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EVALUATION OF CT DOSE QUANTITIES INFLUENCED BY PATIENT SIZE USING THE INDIGENOUS EXTENDED-SIZE CT PHANTOMS

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Abstract

Objective: This work evaluates the CT dose quantities influenced by patient-size validating the indigenous extended-size head and body phantoms. This study also reports the size-specific dose estimation and the effective dose to access optimized patient radiation dose.

Methods: Patient-specific data was collected comprising brain and abdomen CT examinations. Two ion-chambers; farmer-type and Pencil-type ion-chamber, were used for dose measurements. Two customized phantoms with a diameter of 20 cm and 35 cm were developed as CT head and body phantom respectively, and the central and peripheral doses were investigated. Dosimetric quantities like $CTDI_{vol}$ and DLP were calculated and were compared with standard CT protocols. Size specific dose estimates (SSDEs) and effective doses (E_{eff}) were also calculated.

Results: For free-in-air dose measurements, using 20 cm head phantom, the relative difference between Farmer and Pencil ion-chamber, was found to be 2.5 % at the center and 5.8 % at the peripheral region. A percentage difference between central and peripheral dose values was observed at 23% for the Pencil ion-chamber and 26% for the Farmer-type ion-chamber. Whereas, in-phantom measurements using head and body phantom, $CTDI_{vol}$ values ranged from 22.1 – 71.3 and 4.4 – 36.7 (mGy), whereas the DLP values had a range of 286 – 1125 and 55 – 804 (mGy.cm) for the brain and abdomen examinations, respectively. The average for $CTDI_{vol}$ was noted as 49.5 and 16.6 (mGy) and for DLP it was noted as 717 and 411 (mGy.cm) respectively for both examinations. Size-specific dose estimation for the $CTDI_{vol}$, showed a percent difference of 11.7% and 2.3% respectively. The effective dose influenced by patient size was observed to be tolerable at 1.3 mSv and 7.6 mSv, respectively, for head and body examinations.

Conclusion: Dose quantities are comparable to European DRLs for determining the effective dose-to-organ as an attribute of radiation exposure when a patient undergoes CT examinations. The measurements obtained can be used as baseline data to help forecast malfunctioned output of the unit in the future. This study contributes to an experimental validation of the existing knowledge based only on the MC study. The result also signifies that the x-ray radiation from CT can also be quantified using a Farmer-type chamber.

Keywords: CT Simulator, CT Phantom, CTDI, Size-specific dose estimates, Effective dose

1. Introduction

In the early 1970s, a very important diagnostic machine in medical science was launched as Computed Tomography (CT). The prevalent use of CT demands precise dose assessment. Historical data shows that 40% of the resulting dose was due to the contribution of exposure to X-rays in diagnostic cases in the UK in 1997 [1]. As an estimate, about 70% of the total dose when a patient undergoes x-ray examinations is contributed by CT. Computed tomography, as an imaging modality, is associated with a comparatively more effective dose for patients than conventional diagnostic techniques [2]. Radiation dose from the x-rays of CT is characterized by measuring the dose quantity called CTDI - the CT dose index [3]. The CTDI was originally defined for "Axial" scan mode for a scanner with only one detector-row, whereas now, with new generation CTs, all scanning is done in "Helical" mode, taking several image slices at a time. AAMP report no. 96 [3] has been published, which deals with the methods of dosimetry measurements, reporting, and managing the radiation dose in computed tomography.

CT pencil ion-chambers used for measuring the dosimetric parameters must be calibrated in terms of air kerma-length product (KLP) under specific procedures. A typical pencil ion-chamber with an active length of 10 cm long, marketed as DCT10, has a nominal calibration factor, $N_{D,K}$ (nominal) = 70 mGy.cm/nC in terms of air-kerma at STP. The international dosimetric protocols permit dosimetry free-in-air and then convert the result to a dose to a phantom. Moreover, chamber values measured free-in-air are forthright, allowing the direct comparison of results. As a comparison of measurements, defined practice resembles relative dosimetry [4]. Following this perspective, a comparison was made using measurements in-air for the Pencil ion-chamber and Farmer ion-chamber, with two protocols; IAEA TRS-457 [5] and AAPM Report-111 [6].

