Nigerian Journal of Applied Physics (NJAP)
ISSN Print: 3115-5871
ISSN Online: 3115-588X
DOI: https://doi.org/10.62292/njap-v2i2-2026-49
Volume 2(2), June 2026

A Study of the Quantitative Assessment of Radiation Dose Out Put for Tomography View of the Dental Cone Beam Computed Tomography (CBCT) System

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ABSTRACT
The efficiency of machine operation in radiology has been established to be affected by many factors, such as the device used, and the exposure level adopted for a specific type of radiation. The radiation exposure from cone beam computed tomography (CBCT) system is generally higher than that of other diagnostic modalities such as X-ray and computed tomography (CT) scan. This study aims to determine patient dosimetry on CBCT scanners used in Advanced Medical and Dental Institute (AMDI) University Sains Malaysia (USM). The CTDI phantom was scanned using the common exposure settings used in clinical routine (90 kVp, 8 mA) with a full rotation scan. For the standard CTDI method, the dosimeter was placed to measure the doses at the central hole (CTDIc) and the four peripheral holes ($CTDI_p$) of the phantom making a total of five measurement points in the scanning field of view. The percent deviation of the recorded doses at the peripheral points were observed to range from 0.7 % to 10 %. The highest average dose was observed at the hole C which is about 90° and with a position near the x-ray tube. Furthermore, the measured $CTDI_{100}$ value was 3.286 mGy which was within the expected range of ($40\ mGy$) which is the value obtainable in most standard scanning protocols. The difference between the displayed and measured $CTDI_{vol}$ value was 0.814 (19.85 %) which was observed to be within 20 %, which is comparable to that of the Swiss legal requirements (within a $\pm$20 % threshold limit) and the American Association of Physicist in Medicine (AAPM) report 204 of approximately 20%. This study showed the effectiveness of CTDI as a dose index in CBCT scans and also that the CBCT scanner in AMDI USM poses no risk to the patients.

Keywords:
CBCT,
CT dose index,
Computed tomography dosimetry,
Computed tomography phantom,
Imaging.

INTRODUCTION
Medical physics have been a growing research field of great importance at AMDI USM. Evident of this is the availability of the CBCT machine at the imaging unit of AMDI. The need for using this imaging technique has become more important in radiology. Currently, the CBCT scanners have an extensive choice of exposure factors and employ techniques that can significantly influence the radiation dose given to a patient. During a CBCT examination for a specific clinical objective, a quality image should be recorded without any unnecessary dose to the patient.

The efficiency of machine operation in radiology have been established to be affected by many factors, one of which is the device used (Nasir et al., 2016). It has also been established that the radiation exposure from CBCT is generally higher than that of other diagnostic modalities and this has been a major concern (Brink & Morin, 2012; Khawaja et al., 2015; Lorenzoni et al., 2012; Xu et al., 2020). Hence, a comprehensive and accurate assessment of this radiation dose allowable to patient generated by the CBCT to as low as reasonably achievable (ALARA) without exceeding the limit of radiation dose to individual has become necessary (Anam et al., 2016; Nasir et al., 2016; Rajaraman et al., 2020; Terashima et al., 2019; Xu et al., 2020). This is in order because the amount of radiation dose received by a patient during scanning irrespective of its magnitude have the potential of causing changes in the biological system of the patient and eventually increase the risk of cancer to the sensitive organs at risk. This is because of the increased ability of x-rays to induce mutations in DNA, thus increasing the risk of cancer particularly in paediatric and adolescent's patient than in adult. This is due to their longer life expectancy of more than an adult of about 50 years old, and this has been a major concern in the imaging community. Thus, the prescription of CBCT exams for these category of people must be strongly targeted on the individual's needs and its use must be justified based on the situation, such that the benefits outweigh the risks of radiation exposure, and all basic principle must be strictly adhered to (Dawood et al., 2009; Dhillon & Kalra, 2013; Kritsaneepaiboon et al., 2019; Lorenzoni et al., 2012; Theodorakou et al., 2012; Xu et al., 2020).

The degree of the risk is dependent on the amount of radiation dose received. It must be noted that there is no limit of dose below which this effect is zero and cannot manifest (McGuigan et al., 2018; Theodorakou et al., 2012; Yang, 2016). Hence, the need to account for machine dose from any radiological examination which uses ionising radiation (Rajaraman et al., 2020). This radiation dose delivered by the machine during a particular scanning procedure is extremely variable and depends on a number of technical factors such as beam quality, dose rate setting etc. Furthermore, the geometry, shape as well as material of phantom used can also influence the amount of radiation dose that will be delivered from a particular scanning procedure, and the volume computed tomography dose index ($CTDI_{vol}$) is used to calculate the average exposure to a phantom during a CBCT examination.

