NEAR EAST UNIVERSITY
INSTITUTES OF GRADUATE STUDIES
DEPARTMENT OF BIOMEDICAL ENGINEERING

CT DOSE REDUCTION: PROPOSED TO ESTABLISH DIAGNOSTIC REFERENCE LEVELS IN ADDIS ABABA, ETHIOPIA

PHD THESIS

JEMAL EDRIS DAWD

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APPROVAL

We certify that we have read the thesis submitted by **Jemal Edris Dawd** titled "**CT dose reduction: proposed to establish diagnostic reference levels in Addis Ababa, Ethiopia**" and that in our combined opinion it is fully adequate, in scope and in quality, as a thesis for the degree of Ph.D. of Biomedical Engineering.

Examining Committee	**Name-Surname**	**Signature**
Head of the Committee:	Prof. Dr. Ayşe Gunay Kibarar	
Committee Member:	Prof. Dr. Nail Bulakbasi	
	Assoc. Dr. Doga Karaz	
Supervisors:	Assoc. Prof. Dr. Ilker Ozsahin	
	Assoc. Prof. Dr. Dilber Uzun Ozsahin	

Approved by the Head of the Department:

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Prof. Dr. Ayşe Gunay Kibrar
Head of Department

Approved by the Institute of Graduate Studies:

…../…../2022

……..
Prof. Dr. K. Hüsnü Can BAŞER
Head of the Institute

DECLARATION

I hereby declare that all information, documents, results and analysis in this thesis have been collected and presented according to the academic rules and ethical guidelines of Institute of Graduate Studies, Near East University. I also declare that as required by these rules and conduct, I have fully cited and referenced information and data that are not original to this study.

Jemal Edris DAWD

April, 2022

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Jemal Edris DAWD

April, 2022

DEDICATION

In the memory of my mother Teguade Yessuf Mohammed (1955 - 2002) ...

In the memory of my father Endris Dawd Yimer (1944 – 2017) ...

To my family, mainly to my lovely grandson Abdulaziz Jemal ...

To my lovely Brothers and Sister ...

ABSTRACT

CT Dose Reduction: Proposed to Establish Diagnostic Reference Levels in Addis Ababa, Ethiopia

DAWD, Jemal Edris
PhD, Department of Biomedical Engineering,
Assoc. Prof. Dr. Ilker Ozsahin
Assoc. Prof. Dr. Dilber Uzun Ozsahin

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Computed tomography (CT) is delivering comparatively very high dose to the patient. Hence, it requires strict adherence to dose optimization. Basically, dose optimization can be achieved via the setting and medical effecting of DRL. DRL signify the investigation level and fitness of the dose for patients based on its acceptability to ALARA principles without convicting image quality. Currently, CT DRL is not existing in the research area, except locally proposed DRLs. The goal of this research is to introduce applicable CT DRL towards projecting patient dose optimization. This research was done in three phases. The 1st phase included retrospective dose data collection from the health institutions archives and defined local DRL as initial. The 2nd phase required data collection using experimental setup using PMMA phantom and ionization chamber to set new post optimized DRL. The 3rd phase required experimental CT image data gathering using ACR CT accreditation phantom to assess the image quality. The DRLs have been computed based on clinically measureable dose metric. The typical measureable metrics for phase-1 were CTDIv and DLP and for phase-2 studies was $CTDI_v$. The 75th percentile mean dose distribution of $CTDI_v$ and DLP were applied to set DRLs. The essential image quality criteria were testing the compliance of alignment, slice thickness, CT number accuracy, positioning accuracy, in-plane distance accuracy, spatial resolution, low contrast resolution, uniformity as well as noise. Acquired image quality factors were compared with the tolerance quantities. The research applied the 95% confidence level. The phase-1 research revealed that the 75th percentile of $CTDI_v$ for head (wo), head (w), chest (wo),

chest (w), pelvic (wo) and c-spine (wo) and abdomen (mph) were 52.70, 50.78, 16.56, 14.75, 14.20, 37.52 and 13.66 mGy, respectively. Whereas, the 75th percentile of DLP values were 1237, 1459, 625, 565, 728, 605 and 1106 mGy.cm for head (wo and w), chest (wo and w), abdomen (mph), pelvic (wo) and c-spine (wo), respectively. The outcomes of phase-2 study shown that the estimated 75th percentile mean dose distribution of $CTDI_v$ values computed for Philips, Siemens and GE scanners using head phantom were 50 mGy, 31 mGy and 42 mGy, respectively. The corresponding 75th percentile mean dose distribution of $CTDI_v$ values measured for Philips, Siemens and GE scanners using body phantom were 17 mGy, 16 mGy, and 16 mGy, respectively. This study shown wide variations of mean dose distribution among health centers. This could possibly due to variations of CT brand, exam protocol, scan length and application of dose reduction software as well as other technical parameters in use. The outcomes of phase-1 research have advised for medical user because it will empower the medical practitioners by showing optimized dose ranges. However, phase-2 result can be considered as more optimized DRLs values and appropriate for clinical purposes. Hence, the proposed 75th percentile of $CTDI_v$ were 41 mGy 16 mGy measured for head and body phantoms, respectively. The image quality assessments done in phase-3 were in the acceptable range for all tested IQ criteria. However, IQ test variations were seen, this may be due to variation in the CT scanners.

Key words: CT; $CTDI_v$; DLP; optimization of protection; diagnostic radiology; DRL; image quality

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LISTS OF ABBREVIATIONS AND SYMBOLS

AAPM	American Association of physicists in Medicine
ACR	American college of radiology
AD	Achievable Dose
ADC	Analogue to digital converters
AEC	Automatic Exposure Control
ALARA	As Low As Reasonably Achievable
AP	Abdominal Pelvic or anterior posterior
ARPANSA	Australian Radiation Protection and Nuclear Safety Agency
ATCM	Automatic Tube Current Modulation
Axial CT	Sequential CT
AXTM	Anatomical X-ray tube modulation
AXTV	Adaptation of X-ray tube voltage
BEIR	Biological Effects of Ionizing Radiation
BSS	Basic Safety Standard
C	calibration factor for the ion chamber
CBCT	Cone Beam Computed Tomography
CI	Confidence interval
cm	centimeter
CNR	Contrast noise ratio
Covid-19	Coronavirus disease of 2019
CT	Computed Tomography
CTDI	Computed Tomography Dose Index
$CTDI_{100C,PMMA}$	CTDI values measurement of PMMA phantom at its center
$CTDI_{100P,PMMA}$	Periphery PMMA phantom average measurement of CTDI
$CTDI_v$	Volume Computed Tomography Dose Index
$CTDI_w$	Weighted Computed Tomography Dose Index
CV	Coefficient of Variation
D(z)	The out potential of CT tube in the z-axis of rotation
DDM2	Dose Data Med2
DICOM	Digital Imaging and Communications in Medicine
DLP	Dose Length Product
DNA	Deoxyribonucleic Acid

DRL	Diagnostic reference level
DRLs	Diagnostic reference levels
E	Measured value
EC	European Commission
ECG	Electrocardiogram
ECG-CTCM	Electrocardiogram – controlled tube current modulation
EMoH	Ministry of Health of Ethiopia
ESE	Entrance Surface Exposure
EU	European Union
Fig.	Figure
FOV	Field of View
GE	General Electronics
GSR	General Safety Requirement
Gy/A.s	Gray per ampere and second
Gy/C	Gray per coulomb
HCR	High contrast chest
HIPAA	Health Insurance Portability and Accountability Act
HSDP	Health Sector Development Program
HU	Hounsfield Unit
HVL	Half Value Layer
IAEA	International Atomic Energy Agency
IAEA-BSS	International Atomic Energy Agency-Basic Safety Standard
ICRP	International Commission on Radiological Protection
ICRU	International Commission on Radiation Units and measurement
IEC	International Electro-Technical Commission
IPEM	Institute of Physics and Engineering in Medicine
IR	Interventional Radiology
J/kg.C	Joule per kilogram and coulomb
kerma (KERMA)	Kinetic Energy Released per unit Mass of Air
kg	Kilogram
kV	Kilo voltage
kVp	Peak kilo voltage

L	Active length of pencil ion chamber
ℓ	Scan length
LAR	Lifetime Attributable Risk
LCD	Low contrast detectability
LNT	Linear Non-Threshold
lp/cm	Line pairs per centimeter
LS	Lumbar Spine
mA	Milli Ampere
mAs	Millli Ampere second
MDCT	Multiple Detector Row Computerized Tomography
mGy	Milli Gray
mGy.cm	Milli Gray centimeter
mm	Millimeter
mph	Multiphase
MRI	Magnetic Resonance Imagine
ms	Millisecond
mSv	Millli Sievert
MTF	Modulation transfer function
N	Number of image slices per a scan
NDRL	National diagnostic reference level
NRPB	National Radiological Protection Board
NT	Nominal beam width
OBTCM	Organ-based tube current modulation
p	Pitch or periphery
PACS	Protocols and picture archiving and communication system
PHC	Primary Healthcare Unit
$P_{KL,CT}$	Air kerma length product a quantity assessed inside a phantom
PMMA	Poly-methyl methacrylate
QAP	Quality Assurance program
QCP	Quality control program
RadiAnt	Research and Development in Advanced Network Technology
ROI	Region of interest
SI	International System of Units

Spiral CT	Helical CT
SPSS	Statistical Package for the Social Sciences
SR	Special resolution
SSCT	Single slice CT
STD	Standard Deviation
Subject	Patient
T	Slice width per image
TCM	Tube current modulation
UK	United Kingdom
UN	United Nations
UNSCEAR	United Nations Scientific Committee on the Effects of Atomic Radiation
US	United States
USA	United States of America
w_c	with contrast
WHO	World Health Organization
WMA	World Medical Association
w_{oc}	Without contrast
σ_x	Standard deviation
$\bar{x}$	Mean
3D	Three-dimensional

CHAPTER I

BACKGROUND

1.1 Introduction

In this complex world, scientific growth and technological development have motivated the health care area of the world population. This is true in the development of medical imaging technology with an increase contribution in the medical diagnosis of modern health care attention. The modern medical diagnosis using radiology started since the discovery of the x-ray by Roentgen 1895. Since then the knowledge of radiology integrated medical imaging procedures are alarmingly growing and currently it is taking as one of the modern life care system in the world (Paulo, 2015; Paulo et al, 2016).

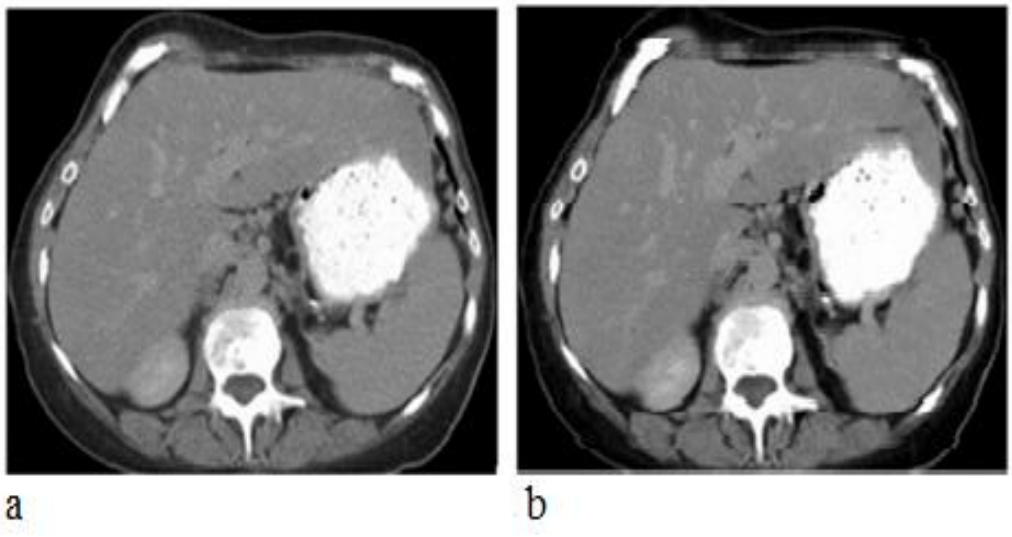
Computed Tomography (CT) is also portions of the current very strong and flexible examination in modern medical imaging technology (such as multi-slice CT and digital imaging) which is the most important irradiating equipment introduced in 1970s. Clinically, CT is influential medical diagnostic tools for patient management, for getting high quality 3D-image data, and for medical management by avoiding surgical interventions. Conversely, CT is great linked with cause risks of carcinogenesis coming from its high burden of radiation dose (Saravanakumar A, et al, 2014; Foley S, et al, 2012). When its radiation dose burden is compared with traditional X-ray, it incurs larger amount of dose to the subject. For instance, CT chest imaging required 100 time more radiation dose than traditional chest X-ray (Li, 2015). The dose complication of CT arises from its dynamic range of exposure. Figure 1-1 illustrates different radiation doses for acceptable diagnostic quality. Hence, application CT for medical exposure is requesting practical application of radiation protection principle called justification, optimization and dose limitation. However, the major difficulty towards dose optimization is mechanisms of identifying acceptable threshold of image quality with sufficiently optimized radiation dosages without jeopardizing the required clinical images (Li, 2015).

Hence, current significant improvements of CT technology including its variety, quality, and speed brought advancement in its clinical applications. Integration of the complex CT scanners technology with professionals into their common CT practices

required for reevaluating the existing CT practice methodology and procedure protocols. The technological growth should be accompanied by considering the awareness and understanding of radiation dose concerns (American Association of Physicists in Medicine, 2008). In future, the application CT for medical exposure is requesting practical application of radiation protection principle called justification, optimization and dose limitation.

Figure I-1

CT images managed at (a) 88 mAs and (b) 256 mAs maintaining other parameters being constant for a 72 kg and 62 years old man (Li, 2015)



During medical exposure of x-ray imaging, the incident x-ray from the CT-tube source penetrates the patient. Using picture forming film or ionization chamber that is sited in the opposite side of the diagnosed patient, it is possible to detect the amount of radiation passing through the patient body (Andrew, 2003). However, the amount of radiation dose deposited in the body of the patient is dependent on the intensity sources, strength of the radiation sources, time and rate of exposure, area of exposure, age, and sex of the patient and so on. In distinction, the attenuation of x-ray radiation in the body is different due to differences in tissues of the body. Hence, the referring medical doctor should access the impacts of each examination result on patient management. The medical imaging technologists should adequately have interested on the justification of the procedures. The operators must be conscious on the selections to minimize patient doses through adjusting procedural parameters to each patient. Pediatric and young patients should be given special attention during examination

procedures. Recent knowledge conclude that appropriate selection of procedural parameters and existence of proper quality control program as well as appropriate usage of recent diagnostic reference levels (DRLs) reduces the patient radiation dosage more than fifty percent (ICRP, 2000a).

It is clear that CT is helping to save countless lives and improve the outcomes of millions of patients. In the hospital, doctors prescribe CT scan diagnosis to the patient, however, the amount of radiation dose delivered different from patient to patient and hospital to hospital. That is due to greater option exposure parameters in CT scanners can highly influences the diagnostic doses of the patient (Saravanakumar A, et al, 2014). Moreover, the lack of some minimum reference levels for the doses delivered to the patient is an indispensable factor for the high doses patients gained from CT diagnosis. In general, when the representative reality of the world practice in radiography medical imaging, it is relied on: (i) the absence of professional practice coordination at all stages; (ii) the communication barriers between current science and professional exercise; (iii) the interval between curricular program of health profession and the new biomedical imaging technological ideas and (iv) the gap between manufacturers and clinical professionals practice (Paulo, 2015).

Diagnostic reference levels for the patient was mentions for the first time through the international commission on radiological protection (ICRP) in 1990. Consequently, ICRP suggested the more details of DRLs, in 1996 the most known report was ICRP 73 (IAEA, 2007). Based on the ICRP, DRLs stands for a system of investigation stages (Joseph et al, 2017) to assist patient dose management in diagnostic and interventional radiology (IR) procedures by optimizing radiological protection by putting method that show abnormally high or low radiation doses to the patient for a specific CT studies (Ruiz et al, 2016). The Irish law definition in it Medical council, "*DRL is dose levels in medical radio-diagnostic procedures or levels of radioactivity in the case of radiopharmaceuticals, for typical examinations for groups of standard-sized patients or standard phantoms for broadly defined types of equipment*" Foley S, et al, 2012).

According to the new ICRP recommendation, DRL is used to optimize the patient doses undergoing medical exposure; which can be used as a benchmark alongside doses from diagnostic procedures that can be compared (Wrixon, 2008). However, DRLs cannot be the dividing line between good and poor radiological practices but it can be used as professional judgments (Ruiz et al, 2016; Joseph et al, 2017). It is also

forbidden to use DRLs for regulatory and commercial purposes. DRLs are functional for medical exposures but not to public and occupational exposures. Therefore, there is no link among DRLs and dose limit as well as dose constraints. In an ideal world, DRLs thought the results of general patient dose optimization of radiation protection in clinical examinations. Actually, it is idealistically problematic. Ruther, it remains better towards picking detected percentile patient dose distribution as the foundational values for setting DRLs. These values should be proposed by professionals. The proposed DRLs should be reviewed at intervals. The review should inculcate the required changes and stability in the observed dose distributions. The DRLs values selected by the professionals could be particular to a country (Ruiz et al, 2016; Joseph et al, 2017; McCullough, 2010). DRLs signify the investigation of dose appropriateness levels in optimizing patient dose and amplifying the aim towards patient protection. However, it is not the ideal dose or absolute upper bound for radiological procedures. DRLs appreciate to attaining image quality at lower doses. Therefore, DRLs can trigger image quality improvements at the appropriate patient dose. The main objectives of DRLs is towards identifying higher dosage levels to find means to reduce those dose without compromising image quality. To keep the doses to medical examinations in hospital is minimum, formerly it is important to estimate the dose to patients as a function of radiographic exposure parameters. When the patient dose during a medical examination in diagnostic and therapeutic radiology is monitored, there is a need for assessing the delivered doses to the patient which need the assessment of image quality. Patient's radiological dose assessment requires professional experts' judgment and it is time taking. Practically, it is difficult to acquire sufficient information about the dose to the patient in hospitals, (Joseph et al, 2017).

1.2 Computed Tomography

CT utilizes X-rays and consistent 3-dimentional images of the inside structures of the body without gaps as well as supplementary artifacts using computerized reconstruction techniques, see Figure 1-1. As Figure 1-1 shown, the patient is placed on a patient bed within the center of the CT scanner. Radiation pulses passes via the patient body when CT tube rounding 360^{o} the subject body. Using fitted circular radiation detectors, the radiation pulses passed through patient body are detected. The computer registers the X-ray transmission data. Then, the cross-sectional image are

produced using the matrix of CT number that is calculated by computer using assigned algorithm (Reiser M, et al, 2009).

Approximately 3.6 million diagnostic radiology examinations were conducted worldwide annually reported until 2008 (UNSCEAR, 2008). Medical radiology contributed 60% - 70% of doses. The radiation dose contributions from CT scanning accounts 25% of medical diagnostic examinations dose (IAEA, 2009). The collective dose of radiation from CT scanning reached approximately 50% in several countries, exclusively 68% in UK due to proportionally high dose nature of CT examinations (ICRP, 2017b). Based on AAPM report no. 204, the annual CT examinations conducted in United States (US) was 80 million exams. The medical exposure of US in 2006 was seven times than the exposure in 1980s (American association of pysicists in medicine, 2011).

The percentage contribution of CT procedures was 12% of medical imaging performed, resulting nearly 50% of the total dose of radiation to American population (Kanal et al, 2017). Overall, the population collective effective dose from the medical examinations of CT in the world fall in the range of 30 - 50% (European Commission, 2008); see Table 1-1 for the summary of medical exposure of CT-scanning collective dose contribution to the world community. The recorded patient dose as a result of frequent CT studies have been risen quickly. However, the patient dose from conventional radiography studies has been reduced by nearly 30 percent in the previous decade (ICRP, 2000b). Therefore, the annual CT exam frequency and the significant radiation dose per CT exam is increasing recently. Subsequently, the population total clinical dose received from CT examination is rising annually and dosage per exam.

Figure I-2

It is computed tomography scanner (MF. Reiser et al, 2009)

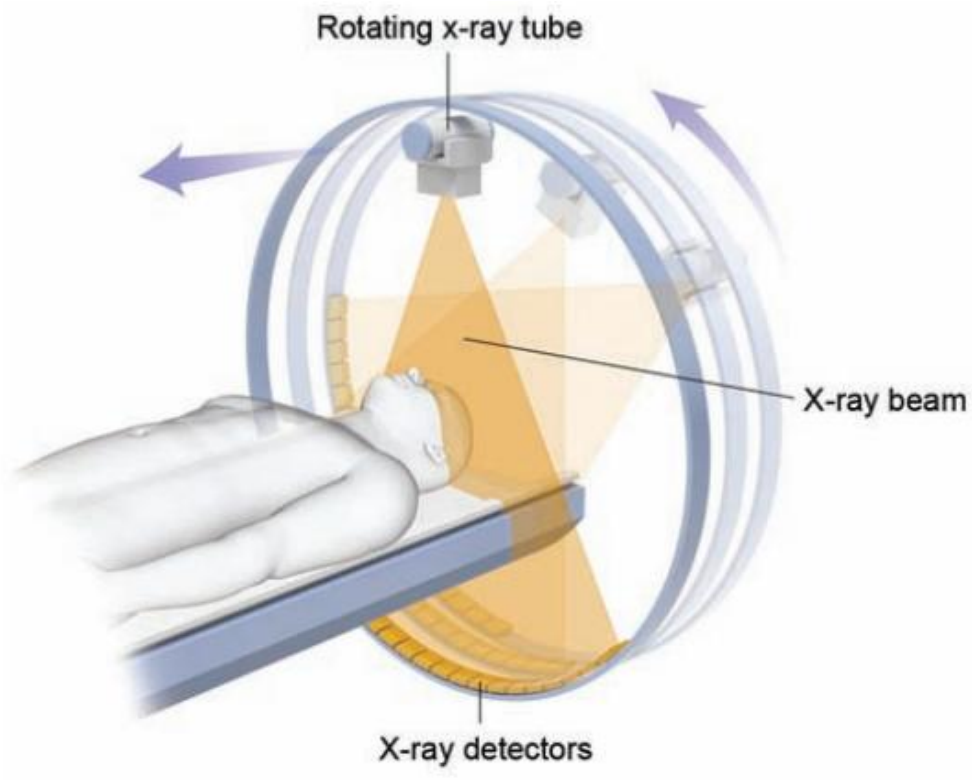


Table I-1

Population collective radiation dosage involvement from the clinical examination of CT

Reference	Approximate Dose Impacts to the population (%)	Dose contribution
(UNSCEAR, 2008)	34 %	Worldwide population
(ICRP, 2017b)	50%	UK population
(Kanal et al, 2017)	50%	American population
(European Commission, 2018)	60%	European population

In radiography, excellent image quality for adequate diagnostic purpose related with higher X-ray exposure (Graingr and Allison, 2015). Similarly, CT examination uses higher exposure to obtain higher quality of image with essential diagnostic information. However, exposure to high dose of ionizing radiation may create reactions of tissue (infertility, skin erythema, hairlessness, cataracts, etc.) and stochastic effects (cancer and genetic effects) (ICRP, 2015). Higher CT dose exposure has also high risks of cancer to the patient. This effects get noticeable concern in medical diagnosis of the world community (ICRP, 2017b). Hence, the clinical application of CT scan required careful quality control program by considering

excellent image quality with highly optimized dose of the patient. Hence, CT examination required the practical application of the principle of ALARA at lower dosages by giving greater consideration towards the required quality of image.

Thus, application of patient dose reduction required the eminent coordination of imaging process professionals with the aim of facilitating and promoting its accomplishment (ICRP, 2017b). The most substantial CT parameters which affect amounts of dose received by a subject are automatic exposure control, Bow-tie/beam shaping filter, abdomen CT at 120 kVp, usage of anti-scatter grid, tube current modulation, selective in plane shielding, thyroid and breast shields (Alsafi, 2016; ICRP, 2017b; McCollough et al, 2009). These are equipment and operator dependent factors. Different CT scanners revealed different clinical examination controller setup method towards a particular medical protocols. Hence, the most applicable approaches to minimize high dose effectively are usage of AEC, cautious referral standards, appropriate selection of examination parameters as well as strict evaluation of protocols (Alsafi, 2016). Table 1-2 shows the strategies of dose optimization methods. The practical application of the fundamental radiation protection principles is highly significant to manage all these CT parameters. These principles are justification and optimization; the next sections briefly pronounce about their relevance in medical diagnosis of CT.

Table I-2

Estimated CT dose reduction with their optimization methods strategies

Dosage reduction methods	Rough dose reduction (%)
Automatic exposure control (AEC) (Alsafi, 2016; McCollough et al, 2009)	20 – 40
Bow-tie/beam shaping filter (ICRP, 2017b)	up to 50
pitch automatic adaptation (ICRP, 2017b)	30 – 50
120 kVp for Abdomen CT (ICRP, 2017b)	20 – 40
Reduction of kVp from 140-100 (ICRP, 2017b)	50
Use of anti-scatter grid (Alsafi, 2016)	up to 50
Tube current modulation (Alsafi, 2016)	30 – 40
Selective in plane shielding (McCollough et al, 2009)	40 – 67 for adult, 30 - 40 for children
Thyroid and breast shields (ICRP, 2017b)	20 - 30

Most recent CT machines provided pre-set image acquisitions parameter settings that helps the operator to instruct exposure to each body parts. These CT image acquisition parameter setting and operator integrity affect the patient dose profile. Hence, the

practitioners should obtain proper training concerning principles of radiological-protection and methods of reducing patient dose without jeopardizing the intended clinical image quality. As Table 1-3 shows the parameters integrity with volume computed tomography dose index ($CTDI_v$) on radiation output that directly correlated with patient dose. Inappropriate use of those parameters will lead the choice of unwantedly high or unwantedly low exposure setup (ICRP, 2000b).

Table I-3

The settings of image acquisition parameter against influences on $CTDI_v$ summary (Siemens healthcare, 2015)

Image-acquisition-Parameters	Correlation-to-$CTDI_v$
Mode of scan	Changes in mode of scan can influence $CTDI_v$
Table increment	Table feed affects $CTDI_v$ through pitch
Pitch factor	$CTDI_v \propto 1/\text{pitch}$
Exposure time per rotation	$CTDI_v \propto$ Exposure time per rotation
Tube current	$CTDI_v \propto$ tube current
Tube voltage	$CTDI_v \propto (kVp_1/kVp_2)^n$ where n~2 - 3
Tube current time product	$CTDI_v \propto$ Tube current time product
Effective tube current time product	$CTDI_v \propto$ Effective tube current time product
Field of measurement	Change in field of size may affect $CTDI_v$
Beam shaping filter	Changes in this factor may affect $CTDI_v$
Detector configuration	Reducing the beam collimation typically, but not always, increase $CTDI_v$

The mathematical image reconstruction principles of CT imaging use X-ray penetration via skinny slices of the subject body to produce cross-sectional images, Figure 1-1. CT shows each pictured slice separately. Properly adjusted narrow X-ray beam is radiated from CT tube to the patient and detectors placed opposite to the patient. The patient body attenuate the photons via absorption and scatter of the beam. The detectors measure the amount of X-ray passes via the patient body as slice images. This quantification is thoroughly repetitive for several times commencing different directions through 360^o rotation of X-ray tube about the patient, Figure 1-1. The computer algorithm assigns CT-number to each pixel in the image. It represents the measurements of transmitted X-rays and uses as data to form an image. CT-pixel numbers are proportional to the difference in a mean X-ray attenuation of the tissue within the voxel and water (Brant W and Helms C, 2012). A Hounsfield (H) scale is used. Hounsfield units are proportional quantities, which can differ from one CT system to another. The Hounsfield unit (HU) ranges for different materials and tissues are illustrated in Table 1 – 4.

The computer algorithms determine the dimensions of voxel. This computer algorithm helps to reconstruct the thickness of the slice. The specifications of slice thickness are between 0.5 mm and 10 mm for most CT units. Obtaining a slice from one 360^{o} tube rotation required one second or less in a recent CT technology. CT allows quick scan acquisitions, greater bone details and shows classified images than magnetic resonance imaging (MRI). Generally, CT examination is restricted to the sequential plane. However, the images can be reconstructed into coronal, sagittal, or oblique planes or 3-D images. MDCT permits cube-shaped isotropic voxels acquisitions. Isotropic voxels have equal length on all coordinate planes (x-axis, y-axis, and z-axis). They permits direct reconstruction of images in every coordinate plane with greater image resolution (Brant W and Helms C, 2012).

Table I-4

The Hounsfield unit (HU) ranges for different materials and tissues (Brant W and Helms C, 2012)

Material	CT-number (HU)
Water	0
Air	-1000
Bone	400 -1000
Very dense bone	3000 - 4000
Soft tissue	40 – 80
Fat	-100 to -60
Lung	-600 to -400

1.2.1 Conventional CT

It acquires image data of one slice in a single tube rotation. The subject grasps his/her breathe, slice image is obtained, the subject release breathes and the bed travels. This imaging sequence is recurring until completing an examination. For any patient of scan volume, this method acquires as a minimum of 2 or 3-times the helical CT total scanning time. Optimization of scanning is more difficult during maximum contrast was given to patient. Slight lung volume alteration in each sequence of breath hold can create considerable deviations in the scanned anatomy of chest and abdomen consequent ‘skip’ regions. Current conventional machines able to simulate helical examinations through ‘cluster’ methods, such that numerous successive scans are taken through a single breathe grasp (Brant W and Helms C, 2012).

1.2.2 Helical/Spiral CT

Spiral CT involves moving the patient table via the opening of CT gantry at constant speed during data acquisition. The substantial artifacts coming from table movement has been compensated via wonderful interpolation. Hence, consistent 3-dimentional images data set can be created without gaps as well as supplementary artifacts. Recently, almost all CT body acquisitions is becoming helical scan standard (Grainger & Allison, 2015). Through a single breath hold, it is possible to acquire an uninterrupted measurements of image data. This method radically advances the image acquisition speed, removes artifacts produced through miss-registration besides dissimilarities of patient breathing, and allows scanning through optimal contrast opacification. Volume acquisition assists high detail 3-dimentional CT angiography, allowing retrospective reconstruction of numerous intersecting slices, and advancing visualization of small lesions. In modern multi-detector CT systems, the influence of pitch factor is greatly reduced in contrast to single slice CT systems. Helical scans allows multiple phase of organ enhancement (arterial, venous, parenchymal, and delayed) (Brant W and Helms C, 2012; Grainger & Allison, 2015).

1.2.3 Multi-detector Helical CT

MDCT is a modern type of helical CT imaging with great speed compared with single slice CT (SSCT), 5 – 8 times faster. Currently, the medical application of several rows of detector rings CT comprising from 2 – 512 detector rings and above are common in the world. In a single tube rotation, acquisition of several slices is permitted with increased proportion of the volume of target area covered. Narrow body scanning (1mm slice) with excellent resolution in any anatomic plane image reconstruction is possible in MDCT. Wide-ranging imaging also possible for high detail CT angiography, colonoscopy plus bronchoscopy. The main disadvantage of MDCT is the cost of maximum patient dose with 3 – 5 times greater than SSCT (Brant W and Helms C, 2012).

1.2.4 CT-Fluoroscopy

It allows real time imaging of the patient. This advanced CT technology radically advances the capability to conduct percutaneous interventions rapidly. When its risk of radiation is compared with conventional CT, it delivers lower radiation to the patient. CT fluoroscopy enables quick reconstruction of images, which gives real time image of lesions, anatomy and needle or catheter location. Nowadays, the main

function of CT fluoroscopy is towards directing drainage, cell removal, and interventional techniques everywhere in the proposed medical patient. CT fluoroscopy is mainly used to regulate needle placements in the body with physiologic motion like the chest and abdomen (Brant W and Helms C, 2012).

1.2.5 Dual Energy CT

Dual energy imaging CT uses two energy sources of different energies (high- and low-energy) and two highly sensitive x-ray detectors to near-simultaneous cross-examining tissues to know the reaction of tissue at varying energies of radiation (Grainger & Allison, 2015; Khaled ME. and Sandra AO, 2014). The dual energy CT image data sets (image) acquisition requires near-simultaneous recording of these variable energies, usually at 135 or 140 kVp and 80 or 100 kVp (Grainger & Allison, 2015). CT data sets acquisition approaches is dependent on each CT vendor. The source of these two types CT data set (different X-ray spectra) can be obtained by using a quickly alternating potential of a single x-ray tube or through operating two separate x-ray tubes at flexible potentials. Then, the X-ray detector record these high and low energy photons using special techniques of the detector systems (Graingr and Allison, 2015). To map the distribution of energy dependent absorbent materials in the CT image, 3-material decomposition algorithm is functional (Reiser M, et al, 2009).

Dual energy imaging emphasizes the variation of absorption coefficients of materials in space containing elements with high atomic number (like calcium or iodine) and low atomic number (like water or soft tissue) (Grainger & Allison, 2015) The greater variation of absorption coefficients between soft tissues, fat and contrast materials becomes highly clear at different energy levels that helps the professional for tissue lesion's characterization. An excellent encouraging application of dual energy CT in clinical setting related with the mapping of iodine distribution in the liver, kidney or lung and the removal of bone from datasets of angiography (Reiser M, et al, 2009). Kidney stone differentiation is also other clinical application of dual energy CT. It is also important to show the enhancement of contrast via color coding it in the CT image dataset or through subtracting it to get virtual unenhanced images. Hence, at normal dose level of CT scan, the tactic is highly important to acquire clinically significant information.

