ESTABLISHMENT OF LOCAL DIAGNOSTIC REFERENCE LEVELS (DRLs) FOR CT EXAMINATION AND ASSESSMENT ON CURRENT CLINICAL CT PRACTICE AT AMDI, USM

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by

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LIST OF SYMBOLS

c	Centre measurement
d	Nominal slice thickness
kV	kiloVolt
kVp	Peak kiloVoltage
K(z)	Air kerma in the phantom
L	Scan length
mAs	Milli-Ampere-Second
mGy	Milli-Gray
mGy.cm	Milli-Gray- Centimetre
n	Number of tomographic sections imaged in a single axial scan
p	Periphery measurements
P	Pitch factor value
T	Nominal width of the tomographic section along the centre of the z-axis
z	Position at given location

LIST OF ABBREVIATIONS

AEC	Automatic Exposure Control
AP	Abdomen-Pelvis
AMDI	Advanced Medical and Dental Institute
CT	Computed Tomography
CTDI	Computed Tomography Dose Index
$CTDI_{100}$	CTDI using 100 mm ionization chamber
$CTDI_c$	Center Computed Tomography Dose Index
$CTDI_p$	Peripheral Computed Tomography Dose Index
$CTDI_{vol}$	Volume Computed Tomography Dose Index
$CTDI_w$	Weighted Computed Tomography Dose Index
DLP	Dose Length Product
DRL	Diagnostic Reference Level
DSCT	Dual-Source Computed Tomography
ICRP	International Commission on Radiological Protection
MDCT	Multi Detector Computed Tomography
MSCT	Multi Slice Computed Tomography
MOH	Ministry of Health Malaysia

MRI	Magnetic Resonance Imaging
NDRLs	National Diagnostic Reference Levels
NTAP	Neck-Thorax-Abdomen-Pelvis
PACS	Picture Archiving and Communication System
QA	Quality Assurance
SSCT	Single Slice Computed Tomography
TA	Thorax-Abdomen
TAP	Thorax-Abdomen-Pelvis
USM	Universiti Sains Malaysia

LIST OF APPENDICES

PENUBUHAN ARAS RUJUKAN DIAGNOSTIK TEMPATAN (DRLS) UNTUK PEMERIKSAAN CT DAN PENILAIAN AMALAN CT KLINIKAL SEMASA DI AMDI, USM

ABSTRAK

Fokus utama kajian ini adalah untuk mencadangkan dan menubuhkan aras rujukan diagnostik tempatan (LDRLs) bagi pemeriksaan CT yang paling kerap dilakukan di Institut Perubatan dan Pergigian Termaju (IPPT), Universiti Sains Malaysia (USM), Pulau Pinang, Malaysia. Sebanyak 1444 pemeriksaan CT bermula dari Januari 2015 hingga Disember 2018 disertakan ke dalam tinjauan rektrospektif dos CT. Berdasarkan protokol-protokol pemeriksaan CT, LDRLs yang dicadangkan telah ditubuhkan pada persentil ke-50 (dos yang boleh dicapai, AD) dan ke-75 (DRLs) daripada taburan indeks dos CT ($CTDI_{vol}$, $CTDI_w$, dan produk panjang dos, DLP). Perbandingan antara LDRLs yang telah ditubuhkan dengan DRLs kebangsaan (NDRLs) daripada Kementerian Kesihatan Malaysia (KKM) dan DRLs antarabangsa semasa turut dilaksanakan. Pemeriksaan CT yang paling kerap dilakukan di IPPT adalah CT toraks abdomen pelvis (TAP) (46.95 %), diikuti dengan CT pelvis (15.44 %), CT abdomen pelvis (AP) (10.46 %), dan CT otak (8.31 %). Kebanyakan LDRLs melepasi NDRLs dan DRL. Daripada audit dos radiasi, hampir semua data dos CT pada tahun 2018 melepasi LDRLs dan pemeriksaan CT TAP, pelvis, AP, otak, dan abdomen dikesan menghasilkan data dos CT yang lebih tinggi. Tidak terdapat perbezaan ketara (nilai-P = > 0.05) antara kebanyakan dos data CT tahun 2018 yang melepasi dan di bawah LDRLs yang dicadangkan. Faktor utama yang menyumbang kepada dos yang tinggi adalah disebabkan oleh jumlah siri dedahan yang tinggi, arus tiub (mAs) yang tinggi, dan kesilapan ketika memposisi pesakit. LDRLs harus

ditubuhkan untuk mengurangkan dedahan radiasi yang tidak wajar dan mengoptimum dos CT kepada pesakit. Sebagai penambahbaikan untuk amalan tempatan semasa, taburan dos tempatan haruslah disemak secara berkala.

ESTABLISHMENT OF LOCAL DIAGNOSTIC REFERENCE LEVELS (DRLS) FOR CT EXAMINATION AND ASSESSMENT ON CURRENT CLINICAL CT PRACTICE AT AMDI, USM