In recent time, all modern imaging device such as CBCT scanners displays the dose values such as volume computed tomography dose index ($CTDI_{vol}$) (mGy) and the dose area product (DAP) (mGycm^2) dose indices before a scanning process, which permit for the determination of the effect of various scan protocols on dose (AAPM, 2011, 2016; Dawood et al., 2009; Terashima et al., 2019). The $CTDI_{vol}$ and DAP are often used to represent the radiation doses from a CBCT scan, but it is not a representation of the absorbed dose to the patient on which the scan was performed (Boone et al., 2019) . They have the drawback of being surrogates for patient dose, as they only include details about the scanner performance for a very particular standard situation (AAPM, 2014).

The $CTDI_{vol}$ refers to a standard procedure in the comparison between the radiation output of various CBCT scanner using a standard phantom. This $CTDI_{vol}$ is normally defined for two diameters of polymethylmethacrylate (PMMA) phantoms which represent the body (32cm in diameter) and the head (16cm in diameter) which is used in this study (Anam et al., 2016). The exposure parameter (such as tube voltage, tube current, exposure time) applied during a particular scan has the capacity to influence the $CTDI_{vol}$ and the DAP, but they are not affected by the size of the patient (AAPM, 2011; Brady & Kaufman, 2012). The CTDI only provide information about the scanner output and not the size of the patient, hence, it cannot estimate the patient dose (Terashima et al., 2019). Patient size must be taken into account both in estimation of patient dose and prescribing the correct scanner output setting.

A significant variation in CTDI-based dose metrics can and should be expected, which is a significant implication of need to consider patient size when estimating patient dose and prescribing the appropriate scanner output settings. A broad range of scanner output (CTDI) values are recommended by facilities that modify their CT method according to patient size, whether through the use of human technique charts or automatic exposure management. This is a positive result that demonstrates the facility's ability in "right-sizing" the dose settings according to individual patient body physique. Moreover, even for patients of the same size, variations in image quality standards for different clinical applications and diagnostic tasks lead to variations in the recommended scanner output settings. For instance, even when a patient is being examined for CT colonography and CT enterography, the scanner output should differ significantly.

Accurate measurement are the basis of every effective diagnosis and safe treatment and this fundamental requirement for patient safety must be met, therefore, selecting the appropriate scanner output parameters for the imaging task and patient size is essential for radiation management in CT (McCollough et al., 2011). Since medical imaging from CBCT examinations makes the largest contribution to the radiation dose received by patients from medical exposures. Therefore, this study aims to determine patient dosimetry on CBCT scanners in order to quantify the scanner output using the CT dose index (CTDI)

MATERIALS AND METHODS

UNFORS RaySafe Multimeter

The UNFORS RaySafe Xi multimeter (UNFORS Raysafe, AB, Sweden) is a device used for radiation dose measurement of covering various performance parameters for quality assurance (QA) testing of the dental CBCT unit (as shown in Figure 1). This meter was used for measurement and evaluation of CBCT performance testing (RaySafe Xi CT dosimeter). This meter was connected to the base unit by a long electric cable. The base unit displays all the measurement values that were set by the users. The long electrical cable allows the measurements to be taken and recorded from base unit at the control room, which is outside the scanning room during the phantom scanning for radiation protection to the staff.

Figure 1: UNFORS RaySafe Xi CT dosimeter for CTDI measurement used in this study

The RaySafe, Xi multimeter has a built-in sensor (internal silicon detector) for measurements of wide range of parameters such as tube current (mA, mAs, waveform), exposure time, pulses, pulse rate and tube voltage (kVp). Besides, it can also be used as dosimeter to measure the dose, dose rate, dose-area product (DAP), half-value layer (HVL) all at the same time with a single exposure setting (Stanley et al., 2014). The sensor has a maximum inaccuracy of 0.5 % for time measurement and 2 % for all other measurements. The instrument is controlled by two buttons: On/Off and arrow button for scrolling the measured data. The carefully built liquid crystal display (LCD) clearly displays both the measured values and their units.