Imaging delivered by eliminating one material from others (Grainger & Allison, 2015). For instance, elimination of soft tissue produces iodine maps; elimination of iodine

produces a virtual un-enhanced image; and exclusion of calcium eliminates bones and calcified plaques. Normal looking CT images of good signal to noise ratio can be produced by blending these images together (Grainger & Allison, 2015).

Dual energy CT is twice faster than the conventional MDCT that enables to image the heart. The capability of determining the chemical composition of urinary that provide evidences in treatment selection decision to be either medical or surgical (Brant W and Helms C, 2012).

1.3 Medical Exposure

The medical usage of ionizing radiation is quickly increasing from time to time in the world including Ethiopia. The three main groups of medical practices comprising ionizing radiation exposures are diagnostic radiology, nuclear medicine as well as radiotherapy. Diagnostic radiology signifies a medical imaging procedures involving ionization radiation such that images obtained from CT, which is one of computerized reconstruction techniques in diagnostic radiology.

The greater practical application in medical imaging, higher frequency and the significant dose per examination of annual CT examination consequent rise in the amount of overall population radiation dose. UNSCEAR 2008 reported that computed tomography examinations accounted 34% of the collective dose in medical exposures. As a result, the overall population dose is increasing. Since the invention CT in 1970s, the frequency of CT examinations and its consequent population dose is ongoing to rise quickly. The introduction of helical and multi-slice CT scan examination played great role to reduce scan time, to conduct further exams at a time, to expand some exams scope and to announce new methods plus procedures (UNSCEAR, 2008). However, unnecessary exposures of patient dose to radiation caused due to the ease acquisition of images and the increase in the wide use of CT machines.

The level of patient radiation exposure relies on several technical and physical parameters. The factors that lead to the reduction of patient exposure (ICRP, 1982b) are; (1) the removal of non-useful radiation to beneficial image formation, and (2) the selection appropriate data recording methods to each diagnostic case. In order to foster efficient use of CT for patient diagnosis, it is preferred to: (1) use effectively designed recent CT-equipment, (2) not train experts using equipment and facilities that do not meet current standards, (3) withdraw unrelated design against radio-diagnosis, (4)

withdraw obsolete equipment from use for radio-diagnosis, and (5) perform periodic conduct of QAP to CT facilities. Important technical factors for achieving patient protection without compromising diagnostic information are strict collimation of the beam to the intended target volume, correct adjustment of recording system, applying regulation for the quality of beam of radiation, applying mechanisms to the maximum reduction patient dose, optimized selection of energy ranges of X-ray, beam filtration, wave form, shape and size of the focal spot, scatter beam reduction techniques and devices in general assessing use of modulation transfer functions (ICRP, 1982a).

Medical practice concerning ionizing radiation exposure is classified into three main groups called radiation therapy, nuclear medicine and diagnostic radiology. Diagnostic radiology is defined as images analysis obtained using X-rays including plain radiography (i.e. conventional and digital radiography), mammography, fluoroscopy, CT and others (UNSCEAR, 2010). Diagnostic radiology also includes image guided interventional procedures. Diagnostically image analysis should be done according to acceptable reference levels recommended by the country as well as by the international community. The medical reference level endorsed by the country is very important to protect the patient from clinically unnecessary exposure of ion forming radiation.

Medical practice of radiology is changing rapidly due to high supply of new imaging techniques in consistent to it invention. In many African countries especially Ethiopia, radiologic medical procedures including CT-examination can easily assessable, particularly in urban area, that will grow the net health benefit of the continent (UNSCEAR, 2010). Most important and distinct differences between medical exposures from other non-medical exposure is that clinical exposure is conducted voluntarily to deliver greater benefit than risks. Aside the diagnostic benefit of ionizing radiation to the exposed patient, dose must be reduced in harmony with the principle of ALARA without jeopardizing the image quality.

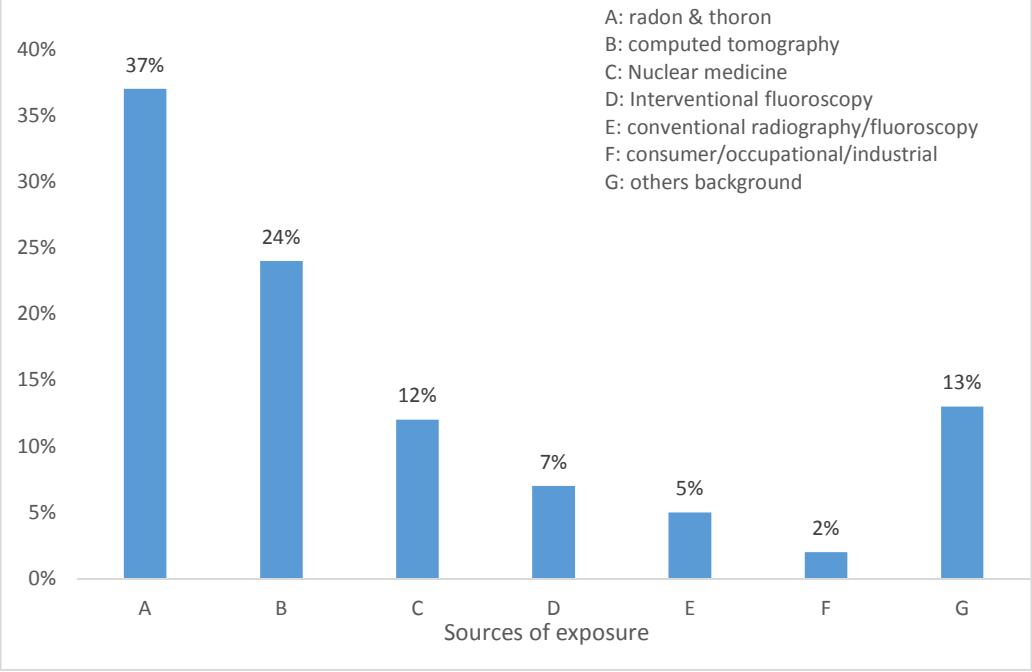
As mentioned earlier, medical radiation exposures are the most contributor towards the collective dose to the exposed individuals among all the other artificial radiation sources. The absorbed dose received by the patients from the medical procedure via digital radiography is lower in comparison to computed tomography procedure involving ionizing radiation (Souza A, et al, 2009; UNSCEAR, 2010). According to Jemal 2016 report in his thesis work, the world average annual dose from medical exposure is 14% on the world community attributable to clinical radiation exposure in

the year 1997 - 2007 (Jemal, 2016; UNSCEAR, 2010). According to Thulani 2012 report, the overall dose budget for the population of America obtained from conventional radiology and fluoroscopic procedure is approximately 5%, however from computed radiography is approximately 24%, Figure 1-2 (Thulani T., 2012). For the case of Ethiopian, the annual dose budget obtained from medical radiology from all radiological exposure has not been recorded, documented and studied yet.

Different diagnosis methods should be explicitly assessed before radiological examination referred. Many studies in the past years reveal that the same radiological examination is receiving different dose above two orders of scale. This high dose received by the patient required reevaluating the procedures. This high patient dose can also be considerably minimized without jeopardizing the medical diagnosis. To accomplish the objective of patient dose reduction, reference levels is practically essential to minimum radiation dose required for protecting the patient. Dose reference levels were primarily recommended in the 1990s by the International commission for radiation protection (ICRP) (Cynthia, 2010; Joseph et al, 2017; Ruiz et al, 2016; Souza A, et al, 2009).

Figure I-3

The diagram shows the sources of radiation exposure in America in 2006 (Thulani T., 2012)



1.4 Biological Risks of Radiation

The diagnostic potential of X-ray was acknowledged after its innovation in1895 by W. Roentgen (Bevelacqua, 2009). The presence of detrimental biological health effect (such as lose of hair and erythema) were quickly recognized by professionals in hospital the need to avoid over exposure. In 1920, United Kingdom proposed the general radiation protection recommendations and forthrightly in1925, the first international congress of radiology was apprehended. Then, international commission on radiological protection (ICRP) was acknowledged in 1950. Since formerly, ICRP published a series of recommendations which reveals understanding on the distractive bio-effect of ionizing radiation is increasing (Bevelacqua, 2009; Cember H, and Johnson E, 2009).

Physical and chemical reaction is initiated by biological tissues due to absorbed radiation which results in biological changes (ICRP, 2007). Improperly operated diagnostic X-ray (like CT-scan) can provide radiation dosage which can sufficiently create cellular reactions like radiation injury or severe radiation reactions. Conversely, when the radio-diagnostic study is conducted properly, such kinds of radiation effects may not occur because the dose is lower than the threshold level for such kind effects. However, the initiation of deleterious biological changes may not have lower dose limits. Even a lower amount of radiation dose can aggravate the risks of neoplasia growth. Additionally, a lower amount of absorbed dose of radiation by the gonads may induce chromosomal changes or mutations which lead to hereditary effects. This adverse health effects of radiation exposure are called stochastic effect that means the possibility of incidence of the injury rely on absorbed radiation dosage while the severity of the effect is independent of the dose. Hence, irrespective of the amount of risk to an individual it is possible to say that every increment of a particular radio-diagnostic examination dose may carry some risk. The factors affecting that used to estimate the quantitative relationship between radiation dose and relative risk are the energy absorbed distribution in the body, the rate of radiation dose, the exposed tissue, the cumulative dose and the exposed patient age (ICRP, 1982b).

Originally, most concern have been given to the radiation exposure risks rising from comparatively greater exposure dosages expected through a limited population (ICRP, 1982b). Nowadays, there is increasing concern that deleterious effects could be

predictable from the exposure of enormous numbers of people to low radiation doses of radiation.

The two general groups of adverse biological consequences from exposure of radiation are deterministic and stochastic effects. Deterministic effects are caused by high doses of radiation exposure resulting harmful tissue reaction ending the killing and malfunction of cells. Whereas, stochastic effect is caused by the low dose range of radiation exposure inducing cancer and heritable diseases. Stochastic effect is a probability with no threshold causing development of damage to the genetic materials resulting cancer many years later, and heritable diseases due to mutation of reproductive cells at their offspring in future generations (ICRP, 2007, 2016). As a result, stochastic risks are not preventable. Hence, to prevent unacceptable level of risks, exposure dose should be limited by authorized regulatory body in order minimize their occurrences. Heritable disease induced by ionizing radiation has not been revealed in human however there is genetic risk evidences from laboratory research conducted on animal's sperm and ova to reveal the development of heritable damage (ICRP, 2007).

1.4.1 Deterministic Effects

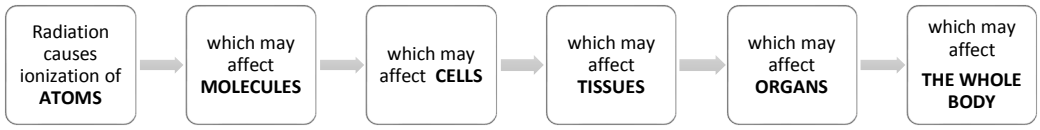
Doses required to produce harmful tissue reactions (deterministic effects) are greatly superior than the dosage limit advised by the scientific body (ICRP, 2007; UNSCEAR, 2000b). Tissue reactions correlated with the malfunction of cells that may lead to impairment of the organ or tissue (A. Tsalafoutas and V. Koukourakis, 2010). This clinically observable serious damage or death of cells, tissues and organs occurs when the delivered dose is above threshold levels, Figure 1-3. The extent of damage happen lonely if enormous percentage of cells in exposed tissues which rely on the quality of radiation delivered, dose rate and the total absorbed doses of radiation (ICRP, 2007). This implies that the severity of damage of cells or tissues or organs increases with an increase in radiation dose. Because of the radio-sensitivity difference between cells or issues or organs, there is variations in injuries due to high dose of radiation. The principal character of most deterministic effects are damage of reproductive capability cells, the growth of fibrotic processes and cell death.

Generally, whether the origin of radiation is natural or artificial; and whether it's the radiation dosage is small or big, the existence of biological effects will be irremovable.

Hence, it is possible to conclude as the following radiation consequence chain in Figure 1-3.

Figure I-4

Chained consequences for harmful tissue reactions of radiation in living matters



The harmful tissue reactions occurred due to radiation exposure are broadly categorized into early and late tissue reaction (ICRP, 2007). Early tissue reactions to radiation requires days to weeks which occurs doses above threshold levels. This tissue reaction is including the damage of rapidly proliferating cell systems comprising hematopoietic tissues, the gastrointestinal tract cell lining, the layer of skin basal cell as well as the reproductive cells of male, Figure 1-4. The inflammatory tissue reactions occur due to the release of cellular factors or loss of cells. Late tissue reactions take several months to years which resulting from direct and consequential types of damage to tissues or organs after exposure. There is a 5% possibility/Sievert for the occurrence of the late radiation effects (ICRP, 2017b). With respect to late tissue reactions, the most sensitive tissues are blood vessels, connective tissues, and lens of the eye.

Figure I-5

After brain perfusion CT procedures, patients upsetting from epilation (Di Zhang, 2012)

1.4.2 Stochastic Effects

The risks of cancer and heritable diseases demonstrated on irradiated and surviving cells. The induction of malignant neoplasms (cancer) due to radiation dose with uncertainty about 100 mSv (100 mGy) or less has been approved from epidemiological and experimental studies (ICRP, 2007). There are no straightforward confirmations to heritable diseases due to radiation exposure to human, however animal experimental explanations debate convincingly to consider future generation in the system of protection.

1.5 Biological Risks of Radiation in CT-Procedures

Diagnostic medical radiography has been increased in Ethiopia from time to time nowadays. It is highly beneficial in detecting the patient problem without surgical opening of the body. However, there are unavoidable potential risks with respect to medical radiography. Therefore, scientific researcher and medical professionals should cooperate with patient activism societies (societies who teach healthcare providers about the world of personalized medicine and assisted patients to change the healthcare system)[1]. The society make easy of understanding and communicating health benefits and risks of medical radiation in detail effectively (Lecomte J-F, Solomon S, Takala J, et al, 2015). Medically diagnosed subjects should be saved from needless as well as unplanned radiation exposure. For quality management besides medical-diagnostic dose optimization, professionals like medical radio-technologists, physicist and radiographers should perform an indispensable role in radiology departments. Therefore, patient-centered care system on radiological protection should be their central issues by national organizations, professionals and relevant stakeholders through implementing laws, guidelines, accreditation and education to cooperate control of biological risks incurred from medical radiation exposure.

Computed tomography (CT) scan uses much higher mAs than that is used in conventional radiography (Di Zhang, 2012). That means huge amount of photon beam penetrate the patient during CT-scanning procedures. The main purpose of diagnostic radiology (CT-scanning) is to produce a sharp shadow picture of a part of the patient even if some amount of radiation is absorbed by body of the patient (Di Zhang, 2012). The epidemiological studies revealed that clinical radiation dose from CT

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examinations sometimes exceeds the known references to accelerate the probability of cancer risks (ICRP, 2000a). According to ICRP publication 87 report, the effective dose in chest CT examination is 400 times more than chest routine radiography dose; and some CT procedures like that of pelvic region is 28.57 times than routine radiograph dose, for detailed information see the Table 1-5 below. In fact, that CT-scanning is prescribed justifiably with very favorable ratios to risk, there is a strong case to be made that several CT studies are being undergone internationally.

Table I-5

Effective doses in CT and radiographic examinations (ICRP, 2000a)

Radiographic examination	Effective dose (mSv)	CT-exam	Effective dose (mSv)	Comparison CT to Radiography
Skull	0.07	Head	2	28.57
Chest PA	0.02	Chest	8	400
Abdomen	1.0	Abdomen	10 – 20	10-20
Pelvic	0.7	Pelvic	10 – 20	14.28 – 28.57

The carcinogenesis effects induced by low dose ionizing radiation, such as doses from diagnostic radiology including CT scanning, is controversial. The induced cancer risk data collection from low dose radiation are mostly imprecise and conflicting. Hence, epidemiologic methods only cannot be applied towards approximating the risk of low dose from radiological studies. However, radiation scientists estimated this risk by extrapolating the linear non-threshold model calculation of risk at high doses. The main claim of using LNT model directed the general conclusion that very small amount of radiation dose can initiate cancer—for example, from diagnostic x-ray sources such as CT.

However, technical and clinical advancement in CT has not impose patient dose reduction per examination. Japanese Hiroshima and Nagasaki bomb survivor scientific research data informed the world community that dose from CT examinations incur a cancer risk induction (ARPANSA, 2008). In the case of pediatric patients of CT examinations, the epidemiological Japanese bomb survivor cancer risk suggestion is highly true such that the risk of probabilistic effects is greater than the over-all people. Furthermore, repetitive CT examinations (like four phasic dynamic CT-examination) have great possibility to consequence high amount of absorbed dose in the target volume, which can near or go beyond the limit for creating deterministic effects.

Therefore, CT procedures must be conducted based on the previously proposed protocols (ARPANSA, 2008; ICRP, 2000b).

Generally, image quality and the corresponding patient dose in computed tomography rely on the choice of CT parameters tailor to individual patient anatomy and diagnostic information essentially required which are used to conduct CT examinations (Alsafi, 2016; ARPANSA, 2008). The most significant CT parameters that determine the amount of radiation dose received by a patient in the control of the CT operator are tube voltage (kVp), milli-ampere second (mAs), nominal beam width (beam collimation), pitch, scan volume, patient body composition as well as diagnostic medical info are essentially required (ARPANSA, 2008; European Commission, 1999; ICRP, 2000a, 2007).

> *"For a particular patient, all other factors being kept constant, the patient effective dose will increase in direct proportion to the mAs and inversely to the pitch. As a result, with single slice scanners it has been good practice to choose the highest value for the pitch and the lowest value of the mAs consistent with obtaining the required clinical diagnosis. Since a pitch value of less than one is corresponding to overlapping scanning in sequential mode, pitch values have usually been chosen in the range of one to two and only in exceptional circumstances should they be less than one. With multi-slice scanners, some manufacturers have tied the selection of mAs and pitch together so that the ratio of the mAs to pitch remains constant when the pitch is altered. Under these circumstances, changing the pitch has little impact on patient dose and pitch values of less than one may be safely used" (ARPANSA, 2008).*

> *"The European Commission has developed quality criteria (ARPANSA, 2008; Commission, 1999) that result in recommendations concerning achievable standards of good practice for CT. These documents provide an operational framework for radiological protection initiatives in which technical parameters for image quality are considered in relation to patient dose. Diagnostic and dose requirements for CT are specified in terms of the quality criteria considered necessary to produce images of standard quality for a particular anatomical region. The subjective image criteria include anatomical criteria that relate to the visualization or critical reproduction of anatomical features. DRLs associated with the examination technique used for standard-sized patients outline the criteria concerning patient dose. Quality criteria have been developed for most examinations, together with examples of technique parameters influencing the dose" (ARPANSA, 2008).*

> *"Multi-slice CT scanners offer a number of clinical advantages, but because of a combination of their unique design characteristics and superior scanning speed, are capable of delivering high patient doses (ARPANSA, 2008; ICRP, 2000b, 2007) unless technical factors are carefully selected by the operator. Practitioners should be mindful that manufacturers of multi-slice CT scanners intend that the Radiation Medical Practitioner modify the default protocols to optimize the image quality/patient dose relationship. Substantial dose reductions without loss of diagnostic image quality can be achieved for even the average patient, by tailoring the technical parameters used in an examination" (ARPANSA, 2008).*

1.6 Benefit and Risk of Radiation in Medical Imaging

Daily, millions of people conduct evaluations of medical imaging to evaluate their health conditions. The growth of imaging technology empowers yielding a capital of convenient medical evidence. However, greatest imaging modalities are related with risks and costs to patients and the society like other medical studies (Khaled ME. and Sandra AO, 2014). The limitations are more pronounced when the examinations fail to produce intended information. For correct management of patient, awareness of the different medical imaging modalities as well as their costs and benefits is indispensable.

The medical applications of radiation involving radiotherapy, diagnostic radiology and nuclear medicine as well as interventional radiology remain the major causes of radiation exposure to human. Human beings are benefiting from advanced medical procedures such as medical diagnosis and treatment of diseases. Good medical practices in medical imaging of CT has tremendous benefits for clinical study by producing detailed information of the internal organs, limiting unwanted radiation doses, enlightening diagnosis, and improving treatment, particularly, medical imaging of CT generate details of tissues image (lungs, brains, abdominal organs, bones and blood vessels) inside the body, remove surgery, painless, accurate and fast to diagnose patients.

However, in developed countries, the average level of radiation exposure due to CT for only a few percent of examinations is approximately 50 percent of the exposure involving medical diagnosis (UNSCEAR, 2000b). CT scanners radiate x-rays to diagnose patients. The diagnostic x-ray will cause variations of the living matters building blocks starting at molecular, cellular, tissue and organs. Different tissues absorb x-rays different amount of radiation dose. Consequently, contrasts provide the anatomy and diseases detailed pictures of the target in the patient. X-rays absorbed by the tissues above certain threshold limits can cause biological effects attributable to sequential physical and chemical processes (C. Schmidt, 2012; UNSCEAR, 2000a). Biological effects of x-rays in patient arise instantaneously after the radiation passage via tissues resulting killing of cells. The cumulative effects of this effects produces the damage of tissue and organ. The dangerous cases of this effects will be death of the individual (UNSCEAR-vol II, 2008). Scientists concluded that ionizing radiation of

low doses could cause late carcinogenic and non-carcinogenic effects including genetic effects (Grainger & Allison, 2015).

The absorbed radiation by tissues able to breakdown the chemical bonds. This will result the releasing of charged ions which can break genetic code carriers called DNA. If the DNA break is unable to be repair, cancer of several organs will be incurred (Schmidt C, 2012; UNSCEAR, 2000b).

Epidemiological studies done on patients conducting medical examinations shown that low radiation doses less than 100 mSv given at low dose rates did not revealed reasonable increase in risks of cancer (Schmidt C, 2012; Gerber & Gibbons, 2010; UNSCEAR, 2000b). Nowadays, cancer risk estimation from CT examinations remain controversial topic. Researchers cannot formally have released convincing conclusion that CT scan can cause a risk of cancer. The prospective studies are still on going to produce conclusive results on this issues. For now, scientists approximate cancer consequences of CT scanning based on dose response models from patients treated with radiation and Japanese atomic bomb survivors (Schmidt C, 2012). LNT model suggesting that risk is directly proportionate to the amount of dosage acquired. The biological effects of ionizing radiation (BEIR) model suggests that "there is no safe level of ionizing radiation exposure; carcinogenic effects are assumed to follow a linear dose response, meaning even the smallest exposure carries some level of cancer risk". The BEIR VII model produces lifetime attributable risk (LAR) elements that approximate a probability of cancer incurred on an individual is proportional to the radiation dosage based on LNT model. The expected cancer from exposure of population can be obtained by multiplying the total of individuals irradiated to a certain dosage by the LAR (Schmidt C, 2012). Hence, the consequences of high probability of risks to the patient from diagnostic radiology have led the developments of managing patient dose mechanisms called diagnostic reference levels.

1.7 Diagnostic Reference Levels

Diagnostic radiology remains the greatest source of manmade radiation exposure (IAEA, 2002). CT is categorized amongst the greatest contributors of radiation absorbed dosages in tissues ranging from 10 – 100 mGy in medical radiology (ICRP, 2000a). It accounts 43 percent of the global total collective effective dosages (UNSCEAR, 2010). If this CT studies is inclining to repeat, then the dosages may habitually near or surpass the limits of cancer incidence that clearly witnessed in

people (ICRP, 2000a). Therefore, the international community has great interest for establishing DRLs for CT to protect the patient from the deleterious effect of medical radiation. In the habit of radiography, repeated radiography examination is common.

The mostly pronounced radiation protection principles are justification, dosage restriction, and optimization. Sometimes in the field of diagnostic radiography dose limitation is substituted by diagnostic reference levels without compromising image quality. Diagnostic reference level (DRL) represents the investigation dose level at which the dose is appropriateness for patient diagnosis. The dose delivered to the patient during diagnosis should follow the principle of ALARA without compromising image quality. That means DRLs are an important tool to optimize dose and acting as an activating level towards quality improvement in diagnostic radiology. Therefore, DRL is used to minimize the total patient dose-range observed in clinical practices. According to a broad investigation conducted using stated dose measurement protocol and phantom, the DRLs can be established at the 75^{th} percentile or 95^{th} percentile of the dosage distribution. DRLs can be developed for specific regions or for the country; which required periodical review to show variations in the average & standard deviation of the dosage distribution (McCullough, 2010).

DRLs was recommended by ICRP for the first time in 1990. However, the greater comprehensive recommendation was presented in 1996 (McCullough, 2010), which in short said DRLs is suggested by the commission for patients, which is a system of investigation level to optimize the patient radiation doses. DRL is not suitable to use as a regulatory purpose, because it is not the separating line between good or bad radiological procedures; however, it is considered as additions to the professional decision. DRL is linked to medical exposure, but neither applied to radiation dose limit or constraints nor applied to professional and public exposure. In an ideal world, DRLs thought the results of general patient dose optimization of radiation protection in clinical examinations. Actually, it is idealistically problematic. Ruther, it remains better towards picking detected percentile patient dose distribution as the foundational values for setting DRLs. These values should be proposed by professionals. The proposed DRLs should be reviewed at intervals. The review should inculcate the required changes and stability in the observed dose distributions. The DRLs values selected by the professionals could be particular to a country (Ruiz et al, 2016; Joseph et al, 2017; McCullough, 2010).

DRLs should be illustrated based on the simply as well as reproducibly quantified dosage metric exhausting clinical based method settings. The typical measurable quantities of computerized tomography (CT) scan are CTDI) and DLP. The dosage of radiation can be measured directly or indirectly (American Association of Physicists in Medicine, 2008).

1.8 Uses of Diagnostic Reference Levels

According to McCollough 2010 report, the use of DRLs remains approved through qualified experts besides governing organization in worldwide comprising ACR, AAPM, ICRP, UKHPA, IAEA, EC, and others. DRL remains developed at the 3rd quartile's mean/median dose distribution obtained from paramount data. The data should be collected from large, small, public and private facilities using an approved protocol, calibrated ionization chamber and phantom. DRLs can be developed for a region or country specific which requires periodical review (McCullough, 2010).

McCollough 2010 said that the main use of DRLs is to optimize the dose for the patient, particularly to decrease the patient dosage and to reduce the range of doses in clinical practices. The McCollough 2010 report emphasizes that there existed a 30 percent reduction of radiation dose between 1984 and 1995 and a 50 percent mean decrease between 1985 and 2004 because of DRLs use thought out UK (McCullough, 2010). This dose reduction data reflected that there were equipment dose efficiency improvements. When the dose exceeded the reference dose, it triggered investigations to find dose reduction strategies without arming diagnostic examinations. That means, the doses above the 3rd quartile will be drawn down below 3rd quartiles and the net effect dose distribution becomes narrower, which means lower average dose (McCullough, 2010).

Based on the ICRP 2002 review document said that the main purpose of DRLs helps to avoid doses to the patient that is not contributing to the clinical purposes during radiography. That is done by comparing the clinical procedure doses to the benchmark DRL. When clinically delivered radiation doses exceed the normal reference level, it is triggering an appropriate investigation to reduce the abnormal doses to be narrower (McCullough, 2010; Joseph et al, 2017; Wrixon, 2008). Therefore, DRL is applied to minimize the high or low dosage quantities distribution; encourage a narrower dose value ranges which characterize good practices; and encourage development near to optimal dose value ranges for specific examination.

CHAPTER II

THE STUDY OBJECTIVES

2.1 Health Status of Addis Ababa

Ethiopia was the home of old civilization of hominid habitation in the world with rich diversity of people and culture. It is located in Eastern Africa comprising neighbors of Sudan, Kenya, Somalia, Eritrea and Djibouti, Figure 2-1. The total area comprises 1.1 square km. Its official language is Amharic, and the regional languages are also officially acknowledged. According to UN 20017 population prospectus, the population of Nigeria (190.886 million) is the first in Africa and Ethiopia (104.957 million) is the second (United Nations, 2017). The 2007 population census of Ethiopia estimated with percentage contributions of Oromo, Amara, Somali, Tigray and others were 35.5, 26.91, 6.2, 6.07 and 26.32, respectively.

Figure II-1

Map of Ethiopia (World Bank, 2015)

The EMoH authorized towards framing the country's health policies and plans and establishes criteria in communication to hierarchical organization of the country such as regional health bureaus. Ethiopia is divided into administrative regions which has its particular Regional Health Bureau. The regions are progressively organized and subdivided in the form of region to zone, zones to woredas, and weredas to kebeles; each with Zonal Health Department, woredas Health Office and Kebeles health care service office, respectively, Figure 2-2. All are autonomous for bottom-up and top-down planning processes. These hierarchical organization of the country facilitated the effort to expand right to use, quality, democratization and decentralization of healthcare deliveries and systems in Ethiopia. The service provision and organization of healthcare has been concentrated at the kebele level.

Figure II-2

Administrative map of Ethiopia

Even though, the history of Ethiopian modern health system is comparatively short. Nowadays, new health reform programs and policies has been declared to meet the millennium development goals by exercising an appropriate health service

administration and management systems. The current health policy has been enacted in 1993. Hence, the policy "… focuses on the significance of attaining opportunity towards using to the essential primary healthcare delivery packages by the population throughout the nation by applying decentralized management systems" (Aynalem A, 2014). The fundamental concepts of country's healthcare policies stand on equity, democratization, decentralization, prevention, promotion, accessibility, acceptability healthcare deliveries to all population (Gevernment of Ehiopia, 1993).

The implementation of health policy existed through a 20 years HSDS, that has been consecutively categorized into a sequences of five years health sector development program implementation through a centralized system of governance in accordance to millennium development goals and national plans (Abebe A and Catriona W, 2015).

The nation's policy preparation and the first HSDP development have been the consequences of strict review and research on the root causes, nature and degree of the main health complications. The newly emerging problems on health has been also anticipated (EMoH, 2010).

The current Ethiopia's health policy has four successive phases of HSDP (EMoH, 2010; Richard, 2009). The program began with … a PHU which exists at kebele level in addition to create the basis of health care system. Currently, the country has been completing the IV health sector development program implementation; and investing plenty of money to improve the nation health care.

The 1993 health policy of Ethiopia stated in its preamble as "Ethiopia is placed among the least privileged nation in the world due to life expectancy, communication disease and malnutrition that consequently create high mortality and morbidity" (Government of Ethiopia, 1993). For instance, greater than 75 percent of the country's disease burden was related with communicable and preventable diseases (Abebe A and Catriona W, 2015; Aynalem A, 2014). That is why the main emphasis of primary health care system are sanitation and environmental health, disease control and prevention, as well as family health deliveries (Rad-aid, 2007).

Currently, it has been recorded remarkable progresses towards improving the country's health records in the previous two decades. But, the poor health records of the nation resulted greater rate of mortality and morbidity compared to other African countries. According to Ethiopian the ministry health HSDP-IV (2010/11 – 2014/15) 2010

report, life expectancy of Ethiopian population of is 53.4 and 55.4 for male and female, respectively (FMoH, 2010), which shown major progress compared to the low life expectancy record before 1993. Ethiopian population life expectancy of 54 recorded in 2010 (FMoH, 2010) is increased to 64 in the recent 2017 report (WHO, 2017).

The 1993 health policy of the country stated that *"diagnostic and supportive services for health care shall be developed by strengthening the scientific and technical bases of health care, facilitating prompt diagnosis and treatment, and providing guidance in continuing care"* (Government of Ethiopia, 1993). Hence the country health care service infrastructure has been significantly grown. For instance, medical radiography, not stated explicitly in the primary health care unit (PHU), can be exploited to advance and simplify diagnosis and monitoring of communicable diseases (Rad-aid, 2007). Utilization of CT in health care services is highly appreciated and common in the world to facilitate fast diagnosis of the patient.

2.1 Radiological Status in Ethiopia

The government of Ethiopia has been devoting comprehensively to strengthening the health systems and upgrading the population's health record. The Ethiopian Ministry of Health is working towards meeting the government's development agenda (and the millennium development goal) and vision in the health sector (EMoH, 2015) through maximizing the application diagnostic imaging facilities throughout the country. The national care health policy of Ethiopia is a key of basic social services that are directly connected to the national growth and development plan and social health that is targeted to satisfy the less privileged rural population which consists about 81% (FMoH, 2010) of the total population of Ethiopia.

The national government healthcare policy is designed to deliver reasonable and affordable excellence wellbeing care deliveries to every population sectors (Gevernment of Ehiopia, 1993). Diagnostic medical imaging services have been playing a vital role in the healthcare system by providing high quality of services all over the world (Kwadzo, 2016) including Ethiopia. Diagnostic radiology empowers medical practitioners for their precise decision in delivering quality service to the end users by applying best practice approach to their clinical effort. The application of diagnostic medical imaging, particularly CT, is increasing throughout the country. In this case, diagnostic x-ray departments are established in all involving health institutions. The diagnostic medical imaging working at health institutions comprising

both conventional and digital radiography including CT. According to the report obtained from the Ethiopian radiation protection authority, the total numbers of CT distributed in the entire country were 92; out of figure 38 percent distributed in Addis Ababa and the remaining 62 percent has been distributed to the rest of the country. Hence, special attention should be given to Addis Ababa's diagnostic imaging facilities towards applying essential ethics of justification and optimization when conducting the procedures. Radiation exposure in medical practices must be confirmed through evaluating the benefits and risks (always satisfactory net benefits must be greater the risks) by applying alternative methods of dose optimization that exclude unwanted patient dose. The optimization perspective requires the medical practitioner's ability to ensure to use the minimum amount of radiation to be used to the intended diagnostic objective. To make easy for the medical practitioner, the application of DRLs as an instrument for optimization of protection to the patient to all radiologic procedures. Therefore, DRLs should be developed and implemented into the health care system to ensure optimized patient dose delivery and consequently improves diagnostic values in the country, especially, in Addis Ababa.