ABSTRACT

The main focus of this research is to propose and establish the local diagnostic reference levels (LDRLs) for the most frequent CT examinations performed at Advanced Medical and Dental Institute (AMDI), Universiti Sains Malaysia (USM), Penang, Malaysia. A total of 1444 CT examinations from January 2015 until December 2018 were included in a retrospective CT dose survey. Based on CT examination protocols, the proposed LDRLs were established at 50th (achievable dose, AD) and 75th (DRLs) percentile of CT dose index distribution which are CTD_{Ivol}, $CTDI_w$, and dose length product, DLP. Comparison between established LDRLs with the national DRLs (NDRLs) from Ministry of Health (MOH) Malaysia and other recent international DRLs were performed, respectively. The most frequent CT examinations performed at AMDI were thorax-abdomen-pelvis (TAP) CT (46.95 %), trailed by pelvis CT (15.44 %), abdomen-pelvis (AP) CT (10.46 %), and brain CT (8.31 %). Most of the LDRLs surpassed the NDRLs and other recent international DRLs. From the radiation dose audit, most of the 2018 CT dose data exceeded the LDRLs; TAP, pelvis, AP, brain, and abdomen CT examinations were noted to have higher CT dose data. There were no significant differences (P-values = > 0.05) between most of 2018 CT data that exceed and below the proposed LDRLs. The main factors that contribute to higher dose were due to higher number of CT sequences, higher tube current (mAs) settings, and error in patient positioning. The LDRLs must be established to reduce unjustified radiation exposure incidence and to optimise the

CT dose to the patients. For improvement of current local practice, the local dose distribution must always be revised regularly.

CHAPTER 1 INTRODUCTION

1.1 Background of the study

An x-ray machine that can acquire anatomical structures images from various angles is called computed tomography (CT) scanner. An image that demonstrates a cross section of the region being acquired was produced by the data that being processed by a computer. The external and the internal region of the body can be seen on the CT axial scan. Numerous aspects of the region being acquired can be seen since the CT examination can develops slices of the images. Screen monitor is used to display the CT images, or it can be printed on films for further investigation. A CT scanner that has an ability to move its x-ray tube around the patient like a coil pathway and permits constant data with no gaps between the images is called a spiral or helical CT. A specific contrast media that can be ingested as liquid or introduced into a vein can made the CT image even clearer. Rapid advancement of technology had allowed CT scan as a diagnostic equipment with greater spatial resolution and less scanning times contributing to overly increased number of clinical utilisations such as CT urography, CT angiography, and many more (Power et al., 2016).

Due to rapid development of technology in diagnostic imaging, CT imaging had been utilised more commonly nowadays, as it allows physicians to instantly diagnose the diseases, accurately, and harmlessly in comparison to different imaging modalities. In contemplation of evaluating the hazard associated with ionising radiation exposure to the patient, particulars factor should be accustomed. Compared to other conventional x-ray imaging, CT imaging is well known by its character to contribute larger radiation doses. This study contemplated to assess the patient doses and to establish the local diagnostic reference level (LDRL) for all CT examinations performed at Advanced Medical and Dental Institute (AMDI), Universiti Sains Malaysia (USM).

This study will provide a better understanding and awareness on dose optimisation among imaging staff through the establishment of local diagnostic reference levels (DRLs) for all common CT examinations performed at AMDI, USM. Besides that, this study will also help to improve the local dose allocation by minimising the frequency of unjustified radiation exposure and suggested corrective method for radiation dose that exceeded the reference level at AMDI, USM. For the institutional radiation protection program, it is also crucial to document the local dose distribution and establish the DRLs.

1.2 Problem Statement

The rising numbers of radiology examinations using CT imaging has been aware lately. For that reason, it has been expected that the cumulative radiation doses from medical exposure will also be increased proportionately to an increasing of radiology examination and procedure number. Based on United Nations Scientific Committee's Effect of Atomic Radiation's (UNSCEAR) report, the total effective dose from CT examinations of whole populations have slightly elevated from 41% (1991 – 1996) to 43% (1997 – 2007) of the total annual population dose (UNSCEAR, 2010). In addition, for the past 10 years; the amount of CT usage have been raised rapidly in the USA and elsewhere (Pearce et al., 2012). Furthermore, there was further proof that showed that CT scan undoubtedly demonstrates higher dose compared to other imaging modalities (Smith-Bindman et al., 2015). Those who were undergoing multiple CT examinations had 6.8% cancer-related deaths and this statistic had elevated concerns from the media and the literature (Shah et al., 2013).

According to several studies, CT is frequently utilised modality for brain imaging (Langner, 2015). Besides that, it has been reported that 39% of patients received cumulative radiation exposure from repeated exposures of brain CT (Mettler et al., 2000; Lee et al., 2007; Mettler et al., 2009). International Commission on Radiological Protection (ICRP) introduced the term effective dose (ED) as an index that indicates the hazard of non-uniform exposure that consider equivalent whole-body exposure and the organs or tissues radiosensitivity (ICRP, 2007). According to 42 CT dosimetry studies, brain CT examination has wider range of ED values (0.3 – 8.2 mSv) (Pantos et al., 2011). To evaluate the current approach in routine clinical practice and to optimise the use of radiation exposure in AMDI, USM; a comparative study of local DRLs with NDRLs was performed. Besides that, this study also helps to improve local dose allocation by minimising the frequency of unjustified radiation exposure and suggested corrective method to decrease the radiation dose in AMDI, USM.

1.3 Aims and Objectives

To assess the current CT practice by the establishment of local DRLs for CT examination at AMDI, USM, Penang.

Specific Objectives

i. To survey the CT patient's attributes on the CT dose distribution at AMDI, USM.
ii. To establish a local DRLs for CT imaging in AMDI, USM.
iii. To compare the established DRLs with the national DRLs provided by MOH and international DRLs.
iv. To assess the current local practice on acquisition protocol settings for CT examination at AMDI.