In radiology and CT dose estimation, Diagnostic Reference Levels (DRLs) are used as guidelines to help identify if the radiation dose in a routine scan is unusually high or low. They're not strict limits but reference points based on typical doses for standard-sized patients. DRLs help healthcare providers spot outliers and adjust their practices to ensure patients get the safest, most effective care. For CT scans, the most common dose measurements to establish the DRLs are $CTDI_{vol}$ and DLP. Recent CT systems display the $CTDI_{vol}$ and DLP dose indices before & after the CT scan exam to perform. This display indicates information concerning only the radiation output and does not estimate the dose on a size basis, as $CTDI_{vol}$ is determined by using only a standard size CTDI phantom [7]. Mostly, the dia of phantom used is also displayed on the console along with the above two indices or can be found in the DICOM dose-report. The dose quantity' $CTDI_{vol}$, which is a distinct metric of the radiation beam output of a CT scanning system, signifies the mean radiation dose to a 15 cm long cylindrical phantom for a multi-scan exam with patient couch incremented between CT scans. This issue was addressed in AAPM TG-204 [8] by developing a new dose estimation index, "Size-Specific Dose Estimate (SSDE)". A number of studies have established that SSDE values can estimate the radiation dose to patients more accurately [9, 10]. The peripheral $CTDI_{100}$ may lead to an overestimate of the surface dose by up to a factor of 2. Peripheral $CTDI_{100}$ values are rational estimates to assess the surface doses for sequential and/or spiral CT exams [11].

Recently, the distribution of CT dose in phantoms of physical diameter ranging from 8 to 40 cm was inspected by Choirul Anam [12]. The author explored that for the phantoms with water equivalent dia > 14 cm, the dose at the central position of the phantom was observed to be less than the dose measured at the periphery. The relative difference in dose measurements between the central and the peripheral positions was higher for large size phantom. Dose distribution is also affected by tube-voltage (kV). The difference decreases when phantoms with large dia are used for high kV. An exponential decrease in size-specific weighted dose values ($D_{s,w}$) was reported over an increasing water equivalent dia.

The influence of patient-size on estimating the CT dose was studied with the dose quantity $CTDI_w$. The Monte Carlo simulation based study for extended head and body phantoms with cylindrical dia ranging from 8 to 40 cm, was presented by Haba T. et al. [13, 14]. The author showed that the $CTDI_w$ equation with weighting factor at the center, W_c = 3/8, and at the periphery, W_p = 5/8, can provide a more precise average dose as compared to those suggested by conventional (initially proposed by Leitz et al.) and Bakalyar's equation [15]. The result showed that the maximum % difference between the values of average dose and $CTDI_w$ for the conventional, Bakalyar, and Haba's weighting-factors was +16%, -12% and -6% respectively. Moreover, it was hypothetically proposed that the size correction factors, g(d) calculated for the weighting-factors were "in good" agreement and reported as a 17% difference compared to the values in AAPM TG-204 [8]. The current work provides an experimental basis for evaluating the x-ray dose quantities associated with the patient's physical size in a CT unit using two indigenous phantoms, which resemble the geometrical extensions of standard CT dosimetry phantoms.

2. Materials and Methods

2.1................ Radiation detectors and Phantoms

A 16 slicer Toshiba CT Simulator (Model: Aquilion LB) with image acquisition of helical and axial was used in this study. Two ion-chambers, first one- a cylindrical pencil-shaped CATO ionization-chamber (Type DCT10, Sr. No. CTDI-0853) with active volume 4.9 cm^3, active length 10 cm, and outer dia 9 mm was used as detector connected via chamber adopter to an electrometer (RTI Group AB, Sweden). The Pencil ion-chamber with a calibration factor of 74.6 x 10^6 Gy.cm/C (8512.1 x 10^6 R.cm/C), was capable of measuring radiation quality ranging from 80 kV to 150 kV and was developed for CT dose measurements. The second one- a Farmer type ion-chamber (Sr. No. 30013, PTW Freiburg GmbH) with an electrometer (PTW UNIDOS, Sr. No.1005) was used to replicate the same setup for measurements. Measurements were obtained free-in-air as well as using two CT dosimetry phantoms. Though, the standard dosimetry phantoms were not provided with the newly installed CT system by the vendor, the inadequacy of standard head/body phantoms for practical setup is accomplished using relatively large-sized phantoms customized locally. At first, a solid PMMA cylindrical afterloading calibration phantom (Krieger T9193) available with a 20 cm diameter and 12 cm length, shown in Figure-1, was used to mimic the geometry of a standard head phantom (dia 16 cm and length 15 cm) by customizing its inserts with a cavity similar to the geometry of both the Pencil ion-chamber and the Farmer ion-chamber. Secondly, a hollow acrylic cylinder (dia = 35 cm and length = 45 cm), shown in Figure-2, simulating the geometry of the adult body, was built with two acrylic inserts; one at the center and the other at the periphery of the circle at 2 cm depth. The inserts are open at both ends, permitting the ion-chamber to travel freely for the ease of required setup. The cylinder may be designed with four peripheral inserts but in this study having only one peripheral inserts, the phantom is rotated to position the insert at desired position/angle. The cylinder was filled with water. Phantom size correction factors were used to tolerate the accuracy of measurements comparable to those of conventional head and body phantoms of standard size.

