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TOPIC HIGHLIGHT

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Radiation dose measurements in coronary CT angiography

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Abstract

Coronary computed tomography (CT) angiography is associated with high radiation dose and this has raised serious concerns in the literature. Awareness of various parameters for dose estimates and measurements of coronary CT angiography plays an important role in increasing our understanding of the radiation exposure to patients, thus, contributing to the implementation of dose-saving strategies. This article provides an overview of the radiation dose quantity and its measurement during coronary CT angiography procedures.

Key words: Coronary computed tomography angiography; Dose measurement; Dose quantity; Multislice computed tomography; Radiation dose

Core tip: Various dose parameters are used for measurement of radiation dose associated with coronary computed tomography (CT) angiography. It is important to be aware of the dose quantity and measurement in order to achieve the low-dose coronary CT angiography protocol. This article provides an in-depth review of the dose quantity and dose measurement parameters that are commonly used in coronary CT angiography.

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INTRODUCTION

The introduction of latest multi-slice computed tomography (MSCT) technology has emerged as a useful diagnostic imaging modality for the noninvasive assessment of coronary artery disease. The recent advances in the spatial and temporal resolution with thinner detector widths and the low helical pitch values being required for data acquisition in cardiac computed tomography (CT), mainly in retrospective ECG-gating coronary CT angiography (CCTA) mode, however, resulted in increased radiation dose. Compared with plain film radiography, CT examination produces significant higher radiation dose, resulting in a marked increase in radiation exposure to patients. However, the main concern of exposure to ionizing radiation is the potential risk of radiation-induced cancer, and this has raised serious concerns in the literature[1].

Risks associated with radiation exposure are manifested as either deterministic or stochastic effects. Deterministic effects occur when the radiation dose reaches a threshold dose level. The threshold level in deterministic effects varies in different subjects and the damages are significantly related to the amount of dose received. Skin injury, hair loss and cataract are the examples of deterministic effects associated with radiation dose. For example, skin injuries range from skin erythema, moist desquamation, epilation, laceration to necrosis if the skin is exposed to radiation dose beyond the threshold level of 2 Gy[2]. On the other hand, stochastic

effects can be defined as an effect that occurs without any dose threshold. It happens at all time and the damages are not depending on the amount of dose received. Ionizing radiation-induced cancer and genetic changes belong to the stochastic effects. However, previous studies have reported that the increment of radiation dose could increase the chance of developing cancer[3].

Radiation dose estimates for cardiac CT examinations are best expressed as the CT volume dose index ($CTDI_{vol}$), dose-length product (DLP) and effective dose (E). These parameters are precisely defined to allow comparisons of the radiation doses among different CT imaging protocols. The dose received by a patient from a given CT examination is commonly estimated using $CTDI_{vol}$ or DLP value available on the scanner console[4]. Other than $CTDI_{vol}$, DLP and E, there were several radiation dose parameters widely used in CT study in order to measure or quantify the radiation dose of CT scanning procedure. Therefore, the purpose of this article is to provide an overview of the radiation dose quantity and its measurement during CCTA procedures.

RADIATION DOSE QUANTITY AND MEASUREMENTS

CT dose index

The fundamental radiation dose parameter in CT is the computed tomography dose index (CTDI). $CTDI_{100}$ is a measured parameter of radiation exposure which is more convenient than the CTDI and it is regarded as the measurement of choice performed by medical physicists in the clinical setting. Initially, $CTDI_{100}$ is measured by a 100-mm long pencil-shaped ionization chamber in two different cylindrical acrylic phantoms (16 and 32-cm diameter) which was placed at the iso-center of the CT scanner. Most manufacturers use a 16 cm phantom for head and 32 cm phantom for body examinations during CTDI calculation[5]. The $CTDI_w$ is the weighted average of the $CTDI_{100}$ measurements at the center and the peripheral locations of the phantom. This parameter reflects the average absorbed dose over the two-dimensions (x and y dimensions) of the average radiation dose to a cross-section of a patient's body.