1.4 Significance of Study

The significance of this study is to provide a better understanding and awareness on dose optimisation as the dose guideline that will be beneficial for the establishment of DRLs for all CT examinations performed at AMDI, USM. This study also significant to evaluate the current clinical routine and to optimise the use of radiation exposure at AMDI, USM. Besides that, this study also helps to improve local dose allocation by minimising the frequency of unjustified radiation exposure and suggested corrective method to decrease the radiation dose in AMDI, USM.

1.5 Limitation of Study

The constraint of this study is the limited sample size for certain CT examinations or protocols that restricted the establishment of local DRLs (the minimum sample size is 10 patients as practiced by study in Ireland) (Lee et al., 2007; Foley et al., 2012). Besides that, the scan coverage for several CT examination performed at AMDI, USM were varied due to the extension of the pathologies. The comparative study on the dose data was performed for only few selected CT protocols depending on the available CT protocols published by Ministry of Health (MOH) in NDRLs since the CT protocols are not standardised and the NDRLs did not specify the detailed procedure for each reference value. Another limitation of this study is the current Malaysian DRLs used for this study are based on the dose reviews documented from 2007 to 2009, which are outdated and comprised of dose data obtained from single slice computed tomography (SSCT) scanner. Theoretically, the doses received from multi-slice computed tomography (MSCT) scanner are slightly higher than SSCT scanner. Some of CT examinations/protocols are also not updated in NDRLs such as brain-neck, neck, neck-thorax, neck-thorax-abdomen-pelvis, thorax-abdomen-pelvis, thorax-abdomen, and abdomen- pelvis. Another limitation of this study is the availability of all patient's weight data was limited. This is because the practice of recording patient's weight for CT examinations was only made compulsory at this centre starting on the year 2018. All CT examinations at Imaging Unit, IPPT, USM utilise the automatic exposure control (AEC) mode during the exposure. It is a feature in recent CT technology that aims to automatically modulate the tube current to adjust variation in attenuation based on patient size and shape (McCollough et al., 2006). Several studies indicate that patient's body habitus and weight have significant impact on CT radiation dose including CTDI and DLP dose metrics when AEC is employed

(Mulkens et al., 2005; Schindera et al., 2008; Israel et al., 2010). Thus, patient weight and size are the important reference parameters used during the modification of exposure setting for dose optimisation.

CHAPTER 2 LITERATURE REVIEW

2.1 Computed Tomography

Computed tomography (CT) is an effective clinical modality for the diagnosis which has capability to supply high quality three dimensional images and has few important advantages such as allowing rapid images acquisition, more precise diagnosis, and prevention of unnecessary surgical procedures (Foley et al., 2012). In order to decide internal bleeding, investigating cancer cases, and planning for surgical; CT scan has become the baseline for a variety of clinical examinations (RehaniandBerry, 2000; ICRP, 2001; Kalra et al., 2004). Besides that, one of the advantages of CT is its affordability compared to other modality like magnetic resonance imaging (MRI) which is way more expensive (Yu et al., 2009).

Rapid advancement of CT technology had introduced multi-slice CT scanner (MSCT) which can produce higher image resolution, better detection of smaller abnormalities, and faster image acquisition in comparison to the older single-slice CT scanner (SSCT). In comparison to radiography and ultrasound imaging technique, CT imaging has the highest sensitivity (95%) and specificity (98%) to evaluate urinary stone (Abramson et al., 2000). Besides that, invasive angiography has nearly been replaced by CT angiography as the primary examination because of its clinical value (McCollough et al., 2015). CT imaging has nearly made many of preliminary surgical procedure obsolete where it has reduced the demand for trauma surgery from 13% to 5% since its introduction in the 1970s. Besides that, major usage of CT imaging in medical procedure has been confirmed to reduce the total number of patients needing ward entrance (Rosen et al., 2000; Rosen et al., 2003).

The basic principles of CT system can be divided into three phases which are data acquisition, image reconstruction, and image display (Romans, 2011). Data acquisition can be defined as the process of collecting x-ray transmission that passed through the patient and fall onto the electronic detectors (Seeram, 2015). The x-ray photons are produced in x-ray tube inside the CT gantry when rapid-moving electrons from the tube cathode hit the metal target (anode) resulting the electromagnetic energy. By heating up the filaments inside the x-ray tube, the electrons are ejected and produce x-ray photons. The high tube voltage (kV) generated by generator will be transmitted to the x-ray tube, that will propel the electrons from tube filament to the anode. The focal spot is the area on the anode that will be struck by electrons to produce x-ray beam. The tube current (mA) controls the quantity of the propelled electrons that will strike the target and the number of x-ray photons (beam intensity) produced. The kVp setting controls the energy of the electrons that strike the target and penetrating power of the x-ray photons. Thus, increasing the kVp will increase the x-ray beam's intensity and quality (Romans, 2011). During the scanning, the x-ray tube and detectors will rotate around the patient that is positioned at the isocentre of the CT gantry. The x-ray photons will pass through the patient's body, attenuated and will be measured by the detectors (Seeram, 2015). The detected energy of the x-ray photons will be converted into light by the detector that made -up of solid-state scintillation material. After that, the light intensity will be converted into electrical current. Data acquisition system will sample and convert the electrical current into digital signal before it will be transmitted to the central processing unit (CPU) (Romans, 2011).