Figure-1: Solid PMMA cylindrical phantom (Krieger T9193) used as a CT head phantom.

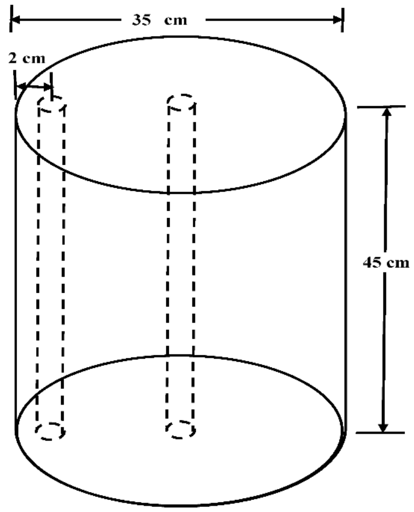

Figure-2: Diagram of cylindrical water phantom (CT body phantom) customized in-house.

2.2................ Scan details and Dose measurements

Two clinical protocols within the typical scan range used for routine CT examinations, namely the brain and the abdomen, were selected in the current study. Data from CT exam parameters like kVp, mA or mAs, number of slices, slice thickness, pitch, and couch increment were taken into consideration. This study was not a clinical investigation based on the physical participation of humans or animals, therefore, neither the consent from the patients nor the approval from the ethical committee was required. Measurements were performed by setting the head & body phantoms in succession. At first, the ion-chamber connected to the UNIDOS electrometer was placed in the central hole of the phantom and aligned to the isocentre [16]. Setup accuracy and phantom alignment was ensured by acquiring the single scanogram (or scot view) before measurements. A scanogram was used to select the volume or slice to be imaged. The Brain protocol was selected and scanning was performed in axial mode with 120 kV and 200 mAs. Charge measurements were taken by changing the position of the ion-chamber at the center and periphery, and corrected for temperature and pressure. The procedure was repeated to obtain measurements for the body phantom using the abdomen protocol. Dosimetric quantities were estimated using equations, summarized in Table-1. The values of $CTDI_{vol}$ and DLP displayed on the console, incorporated by scanner software at the completion of each scan, were recorded for further comparison with the values obtained experimentally.

A. In-Air measurements

In-air measurements were performed using head phantom only. The In-air measurement using the body phantom are not included in this study. The above Figure-1 shows the measurement setup by head phantom, with the detector placed at the axis of tube rotation and then the left upper/lower and right upper/lower lateral sides 8 cm off the center making a 45^0 angle above the frontal plane. The two ion-chambers; Farmer-type and Pencil-type, were irradiated free-in-air with the same scan conditions derived from AAPM Report-111 [6]. Scan length was selected 8 cm, such as 4 cm along each side of the center of the ion-chamber. Scanning was performed in helical scanning mode with 120kV, 200 mA, tube-rotation time 1 second, slice thickness 10 mm (1 cm), slice interval 10 mm and pitch factor, PF = 1. The longitudinal beam width (range or scan length) was taken at 80 mm and the FOV was selected to be in the head field. The measuring setup was designed to mimic the standard geometry of a phantom to reproduce the same position of the ion-chamber. The in-air measurements at positions simulating the center and periphery of the phantom were performed to provide a direct comparison between dose distribution in-phantom and in-air under similar geometric conditions. These measurements help in understanding the variation in radiation exposure at different locations within the phantom and provide a reference for normalizing CTDI calculations. While in-air measurements at the isocentre along the central axis are commonly used for standardization, our approach was intended to better approximate the dose deposition pattern within the phantom by mimicking the locations where the in-phantom detectors are placed. This method allows a more relevant comparison when determining weighted CTDI ($CTDI_w$) and volumetric CTDI ($CTDI_{vol}$).

B. In-Phantom measurements for dose quantities

In-Phantom measurement were performed using both head & body phantoms. Two protocols, AAPM TG-111 [6] and TG-233 [17], were considered for the methodology of dosimetric measurements. Scan conditions are followed as used clinically in practice as well as provided by the manufacturer to obtain optimized measurements in determination of CTDI for a single axial scan and DLP in a complete examination. The scan protocol used in routine examinations was 120 kVp, 200 mAs, 10 mm for the brain and 120 kVp, 300 mAs, 10 mm for the abdomen region.

