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Estimation of effective dose using the dose length product in chest computed tomography procedures

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► ***Original article***

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ABSTRACT

Background: Approximation of radiation risks in computed tomography (CT) requires knowledge of specific organ doses. A Rando phantom and thermoluminescent dosimeters (TLDs) provide a proxy for *in-vivo* measurements. In this study, measured chest CT doses were used to calculate dose length product (DLP), a dosimetric needed for estimation of effective dose (E). ***Method and Materials:*** Ninety-five calibrated TLDs embedded at peripheral and central positions of Rando phantom chest slice measured chest CT dose during imaging using Phillips Brilliance 64-slice CT scanner. Three measurements were conducted each with new TLDs. Irradiated TLDs were read with a Harshaw TLD reader (Model 3500). One-way ANOVA test verified statistical significance of TLD measurements. TLD doses were used to calculate chest CT dose given as dose length product (DLP), a product of chest slice CT dose measured by volumetric CT dose index ($CTDI_v$) multiplied by scan length. Consequently, E was calculated as the product of DLP and k, an adult chest conversion factor published by International Commission on Radiological Protection Publication 103. ***Results:*** Differences in mean TLDs measurements were statistically significant (p=0.032). The mean chest slice peripheral and center doses were 3.61 ± 0.6 and 4.60 ± 0.31 mGy respectively. Adult chest CT dose was 178.8 ± 15 mGy. E was estimated as 2.5 ± 0.21 mSv. It is than the range (5.6 – 9.3 mSv) found in literature. ***Conclusion:*** E relates radiation exposure to stochastic effects. The estimated value (E = 2.5±0.21 mSv), reveals that chest CT protocol used was optimized.

Keywords: *Computed tomography, dose length product, thermoluminescent dosimeters, effective dose, optimization.*

INTRODUCTION

The advent of computed tomography (CT) imaging and development of scanner technology has made it possible to acquire high quality images [1, 2], in a fraction of seconds [2]. CT images show precise anatomic information making it easy to plan and execute therapeutic procedures successfully [1]. Reliance on CT images has significantly reduced exploratory surgeries [3]. Currently, CT imaging has seen applications expanding from cancer diagnosis to trauma screening [4]. Despite noticeable contributions of CT imaging in healthcare, concerns that X-rays used to produce CT images are carcinogenic have continued to increase [5, 6]. Of further concern is that it has since been established that any radiation exposure regardless of quantity possess health risk to patients [7]. It is therefore important that all potential health risks arising from CT imaging be reduced. In this regards, all clinical establishments should from time to time undertake radiation dose assessment. Clinicians have since embraced effective dose as a metric for radiation dose assessment. This choice is based on the opinion that effective dose provides a single value that measures the risk of cancer [8]. Effective dose is a concept originally developed for radiation protection purposes. It was primarily established to monitor compliance with regulatory limits. It also

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regarded as an essential tool for prospective dose assessment in radiological protection for the purpose of planning and optimization [8]. Although effective dose provides an acceptable metric for estimation of stochastic effects of radiation, it cannot be directly measured in vivo. However, it can only be estimated. Estimation of effective dose for patients undergoing CT imaging requires knowledge of specific organ dose [9], and the conversion factors commonly referred to as k-factors. The numerical value of effective dose in a clinical setting is considered as the product of the organ dose and the k-factor for the particular organ. International Commission on Radiological Protection Publication 103 published k-factors for various body organs for both children and adults [10].

The k-factors were derived from calculations involving use of computational human phantoms coupled with Monte Carlo transport simulation of CT X-ray beams [8]. Using the specific organ dose given by the dose length product (DLP) and the k-factor, the effective dose (E) can be calculated using equation 1:

$$E = k \times DLP \quad (1)$$

Where k is the k-factor measured in mSv mGy^{-1} cm^{-1}, E is effective dose measured in mSv and DLP is the dose length product measured in mGy.cm [10].

This study introduces a methodology for estimation of effective does in adult chest CT procedures using the Rando phantom and thermoluminescent dosimeters. Most medical personnel are already familiar with these apparatuses. Both apparatus are readily available in almost all clinical settings.

MATERIALS AND METHODS

Thermoluminescent dosimeter (TLD) dosimetry

Lithium fluoride thermoluminescent dosimeters doped with Mg and Ti (TLD-100), (Harshaw Chemical Company, OH, USA) were used in this study. The dimensions were 3.2×3.2×0.9 mm^3. TLD-100 chips have good tissue equivalence at X-ray photon energies delivered in CT imaging [11]. Furthermore, they are renowned for their high responsiveness to radiation [12]. However, these TLDs do not directly measure absorbed dose but electrical charge, hence they had to be calibrated before use [13]. In order to reduce chances of contamination, a vacuum tweezer was used when handling TLDs during experimentation.