The $CTDI_{vol}$ is different from $CTDI_w$ where $CTDI_{vol}$ represents the average radiation dose over the volume scan (x, y, and z directions) while $CTDI_w$ represents the average exposure in the x-y plane only. $CTDI_{vol}$ is the weighted CTDI divided by the pitch, or $CTDI_{vol} = CTDI_w$/pitch and it is measured in mGy. The $CTDI_{vol}$ is now the preferred radiation dose parameter in CT dosimetry. $CTDI_{vol}$ is commonly used in clinical practice due to its accessibility to the radiologists and CT operators as it specifies the radiation intensity used to perform a specific CT examination and not to quantify how much radiation that each patient receives from the CT examination[6]. Rather than the dose to a specific patient, $CTDI_{vol}$ is a standardized index of the average dose delivered from the scanning series. $CTDI_{vol}$ is available to be displayed on the control console. This allows the clinicians or operators to compare the radiation doses that patient receive from different imaging protocols. $CTDI_{vol}$ can also be used in turn to determine DLP.

Dose-length product

The dose-length product (DLP) is an indicator of the integrated radiation dose of an entire CT examination. The DLP is an approximation of the total energy a patient absorbs from the scan. It incorporates the number of scans and the scan width, *e.g.* the total scan length, while in contrast $CTDI_w$ and $CTDI_{vol}$ represent the radiation dose of an individual slice or scan. Therefore, DLP increases with an increase in total scan length or variables that affect the $CTDI_w$ (*e.g.* tube voltage or tube current) or the $CTDI_{vol}$ (*e.g.*, pitch). Because scan length is expressed in centimeters, the SI unit for DLP is mGy·cm. Similar to $CTDI_{vol}$, DLP is also available on the operator's console.

Absorbed dose and equivalent dose

Absorbed dose is an amount of energy that is deposited in a unit of mass of matter (tissue). It is measured in gray (Gy) with 1 Gy equivalent to 1 joule per kilogram. Each type of ionizing radiation produces different biological effect. For instance, the biological effect on tissue which is exposed to 1 Gy α radiation is more harmful than 1 Gy of X-rays. This is because α particles are more heavily charged and slower than x-rays. Therefore, α particles lose much more energy along the travel path before reaching the target[7]. However, the quantity of equivalent dose is used to compare all types of ionizing radiation equally on the biological effect. Equivalent dose is measured in Sievert (Sv). Equivalent dose is obtained by multiplying the absorbed dose with the radiation weighting factor (Table 1).

Effective dose

The most important parameter in CT imaging is the effective dose (E), which is valuable in assessment and comparison of the potential biological risk of a specific examination. E is a sum of equivalent doses in organs of the body that are considered radiosensitive. It is a uniform whole-body dose that has the same nominal radiation risk of carcinogenesis and induction of genetic effects as any given non-uniform exposure[8]. Each organ in human body has different radiosensitivity with some organs more sensitive to the risk of damage than the others. E can be estimated by multiplying each equivalent dose by a relative organ with the tissue weighting factor related to the risk associated with that organ and summing overall exposed organ. International Commission on Radiological Protection (ICRP) publication 103 released in 2007 has recommended values for the tissue weighting factors with major changes different from the previously published ICRP publication 60[9,10] (Table 2).

The SI unit of estimating E is the sievert (Sv) or millisievert (mSv). The weighting factors used for individual

Table 1 Radiation weighting factor for various type and energy range

Type and energy range	Radiation weighting factor, W_R (ICRP-60)
Photons, all energy	1
Electrons, muons, all energy	1
Neutrons < 10 keV	5
10 eV-100 keV	10
> 100 keV-2MeV	20
> 2–20 MeV	10
> 20 MeV	5
Protons > 2 MeV	5
Alpha particles, fission fragments and heavy nuclei	20

Adapted from Ng *et al*[7]. ICRP: International Commission on Radiological Protection.

Table 2 Tissue weighting factor comparison between International Commission on Radiological Protection publication-103 and publication-60

Organs	Tissue weighting factor, W_T	
	ICRP-103	ICRP-60
Colon	0.12	0.12
Lung	0.12	0.12
Red bone marrow	0.12	0.12
Stomach	0.12	0.12
Breast	0.12	0.05
Gonads	0.08	0.20
Bladder	0.04	0.05
Liver/Oesophagus	0.04	0.05
Thyroid	0.04	0.05
Bone surface/skin	0.01	0.01
Brain	0.01	-
Salivary glands	0.01	-
Remainder tissues	0.12[1]	0.05[2]

Adapted from Ng *et al*[7]. [1]Remainder tissues in International Commission on Radiological Protection (ICRP)-103: adrenals, kidneys, muscle, small intestine, pancreas, spleen, thymus, uterus/cervix, prostate, extra-thoracic region, gallbladder, heart, lymphatic nodes and oral mucosa; [2]Remainder tissues in ICRP-60: adrenals, kidney, muscle, small intestine, pancreas, spleen, thymus, uterus, upper large intestine and brain.

tissues are based on a statistical analysis of the increase in the long-term incidence and mortality for cancer determined from a life span study of the survivors in Japan during the atomic bomb explosion[11-13]. Usually, tabular data of conversion coefficients are available to estimate E from entrance skin dose for radiography[14,15], from dose area product (DAP) for fluoroscopy[16,17], or from $CTDI_{vol}$ or DLP for CT[18]. The goal is to convert the higher radiation doses delivered to a small portion of the body into an equivalent uniform dose to the entire body that carries the same biological risk for causing radiation-induced fatal and nonfatal cancers.