For $CTDI_{100}$ measurements, phantoms were aligned with their axis at isocenter clinically. The head phantom was positioned in a holder while the body phantom was placed directly on the couch. An ion-chamber was inserted into the central hole in the phantom and connected with the electrometer. Other peripheral holes were filled with acrylic plugs/inserts. The ion-chamber was then positioned at each peripheral position (12, 3, 6 and 9'o clock) and measurements were obtained. The obtained values of $CTDI_{100}$ were then used to calculate more dose quantities; $CTDI_w$, $CTDI_{vol}$, DLP, E_{eff} and SSDEs. Furthermore, all measured and calculated values were compared with the corresponding dose quantities displayed on the scanner console.

Table-1: A description of dose quantities expressed for the quantification of CT dose.

Symbol	Dose Quantity	Equations
CTDI	CT Dose Index (dose descriptor)	General dose description for CT. $CTDI_{100} = \frac{1}{NT}\int_{z=-50mm}^{+50mm} D(z)\,dz$ (Air-Kerma, in mGy)
MSAD	Multiple Scan Average Dose	As CTDI but corrected for pitch. Average dose from a series of scans over an interval of length (l). $\text{MSAD} = \frac{1}{Pitch} \text{ x CTDI} = \frac{NT}{l}\text{CTDI}$ Where N is the number of scans, T is the nominal scan width (mm), l is the distance between scans (mm) and N x T is the total nominal scan/beam width.
$CTDI_{100}$	CTDI (100)	Current definition of CTDI. $CTDI_{100} = \frac{Rdg\ x\ C_{T,P}\ x\ K_{el}\ x\ N_x\ x\ f_{med}}{N\ x\ T}\ x\ 1000\ mm$ Or $CTDI_{100} = \frac{X(rad)\ x\ C_f\ x\ \ L(mm)}{N\ x\ T\ (mm)}$ (for charge measurements) where $X(rad) = \frac{Q}{m_{air}}\left(in\frac{C}{kg}\right) = \frac{Q}{m_{air}}.\frac{1}{2.58\ x\ 10^{-4}}\ (in\ R).f_{med}$
$CTDI_w$	Weighted CTDI	Main descriptor of local dose, proposed for single axial rotation. $CTDI_w = W_C \cdot CTDI_{100,C} + W_P \cdot CTDI_{100,P}$ (1-General Form Eq.) $CTDI_w = 1/3 \cdot CTDI_{100,C} + 2/3 \cdot CTDI_{100,P}$ (2- Conventional/Leitz et al.) $CTDI_w = 1/2 \cdot CTDI_{100,C} + 1/2 \cdot CTDI_{100,P}$ (3- Bakalyar et.al.) $CTDI_w = 3/8 \cdot CTDI_{100,C} + 5/8 \cdot CTDI_{100,P}$ (4- Haba et al.) Where C and P are abbreviated for measurement at the center and the periphery respectively.
$CTDI_{vol}$	Volume CTDI	As $CTDI_w$ but corrected for pitch in Spiral/Helical mode.

		$$\text{CTDI}_{\text{vol}} = \frac{\text{CTDI}_{\text{w}}}{\text{pitch}} = \frac{NT}{l}\,\text{CTDI}_{\text{w}}$$ Where pitch is the ratio of couch increment per rotation and beam width. (Pitch = Table travel per rotation/Beam Width=l/NT)
DLP	Dose Length Product	Includes the irradiated volume and represents the overall exposure for an examination. $\text{DLP} = \text{CTDI}_{\text{w}} \times N \times T$ or $\text{DLP} = \text{CTDI}_{\text{vol}} \times L$ (Helical Mode Scanning, L: Scan Length)
E_{eff}	Effective Dose	Gives the total radiation dose received to Organ in CT exposure. $E = \text{DLP} \times \text{k-factor}$ Where k is the conversion factor (C.F).
SSDE	Size-Specific Dose Estimates	$\text{SSDE} = \text{CTDI}_{\text{vol (32cm)}} \times g(d)$ Where $g(d) = A_0 \times \exp(B_0 \times d)$ represents the Monte Carlo (MC) based phantom Size Correction Factor (SCF). It is a function based on the patients' effective diameter d. A_0 and B_0 are the exponential regression coefficients for CTDI_{100} (Axial) dosimetry model through each weighting-factor of CTDI_{w} equation. SCF depends on the Water Equivalent Diameter (WED) or Lateral Dimension of the patient.

