

Establishment of Diagnostic Reference Levels and Estimation of Effective Dose from Computed Tomography Head Scans at a Tertiary Hospital in South Africa

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ARTICLE INFO	ABSTRACT
Article type: Original Article	Introduction: Head scans are the most frequently performed computed tomography (CT) examinations worldwide. However, there is growing concern over the probability of increased cancer risks among the exposed populations. Diagnostic reference levels (DRLs) identify radiation dose that is not commensurate with clinical objectives. The aim of this study was to establish DRLs for CT head procedures and estimate effective dose (ED).
Article history: Received: Jan 25, 2019 Accepted: Jul 05, 2019	Material and Methods: The dose absorbed by the head slice of a Rando Alderson phantom was measured using calibrated lithium fluoride thermoluminescent dosimeters (TLDs) exposed to a CT scanner operated on clinical parameters. The measurements were done at the periphery and center of the slice, and repeated twice with a new set of TLDs. The radiation dose absorbed by the TLDs was read using a Harshaw TLD reader, Model 5500. The measured doses were used to calculate the weighted CT dose index (CTDI _w), CT dose index volume (CTDI _v), and dose length product (DLP). Finally, the ED was calculated using the formula; ED = $k \times \text{DLP}$, where k was considered as 0.0021.
Keywords: Effective Dose Computed Tomography Dose Absorbed Dose	Results: The mean absorbed dose was 30.9 mGy, while the established CTDI _v and DLP values for the head protocol were 40 mGy and 990 mGy.cm, respectively. Additionally, the ED was calculated as 2.1 mSv. These values compared well with some international values. Conclusion: According to the results of the present study, the established CTDI _w , DLP, and ED for head scan were well-compared with some international values, except in the cases using different scan lengths and scanner algorithms.

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Introduction

Computed tomography (CT) imaging has revolutionized medical imaging since its emergence during 1970s [1]. The introduction of multi-detector CT (MDCT) scanners, which allows for the fast acquisition of three-dimensional improved-quality images, increased the demand for CT modality [1-5]. Therefore, the use of CT imaging as the preferred modality has contributed to successful surgeries, improved diagnoses, and cancer treatment [4, 5].

The widespread use of CT imaging modality has decreased the need for emergency surgeries from 13% to 5% [5]. However, despite the well-publicized patient benefits, there is growing concern about the high dose delivered by this imaging modality [2-5]. For instance, the low dose radiation from the X-rays of an operational CT scanner raises the risk of cancer among the exposed population [2, 6].

Brenner et al. established that CT examinations contribute disproportionately to the collective diagnostic dose [4]. In the United States of America, about 70 million CT scans are performed every year. The high demand for CT examinations makes it the

main contributor to human exposure, contributing 49% of manmade medical radiation despite accounting for only 17% of medical radiation-based exposures [7].

A study performed by Okei et al. [8], it was reported that in the United Kingdom the CT examinations constitute about 40-47% of collective dose arising from all medical exposures, while CT modality is responsible for only 3-5% of all X-ray-based examinations. Additionally, Tsushima et al. [9] found that the introduction of MDCT substantially increased CT examinations to an estimate of around 29.9 million a year in Japan. Another study carried out by Chipiga et al. [10] in Russia revealed that just in 2015, 8 million CT scans were performed, and the CT doses were within the range of 50-100 mSv. All in all, the CT modality was found to account for 45% of the collective dose to the Russian population [10]. Furthermore, Wardlaw [11] found that a total of 4.3 million CT scans were conducted in Australia within 2010-2012, compared to 4.4 million carried out in 2012 alone [11]. In addition, Van der molen et al.

revealed that about 1.16 million CT scans were prescribed just in 2010 accounting for 47.5% of the total dose to the Dutch population. Therefore, one of the major concerns about the widespread use of CT is the associated increased radiation exposure incurred by patients and the risk of cancer induction [12].

The increased attention to the risk of radiation-induced cancers resulted in calling for the reduction of CT doses and prioritization of ionizing radiation protection of patients. Protection of patients from ionizing radiation is based on the justification of the prescribed examination [13, 14]. The protection principle requires the optimization of all CT doses, meaning that they should be kept as low as possible [13] and consistent with the clinical objective. To this end, the International Commission on Radiological Protection (ICRP) proposed the establishment and implementation of diagnostic reference levels (DRLs) [14]. In this regard, the use of DRLs facilitates the identification of patient doses that are unusually high or low for specified CT imaging procedures. Therefore, the DRLs as a form of investigation level can be considered an optimization tool [14, 15].