Image reconstruction can be defined as mathematical techniques utilised by the CPU to reconstruct the CT image. Recent CT scanners utilise iterative reconstruction algorithm or also known as algebraic reconstruction technique (Seeram, 2015). An iterative reconstruction involves few steps during image reconstruction including image assumption, computing projections from the images, comparing images with the original projections data, and lastly updating the image differences between the calculated and the actual projections (Romans, 2011). For spiral or helical CT scanners, filtered back projection and interpolation algorithms is used for fan beam-image reconstruction and also known as analytical reconstruction algorithm (Seeram, 2015). The system accounts for attenuation properties of each ray sum and correlates it to the position of the ray, which is called as attenuation profile. Back projection is a process where the data is converted from the attenuation profile to a matrix. However, the main disadvantage of back projecting data is the blurring effects and production of streak artifacts if the projection is limited. The filtering process is applied on the scan data prior to back projection reconstruction to minimise streak artifacts and this technique is known as filtered back projection (FBP).

Iterative reconstruction maintains image quality and reduced radiation dose compared to normal-dose filtered back-projection. Willemink et al., (2013) found that iterative reconstruction can reduce radiation dose by 23% to 76% without compromising on image quality (Willemink et al., 2013). This finding was supported by several studies on thorax, coronary angiography and abdomen CT examination (Den Harder et al., 2015; Den Harder et al., 2016; Ellmann et al., 2018). In the final stage of CT principle, the reconstructed images processed by the CPU will be displayed on the display monitor (Seeram, 2015). Image is displayed on the workstation monitor for further image processing and modification to produce a good quality image for diagnosis evaluation by radiologist. The degree of beam attenuation of anatomical structures in CT images is expressed in Hounsfield units (HU) or CT numbers that represent the pixel density values. The CT number for dense materials such as bone is assigned as 1000 HU while the less dense material such as air is assigned as -1000 HU. The range of displayed HU for particular image is selected by window width meanwhile the centre of displayed HU range is determined by window level (Romans, 2011).

One of the advancements in CT technology is dual-energy computed tomography (DECT). DECT can be defined as the CT that utilises two photon energy spectra. It is also known as spectral CT (Johnson, 2012). In 1970s, the first investigations regarding DECT were made; but it was not clinically utilised due to several factors like long scanning time resulting patient movement between the scans, postprocessing complexities, and limited spatial resolution (Johnson et al., 2011). The simultaneous acquisition of DECT data was made possible in 1980s where the tube voltage was changed rapidly between high and low kV settings during the rotation of tube-detector resulting in two sets of raw data (Nikolaou et al., 2019). Different tube potentials generate different energy spectra to provide maximum attenuation difference and least overlap between the spectra, the tube potentials selection of 80 and 140 kV is frequently utilised with additional filtration at the highest kVp. There are several DECT technical methods which have different advantages and disadvantages such as sequential acquisition, rapid voltage switching, dual source computed tomography (DSCT), layer detector, and quantum counting detector. Sequential acquisition is accomplished by a series of subsequent rotations at alternating tube voltages and stepwise table feed. This technical method requires least hardware capabilities, but it causes long delay between both sequence acquisitions.

Meanwhile the rapid voltage switching method utilises high and low tube voltage value alternation with the transmitted data are gathered for each projection for two times. Even though this method requires less technical effort, but it takes long acquisition time due to system's rotation speed is reduced to comprise with the additional projections acquisition and with the up and down of the voltage modulation's time. At low voltages, this method has limited photon output resulting high noise. Another DECT technical method is dual-source CT, which utilises two tubes and runs at different voltages with the corresponding detectors mounted orthogonally in a gantry. The disadvantages of this methods are expensive technical setup and cross-scatter radiation due to the orthogonal setup. The advantages of this method are it allows easier selection of exposure parameters by the user, and filter values for both tubes to accomplish an optimal spectral contrast, adequate transmission, and less overlap.

The dual-layer detector approach utilises one x-ray tube that running at constant voltage with different layer of scintillated material detectors. The sensitivity profiles and the spectral resolution are determined by the materials of the scintillator. The disadvantages of this approach are limited contrast of the spectral information and requires high dose. Meanwhile, the quantum-counting detector approach has very high efficiency of the quantum and utilises to distinguish more than two photon energies. Nevertheless, a quick drift of the measured signal resulted by the rapid saturated detector materials is the drawbacks of this approach. Thus, this approach cannot cope with a clinical CT's photon flux and can only use in small animal imaging (Johnson et al., 2011).

2.2 Radiation Interaction with Patient in Computed Tomography

Theoretically, there are five types of interaction between x-rays and matter which are coherent scattering, Compton scattering, photoelectric effect, pair production, and photodisintegration. However, in diagnostic energy range, only scattering (coherent and Compton) and photoelectric absorption will occur. While pair production and photodisintegration interactions happen at higher energy range such as in nuclear medicine imaging and radiotherapy. An interaction between low-energy x-ray with atoms can be referred as coherent scattering or also known as Thompson scattering (Figure 2.1). During this interaction, the target atom become excited when incident x-ray interacts with it and causing it to release its excess energy as a scattered x-ray with the same wavelength and energy as the incident x-rays but in different direction. Coherent or Thompson scattering interaction contributes less to the medical image since it adds slightly to image noise.