3. Results

In-air measurements using the head phantom, were obtained in positions that corresponded to the holes of a real head phantom. For measurement, the central hole position was tested for dose at the center whereas, the other four hole positions, each 8 cm off the center at 45^0 angle (5.7 cm above the frontal plane, and 5.7 cm left & right of the sagittal plane) were tested for peripheral dose measurements. The result shown in Table-2 indicates that the dose quantity for both ion-chambers is higher at the center than at the peripheral positions (i.e. $D_c > D_p$) using a phantom geometry of 20 cm in diameter. Compared with the literature stated above that, the central dose (D_c) measurement values can be higher and overestimated than skin or peripheral dose (D_p) values by a factor of up to 2. This means that the central dose can increase twice as much as the peripheral dose or generate a difference of up to 100% between these two positions. In the current study, the dose measured at the centre was 23% and at the periphery it was 26% (a factor of maximum 1/4 of central dose). Comparing the measurements of two chambers at a specific position of phantom geometry, the results were almost similar, showing a relative difference of 2.5% at the center and 5.8% at the peripheral boundaries. These values resemble those of the $\text{CTDI}_{\text{in-air}}$ model. Remarkable measurements are noticed at two upper positions and the two lower positions with identical values in the case of the Farmer ion-chamber, which conforms to no relative uncertainty in measuring setup. We found the Farmer ion-chamber to have stable and acceptable measurements, without the influence of FOV for head exams. The IAEA protocol, in its code of clinical measurement practice, considers uncertainties in measurements of 6.3% (1SD) acceptable in diagnostic radiology. Furthermore, the radiation dose quantification using Farmer ion-chamber concludes that the maximum relative error of 5.8% at peripheral sites is well within the tolerance for clinical measurements.

In-phantom measurements using the head & body phantoms, are obtained for the calculations of CTDI dose quantities. The result of CTDI measurements are shown in Table-3. The CTDI_{vol} calculated in the current study is comparable with the literature as shown in Figure-3. The values of CTDI_{vol} and DLP measured experimentally are compared with the values displayed on the console, as shown in Table-4. The DLP values acquired for the brain and abdominal (or abdomino-pelvis) exams, are in the range of 286-1125 and 55-804 mGy.cm respectively. The values of CTDI_{vol} calculated are widely different (upto 4 times in the brain exam), ranging from 22.1-71.3 and 4.4-36.7 mGy for the head and body phantom measurements respectively. Patient dose quantities are significantly affected by technical factors, such as scan parameters and the size of the phantom or patient-size.

Size correction factors, g (d) (also called the conversion factor for $CTDI_{vol}$) for each weighting-factor in the CTDIw equation, are shown in Table-5. The values of the conversion factor in TG-204 for effective dia 16 cm and 32 cm are based on the use of PMMA cylindrical phantom for $CTDI_{vol}$. The results of SSDEs (Table-6) estimated with TG-111 methods performed with weighting factors of 3/8 and 5/8 for the $CTDI_w$ equation were reasonably validated for the extended size phantoms. The effective dose (E_{eff}) with a product of DLP and conversion factor (k) (also expressed as E_{eff}/DLP in units of µSv/mGy-cm) providing the total radiation dose received to an organ exposed to x-rays from a CT beam, was determined for a 20 cm head and a 35 cm body phantom. The results of E_{eff} are shown in Table-7.

Table-2: Free-in-air measurements using two different ion-chambers at central and peripheral positions simulating the head phantom-geometry.

Phantom (geometry simulated as measurement positions)	Detector	In-Air meas. (mGy)		Percent Difference b/w D_c and D_p (%)	D_c values normalized to mAs (mGy/mAs)
		Dose at Center D_c	Dose at Periphery D_p(8 cm, 45^0)		
Head Phantom d=20 cm	Pencil Ion-Chamber	52.3	40.1	23	0.262
	Farmer Ion-chamber	51.0	37.9	26	0.255
	Relative difference (Chamber) (%)	-2.5	-5.8	---	---

Table-3: Result of CTDI measured in-phantom.

Phantom	Chamber Position	$CTDI_{100}$ (mGy)	$CTDI_w$ (mGy)	$CTDI_{vol}$ (calculated) (mGy)	$CTDI_{vol}$ (console) (mGy)	Percent Difference (%)
Head	Center (c) Periphery (p)	37.33 49.21	45.3	45.3	47.1	3.9
Body	Center (c) Periphery (p)	17.31 21.88	20.4	20.4	19.5	-4.4