The International Electrotechnical Commission recommended the CT dose index volume (CTDI_{vol}) and the dose length product (DLP) as dosimetric for CT. The CTDI_{vol}, which is measured in milligray (mGy), is a standard measure of radiation output in a single slice of a CT scanner allowing the user to compare different scanners and scan protocols. On the other hand, the DLP quantifies the total radiation output of the CT scanner, and in so doing unveils the radiation dose received by patients during the CT scan. Mathematically, it is the product of the CTDI_{vol} and scan length. The DLP is measured in milligray-centimetres (mGy-cm) [16].

The effective dose (ED) is a dose quantity to a given particular organ in an irradiated volume weighted according to the radiosensitivity of the organ [17]. It may be used to compare the stochastic risks of various examinations and measure the risk of cancer induction [18, 19, 20]. The ED is calculated by multiplying the conversion factor with DLP value for a particular anatomical region using age and region-specific conversion factors provided by the ICRP publication 102 [21]. It is measured in millisieverts (mSv) [18, 21]. This study aimed to determine CTDI_{vol} and DLP values for CT head scans with the purpose of estimating the ED attributed to the head procedure.

Materials and Methods

Alderson Rando phantom (Figure 1) and lithium fluoride thermoluminescent dosimeters (TLDs) with the cross-section of 3×3 cm² and thickness of 0.9 mm (TLD-100, Harshaw-Bicron, Cleveland, OH, USA) were used in this study. A Harshaw 5500 TLD reader (Phoenix Dosimetry Ltd, USA) was also applied for the measurement of absorbed dose by the TLD chips.

The initial process involved annealing 30 TLDs chips for 1 h at 400°C, followed by fast cooling prior to individual calibration (i.e., a process aimed at the compensation of random response to the same radiation). Thereafter, the TLDs were placed on a thin Perspex slab with a source to surface distance of 80 cm, field size of 10×10 cm, and depth of 5 cm. In the next stage, they were irradiated with a cobalt-60 (⁶⁰Co) teletherapy photon beam of 50 cGy for 1.06 min.

After irradiation, the TLDs were annealed at 100°C in order to free the trapped electrons before readout. The irradiated TLDs were read out after 24 h (i.e., a time frame allowing for the elimination of low temperature peaks). These TLDs were classified as “element correction coefficient (ECC)” that is a relative response to irradiated dose from the mean. The ECC for each individual TLD was calculated in this once-off calibration process using Equation 1 [22]:

$$ECC_j = \frac{\langle Q \rangle}{Q_j} \quad (1)$$

Where, $\langle Q \rangle$ corresponds to the average charge of a set of TLD chips, and Q_j refers to the integrated current measured for TLD. In this regard, only TLDs with a variation of < 3% were selected to be used in the measurement of dose distribution in the head slice. This batch of TLDs was then read out in order to calculate the reader calibration factor (RCF) using Equation 2 [22]:

$$RCF = \frac{\langle Q \rangle}{E} \quad (2)$$

Where, $\langle Q \rangle$ represents the average charge of a set of TLD chips, and E is the radiation exposure delivered to that set. The TLD measurements were performed using a Harshaw 5500 TLD reader. After heating the light output was analyzed by a photomultiplier tube (PMT). The PMT provides an output current that is directly proportional to the chips radiation exposure which is calculated using Equation 3:

$$Exposure = \frac{ECC * Charge}{RFC} \quad (3)$$

The Rando phantom (Figure 1) represents an average person comprising of 2.5 cm-thick slices that are transacted horizontally with holes filled up with pins that are either bone, soft-tissue or lung tissue equivalent. Lithium fluoride (LiF-100) TLDs were chosen due to their efficiency and flat energy response within the range of X-Ray beam qualities used in diagnostic radiology [23, 24]. The TLDs met the technical requirements, including dose detectability of 50-100 μGy as proposed by Burke and Sutton [24], standard deviation of TLD batch as 5%, while the standard deviation of readings at 0.1 Gy was maintained to be < 30% [23, 24].



Figure 1. Alderson Rando anthropomorphic humanoid phantom used for the estimation of radiation dose output from a computed tomography scanner during head scanning

The TLDs were placed at the center and periphery of central head slice (Figure 2) to measure the absorbed dose by the head during CT scanning. Three measurements of the absorbed dose were conducted on head slice using newly calibrated TLDs for each measurement.