2.2 Ethiopia's Legal Framework on Ionizing Radiation

The national radiation protection authority was established in 1993 targeting towards protecting the general public as well as the environs from the deleterious effects of ionizing radiation. The proclamation has been revised by "Radiation Protection Proclamation No. 571/2008" in 2008 which focus on radiation safety including non-ionizing radiation. Currently, by analyzing the existing regulatory system and benchmarking international (IAEA) standards and other important national regulatory working systems, the new proclamation has been approved by the parliament "Radiation Protection Proclamation No.1025/2017" in 20017. The preamble of proclamation no.1025 stated as "the application of nuclear technology as well as radiation is rising from time to time in the social and economic activities in the country with parallel consequences health risks" (Federal Negarit Gazette, 2017). According to the proclamation No. 1025 description, its objective is to protect individuals from radiation hazards, this shows the authority has empowered to take any measure like setting DRLs to protect the current as well as future generation of the country from the hazards of radiation. For this purpose, Practice specific guidelines for diagnostic and interventional radiology and dental radiology which are compatible with the

proclamation 1025, IAEA-BSS and other relevant documents, are established and implemented. For medical exposure, authorization applicants are required to submit a description of their radiation protection program that includes details on justification, optimization, responsibility, medical dosimeter, calibration, quality assurance, dose constraints as well as investigation of accidental medical exposure (Federal Negarit Gazette, 2017).

In proclamation 1025, the article quoted relevant to medical imaging is article 14(4) "… the direct or indirect exposure of radiation from authorized practices must confirm the practicability of ALARA principle lower than the permissible dose limit" (Federal Negarit Gazette, 2017), this implies that the development and application DRLs to protect the patient is emphasized in this article. However, the development of reference levels to protect the patient radiation dose has not revealed in the country, yet.

That is why this document is convinced to emphasis patient's clinically unwanted dose reduction mechanism in the hospitals in Addis Ababa. This research is mainly focused on the development of DRLs to be used by the hospitals and clinics who are concerned with diagnosis using CT. By the way, the DRL concept is principally introduced ICRP in 1996 (ICRP, 2017b; McCollough, 2010). And settled further to bring to practical application then guidance in real-world has been started in 2001 (ICRP, 2017b). In African countries including Ethiopia, DRLs has not become fully practical to investigate patient dose except the research work done in some of the continent like in South Africa, Egypt, Kenya, Cameroon, Nigeria, Sudan etc. (Ali, Elawad, & Ibrahim, 2016; Joseph et al, 2017; Korir et al., 2014; Moifo et al., 2017; T. Nyathi, 2012; Salama et al., 2017) until now. Although Egypt has been revealed national DRLs for CT in 2017 for the first time, professionals have expressed their doubt concerning about the dose of radiation. Because, the dose for different examination is using consistently higher than the established DRLs[2]. Therefore, they need to improve for the selection of exposure parameters for different examination and the dose to the patient. Ethiopia has not established it national DRLs so far.

On the other hand, European commission adopted DRLs to its member state principally through 97/43/Euroatom directive in 1997 and defined DRLs as "the radiation dose levels in medical diagnostic for representative exams for standard man

[2] https://www.auntminnie.com/index.aspx?sec=log&itemID=117732

or phantoms for generally well-defined equipment kinds" (Paulo, 2015) in its article 2. The previous directive was strengthening with directive. The 97/43/Euroatom directive was forcing its member state by saying "each member must develop and use DRL in clinical exams using radiology". The 2013/59/EURATOM directive stressed and generalized the need of DRLs to its member state by saying "you shall confirm the development and periodic review as well as application of DRLs in medical exposure" in article 56(2). Look at how DRLs significance to country has given great attention by the international community. This implies, Ethiopia is too late to develop DRLs involving CT procedures which is a common practices of exposing patients to ionizing radiation going to hospitals for medical practices.

2.3 Statement of Problems

Ionizing radiation has helpful benefits in the diagnosis and therapy of patients. However, besides its use, the biological deleterious effects of radiation on patients, the public and the environment are highly inducing tension to the world population. The tension is the controversial ill-effects of radiation, which is a growing concern in the world (Nyathi, 2012). This is more series in case of CT examination, because it delivered higher dose to the patient with higher IQ obtaining (ICRP, 2017b). Recent research outcome revealed that at lower radiation dosages, sufficient diagnostic information can be achieved. Therefore, the scientific body should find scientific means to minimize these effect by putting procedures and dose references. Even though, the manufacturer of the X-ray machines provides dose guidance levels for each radiology department to aware them the dosage given to the subject as well as the dose that reaches image detectors, dose optimization towards the patient is highly essential (Gibson, 2011). Because, for instance in Screen-film radiography overexposed and underexposed are expressed in terms of the darkness or whiteness of the film after exposure, i.e. dark film show overexposure and white film express underexposure, whereas in CT scan, the higher the exposure the higher the quality of image produced. In digital radiography (CT-scan), exposure indicators are the only means to control overexposure and underexposure. This exposure indicator formulated by the manufacturer in CT-scan has been arranged by targeting maximum image quality delivery by ignoring the optimum doses to the patient. This implies that, the higher to dose to the patient the higher probability of inducing radiation related ill effect (cancer or leukemia) to the patient.

Currently, medical diagnostic procedures in radiology involved the largest portion percentage of the entire public dosage from artificial radiation. Mainly, patient exams spending CT (CT-scan - digital radiological techniques) accounting more than 34% of the total collective effective radiation dose (UNSCEAR, 2000b), which were expected to offer the potential for dose reduction. In practice, however, higher doses are correspondingly possible than general radiography (IAEA, 2007). Therefore, mechanisms should be created to optimize and control these dose such as developing procedural dose guidance levels (i.e. DRLs) specific to a nation or a region.

According to IAEA 2007 report, above 90 percent of the population dose contributor is being diagnostic X-ray from artificial sources sourced from medical ionizing radiation (above 90%). Because large amount of X-ray exams is being conducted every year. The typical highest organ doses (ranging from 10-100mGy) resulted from CT-scan procedures. This value is estimated below the dose needed to induce deterministic effects. But, every practices of X-ray can increase the induction of stochastic effects like tumor and/or hereditary (IAEA, 2007).

This higher patient exposure has been tumultuous in the radiology department in Ethiopia, especially, the phobia developed by radiation officers from radiation hazard scenarios of CT scan. However, hospitals and clinics have not been giving concerns about the patient's radiation dose receiving during CT scans. Because radiologists/radiographers follow the philosophy of 'more is better' in patient exposure that is delivering high CT dose. Most CT medical diagnosis of patients are repeated so many times. The dose received from CT scanning is confirmed much higher than the dose from conventional radiography. However, there are confirmed evidences that medical exposure can be substantially reduced without affecting the required diagnostic information. Conversely, the manufacturers of CT scan have been working by focusing on the modality centered on speed to maximize the data acquisition; this will true at elevated radiation dose to the patient. Furthermore, international papers revealed that there are great variations on the patient's dose for the same types of CT examinations carried on different patients in different hospitals. Most likely, the situation in CT practice in Ethiopia is also the same.

The world radiologists are being confused between minimizing patient exposure and image quality. Most of the time they are using higher exposure parameters to get high quality of radiographic image to deliver appropriate diagnosis of the patient. However,

in diagnostic radiology, patient dose evaluation is needed to confirm the proper performance of the X-ray equipment with the ultimate goal of patient dose management. Most X-ray machine manufacturer recommended to the user to use dose guidance levels during diagnosis which is the current practice in the globe. Derivation of dose guidance levels required wide scale data with frequent review as emerge of techniques and new technology. The EC and International Atomic Energy Agency (IAEA) as well as some other countries have been implementing the same approach (IAEA, 2007).

The radiologists will adapt exposure indicators as normal exposure values to the patient and consequently, they will not review exposure indicators which finally result from exposure creep (Gibson, 2011). Therefore, the nation should establish DRLs to optimize the radiologic dose to the patient. The DRLs will impetus the radiology department to develop its own exposure latitude.

According to the current international researches, during X-ray diagnostic and procedures, the patient dose varied from hospital to hospital by a factor of 100; which affects the image quality (Nyathi, 2012). Moreover, the numerous researches on patient radiation doses strengthen the worries of radiation-induced diseases like cancer in the world. Therefore, every radiology institute should in-place the mechanisms of optimizing patent doses without affecting clinical image quality.

In the diagnostic procedure, the minimum acceptable dose references should be set by the nation as the national DRL for each procedure to optimize doses. In addition to the heavy burden of diseases consequences of great mortality and morbidity, currently, Ethiopia is also facing the potential health effects of radiation with a growing prevalence of x-ray facilities in the country; which is the most requested examination in many clinical situations. This medical procedure is the most significant source of radiation exposure to the community, which consequently will result in the development of cancer cases burden in the country. Conversely, such kinds of extensive research has not been done regarding the cancer case burden of radiation in the country, yet. In this regard such fundamental as well as recently evolving health complications, the development of the minimal DRL is indispensable which should be population-specific.

ICRP introduced the recommendation of DRLs in 1996. In 1997, the European Union approved the council directives 97/43Euratom. However, the African Union has not developed any directive related to the development of DRLs associated to radiology. But, some African countries like South Africa, Nigeria, Sudan, Kenia, Egypt, Cameroon, and others tried to put the base line to establish their own DRLs. Nevertheless, such kind wide radiological study related DRL on CT has not been done in Ethiopia yet, except some institution specific works. Therefore, this research work will fill the gap reflected in this perspective, publicized dose references to the concerned bodies for their action and make ease for the healthcare issues to improve patient dosage controlling mechanisms in the Addis Ababa. Hence, the objective of this research work will identify the possible ways of minimizing patient dose to commensurate the dose to clinical purpose by setting the DRL for the selected CT procedures in the country.

2.4 Goal of the Research

This research was designed to be conducted in three phases in order to achieve the defined objectives. The first phase included collection of retrospective data from the health institutions archives and define the first local DRL per anatomical regions. The second phase required the collection of patient data using phantom and pencil ionization chamber to establish new post optimized DRL per anatomical regions. The third phase required the collection CT images for common procedures using ACR CT accreditation phantom to assess image quality.

The research primary aim is towards developing regional (Addis Ababa, Ethiopia) DRL to common procedures of CT-scan examinations which help to avoid unwanted radiation dose to patient's clinical purpose in medical imaging and to use the possible minimum patient dosage without compromising IQ. The statistical quantities of the appropriate health facilities data were compared with acceptable reference recorded internationally.

2.4.1 Specific Objectives

The specific objective of the study work was defining DRLs of the most frequent diagnostic procedures in CT-scan examinations of adults. Critically, the specific objectives emphases to:

1. Collect sufficient data so as to recommend a methodology aiming at minimizing doses in adults, when feasible;
2. Quantify $CTDI_v$ as well as DLP in the best common CT-scan examinations and estimate DRLs;
3. Compare the DRL values obtained with published international works of literature;
4. Optimize examination procedures in order to improve radiographers' best practice;
5. Encourage a best range of DRL values for common clinical imaging protocol; and
6. Set a comprehensive framework for future studies covering the whole of the country.
7. Publication activity and write up of the thesis completion

The most frequent diagnostic procedures in CT-scan examinations were abdomen, chest, head, cervical spine and pelvic. The selection of CT-protocols was based on existing literature from survey of frequently performed CT-scan in Addis Ababa. The study is designed based on scientific methods of diagnostic radiology to foster the optimization of clinical radiation doses of a patient deprived of negotiating the diagnostic IQ practiced in the Addis Ababa region.

2.5 Significance of the Study

This research is the first in its kind in Ethiopia corresponding to the aforementioned problem above. The study was contributed a great deal of importance with respect to patient dose optimization. Additionally, the establishment of DRL will improve the quality of dosage without jeopardizing the radiogram quality present and future radiological procedures in Addis Ababa, Ethiopia. Furthermore, the development of DRLs can enrich the capability of radiographers towards dose optimization congruent to ALARA principles. Correspondingly, the measure taken to dose optimization will help to reduce the complexity arising from radiation exposure due to medical interest. Hence, the success of this research will comprehend towards the achievement of the aims of the ministry of health of Ethiopia as well as the goal of WHO as a whole regarding the patient's radiation dose, which means it will improve patient safety and concert of the image processing machines. The goal of this research is to develop of regional DRL which help the procedure practitioners to use the possible minimum

radiation dose to diagnose patients, that aims to provide intense knowledge regarding radiation patient doses which fill the knowledge hole existed in radiographic practice in the region of Addis Ababa. The inclusive method of this research will open the eyes for further study to reach the whole of the country to communicate the challenges of radiography to the patient's dose. Generally, the importance this research will extend to the development of quality standards patient care culture and/or policy quality service by the radiologists and radiographers in the x-ray facilities in Addis Ababa, Ethiopia. Additionally, the respective regulatory bodies of the country will be initiated to accept and structure radiological research findings as part of service delivery policy.

2.6 Scope of the Research

The research carried towards directing radiological topics connected to essentially DRLs to patient. Because DRLs is mainly important in radiation protection in X-ray medical imaging. The research scope was restricted predominantly on medical diagnosis accomplished in Hospitals in the city of Addis Ababa. For the DRLs, the most frequent diagnostic procedures in CT-scan examinations that were used to establish DRLs are abdomen, chest, head (brain), cervical spine and pelvic. According to a broad investigation conducted using stated dose measurement protocol and phantom, the DRLs can be established at the 75^{th} percentile of the dosage distribution (McCullough, 2010). The outcome of the research was compared with international records and references. The following chapter explains the details, importance and establishment of DRL in the international point of view and national perspectives including the mathematical brief in developing DRL in CT.

CHAPTER III

LITERATURE REVIEW

3.1 Overview

The application of CT machine introduced new anatomical imaging of human body in the medical fields has been increasing since its invention by Godfrey Hounsfield in 1969. Medical imaging with CT is increasingly utilized in health care for diagnosis of diseases. It has become commonly accessible and are frequently used. The CT technology have become sophisticated and advanced. Currently, advanced 3-dimentional imaging and 4-dimentional dynamic imaging becoming common health facilities (Grainger & Allison, 2015). 3.6 million medical diagnosis using X-ray imaging including CT examinations has been conducted annually until 2008 (UNSCEAR, 2008). Current investigations in medical centers revealed that diagnostic radiology contributed 60 – 70 percent patient dose annually (IAEA, 2009). Particularly, CT procedures accounts 25 percent of annual patient dose receiving from all X-ray studies (IAEA, 2009). As a consequence, fears about short and long-lasting radiation risks developed to world community (Grainger & Allison, 2015). This high dose delivery to patient enforces the world community to develop dose reduction mechanisms. According to the IAEA basic safety standard publication and international commission on radiological protection recommendation and guidance, diagnostic reference levels are essential approaches to optimize the patient dose without compromising the image quality in medical diagnostic fields (Japan, 2015). The reporting professionals and the CT protocols are main factors of image quality in CT (IAEA, 2009).

The fast image acquisition and enhanced image quality of computed tomography scanners are due to the invention of multi detector. Nevertheless, radiation dose concomitant with CT procedures and the radiation induced cancer are a pronounced concern for patients such as pregnant woman, children and teenagers. The design and acquisition protocols are parts of many parameters of CT that determining the amount of radiation a patient received (ICRP, 2017a).

UK and USA were implemented patient dose surveys at the national level on diagnostic radiography in the beginning of 1905s. Then, in the 1980s, entrance surface

exposure (ESE) for patient was measured by public health England in United Kingdom (Brink A and Miller L, 2015). The main basis of recommendations for patient dose from radiography was results similar works in the world. When Ethiopia is compared with US, UK and some Europe in the development of diagnostic reference levels, the three of them established their DRL before 2000, see Table 3-1 to see dose distributions for some examinations of radiology. However, Ethiopia has not developed yet in the 2020s.

Table III-1

Contribution of radiation doses from different radiography (MK Abdulkadir, 2015)

Examinations	Percentage of diagnostic image studies	Percentage of radiation exposure
CT	17	49
Interventional	4	14
Radiography	74	11
Nuclear medicine	5	26

ICRP presented the idea of DRL in 1990 by publication 60, and first recommended in 1996 by ICRP publication 73 (J-F. Lecomte, S. Solomon, J. Takala, T. Jung, P. Strand, C. Murith, S. Kiselev, W. Zhuo, F. Shannoun, 2015). DRLs deliver mechanisms for a facility to compare their patient dose with the national DRL benchmarks (Brink A and Miller L, 2015). The system of DRL establishments requires setting radiation dosage, determining quantities to set DRLs, deciding the units to set DRLs, standardizing the dose quantification methods, collecting data, and the practical application methodology of DRL (Japan, 2015). This also involves the setting of DRLs at the 3rd quartile mean (median) dose distribution of patient data ($CTDI_v$ and DLP) such that when the facilities mean (or median) patient dose distribution is above the NDRL, they are counseled towards reviewing their procedures to decrease dose lower than the DRL. Hence, DRLs promotes justification and optimization principles of radiation protection. The ICRP recommends that “if it is found that procedures typical dose exceeded subsequently the relevant DRL, then the reason (procedures and equipment) should be resurveyed in order to define whether the protection has been sufficiently optimized. If not, measures aimed at reduction of the doses should be taken” (Brink A and Miller L, 2015).

DRLs produces synchronized proofs about patient dose that helps for professional judgments. In addition, it is highly significant to promote instant investigation

whenever unusually high or unusually low patient doses is noticed at the health facilities. It requires periodic review (in 3 to 5 years) to improve excellent practices at lower patient doses (ICRP, 2017b) deprived of compromising the patient care and image quality. Table 3-2 shows national DRLs for some countries in the world for selected CT procedures.

Table III-2

International DRLs of some countries for a few CT procedures

Country	Descriptor	Head	Chest*	Chest**	Abdomen	Pelvic	C-spine
IAEA CRPR (Tsapaki V, Aldrich J, Sharma R, et al, 2006)	$CTDI_w$	47	9.5		10.9		
	DLP	527	447		696		
EU (European Commission, 2014b)	$CTDI_v$	60	10	10	13-35		
	DLP	1000	400	400	460 - 1200	450-650	400-600
USA (Kanal et al, 2017)	$CTDI_v$	56	12	13			28
	DLP	962	443	469			562
Japan (Japan, 2015)	$CTDI_v$	85	15	15			
	DLP	1350	550	550			
Canada (Health Canada, 2016)	$CTDI_v$	82	14	14			
	DLP	1302	521	521			
UK (Kanal et al, 2017)	$CTDI_v$	60	12	12	14□		21□
	DLP	970	610	610	910□		440□
Australia (Kanal et al, 2017)	$CTDI_v$	60	15	15			
	DLP	1000	450	450			
Korea (S.w. Yoon, 2018)	$CTDI_v$	63.7		7.3	10.58		17.89
	DLP	1119.4		297.05	1511.41		434.04
Ireland (S J Foley et al, 2012)	$CTDI_v$	58	9	9	13		19
	DLP	940	390	390	1120	570	420
Syria (Ataç G, Parmaksız A, İnal T, 2015)	$CTDI_v$	60.7	22	30.5	24.1	27.5	
	DLP	793	520	133	721	542	
Turkey (Ataç G, Parmaksız A, İnal T, 2015)	$CTDI_v$	66.4	11.6	11.3	13.3	19.4	
	DLP	810	289	283	204	421	
South India (Saravanakumar A, Govindarajan K, 2016)	$CTDI_v$	47	10		12		
	DLP	1041	445		550		

CTDIv=volume computed tomography dose index, DLP=Dose length product, ∗=chest without contrast, ∗∗=chest with contrast and □=(NRPB, 2020)

In 1997 after 1996 ICRP recommendations, the European Union approved the council directives 97/43Euratom with article 4 (sub-article 2) said each EU "member states shall promote to the develop and the use of diagnostic reference level for radio-diagnostic examinations" (European Commission, 1997); but the council directives 97/43Euratom has been repealed by the council directives 2013/59/Euratom on 5 December 2013 (European Commission, 2014a), its article 56 said "member states shall ensure the establishment, regular review and use of diagnostic reference levels for radio-diagnostic examinations"; for some countries DRLs, look at Table 3-3.

Optimization of radiation exposures in medical imaging techniques initiatives have been conducted at international, regional and national levels nowadays. Optimization of patient dose in medico-radiology via the implementation of DRL as a tool have been endorsed and popularized by the work done by ICRP, WHO, IAEA and multi-society campaigns pursued by professional societies activities enhancing implementation of optimization (Lecomte J-F, Solomon S, Takala J, et al, 2015). For instance, international atomic energy agency BSS (GSR part-3 – paragraph_3.160) said that "registrants, licensees and radiological medical practitioners should ensure that protection and safety are optimized for each medical exposure" (IAEA, 2011). According to ICRP, DRL can be defined as such "a form of investigation level, applied to an easily measured quantity, usually the absorbed dose in air or tissue-equivalent material at the surface of a simple standard phantom or a representative patient" (Japan, 2015). These definitions of DRL strongly stresses that it is not a dose limit or dose constraints and does not give a dividing line between good and poor radiology practices.

Table III-3

Current national diagnostic reference level (NDRL) in CT in EU member state (European Commission, 2014b)

Country	Descriptor	Head	Neck	Chest	Abdomen	Pelvic	AP	C-Spine
EC	CTDIv	60		10	13 - 35			
	DLP	1000		400	460-1200	450-650		400-600
Switzerland	CTDIv	65	20	10		20	15	30
	DLP	1000	500	400		500	650	600
Norway	CTDIv	70	20	15	18			20
	DLP	1000		400	800			400
Slovenia	CTDIv			15	17			
	DLP			475	555			
UK	CTDIv	60		12	14		13	28
	DLP	970		610	560		510	600
Germany	CTDIv							
	DLP	900		400	900	450		
France	CTDIv	65		15				
	DLP	1050		475				
Luxembourg	CTDIv							
	DLP	1000	440	270				440
Sweden	CTDIv	75		20	25			
	DLP	1200		600				

CTDIv=volume CTDI, DLP=Dose length product, AP=abdominal pelvic

For example, as stated by McCullough 2010 record commencing the nationwide dose investigation of UK, a 30 percent drop in radiation doses from 1984 to 1995 and 50 percent fall from 1985 to 2004 were recognized due to DRLs practices in the whole country (McCollough C, Branham T, Herlihy V, et al, 2011). Nevertheless, most African do not develop DRLs to protect their patients from abnormally high or low dose practices of diagnostic radiology without jeopardizing the intended image quality. Since then, the African Union has not developed any regulations or directives related to DRLs associated to radiology practiced in Africa. Table 3-4 shows DRLs benchmarks of some African countries for a few CT procedures, empty cells in the table shows the nonexistence of a corresponding data. This indicated that certain African countries such as Nigeria, Sudan, Egypt, Kenia, South Africa, Ghana, and others irritated to develop their individual standards for their upcoming DRL settings.

Table III-4

DRLs benchmarks of certain African countries for a few CT exams

Exam	Descriptor	Head	Chest	$Chest_{HR}$	Abdomen	Pelvis	LS	AP
Nigeria (Yurt et al, 2020)	$CTDI_v$	61	17		15*			20
	DLP	1310	735		757*			1486
Sudan (MK Abdulkadir, 2015)	$CTDI_v$	65	11.5		11.6			
	DLP	758	327		437			
Kenya (Korir, Wambani, Korir, Tries, & Boen, 2016)	$CTDI_w$	61	19		20	21	20	18**
	DLP	1612	895		1842	1928	712	1182**
Cameroon (Moifo et al., 2017)	$CTDI_v$		52				25	15
	DLP		1151				769	716
Morocco (Semghouli S. et al, 2017)	$CTDI_v$							
	DLP	1408						
Egypt (Ekpo E et al, 2018)	$CTDI_v$	30	22	22	31			31
	DLP	1360	420	420	1425			1325
Tanzania (Muhogora et al., n.d.)	$CTDI_w$	60	30	35	35	35	35	
	DLP	1050	650	280	780	570	780	
Algeria (Khelassi-Toutaoui N, Merad A, Tsapaki V, 2020)	$CTDI_v$	50			25		35	
	DLP							
South Africa (M. Nyathi & Shivambu, 2019)	$CTDI_v$	32	32					7
	DLP	767	593					386
Cote DIvoire (Monnehan G, Silue K, Djagouri K, 2017)	$CTDI_v$	50.9						
	DLP	982.879						
Tunisia (Muhogora W, Ahmed A, Beganovic A, 2009)	$CTDI_v$	24.3				25.4		
	DLP	874				599		

$CTDI_v$ = volume CTDI, DLP=Dose length product, $chest_{HR}$= high resolution chest, LS= lumbar spine and AP=abdomen/pelvic; *=(European Commission, 2008), **=(IPSM, 1992)

Although dose limits must not be exceeded, DRLs may be exceeded if clinically necessary. DRLs also differ from dose limits for occupational exposure because they

are not used to constrain individual patient exposures; this is because a dose higher than the standard dose may be required depending on the patient's body size and weight. DRLs are a tool for identifying facilities with unusually high doses and for promoting the optimization process (ICRP, 2017a). Separate DRLs have been established for each country and/or region because equipment and procedure protocols can vary between different facilities in countries or regions. DRLs are set for patients with standard sizes or for standard phantoms in easily measurable and highly reproducible dose metrics; they should not be set as effective doses. Image quality is considered when setting DRLs. When setting DRLs, dose measurements are first performed using a previously standardized method for each type of radiation examination. DRLs are generally set as the 75th percentile of the typical dose distribution (mean or median) for a patient or phantom measurement. However, this does not necessarily apply to examinations with narrower dose distributions because of the effect of optimization (Japan, 2015).

When it is said reduced mean (median) radio-diagnostic dose, it doesn't mean modification in clinical required image quality as examinations proceed at consecutively practiced at lower radiation exposure. Medically essential quality of image shall not be compromised in the process of DRL establishment. However, the subjective aim of DRL is that medical exposure levels should fulfil ALARA principles proportionate with the aim of CT-procedures. Clinical diagnosis of patients should be conducted consistently below the national DRL without jeopardizing the required image quality.

Based on the medical practice and equipment changes in a health facility, periodic review of the DRLs is mandatory. To accomplish the review, large amount of data collection regarding patient dose is highly significant. Recently, the application of technological data collection method made easy to conduct dose survey in a country.

Most countries in the world established DRLs at the 75th percentile to detect doses of clinical practices. When the doses of clinical practice are above the established DRLs, they take corrective actions. However, they abandon optimizing radiation doses below the 75th percentile. The optimization of radiation doses between 25th percentile and 50th percentile is known as achievable dose (AD). AD is significantly applied in the modern CT technology to improve protocols (ICRP, 2015). DRLs will provide incentives for optimization for the 25% of facilities that use dose over the DRLs for a particular

examination. Then, the researcher recommends the use of ADs to encourage optimization for the 75% of facilities that already use dose within the current DRLs. The use of diagnostic reference ranges (DRRs) have also been proposed as a useful approach for quality improvement. DRR can indicate investigation or action levels ranging from the lower value, below which reduced image quality may not be diagnostic, to the upper values, above which the dose may be in excess.

Generally, the key and complementary radiological safety principles are highly required to achieve the principles of dose reduction to the medical patient. "*The goal of radiation protection against ionizing radiation protection in medical imaging is to restrict radiation dose to the staff, general public and patients to maintain below the level at which deterministic effects occur and the probability of stochastic effects is limited to an acceptably low level. To achieve this, the ICRP recommends the application of three principles called justification, optimization and dose limitation"* (Graingr and Allison, 2015).

3.2 Justification of Medical Exposure in CT

Most physicians who are working with medico-radiology have little understanding about the effects of ionizing radiation and/or its perspective radiation safety. Any philosophy of radiation protection should make allowance for clinical practice of the real world, unless it is convicted to failure (IAEA, 2001). Therefore, physicians should be equipped with concrete information about the how to protect individual patient from the deleterious effects of radiation during clinical diagnosis. Because physicians (radiologists and referring clinician) have indispensable roles to ensure justified exposure of patients. ICRP philosophy (ICRP, 2007) stand on the justified use of radiation for clinical diagnosis.

European commission council directive 2013/59/Euroatom defined justification as '*decisions introducing a practice shall be justified in the sense that such decisions shall be taken with the intent to ensure that the individual or societal benefit resulting from the practice outweighs the health detriment that it may cause*' (European Commission, 2014a). This indicates that the referring medical professional required to ensure justified diagnosis of patient. Particularly, the radiographers should use every technical parameters to reduce each patient dose as much as reasonably achievable (ICRP, 2000b). According to ICRP report, patient dose can be reduced by greater than 50 percent by using DRLs properly, adapting excellent culture of quality control program

and appropriate selection of technical parameters (ICRP, 2000b). There are two principal facts for defining patient dose in medical imaging. These are (1) to establish and provide standards of good practices and (2) to evaluate harms incurred from medical diagnosis that is aiming for justification of the practice and assessing the risks (IAEA, 2001, 2007).

The justification principles stated that any decision that alters the radiation exposure situation should do more good than harm (ICRP, 2007). The three levels of ICRP justifications (ICRP, 2015) for the use of radiation in medicine are: (1) the use of radiation in medicine is acceptable when it results in more good than harm to the patient. It is now taken for granted that the use of X-rays in medicine is justified., (2) a specified procedure with a specified objective is defined and justified (example, a cone beam computed tomography, CBCT, examination for patients showing relevant symptoms, or a group of individuals at risk of a condition that can be detected and treated), and (3) the use of radiation in an individual patient be justified (example, the particular CBCT application should be judged to do more good than harm to the individual patient). Principle (1) clarified that CT examination is justified to apply to individuals, however, principles (2) and (3) are source related and apply in all exposure situation to individuals (ICRP, 2007).

CT examination request must be ordered by qualified experts (medical or dental practitioners). The radiologists must be properly educated and experienced in CT technology and radiation protection as well as have sufficient understanding regarding substitute imaging methods that can deliver comparable clinical information with less radiation exposure of patients (ICRP, 2000b, 2015) (i.e. weather the required information can be other techniques like ultrasound, MRI, conventional radiography, etc.). Under justification resource and cost availability should be considered in addition to the potential clinical benefit of radiation exposure to the patient. Therefore, justification stands for a collective accountability among clinician and radiologists. Clinical guide-lines tailored to proper and acceptable CT examinations should be available to clinician and radiologists. This avoids repetition of CT examinations without clinical justification. Pregnant women should not do CT scanning, even if abdomen and pelvis examinations are wisely justified. If done, the fetus absorbed dose should not be greater than 40 mGy (ICRP, 2000b).

3.3 Optimization of Protection in CT

Based on the radiological protection principle of optimization, limiting the radiation dose received from X-ray medical imaging that is highly required without jeopardizing the importance of diagnosis. Once CT examinations have been justified, the principal responsibility for ensuring protection of medical workers and patients should be done carefully, successfully with excellent systems which can be collectively called as the optimization principle (ICRP, 2015). The optimization principle is "the likelihood of incurring exposures, the number of people exposed, and the magnitude of individual doses should be kept as low as reasonably achievable (ALARA), taking societal and economic influences into account" (ICRP, 2007). The first duty for optimization of CT lies with the CT facilities. That means the CT facilities should assure that the examinations are conducted with the lowermost possible patient doses without jeopardizing the required image quality for the medical diagnosis (ICRP, 2000b). Because, the main objective of optimization is a revolutionary iterative practice intended at avoiding or minimizing upcoming exposures to achieve the best level of patient protection (ICRP, 2007); and to deliver adequate investigative information (ICRP, 2000b). Clinician explain the target volume of the patient body to be examined and the scope of the desired examination.

Normally, modern CT is provided with customized pre-set parameters for exposure that enables to order for each body parts of the patient. This integrated CT setting parameters consequent for the selection of unreasonably high exposure factors (ICRP, 2000b). Nevertheless, the radiologist, technologist and operator should be trained and had capability and responsibility for reducing the dose of radiation commensurate with sufficient quality of image. Because the operators have good control over image acquisition factors such as applied voltage, tube current time product (mAs), time of gantry rotation, slice thickness (collimation) and patient bed feed per 360^{o} (ICRP, 2000b).

Recent CT technologies are armed by sophisticated methods towards reducing dose outputs of CT-tube such as organ-based tube current modulation, anatomical X-ray tube modulation, adaptation of the X-ray tube voltage to the patient's anatomy and the planed examination type, dynamically adjustable pre-patient controllers, Electrocardiogram (ECG)-controlled tube current modulation and iterative reconstruction (MF. Reiser et al, 2012). The development of MDCT starting from 4-

slice to the modern 512-slices brought great enhancement to its technical advancement irrespective of considering their greater potential of high dose burden to the patient. Clinical experience strongly responded that advancement of detector rows quantity in MDCT cannot be a guaranty to better quality of clinical performance. Great work is waiting for the mechanism to reduce radiation dose parallel to its advancement.

3.3.1 Anatomical X-ray tube modulation

The X-ray tube current adaptation towards the patient's structure (body shape as well as size) is the best effective mechanism to minimize radiation dose. This is achieved by adjusting mAs settings to individual patient manually or automatically. Automatic exposure control is used to select anatomical tube current modulation (AXTM) automatically. This anatomical tube current modulation method adjusts the current yield in the z-direction towards retain sufficient dosage during examination of different body regions like chest and abdomen-pelvic. Angular tube current modulation is also conducted when asymmetric body regions having intensely changing X-ray attenuations are exposed, like shoulders and pelvis. Practical application of anatomical dose modulation can reduce radiation dose from 20 – 68% relying on the area of exposure without compromising the clinically intended image quality (MF. Reiser et al, 2012).

3.3.2 Organ-based tube current modulation

This can be a modified organ-based tube-current modulation (AXTM) that is carefully-chosen for specific organ dose reduction like female breast. With benefit controversy, breast shield made from bismuth is used for dose reduction purpose. The tube current can be reduced using this modulation for adjustable angular range. The estimated local dose reduction is from 20 – 35% to the breast or thyroid gland without jeopardizing image quality (MF. Reiser et al, 2012).