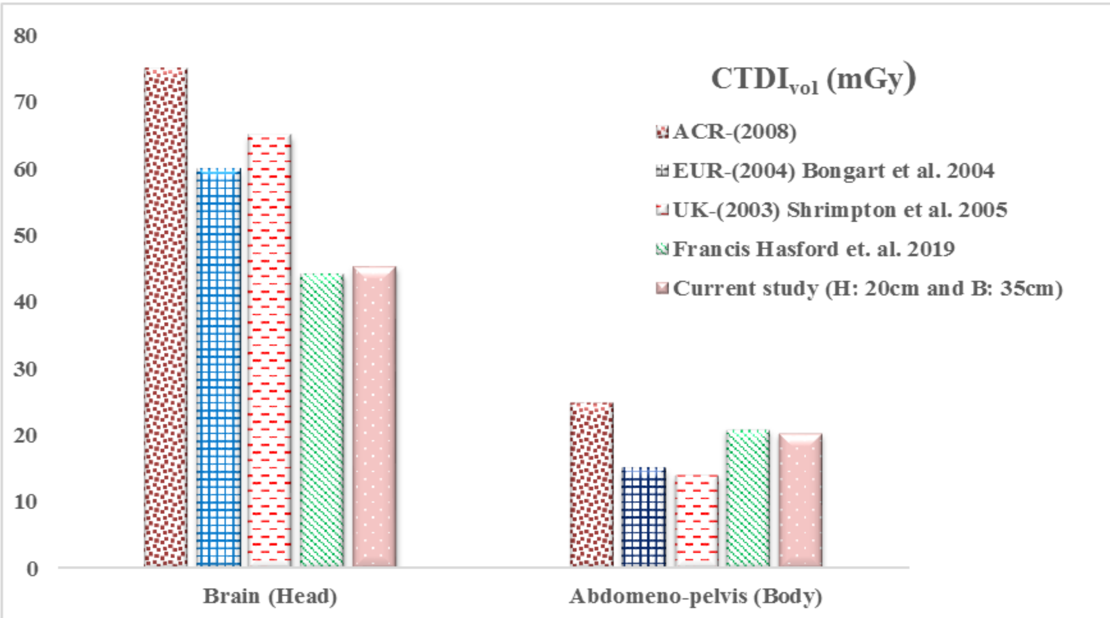

Figure-3: A comparison of $CTDI_{vol}$ calculated in current study with the literature values.

Table-4: Comparison of calculated $CTDI_{vol}$ and DLP values with the values displayed on console.

Phantom (Protocol) (different scan parameters)	$CTDI_{vol}$ (console) mGy	DLP (console) mGy-cm	$CTDI_{vol}$ (calculated) mGy	DLP (calculated) mGy-cm
	25.2	270	22.1	286
Head	47.1	482	45.3	622
(Brain)	60.0	823	59.2	834
	75.3	1040	71.3	1125
Mean	51.9	654	49.5	717
Range	25.2 - 75.3	270 - 1040	22.1 - 71.3	286 - 1125
% Diff. DLP				-9.6
	7.1	38	4.4	55
Body	10.0	239	4.9	279
(Abdomino-Pelvis)	19.5	468	20.4	505
	25.1	765	36.7	804
Mean	15.4	378	16.6	411
Range	7.1 - 25.1	38 - 765	4.4 - 36.7	55 - 804
% Diff. DLP				-8.8

Table-5: Size-Correction factors, g(d) or the Conversion factors for $CTDI_{vol}$ (Console Displayed) selected from guidelines, used to correct the $CTDI_{vol}$ for patient (or phantom) size.

Phantom dia (d) (in cm)	Size-correction factors (SCF), g(d)							
	TG-111 (Helical) *			TG-111 (Volume)*			TG-204**	
	Conv. (1/3, 2/3)	Bakalyar (1/2, 1/2)	Haba (3/8, 5/8)	Conv. (1/3, 2/3)	Bakalyar (1/2, 1/2)	Haba (3/8, 5/8)	(based on 16cm PMMA phantom)	(based on 32cm PMMA phantom)
16	2.07	2.20	2.09	2.12	2.29	2.17	1.01	2.06
20	1.78	1.87	1.79	1.81	1.92	1.85	0.86	1.78
32	1.13	1.14	1.12	1.14	1.13	1.14	0.54	1.14
35	1.00	1.01	1.00	1.01	0.99	1.01	0.48	1.02

***** MC (water model) based size-correction factors for each weighting factor in $CTDI_w$ equation (Refer to Haba-2020).
******Conversion factors are based using 16 cm and 32 cm PMMA cylindrical phantoms for CTDIvol (Refer to TG-204).

Table-6: Size-specific dose estimation (in mGy) to account for the influence of patient size on $CTDI_{vol}$ during the two CT examinations.

