low contrast resolution is greatly impacted by the accompanying image noise.

Results

In this result, the CT radiation dose parameter was measured using an axial scan of the head and body PMMA phantoms. The measured volume computed tomography

Fig. 4 The contrast-to-noise ratio (CRN) measured into the selected region of interest (ROI)

dose index (CTDIvol) was 10 mGy and the DLP was 680 mGy*cm for the head scan technique, whereas the volume computed tomography dose index (CTDIvol) in the body scan technique was calculated at 6.1 mGy and the DLP at 587 mGy*cm, respectively. The radiation dose delivered by the multi-slice computed tomography (CT) scanner matched the relevant values in Table 2 with good accuracy, and the DRLs values for head and body CT PMMA phantom measurement were lower than the selected internationally accepted DRLs in Table 3.

The choice of X-ray tube voltage and tube current affected the quality of the CT image as well as the radiation dose received by the patient. Using quality assurance phantom to establish the CT image quality parameters, tests were carried out on the image's resolution, noise, linearity, and CT number accuracy.

CT number accuracy

The CT number, which defines the X-ray linear attenuation coefficient, is the normalized value of a pixel in a CT image. It is 0 for pure water, −1000 for air, and 900 for bone [15]. The measured CT numbers of air, water, bone, polyethylene, and acrylic were in good agreement with the CT numbers in the actual values. We compared the measured and actual values of the CT number accuracy of these materials in the brain protocol at 120 kV and 500 mAs, as shown in Table 4. A homogeneous object's mean CT number should differ by no more than 8HU between its peripheral and central regions. The image quality parameter was obtained for the CT number accuracy of bone value, which was not accepted for the tolerance values for the ACR phantom [12]. In these studies, the CT number accuracy of bone, air, water, polyethylene, and acrylic were accepted as the reference values for quality assurance phantoms. Air, acrylic, bone, polyethylene, and water all have CT number accuracy of -998HU, 124HU, 846HU, -91HU, and 0HU, respectively [12].

Noise measurement

Noise is the standard deviation of CT number (HU) in the selected region of interest in a medium. The contrast-to-noise ratio (CNR) is used in Fig. 4 to evaluate the signal level in the presence of noise. It is computed by dividing the average grayscale of a region of interest (ROI) by the background region of interest (ROI) [16]. Using Eq. (4), the CT number of grayscales of the selected region of interest is equal to 63.9HU, the CT number of the background region of interest is equal to 54.7HU, and the standard deviation (SD) of the background is equal to 4HU. Then the contrast-to-noise ratio of the image was equal to 2.30 HU, so the noise measurement was accepted. The image noise value of cylindrical PMMA chest phantom is 2.47% [17]. The standard variation for image noise in the center of a water phantom is ± 3HU [18]. For adult head, pediatric head, and adult abdomen protocols, the contrast-to-noise ratio (CNR) must be greater than one, and for pediatric abdomen protocols, the contrast-to-noise ratio (CNR) must be more than 0.5 HU [12, 19].

Uniformity measurement

The uniformity is calculated as the mean value of the standard deviation (SD) of a peripheral and a central region of interest (ROI). The SD in the peripheral and central area were 3.7HU, 3.5HU, 7.3HU, 3.7HU, and 4HU, respectively. The mean values of these five regions

Table 2 CTDIvol and DLP test result for head and body PMMA CT phantom at 120kv and 100mAs

Body region	PC readings at different phantom locations (mGy)					CTDIvol (mGy)		DLP (mGy*cm)
	Center	3	6	9	12	Calculated	Displayed	
Head	2.29	2.48	2.31	2.46	2.60	10	10.2	680
Body	0.77	1.81	1.48	1.82	1.65	6.1	6.31	587

Table 3 Compare the DRLs of the head and body CT phantom with the corresponding other CT phantom study international