A total of three hundred TLDs were used in the study. Firstly, they were cleaned with alcohol to remove impurities and then wiped dry with cotton wool. TLDs were then annealed in three batches in an oven (PCL_3, PTW-Freiburg, German) at 400^{0}C for one hour in order to reset trap structure and also to eliminate electrons in residual traps. Annealed TLDs were kept in ultraviolet environment at room temperature for 24 hours to enable thermoluminescent (TL) peak to fade out. Thereafter, a calibrated orthovoltage machine (Gulmay, German) was used to deliver uniformly 1.00 Gy to each of the three TLD batches while placed on a thin Perspex slab with source SSD = 80 cm, field size = 10 × 10 cm^2, depth = 5 cm, in the isocentre. After irradiation, a TLD reader LTM (Model 3500 with WinREMS, Saint-Gobain Crystals & Detectors Measurement Products, Ohio, USA) was used to read response of TLDs [14]. A batch comprising 10% of TLDs (known as calibration TLDs) was then used to generate an element correction factor (ECC) using equation 2:

$$ECC = \frac{\langle Q \rangle}{Q_i} \quad (2)$$

Where <Q> is average charge integral of calibration TLDs and Q_i is the individual charge [14].

TLDs that fell within ±10% range of calibration dosimeters were selected for generating Reader Calibration Factor (RCF) using equation 3:

$$RCF = \frac{\langle Q_c \rangle}{D} \quad (3)$$

Where Q_c is the average charge of the set of calibration and D is dose absorbed (1.00 Gy) by the TLDs upon irradiation [14]. Once the ECC and the RCF have been established, they are stored

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in the reader system for application when reading the TLDs exposed during the chest CT procedure.

The TLDs used to measure the dose absorbed by the chest during imaging with the Phillips Brilliance 64-slice CT scanner (Phillips, Health Care Bothell, WA, USA) were latter read using the Harshaw reader (Model 3500 with WinREMS, Saint-Gobain Crystals & Detectors Measurement Products, Ohio, USA).

A Photomultiplier tube (PMT) captured light released by TLDs resulting in an output current directly proportional to radiation exposure of TLDs [15] expressed as:

$$Exposure = \frac{ECC \times \text{charge}}{\text{RFC}} \quad (4)$$

Anthropomorphic phantom

A Rando phantom (The Phantom Laboratory, Salem New York, United States of America) Figure 1a) was used in the study. It simulates a male adult person of height 175 cm and mass 73.5 kg. The phantom consists of slices of thickness 2.5 cm that are transacted horizontally, each with holes filled up with pins that are bone equivalent, soft-tissue equivalent or lung tissue equivalent. Lung and tissues (atomic number 7.3) have densities of 0.32 g/cm^3 and 0.985 g/cm^3 respectively. These tissues are ideal for patient dosimetry at photon energies characteristic for CT [11]. Furthermore, the Health Canada Safety Code 35 [16], recognizes use of a phantom data as an alternative for patient data when computing dose length product (DLP) values. Roche *et al.* [17], also established that upon using "standardized patients" the weight criteria becomes insignificant. In this regards a Rando phantom was be used as a good proxy for an adult patient.

Placement of TLDs on chest slice

Ninety-five calibrated TLDs chips (white spots) were implanted on peripheral and centre of chest slice (figure 1b). The central position was denoted C while the remaining four peripheral positions were denoted A, B, D and E. The slice was then securely reassembled to its original position on the Rando phantom (figure 1a).

CT imaging

A Phillips Brilliance 64-slice CT scanner (Phillips, Health Care Bothell, WA, USA) programmed on chest CT protocol (Figure 1c) was used to scan the chest of the Rando phantom with TLDs embedded as described previously. The scan parameters used were as follows: tube voltage (120 kVp), mAs (30-300), scan time (0.5 -10 s) and pitch (0.7-1.4 cm). The activated automatic exposure control (AEC) was crucial in adjusting the current to match the body size of the phantom resulting in appropriate adjustment of mAs and dose reduction.