Meanwhile, Compton scattering (Figure 2.2) interaction happens between outer-shell electrons and moderate-energy x-rays. During this interaction, the incident x-ray hit the outer-shell electrons causing it to be ejected from the atom (as Compton electron. The results of this interaction are x-ray energy reduction, x-ray direction alteration (scattered x-ray), and ionisation. Compton scattering is considered crucial in x-ray imaging as it can reduce image quality and produce hazard in medical imaging, as high amount of scattered radiation can cause unnecessary dose to the patient during the scanning. Thus, the radiographers could receive high occupational radiation exposure. Another type of radiation interaction is photoelectric effect (Figure 2.3). During this interaction, a total energy absorption of the incident x-ray photon occurs and lead to ionisation of an inner-shell electron. Then, the incident photon vanishes and the K-shell electron is ejected from the atom and is known as photoelectron. Higher atomic number atoms have higher K-shell electron binding energies, but lower kinetic energy of the photoelectron. Meanwhile, lower atomic number atoms have lower binding energies, but the kinetic energy of the photoelectron is the nearly same as the incident x-ray. Vacancy in the K-shell is instantly filled by the dropping of an outer-shell electron (L-shell) with the emission of x-rays. The emitted x-rays have the similar manner as a scattered radiation and does not contribute to diagnostic value.

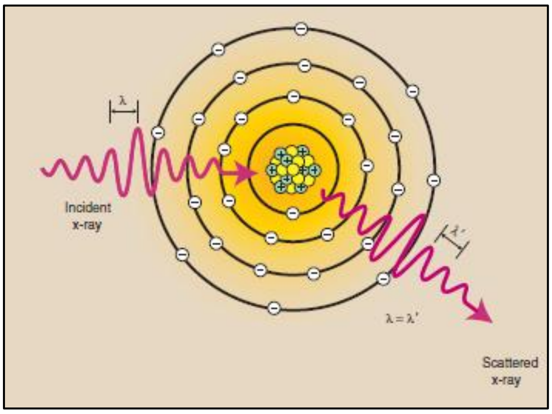

Figure 2.1 Coherent Scattering (Bushong, 2016)

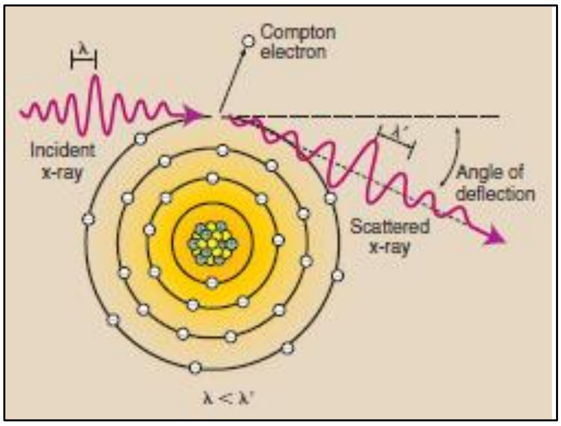

Figure 2.2 Compton Scattering (Bushong, 2016)

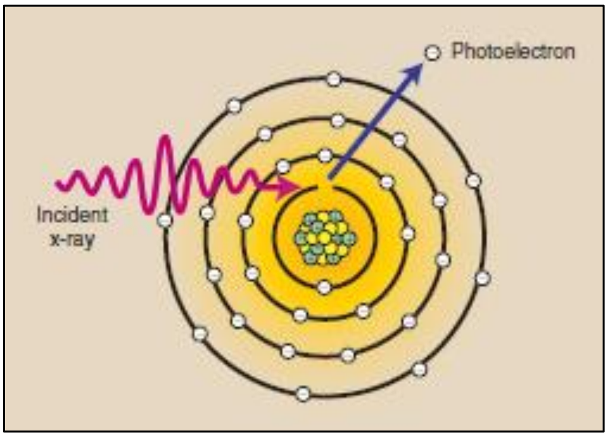

Figure 2.3 Photoelectric Effect (Bushong, 2016)

The highest contribution of man-made ionising radiation exposure to human is from medical diagnostic imaging. Rapid progress of modern technology and establishment of multi-detector CT (MDCT) has surprisingly increased and grew the utilisation of CT in the recent years (Prokop, 2003). Since the innovation of computed tomography in medical practice in 1973, the application of CT imaging for medical diagnosis have been expanded (Mettler et al., 2009). Furthermore, due to recent CT technology advancement, the scanning can be conducted smoothly and rapidly as compared to conventional technology and the operators tend to utilise the CT scanner excessively (Tsapaki and Rehani, 2007). Due to this, CT imaging has been widely used in clinical application since its initiation (Smith-Bindman et al., 2012). From the previous report by United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR), CT procedure has contributed less than 15% of worldwide cumulative dose from clinical x-ray procedures in the period of 1985-1990 and increased to more than 30% in the period of 1991-1996 (United Nations Scientific Committee on the Effects of Atomic Radiation, 2000). CT procedure contributes almost half (49%) of total cumulative dose even though it only represented 16% of total medical imaging procedure as reported in the USA, 2009 (Schauer and Linton, 2009). This situation has led to critical concern on the radiation doses given to the patients during CT examinations and health related hazard induced by the exposure from ionising radiation received by the patients (McCollough et al., 2009).

Unnecessary exposure to the patient may contribute to irrelevant hazards by cause of stochastic effects (Holmberg et al., 2010). DNA double-strand break which cannot be completely repaired and linked with organ sensitivity to ionising radiation produces lesion and it is established as a risk associated with ionising radiation. There were also reported evidences on the molecular lesion associates with the risk of ionising radiation that lead to chromosomal aberrations, mutation, and cell killing (Baert et al., 2007). Non-target organs or tissues like the ovaries, uterus, and testis within the scanning field of view (FOV) during abdominal or pelvic CT scans, breast in thorax CT scans, and lens of the eye in the brain CT scans are the major concern since most of them are highly radiosensitive organs (Abdulkadir, 2015). The progress of radiation-induced cancers and leukaemia were correlated with low-dose radiation exposure according to earlier epidemiologic studies (Royal, 2008). Based on extrapolating data of Japan’s atomic bombs survivors, there were connection between radiation and subsequent development of neoplasia (Power et al., 2016). Besides that, according to a study conducted in Australia from the year 1985 until 2005, the cancer incidence was 24% greater for the patients who were exposed to CT scans, than those who are unexposed (Mathews et al., 2013).