The *E* can be estimated by multiplying the DLP with a conversion coefficient factor (*E*/DLP), *k* (mSv/mGy per centimetre). The *E*/DLP value of 0.026 or 0.028 mSv/mGy per centimetre was applied for coronary CT study since this value was likely to be more accurate for estimation of radiation dose associated with cardiac CT compared to the chest CT (0.014 or 0.017 mSv/mGy per centimetre)[10,19,20]. If no dose-saving strategy is applied, it is estimated that effective doses of coronary CT angiography may reach up to 30 mSv in patients undergoing cardiac CT imaging, thus, there is potential risk of associated radiation-induced malignancy[21].

Gosling *et al*[20] compared the effective dose using the latest ICRP 103 tissue-weighting factors with that calculated with previously published chest conversion factors. Their results showed that the use of chest conversion factors (0.014-0.017) significantly underestimated the effective dose when compared to the dose calculated using the conversion factor of 0.028. A conversion factor of 0.028 would give a better estimation of the effective dose from prospectively ECG-triggered coronary CT angiography. Appropriate conversion factors are needed to accurately estimate effective dose. A conversion factor of 0.014 or 0.017 is commonly used in many cardiac CT studies to estimate the effective dose associated with coronary CT angiography, thus, this could lead to variations in the reported effective dose. As a result, the DLP or $CTDI_{vol}$ is recommended to compare the radiation exposure of coronary CT angiography[22].

Background equivalent radiation time

Background equivalent radiation time (BERT) is used to explain the dose to the general public without complicated scientific units, terminology or concepts. It converts the radiation dose to an equivalent period of natural background radiation in days, weeks, months or years to which the entire population is exposed every day from natural radioactive substance in the air, internal, terrestrial, cosmic and environment. For example, it is more likely for patient to easily understand that "your chest X-ray dose is about equal to 3 d of background radiation" rather than "you have received 0.02 mSv for your chest X-ray examination"[7]. BERT is not used to provide a high level of diagnostic accuracy, but to relieve anxiety about radiation by giving an understandable and satisfactory answer (Table 3)[23].

Entrance skin dose

Entrance skin dose is an amount of energy imparted per gram of tissue at the entrance surface. It is also known as surface absorbed dose (SAD). About 1 Gy is equal to 1 millijoule per gram of energy deposited by the X-rays. Entrance skin dose can be obtained by multiplying the radiation exposure measured in the air at the skin by a factor, f for the tissue. The f factor is a quantity of radiation dose exposure conversion measured in the air (coulomb per kilogram at the standard temperature and pressure) to an equivalent radiation dose absorbed in tissue (grays) at the same location. However, entrance skin dose is not an indicator to measure radiation risks except for skin erythema, but it is useful for organ dose calculation especially in a computer-based program that is involved with Monte Carlo simulations[14,15].

Table 3 Estimated effective doses for diagnostic medical exposures associated with background equivalent radiation time and lifetime fatal cancer risks from National Radiological Protection Board

X-ray examination	Estimated effective dose (mSv)	BERT[1]	Fatal cancer risk per examination[2]
Limbs and joints (exclude hip)	< 0.01	< 1 d	1 in a few millions
Dental (single bitewing)	< 0.01	< 1.5 d	1 in a few millions
Dental (panoramic)	0.01	1.5 d	1 in 2 million
Chest (single PA)	0.02	3 d	1 in a million
Skull	0.07	1 d	1 in 300000
Cervical spine	0.08	2 wk	1 in 200000
Thoracic spine	0.7	4 mo	1 in 30000
Lumbar spine	1.3	7 mo	1 in 15000
Abdomen	0.7	4 mo	1 in 30000
Hip	0.3	7 wk	1 in 67000
Pelvis	0.7	4 mo	1 in 30000
Intravenous urography	2.5	14 mo	1 in 8000
Barium swallow	1.5	8 mo	1 in 13000
Barium meal	3	16 mo	1 in 6700
Barium follow-through	3	16 mo	1 in 6700
Barium enema	7	3.2 yr	1 in 3000
CT head	2	1 yr	1 in 10000
CT chest	8	3.6 yr	1 in 2500
CT abdomen/pelvis	10	4.5 yr	1 in 2000

Adapted from Ng *et al*[7]. [1]Natural background radiation based on Australia average = 2.4 mSv per year; [2]Appropriate lifetime risk for patients from 16-69 years old: paediatric = 2x; geriatric = 5x. BERT: Background equivalent radiation time.