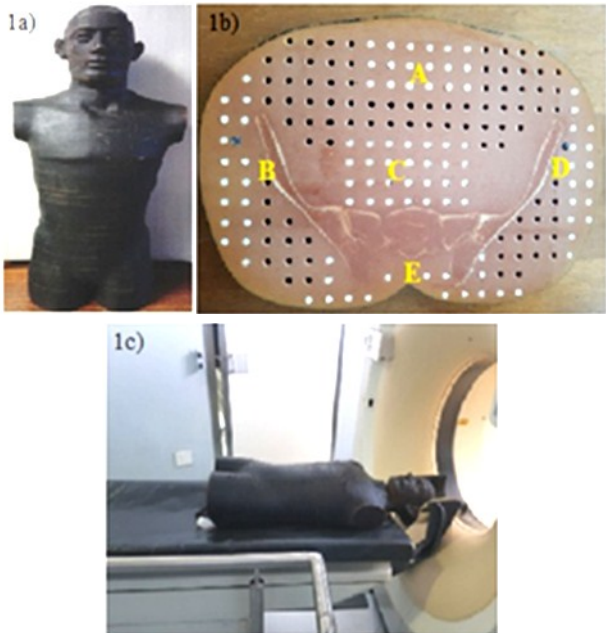

Figure 1. Experimental set up for measurement of radiation dose absorbed by chest during chest computed tomography (CT) examinations. **1a)** shows the Alderson Rando Phantom, **1b)** chest slice indicating positions A-E where TLD chips (white spots) were implanted, **1c)** Alderson Rando phantom placed on supine position ready for measurement of dose absorbed by chest during chest CT procedure.

Dosimetry

The dose absorbed by the TLDs exposed to the X-rays (low dose ionizing radiation) from the Phillips Brilliance 64 CT scanner (Phillips, Health Care Bothell, WA, USA) was read using a reader type LTM (Model 3500 with WinREMS, Saint-Gobain Crystals & Detectors Measurement Products, Ohio, USA). The measured doses were then used to calculate the weighted CT dose

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Index ($CTDL_w$) values using equation 5:

$$CTDI_w = \frac{1}{3}CTDI_{centre} + \frac{2}{3}CTDI_{periphery} \quad (5)$$

where, the weighting factors: 1/3 and 2/3 represents the weighting factors for the central and periphery positions of the chest slice. $CTDL_w$ represents dose in the x (horizontal direction) and y (vertical direction) [18].

Weighted CT dose Index ($CTDL_w$) was used to calculate the volumetric CT dose index ($CTDL_v$) using equation 6 [18]:

$$CTDL_v = CTDL_w \div Pitc \quad (6)$$

where pitch is the table travel per rotation relative to beam collimation.

The $CTDL_v$ is a "fixed" parameter measured in mGy, it gives radiation intensity directed to the patient [19]. It is an indication of absorbed dose in the irradiated slice within the scan range.

Using $CTDL_v$, the DLP was calculated using the equation 7:

$$DLP = CTDL_v \times L \quad (7)$$

where, L is the scan length measured in centimetres (cm) [18].

A DLP value measures the total amount of radiation output from the CT scanner. It is measured in milligreys. centimetres (mGy.cm) [20]. In this study, it was equated to the CT dose absorbed by the chest during chest CT imaging. Lastly, E was estimated using equation 8:

$$\text{Effective dose (E)} = k \times DLP \quad (8)$$

Where k = 0.014, a conversion factor for adult chest as described by International Commission on Radiological Protection publication 103 [10].

Statistical analysis

A Statistical Package for Social Sciences (SPSS) software version 26 was used to obtain the mean of TLDs doses measured at the peripheral and centre of the Rando chest slice. A one-way ANOVA test was used to test statistical significance of the TLD measurements.

RESULTS

Table 1 shows scan parameters (the scan length and pitch) applied during acquisition of three sets of measurements of chest CT dose at the peripheral and centre of the of the Rando chest slice. A one-way ANOVA test confirmed the statistical significance of the three sets of measurements (p = 0.032).

The TLD dose measurements (Table 1) obtained at the centre and peripheral positions of the Rando chest slice irradiated with the Phillips Brilliance 64 CT scanner programmed on adult chest CT protocol were used to measure the dose absorbed by the adult chest (table 2).

The $CTDL_w$ (table 2) is the sum of one third mean slice CT dose measured at central position C and two thirds mean peripheral CT dose measured at positions A, B, D, and E. Diving the $CTDL_w$ by the pitch gives $CTDL_v$, a dosmetric parameter found on display in modern CT scanners. The $CTDL_v$ represents the dose absorbed by the adult chest slice. The product of $CTDL_v$ and the scan length gives the dose absorbed by the adult chest during chest CT procedure. The dose absorbed is represented by the metric called dose length product (DLP), which upon multiplying by the k-factor (k = 0.014) [10] gives the estimated effective dose for the adult chest CT procedure.