3.3.3 ECG- Controlled-Tube-Current Modulation

Electrocardiogram – controlled tube current modulation (ECG-CTCM) able to reduce radiation dose of the heart examined by ECG-gated helical studies (MF. Reiser et al, 2012). Based on the electrocardiogram of the patient, the current from the X-ray tube remains modulated in the course of data acquisition.

3.3.4 Adaptation of X-ray tube-voltage

Adaptation of X-ray tube-voltage (AXTV) applies principle of regulating voltages of the X-ray tube to reduce the radiation dose based on the individual patient size and the intended clinical image. This practice is mostly common in contrast-enhanced CT procedures like angiographies. Reducing the tube voltage, increases the contrast material (i.e. iodine)-to-noise ratio proceeding by equal radiation dose of angiography examinations because at lower kV, iodine contrast increases. Hence, by lowering the kV, it is possible to reduce patient dose for the required diagnostic contrast-to-noise ratio (Reiser M, et al, 2012).

In clinical practice, that is problematic towards choosing the correct kV-settings manually, correspondingly selecting the respective adapted mAs for an individual typical scan is also difficult. Hence, automatic tube voltage selection software has been developed recently. In advance of individual scan, the software analyzes attenuation of a subject's target volume through localizer scan (scout view, topogram) evaluation. Then, the software recommends optimized settings of kV and mAs based on the type examination and limitations of the system like maximum system load and availability of maximum tube current (Reiser M, et al, 2012).

3.3.5 Iterative-Reconstruction

The primary goal of reintroducing iterative reconstruction to CT to advance quality of image, lower image noise and improve image resolution (MF. Reiser et al, 2012). Iterative loop has been familiarized in the image reconstruction process of iterative reconstruction. The fundamental circumstances of continuing iterative process of optimization protection involves: (1) assessment of the exposure situation; (2) choosing a proper reference level or constraint values; (3) differentiating the possible protection alternatives; (4) choosing the best alternatives from the fundamental situations; and (5) application of the nominated alternative.

3.4 Factors Influencing Optimization of Patient Dose in CT

Dose reduction technologies have been influenced by the performance of equipment and operator capability. Justified CT examination must be conducted with great attention of optimal selection of the acquisition parameters that requiring critical review of patient history and referred clinical information before imaging. Anyhow, all dose reduction strategies need proper applications of recommended dose

minimization technologies and strategies (McCollough et al, 2009; Thakur, Mclaughlin, & Mayo, 2013).

3.4.1 Operator Vital Factors

The scan parameters in different computed tomography scanner is adjusted differently depending on their manufacturers. Hence, facility owners should train their staff to have appropriate understanding of their CT scanner capabilities and function in practice. Generally, image quality and the corresponding patient dose in computed tomography rely on the choice of CT parameters tailor to individual patient anatomy and diagnostic information essentially required which are used to conduct CT examinations (Alsafi, 2016; ARPANSA, 2008). Standardized CT parameters are required for proper calculation and comparison of radiation doses (Øberg & Andersen, 2011). The significant parameters that determine the amount of dose to the patient in the control of the CT operator are tube voltage (kVp), tube current (mA), rotation time, beam collimation (image thickness), pitch (ICRP, 2000b), patient thickness (Abdulkadir M, 2015), and automatic exposure control (AEC). These parameters have been highlighted as follows.

3.4.1.1.Tube current (mA)-Time Product (mAs)

Tube current time product (mAs) is the product of tube current in mA by the scan time in seconds (IAEA, 2007). Tube current (mA) determines the proportional quantity of photons produced from X-ray tube per second. The mA determine the amounts of radiation dose (ICRP, 2000b). The quantity of dose delivered to an individual depends on size and target volume (ICRP, 2000b). Keeping all other CT parameters are constant, patient dose and applied mAs are directly proportional and this parameter impacts the noise in the image quality considerably (ICRP, 2015). Decreasing milli-Ampere second (mAs) decreases dose to the patient meaningfully and prolongs CT tube life. The mA regulates the quantity of photons per unit seconds proportionally, while mAs signifies the quantity of photons per fixed period of exposure. Therefore, dose rate related with mA while radiation dose related with mAs. In a series of CT generations, the main focus is to reduce scanning time. To get CT image in fractions of seconds requires the intensity of X-ray exposure should be very high. To obtain greater X-ray intensity, the time of exposure should be very short. The higher the radiation output in CT, the high quality of image is obtained such that CT X-ray tube is planned towards delivering excellent output of radiation with heat capacity and

dissipation (ICRP, 2000b, 2015, 2017b). For a fixed mAs value, lowering irradiation time required proportional increase of the current. Lowering of mA lacking parallel rise in irradiation period results towards increasing noise; this consequently degrade the image quality. Hence, there should have always a balance between dose delivered to patient and image quality (ICRP, 2017b).

Currently, manufacturers of all types of CT scanners comprises automatic tube current modulation (ATCM), which minimizes a tube current. In addition, ATC regulates the quantity of applied radiation doses. The patient radiation dose can be reduced using tube current modulation by 30% – 40% per scan (ICRP, 2017b). The mA can be regulated for Z-direction and x-y-direction scan lengths of the body when the X-ray tube rotates about the patient. Nevertheless, these systems are implemented in different ways by different CT manufacturers. Certain manufacturers like general electric and Toshiba apply a measure of image quality depend on the noise level in the image. The mA increases with patient size proportionately in that models. Other systems (like Siemens and Philips) uses comparisons using reference mAs or image. Thus tolerating high image noises resulting from large patient. The higher noise is tolerated to images from large patient which has improved organ departure as a result of interposed fatty tissue (ICRP, 2017b).

3.4.1.2.Tube Voltage (kVp)

The quality of radiation stands for the penetration power of radiation produced from an X-ray tube. It is commonly illustrated using applied voltage and beam filtration (ICRP, 1982a). The factors affecting radiation quality are peak voltage applied to tube, filtration, and high voltage waveform. The consequence of high voltage applied to the X-ray tube resulted high penetration of radiation beam and reached easily at the image receptor. Because voltage defines the energy of the electrons and consequently the incident X-ray's energy distribution (Øberg & Andersen, 2011). Nevertheless, the contrast between soft tissue and bone will be reduced at higher voltage (ICRP, 1982a).

The CT parameters that regulate X-ray kVp settings (energy spectrum) on the control console must retains as low as reasonably achievable without jeopardizing the image quality with sufficient clinical information. Applied voltage and tube current are determining the patient's overall radiation dose which are user-selectable variables on the control console. The influence of applied voltage settings on radiation dose and quality of image is highly problematic. The higher energy of radiation, the lesser

interaction of radiation with tissue. This results bad tissues contrast. However, higher energy photons penetrate the tissue easily to reach the detector to form image. The anatomy of patient affects the settings of mAs and kVp, regardless of usage of contrast medium, and the design factors of detectors, frame rates and systems of filters (ICRP, 2015).

The majority of CT scan uses 120 kV tube potential for many years (ICRP, 2000b) but can vary from 80 kVp – 140 kVp. Instead of high mA values, higher kV could be used to examine body parts of high absorption (Abdulkadir M, 2015). As explained above, higher kVp produced hard beam with the consequence of higher penetration of medium (MK Abdulkadir, 2015) with less surface radiation absorption (ICRP, 1982a). In principle, lower tube potential can deliver good quality of image and then resulting lower patient dose (ICRP, 2017b).

3.4.1.3. Scan length

The scan length (in mm) regulates the volume of radiation exposed patient (ICRP, 2000b). It remains directly proportional to a radiation exposure. Scan length determining the scan width of the patient body in the z-axis (Alsafi, 2016). Scan length should cover the diagnostically important parts of the patient body such that increasing the total scan length will irradiate a large part of the patient to radiation (Øberg & Andersen, 2011). It should be adjusted based on ALARA principle at the lowest possible value without compromising the desire of clinical request to be responded. Unlikely, through the introduction of fast CT machines, coverage of the scan length increases to examine several body parts at a time either separately or completely, such as rapid examination of patients with massive trauma of the thorax+abdomen+pelvis or head-to-pelvis examinations (ICRP, 2017b). This high scan length will increases the dose of radiation to the patient, hence, this high scan length will be the major factor for the development higher value of DRL (Abdulkadir M, 2015). Therefore, it is required to aware referring medical professionals towards dose concerns of repetitive studies, wrong exam request of anatomy or non-clinical essential exam request (Alsafi, 2016; ICRP, 2000b, 2017b).

3.4.1.4.Slice thickness

The controlling mechanisms of CT scanner parameters are dependent on the manufacturers and models. For example, certain CT scanners selects thinner images can consequence in noisier images. Whereas, the remaining CT scans retain the thinner

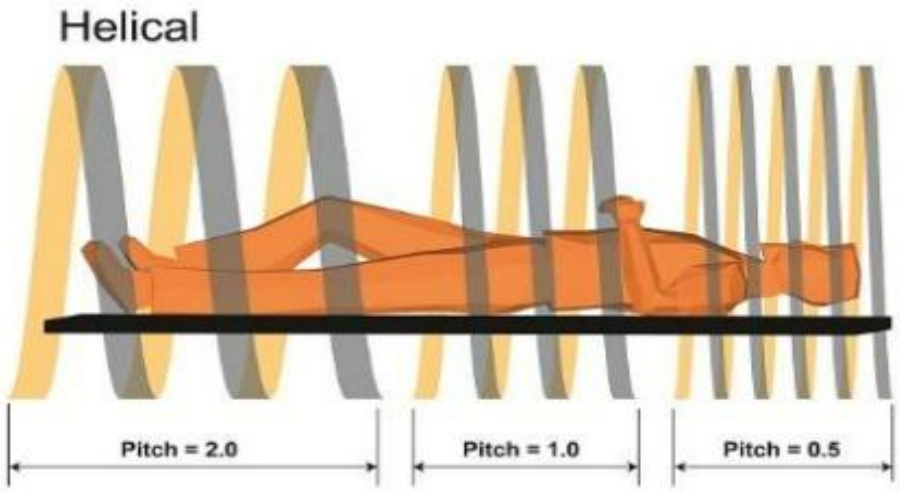
image quality through increasing the mA and kVp. In practice, the demand for high spatial resolution in CT examination contribute to high patient dose. This lead the selection of thinner slices (ICRP, 2017b). The thinner the slices, the higher the dose to the patient with greater spatial resolution (Abdulkadir M, 2015). However, thicker slices are less noisy (Alsafi, 2016) than thinner slices. High resolution computed tomography needs thin slices characteristically of 1 or 2mm that is lonely likely by increasing the mA (ICRP, 2017b).

3.4.1.5.Pitch

Pitch factor stands for the fraction of table feed (or table increment or continuous table advance), measured in mm, per 360^{o} tube rotation of the CT in the z-direction between consecutive scans to the total beam width (nominal slice width or nominal width) (American Association of Physicists in Medicine, 2008, 2010; IAEA, 2007). Pitch (p) calculates the longitudinal gap (p>1), overlap (p<1), or contiguity (p=1) between consecutive collections CT images (see Figure 3-1).

Figure III-1

The effect of pitch on an exposed area of patient (Enriquez, 2015)



In helical scan, patient radiation dose and pitch factor are inversely proportional to each other. Moreover, assuming mAs per rotation is kept constant (several general electric and Toshiba machines), such that the consequence of pitch increment can minimize the radiation dose while reducing of the pitch can increase the radiation dose (ICRP, 2017b). For example, supposing all factors except pitch are kept constant, pitch factor increased by two-fold will results a 50% reduction in the patient dose (Alsafi, 2016; ICRP, 2000b). If the pitch is increased from 1 to 1.375, the patient dose will be

reduced by 27% (Alsafi, 2016). Several Siemens and Philips model manufacturers retain the same radiation dose by regulating the mA as the pitch factor is altered (ICRP, 2015, 2017a).

3.4.1.6.Full Scan Time

Currently, modern CT gantry rotation time is capable to decreased up to 0.4 seconds and consequently an increase of the noise and absorbed dose reduction (Øberg & Andersen, 2011). Applying strict regulation of each CT irradiation, optimization of scan time is mandatory. Because the amount of dwell of time and patient dose from each exposure can be restricted. The aim of radiologist and operators should be to select a scan time as short as possible without jeopardizing the image quality, especially for chest and abdominal scan such that heart gesticulation and peristalsis may deteriorate image quality. By controlling the total scan time, patient dose can be controlled for each examination. Scan time, slice thickness and number of projection have proportional relationship to patient dose, such that short total scan time, large slice thickness and decreased number projection deliver relatively reduced amount of radiation dose to the patient (ICRP, 2015, 2017b; Abdulkadir M, 2015).

3.4.1.7.Automatic Exposure Control Techniques

Automatic exposure control arrangements adjust the exposure parameters to acquire a preferred quality of images and adapt the dosage towards the particular tissues or organs of the patient body due to different thickness of the anatomy of the patient. It has an electronic sensor that identifies the amount of signals manufactured at the detector, additionally adapts the generator (X-ray) to modulates (decrease or increase) the mA in real-time, mostly applied voltage, in relation to attenuation of a patient (thin or thicker body Sze) in a certain exposure direction (both in the angular and longitudinal or z-direction). Therefore, each projection can produce regular image quality according to AEC modulated exposure factors. And hence, its variance is dependent on patient thickness and applied as a checkup loop to regulate the X-ray source centered on reaction from the detector. This system reported 20% - 40% reduction of patient dose in (Alsafi, 2016; ICRP, 2015; Øberg & Andersen, 2011). The type of AEC influences patient dose reduction and overview of related vendors with their AEC is given in Table 3-5.

Table III-5

AEC and its Important Vendors (Øberg & Andersen, 2011)

Vendor	Automatic exposure control		
	Slice plane	Longitudinal plane	Combined
Philips	DOM		Z-DOM
Toshiba	3D	Real E.C.	Sure Exposure
GE	Smart scan	Auto mA	Smart mA
Siemens	Care Dose		Care Dose 4D

In the case of AEC, fatty patient exposed to higher radiation compared with thinner patient but their image quality is the same. This implies that AEC deliver patient specific exposure to the patient. The AEC angular and z-direction alterations of the mA give rise towards dose variation on the patient. For example, assume the operator designated quality of image factors are constant, the lateral-direction dose is greater than AP-direction due to angular variation. On the other hand, the z-direction AEC regulates the value of tube current in the direction of superior-inferior causing a larger abdomen and pelvic doses compared with chest (ICRP, 2015). Generally, CT systems with AEC software has excellent ability of maximization of patient protection than that misses the software.

3.4.2 Equipment Vital Factors

3.4.2.1 Detectors

The whole digital detectors arrangement used in CT have a high effective dynamic ranges setting. Quantum detection efficiency and geometric efficiency are the two dose-relevant characteristics of a detector (Abdulkadir M, 2015). Detectors uses direct and indirect translation of the incoming X-ray photons energy to produce an output electrical signals (Cember H and Johnson E, 2009; ICRP, 2017b) which interpret the incident signal on it into an image. The detector geometric factors, motion of the X-ray tube, and motion of detectors influence the distribution of the dose (ICRP, 1982a). Detector quantum efficiency quantifies the image quality generated by the image receptor originating from particular amount of radiation dosage to the detectors (ICRP, 2015). Detectors with high quantum efficiency, rapid response and low afterglow of the scintillator deliver high quality of image and hence allow significantly a smaller amount radiation dose to the patient (ICRP, 2015, 2017b; Abdulkadir M, 2015). This implies the radiation dose recognized by detectors and the produced quality of image have direct relationship. The consequence of high doses deliver excellent image quality

in the absence of saturation seen in film based imaging methods, and hence directly burdened on the patient dose that need optimization (ICRP, 2017b).

3.4.2.2 Collimator

Perfect CT must expose the desired target size beside z-direction using a uniform radiation dosage which must reduce quickly outer a target size (ICRP, 2015). The slice thickness, detector configuration, and beam collimation are highly integrated in multi-detector CT (MDCT) systems (Alsafi, 2016). In the case movable detector system, some amount photons may not get the detector while X-ray beam and detector dimensions are generally congruent. Therefore, careful attention must be given during beam collimation. Because any production of photons outside a detector contribute needless patient dose. The field of view (FOV) is defined by the distance between X-ray tube and the image receptor (detector). FOV must be properly regulated according to the intended anatomy of the patient. The scatter noise in the projection data and area of irradiated fields are linearly related; that means as area of irradiation increases, the scatter noise also increases. Generally, the main objectives of tight beam collimation are to lower the dose, to lower scatter radiation, to advance image quality and to keep unwanted irradiation of the body parts.

If the primary beam is poorly collimated outside patient target volume, then it may increase the patient and occupational dose (ICRP, 2015). This implies that tightly collimated X-ray beam to the scan FOV is mandatory to exclude any adjacent sensitive organs in order to answer the clinical enquiry at hand. When primary beam incident covers outside the detector (image receptor), over beaming is occurred. Hence, some beam is not functioning for imaging. Rather, it is exposing the patient unwantedly. Pre patient collimators are located between the X-ray tube and the patient to define the collimated beam width and therefore reduce clinical radiation dose to the patient (Alsafi, 2016; Abdulkadir M, 2015). This method lowers over-beaming via gauging beam position every specific time (milliseconds) as well as relocating the opening of the beam as required. The focal spot tracking systems allows a narrower dose profile than non-focal spot tracking X-ray tube CT systems (Alsafi, 2016). Post-patient collimators are positioned between the patient and the detector commonly facing the detector to reject scatter radiation that improves the quality of the image however expenses the efficiency of the dose (Abdulkadir M, 2015).

3.4.2.3 Beam Shaping Filters

There are two types of filters called inherent and added filters, these are collectively called total filters. The main purposes of filters is to attenuate lower energy X-ray beam and don't contribute in imaging that mostly be absorbed in the patient body (Alsafi, 2016). Hence, the usage of a filter resulting in the reduction in dose rate during a tube is functioning at a particular voltage and current. Increasing filtration caused in a significant reduction in dose (ICRP, 1982a) and reduce image noise.

Use of bowtie filters (beam shaping filters) attenuate soft X-ray beam and hardens, reduce the primary to scatter ratio and reduce beam fluencies heterogeneity at the image receptor. Beam shaping filters reduce the patient dose by reducing scatter contributions in CT imaging (ICRP, 2015). Different studies revealed that the use of bowtie filters reduces the dose of radiation by 50% compared with conventional filters (Alsafi, 2016). Filters can be also categorized in to smooth and sharp, such that smooth filters reduce noise while sharp filters increase image noise. The appropriateness of filter selection, which influences the role of TCM (tube current modulation), be governed by the imaging task. On some CT model, the image noise can be increased according to the choice of the sharper filter. This will affect the TCM to increase the mA. This resulting an increase of the quantity of dose to retain similar level of noise. Other models of CT, the image appearance drive an alteration. However, the quantity of radiation dose can be stayed comparatively unaffected (ICRP, 2015). Software noise reduction filter has better result in high contrast exam like chest CT study (Alsafi, 2016).

3.5 DRLs in the Global Background

1996 was the opening year for the term DRLs by the ICRP by its publication 73. The idea of DRL gradually developed into practical guidance in 2001. Publication 103 of ICRP in 2007 stated that "one of the principles of optimization of protection in medical exposures is implemented through the use of diagnostic reference levels (DRLs). The DRL has proven to be an effective tool that aids in optimization of protection in the medical exposure of patients for diagnostic and interventional procedures". The concept of DRLs values are employed only for medical exposure, not for occupational and public exposure. (ICRP, 2007, 2016; McCollough, 2010). DRLs cannot be employed to radiotherapy. DRL is used as an effective tool to optimize the patient dose in various imaging modalities. The medical aim of diagnostic radiology is getting

outstanding quality of images and sufficient diagnostic information. In addition to clinical purpose exposure, protecting the patient from unwanted radiation is highly significant. Hence, professional should use DRLs properly to optimize their patient dose as low as reasonably achievable.

> *According to IAEA new BSS "the registrant and licensee must confirm developed DRLs for the accepted radiological examinations and review should be conducted to determine the satisfactory optimization of protection and safety for patient medical dose so that the dosimetry must be documented. These protection and safety of patient can be sufficiently upgraded through possessing the dose to the patient below the DRLs in commensurate to medically adequate quality of image" (IAEA, 2011).*

DRL is a tool that help to regulate abnormally high or low levels of patient medical dose. When the patient dose significantly above or below the working DRLs, review of the DRL is needed to conducted. Because significantly high or low patient dose will not provide useful medical information or benefit to the patient. However, radiation exposure of the patient is always for the direct use of the patient. Therefore, the concerned body need to confirm the adequate optimization of patient protection by taking the required corrective actions (ICRP, 2007). Using DRLs as regulatory determination, or commercial aim or dose limit or dose constraint is clearly forbidden (ICRP, 2017a; McCollough, 2010; Paulo, 2015).

The principles of radiation protection structure across the world is based on the ICRP recommendations (Paulo, 2015) which are applied in all exposure situation including medical exposure (IAEA, 2011). The ICRP principles of radiation protection are justification, dose limitation and dose optimization (IAEA, 2011; ICRP, 2007; Paulo, 2015). Conducting unjustified medical diagnosis to the patient is extremely prohibited and the dose should not have exceeded the specified dose references stated by the nation's regulatory authority. Therefore, responsible bodies should select the values of reference levels to optimize and protect patients in case of planed exposure situations (ICRP, 2007).

The principles of optimization of protection required that individual radiation doses must bas ALARA in any exposure situation by considering the economic and social factors of the society. ICRP publication 103 (2007) stresses the key role of optimization principles in which it is practical in all exposure situation in the same manner. Dose restriction stated by a particular nation are applied towards the reference person. These restrictions are called dose constraints and reference levels for planned

exposure, and existing and emergency situation exposure, respectively. The incurring medical dosage greater than the reference levels must be automatically disallowed at the start of preparation of medical diagnosis. When the medical dose is frequently exceeding the dose restriction, investigation must be conducted at that medical facility and the reasons should be justified clearly (ICRP, 2007).

The principles of dose limitation cannot be applied to medical diagnosis. Because, medical diagnosis is conducted intentionally for the benefit of the patient. The principle of dose limitation works only for occupational and public exposure in diagnostic and interventional planned exposure situation. This dose restriction should not surpass from the limit given by the nation or in general by ICRP. The dose limitation refers individual total dose incurred from controlled radiation sources in planned exposure. Planned exposure situations include the use of X-rays in medical diagnosis.

DRLs does not has direct linkage to dose constraints and dose limits proposed by the nation. Because the DRLs numerical values are applied only for medical imaging. This DRLs values are computed based on the basis of a percentile point (in this case 75^{th} percentile) on the observed dose distribution to the reference patient (ICRP, 2007). The numerical values of the DRLs must be developed by concerned scientific bodies. Review of the DRLs should be conducted periodically based on the professional recommendation (mostly it is 3 to 5 years). The review should consider the DRLs values stability and continuous alteration of the mean/median dose distributions. Each country must develop its own DRLs values based on easily quantified patient dose related metrics.

3.6 Application of DRLs in CT

DRLs provide synchronized evidences about the dose received by the subject. This helps the medical scientists for their prominent decision about the patient diagnosis. DRLs also gives signals to unusually high patient at the health facilities. This drives encourage immediate investigation (Paulo, 2015). Generally, the inescapability of DRL for a country in medical diagnosis is to reduce the limitation of dose dispersion, to harmonize and expand good practice, to narrow large dispersion of doses, and to create systematic supervision for unwanted radiological doses. The main objective of DRL is optimization of the medical dose of patient not minimization of the doses.

When the change in equipment at a health facility, its present performance must be investigated. This is done to confirm its compliance to the DRLs protocols. If significantly higher or lower dose report is found, its reason will be assessed to optimize the dose. Following these all processes, the facility is reevaluated to confirm its typical dose is below the relevant DRLs. Japan recommend annual review of protocols and practices (Japan, 2015). Because protocols and practices should not periodically change intentionally. New CT equipment required initial assessments for the established protocols before operation on patient. Even reassessment will be needed within 3 -6 months of operation (Japan, 2015). These all processes will be going without neglecting the adequate image quality required for the intended diagnostic objectives. Even if the examination is justified, essential medical information may not be obtained from patient exposure for extremely low doses. Because, extremely low radiation doses result unsatisfactory image quality. This implies, the patient can be ordered for repeated exposure.

Dose optimization process started with comparing health facility's dose with the relevant DRLs. But, dosimeter (that is used to monitor the absorbed dose) is required at a facility for dose comparison in medical settings. Practically, obtaining dosimeter for medical setting is highly difficult. A promising solution used for this obstacle is the use of substitute dose values such as dose values computed using conventional software or dose values displayed on equipment's control console. Furthermore, mechanisms of obtaining dose values using dosimeters or phantoms based surveys can be established. Then, the dose values can be used to develop local, regional or national DRLs for the selected examination protocols for the CT clinical application. The DRLs obtained based on standard phantoms can also compared with the international records for generalization.

3.7 Determining DRLs

Appropriate DRLs development required gathering of sufficient amount of patient dose data (20 – 50 data from each facility) (ICRP, 2017b), then the researcher decided how the local, regional or national DRL values were established. The mean or median patient dose data distribution for each kinds of examination uses to drive the DRL quantities for each facility.

Outliers and data with gross errors contribute nonsensical quantities for the DRLs values because they have significant effect on the mean dose distribution (ICRP,

2017b). However, rare outliers have minimal consequence on the median dose distribution. These outliers and data with gross errors should be excluded from main data analysis. One of the exclusion method is considering rejecting the 5 percent highest and lowest tails of the data distributions (ICRP, 2017b) with greater care of their effect. The sources of outliers and data with gross errors are wrong data entry or unusually large patients that delivering unusually high or low values.

This unusually high value data is common in the collection large number of data and methods should be proposed to exclude. Facilities with this uncommon data values required future focus for optimization. Most of the time, the typical DRL values distributions obtained from multiple health centers are nearly log-normal. The form of the skewed pattern of the distribution of a DRL quantity has been repeated many times in surveys throughout the world, from many different types of examinations and for many DRL quantities as there are inevitably always a few facilities where optimization has not been fully implemented.

The DRL development required large amount of sample of health centers at the regional or national level. The DRLs values are defined at the third quartile of the mean/median dose distribution of the data set. This third quartile has no scientific basis chosen arbitrarily. It is used as a preliminary divider among adequate and extreme values. Nevertheless, the 3rd quartile commonly lies well below the greater tail of the data scatter, and used as a valuable indicator for centers classification whose outcomes lie to the higher tail of the distribution. Setting the DRLs quantities at the 3rd quartile of the distribution is highly reasonable (ICRP, 2017b).

CT modality is comparatively delivering higher dose to the patient. Hence, it should attain greater priority for optimization. The mean, median and DRL values can be used in assisting patient dose and image quality optimization. Since it possibly gives an improved director for judging good practice. Bur, after developing DRLs values, most country abandon updating their DRL. It is not static. It requires periodic review. DRLs is used as a consultancy tool for investigation of the facility. The investigation descriptor values are seen when the site mean/median values are higher than the national DRL quantities. Improvements are also required when the site median values of the DRL values are below the national DRL quantities. Using of median values for DRL establishment has be recognized by the ICRP (ICRP, 2017b).

3.8 Special Emphasis on DRLs in CT

Technically, different types of CT scanner (both single and multi-detectors CT-scanners) required different DRL, because patient dose levels delivered from different generations of CT scanners will be different. But, developing different DRL for each of the CT generation can make more problematical for its practicability. This implies that DRL commonly prepared for all equipment generations by considering its frequent update due to the potential effect of equipment and knowledge advancement on patient doses (European Commission, 2018). The medical and dental exposure of CT procedures delivered about 50% of collective effective dose, because CT procedures produce high dose compared with other diagnostic imaging techniques. Since CT introduction to medical diagnosis application worldwide, there existed a dramatic increase of CT examinations frequency annually (ICRP, 2017b; UNSCEAR, 2008).

3.9 Image Quality and CT

3.9.1 Introduction to Clinical Image

The medical application of CT emphases on patient dose and protocol assessments. This leads image quality benchmarks establishment (IAEA, 2001). In the medical application of CT, the central focus of medical professionals should be realizing obedience of patient dose optimization with the DRLs for the particular exam. Practically, the typical exam doses in a facility is expected to be below the relevant DRLs. Nevertheless, the lowest possible dose of the patient does not signify achievements of excellent clinical image quality needed for the intended diagnostic information. Rather, the unsatisfactory image quality will mislead the medical professional clinical decision for the patient. During optimization of patient protection through using DRLs, the quality of image needed for the clinical purpose must not be compromised. Since, the main objective of medical professionals is obtaining acceptable quality of image that provide all the required diagnostic information for the intended medical aim. DRLs is not a measure of image quality in patient diagnosis. This implies that patient dosages above DRLs or below DRLs does not express poor quality of images. In other words, adequate image quality does not relate with achieving the DRLs or unacceptable image quality has no relationship with patient doses below DRLs. Rarely, bad quality of image can deliver satisfactory medical information. This indicates that DRL is developed by optimizing the performance of

procedures to obtain adequate image quality with minimized patient dose (ICRP, 2015).

Images obtained from either too low doses or too high doses provided unsatisfactory quality for the intended diagnostic information. Hence, the patient will be over exposed from repeated examination. Therefore, optimization of protection must equilibrate patient dose and image quality. However, the relationship between radiation dose and image quality are pintsized (IPSM, 1992). Good imaging performance of the facility should complement the clinical demands for the patient exam by keeping the principles of ALARA (European Commission, 1996). According to the indication of patient examination, DRLs setting from dose data collection must follow the daily quality control program (QAP) while deciding adequate quality of image (European Commission, 2018). In order to attain sufficient image quality, careful choice of technical factors is mandatory. The parameters monitor the dose and image quality. Additionally, consistent inspection of CT machine performance and image quality factors should include into the daily QAP (European Commission, 1996). Proper image quality factors per examination dose depends on contrast media (high and low contrast image resolution), the imaging systems performance (image homogeneity), professionals skills in imaging parameter choice, modulation transfer function, the system transfer factor as well as noise power spectra (Conference of radiation control program directors, 2015; ICRP, 2017b). Patient dose can be reduced by 30 – 40% per CT scan examination by using TCM. The appropriate level quality of image for CT examinations has been arguing amongst medical professionals. The several factors that donate image quality when protocols of imaging for new scanners set up must be consulted in detail. Those parameters are related to detectability of low contrast as well as special resolution of the displayed image (ICRP, 2017b).

3.9.2 CT Image

This review includes CT image quality on the common clinical basis. CT has direct digital image processing techniques since its application is mainly focused on medical diagnosis. Its application is increasing from time to time for most medical diagnosis modalities; and its significance is progressively more in medical care (Deserno TM, 2011). The principal objective of medical diagnosis is knowing the patient central reasons of illness. Thus, recent technology of imaging device like CT revolutionized the healthcare system through watching into the patient body with reduced risks (JD.

Enderle, Bronzino, 2012). The CT imaging techniques has four consecutive main categories. These are image acquisition, digitization, processing and display. In CT, the image acquisition device is its gantry (i.e. documentation camera) (Cierniak R, 2011) that uses the in-built transducer to transform the oncoming radiation through the patient body towards analogue signals (electronic current) before arriving the computer. Image digitization step transforms the analogue electronic current signals to 2-D digital signals (digital images) form using the analogue to digital converters (ADC). ADC bonds the image acquisition interface to the computer. The final output of digitization is a digital image that is tailored to be stored in the computer. The digital image formation processing is the task of image processing step that is completed through the computer itself. Array processer is a dedicated hardware constructed into the computer to accelerate its processing time. The end outcome of image acquisition, digitization and processing is the digital image. Digital images are the collection of different pixels. Pixel stands for picture and element. Each pixel in digital images are assigned individual brightness or pigment standards. Pixels be capable of proficiently processed, accurately assessed and readily accessible at all locations all together. These are done using protocols and proper communication networks like the digital imaging and communications in medicine (DICOM) protocols and picture archiving and communication system (PACS), correspondingly. Currently, the complete of digital image processing spectrums are appropriate in medicine according to digital imaging systems (Deserno TM, 2011).

The complexity of CT Image quality remains problematic towards well-defining accurately. The WHO defined quality assurance program (QAP) in diagnostic radiography as: *"An organized effort by the staff operation a facility to ensure that the diagnostic image produced are of a sufficiently high quality that they consistently provide adequate diagnostic information at the lowest possible cost and with the least possible exposure of the patient to radiation. Registrant and licensees must establish a comprehensive QAP for medical diagnosis with the participation of appropriate medical physicists, taking into account the principles established by the WHO" (IAEA, 2012).*

However, the unforgettable image quality factors in CT are radiation dose, image noise, resolution and quantum mottle, which are briefly explained in section 4.3. Image artifacts in CT also caused by mechanical misalignment. Patient motion, non-

uniformity of detector, beam hardening, aliasing and partial volume effects (Dendy & Heaton, 2001). To overcome all the affirmation causes of image quality deterioration in CT, adequate quality assurance program including image quality analysis should be implemented prospectively or retrospectively in a regular basis in any CT facility.

Image quality denotes a linear attenuation coefficients (μ) map of effective photons that pass through the patient's anatomy. μ denotes the quantification of X-ray photons interaction with matter (in this case the composition of patient's anatomy). That means, it signifies the exponential path change probability of photons through scatter or absorption. Therefore, the quantity of μ remains relied on density of the material and photons energy. The techniques of mathematical reconstruction are applied to approximate the effective (μ) value matrix that containing CT image from photon quantification (Reiser M, et al, 2012).