A study conducted in Taiwan found that patients who were diagnosed with non-Hodgkin lymphoma and exposed to multiple CT examinations were linked with an increased risk of leukaemia and thyroid cancer (Shao et al., 2019). A single phase of CT abdomen examination yields an effective dose around 10 mSv, but if the patient undergoes repetitive multiphasic CT examinations especially for cancer screening cases, the cumulative doses will be around 100 mSv. There is a clear evidence of radiation-induced cancer risk with the dose of above 100 mSv (Council, 2006). Even though the linear no-threshold theory of the excess cancer risk due to low dose radiation is still debatable, the epidemiological data clearly imply increased cancer risk in the medical imaging examinations for 10 to 100 mSv range, and this is relevant to many CT examinations. Furthermore, there is 5% excess risk of death from cancer with 100 mSv dose, and this value is used as limit (ICRP, 1991; Meinhold, 1993; Roslee et al., 2020). Table 2.1 shows the effective doses (in mSv) received from exposure to ionising radiation delivered by common CT examinations for adult patients.

Table 2.1 Effective Doses for Common CT Examinations (Mettler et al., 2008)

Common CT Examinations	Average Effective Dose (mSv)	Value of Effective Dose (mSv) Reported in Literature*
Head	2	0.9 – 4.0
Neck	3	-
Chest	7	4.0 – 18.0
Chest for pulmonary embolism	15	13 - 40
Abdomen	8	3.5 - 25
Pelvis	6	3.3 - 10
Three-phase liver study	15	-
Spine	6	1.5 - 10
Coronary angiography	16	5.0 - 32
Calcium scoring	3	1.0 - 12
Virtual colonoscopy	10	4.0 – 13.2

*Value of Effective Dose (mSv) reported in literatures (Yeoman et al., 1992; Mini et al., 1995; Hopper et al., 2001; Bassim et al., 2005; Brenner and Georgsson, 2005; Einstein et al., 2007; Huda and Vance, 2007; Martin, 2007)

Radiation dose received from medical imaging was also comparable with the time needed to attain the similar effective dose from background radiation and it is known as Background Equivalent Radiation Time (BERT) (Dougherty, 2009). BERT is used as an alternative method for diagnostic imaging practitioners to ease patient's concern and fear by providing a simple way of explanation on the radiation's hazard. The BERT method is achieved by comparing the amount of radiation received from medical exposure with the natural background radiation received over a particular period of time like days, weeks, months, or years (Sherer et al., 2014). Most of the scientific concepts, terminologies, and measurements of radiation levels are difficult to be understood by public, thus the BERT method was used to overcome that problem by using simpler terms that could be easily understood by them. The annual dose from background radiation is about 3 mSv/year (Nickoloff et al., 2008). In comparison, the radiation received from exposure to brain, thorax, and abdomen-pelvis CT examinations are equivalent to approximately 1, 3.6, and 4.5 years of background radiation received while spending in natural surroundings, respectively. Nevertheless, the BERT method does not indicate radiation risk since it is merely a method for comparison (Sherer et al., 2014). Thus, justification and optimization should be made for all radiation exposures from medical purpose (ICRP, 1991). Diagnostic reference level (DRL) had been introduced in 1996 concerning to improve patient's radiation protection (ICRP, 1996). Despite the fact that CT gives huge radiation dose to patients, its benefits can significantly exceed the hazard if all related practitioners have the knowledge on utilisation of the modality accordingly (Abdulkadir, 2015).

2.3 Computed Tomography Dosimetry

In order to assess medical radiation exposure in Malaysia, medical radiation survey was performed to gain qualitative and quantitative data of the doses for diagnostic procedures (MOH, 2013). Scanner radiation output, organ dose, and effective dose are several ways to quantify the radiation dose in CT (Abdulkadir, 2015). There are two standard phantoms sized 16 cm for head and 32 cm for body, known as CTDI phantoms (Figure 2.6) used in the determination of the dose length product (DLP) and volume CT dose index ($CTDI_{vol}$). $CTDI_{vol}$ and DLP were the CT dose quantities associated with the scanner output suggested by European Commission (EC, 1999).

2.3.1 Computed Tomography Dose Index (CTDI)

CTDI (as defined in Equation 1) serves as crucial CT-specific dose amount which appoint the peak of dose profile ahead z-axis with the total summation of the CT dose (Dixon, 2019). CTDI is determined by the equation given below (ICRU, 2012):

$$CTDI_{\infty} = \frac{1}{nT}\int_{-\infty}^{\infty} K(z)dz \text{ (Dixon, 2019)} \qquad 1$$

where (n) and (T) symbols depict as the number of tomographic areas acquired in one axial scan and the nominal wideness of the tomographic areas ahead the middle point of the z-axis. Meanwhile, the air-kerma in the phantom depicted as (K(z)); (z) depicts the position at given location; and (d) represents nominal slice thickness. Thus, CTDI is represented by the region of the dose profile divided by the nominal beam wideness.