Figure 2 shows a comparison of effective dose (2.5 mSv) for chest CT examination obtained in this current with international values. E was estimated multiplying mean DLP value (table 2) with the adult chest conversion coefficient (k = 0.014) provided in ICRP publication 103 [10].

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Table 1. Doses measured with thermoluminescent dosimeters embedded on the centre and peripheral positions on a Rando phantom chest slice irradiated with a Phillips Brilliance 64 CT scanner.

Activity	Pitch	Scan length (cm)	Measured doses					p-value
			Position A (peripheral) (mGy)	Position B (peripheral) (mGy)	Position D (peripheral) (mGy)	Position E (peripheral) (mGy)	Position C (centre) (mGy)	
Measurement 1	0.88	40	2.88	3.99	2.84	2.80	3.99	0.032
Measurement 2	0.88	40	3.53	4.08	4.60	4.55	4.60	
Measurement 3	0.88	40	3.24	4.10	2.90	3.85	3.44	

One-way ANOVA test confirmed that the mean difference in the three sets of measurements (TLD doses) at the centre and peripheral of the chest slice are statistically significant (p = 0.032). The scan parameters (pitch = 0.88 cm and scan length = 40 cm) employed during irradiation of the TLDs with a Phillips Brilliance 64 CT scanner were identical to those used in clinical settings during chest CT procedures.

Table 2. Adult chest CT dose and estimated effective dose in chest computed tomography imaging.

Pitch	Scan length (L cm)	Mean measured CT doses using Thermoluminescent dosimeters embedded on the Rando phantom chest slice (mGy)		$\frac{1}{3}CTDI_{centre}$ (mGy)	$\frac{2}{3}CTDI_{peripheral}$ (mGy)	$CTDI_w$ (mGy)	$CTDI_v$ (mGy)	DLP (mGy.cm)	Effective dose (E) $E=k\times DLP$ (mSv)
		Mean CT doses absorbed at the peripherals positions of chest slice (A, B, D and E)	Mean CT dose absorbed at centre of the chest slice (position C)						
0.88	40	3.61±0.67	4.60±0.31	1.53±0.22	2.4±0.21	3.93±0.30	4.47±0.38	178.8±15	2.5±0.21

weighted CT dose Index ($CTDL_w$) represents sum of a third of the central chest CT dose and two thirds of the chest CT peripheral dose. In modern CT scanners, $CTDL_w$ is quoted as volumetric CT dose index ($CTDL_v$). The $CTDL_v$ represents CT dose absorbed by chest slice. Multiplying $CTDL_v$ by scan length gives dose absorbed chest during chest CT imaging, this value is represented by the dose length product (DLP). The DLP value when multiplied by k-factor (k = 0.014, publish by International Commission on Radiological Protection)10 for adult chest gives an estimate of effective dose in chest CT procedures.

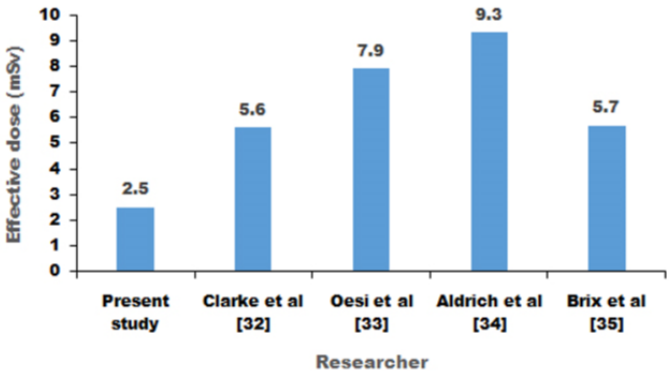

Figure 2. Comparison of estimated effective dose in adult chest computed tomography examination obtained using a Rando phantom and thermoluminescent dosimeters with values in literature. The present study was conducted using a Phillips Brilliance 64-slice CT scanner computed tomography (CT) scanner fitted with fitted with improved Dose Optimization Software (Automatic Exposure Control (AEC)). Comparative estimates were much higher since older models of CT scanners were used. These were not installed with AEC software that automatically reduces tube current in accordance to patient size.

DISCUSSION

Technological development of CT detectors that saw advent of multi-slice CT (multi-detector CT (MDCT)) scanners has made computed tomography an indispensable medical imaging modality. Currently, the demand for CT surpasses that of other radiological imaging modalities. MDCT scanners are renowned for their ability to acquire high resolution three dimensional images in a fractions of seconds [2]. Acquired CT images facilitate confirmation or exclusion of a diagnosis with improved conviction [2]. Furthermore, the availability of high resolution images has played a crucial role in the reduction of exploratory surgeries from 13% to 5%, eventually improving patient care and management. However, the benefit from any imaging modality hinges on sufficient understanding of its technical parameters and appropriate use or application [21].