The medical question of image quality and radiation dose required intense optimization and balances between image quality and dose. The efficiently developed scanner design and correctly applied radiation dose efficiency parameter are capable to compare different scanner marks with respect to image quality and dose balances (Verdun et al., 2015). The balance of these factors is relied on the medical question and types of study. The capability of the operator significantly affects the performance of the scanner and image quality. The CT machine design and its scan settings are meaningfully affect the quality of image and radiation dose, as shown in Table 3-6.

Table III-6

Image quality and radiation dose can be affected by CT machine design and its corresponding scanning settings (Verdun et al., 2015)

CT Machine Design parameters	Exam protocol parameters
✓ Material of the detectors ✓ Configuration of the detectors ✓ Number of detector rows ✓ Rate of data acquisition ✓ Software corrections ✓ Filtration ✓ Focal spot size ✓ Geometry (i.e. focus-axis, focus-detector distances)	✓ Medical application ✓ Tube current, voltage, focal spot size ✓ Image reconstruction algorithm ✓ x-ray collimation width, detector acquisition width ✓ Reconstructed image slice thickness ✓ Pitch ✓ Interpolation algorithm

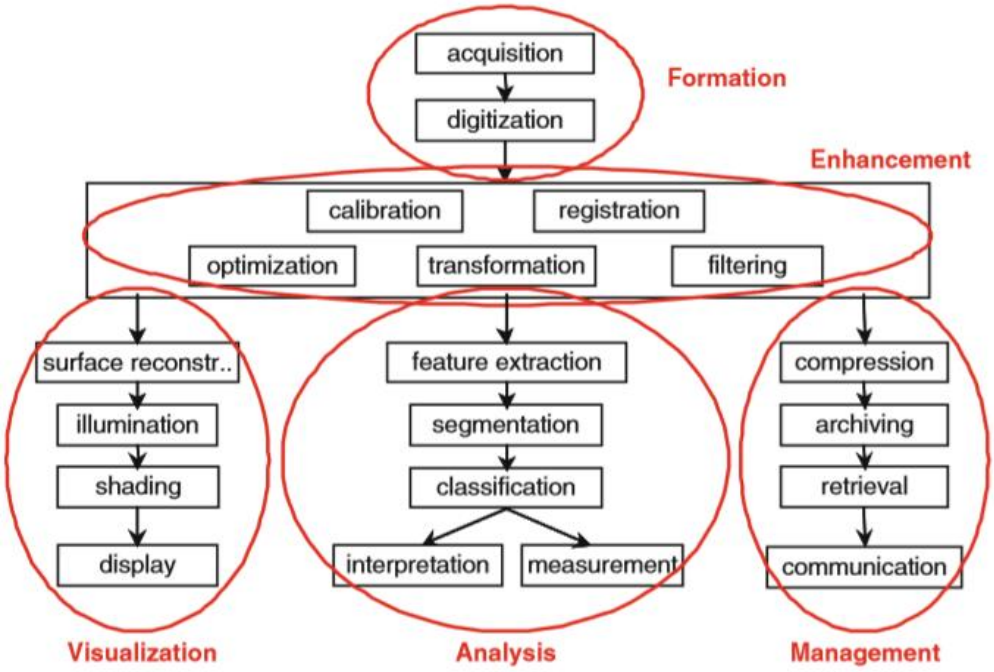
3.9.3 Image processing Steps

The digital image processing has four main parceled categories including image formation, visualization, analysis and management (Figure 3-2). The image enhancement algorithm is assigned as pre and post digital image processing in all other categories. Image formation encompasses image acquisition steps and the matrix of digital image realization. Image visualization clarifies to all kinds of manipulation of digital image matrix, bringing about the optimized image (Deserno TM, 2011).

Image analysis comprises the entire stages of digital image processing that are used for quantitative measurements and abstract interpretations of diagnostic images. These hierarchy of digital image processing need prerequisite understanding on the general condition of the image that should be incorporated into the algorithms on high level of abstraction. Hence, the digital image analysis process is highly specific as well as established algorithms able to transport often directly into other application domain. Image management includes the entire methods that deliver the effective storage, communication, transmission, archiving and access (retrieval) of image data. Hence, image management process includes all techniques of telemedicine (Deserno TM, 2011).

Figure III-2

Diagram of digital image processing hierarchy (Deserno TM, 2011)



3.9.4 The quality of the Clinical CT Images

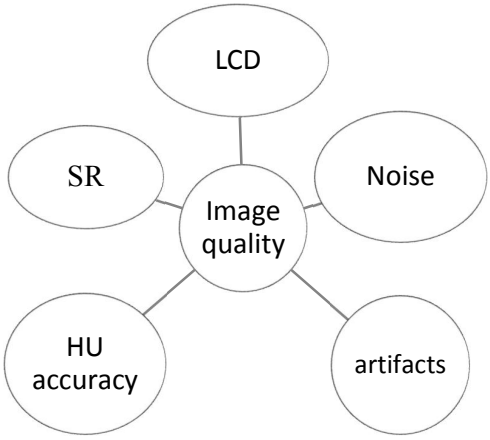
Understanding of image quality by image scientists required their capability of explaining vision, color, computational and behavioral sciences (Virtanen et al, 2020). The ideal science of CT image is related with the interaction of radiation with matter and the process of mathematical reconstructions applied to create clinical image from the projection of X-rays (Reiser et al, 2012). Image quality defines the total image fineness and its features qualitatively (Reiser et al, 2012). Clinical CT image characterizes the cross-sectional map of effective photons linear attenuation coefficients of the anatomy of the patient body (Hendee & Ritenour, 2002).

The main goal of clinical image is to deliver diagnostic information that enabling the (1) revealing patient injury or illness; (2) explaining the magnitude of injury or illness; (3) diagnosing the fundamental reasons of the injury or illness; (4) guiding illness or injury treatments; (5) controlling the efficiency of the treatment and its consequence (Hendee & Ritenour, 2002). The paramount purpose achievements of the medical image to clarify the presence or absence of sickness is defined through the quality of image.

The quality of image predicts the clarity of diagnostic information displayed on the images like the changes in the triggered through sickness on the patient's physiology, anatomy, and functional cavity. Because, image quality should deliver clear diagnostic information for the medical professionals. The image quality can not only depend on the performance of imaging system and image processing but also affected by the detectability of disease or injury by imaging system, correctness of the image, and area of the patient body scanned (Hendee & Ritenour, 2002). However, image quality criteria should be delineated by scientific professional to clearly understand the mechanisms of crosschecking the image quality created by the imaging system and image processing. According to image scientists, the main components of CT image quality produced are Hounsfield unit (HU) accuracy, low contrast detectability (LCD), noise, special resolution (SR) and artifacts (Hendee & Ritenour, 2002; Reiser et al, 2012), Figure 3-3.

Figure III-3

Influence to image quality (Hendee & Ritenour, 2002; Reiser et al, 2012)



Hounsfield Units (HU) Accuracy

The computer algorithm assigns CT-number (informally called Hounsfield Units) to each CT-pixel based on CT attenuation map. Hounsfield Units (HU) represents the measurements of transmitted X-rays through the patient's anatomy and uses as data to form an image. CT-pixel numbers are proportional to the difference in a mean X-ray attenuation of the tissue within the voxel and water (Brant W and Helms C, 2012) (Reiser M et al, 2012), for further information about HU, refer Table 3-7, and section 4.4.1. Mathematical expression of HU is given as follows (Reiser M et al, 2012):

$$HU_s = \frac{\mu_s}{\mu_w} x\ 1000 - 1000$$

where μ_s is the effective linear attenuation coefficient of scanned sample and μ_w is effective linear attenuation coefficient of water (μw = 0 HU).

Table III-7

The Hounsfield unit (HU) ranges for different materials and tissues (Brant W and Helms C, 2012)

Material	CT-number (HU)
Water	0
Air	-1000
Bone	400 -1000
Very dense bone (metal prosthesis and dental fillings)	3000 - 4000
Soft tissue	40 – 80
Fat	-100 to -60
Lung	-600 to -400

Spatial Resolution

Image spatial resolution can be expressed as the capability of producing images of small objects that have high contrast subject with excellent spatial information in the CT scanner. Spatial resolution mostly signifies to objects that have high contrast nature. Even though, image special resolution is allowed in 3-dimentional CT scan reconstruction, it is well-defined in the tomographic plane. Estimation of spatial resolution is done by quantifying using either a resolution patterns (high contrast test tools) through optical inspection or calculating modulation transfer function (MTF) or calculating MTF using proper test object (IAEA, 2012). MTF is resulting from image bread or wire. These wire or bread used to calculate the edge spread function or point spread function. MTF can be obtained using DICOM image transfer standard and functional software that can installed on individual computer. Frequency comparisons is required to understand the fall of MTF curve at 50% and 10% levels. When high contrast subjects are used, the subsequent value is line pairs per centimeter (lp/cm). When MTF is estimated the resulting quantity is cycle per centimeter (c/cm).

Low Contrast Detectability

Imaging contrast describe the capability of the imaging system towards distinguishing small variations between the target volume and its surrounding of the patient body (Hendee & Ritenour, 2002). Hence, image contrast is the variance in clarity between the target volume and its surrounds (Khaled ME. and Sandra AO, 2014). In clinical imaging, the image contrast is the result of sophisticated interactions amongst the anatomic and physiologic contributions of the target volume, the characteristic of the imaging system, the characteristics of the receptors used as well as the professional skills to extract an image. In clinical image contrast expression, greater differences in the brightness of variable tissue types (see Figure 3-4) is required for easy distinguishing of the tissues from each other (Khaled ME. and Sandra AO, 2014). The components of image contrast (Figure 3-5) are categorized into four influences, these are intrinsic contrast, imaging technique, contrast agent and receptor contrast.

Figure III-4

Images with different brightness such image (A) is with low contrast and image (D) is with high contrast (Khaled ME. and Sandra AO, 2014)

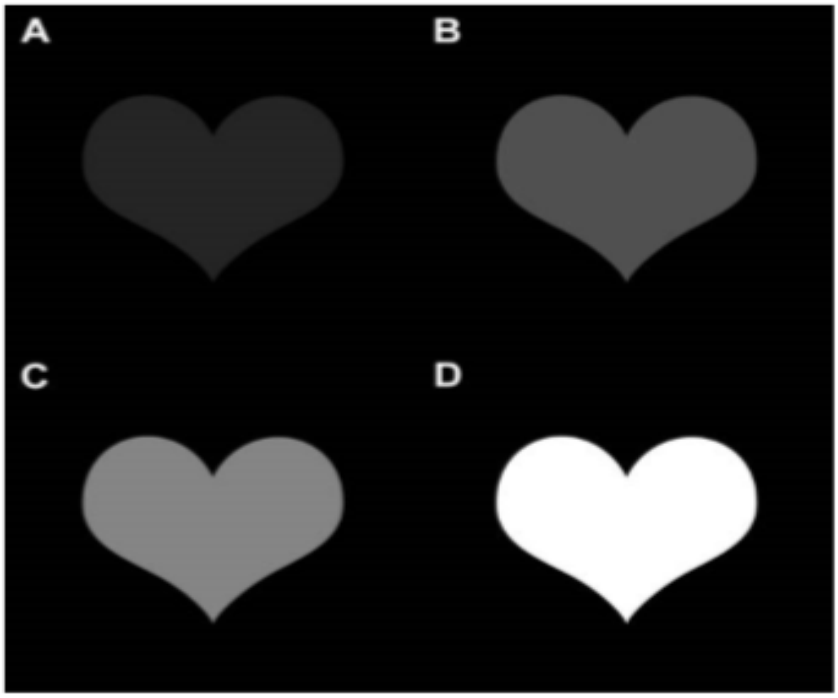


Figure III-5

Image contrast influencing factors (Hendee WR and Ritenour ER, 2002)

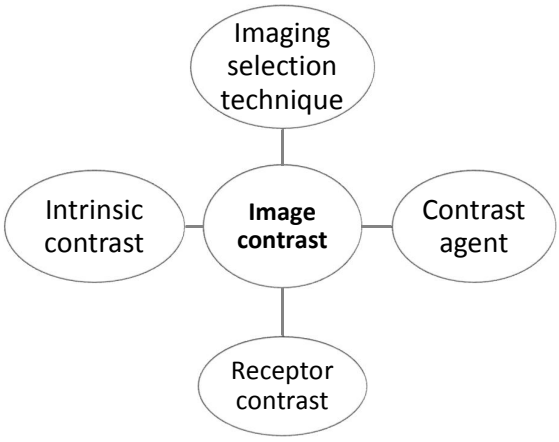


Image Artifacts

Every clinical images contain some amount of image artifacts. Image artifacts cannot represent diagnostically intended realistic tissue structure (Hendee & Ritenour, 2002). Artifacts create troubles of image understanding by misleading as well as providing

useless or mistaken diagnostic medical information to the image scientists or radiologists or physicists. Hence, image artifacts damage quality of image and vague image anatomy. In generally, the causes of artifacts are corresponding to operator skills, software in use, patient and the CT machine itself. Furthermore, it has several root sources like problems in installation process, servicing, calibration, and electronic components. Hence, greater attention should be given when CT machines installation, when assembling electronic components, servicing and calibrating the facility regularly. Careful installation room preparation must include its stability, level and capability to sustain the scanner weight. Electronic components assembly must be conducted carefully with electronics experienced engineer. After components assembly, critical assessment of acceptance test must be conducted at primary medical phase. Periodic and daily scanner calibration should be performed by concerned professionals. The facility quality assurance program should incorporate image quality display monitor. The constituents of image artifacts in CT are categorized into four major groups called shading, streaking, aliasing and rings (IAEA, 2012; Khaled ME. and Sandra AO, 2014), Figure 3-6.

Figure III-6

Factors in image artifacts

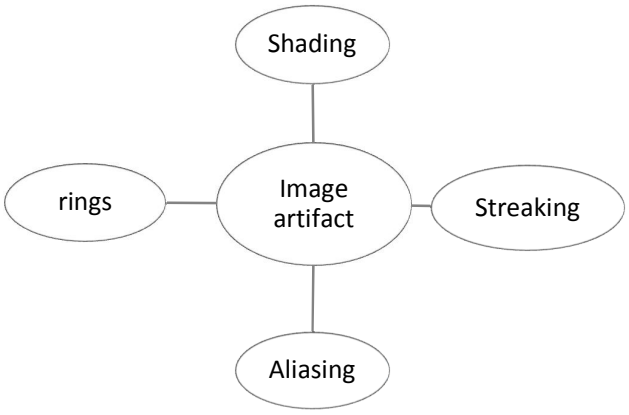


Image Noise

Reviewing the image science concerning entity of detectability in the existence of noise is highly sophisticated article (Reiser et al, 2012). Anyhow, all medical image consists of destructive information which cannot be applied for representing the current illness of the patient and for the required medical diagnosis. This unwanted

random variation of information in the ideal clinical image confuses the image professional's view. This inappropriate information in the image is called image noise (Hendee & Ritenour, 2002). Noise attempts towards minimizing the structures and objects brightness, particularly for low contrast structures and objects.

> *Because, once a uniform object is imaged on a CT scanner, studies of the CT values for individual pixels in a localized area shows that the CT numbers are not all the same, but fluctuate around a mean value. This random variation is called image noise and is due primarily to the statistical nature of X-ray production and interaction with matter (IAEA, 2012).*

Image noises have precise texture of appearance, which relies on different factors. The main causes of noise are post-patient detected X-ray intensity, radiation dose, scanner efficiency as well as processes of image reconstruction. Image noise and radiation dose have inverse relationship. It is highly associated with dose utilization. The factors influence the appearance as well as noise intensity retained in the image. Image noise obscures small low contrast objects. The main components of image noises are radiation noise, structure noise, quantum mottle and receptor noise, Figure 3-7.

Figure III-7

Image noise influencing parameters (Hendee & Ritenour, 2002)

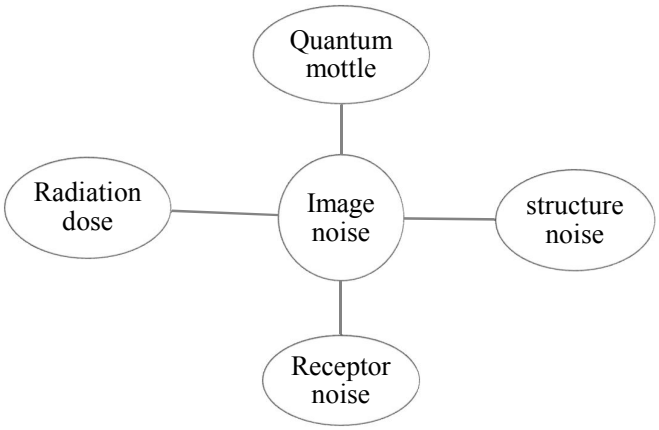


Image noise can be expressed in terms of the standard deviation of Hounsfield number (σ) within a region of interest (ROI). Noise comparison between different CT scanners with different contrast scales can be obtained using normalized standard deviation, S as follows (IAEA, 2012):

$$S = \frac{\sigma_{water}}{CT_{scale}}\ x\ 100\%$$

where σ_{water} is the standard deviation of the pixel values within an ROI, and given as:

$$CT_{scale} = CT_{water} - CT_{air}$$

where, CT_{water} and CT_{air} are the CT values for water and air, correspondingly, with CT_{water} value is zero and CT_{air} values is -1000 HU. Therefore, the CT_{scale} value is become 1000 giving that is linear regarding the linear attenuation coefficient (IAEA, 2012) and hence:

$$S = \frac{\sigma_{water}}{1000}\ x\ 100\%$$

Contrast to Noise Ratio

Contrast noise ratio (CNR) of CT examination uses an enormous amounts of photon flux in image acquisition to attain low noise images. Thus, this consequence in greater radiation dosage to the patient. Those images permit the differentiation of low contrast objects, showing significantly small variations in X-ray attenuation in the tissue due to differences in density or composition. In diagnostic medical imaging, the image noise for a uniform material is commonly an excellent cursor of visualizing capability of low contrasts commonly measured in CNR. To measure CNR, the contrast of two images are obtained by the mean CT numbers difference within nominated ROIs and is divided by the average noise for these two ROIs (Khaled ME. and Sandra AO, 2014):

$$\mathrm{CNR} = \frac{\mathrm{CT}_1 - \mathrm{CT}_2}{\frac{(\sigma_1 + \sigma_2)}{2}}$$

This factor is beneficial when optimizing a CT studies protocol for a particular contrast situation for instance tissue density contrast, iodine contrast and air contrast.

Uniformity of Image Noise and CT Number

Throughout the examination field of view, uniform water filled phantom scanning must deliver a CT image with comparable noise values and pixel quantities.

Conversely, water filled phantom scanning regularly display differences in noise quantities and CT number (informally called Hounsfield units). The differences in CT number and image noises are clear when the phantom is encircled with great contrast materials, when extremity of big phantom is investigated, and when phantom is not correctly centered at the isocenter. The inappropriate isocenter setup of phantom highly pronounce the variation (IAEA, 2012). CT number is related with the linear attenuation coefficient of the medium (Hendee & Ritenour, 2002). The CT numbers for most CT scanners are between -1000 for air and +1000 for hard bone. The CT number for water is zero. CT number is given by:

$$CT\ number\ = \frac{(\mu\ -\ \mu_w)}{\mu_w} = \frac{\mu}{\mu_w}$$

where, μ is the linear attenuation coefficient of a material and μ_w is the linear attenuation of water. Higher CT number correspond to brighter CT image and lower CT number to darker CT image (Hendee & Ritenour, 2002). Lowering of CT number at the center of water phantom is called "cupping". High CT number at the center of water phantom is called "peaking". Both result used to describe image quality.

Medical physics professionals and medical image scientists are highly interested in the image quality of imaging systems. Because, medical image quality should provide sufficient diagnostic conclusions for prompt detection of illnesses (Elnour et al., 2017). During CT image, high image quality is mostly related with high doses to the patient. Hence, to optimize the quality of CT image, optimization of patient dose and other CT dosimetry are highly essential. In Ethiopia, researches on patient dose and image quality in CT have not be conducted yet. Hence, the goal of this research is to assess CT image quality factors with the intention of optimizing imaging protocols.

Common Mistakes of CT scan Operators

Professionals do common mistakes during ACR CT phantom image quality measurements. The most important sources of professional error are the techniques of observing medical images using eyes. The diagnostic consequence is highly related with the capability of the image scientists towards comprehending the required medical image information. The image scientist observes and read the diagnostic medical

image subjectively using his eye (Lee, Nam, Jang, & Kim, 2021). Then, they generate subjective decision based on his understanding. This will lead to aggravate the error frequency of medical diagnosis and varying medical outcome may be recorded for the same signals(Lee et al., 2021). McCollough et al (McCollough et al, 2004) mentioned the most common errors done by medical professionals as follows in Table 3-8.

Table III-8

The most common errors of an operators (McCollough et al, 2004)

Common Errors of Operator
Non-standard use of scan parameter. Recommended ACR CT phantom scan parameters are provided by professionals
Inappropriate configuration of detector (N and T)
Improper table feed or pitch computation
Ignoring SMPTE patter or alternate video test pattern submitting
Bad phantom alignment (central wire not centered in ramp, all 4 BBs not the same brightness)
Placing images in the wrong positions (boxes) on the films submitted to the ACR
Filming images with the wrong window width and window level settings
Wring size or wrong position of ROIs
Submitting images having obvious artifacts or other deficient results
conducting CTDI scans using the wrong detector configuration (N and T values)
Not submitting the printed Excel dose calculator spreadsheet
Not noting the difference between mA, mAs and effective mAs (=mAs/pitch) when completing the forms
Not displaying sufficient technical parameters on the printed films (all scan parameters should be shown)
Positioning ROIs such that annotation covers important portions of the image
Using too small or too large reconstruction FOV

CHAPTER IV

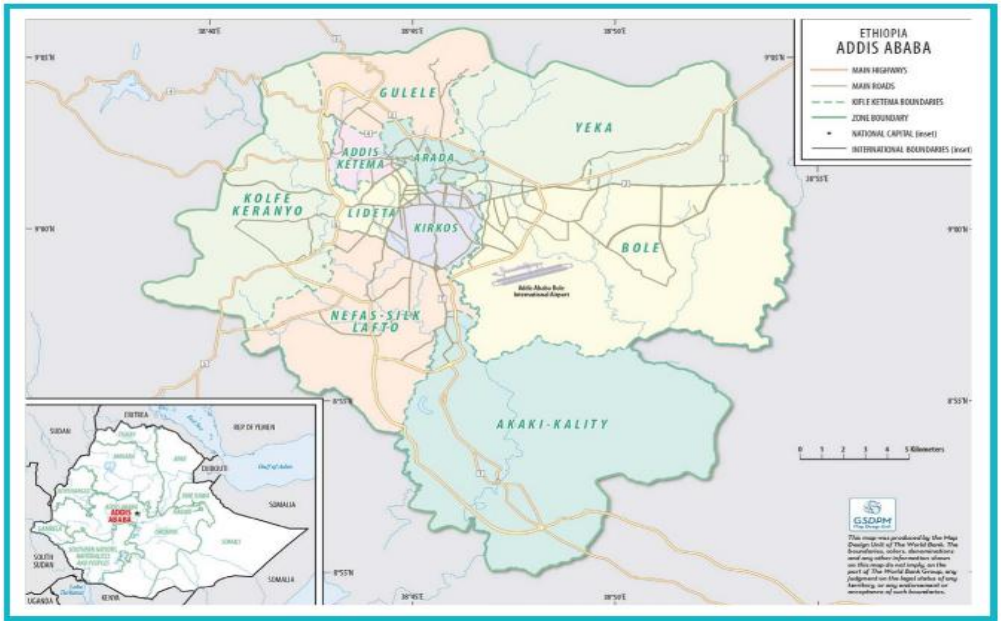
MATERIALS AND METHODS

4.1 Study area

Ethiopia is land-locked country found in the horn of Africa, which is the second most populated nation next to Nigeria. Addis Ababa (AA) is its capital town and seat for the Federal Democratic Republic of Ethiopia (FDRE). AA is the hearts of administration, transportation, economics, and social culture. It is the fastest booming city in the eastern Africa. AA is also the home for the head quarter of African Union (AU) and diplomatic capital of Africa. It hosts also several international organizations like United Nations Economic Commission for Africa (UNECA). The population live in AA are estimated to be twenty-five percent of urban population live in the nation (World Bank, 2015). Generally, Addis Ababa can be called it is the political capital of Africa. It is covering about 647 km^2 (World Bank, 2015). Geographically, the coordinate of Addis Ababa is 9º1′48″N 38º44′24″E, Figure 4-1. AA was established by empress Taytu and emperor Minilik-II in 1886. It is most populated and largest city of the nation. AA is self-governing chartered city of the nation. It is divided into 11 sub-cities.

Figure IV-1

Maps of Ethiopia and Addis Ababa (World Bank, 2015)



4.2 Geography and Climate

The geographical location of AA is located on a plateau at an altitude of 2,408 meters (UN-HABITAT, 2003) in the central region of Ethiopia (officially called FDRE). FDRE is one of the oldest independent civilization country in the world. It is consisting 1.1043 km^2 comprising the major landmass (EMoH, 2010). Ethiopia shared boarders at the east to Eretria, northeast to Djibouti, East to Somali, west Sudan and South Sudan, and south to Kenya. Ethiopia's geographic co-ordinates are in between 8:00N and 38:00E. Ethiopia has great variety of geographical appearance. It has a topography raging from high peaks to low depressions. The high peaks reach 4,550 meter above sea levels. The low depressions also reach 110 meters below sea level. The most common climate of Ethiopia is monsoon (EMoH, 2010).

4.3 Demographic Situation

According to UN world population prospect 2017 report, the population size of Ethiopia is around 104.957 million; the male to female population ratio is around 0.997. In particular, Addis Ababa is comprising around 3.316 million population (United Nations, 2016), which is 25 percent of urban population in Ethiopia (World Bank, 2015). However, many current approximations put the population near towards 5 million (Erena eta al, 2017). The age structure of Ethiopia's population has been persisted largely young under the age of 15 years with 41% that telling greater fertility rate of the population; and 54% of the total population is accounting in the age range between 15 – 59 years; whereas 5% of the total populace fall in the age category greater than 65 (United Nations, 2017). The gender fraction among female and male exists nearly equivalent. From the total population, 19% are living in urban.

The annual economic growth of AA is registered by 14 percent. Most strong economic activities undertaking within the city contributing around 50 percent to the country's GDP. That is why it is said that AA is the development engine for the country.

4.4 Health Status of Addis Ababa

The main health problems of Ethiopia are avoidable infectious sicknesses and nutritious syndromes. Ethiopia has been prepared major progresses towards improving the population health status in the previous twenty years however great percentage of illness and death due to poor healthcare systems compared to other African countries.

According to the ministry of health of Ethiopia – HSDP-IV in 2010 report, the life expectancy of the population of Ethiopia is 53.4 and 55.4 for male and female respectively. The current health policy of Ethiopia has been published in fifty years setting followed by four successive phases of health sector development program - HSDP (EMoH, 2010; Richard, 2009).

The critical reviews and scrutiny of the nature, magnitude and root causes of the prevailing health problems of the country resulted the formulation of policy and HSDP. The wider awareness of the newly emerging health problems is also contributed the policy formulation and the development of the first HSDP. Since the development of HSDP, the Federal Ministry of Health has formulated as well as implemented several policies and strategies. The policies and strategies were designed to improve healthcare framework in the country (EMoH, 2010).

4.5 Research Materials

The materials required for this study were Dose-length product (DLP) meter-100 mm long pencil ionization chamber, poly-methyl methacrylate (PMMA) phantom (head and body phantoms) representing average adult patients, caliper, laptop computer, complete MagicMaX software, RadiAnt DICOM viewer 5.5.1 (Digital Imaging and Communications in Medicine) and SPSS software.

The researcher was proceeding this great work with patience in the wave of covid-19 since the beginning of the research, for the success of developing possibly first regional diagnostic reference levels, and to put in place bench mark for further expansion of this work in whole of the Country.

4.6 Methodology of Developing DRLs

The primary steps to setting DRL values must be attached to well-clarified general or specific medical and methodical requirements aimed at tasks of clinical diagnosis using computed tomographic. A particular numerical values of DRL set for a country cannot used appropriately for another country, although the area of scanned body is being the similar. Generally, DRLs values must be according to data obtained from registries or surveys for the majority of patient examination types (ICRP, 2017b). Practically, it is difficult to knowing which DRL values are fairly low enough and which quality of images are fairly good enough to deliver the intended medical

information. Anyhow, DRL values derived from surveys or registries data will bring the majority medical professional to agree with that a particular value of the DRL.

Phantoms is valuable for measuring overall performance of CT radiology with AEC for comparison and checking its performance of several X-ray facilities (ICRP, 2017b). However, using phantom based surveys for setting DRL values is not recommended. Because data obtained from phantom based surveys cannot essentially reveal the medical and technical parameters for medical purposes. Furthermore, data based on phantom cannot include worker contribution and cannot include the protocols applied for data getting from surveys in similar data collection methods of different patients. To use phantom based data for setting DRL values, the complete techniques of measurement set up should be constructed on patient measurements. When setting the DRL values priority should be given for the most common protocols conducted in the facility. Setting the DRL values for the lower priority examinations are still recommended. However, the lower priority protocols surveying is more difficult; its population dose contribution is also lower than the high priority of most common protocols conducted in the facility. Therefore, the early stages of setting DRL values dose survey should not include the examinations marked as lower priority.

Undertaking valid comparison of surveyed data is mandatory to set the national or regional DRL values. The DRL values should be developed for each of the particular protocols. To make the comparison of DRLs more meaningful, it should between similar protocols. Furthermore, data survey for developing the DRL values should contain large sufficient and adequately diverse group of facilities to exemplify the range of practices within the nation for specific protocol. Nowadays, the production new technology in medical fields are increasing from time to time. Hence, DRL values should be linked to technology appropriately. Because, these emerging medical technologies may achieve varying doses to the patient for the intended diagnostic image quality (ICRP, 2017b).

Within a region or nation, the CT technology and practices will vary. That is why scientific professionals recommended that the DRL values should represent the applied procedures in the whole of region or nation. If two procedures have different quantities for the DRL values, setting two different DRL values for each procedure is recommended. This is particularly true if new methods are incorporated. The most acceptable bases of setting DRL values for a region or nation are patient based data

(ICRP, 2017b) rather than using phantom based data. But, using other bases of data (like phantom based data) for developing DRL values can deliver essential information.

4.7 Phase – 1 – Method

The research was designed by referring data from four-government hospitals (there was two CT machine in one of these hospital), one-private hospital and three private diagnostic centers who are experienced in clinical CT-examinations. Those hospitals and diagnostic centers (9 out of 35, i.e. 25.71%) were selected randomly by considering scientific data representativeness to determine the amount of data required depending on their registration period by the regulatory authority, number of workload and willingness to implement measures of basic quality controls. Before starting data collection, ethical clearance was granted from Addis Ababa health Bureau for private institutions (appendix – I) and from each government hospitals (appendix – II). Moreover, ethical clearances were granted from Near East University (appendix – III) and Ethiopian radiation protection authority (appendix – IV). The study was conducted to the most frequent performed CT procedures identified by the researcher such as head, chest, pelvic and cervical spine. The phase-1 data collection period was from August 2019 to February 2020.

Retrospective data was derived from the archive of radiology information systems (Figure 4-2 and 4-3). Retrospective data was collected using computer with complete RadiAnt DICOM Viewer 5.5.1 and SPSS software. RadiAnt DICOM Viewer uses to view accessible radiogram of patient data related with exposure parameters and radiation dose information. SPSS was used to conduct statistical analysis. The application of automatic tube current modulation Software was also considered in all CT. Data collection booksheet-1 (Appendix – V) contained the required parameters for computed tomography dose index (CTDI) estimation for each CT examination was appropriately prepared. The data collection booksheet-1 designed for all patients in order to maintain consistency of the information. Based on the format, retrospective data were collected from health centers patient dose archive. At least 20 patients reflecting both sexes per examinations was collected from each facility.