Figure 2. Alderson Rando phantom head slice indicating the positioning of thermoluminescent dosimeters used to measure the absorbed dose during a head scan

Once the TLDs were secured in the head slice in the positions of A, B, and C (Figure 2), the Rando phantom was reassembled to its original shape and placed at the isocentre of the CT scanner (Figure 3).



Figure 3. Alderson Rando phantom placed on a supine position ready for measurement of absorbed dose on the head

Table 1. Scan protocols and parameters for computed tomography scanner used in the present study

Scan protocol	Scan parameters			Pitch
	Tube voltage (kVp)	mAS	Scan time(s)	
Head	120	30-300	0.5-10	0.7-1.4

The dose distribution on the head slice was measured according to the head protocol. Two more measurements were conducted using a new set of TLDs. The scan parameters employed in head scanning are presented in Table 1.

Post-irradiation of the TLDs was embedded in the head slice, and the absorbed dose was read out using the Harshaw 5500 TLD reader. Table 1 shows the scan protocols employed during CT scan acquisitions. Based on the values of doses read from TLDs, the mean $CTDI_w$ was calculated using Equation 4 [25, 26]:

$$CTDI_w = \frac{1}{3}CTDI_{centre} + \frac{2}{3}CTDI_{periphery} \tag{4}$$

Where the weighting factors, namely $\frac{1}{3}$ and $\frac{2}{3}$, represent the values for the central and peripheral positions of the selected slice. The calculated value of $CTDI_w$ display dose in the x (i.e., horizontal direction) and y (i.e., vertical direction). The calculated values of $CTDI_w$ were then used to determine $CTDI_v$ using Equation 5 [26]:

$$CTDI_v = CTDI_w \div Pitch \tag{5}$$

Based on Equation 5, the DLP was calculated using Equation 6 [26]:

$$DLP = CTDI_v \times L \tag{6}$$

Where, L is the scan length.

Results

Table 2 presents the DLP, CTDI_v, and ED values for the head procedures. The DLP was calculated as the product of scan length and CTDI_v, while ED dose was calculated as a product of conversion factor *k* and dose

length product, where *k* was considered as 0.0023 mSv.mGy⁻¹cm⁻¹[21].

Table 2. Dose length product, computed tomography dose index volume, and effective dose values for routine head scans

Measurement	Pitch	Scan length (L, cm)	Measured doses				$\frac{1}{3}CTDI_c$	$\frac{2}{3}CTDI_p$	$\frac{1}{3}CTDI_c + \frac{2}{3}CTDI_p$	$CTDI_v = \frac{CTDI_w}{Pitch}$	DLP = CTDI _v × L	ED = <i>k</i> × DLP (mSv)
			Position A (peripheral)	Position C (peripheral)	Mean values A and C	Position B (centre)						
Measurement 1 [1 st scan]	0.78	25	31.05	26.75	28.9	29.84	19.89	9.63	29.52	37.85	946.25	1.99
Measurement 2 [2 nd scan]	0.78	25	27.24	27.92	27.58	32.95	21.97	9.19	31.16	39.95	998.75	2.10
Measurement 3 [3 rd scan]	0.78	25	29.84	30.22	30.03	32.94	21.96	10.01	31.97	40.99	1024.75	2.15
Mean value	0.78	25	29.39	28.30	29.81	31.91	21.27	9.61	30.88	39.60 ~ 40	989.92 ~ 990	2.10

**k* = 0.0021 is a conversion factor for adult head
 **k* factor is measured in mSv/mGy.cm and is age dependent
 **DLP is measured in mGy.cm