Phantom (Protocol)	$CTDI_{vol}$ (console) mGy H:16, B:32	$CTDI_{vol}$ corrected for patient size (or SSDE) mGy	$CTDI_{vol}$ (calculated) mGy H:20, B:35	Percent Difference (Corr. Vs Cal.) (%)
Head (Brain)	47.1	40.5	45.3	-11.7
Body (Abdomen)	19.5	19.9	20.4	-2.3

Table-7: Estimation of Effective dose, E_{eff} to patient organs at 120 kV.

Phantom (organ)	DLP (console) (mGy.cm)	DLP (calculated) (mGy.cm)	E_{eff} (Typical value) (mSv)	E_{eff} (calculated)* (mSv)
Head (Brain)	482	622	1.0	1.3
Body (Abdomen)	468	505	7.0	7.6

*k-factor taken as: 0.0021 (Brain) and 0.015 (Abdomen)

4. Discussion

This work was intended to evaluate the dose parameters influenced by the patient size using Pencil ion-chamber and Farmer ion-chamber measured in-air, acrylic head phantom and body water cylinder extended than standard size. As an author's understanding, this study is the first to quantify the dose in water cylinder simulating the adult body. Patient radiation doses are accessed for optimization purpose and to verify the performance of newly installed CT simulator. Size-specific dose estimates are also determined.

For in-air measurement, the IAEA TRS-457 [5] requires that CTDI measurement be performed in-air as well as using an acrylic head/body phantom to calculate $CTDI_w$ and further, the $CTDI_{vol}$ value that is a key parameter usually stated on the CT scanner console. Additionally, the AAPM Report-111 [6] states there was "no consensus" concerning the use of a specific phantom type to determine the CTDI values. Though, the phantom shape or geometry was not fundamental for CT measurement, its uncertainties were considerable when measurements in a phantom are converted to the CTDI values for another phantom. In this context, the easier way was adopted for the result comparison by measuring the dose in-air (or air-kerma) using two different ion-chambers. The result in this study shows excellent equivalence of two types of measurements. When the measurements are normalized to a pitch factor, P.F $\cong$ 1, the results obtained are nearly identical. Besides the high dose at peripheral measurements, the in-air measurements show a very small difference in dose values at the center normalized to mAs.

For brain CT exam, the mean DLP value obtain in this work (717 mGy.cm) is comparable to reference dose values reported in literature—Bahreyni Toosi et al. -2012 (564 mGy.cm) [18], Torp et al. -2001 (740 mGy.cm) [19], Hidajat et al. -2003 (587 mGy.cm) [20] , Shrimpton et al. -2005 (787 mGy.cm) [21], Hatziioannou et al. -2003 (677 mGy.cm) [22], and is satisfactory lower than standard value (1050 mGy.cm) for head exam defined in European DRL (EUR 16262 -2000) [23].

For the abdomen CT exam, the mean DLP value obtained in this work (411 mGy-cm) is much lower than the standard value (780 mGy.cm) defined in EU DRL. The wide difference in the range of calculated $CTDI_{vol}$ is upto 4 times, and the percent difference relative to scanner displayed values is 9.6% for head exam and 8.8% for abdomen exam. This is mainly due to the fact that the scanner displayed values do not account for patient size but correspond only to the standard diameter. Additionally, the TG-204 [8] reports that without recognizing the distinction between the phantom sizes, interpreting the scanner-displayed values as patient dose may lead to underestimation upto a factor of 2 to 3. Even if a quantity is not to be comparable, the author suggests' these values can be used as a baseline data for the evaluation of CT performance and the prediction of malfunctioning of a CT system.

Determining the SSDEs, the values of conversion factors for CTD_{Ivol}, g(d) calculated as A_0 x exp (B_0 x d) are taken as 2.09, 1.79, 1.12, 1.00 for phantom dia 16, 20, 32, 35 respectively, where exponential regression coefficients A_0 = 3.90 and B_0 = -0.039 are taken from Haba's work (Haba-2020) with each weighting factors (3/8 and 5/8) as a good agreement with those from AAPM TG-204 of the $CTDI_w$ equation. For clinical submissions, SSDEs are used to describe how patient size has been taken into account to assess patient dose from $CTDI_{vol}$, an index of scanner radiation output. TG-204 also illustrates that the uncertainties in reporting the CT radiation dose, e.g., $CTDI_{vol}$, approach 20%. The value of SSDEs would not be used further to estimate the modified DLP (DLP = $CTDI_{vol}$ x L) and would not be used in computing the E_{eff} (E_{eff} = DLP x k-factor).