Critical organ dose

Critical organ dose (COD) is more commonly reported in the literature for radiologic examinations. Critical organ dose refers to the energy deposited per unit mass to individual critical organs for which the radiosensitivity and radiation dose are high. Its unit of measurement is usually milligrays, which is equivalent to millijoules per kilogram. COD can be used to assess the risks of irradiation beyond cancer induction for certain organs; for example, other potential biological effects can include skin erythema, cataracts, fetal abnormalities, haematologic effects, vascular damage, and effects on the central nervous system.

Critical organ dose may be determined by other dose descriptors, such as entrance skin dose or dose area product, by using tables or software programs that are based on Monte Carlo calculations for standard patient sizes[14,15]. Also, the critical organ dose values for various organs, along with their corresponding weighting factors, can be used to calculate the effective dose[9,24]. In clinical practice, knowledge of organ doses and the carcinogenic sensitivity of certain organs can lead to better collimation and patient positioning to reduce the risks from exposure to radiation.

Diagnostic acceptable reference level

Diagnostic acceptable reference level is also known as diagnostic reference level (DRL). DRL values are published based on the nationwide evaluation of X-ray trends surveys[23,25]. The data values can be used as a reference point to ensure that all current clinical practice involving radiation in radiological investigations are safe. However, ESD, DAP, or $CTDI_{vol}$ values that are greater than those of DRL may be attributed to the patient's size, the complexity of the clinical case, equipment malfunctions, or suboptimal protocols. Some of the higher values may be unavoidable; however, many of the higher values can be avoided. When patient doses appear to be above those of DRL, especially when they are consistently higher, investigation and assessment are required. If suboptimal protocols or equipment deficiencies are the cause of the higher dose levels, necessary strategies must be undertaken to reduce the radiation dose.

Radiation dosimeter

Radiation dose in clinical practice can be measured accurately by using a dosimeter. There are a number of dose measurement tools with different methods being used to measure the radiation dose absorption. The value of absorbed dose is determined indirectly by measuring the radiation effect through ionization of air, fogging of photographic emulsion, thermoluminescence, scintillation and ionization of a semiconductor. However, the most commonly used method in radiation dosimetry is thermoluminescence dosimeter (TLD)[26].

Thermoluminescence phenomenon

Thermoluminescence is a condition where the light is emitted from a heated crystalline material which is made up of lithium fluoride (LiF) or calcium fluoride (CaF_2) phosphors. When the crystalline is exposed to the radiation, electrons in the crystal are pulled out from valence band to the conduction band by a small amount of energy. However, without enough energy, some of the electrons are trapped into one of the isolated levels provided by impurities in the crystal. It will remain immobilized at that state until energy is supplied to release it (usually by heat). Thus, the electrons leave a positive hole in the valance band. By heating the crystal, the trapped elec-