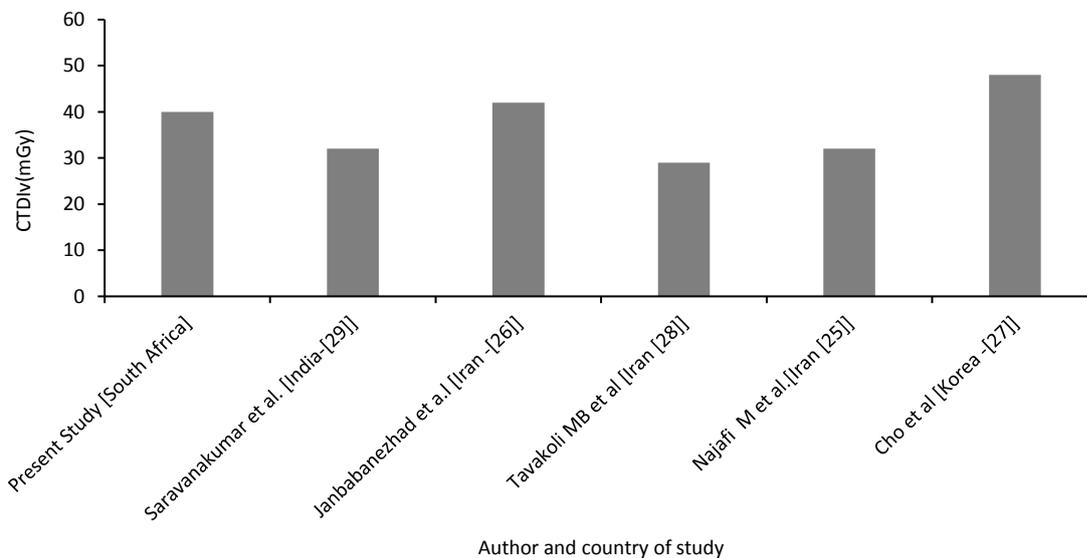


Figure 4. Comparison of the values obtained for computed tomography dose index volume in the current study and international values

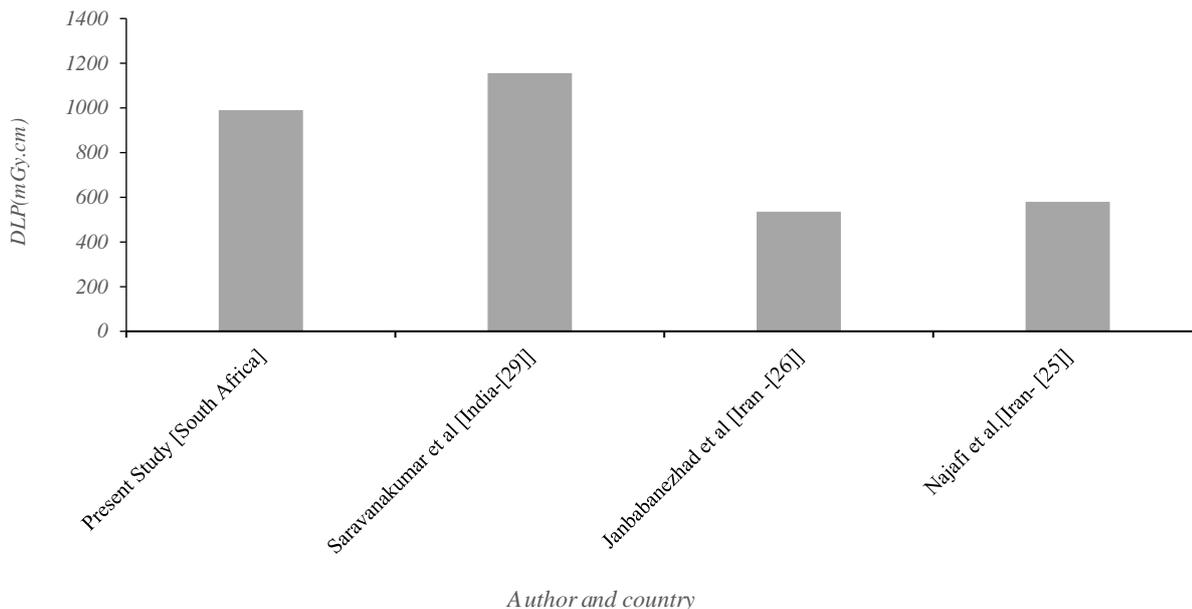


Figure 5. Comparison of dose length product obtained in the current study with international values