The R/F meter consist of a flat plane ionization chamber for easy adaptability over the detector of the CBCT unit. This device, which is a multiparameter x-ray meter, has an internal silicon detector that can measure kVp, dose rate, and time with a maximum inaccuracy of 0.5 % and 2 % for all other measurements.

CTDI phantom (sized 16 cm)

The CTDI phantom is a commercially available phantom that is suggested by the SEDENTEXCT for CT dose measurement. In this study, the 16 cm diameter CTDI phantom was used for the evaluation of the measured CBCT dose. This phantom is made up of poly-methyl methacrylate (PMMA) cylindrical blocks with a density of 1.2 ± 0.01 g.cm^{-3} with 5 holes precisely dimensioned for the insertion of pencil ion chambers during the dose measurement (Abouei et al., 2015). The phantom is cylindrical in shape and has a height and diameter of about 15 cm and 16 cm, respectively that is a convenient choice as a standard reference size for the dose measurement of head (Li et al., 2017; Yu et al., 2010). It is a cylindrical PMMA which represents an adult human head (size 16 cm).

This phantom is used to quantify the scanner output (CT dose index, CTDI) for head CBCT examinations (McCollough et al., 2011). The phantom has five holes, one is positioned at the center and 4 holes at the peripheral part of the phantom (Akpochafor et al., 2019; Sharma et al., 2006). Each of the peripheral hole is positioned at the 0°, 90°, 180°, 270° of the peripheral positions, with diameter of each hole is 1.3 cm (Figure 2). Each hole is used to accommodate the insertion of the ionisation chamber and each of the holes is positioned at 10.0 mm from the edge of the phantom (Boone et al., 2019; Sookpeng et al., 2016).

Alongside this phantom are 4 fillers (plug or rods) that are made up of acrylic (PMMA) with their diameter and height corresponding to the diameter and height of the phantom holes (AAPM, 2011; Abouei et al., 2015; Araki et al., 2013; Buckley et al., 2018). These rods are provided for insertion into the empty holes of the phantom during CTDI measurement. During phantom scanning, only one hole contains the ionisation chamber or dosimeter, while other empty holes should be inserted with the PMMA rods during each scan to avoid air within the field of view (FOV) and reduce inaccuracy in dose measurement (Boone et al., 2019; Pauwels et al., 2012).

Figure 2: The 16 cm and 32 cm cylindrical CTDI phantom with the acrylic plug or rods to be inserted at the five holes within the phantom

CT Dosimetry

The radiation dose output for the tomographic view of the CBCT was evaluated using the CTDI phantom (size 16 cm). The CT dosimeter (UNFORS Raysafe multimeter) with height of 100 mm was inserted into and securely fit the hole of the phantom. Also, 4 fillers (acrylic plug or rod) with diameter and height corresponding to the size of the holes of the phantom was used for insertion into the holes of the cylinder phantom that was not containing the CT dosimeter during phantom scanning (Boone et al., 2019; Pauwels et al., 2012). Each hole was labeled as A, B, C, D and E for the purpose of easy identification (Bamba et al., 2013).

The UNFORS Raysafe CT dosimeter was then connected to an electrometer and radiation monitor controller and then the phantom was carefully positioned on the CBCT holder with similar setup as the constancy test procedure to imitate patient positioning during a head CBCT scan (as shown in figure 3). The patient positioning lasers light were used as a guidance during the phantom positioning (Buckley et al., 2018). The phantom was positioned on the height-adjustable table (Figure 3) and aligned at the isocentre within the field of view (FOV).

Figure 3: The CTDI phantom sized 16 cm with the UNFORS CT dosimeter were positioned on the CBCT holder during the dose measurement

The phantom was scanned using the common exposure settings used in clinical routine with a full rotation scan. For the standard CTDI method, the dosimeters was placed to measure the doses at the central hole (CTDIc) and the four peripheral holes ($CTDI_p$) of the phantom, a total of five measurement points in the scanning field of view (Allisy-roberts & Williams, 2008; Yu et al., 2010). The CTDI varies across the field of view (FOV). The weighted CTDI ($CTDI_w$) is used to calculate the average CTDI across the FOV (AAPM, 2008). The $CTDI_w$ was calculated using Equation 1 (Buckley et al., 2018; Sharma et al., 2006; Siewerdsen et al., 2019; Sookpeng et al., 2016).