5. Conclusion

All the dosimetric quantities are comparable to European DRLs for determining the effective dose-to-organ as an attribute of radiation exposure when a patient undergoes CT examinations. The conversion factors applied to the displayed dose index $CTDI_{vol}$ takes advantage of estimating patient dose in routine dosimetry as these factors take account of the patient size. This study not only contributes to an experimental validation of the existing knowledge based only on the MC studies but also illustrates that the x-ray radiation from a CT unit can be quantified, in an easier way, with a

Farmer ion-chamber within acceptable limits. The results can be used as baseline values or reference to predict any deviation from standards or malfunction of the newly installed CT simulator, in future.

References

1. Shrimpton, P. and S. Edyvean, *CT scanner dosimetry*. The British journal of radiology, 1998. **71**(841): p. 1-3.
2. Schauer, D.A. and O.W. Linton, *NCRP report No. 160, ionizing radiation exposure of the population of the United States, medical exposure—are we doing less with more, and is there a role for health physicists?* Health physics, 2009. **97**(1): p. 1-5.
3. Medicine, A.A.o.P.i., *The measurement, reporting, and management of radiation dose in CT: AAPM report no. 96. AAPM website*, 2015.
4. Stefanovski, Z., S. Bidikova, and D. Nikolovski, *Testing the new AAPM formalism for the evaluation of radiation dose in x-ray computed tomography*. 2010.
5. IAEA, *Dosimetry in Diagnostic Radiology: An International Code of Practice, Technical Reports Series no. 457*, 2007, International Atomic Energy Agency Vienna.
6. Li, C.L., Y. Thakur, and N.L. Ford. *Dose comparison between CTDI and the AAPM Report No. 111 methodology in adult, adolescent, and child head phantom*. in *Medical Imaging 2017: Physics of Medical Imaging*. 2017. International Society for Optics and Photonics.
7. McCollough, C.H., et al., *CT dose index and patient dose: they are not the same thing*. Radiology, 2011. **259**(2): p. 311-316.
8. 204, A.T.G., *Size specific dose estimates (SSDE) in pediatric and adult CT examinations*.
9. Moore, B.M., et al., *Size-specific dose estimate (SSDE) provides a simple method to calculate organ dose for pediatric CT examinations*. Medical physics, 2014. **41**(7): p. 071917.
10. Supanich, M. and D. Peck. *Size-specific dose estimate as an indicator of absorbed organ dose in CT abdomen and pelvis studies*. in *Radiological Society of North American Scientific Assembly and Annual Meeting Program, Chicago, IL*. 2012.
11. Bauhs, J.A., et al., *CT dosimetry: comparison of measurement techniques and devices*. Radiographics, 2008. **28**(1): p. 245-253.
12. Anam, C., et al., *Comparison of central, peripheral, and weighted size-specific dose in CT*. Journal of X-ray Science and Technology, 2020. **28**(4): p. 695-708.
13. Haba, T., M. Kobayashi, and S. Koyama, *Size-specific dose estimates for various weighting factors of CTDI equation*. Physical and Engineering Sciences in Medicine, 2020. **43**(1): p. 155-162.
14. Haba, T., et al., *New weighting factor of weighted CTDI equation for PMMA phantom diameter from 8 to 40 cm: A Monte Carlo study*. Medical physics, 2017. **44**(12): p. 6603-6609.
15. Bakalyar, D., *SU-FF-I-28: A critical look at the numerical coefficients in CTDIvol*. Medical Physics, 2006. **33**(6Part3): p. 2003-2003.
16. Hasford, F., et al., *Determination of dose delivery accuracy in CT examinations*. Journal of Radiation Research and Applied Sciences, 2015. **8**(4): p. 489-492.
17. Samei, E., et al., *Performance evaluation of computed tomography systems: summary of AAPM task group 233*. Medical physics, 2019. **46**(11): p. e735-e756.
18. Bahreyni Toossi, M.T. and M. Bahrami, *Assessment of patient dose from CT examinations in Khorasan, Iran*. Iranian Journal of Medical Physics, 2012. **9**(4): p. 233-238.
19. Torp, C., et al., *Use of the EC quality criteria as a common method of inspecting CT laboratories-a pilot project by the Nordic radiation protection authorities*. 2001.
20. Hidajat, N., et al., *Survey of conventional and spiral CT doses*. Radiology, 2001. **218**(2): p. 395-401.
21. Shrimpton, P., et al., *Doses from computed tomography (CT) examinations in the UK-2003 review*. Vol. 67. 2005: NRPB Chilton.
22. Hatziioannou, K., et al., *A contribution to the establishment of diagnostic reference levels in CT*. The British journal of radiology, 2003. **76**(908): p. 541-545.

23. Jessen, K., et al., *EUR 16262: European guidelines on quality criteria for computed tomography*. Luxembourg: Office for Official Publications of the European Communities, 2000.