Figure IV-2

Sample of patient data archive of radiology information systems

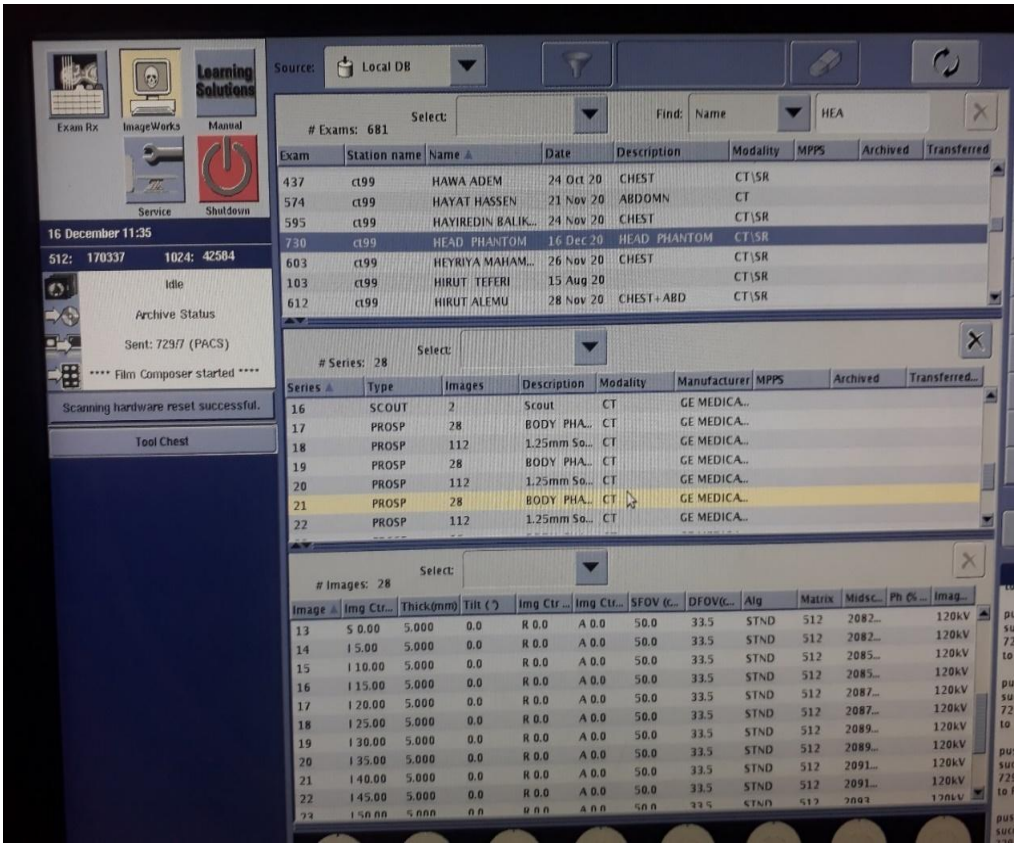
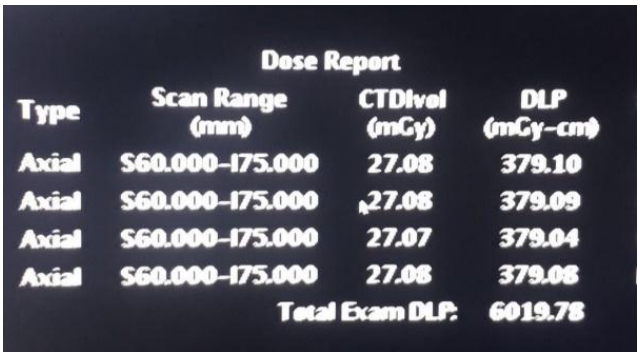


Figure IV-3

Retrospective data obtained from archives of radiology department



A total of 1195 adult patients' data were collected. From each health facilities, a total of 200 patients' data were collected from September 2019 to March 2020. All data was collected to adult patients. The data includes name of institutions, date of examination, details of the CT system (manufacturer, date of manufacture, number of detectors, date of installation, and model), patient dose factors (such as kVp - applied voltage in

kilovolt, mA – amount of current for the X-ray tube in milli-ampere, mAs - tube current time product, T = nominal slice thickness for single slice; t = total irradiation time; NT = normal width of irradiated beam for multi-slice machine; n = number of slices acquired simultaneously; $CTDI_w$= weighted CTDI, $CTDI_v$= volume CTDI and DLP = dose length product with their corresponding axial projection were gathered accordingly (Appendix – V). Patient data such as sex, and age of patient were collected. Patient body weight has not been restricted to reflect the actual practice in the facilities. This was done to define the age group standard patient of the region to permit forthcoming comparison within the country as well as with international literatures. Finally, the data was transferred to excel sheets and the SPSS software for mathematical analysis.

Statistical analysis of all data obtained for this research was recorded and performed by statistical Package for the Social Sciences (SPSS) software. Measureable values are articulated 25^{th} percentile, 50^{th} percentile, 75^{th} percentile and mode. The average $CTDI_v$ and total DLP were computed as typical quantities useful for further study. The 3^{rd} quartile values of median DLP and $CTDI_v$ data dose distribution from each examination were used to set DRLs for Addis Ababa. The median values of $CTDI_v$ and DLP were also used as proposing "achievable dose".

4.8 Phase – 2 – Method

The research was conducted in Addis Ababa's health facilities. It uses poly-methyl methacrylate (PMMA) phantoms (head and body) and pencil ionization chamber to determine CT dose. The ultimate goal of the research was to establish acceptable and practicable CT DRL. At present, 92 CT machines were found in the whole of the country. Addis Ababa comprises 35 (i.e. 38.04%) CT machine. 22 CT facilities out of 35 (62.9%) were selected randomly for this research. However, only 9 of 21 CT facilities were functioning, other 13 CT facilities were not working with different reasons at the time of data collection. The selection criteria were depending on their registration period by the regulatory authority (ERPA), number of workload and willingness to implement measures of basic quality controls. The duration of data collection was from July 2020 to December 2020. Adult PMMA phantom based experimental CT data sets were collected from three principal government hospitals (in one of the hospital, there was two CT machine), three private owned hospitals and three private diagnostic centers. The research consists of different CT models and manufacturers. This includes Philips (Bright speed, optima CT540 and DU 5008C), Siemens (SOMATOM go top and SOMATOM scope) and GE (Revolution ACT, optima CT660 and optima CT520). The details of the CT systems used were listed in Table 4-1.

Table IV-1

Details of CT scan facility used this study

Hospitals	Type of Facility	CT Model	No of Detectors	Brand	Date of Installation
A	Private	Bright Speed	16	Philips	2018
B	Government	Optima CT540	64	Philips	2010
C	Government	DU 5008C	16	Philips	2011
D	Private	SOMATOM Scope	16	Siemens	2015
E	Private	SOMATOM scope	16	Siemens	2017
F	Private	SOMATOM go. Top	128	Siemens	2019
G	Private	Revolution ACT	16	GE	2019
H	Government	Optima CT660	64	GE	2015
I	Private	Optima CT520	16	GE	2020

Table 4-2 summarized prospective CT dose data gathered based on the image acquisition parameters.

Table IV-2

Average exposure parameters for Philips, Siemens and GE CT scanners used in this study

Brand	Phantom Types	Scan Type	mA/mAs	kVp (v)	Scan time(s)	T (mm)	No. images	Scan L(mm)	Feed (mm)	NT (mm)
Philips	Head	Axial	150 – 314	120	1.104	2	8	100	10	13
	Body	Axial	180 – 314	120	0.923	5	2	100	10	13
Siemens	Head	Axial	220 – 270	110-130*	1.473	5	2	100	10	21
	Body	Axial	220 – 250	100-110*	5.008	3	3	100	10	8
GE	Head	Axial	150 – 250	120	0.868	4	4	100	10	14
	Body	Axial	180 – 250	120	0.713	4	9	100	10	25

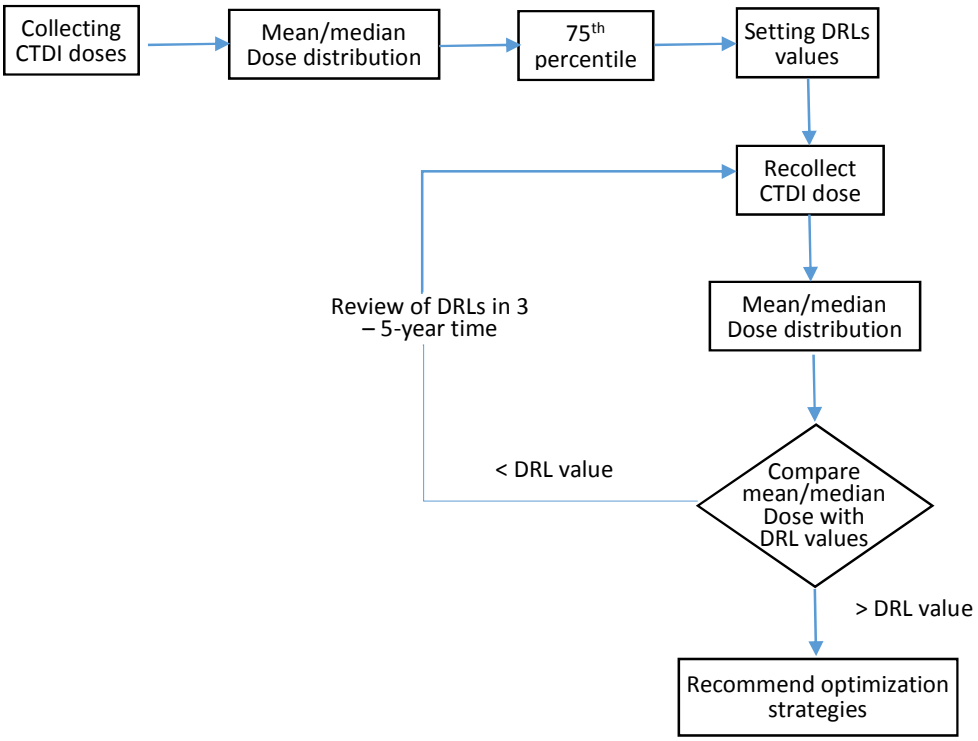
*T_feed=table feed, Aver.=average, T= slice thickness, *=effective tube current in mAs for Siemens CT scanners*

Quality control measurements (mechanical, electrical and radiation safety) were carried out by concerned Ethiopian Radiation Protection Authority in all the facilities before the required data collection. The quality control measurements were conducted according to international guidelines (American College of Radiology, 2018) to evaluate the performance of the X- ray generator and x ray tube in the CT facilities. Ethical clearance was granted from Addis Ababa health Bureau for private institutions (appendix – I). Because of the authorization given to each government hospital to grant ethical clearance, it was obtained from them (appendix – II).

The data were analyzed based on the following flow chart for both phase – 1 and phase – 2 research methodologies to develop DRLs values and its audit cycle, Figure 4-4.

Figure IV-4

Flow chart to develop DRLs values and its audit cycle (ICRP, 2000b)



4.8.1 Data Collection Tools

Prospective radiation dose data was collected by using pencil ionization chamber, PMMA phantom, MagicMax software and their accessories. The CT examination was performed by poly-methyl methacrylate (PMMA) phantom. The technical specification of the phantom was 16 cm and 32 cm diameters mimicking average adult human head and body, respectively. The length of the phantom is 15 cm elongated (Figure 4-5). Calibrated pencil ionization chamber (appendix – VI and VII) containing the MagicMax software (100 mm long and 4.9 cm^3 active volume with model: DCT10-MM and Serial number: 2640) connected with MagicMax USB-multi-meter (appendix – VIII) with serial number: G19-0860 was used to measure CT tube output dose, the set up was as presented (Figure 4-8).

Figure IV-5

Research materials: (1) body phantom, (2) head phantom, (3) solid PMMA rods and (4) Ionization chamber DCT10-MM: for CT applications

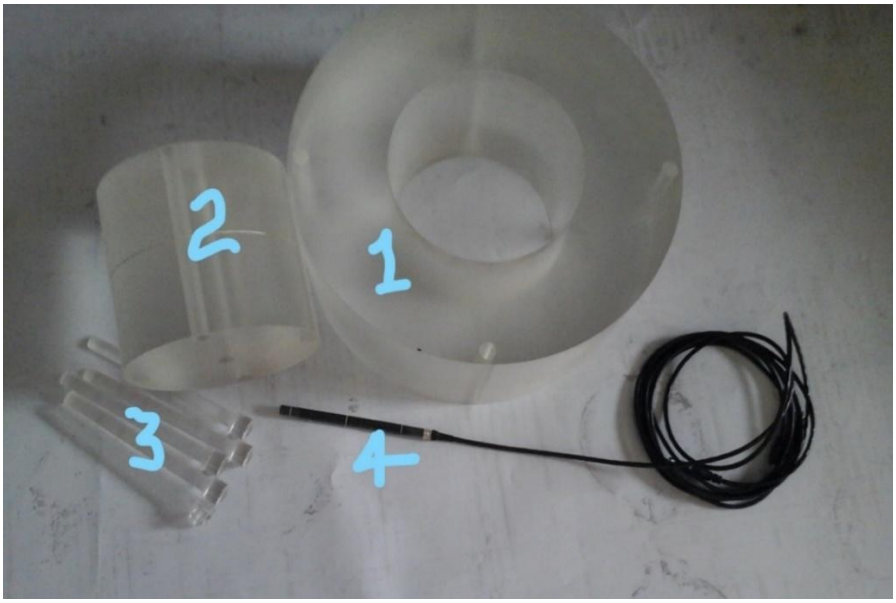


Figure IV-6

(1) The MagicMax universal with channels for multi-detector, (2) pencil Ionization chamber DCT10-MM with its PMMA tube holder: for CT applications

To measure the CT tube output dose, standard dosimetry phantoms (head and body) were used by planning each on the patient bed once at a time. First of all, the dosimetry body phantom was positioned on the patient bed. Following this, solid PMMA rods were plugged into the four peripheral cylindrical openings to eliminate the interference of atmospheric air (Figure 4-7 and 4-8). Then, the standardized ionization chamber was placed into the PMMA tube holder. Hence, the central cylindrical probe hole of the body phantom was plugged in using the tube holder together with the standardized ion chamber. Next, pilot scan (scan projection radiograph) image of the body phantom was conducted to adjust line configuration alignment. This helps to choose the phantom volume required to be studied. This is to check the coincidence between the scanners discounter (i.e. the point in space via which the central rays of the radiation beams pass) and the phantom isocenter. Subsequently, the scanners and phantom isocenter would line up in the Z-axis of the scanner. The output signal of the standard ionization chamber (model: DCT10-MM and Serial number: 2640) was recorded and displayed on the laptop computer. USB-multi-meter (REF: VD0202010, G19-0860Ver 02) was used to connect the chambers with the laptop computer.

Figure IV-7

Experimental set up of CT scan for CTDI measurement in one the radiology department

Figure IV-8

Experimental set up: (1) head phantom positioning and (2) Body phantom positioning. Pictures showing CTDI measurement set up using adult PMMA phantom viewing that Pencil ionization chamber positioning (at center) at different radiology department. (Image source: Researcher archive)

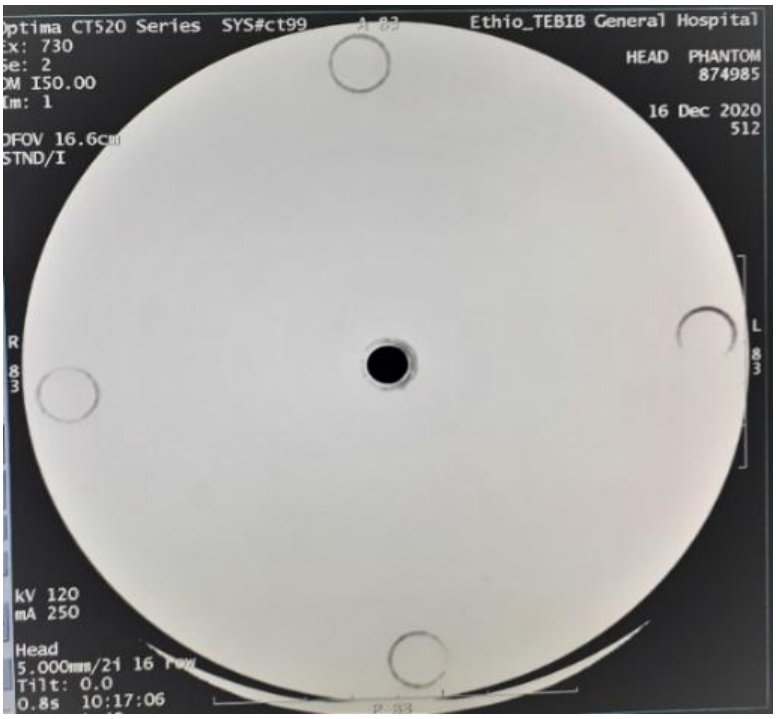
Figure IV-9
Setting up phantom and chamber in one of government teaching hospital

All dosimetry data can be recorded and displayed or exports as excel files by means of previously installed MagicMax software on the laptop computer (Figure – 14). The recording platform was set to zero before displaying the dose for each sequential or axial scan mode of the CT scan. Then and there, the accumulated dose values were measured for the central and peripheral body phantom by changing the position of the chamber from central probe hole to the other peripheral probe hole (Figure 4-9), based on the exposure parameters offered in Table 4-2.

Likewise, the procedures done on the body phantom measurements were repeated for head phantom to obtain the dose values. Each CT examination measurement were conducted at room temperature and pressure.

Figure IV-10

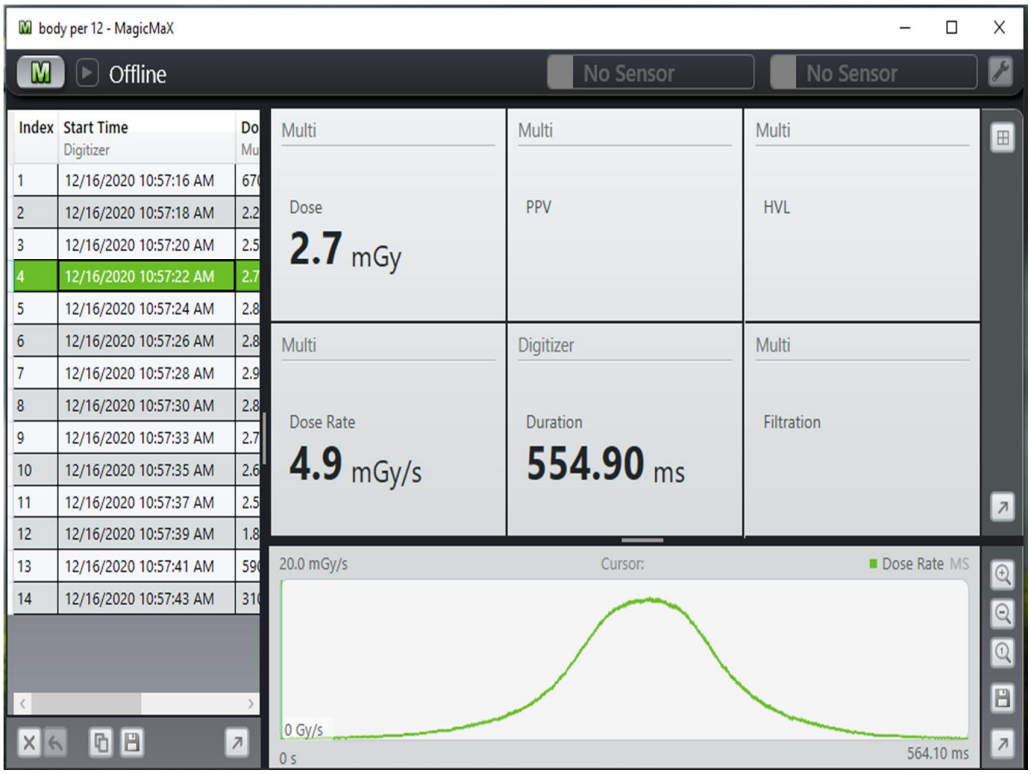
Measurement can be done by changing the position of the chamber from central probe hole to the other peripheral probe hole



The MagicMax software installed on the laptop delivers the settings and directions to the MagicMax universal for quantities and processes the measured data and displays or exports the results as excel files, Figure 4-10. The data are gathered by a microcontroller and transferred to the laptop through a great speed USB interface in a real time that permits long term wave form recording at highest time resolution, Figure 4-11.

Figure IV-11

Displays of long term wave form recording at highest time resolution for the measured data

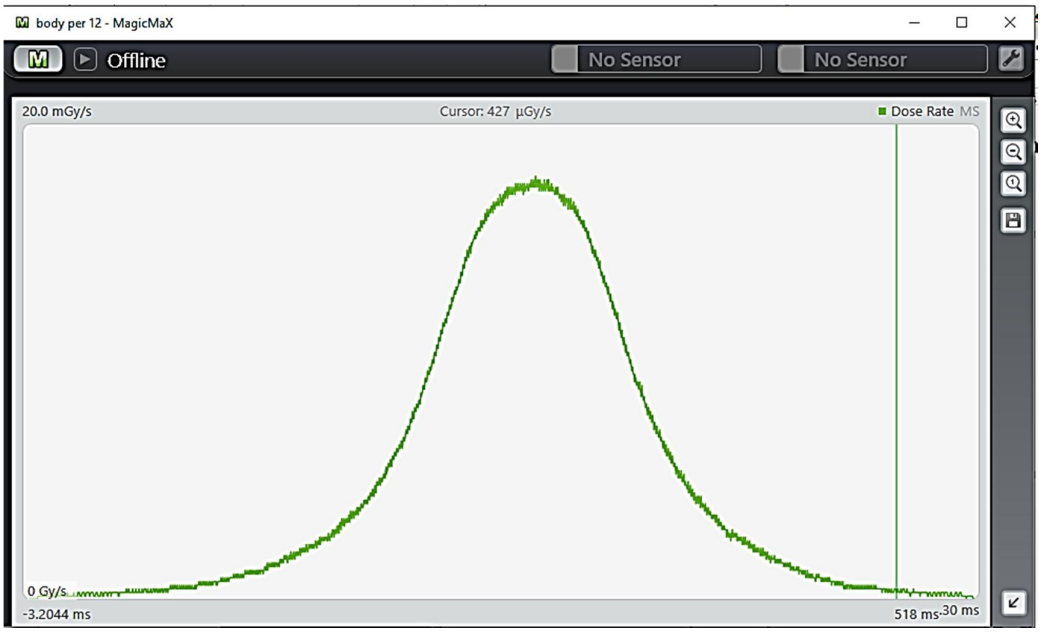


The CT-scanner machines were checked its performance by authorized body (Ethiopian Radiation and Nuclear Regulatory Authority) before data collection. Data collection booklets (appendix – IX – X) contained the required parameters for volume computed tomography dose index ($CTDI_v$) estimation to the patient for each radiographic examination was appropriately prepared. The recent appropriate dose related quantities for medical CT examinations are $CTDI_v$ and DLP values. But, this phase of researched used CTDIv as dose indicator to propose DRLs.

As done in phase-1 data collection, at least five exposures of each phantom was conducted in each of participated health facility. Finally, the data was transferred to excel sheets and the SPSS software for mathematical analysis.

Figure IV-12

The real time that permits long term wave form recording at highest time resolution for the data are gathered by a microcontroller and transferred to the laptop through a great speed USB interface



4.8.2 Sampling procedures

Initially, scientific random pattern sampling procedures of facilities was carried out prior to sampling to determine the amount of data. All data were collected from hospital and diagnostic centers who were experienced in clinical CT for the most frequent CT diagnostic procedures.

Based on the authentication of radiologists and/or radiographers, all examination procedures exposure factors and CT doses was collected from hospitals and clinics at their normal clinical practices. During data collection, the radiologic technical set-ups for imaging procedures was adjusted by the radiologists and/or radiographer without researcher interference. All examination was performed according to the technique used in each institution.

4.8.3 Data Collection

Data gathering booklets were used to gather data and retain consistency of the information for all examinations (appendix – IX – X). All data was collected using PMMA and ionization chamber at each hospital and diagnostic center. The data collection sheet includes information such as name of institutions, date of examination, details of the CT system (manufacturer, date of manufacture, number of detectors, date

of installation, and model), medical diagnosis factors (for example, kVp - applied voltage, mA - tube current in milli-ampere, mAs - tube current time product, T = nominal slice thickness for single slice; t = total irradiation time; NT = normal width of irradiated beam for multi-slice machine; n = number of slices acquired simultaneously; $CTDI_{100,c}$ = measurement at the center of the phantom; $CTDI_{100,p}$ = average of measurements at four different locations about the same phantom periphery) and radiographic procedure such as examination with their corresponding axial projection were collected. For more information, see Table 4-2.

4.9 Quantities of Dose in CT

Assuring the practical application of justification and optimization for protecting patients in diagnostic radiology, large scale electronic patient dose data from radiological information systems is significant to set DRL. The two main reasons for patient dose measuring or estimating are: 1) to give mechanisms of establishing and testing standards of acceptable work culture in order to promote the optimization of patient protection; and 2) to the absorbed dose determinations in the patients (F. Ebrahimzadeh, 2015). The patient radiation dose can be quantified in different methods. From several dose descriptors of CT scanners radiation output, the principal dosimeter quantities are CT dose indices such as weighted CTDI, volume CTDI and DLP values. ICRP recommended CT indices for the aim of developing the DRLs quantities are volume CTDI and DLP assessments of patients, however, .this researched used CTDIv as dose descriptors to propose DRLs. The CT dose indices reveal the quantities of ionizing radiation used to accomplish the clinical imaging during a complete revolution of the CT tube.

CT dose measurements may be made free-in-air or alternatively in phantoms using measurements in two cylindrical standard phantoms of head and body and a pencil ionization chamber. The combination of measurements made at the center (c) and at the periphery (10 mm below the surface denoted as p) to know the incident air kerma-length product (in mGy.cm). It is the integral of the air kerma free in air along a line of length parallel to the axis of rotation of the CT-machines (Council, 2004; IAEA, 2007; UNSCEAR, 2008). It can be calculated by.

$$\boldsymbol{P_{kL,CT} = \int_{L} k_a(L) dL}$$

Where, $P_{KL,CT}$ is air kerma length product a quantity assessed inside a phantom.

4.9.1 CT Air KERMA Index

CT dose is first defined in 1981 for computed tomography. It is a quantity of radiation dose for a single slice in CT procedures. The measurement is done at a stable applied voltage for a single axial rotation in the z-axis of the CT scanning. Then, $CTDI_{100}$ can be defined as the ratio of integral of the CT dose, D(z), measured over 100 mm ion chamber and the beam width (NT). Hence, the quantity of radiation dose measured for multi-slice scanner using phantoms and 100 mm ionization chamber, $CTDI_{100,PMMA}$, is given by (IAEA, 2007):

$$\boldsymbol{CTDI_{100,PMMA} = \frac{1}{NxT}\int_{-50}^{+50} D(z)\, dz}$$

where, T is the slice width per tomographic sections, N is the number of image slices per an examination (i.e. the number of CT images formed by a single x-ray tube rotation) and D(z) is the out potential of CT tube rotation in the z-direction. The subscript 100 represents for the radiation dose measurement using 100 mm ionization chamber. When it comes down to practice, the $CTDI_{100}$ can be obtained for central and peripheral axes for head and body protocols using PMMA phantoms are given by (Albngali, Shearer, van der Putten, Tuohy, & Colgan, 2018):

$$\boldsymbol{CTDI_{100,PMMA} = \frac{E.L.C}{NT\ (mm)}}$$

where, E is the measured dose value (in mGy), C is the pencil ionization chamber calibration factor and L is the active length of chamber (mm).

Chamber measurement, $CTDI_{100,\ reading}$, can simply calculated as:

$$\boldsymbol{CTDI_{100.reading} = \frac{E.L.C}{T}}$$

4.9.2 Weighted – CTDI Per Sequence

The weighted $CTDI_w$ (CTDIw) provides the dose measurement of a particular phantom in a weighted average of the central and peripheral positions of the phantom (Mourão A, Aburjaile W, 2019) for a single slice CT procedures. It can be computed by (IAEA, 2007):

$$CTDI_w = \frac{1}{3}CTDI_{100c,PMMA} + \frac{2}{3}CTDI_{100p,PMMA}$$

where, $CTDI_{100C,PMMA}$ is PMMA phantom measurement at the center and $CTDI_{100P,PMMA}$ denotes the CTDI values for the average measurements of periphery PMMA phantom. The subscripts c refers the central phantom measurements and p represents the average measurements conducted at the four peripheral locations of the phantoms. The standardized unit for weighted CTDI is Gy/C or Gy/A.s or J/kg.C.

4.9.3 Volume – CTDI Per Sequence

$CTDI_v$ is representing patient's absorbed dose in the target volume associated with a specified exam protocol (such as kVp, mAs). It quantifies the beam intensity of the CT radiation. $CTDI_v$ is the ratio of $CTDI_w$ by helical pitch or sequential scan spacing as given below:

$$\boldsymbol{CTDI_v = \frac{CTDI_w}{pitch} = \frac{CTDI_w}{p}}$$

where, CT-pitch factor is defined as the fraction of the patient bed travel (in cm) to the total beam collimation (mm) in the Z-axis consecutive scans/rotation (American Association of Physicists in Medicine, 2008; IAEA, 2007) for multi-slice machine of helical scanning is given by:

$$\text{Pitch} = \frac{\text{patient bed travel per gantry rotation}}{\text{nominal beam width}} = \frac{\ell}{\text{NT}}$$

where, the scan length, ℓ, for each exam protocol can be computed using (Campeloa M, 2016) the following formula:

$$\ell = L + NT + 15mm$$

where, L denotes the active length of the chamber (mm) and ℓ represents the patient bed travel in horizontal direction.

4.9.4 DLP for complete examination

Dose index is used as an indicator of overall exposure for a complete CT examination. This allows comparison of performance against a DRL set for the purpose of promoting optimization of patient protection and to allow computation of the effective dose (IAEA, 2009). The DLP is the multiplication of $CTDI_v$ and scan length for quantification of a phantom as given by (IAEA, 2007):

$$DLP = CTDI_v \times L$$

where, L is the total scan length (mm) reliant on the border of exposure volume of CT examination. For sequential scanning of several examinations, the sum of all the DLP values of each sequences represent the overall energy given by a scan protocol.

4.10 Data Processing and Statistical Analysis Plan

Statistical analysis of all data obtained for this research was recorded and analyzed using SPSS software. Measureable variable is articulated as arithmetic mean (signified as mean), first quartile, 2nd quartile, third quartile, and mode (a dataset appear mostly often). Descriptive statistics was applied for statistical analysis of the data.

In the first phase of this research, the average CTDI and Weighted CTDI per sequence, and volume CTDI per sequence and total DLP per complete examination of a procedure were computed as a basis for further study. The representative radiation dose in each facility was approximated from the median values of $CTDI_v$ and DLP data collected from each data site in the first phase of this research. Hence, the 3rd quartile values of the median dose distribution of these $CTDI_v$ and DLP values were used to propose regional DRLs for the first phase research that can be functional for CT modality in Addis Ababa.

However, the 2nd phase research used only CTDIv to propose its DRLs. From the same distribution of medians, ‘‘achievable dose” was proposed from the $CTDI_v$ values for each facility. These first and second phase research values were applied to indorse

consistency with the proposed DRLs as well as international recommendations. Therefore, this empower comparisons of our research outcomes with those of other countries succeeding those recommendations. In this research "achievable dose" was calculated at the 50^{th} percentile of the mean dose distribution for $CTDI_v$ and DLP values.

4.11 Phase – 3 – Materials and Methods

This prospective experimental research focused on diagnostic medical factors of CT examination to assess the CT image quality. This research was conducted at nine CT facility in Ethiopia at Addis Ababa to assess and analyze several image quality factors and inconsistency quantified all over the nine different CT modalities and to assess the recent procedures for quality assurance controls of CT schemes at those facilities. The period of data collection was from July 2021 to December 2021.

The CT image acquiring protocols were tested for their quality assurance by the recognized authority (ERPA) in 2021 before data collection. The quality assurance test frequency was differed among different radiology facilities. The test frequency of the CT scanners is scheduled once in a year by the regulatory authority. Their essential image quality criteria are testing the compliance of alignment, slice thickness, CT number accuracy, positioning accuracy, in-plane distance accuracy, high contrast (spatial) resolution, low contrast resolution, CT number uniformity and image noise using accredited image quality phantom, like American college of radiology (ACR) CT accreditation phantom.

ACR CT accreditation phantom was used to collect the CT image quality data sets from the aforementioned CT facilities. The image quality data measurement using the ACR CT phantom was set up as shown in Figure 4-13. This ACR CT phantom (a solid water phantom comprising four modules, Figure 4-14) is principally made from water equivalent materials and specifically designed towards assessing CT image quality (slice width, CT number accuracy, positioning accuracy, high contrast (spatial) resolution, low contrast resolution, CT number uniformity and image noise). The geometry of each module is 20 cm diameter and 4 cm length comprising white alignment mark coats to show alignment laser. Head, foot and top marks are also highlighted to assist measurement setup. Module – 1 is working to evaluate positioning and alignment, slice thickness, and CT number accuracy. Module – 2 is also functioning to measure low contrast detectability. Module – 3 specifically manufactured from uniform tissue equivalent materials used to measure CT number uniformity. Module – 4 is also functioning to measure high contrast (spatial) resolution (Mulyadin et al, 2018). The researcher was focused on CT number, image noise, HU standard deviation (image noise), mean HU value, low contrast detectability, and number of distinguishable high contrast patterns measurement to assess image quality

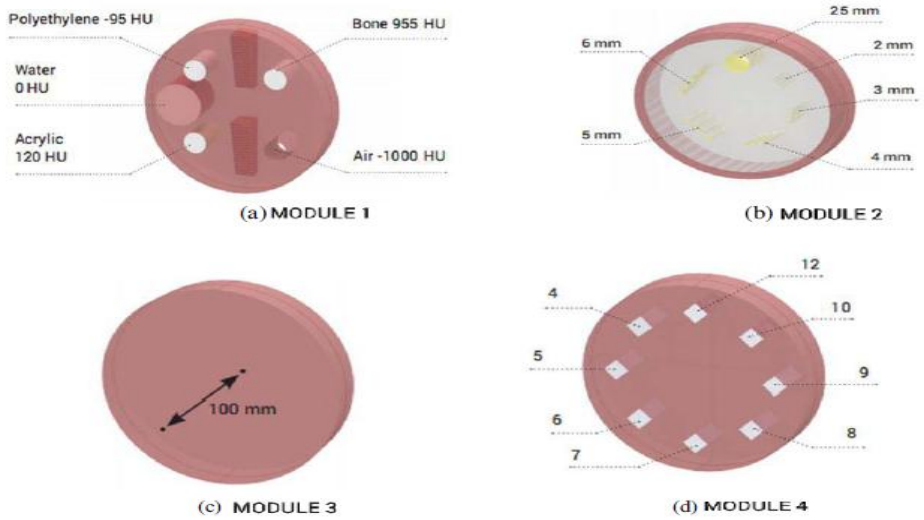
factors. The research consists of different CT models and manufacturers. This includes Philips (Bright speed, optima CT540 and DU 5008C), Siemens (SOMATOM go top and SOMATOM scope) and GE (Revolution ACT, optima CT660 and optima CT520). The details of the CT systems image acquisition parameters used were listed in Table 4-3. That prospective CT dose data were gathered using previously prepared data collection format (appendix – XI) developed by the researcher based on the image acquisition parameters as summarized in Table 4-3.