Head scan	CTDIvol (mGy)	DLP (mGy*cm)	Body scan	CTDIv	DLP
This study	101	680	This study	6.1	587
UK (2003)	56	690	UK (2003)	14	–
UK (2011)	58	890	ACR (2008)	25	–
India (2014)	32	875	Norway (2018)	13	–
Ireland (2012)	64	857	Sweden (2019)	12	–
Sudan (2011)	65	758	Europe (2004)	15	–

Table 4 Comparison of the DRLs of the head and body CT phantom with the corresponding data for real patients study international

Study	Year	Country	Design	Dose index	Head (DRLs)	Body (DRLs)
This study	2023	Italy	CT phantom	CTDIvol (mGy)	10	6.1
				DLP (mGy*cm)	680	587
Japan	2019	Japan	Human	CTDIvol (mGy)	80	–
				DLP (mGy*cm)	1500	–
Slovenia	2020	Slovenia	Human	CTDIvol (mGy*cm)	76.9	18.7
				DLP (mGy*cm)	1942	1116.2
Egypt	2017	Egypt	Human	CTDIvol (mGy)	30	31
				DLP (mGy*cm)	1360	1425
Canada	2016	Canada	Human	CTDIvol (mGy)	82	–
				DLP (mGy*cm)	1302	–
South India	2018	South India	Human	CTDIvol (mGy)	68	–
				DLP (mGy*cm)	1120	509

of interest were 4.4 HU, so the measured uniformity was accepted.

Discussion

The goals of this study were to assess the characteristics of image quality using a quality assurance phantom and head and body PMMA phantoms, as well as to assess CT dose indexes such as CTDIvol and DLP. The safety of CT equipment was tested using two different techniques dosimetric evaluation performed first, while image quality evaluation came in second. The noise, uniformity characteristics of the image quality and CT number accuracy of a material were accepted to the tolerance values for quality assurance CT phantom and to the measured CT dose index parameter. However, the volume computed tomography dose indexes (CTDIvol) and dose length product (DLP) were also accepted to the standard limits in Table 3. The values of the image quality parameter and radiation dose obtained by a 16-slice computed tomography scanner in this study were acceptable. The image quality parameter was obtained for the CT number accuracy of bone value, which was accepted to the tolerance values using the quality assurance phantom compared to the American College of Radiology (ACR) phantom in the literature [15]. Image quality test and radiation dose estimation are very important test in acceptance of any CT scanner after installation and maintenance approve that the image parameters are acceptable by using quality assurance phantom (Table 5).

Table 5 The measured CT number accuracy of different material in quality assurance phantom axial brain protocol at 120kv and 500mAs

Material	Actual (HU)	Measured CT number (HU)	SD (HU)	Result
Air	−1000	−66.7	3.5	Pass
Bone	900	936	7.3	Pass
Water	0	3.2	4	Pass
Acrylic	120	138	3.7	Pass
Polyethylene	−95	108	3.7	Pass

Conclusions

A computed tomography scanner uses ionizing radiation in comparison with other imaging modalities and produces images with a higher radiation exposure. For any CT scanner to remain safe, radiation dose and image quality assessments were essential. If the images generated accurately reflect the attenuation values of the X-ray beam caused by the body tissue as shown on the CT scanner, that is a sign of good image quality. All image quality parameters were acceptable because they were within the tolerance range, and also the measured CT radiation dose for a 16-slice CT scanner was less than the chosen international dose reference level. We advise that testing image quality parameters and radiation dose for CT scanners plays a crucial role in getting the optimal performance of the CT system and also helps each radiological department build a quality control and quality assurance program.

Abbreviations

CT	Computed tomography
PMMA	polymetaylenmetaAcrylate
CTDI	Computed tomography dose index
SD	Standard deviation
ROI	Region of interest
CNR	Contrast to noise
ACR	American College of Radiology
DLP	Dose length product
AAPM	American Association of Physics in Medicine

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Declarations

Ethics approval and consent to participate
Not applicable.

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Competing interests
The authors declare that they have no competing interests.

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