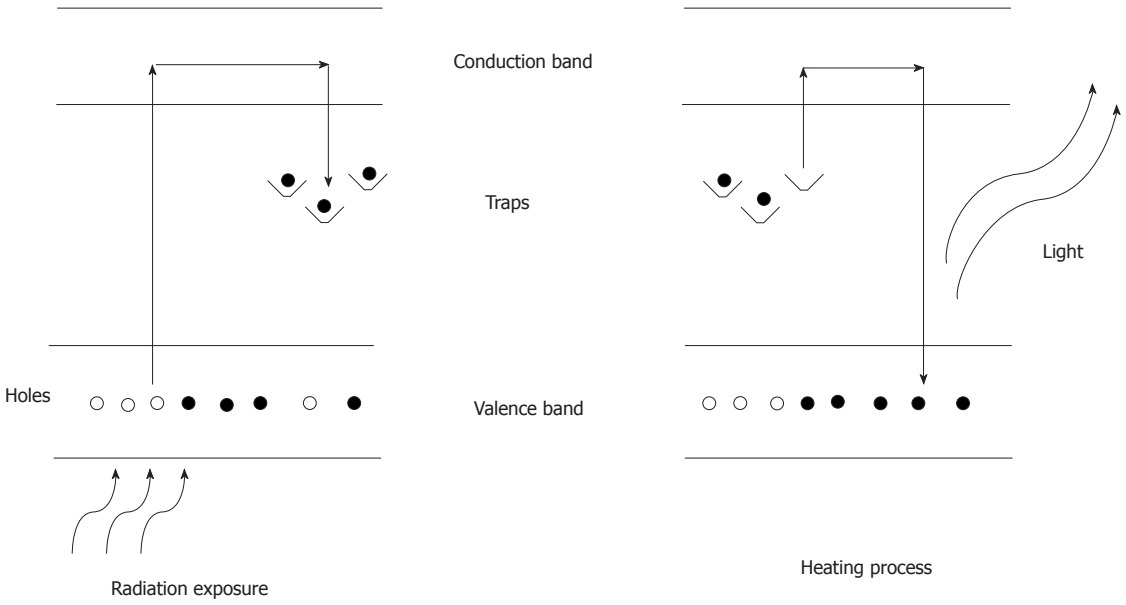

Figure 1 Process of light emission from the radiation exposure in the thermoluminescence phenomenon.

trons will elevate and return to the valence positive hole. A photon of visible light is emitted during the process of returning electrons from the trap to the valence band (Figure 1)[11]. The total light emitted is counted where the measurement for the number of trapped electron indicates the absorbed radiation. Surprisingly, it can be used even after a month of storage.

Several types of TLD are commercially available for a wide range of applications. For instance, LiF: Mg, $Li_2B_4O_7$, $CaSO_4$: Dy, Al_2O_3, CaF_2: Dy and CaF_2: Mn[27]. In diagnostic radiology, LiF: Mg, Ti or usually known as TLD-100 was chosen for dosimetry purposes in clinical radiation measurement. In fact, it was the first material used in diagnostic radiology and one of the most utilised materials when compared to others[28]. TLD with LiF: Mg, Ti material is chosen because of the physical shape which is small, light and convenient for local measurement during the radiological examinations. Apart from physical appearance, it is able to measure entrance surface absorbed dose at the reference point at specific organs without obscuring an image due to the radiolucency specification[27]. Moreover, it has high reproductive capability, thus it can be used repeatedly. The materials are sensitive to detect radiation exposure in a range between 10 μGy and 10 Gy, in addition to having a good linear relationship between thermoluminescence readout value and dose absorption up to 1 mrad.

CT dose measurement

Effective dose in CT can be easily estimated by a simple calculation through multiplying the DLP with a conversion coefficient factor (E/DLP). Huda, Ogden, and Khorasani in their study introduced a new approach to determine the E[8]. They suggested that E can be calculated from DLP by using ImPACT software package which is based on Monte Carlo simulation performed by the National Radiological Protection Board[29]. Yet, the accuracy of this system is undisputable when Huda, Ogden, and Khorasani compared those E calculations with other software packages like CT-expo and ImpactDose. As a result, there were approximately 5% differences between E/DLP values according to each software package and it was not statistically significant[8]. CT-Expo is a program run on Monte Carlo dosimetry data while ImpactDose is a personal computer based-program that calculates ED values for arbitrary scanning parameters and anatomic ranges[30]. However, the E values still can be calculated manually by multiplying the DLP values with the conversion coefficient factor in CT imaging based on individual organs and tissue weighting factors published by the ICRP 103[10,16,31]. Using CT dose reporting packages is an advantage because they are easy to use and produce quick results. However, it must be recognised that there are deviations between the different software packages, and users should understand this and be familiar with different terminologies used in order to provide accurate dose reporting for a consistent comparison[30].

In conclusion, it is important to be aware of the amount of radiation dose produced from cardiac CT scanning. The quantification of the radiation dose is a crucial issue that must be addressed by both practitioners and the operators in determining the correct and accurate dose measurement. With sufficient knowledge of radiation dose terminology and dose quantification, the understanding of radiation dose safety and radiation awareness will be accordingly increased when performing coronary CT angiography examinations. Various dose-saving strategies have been undertaken in the past decade to lower radiation exposure to patients who undergo coronary CT angiography, with effective dose ranging from 10 mSv to as low as 1 mSv. Details of these dose reduction techniques will be discussed in Part Ⅲ of this series.

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