2.3.2 Computed Tomography Dose Index 100 ($CTDI_{100}$)

Cylindrical pencil shaped air ionization chamber is used to measure $CTDI_{100}$ (Equation 2) (Dixon, 2019) by inserting it into the provided holes (first at the centre and then into both sides of phantom's periphery) in the polymethylmethacrylate (PMMA) CTDI phantom (Figure 2.6). $CTDI_{100}$ is revised from the dose profile being accumulated over the range -50 mm to +50 mm reciprocal to the middle point of the beam as Equation 2 given below (Dixon, 2019):

$$CTDI_{100} = \frac{1}{nT}\int_{-50\,mm}^{+50\,mm} K(z)dz \qquad 2$$

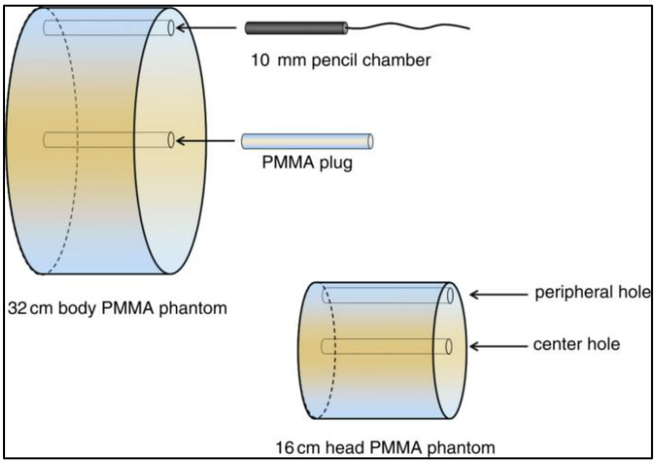

Figure 2.6 The CTDI phantoms of standard sizes which made up of tissue equivalent material, PMMA (UC Davis Healthcare, 2012)

2.3.3 Computed Tomography Dose Index Weighted ($CTDI_w$)

The average weighted dose of the $CTDI_{100}$ is referred as $CTDI_w$ (as shown in Equation 3) where it is measured at the middle point and the phantom's outer edge. It also serves as average of the absorb dose in a single cross section. Therefore, $CTDI_w$ is usually depicted as Equation 3 (Dixon, 2019):

$$CTDI_w = \frac{1}{3}.CTDI_c + CTDI_p \quad (3)$$

Measurement at the centre of the phantom is referred as (c) meanwhile the two periphery measurements is referred as (p).

2.3.4 Computed Tomography Dose Index Volume ($\mathbf{CTDI_{Vol}}$)

Introduction of volume CTDI ($CTDI_{vol}$) (Equation 4) (Dixon, 2019) helps to determine radiation doses from different scan and series protocols. It is important to alert any change in radiation doses related to a pitch factor. Thus, for an accustomed scan volume of the average dose is outline as Equation 4 (Dixon, 2019):

$$CTDI_{vol} = \frac{CTDI_w}{p} \quad (4)$$

where (p) represent a pitch factor value.

2.3.5 Dose Length Product (DLP)

In conjunction with the number of acquisition slices and a scanning range, DLP (Equation 5) (Dixon, 2019) is proposed to obtain the dose from a conclude sequence. DLP also can be a measurement substitution of the whole energy absorbed by the phantom with the units of (mGy.cm) (ICRU, 2012). The amount of DLP can be determined by multiplying $CTDI_{vol}$ with the scan length (L) (Dixon, 2019):

$$DLP = CTDI_{vol}\ x\ L \quad (5)$$

2.4 Diagnostic Reference Level

DRL can be explained as a patient dose level or administered activity which related to the amount of radioactive material for a certain procedure used in medical imaging (ICRP, 1996). DRL can also be defined as dose degrees in clinical diagnostic practice for common examinations for number of individuals collectively of normal-sized patients or normal phantoms for broadly described categories of imaging modality (IR(ME)R, 2000). The first country to establish national DRLs was Norway in 1987 and was presented for 6 conventional radiological examinations in 1996 (Godske Friberg et al., 2010). The DRL is highly recommended to prevent unnecessarily higher dose to patient and serve as a guideline for medical exposure from different examinations. Besides that, it also significant in case of patient dose or applied action is acutely high or low for certain imaging examinations (MOH, 2013).

Diagnostic reference level is also planned to improve radiation protection to the patients by allowing comparison of current and similar CT examination with the same purpose and technique (IPEM, 2000). For that reason, optimisation of protocols and standardisation of scan parameters is suggested since significant variation in doses and scanning protocols were observed in several different surveys (Livingstone and Dinakaran, 2011). In order to put the amount which contribute a critical evidence of patient exposure, the DRLs should be well defined and used the most common indicators for CT examinations such as CT Dose Index (CTDI) and Dose-Length Product (DLP) (MOH, 2013). However, International Commission on Radiological Protection (ICRP) suggested the use of $CTDI_{vol}$ and DLP dose metrics for DRL instead of $CTDI_w$ and DLP as what have been practiced by Malaysian Minister of Health

(MOH, 2013; Vano et al., 2017). Normally, after the CT examination is completed, the CTDI and DLP values are readily displayed on the work-station screen monitor.

MOH advocated for studies to survey medical imaging doses across the country to enhance national, local, or institutional dispersion of recognized results by lowering the prevalence of the needless high or low amounts. Furthermore, this also can help accomplishment of a restricted extent of values that serve as great method for extra definitive clinical diagnostic imaging procedure. The national guideline of CT DRLs as recommended by Ministry of Health (MOH) Malaysia is shown on Table 2.2 (MOH, 2013).