Figure IV-13

Multimodality ACR image quality phantom setup used in the CT image analysis (Gammex, 2020; Mulyadin et al, 2018)

Figure IV-14

The four ACR CT accreditation phantom modules description (Gammex, 2020; Mulyadin et al, 2018)



The researcher observed twelve ACR phantom images at each radiology department of the participated health facility. The head (brain) in axial scan mode and body (abdomen) in helical scan mode were used for the CT image acquisition protocols of adult patients based on image acquiring parameter. The influence of scanning factors that consequently affect the clinical examination protocols is illustrated in Table 4-3. The phantom scanning parameters (appendix – XII) were derived from ACR CT accreditation phantom instruction. The phantom was aligned in the coronal, sagittal and axial plains. The ACR phantom was centered on the patient bed and fasten stationary.

Table IV-3

Average image acquisition parameters used in image QA testing

Parameters	Adult head	Adult Abdomen
kVp	120	120
mA	280	260
Time per rotation (s)	0.8	0.8
Z-axis Collimation (T, in mm)	5	5
Number of Data Channels (N)	8	8
Table Increment (mm, or mm/rot)	10	10
Pitch		0.5
Scan FOV (cm, name)	22.3	Large
Axial (A) or Helical (H)	A	H
Display FOV (cm)	23.8	21
Reconstruction algorithm	STD	STD

4.11.1 CT Number Linearity and Slice Thickness Estimation

The ACR CT accreditation phantom was carefully positioned with head first orientation as patient into the center of the gantry. The CT scanner's internal and external sets of alignment laser light was correctly placed over the mark line matching to the module-1center. The table location was set at zero landmarks. Head examination protocol was used to obtain single axial scan at that zero landmark position with less than 2mm scan width for phantom alignment. The ACR CT phantom scanning parameters were according to Table 4-2. The images were filmed at ROI = 200.00 mm^2 as well as WW = 400 HU and WL= 0 HU values.

a) CT Number Linearity

The CT number of all materials may differ due to beam spectra of the X-ray system, beam hardening, beam scattering, and others. The materials contained in the different image quality accreditation phantom are assigned with specific CT numbers unveiling mean quantities acquired from several models of CT scanner, Table 4-4, and Table 4-5. To obtain the CT number value of a particular material in the ACR CT phantom, a circular ROIs were located on each cylinders of acrylic, polyethylene, bone, air and water to measure and record their corresponding CT numbers. The CT number values should be within the program tolerance for the acceptable values corresponding to the material. The relationship between the CT number and electron density correlation curve was illustrated in the result and analysis section.

Table IV-4
CT number and limits of tolerance for the various materials in ACR CT accreditation phantom (McCollough et al, 2004)

Material	Mean CT number tolerance (HU)
Air	-1005 to -979
Acrylic	+110 to +130
Water	-7 to +7 (0±5)
Polyethylene	-107 to -87
Bone	+850 to +970

Table IV-5
Image quality criteria and their quantitative test tolerance

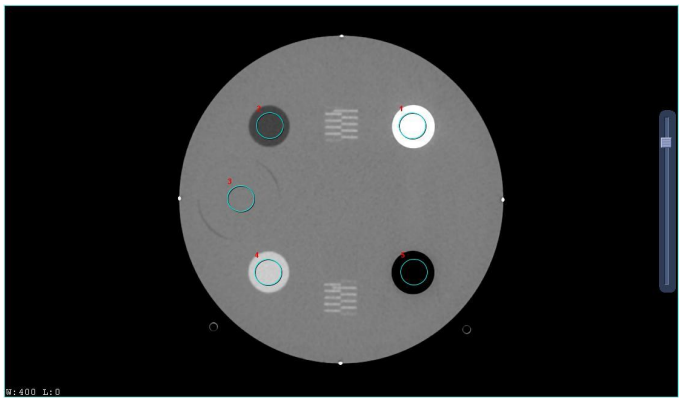
Test	Acceptable	Achievable
CT number	±5 from baseline value	±4
Uniformity	±10	±4
Noise	±25 of the baseline value	±10 of the baseline
Artifacts	No artifacts that have the potential to compromise diagnostic confidence	No visible artifacts

b) Slice Thickness and CT number of Water versus Slice Thickness

The discrete wires located on a ramp represent the width of CT image. The central wire of the slice width ramps at the 12 o'clock and 6 o'clock were clearly visible and symmetrically positioned. The spacing between consecutive wires is 0.5 mm with respect to the Z-direction (McCollough et al, 2004). Therefore, slice thickness of an image was determined by counting the total number of well-visible wires and dividing the sum by 2. Note, all the four steel BBs at 12, 3, 6, and 9 o'clock must be clearly visible and symmetrically positioned, Figure 4-15.

Figure IV-15

ROI for each tissue equivalent materials (Image source: Researcher archive)

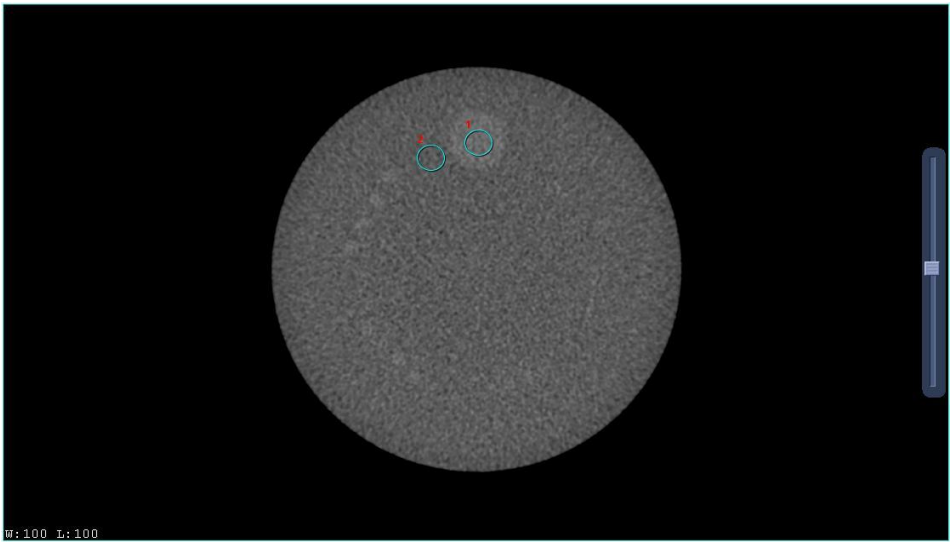


4.11.2 Estimation of Low Contrast Detectability

The ACR CT accreditation phantom was carefully positioned with head first orientation as patient into the center of the gantry. The CT scanner's internal and external sets of alignment laser light was correctly placed over the mark line matching to the module-2 center. The table location was set at zero landmarks. Adult abdomen technique was used to obtain single axial scan at that zero landmark position. The images were filmed at ROI = 100 mm^2 as well as WW = 100 HU and WL= 100 HU values. Module 2 contains a series of four cylinders (each at 6 HU (0.6%) variation from the background material (Gammex, 2020)) with different diameters of 2, 3, 4, 5, and 6 mm (Figure 4-16). To know the low contrast resolution (detectability), determine the cylinder set having the smallest diameter for which all four cylinders are clearly seen. At least, all four 6 mm rods cylinders should be seen brightly.

Figure IV-16

Low contrast resolution assessment by using the four cylinder groups (Image source: Researcher archive)



A circular ROI was located on the 25 mm water cylinder (to measure CT#1). Another circular ROI was placed between the small and big cylinder (to quantify CT#2). This was done to measure and record their corresponding CT numbers and noise. The standard deviation displayed represent the contrast noise of the image. Then, the low contrast resolution (detectability) can be also obtained by measuring CNR. To measure CNR, the contrast of two images are obtained by the mean CT numbers difference within nominated ROIs (CT#1 – CT#2) and is divided by the average noise for these two ROIs (Khaled ME. and Sandra AO, 2014):

$$\mathrm{CNR} = \frac{\mathrm{CT}_{\#1} - \mathrm{CT}_{\#2}}{\frac{(\sigma_1 + \sigma_2)}{2}}$$

4.11.3 Proposing, Uniformity, Noise and In-Plane Distance Accuracy

The ACR CT accreditation phantom was carefully positioned with head first orientation as patient into the center of the gantry. The CT scanner's internal and external sets of alignment laser light was correctly placed over the mark line matching to the module-3 center. The table location was set at zero landmarks. Adult abdomen technique was used to obtain single axial scan at that zero landmark position. The images were filmed at ROI = 400 mm^2 as well as WW = 100 HU and WL= 0 HU

values. A circular ROI was located at the center and edges (i.e. at 3, 6, 9 and 12 o'clock) to measure CT#1, Figure 4-17. Then, CT number and noise was recorded into the previously prepared datasheet. Figure 4-17 (a) shown position selection of center and edges ROI for noise and CT number uniformity measurement. The center ROI used for noise measurement, and center and edge ROIs used for CT number uniformity measurement.

In ACR CT phantom, noise was quantified as the STD of voxel values using the center ROI of test image. While, the CT number uniformity value for all four edge ROIs can be obtained by quantifying the mean CT number with a ROI at the center and four edge locations value and computing as give below:

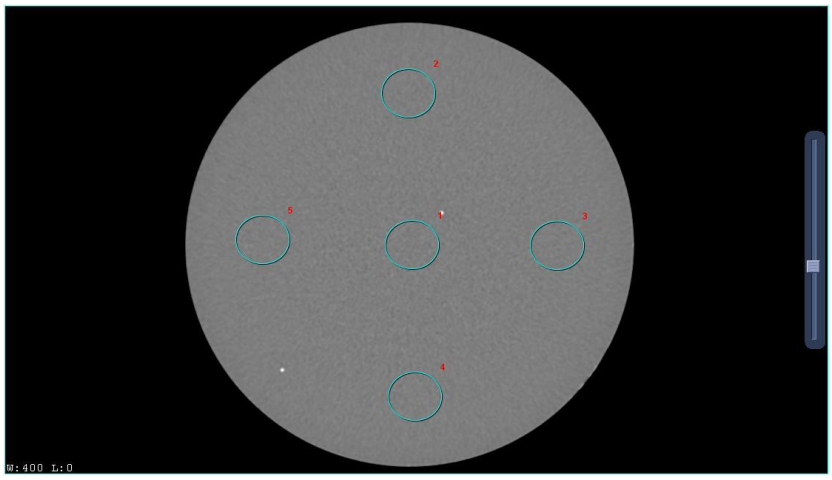
$$\textbf{CT number Uniformity value} = |\textbf{center mean CT\#} - \textbf{edge mean CT\#}|$$

where, the edge to center CT number must measure ≤ 5 HU and the center CT number should equal 0±5 HU (ACR, 2004).

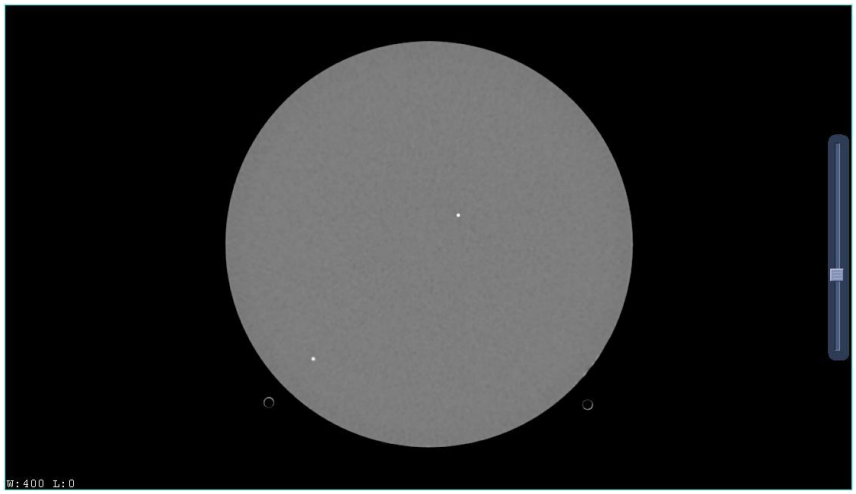
The in-plane distance accuracy or section sensitivity profile assessments can be done through using the two very small 0.28 mm steel BBs, see Figure 4-17 (b).

Figure IV-17

Illustrates an image of module 3 comprising tissue equivalent material: (a) towards measuring CT number uniformity and (b) in-plane distance accuracy and section sensitivity profile assessment were also conducted using the two very small 0.28 mm BBs.



(a)



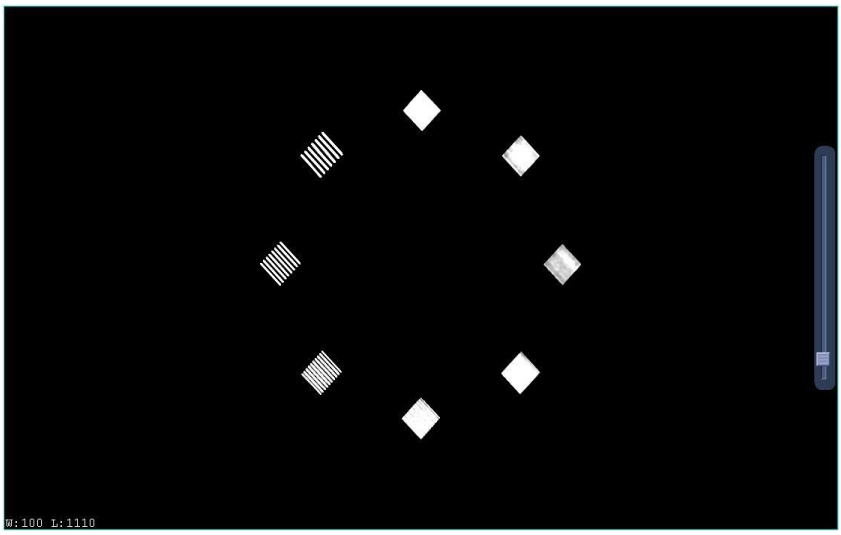
(b)

4.11.4 Estimation of High Contrast Resolution

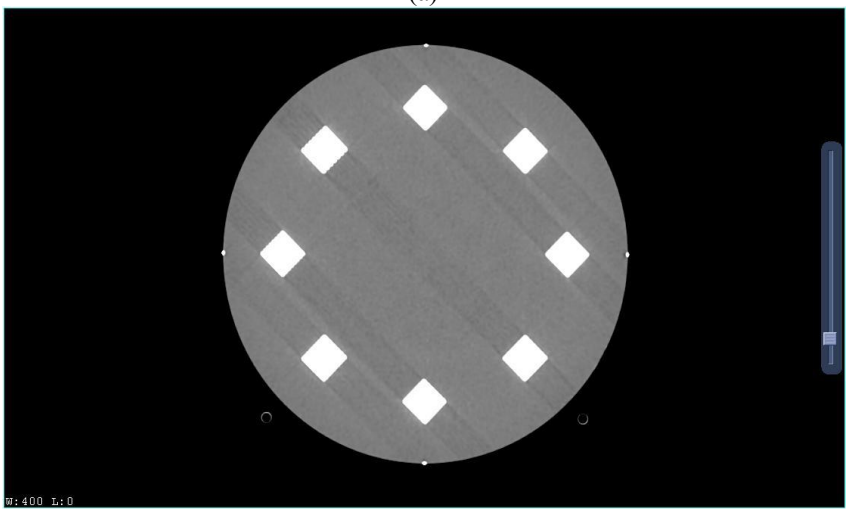
The ACR CT accreditation phantom was carefully positioned with head first orientation as patient into the center of the gantry. The CT scanner's internal and external sets of alignment laser light was correctly placed over the mark line matching to the module-4 center. The table location was set at zero landmarks. HRC and abdomen techniques were used to obtain single axial scan at that zero landmark position. The images were filmed at WW = 100 HU and WL= 1100 HU values. The high spatial resolution can be determined using spatial frequencies of the eight aluminum bar patterns that is well-matched to 4, 5, 6, 7, 8, 9, 10, and 12 line pairs per centimeter (lp/cm) (ACR, 2004), Figure 4-18. The highest spatial frequencies for the bars and spaces should be brightly seen at lowered room light. At least 5 lp/cm and 6 lp/cm must be determined for abdomen and HRC, respectively (ACR, 2004). Then, the data was recorded on worksheet.

Figure IV-18

High contrast resolution bar patters. (1) accurate image viewing window for estimating spatial resolution; (2) imprecise image viewing window and level settings that show streak artifacts caused by the high attenuation of the bars (Image source: Researcher archive).



(a)



(b)

Quality control measurements (mechanical, electrical and radiation safety) were carried out by concerned Ethiopian Radiation Protection Authority in all the facilities before the requiring data collection. The quality control measurements were conducted according to international guidelines (American College of Radiology, 2018) to evaluate the performance of the X- ray generator and x ray tube in the CT facilities.

The acquired image quality factors were compared with the tolerance quantities. DICOM files for the image of every image quality investigation were copied into CD.

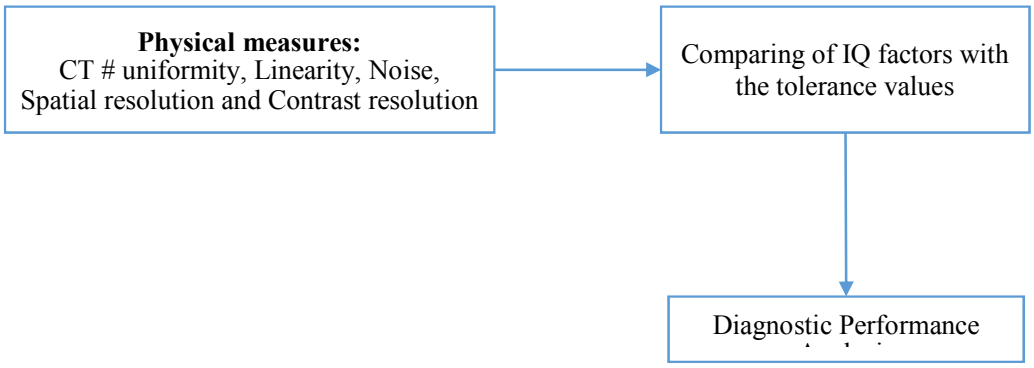
All images were accessed using the eFilm Lite™ 3.4 software (2010) and Media Viewer (provided by GE healthcare) installed into the laptop. All data analysis of CT numbers calibration, low contrast resolution, CT number uniformity, high contrast resolution, and slice thickness were conducted using this specifically grown software for viewing the CT images. Image noise computation was performed using SPSS software. The research applied the 95% significance levels.

The previously recorded image noise levels as well as low contrast detectability representing the routine phantom quantities for HRC and Abdomen protocols conducted were recorded and compared. Using the ACR phantom image analysis, the CT number, HU standard deviation, mean HU value, and number of distinguishable high contrast patterns were also recorded and inter-center comparison was conducted.

Statistical analysis of all data obtained for this research was recorded and analyzed using SPSS software. The average CT number, image noise, HU standard deviation, mean HU value, low contrast detectability, and number of distinguishable high contrast patterns were computed as a basis for further study, Figure 4-19. These values were applied for indorsing consistency with the proposed international recommendations. Therefore, this empower comparisons of our research outcomes with those of other countries succeeding those recommendations.

Figure IV-19

Flow chart of image quality analysis



4.12 Data Quality Management

Quality control measurements was carried out according to international guidelines (IPEM, 2005; AAPM, 2002) towards assessing the X-ray tubes and X-ray generators performance testing in CT. Furthermore, environmental and visual safety inspection were conducted at each facility participated in this research. Calibrated DLP meter and

standard dosimetry phantom were used for patient data collection following phase – 1 data collection. The calibration test was carried out and granted by the authorized body.

4.13 Data comparison for certainty

The uncertainty in the measurement describes a measure of how the measurement is wrong (IAEA, 2007). The purpose of every measurement is to get the value of a parameter or quantity. Uncertainty mentions to the level of confidence that can be an approximate of population central dose values. The major sources of uncertainty are specific parameters variability and measurement accuracy. Uncertainty is an essential factor in all extrapolation procedures. Principally, it is important in evaluating the effects of low dose range radiation doses (ICRP, 2007).

4.14 Confidence intervals

The 95% confidence-intervals (CI) for each examination protocol was calculated for the median DLP, weighted CTDI and volume CTDI about the 75th percentile level dose distribution (Commonwealth of Australia, 2015; H.Cember and E. Johnson, 2009) using:

$$\mathbf{95\%}\boldsymbol{CI} = \bar{\boldsymbol{x}} \pm \mathbf{2}\boldsymbol{\sigma}_{\bar{\boldsymbol{x}}} = \bar{\boldsymbol{x}} \pm \mathbf{2}\sqrt{\frac{\boldsymbol{\sigma}_{\boldsymbol{x}}}{\boldsymbol{n}}}$$

where, $\bar{\boldsymbol{x}}$ is the 75th percentile of facilities reference levels computed for the mean/median distribution of $CTDI_v$ and DLP for each protocol, $\boldsymbol{\sigma}_{\boldsymbol{x}}$ the variance of the possible sample size each protocol, and n is the sample size of each protocol.

4.15 Coefficient-of-Variation

Coefficient-of-variation (CV) describes the fraction of the standard deviation (σ_x) to the mean value ($\bar{\boldsymbol{x}}$) of an observed sample. The following equation capable to estimate CV as:

$$\boldsymbol{CV} = \frac{\boldsymbol{\sigma}_{\mathbf{x}}}{\bar{\boldsymbol{x}}}(\mathbf{100})\%$$

4.16 Ethical Consideration

The World Medical Association (WMA) has developed the Declaration of Helsinki as a statement of ethical principles for medical research involving human subjects, including research on identifiable human material and data. For this research, written consent will not be need to obtain from all patients, because the required data from patient was retrospective cohort data from patient data archiving software for phase one and for the second phase of data collection, it was experimental using CT- phantom and pencil ionization chamber with its complete accessories.

However, Ethical clearance approval for conducting the research was granted from Addis Ababa health Bureau (appendix – I), Ethiopian radiation protection authority (appendix – IV) and the Near East University (appendix – III) as well as from respective governmental hospital's college of health science participated in this research (appendix – II). Support letter written 'To Whom it may Concern' was granted from Graduate School of Applied Sciences at Near East University to request assistance for the research from concerned bodies (appendix – XIII). These approval was followed to protect the personal information of the research participant's privacy and confidentiality. The accountability and responsibility for keeping confidentiality is always a duty of the researcher and medical professionals. This is cannot be the duty of the research subjects.

4.17 Confidentiality of Patient Data

Clinical data collection was performed from hospital and diagnostic centers which was acquired ethical approval from Addis Ababa health Bureau, Ethiopian Radiation Protection Authority (ERPA), St. Paul Hospital's Millennium Medical College Research directorate and Black Lion Hospital's Addis Ababa University College of health science Research Directorate. Patient consent may not be needed. However, confidentiality of the patient shall be secured according to the guideline of USA Health Insurance Portability and Accountability Act (USA_HIPAA) of 1996.

4.18 Dissemination of Results

The results of this research was disseminated to relevant stakeholders participated in the research as well as for others for the successful service delivery policy. Because the inclusive method of this research will open the eyes for further study to reach the whole of the country to communicate the challenges of radiography to the patient's dose optimization. It will extend to the development of quality standards patient care

culture and/or policy quality service by the radiographers in the x-ray facilities in Addis Ababa, Ethiopia. Additionally, the respective regulatory bodies of the country will be initiated to accept and structure radiological research findings as part of service provision program.

CHAPTER V

RESULTS AND DISCUSSIONS

5.1 Introduction

The primary aim of this research presented in this dissertation was: (a) to spread the current knowledge of developing DRLs and its practical application in CT procedures; (b) to exploit the detailed dose knowledge of this research to patient dose reduction without jeopardizing the required diagnostic image quality; and (c) to point out the strategies to reduce dose while maintaining diagnostic outcome. In this research, thorough prospective experimental processes were intensively used to get detailed and accurate dose values in order to deliver full prospective radiation doses from CT-scanning. Hence, the research contributes to the development of DRLs for the doses that can be implemented to CT for the selected common procedures.

This chapter delivers the results of phase – 1 retrospective data for the anatomical areas for which DRLs is expected to propose. This section also explained the comparison of results with international records. Statistical analysis of all data obtained for this research was performed using SPSS software. Measureable metrics articulated such as mean, mode, 25^{th}, 50^{th} and 75^{th} percentiles were presented by descriptive indicators. The $CTDI_v$ per sequence, DLP per examination and other expressive parameters were presented using percentages, table as well as figures for the common protocols of CT examinations.

This chapter also provides the results of phase – 2 prospective data assessment for the head and body phantom which represents human head and body for which DRLs is expected to propose. This section also explained the comparison of results with international records. Statistical analysis of all data obtained for this research was performed using SPSS software. Quantitative variables articulated as mean, 25^{th}, 50^{th} and 75^{th} percentiles were presented using descriptive statistics. The $CTDI_v$ per sequence and other communicative parameters were presented using percentages and table for the common protocols of CT examinations.

Phase – 3 results provided the result drawn from ACR CT accreditation phantom data collections. This section also explained the comparison of results with international records. Statistical analysis of all data obtained for this research was performed using

SPSS software. The CT image quality were measured using homogeneous ACR CT phantom according to the quality assurance control recommendations provided by AAPM, IEC and IPEM. ICRU report 54 recommended that statistical decision theories must be implemented in clinical imaging. The results of this study were illustrated for each HRC and abdomen protocols for each image quality parameters.

The limitations of phase – 1, 2 and 3 researches were described in details. This chapter also presented discussions about the research results and the conclusions drawn from the research. The final goal of any research and standardization start with the dissemination of concrete information to the end users. Hence, this research derived the main idea of the research to propose the practical application of DRLs focusing CT in Addis Ababa's health facilities radiology department.

5.1 Phase – 1 – Results

The anatomical area for which DRLs are calculated were head and chest each without and with contrast media, abdomen (multiphase)[3], pelvic and c-spine each without contrast medium applications. Since there are differences among CT scanners and exam procedures, the results shown substantial divergences in measured radiation dose values.

The data were analyzed for its mean, standard deviation, coefficient of variations, 95% confidence interval, median, minimum, maximum, 75th percentile, and 25th percentile for eight multi-slice CT and one dual slices scanner. The details of the CT scanners used in this study were summarized in Table 5-1.

Table V-1

Details of CT systems used in the study

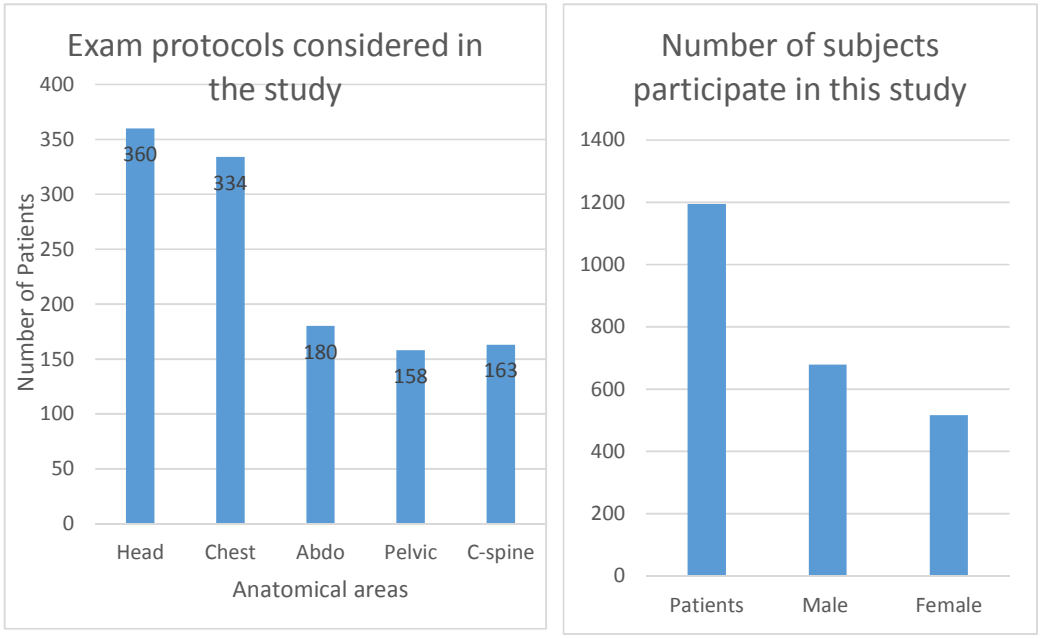
Center	Manufacturer	Model	Slice(n)	Installation date	Types of facilities
A	Philips	GE Bright Speed	16	2018	Private
B	Siemens	SOMATOM go. Top	128	2019	Private
C	GE	Optima CT660	64	2015	Government
D	Siemens	SOMATOM definition AS	20	2015	Government
E	Neusoft	NeuViz Dual	2	2009	Private
F	Neusoft	NeuViz 16	16	2013	Private
G	Philips	Optima CT540	16	2015	Government
H	Philips	Brilliance CT 64 slice	64	2013	Government
I	Philips	Pantry Brilliance CT 64	64	2010	Government

[3] Multiphase abdomen means four phasic dynamic CT examination like abdomen without contrast, arterial, PVP 3-5 minutes delay and 8 minute delay examinations. In this document you will find both multiphase and four phasic dynamic terms interchangeably to illustrate abdomen studies.

According to Figure 5-1, brain scans (30.13%) was the most common examination considered in the study followed by chest (27.95%), abdominal (15.06%), c-spine (13.64%) and pelvic (13.22%). From the total of 1195 patients, 57% were male and 43% were female, which were provided in Figure 5-1below. Also, ~56% of the facilities were government hospitals and the rest ~44% were private hospital and diagnostic centers.

Figure V-1

Graphical description to patients participated in the study



5.1.1 Analysis of Dose Measurements

The data collected by noting down from CT control console from nine CT facilities were analyzed. The results were presented as described in Table 5-2 and 5-3. Each of the corresponding CTDI and DLP values in each protocol has its own deviation from the true mean value. The standard deviation (STD) was also calculated and shown in Table 5-2. Table 5-2 also presented mean, minimum, 25th percentile, 50th percentile, 75th percentile and maximum values as well as the 95% CI for the 75th percentile CT dose distribution. The 75th percentile values of CTDI and DLP for each of the protocol were compared with international records (see Table 5-4 below).

For the values shown in Table 5-2, the mean $CTDI_v$ values for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast) were 49.24mGy, 46.47 mGy, 11.91 mGy, 11.17 mGy, 13.32 and 24.68±16.0 mGy, respectively. The STD for the mean CTDIv were 14, 7, 5, 4, 4, 6, and 16 mGy for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The computed minimum values for CTDIv were 26.69, 30.76, 5.83, 5.18, 6.55, 4.24, and 5.69 mGy for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The 25^{th} percentile values for the CTDIv were 42.87, 40.51, 6.96, 7, 8.68, 9.25, and 11.22 mGy for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The calculated median of $CTDI_v$ were 47.73 mGy, 47.77 mGy, 10.44 mGy, 9.15 mGy, 9.74 mGy, 12.72 mGy, 12.25 mGy, 11.64 mGy, 12.83 mGy and 18.63 mGy for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The 75^{th} percentile of $CTDI_v$ for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (without contrast) and c-spine (without contrast) were 52.70 mGy, 50.78 mGy, 16.56 mGy, 14.75 mGy, 13.66 mGy, 15.20 mGy and 37.52 mGy, respectively. The 95% CI for the 75^{th} percentile for $CTDI_v$ (in mGy) were 49 – 54, 50 – 53, 16 – 19, 13 – 16, 12 – 15, 12 – 16, and 35 – 40 for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively.

Table V-2

Mean, STD, minimum, maximum, 75^{th}, 50^{th} and 25^{th} Percentiles values for $CTDI_v$(mGy) and DLP(mGy.cm) for selected procedures

Exam	Contrast	Descriptor	Mean	STD	Min.	25^{th} Perc.	Median	75^{th} Perc.	Max.	95% CI for 75^{th} percent
Head	without	$CTDI_v$	49.24	14	26.69	42.87	48.2	51.57	80.2	49 - 54
		DLP	1171	617	542	827	1033	1237	2696	1241 - 1272
	With	$CTDI_v$	46.47	7	30.76	40.51	48.2	51.79	56.37	50 - 53
		DLP	1160	371	738	827	1158	1459	1746	1219 - 1238
Chest	Without	$CTDI_v$	11.91	5	5.83	6.96	9.52	17.13	20.96	16 - 19
		DLP	438	216	170	286	370	625	839	591 - 610
	With	$CTDI_v$	11.17	4	5.18	7	8.8	14.09	20.96	13 - 16
		DLP	419	205	171	274	360	565	838	511 - 530
Abdomen	multiphase	$CTDI_v$	11.53	4	6.55	8.68	11.11	13.66	20.04	012 - 15
		DLP	538	202	287	348	528	728	886	679 - 698
Pelvic	Without	$CTDI_v$	13.32	6	4.24	9.25	12.25	14.01	27.27	012 - 16
		DLP	489	215	164	365	419	605	926	576 - 595
C-spine	without	$CTDI_v$	24.68	16	5.69	11.22	17.95	37.76	53.26	35 - 40
		DLP	830	567	285	387	644	1106	2054	1138 - 1171

STD=standard deviation, Min =minimum, Max.=Maximum, and Max. – Min.=difference between maximum and minimum

Table 5-2 also presented the mean values for DLP in mGy.cm were 1171, 1160, 438, 419, 538, 489 and 830 for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The recorded STD for the DLP values were 617, 371, 216, 205, 202, 215, and 567 mGy.cm for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The minimum values for recorded DLP were 542, 738, 170, 171, 287, 164, and 285 mGy.cm for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The 25^{th} percentile of DLP for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast) were 827, 827, 286, 274, 348, 365 and 387 mGy.cm, respectively. The calculated median of DLP were 1033, 1158, 370, 360, 528, 419 and 644 mGy.cm aimed at head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast), respectively. The 75^{th} percentile of DLP were 1237, 1459, 625, 565, 728, 605 and 1106 mGy.cm aimed at head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast) correspondingly. The maximum recorded values for the DLP were 2696, 1746, 839, 838, 886, 926, and 2054 mGy.cm for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast) respectively.