$$CTDI_w = \frac{1}{3}CTDI_C + \frac{2}{3}CTDI_P \qquad (1)$$

Where the factor $1/3$ and $2/3$ is a contribution from the relative air-kerma if the air-kerma in the CTDI phantoms reduces linearly from the periphery to the centre (Akpochafor et al., 2019). Similarly, two methods (Indices) exist for the measurements of CBCT dose index. Dose index 1 (DI 1) involves the averaging of the dose measurements along the diameter of the phantom (Abouei et al., 2015; Li et al., 2017), and it requires the measurements of the dose from a number of various points along the diameter of the phantom and then the DI 1 will be evaluated as the average dose along the diameter of the phantom (Equation 2) (Abouei, 2014; Abouei et al., 2015)

$$DI1 = \frac{\sum_{i=1}^{3} D_i}{3} \qquad (2)$$

According to SEDENTEXCT project recommendation, the dose measurements should involve many points along the diameter of the phantom. Hence, this method may not be applicable to this research as it may give result slightly different from SEDENTEXCT recommendation since the phantom used in this study only have 3 holes along its diameter. Similarly, Dose index 2 (DI 2) which is appropriate for this study, the dose was obtained by measuring the dose at the central point and at the peripheral area of the cylinder (Li et al., 2017) by applying equation 3 (Araki et al., 2013; Li et al., 2017)

$$DI2 = \frac{C + \mu_p}{2} \qquad (3)$$

Where C is the dose measured in the central hole of the phantom and μ_p is the average of all peripheral dose measurement. The scan was then repeated for another three times using the same exposure parameters. This was to check the reproducibility of the measured dose. Then an average of all the three measurements was taken as the dose for each point.

RESULTS AND DISCUSSIONS

CT Dosimetry

Following the procedure proposed by the SEDENTEXCT, the dose was measured using the dosimeter in all holes on the CTDI phantom by positioning the centre of rotation of the x-ray beam at the phantom's centre as well as in the periphery area. The CT doses were measured using the same exposure parameters as in typical clinical setting. The measured dose values were displayed on the monitor that is attached to the dosimeter. The result of the measured dose is presented in Table 1.

Table 1: Exposure parameters of the cone beam CT machines along with CTDI values

Exposure setting	$CTDI_c$, Hole A (mGy)	$CTDI_p$ (mGy)					Measured $CTDI_{100}$	Mean $CTDI_{100}$	% Deviation between set & measured $CTDI_{100}$
		B	C	D	E	Mean			
90 kV, 8 mA	2.967	2.988	4.898	2.995	2.892	3.443	3.285	3.286	19.85
	2.981	2.997	4.897	2.979	2.881	3.439	3.286		
	2.976	2.998	4.894	2.989	2.893	3.444	3.288		

During the scan, the distribution of the dose from points (A–E) was performed with the hole A centred in the phantom. Following the placement of the isocentre at point A, the dose at point C was revealed to be the highest in the full rotation scans (Figure 4). Besides, the percent different of the recorded doses at points A, B, D and E were observed to range from 0.7 % to 10 %.

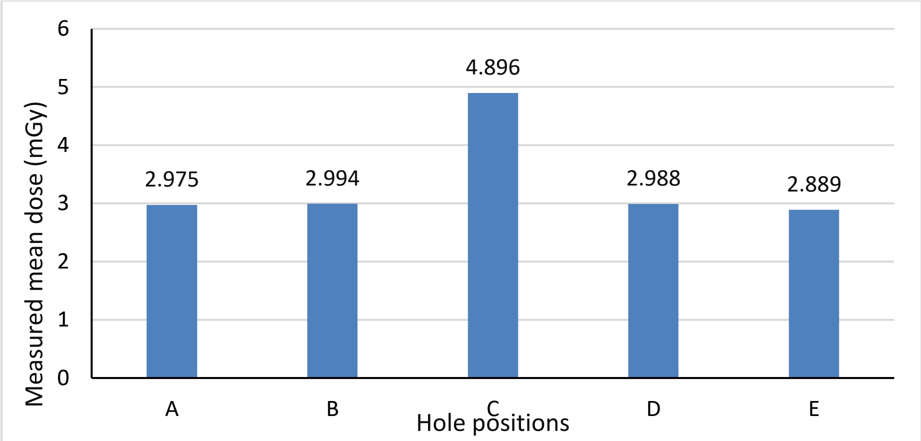

Figure 4: Distribution of measured dose at different holes (A–E) on the CTDI phantom