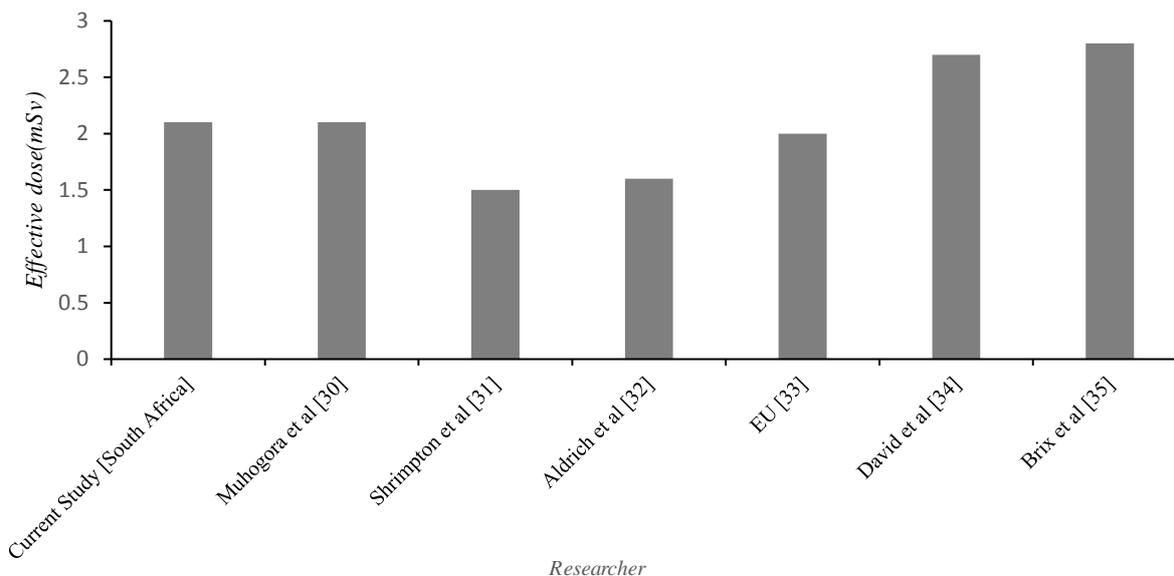


Figure 6. Comparison of effective dose value obtained in the current study with international values

The $CTDI_v$ value obtained in this study well-compared with the international values (Figure 4); moreover, it allowed for phantom implementation. The established DRL for $CTDI_v$ (40 mGy) was comparable to 42 mGy, established in a study conducted by Janbabanezhad et al. in Iran [26]. However, this value (40 mGy) was better than that of a study conducted by Cho et al. in Korea reporting a value of 48 mGy [27]. Although our value was in line with that of the study conducted by Janbabanezhad et al. [26], and lower than the one carried out by Cho et al. [27], there still remains the need for the optimization of head protocol in our center. A comparison with other international values

revealed that $CTDI_v$ obtained in this study was high, compared to 29, 32, and 32 mGy, established by Tavakoli et al. [28] and Najafi M et al. [25] in Iran and Saravanakumar et al. in India [29], respectively.

The DLP (990 mGy.cm) established in this study compared well with international values (Figure 5). However, it was lower than 1155 mGy.cm reported by Saravanakumar et al. [29] in India and higher than 536 mGy.cm and 580 mGy.cm presented by Janbabanezhad et al. [26] and Najafi et al. [25] in Iran. Therefore, it is still essential to optimize the head procedures in our institution.

The ED of 2.1 mSv established in this study was found to be higher than some international values, such as 1.97 [30], 1.5 [31], 1.6 [32], and 2.0 mSv [33]. However, it was comparable to some international values, such as 2.7 [34] and 2.8 mSv [35], (Figure 6).

Discussion

The current study successfully established the DRLs and ED for head CT procedures conducted at a tertiary hospital in South Africa, using a Rando phantom and TLDs. The CTDI_v (40 mGy) established in the current study compared well with some international values. However, it was found to be higher than some values reported in other regions (Figure 4), therefore suggesting the need for dose optimization.

The established CTDI_v value (40 mGy) was very close to that established in a study conducted by Janbabanezhad et al. (42 mGy) [29]. Moreover, the same tube voltage (120 kVp) was used in both studies. Therefore, the tube voltage can be concluded to have a similar influence on dose absorption in both studies. However, the slight difference in CTDI_v values may be attributed to the operator's experience and difference in tube currents used. Additionally, the use of a shorter scan length (12 cm) by Janbabanezhad et al. [26] can be ascribed to differences in DLP values (i.e., 990 mGy.cm for this study versus 536 mGy.cm established by Janbabanezhad et al. [26]). The scan length directly affects patient dose; consequently, it should be limited to areas relevant for diagnosis in order to reduce patient dose [29]. The DLP is the product of CTDI_v and scan length [26]; therefore, a longer scan length increases the value of DLP.