Good practice in CT imaging requires the use of patient size specific protocols. These should be tailored in accordance to patient size and age

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as well as the region of imaging and often clinical indication. Patient tailored protocols minimize patient dose without affecting the diagnostic quality of acquired images [22]. European 2014 Council Directive 2013/59/ EURATOM advocates that clinicians prescribing ionizing radiation should inform the patients about the risks and benefits of ionizing radiation prior to medical exposures [23].

For patients to understand the risks of CT imaging, the risk must be explained in comparison to a year's effective dose arising from naturally occurring radiation already familiar to them. However, the effective dose is not a directly measurable quantity. Nonetheless, it can be estimated from a CT dose metric called Dose Length Product (DPL) and conversion factors as published by International Commission on Radiological Protection publication 103 [10]. In this study, an anthropomorphic phantom and TLD-100 chips were used as a *proxy* for in vivo measurement of patient chest CT dose. Use of a "standard phantom" was in line with the Health Canada Safety Code 35 [24] and European Commission of 1999 standards [25]. These bodies are of the view that data acquired using a "standardized phantom" maybe used as an alternative for patient data when establishing DLP values.

A DLP value quantifies total amount of radiation received by patient during a single scan. As such, the metric is an indirect method for measurement of absorbed dose [26]. A comparison of DLP value (178.8 ± 15mGy.cm) established in this study with international values showed that 178 < 285 mGy.cm for France [27], 178 < 361 mGy.cm for UK [28], 178 < 550 mGy.cm achieved in USA [20] and 178 < 450 mGy.cm achieved in Australia [29]. A possible explanation for these differences could be that different scan lengths were used. Furthermore, other contributory factors could have been use of different tube voltages. In the event tube voltage could have been the same, then tube current used needed to be analyzed in order to convincingly explain above differences. However, information on tube current was not available for all studies. The data in this study was acquired using a Phillips Brilliance 64-slice CT scanner installed in 2017, while the comparative data was acquired using CT scanners which were 10 or more years older. Modern CT scanners unlike the older versions, are fitted with improved Dose Optimization Software (Automatic Exposure Control (AEC). The latter enables the scanner to adjust the tube current according to patient size on the basis of two techniques that include Automatic Current Setting (ACS) and Automatic Tube Current Modulation (ATCM). The two may be activated jointly or separately [30]. Furthermore, the Philips Brilliance 64 slice CT scanner was installed with iterative reconstruction techniques. A study by Thakur *et al*. [31], established that the use of iterative reconstruction techniques significantly lowers patient doses.

Furthermore, Roche *et al.* [17], established that older CT scanners were not installed with dose saving software. As a result, they deliver high doses to patients compared to the new scanners that come with dose saving software. Figure 2, compares the estimated chest CT effective dose (2.5 ± 0.21 mSv), with international values. From figure 2, it can be observed that 2.5 < 5.6 mSv [32], 2.5 < 7.9 mSv [33], 2.5 < 9.3 mSv [34], 2.5 < 5.7 mSv [35] established by other researchers. The differences are likely to have been greatly influenced by the use of different imaging parameters for the same chest protocol as well as type and age of the CT scanner used.

Lastly, the dependence of the effective dose on DLP values implies that where DLP values were high, the effective doses also becomes high since the effective dose = k × DLP. Furthermore, the DLP values used in estimating the effective dose (figure 2) were acquired from different population backgrounds. In this regard, it is worth mentioning that demographics of people differ from one country to the other, thus influencing scan length differently. The scan length influences the DLP values. The latter in turn influence the estimated effective dose.

Study limitations

The main limitation of the study was use of the phantom to measure chest CT dose. Use of the patient may have been better since it

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incorporates both controllable (imaging technique, tube voltage, tube current) and uncontrollable (patient orientation, collimation and distance) factors. Although use of phantom results in almost similar exposures, it only addresses controllable factors.

CONCLUSION

Effective dose is a single numerical value that relates radiation to stochastic effects. It can also save as a tool for comparing radiation imparted to patients by various CT scanners. However, it is not directly measurable in vivo. As such, this study successfully used an anthropomorphic and TLDs to measure radiation dose delivered on chest by a Phillips Brilliance 64-slice CT scanner (Phillips, Health Care Bothell, WA, USA) programmed on chest CT protocol. Successful measurement of CT dose delivered on the chest facilitated calculation of DLP, a value multiplied by adult chest conversion factor (k = 0.014 mSv $mGy^{-1}.cm^{-1}$) to obtain an estimate of effective dose in chest CT procedures.

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