From Table 1, the highest average dose was observed at the hole C (90°) and the position is near towards the x-ray tube (radiation source). This was expected because, this was point where the ion chamber became closest to the x-ray tube, hence probably accounted for this high dose measurement. Similarly, a reduction of approximately 41 % was observed at the peripheral area of the phantom which was farthest (about 0.6 m) from the radiation source. This is expected since radiation dose decreases in accordance with the inverse square law. Thus, the dose distribution was observed to be a symmetric dose distribution. Furthermore, the measured $CTDI_{100}$ value was 3.286 mGy which was within the expected range of (40 mGy) which is the value obtainable in most standard scanning protocols (Allisy-roberts & Williams, 2008). The difference between the displayed and measured $CTDI_{vol}$ value was 0.814 (19.85 %) which was observed to be within 20 %, which is in conformity with Swiss legal requirements (limit of ± 20 %) and the AAPM report 204 (AAPM, 2011; Allisy-roberts & Williams, 2008; Racine et al., 2017). Besides, the length of the ionization chamber (100 mm) employed for the dose measurement was enough to produce a precise evaluation with the FOV used in this study. This ionization chamber utilized in this study can be utilized also in other CBCT machines with medium FOV (up to 6 cm) but, it may not be proper for system whose FOV is quite large. This is simply because the scattered radiation may not be completely accounted for, hence, resulting in an underestimation of the dose values. The CTDIvol value which was reported on this scanner for the 16 cm PMMA phantom was used with size dependent factor. The factor for this phantom was based on the phantom's effective diameter.

For the head phantoms, CTDIs measured at the periphery were similar to each other and were 1.9-2.0 times smaller than that at the center. The dosimetry for the cone-beam geometry has become more important due to its widespread applications in CBCT systems for diagnostic imaging. CT dosimetry phantoms are essential tools for quality assurance and accurate characterization of patient dose. The choice of the appropriate material is important. The phantom selected must have properties similar to water. PMMA is a tissue-equivalent material and has been used as the established phantom material for CTDI assessment. Besides, nylon is also recommended to be used for the phantom in CTDI assessment if PMMA is not available. This is because nylon is also a suitable material for this purpose as it has properties identical to the human tissue. In addition, its physical density and attenuation coefficient are equivalent to water (Khallouqi et al., 2025; Sookpeng et al., 2016) and comparable to PMMA. However, since nylon is not a transparent, the beam from the laser light may not pass through the phantoms during positioning. However, the positioning can be done basically by observing the light beams on the surface of the phantom. Besides, it is recommended that the dose measurements and the image quality performance testing for the CBCT system should be done at the time of CBCT equipment installation and the result should be used as a reference standard for subsequent assessments, which could be performed once a year. The requirements of local and state regulations should be checked to ensure that all relevant tests are conducted and that their limits are not exceeded. These evaluations should be carried out by or under the supervision of a qualified medical Physicist or a trained professional licensed by the regulatory authorities before the unit is used in a typical clinical setting and yearly after wards.

CONCLUSION

In this study, the CTDI concept to CBCT dosimetry was applied. The applicability of CTDI as a dose index for CBCT was also validated and incorporated the CTDI with beam collimation to estimate the total CBCT scan dose. Using this strategy, we successfully estimate the CTDI for a commercial CBCT system in AMDI. We hope that these CTDI values are useful to medical physicists when they estimate the CBCT dose for the CBCT system.

It is important that measurement of the radiation output of a CBCT system can be easily and practically measured in a consistent and robust manner and the $CTDI_{vol}$ fulfil this requirement which is defined both in the United States and in international regulatory guidelines, and the equipment used to measure it is readily available worldwide.
The $CTDI_{vol}$ tells the medical physicist precisely how the machine operates, and it can be used, in conjunction with information regarding patient size and the scanned anatomy, to estimate patient dose. Dose estimates can be for organ dose, skin dose, or a mean dose in the center of the scan volume. However, the CTDI results are not estimates of the patient's dosage. CTDI is the tachometer in a CBCT scanner—not the speedometer. Estimates of individual patient risk, and epidemiologic studies assessing potential late effects, must use patient size–specific dose estimates—they cannot use only scanner output. Instead, by utilizing the established exponential relationship between patient size and absorbed dose, scanner output values can be used to predict patient size-specific doses.

ACKNOWLEDGEMENT
The authors would like to thank all staff at Imaging Unit of AMDI, USM, Penang, Malaysia for their help and support throughout this work.

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