A further analysis of Figure 4 shows that despite the use of the same tube voltage (120 kVp) in both the present study and a study conducted by Cho et al. in Korea [27], the latter established a much bigger CTDI_v value (48 mGy), compared to 42 mGy established in this study. This discrepancy could be ascribed to difference in tube currents applied (mAs). In this respect, Cho et al. [28], used 250 mAs, while in the current study, a lower mAs was utilized. Furthermore, Korean and South African patients are significantly different, and radiographers are also trained differently in these two countries. Therefore, this inconsistency can be attributable to patients' demographics and radiographers' operating skills.

In another phantom study, Saravanakumar et al. [29], established a CTDI_v of 32 mGy using the same tube voltage (120 kVp) as the one used in the current study. The differences in the values (32 mGy < 40 mGy) may be attributed to the differences in the tube utilized in the current study. The use of different scan lengths in these two studies (i.e., 21.5 cm used by Saravanakumar et al. [29] and 25 cm used in the current study) contributed to large differences in DLP (i.e., 925 and 990 mGy.cm). As previously observed, patient dose escalates with the increase in scan length. In this regard, a shorter scan length will result in a smaller value of DLP since DLP is the product of scan length and CTDI_v.

In another phantom study, Tavakoli et al. [28] established a lower CTDI_v value (29 mGy), compared to 40 mGy obtained in the current study. However, both studies used the same tube voltage (120 kVp); therefore, differences may be attributed to the use of a smaller value of mAs in the study performed by Tavakoli et al. [28]. However, the exact value was not specified in their study. Najafi et al. [25], established a CTDI_v (value 32 mGy) lower than 40 mGy established in this study and used a tube voltage of 112 kVp and a scan length of 19.02 cm. This may justify the smaller DLP (580 mGy.cm), obtained by Najafi et al. [25].

Comprehension of the concept of ED in CT examinations by medical technologists and physicians in their institutions will enable them to have a more cautious approach in their practice. Furthermore, patients undergoing CT examinations, as well as their close relatives, should understand the concept of ED and its cumulative effect. The concept of ED

Defines the summation of doses to each organ in an irradiated volume weighted according to the radiosensitivity of each organ.

Radiation among various examinations can be compared using the concept of ED [18]. In this respect, the ED established in the current study was higher than some values reported in literature, such as 2.1 mSv versus 1.97 mSv by Muhogora et al. [30], 1.5 mSv by Shrimpton et al. [31], and 2 mSv by the European Commission [33]. However, the mentioned value was found to be lower than values established in some other studies, such as 2.1 mSv versus 2.7 mSv by David et al. [34] and 2.8 mSv by Brix et al. [35].

The differences in the ED estimated in this study and international values may be attributed to differences in technology. Among all the factors which are controlled by the radiologists and radiology technicians, doses to particular organs depend on tube voltage (kVp), scan length tube current, and scanning time in milliamperes seconds (mAs). As a result, it is not feasible to compare EDs. However, in the light of epidemiological studies on populations exposed to radiation, the results of the study can be put into perspective. For instance, the victims of Japanese atomic bomb who received EDs in a range of 5-150 mSv showed increased malignancies [4]. Additionally, the association of radiation with cancer was observed among 400,000 nuclear industry workers exposed to doses in a range of 5-150 mSv. Although the current study established a lower effective dose of 2.1 mSv, repeated scans, as well as cumulative doses, should be matters of concern since these cumulative doses tend to induce malignancy. Although the conclusion that CT doses are cancer induction agents still remains controversial, it is well agreed that CT doses need to be reduced since CT is a high radiation dose modality.

Conclusion

The DRLs play a significant role in CT dose reduction since they enable radiologists and radiographers to identify low or high doses in their

practice. However, the DRLs should not be viewed as punitive measures, and they can be surpassed depending on the clinical need. In this study, the established CTDI_v of 40 mGy was well-compared with some international values, such as 41 and 48 mGy established by Janbabanezhad et al. [26] and Cho et al. [27], respectively. However, the obtained value in the current study exceeded some international values, such as 32 and 29 mGy established by Tavakoli et al. [28] and Saravanakumar et al. [29], respectively.

The scan length for head CT protocols in our center was apparently longer than the one used in other studies, hence the we obtained were high DLP values. In this regard, the DLP values may be reduced through further optimization of protocols. Although ED was found to be lower than the cancer-inducing range of 5-150 mSv, we are still in need of optimizing the head protocols since repeated scans may tend to raise the risk of cancer induction.

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