Table 2.2 Recommended Malaysian DRLs for CT (MOH, 2013)

Examination Type	$CTDI_w$ (mGy)	DLP (mGy.cm)
Abdomen	12.8	450
Brain	46.8	1050
Cardiac	11.8	870
Chest	19.9	600
Pelvis	39.1	730
Spine/Musculo-skeletal	16.3	390
Thorax	21.3	420
Others	12.3	380

Meanwhile, there was study in the southern region of Malaysia (Johor Bahru) which published the mean values of $CTDI_w$ and DLP for abdomen, brain, pelvis, thorax-abdomen-pelvis (TAP), and thorax CT examinations. The study was conducted at Hospital Sultan Ismail Hospital Sultanah Aminah, and Hospital Permai. The results for establishment of national DRLs and comparable with international DRLs. The mean values of this study is shown in Table 2.3 (Karim et al., 2016).

Table 2.3 Radiation Doses from Computed Tomography Practice at few Hospitals in Johor Bahru,

Malaysia (Karim et al., 2016)

	HSAJB		HSIJB		HPer		HSIJB-R	
Examination Type	$CTDI_w$ (mGy)	DLP (mGy.cm)	$CTDI_w$ (mGy)	DLP (mGy.cm)	$CTDI_w$ (mGy)	DLP (mGy.cm)	$CTDI_w$ (mGy)	DLP (mGy.cm)
Abdomen	11.2	720.5	13.2	263.6	9.9	445.5	-	-
Brain	62.5	884.9	55.4	743.4	64.8	943.3	7.6	203.1
Pelvis	14.7	519.3	-	-	-	-	18.0	539.2
TAP	-	-	11.6	345.3	-	-	-	-
Thorax	10.7	400.4	10.1	170.2	11.3	680.6	10.1	377.9

HSAJB – Hospital Sultanah Aminah, HSIJB – Hospital Sultan Ismail, HPer – Hospital Permai, HSIJB-R –Hospital Sultan Ismail – Oncology Centre, TAP –Thorax-abdomen-pelvis.

Besides that, there were also few established international DRLs (Table 2.4) from different countries used in this study for comparison to assure the reference dose levels are strictly adhered to and dose above threshold can be reviewed. These low- and middle-income countries include Kerala, India (Kumar, 2020), Egypt (Salama et al., 2017), Kenya (Korir et al., 2015), and North-Central Nigeria (Abdulkadir, 2015). These studies have successfully established the national diagnostic reference levels for their countries through national survey and the NDRLs has been reviewed occasionally (Table 2.4). To standardise the optimisation practice associate with ionising radiation, those studies also provide guidelines and recommendations to be complied by individuals, centres, or states which intended to set their national DRLs.

Table 2.4 Established International DRLs by few Countries.

	India (2020)		Egypt (2017)		Kenya (2015)		Nigeria (2015)	
Examination Type	$CTDI_{vol}$ (mGy)	DLP (mGy.cm)	$CTDI_{vol}$ (mGy)	DLP (mGy.cm)	$CTDI_{vol}$ (mGy)	DLP (mGy.cm)	$CTDI_{vol}$ (mGy)	DLP (mGy.cm)
Abdomen	9	482	31	1423	20	1842	15	757
Brain	49	925	30	1359	61	1612	60	1024
Pelvis	-	-	-	-	21	1928	-	-
*TAP	-	-	33	1322	-	-	-	-
Thorax	7	456	22	421	19	895	10	407

*TAP –Thorax-abdomen-pelvis.

The earliest CT DRLs were established by Irish using generated CT dose values displayed on the console for nine most frequently performed CT examinations through radiation doses survey (Foley et al., 2012). With a least of 10 normal sized patients in each region of CT examinations, all $CTDI_{vol}$ (mGy) and DLP (mGy.cm) were documented to calculate its mean values from all 34 multi-slice CT scanners. The rounded 75th percentile was utilised to calculated LDRL and NDRL after gathering all results. From this study, it shows 42% reduction of CT doses compared to previously recommended values (EC, 2000) due to advancement in CT technology in which the single-slice CT scanners are no more in utilise and not included in this study compared to previously reference survey. In addition, $CTDI_w$ dose descriptor used in previous studies has been replaced by $CTDI_{vol}$.

The description of a normal sized patient (70 ± 5 kg) by the (EC, 1999) was expressed as hard to deal with because of 20% of Norwegian population were overweight as recorded in a study in Norway (Godske Friberg et al., 2010). Therefore, an extensive weight group from 55 until 90 kg was utilised as a standard size patients of Norwegian population and this study reveals that the interpretation of standard-sized-patient established by (EC, 1999) was not relevant since the weight of each type of age group and sex differ significantly from every countries.

2.5 Radiation Optimisation

To optimize the radiation protection for medical procedure related to x-ray radiation, a lot of countries such as Australia, Canada, and the Europe had established their local DRL as an effective tool (Heliou et al., 2012). At the same time, to refrain unnecessarily high doses to the patient, Ministry of Health (MOH) in Malaysia has recommended that all government and private health centers that provide medical radiation services to implement DRL as a guideline (MOH, 2013). By setting up the threshold radiation dose value that should not be exceeds during CT examinations, the extreme dose level can be identified (Foley et al., 2012). According ALARA principle, it is essential to apply the DRL in CT examinations since its excessive doses are not directly identified as image artifacts on a film screen. Besides that, it was also recommended by ICRP to establish DRL based on appropriate local, regional, or national data (ICRP, 1996).

All medical institutions should collect their own individual local data in their own setup to compare it with the national DRL as recommended by MOH (MOH,