Therefore, on the basis of the information provided above (in Table 5-2), the 75^{th} percentiles dose distributions for the $CTDI_v$ and DLP for each anatomical area of CT procedures were emplaced in Table 5-3, which can be proposed as first Addis Ababa's DRL values. The researcher do not want to deny some researches done regarding local CT DRLs in Addis Ababa. However, those studies were centered in one health facility to propose local DRLs. Therefore, this recent research were more general by incorporating several health institutions which are functioning in Addis Ababa.

Table V-3

DRL values proposed for the head, chest, pelvic and c-spine as well as abdomen

Exam	Contrast	DRL at 75th percentile		95% confidence intervals	
		$CTDI_v$(mGy)	DLP(mGy.cm)	CTDIv at 75th perc.	DLP at 75th perc.
Head	Pre- contrast	52	1237	49 - 54	1241 - 1272
	Post- contrast	52	1459	50 - 53	1219 - 1238
Chest	Pre- contrast	17	625	16 - 19	591 - 610
	Post- contrast	14	565	13 - 16	511 - 530
Abdomen	multiphase	14	728	012 - 15	679 - 698
Pelvic	Pre- contrast	14	605	012 - 16	576 - 595
C-spine	Pre- contrast	38	1106	35 - 40	1138 - 1171

$CTDI_v$ and DLP have been defined by the International Electrochemical Commission (ECC). The 75th percentile (proposed DRL for the study area) were derived from analysis of retrospective adult-patient data collected in selected health institutes in Addis Ababa. The dose distributions for the anatomical areas for each of the CT protocols analyzed are presented in Figure 5-2 and 5-3 represented from (a) to (n) for head (pre and post contrast), chest (pre and post contrast), abdomen (multiphase), pelvic (pre-contrast) and c-spine (pre contrast) anatomy. For simplicity of understanding, the DRL at 75th percentile being shown by broken horizontal line and written in words on top of each figure.

Figure V-2

Graphical illustrations on statistical DRLs for the two CT- protocols: (a), (c) and (e) show the proposed $CTDI_v$ distributions for head and chest, respectively; and (b), (d) and (f) illustrates the proposed DLP distributions for head and chest, respectively

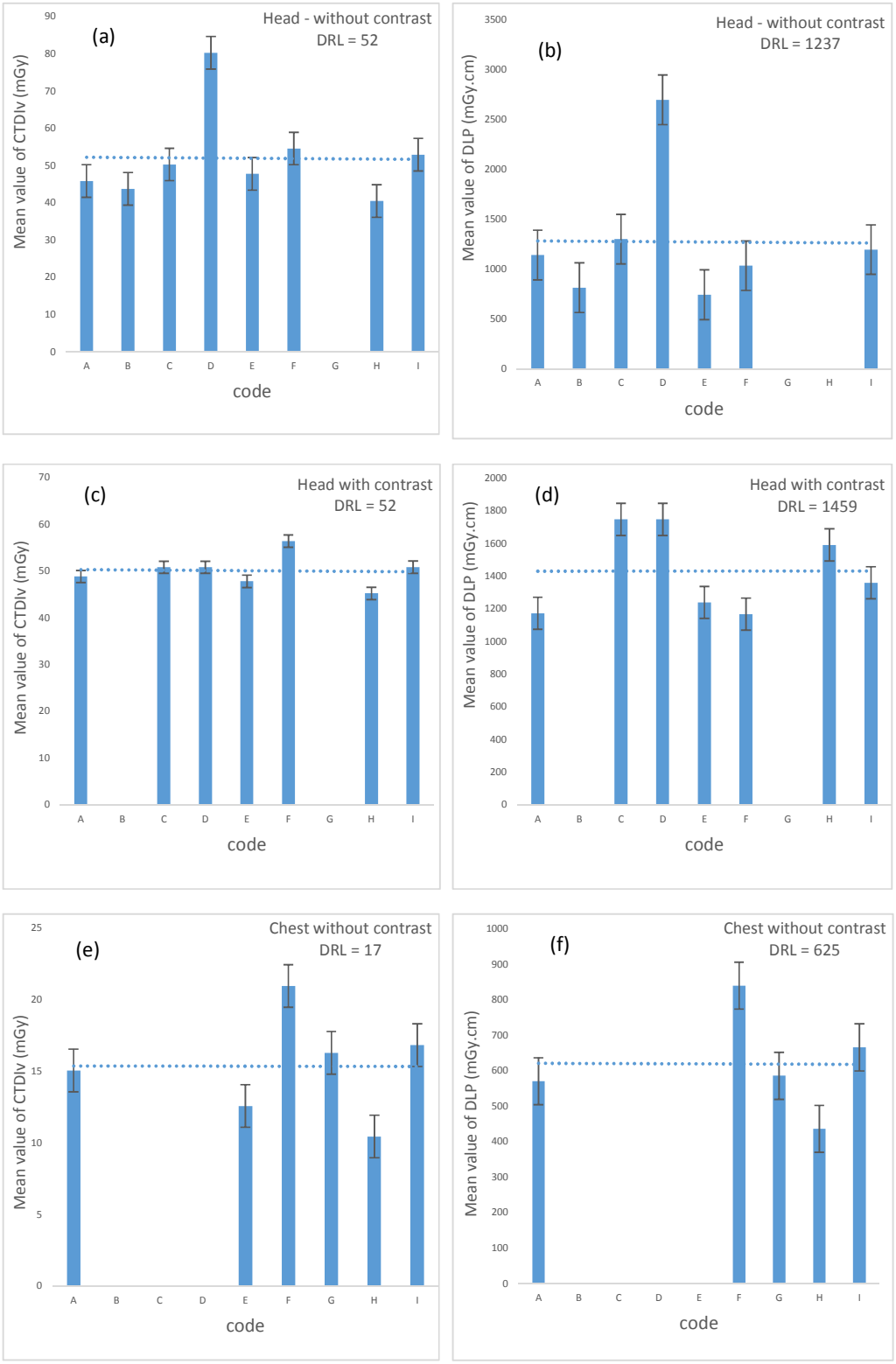
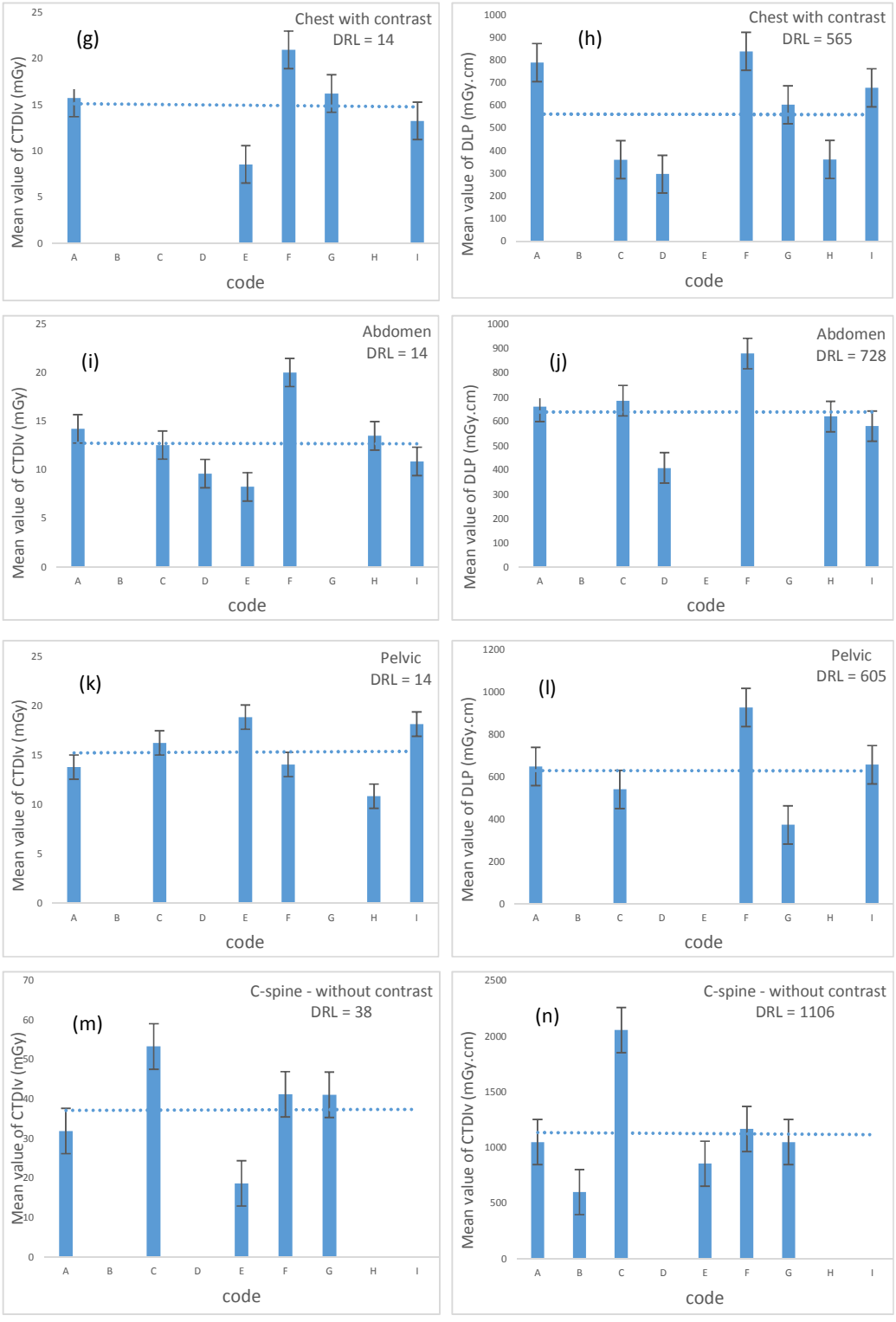


Figure V-3

Graphical illustrations on statistical DRLs distributions for the three CT- protocols: (g), (i), (k) and (m) show the proposed $CTDI_v$ distributions for abdomen, pelvic and c-spine, respectively; and (h) (j), (l) and (n) illustrate the proposed DLP distributions of abdomen, pelvic and c-spine, respectively



5.1.2 Comparison with International Records

The results were compared with considerable data and benchmarks available international DRLs data records**.** Table 5-4 contains DRLs of some countries from the world established for selected anatomical exam of CT procedures. Comparisons of estimated DRL values in terms of the dose descriptors of $CTDI_v$ and DLP with ICRP, European commission and some other countries have been presented accordingly.

Table V-4

International DRLs comparison of DRLs with Addis Ababa

Exam	Descriptor	$Head_{woc}$	$Head_{wc}$	$Chest_{woc}$	$Chest_{wc}$	$Abdomen_{mph}$	$Pelvic_{woc}$	$C\text{-}spine_{woc}$
USA (Kanal et al, 2017)	CTDIv	56		12	13			28
	DLP	962		443	469			562
Japan (Japan, 2015)	CTDIv	85		15	15			
	DLP	1350		550	550			
EU (European Commission, 2014b)	CTDIv	60		10	10	13-35		
	DLP	1000		400	400	460 - 1200	450-650	400-600
UK (Kanal et al, 2017)	CTDIv	60		12	12	14		28
	DLP	970		610	610	560		600
Australia (Kanal et al, 2017)	CTDIv	60		15	15			
	DLP	1000		450	450			
Korea (S.w. Yoon, 2018)	CTDIv	63.7			7.3	10.58		17.89
	DLP	1119.4			297.05	1511.41		434.04
Ireland (S J Foley et al, 2012)	CTDIv	58	66	9	9	13		19
	DLP	940	940	390	390	1120	570	420
Canada (Health Canada, 2016)	CTDIv	82		14	14			
	DLP	1302		521	521			
This study (2020)	CTDIv	53	51	17	15	14	15	38
	DLP	1237	1459	625	565	1821	605	1106

woc=without contrast, wc=with contrast, mph = multiphase

Based on the comparison of the DRL values of $CTDI_v$ for head with contrast material was below the DRLs values of the aforementioned countries in Table 5-4, however the DLP values were slightly above their DRLs values except for Japan and Canada. Addis Ababa's DRL for $CTDI_v$ and DLP values for chest without contrast was all above the value set by stated countries. The $CTDI_v$ and DLP values for chest with contrast material was above the DRL values of the aforementioned countries in Table 5-4 except for Japan and Australia. While the DLP values for chest with contrast materials was all above the values mentioned in Table 5-4. Comparing the $CTDI_v$ and DLP value of Abdomen four phasic dynamic CT protocols, the $CTDI_v$ value was above the value mentioned for Korea however it was below the values of EU and equal to the DRL value of UK. The DLP values of this study for abdomen four phasic dynamic CT was above all the countries mentioned in Table 5-4. When the DLP value of the study was compared for pelvic, the value was above the DRL values of Ireland and EU. The

c-spine value of both $CTDI_v$ and DLP of this study were above the values set by the aforementioned countries. This implies that, the machine operating parameters used by the scanners operating in Addis Ababa for some of the protocols were on the higher side compared to the aforementioned countries in Table 5-4.

5.1.3 Discussions

This study represents the collection of patient data related to CT procedures from Addis Ababa's health facilities patient data archive sources to establish regional DRL. It is clear that every increment of X-ray irradiation to a patient will experience certain risk (Mourão A, Aburjaile W, 2019). To manage the risk from medically irradiated patients, the development of DRL by the nation is indispensable. This is useful to identify optimal practice and to enhance professional awareness with respect radiology in patient dose administration (Kanal et al, 2017). When unusually high radiation dose is being delivered to the patient, DRLs is the best option as an alarm to take correction actions by reviewing DRLs values. Because DRL is a form of investigation level (Joseph et al, 2017) to assist patient dose management (Ruiz et al, 2016). On the contrary, under dose radiological procedure do not provide expected useful medical uses or do not yield expected diagnostic information to the patient (European Commission, 2014b). When the DRLs quantities are always surpassing through a population's quarter, it indicts that investigation is required. This work describes the first study in Addis Ababa - Ethiopia on adult patient's radiation dose related CT.

Wide dose variations were noticeable among health centers participated in this study for all anatomical area. The higher dose values perceived in definite anatomical CT procedures were conceivably due to multiple acquisitions of pre-contrast with simultaneous monitoring contrast scans and protocols techniques variation as well as application of dose minimizing software, even though ATCM software actually present in every CT that able to reduce the dose of radiation delivered to patient and keeping the noise constant during scanning. Hence, the local clinical radiation protection mechanisms should be improved and encouraged. Moreover, those CT procedures practiced in study area should be properly justified and the dose should be reduced.

Even, when the results obtained was compared with international DRL values, higher values have been recorded among CT scanners for some of the procedures in the study area that can be due to the variation to exam protocols and technical parameters in use.

Hence, further investigation may be required to ensure the optimization of patient-dose. To accomplish that, accurate dose reduction approaches should be implemented by the concerned bodies.

When the inter-centers comparison of the mean values (see Table 7-1) of the study, the proposed DRLs were exceeding in some specific sites. For instance, the $CTDI_v$ as well as DLP quantities were (25 mGy and 715 mGy.cm used for center G), (12 mGy and 420 mGy.cm used for center B), (11 mGy and 430 mGy.cm for center B), (9 mGy & 349 mGy.cm for center B) and (6 mGy and 425 mGy.cm for center E) for head (without and with), chest (without and with) and abdomen, respectively were lesser than the proposed DRL values. Conversely, (-29 mGy &-1439 mGy.cm for center D), (-13 mGy & -340 mGy.cm for center F) and (-15 mGy & -900 mGy.cm for center C) were above the proposed DRL values for head (without), pelvic and c-spine, correspondingly. This implies that different centers deliver varying degrees of patient doses for similar types of procedure. Therefore, the practice should be taken into consideration for further attention and study.

Centers with doses beyond and below the proposed DRLs are encouraged to perform serious research through an intention either towards conducting remedial act, or towards giving justification for the greater higher dose usage (Kanal et al, 2017). In addition, they need consistently bring protocols renew and review towards assuring compliance of ALARA principles. From Table 5-2, it is possible to say dose delivery with each center have great variation, this would be due to exam protocol related issues like using higher image quality requirements. This sight highlights sound to give special attention across CT patient examination doses, which requires actions to bring solutions for these inconsistencies.

Table V-5

Comparison of participated health facilities with difference from proposed DRLs values

Exam	Dose Descriptors	X=DRL- Mean								
		A	B	C	D	E	F	G	H	I
Head$_{wo}$	CTDIv	6	8	1	-29	4	-3	25	11	1
	DLP	85	344	-43	-1439	515	223	715	289	82
Head$_{w}$	CTDIv	6	12	1	1	4	-5	21	7	1
	DLP	57	420	-518	-518	490	62	382	168	71
Chest$_{w}$	CTDIv	7	11	11	10	5	-4	1	7	0
	DLP	231	430	325	304	290	-239	15	165	-65
Chest$_{w}$	CTDIv	3	9	5	7	6	-7	-2	5	1
	DLP	161	349	160	224	269	-318	-83	159	-7
Abdomen$_{mp}$	CTDIv	0	7	1	4	6	-6	3	0	3
	DLP	-44	407	43	319	425	-152	248	108	177
Pelvic$_{wo}$	CTDIv	-1	10	1	4	2	-13	1	4	-1
	DLP	1	422	46	167	172	-340	213	229	-41
C-spine$_{wo}$	CTDIv	4	26	-15	32	19	-3	7	27	22
	DLP	108	756	-900	779	700	-11	108	870	511

x=CTDI$_v$ or DLP, CTDI$_v$= volume CT dose index, DL=dose length product, wo=without contrast, w=with contrast and mp=multiphase abdomen

In Table 5-5, positive numeral indicated that the mean $CTDI_v$ and DLP values of the health centers lower than the DRL values of the corresponding dose descriptors. Likewise, the negative values indicated the mean value are higher than the DRL values of the corresponding dose descriptors. The higher the differences of the mean $CTDI_v$ and DLP values of the centers is informing that the local clinical practice of the department should get greater attention to investigate their examination protocols as to optimize dose. Zero value indicate the mean of that value equal to the proposed DRL.

5.1.4 Limitations

This PhD work was based on retrospective data collected from adult patient data archives, of the participated health institutes. Pediatric patients were not included in this study. There were some restrictions that impacts the accuracy of the estimated DRL values of the study. For instance, the data was collected for five months, however, longer time may be required to obtain sufficient data and strengthen the outcome of the research. Covid-19 pandemic was another cause for the limited data collection. Another sources of limitation was the lack of sufficient information regarding clinical indicators for CT, which can lead individual patient dose variations in the same data set because of the technical settings variations related with unlike image quality requirements (S.w. Yoon, 2018). Automatic exposure control and DLP have been influenced by patient weight and height, which were not included in this study. The essential limitation of this research was that a combination of different CT scanners

(GE, Siemens, Philips and Neusoft) were used, this is because of lucking enough number of similar brand of CT scanners distribution in the study area.

Another limitation was on the examination frequency in some centers, which means most exam protocols is not practiced regularly. In addition to the aforementioned limitations, the collected patient data were included data from both new and old CT technology units.

5.1.5 Conclusion

The descriptive statistical dose distribution values for each of the selected anatomical regions has been computed from large amount of data collected from diagnostic facilities in Addis Ababa. Hence, the first retrospective data based local DRLs for selected CT examination protocols has been developed for adult patients. The measurable metrics used in DRLs developments were $CTDI_v$ and DLP at their 75^{th} percentiles of mean/median dose distribution. DRLs inspire radiation protection via optimization of dose to the patient because it gives signals to the health facilities how much radiation doses being used for diagnosis. Wide measured radiation dose variations were noticed among health centers participated in this study. The high dose values perceived in definite anatomical CT-procedures could possibly due to variations of CT brand, protocol and dose reduction software. Hence, local dose optimization according to the proposed DRL should be practiced and encouraged via re-audit their examination protocols, developing quality assurance program and developing clinical audits culture among professionals to minimize unjustified CT procedures. In conclusion, the proposed DRLs for examinations of abdomen, chest, head, pelvic and C-spine can be definitely recognized. It can be also used as a spring board to expand this work in other regions of the country which cumulatively used to adhere all other CT protocols to a national DRLs frame.

5.2 Phase – 2 – Results

The descriptive statistics of dose results has been detailed in Table 5-6 for the studied examinations. The proposed DRLs were also compared with researcher's previous work on CT- DRL (phase – 1 research results) proposed for adult patients in Addis Ababa (Table 5-10) and international recommendations as tabulated in Table 5-11.

The estimation of CT tube output potentials which is expressed as CTDI was computed based on the details given in section 4.9.1. In that detail, $CTDI_v$ signifies the x, y and z axis average absorbed radiation dose at the exposure position of the phantom for a single slice of image at single CT tube rotation, however, measured DLP represents the overall absorbed radiation dose over the 100 mm pencil ionization chamber integration limit based on the scan protocols and parameters. Based on the prospective data collected from 9 health institutions, the mean, standard deviation, minimum, maximum, 90^{th}, 75^{th}, 50^{th}, 25^{th}, coefficient of variation and 95% confidence intervals were calculated for $CTDI_v$ values obtained from broad ranges experimental data analysis using adult head and body PMMA phantoms. Results are presented in Table 5-6 for head and body phantoms measurements. Since there is differences among CT scanners besides exam procedures, the results shown substantial divergences in measured radiation dose values among different brands of CT scan.

All measurements were conducted at brain and chest scan protocols for head and body phantom examinations, respectively, to mimic clinical conditions of adult patients. As detailed in Table 5-6, the typical tube current for head and body phantom examination ranges from 150 – 314 mA and 180 – 314 mA for Philips CT scanners, respectively; while for Siemens scanners was on the average value of 222 – 270 mAs and 222 – 250 mAs for head and body phantoms respectively. For GE CT scanners, the tube current for head and body phantom examination ranges from 150 – 250 mA and 180 – 250 mA, respectively. The representative voltage for head and body phantom examinations ranges between 100 and 130 kVp for Siemens CT scanners. For Philips and GE scanners, the tube voltage used was 120 kVp which has better agreement with the measured data values.

Table V-6

Average exposure parameters for Philips, Siemens and GE CT scanners used in this study

Brand	Phantom Types	Scan Type	mA/mAs	kVp (v)	Scan time(s)	T (mm)	No. images	Scan L(mm)	Feed (mm)	NT (mm
Philips	Head	Axial	150 – 314	120	1.104	2	8	100	10	13
	Body	Axial	180 – 314	120	0.923	5	2	100	10	13
Siemens	Head	Axial	220 – 270	110-130*	1.473	5	2	100	10	21
	Body	Axial	220 – 250	100-110*	5.008	3	3	100	10	8
GE	Head	Axial	150 – 250	120	0.868	4	4	100	10	14
	Body	Axial	180 – 250	120	0.713	4	9	100	10	25

*T_feed=table feed, Aver.=average, T= slice thickness, *=effective tube current in mAs for Siemens CT scanners*

5.2.1 Phantom Dose Analysis

As tabulated in Table 5-7, the mean values of $CTDI_v$ of head and body phantoms for Philips CT scanners were 43, 134 mGy with STD values of 14 and 0.4 mGy, respectively. For Siemens CT scanners, estimated mean of $CTDI_v$ measurements for head phantom were 26 mGy with STD value of 10 mGy, and for the body phantom dose measurement values were 13 mGy with STD values of 1 mGy. The mean of $CTDI_v$ measurements taken in the head phantom for GE CT scanners were 36 mGy with STD of 11 mGy, and the body phantom dose measurement values were 1 mGy with STD of 1 mGy. The general scan protocols and adjusted scan parameters for Philips, Siemens and GE CT scanners were almost similar, however deviation has been realized among the brands of CT scanners assessed for $CTDI_v$ and DLP dose distribution.

Table 5-7 also shown that the estimated minimum values of $CTDI_v$ for Philips, Siemens, and GE CT scanners were (29, 15, 23 mGy), for head examinations, respectively, similarly for body phantom examinations were (13, 13, 15 mGy), for Philips, Siemens, and GE CT scanners, respectively. The estimated maximum values of $CTDI_v$ for Philips, Siemens, and GE CT scanners were (57, 32, 43 mGy) for head examinations, correspondingly; whereas for body phantom examinations were (14, 14, 16 mGy) for Philips, Siemens, and GE CT scanners, respectively. Hence, 48% of $CTDI_v$ deviation was realized among the estimated minimum values of head exams of different CT machines. While body exam shown 13% deviation for $CTDI_v$ and DLP among the minimum values of the different CT machines. The estimated maximum values also shown 44% of $CTDI_v$ of head examinations and among involved CT machines in this study. The 13% deviation was seen for $CTDI_v$ dose distributions

among the maximum values of body phantom examinations the different participated CT machines.

According to Table 5-7, the median values (50th percentile) of $CTDI_v$ for Philips, Siemens and GE CT scanners for head phantom were (43, 31, 41 mGy) with deviation of 27.9% among the scanners, respectively, whereas the median values of $CTDI_v$ for Philips, Siemens and GE scanners for body phantoms were (13, 13, 16 mGy) with deviation 12.5% among the scanners, respectively.

Table V-7

Summarized data of mean, STD, minimum, 25th, 50th, 75th, 90th percentiles, and maximum as well as coefficient of variation (CV) and 95% confidence intervals for the 75th percentile of $CTDI_v$ per sequence and total DLP per examination

Machine Brand	Phantom Type	CTDI	Estimated CTDI Values for phantom Test								For 75th perc	
			Mean	STD	Min.	25th perc.	50th perc.	75th perc.	90th perc.	Max.	CV	95% CI
Philips	head	CTDIv	43	14	29	36	43	50	54	57	0.33	50±4
	Body	CTDIv	13	0.4	13	13	13	13	14	14	0.03	13±1
Siemens	head	CTDIv	26	10	15	23	31	31	32	32	0.37	31±4
	Body	CTDIv	13	1	13	13	13	14	14	14	0.05	14±1
GE	head	CTDIv	36	11	23	32	41	42	43	43	0.31	42±4
	Body	CTDIv	16	1	15	15	16	16	16	16	0.04	16±1

Based on Table 5-7, the corresponding 90th percentile values for the $CTDI_v$ for head phantom examination were 54 mGy; 32 mGy and 43 mGy for Philips, Siemens and GE CT scanners with deviation of 40.7% of $CTDI_v$ among the scanners. Similarly, the equivalent 90th percentile values for the $CTDI_v$ for body phantom examination were 14 mGy and 14 mGy and 16 mGy aimed at Philips, Siemens and GE CT scanners with deviation of 12.5% of $CTDI_v$ among the scanners.

The estimated 25th percentile values for $CTDI_v$ for head and body phantoms were (36, 23 and 32 mGy) and (13, 13, and 15 mGy) for Philips, Siemens and GE scanners respectively, as detailed in Table 5-7.

Table 5-7 also revealed that the 75th percentile values for head phantom were (50, 31, 42 mGy) of $CTDI_v$ and (499, 314, 420 mGy.cm) of DLP with 95% confidence interval values of (50±4, 31±4, 42±4 mGy) of $CTDI_v$ and (499±14, 314±11, 420±12 mGy.cm) of DLP for Philips, Siemens and GE scanners, respectively. The 75th percentile quantities for body phantom were existed for CTDIv and DLP as (13, 14, 16 mGy) and (135, 136, 159 mGy.cm) for Philips, Siemens and GE CT scanners with 95%

confidence interval quantities of (13±1, 14±1, 16±1 mGy) and (135±2, 136±3, 159±3 mGy.cm) for Philips, Siemens and GE CT, respectively.

The coefficient of variation among the CT scanners were (0.33, 0.37 and 0.31) and (0.03, 0.05 and 0.04) measured for head and body phantoms, respectively.

5.2.2 CT dose from Control Console Analysis

The estimated dose data obtained from CT control console for head and body examination protocols in this research was analyzed and the estimated results displayed in Table 5-8. As tabulated in Table 5-8, the mean, STD, minimum, maximum, 90th, 75th, 50th and 25th percentiles as well as coefficient of variation (CV) and 95% confidence intervals for the 75th percentile of CTDI values for Philips, Siemens and GE Ct scanners were presented for head and body exposures protocols. The general scan protocols and adjusted scan parameters for Philips, Siemens and GE CT scanners were almost similar, however deviation has been realized among the brands of CT scanners assessed for $CTDI_v$ dose distribution.

Table V-8

Summarized data analysis obtained from CT control console

Machine Brand	Phantom Type	CTDI	Estimated CTDI values obtained from CT control console								For DRLs	
			Mean	STD	Mini	25th Perc	50th Perc	75th Perc	90th Perc	Maxi	CV	95% CI
Philips	Head	CTDIv	55	18	40	45	50	63	70	75	0.33	63±5
	Body	CTDIv	50	22	31	37	44	59	69	75	0.45	59±5
Siemens	Head	CTDIv	38	11	29	32	35	43	48	51	0.30	43±4
	Body	CTDIv	25	7	19	22	25	27	29	30	0.30	27±3
GE	Head	CTDIv	42	13	29	37	44	49	52	54	0.30	49±4
	Body	CTDIv	19	7	15	15	15	21	25	27	0.36	21±3

Table 5-7 also shown that the median values (50th percentile) of $CTDI_v$ for Philips, Siemens and GE CT scanners for head phantom were (50, 35, 44 mGy) with deviation of 30% among the scanners, respectively, whereas the median values of $CTDI_v$ for Philips, Siemens and GE scanners for body phantoms were (44, 25, 15 mGy) with deviation 65.9% among the scanners, respectively.

Table 5-7 also revealed that the corresponding 90th percentile values for the $CTDI_v$ for head phantom examination were 70 mGy; 48 mGy; and 52 mGy for Philips, Siemens and GE CT scanners with deviation of 31.4% among the scanners. Similarly, the equivalent 90th percentile values for the $CTDI_v$ for body phantom examination were 69 mGy; 29 mGy; and 25 mGy for Philips, Siemens and GE CT scanners with deviation of 63.8% among the scanners.

As detailed in Table 5-8, the estimated 25th percentile values for $CTDI_v$ for head and body phantoms were (45, 32, and 37 mGy) and (37, 22, and 15 mGy) for Philips, Siemens and GE scanners, respectively.

According to Table 5-7, the 75th percentile CTDIv values for head and body phantoms were (63, 43, 49 mGy) with 95% confidence interval values of (65±5, 43±4, 49±4 mGy) for Philips, Siemens and GE scanners, respectively. The 75th percentile quantities of $CTDI_v$ for body phantom recorded as (59, 27, 21 mGy) for Philips, Siemens and GE CT scanners with 95% confidence interval quantities of (59±5, 27±3, 21±3 mGy) for Philips, Siemens and GE CT, respectively.

5.2.3 CTDI Comparison

Table 5-9 presented the 75th percentile CT-tube output dose distribution ($CTDI_v$ and DLP) and percentage variation between data from control console and experimental (phantom and chamber based data) for Philips, Siemens and GE CT scanners.

As tabulated in Table 5-9, the variation was clear and need professional decision. The researcher understands that developing DRLs based on the data obtained from phantom measurement will give lower dose references than CT doses from control console.

Table V-9

The 75th percentile CT-tube output dose distribution and percentage variation between data's from control console and experimental for Philips, Siemens and GE CT scanners

Machine Brand	Phantom Type	CTDI	75th Percentile		Percentage variation
			control console	Experimental	
Philips	Head	$CTDI_v$	63	50	20.1
	Body	$CTDI_v$	59	17	71.3
Siemens	Head	$CTDI_v$	43	31	27.8
	Body	$CTDI_v$	27	16	40.9
GE	Head	$CTDI_v$	49	42	14.7
	Body	$CTDI_v$	21	16	24.5

This is because the best method of optimizing patient's effective radiation dose delivery are applying automatic current modulation, CT-post-processing and lowering CT-tube potential (Costello J & Tucker J, 2013). Hence, these methods are the important tools for lowering CT radiation exposure. Then, considering experimental data (CTDI data obtained from chamber and phantom measurements) is the better tool to obtain lower values of DRLs for the selected CT protocols conducted in Addis Ababa hospitals. Therefore, the DRL values and achievable dose (AD) that can be used

to optimize the patient dose were presented in Table 5-10 for Philips, Siemens and GE machine brands.

Table V-10

The proposed DRLs for selected CT-protocols

Brands	DRLs at 75th percentile		AD at 50th percentile	
	Head	Body	Head	Body
	CTDIv	CTDIv	CTDIv	CTDIv
Philips	50	17	43	13
Siemens	31	16	31	13
GE	42	16	41	16
Average	**41**	**16**	**38**	**14**

5.2.4 Recent and previous Data Comparison

The estimated dose for head and body phantoms in this research was compared with the previous research by in Addis Ababa, Ethiopia, which was based on the retrospective data obtained from CT system control displayed doses that is grounded on the theoretical estimates from internationally accepted software fitted with CT scanners. The respective updated proposed DRL and Achievable dose at 75th percentile and 50th percentile is tabulated in Table 5-10 based on the results obtained from the experimental measurements of adult patient mimicked PMMA phantoms. The previous research proposed DRL values of $CTDI_v$ and DLP recorded as 52 mGy and 1257 mGy.cm for head and 17 mGy and 600 mGy.cm for chest, respectively that were calculated at 75th percentile mean dose distribution. However, the recent study computed DRLs for $CTDI_v$ were 41 mGy and 16 mGy for head and chest, respectively, which were also computed at respective 75th percentile mean dose distribution. When the post DRL proposal was compared with the researcher's previous DRL, there existed with CTDIv numerical magnitudes variations of 26.83% and 6.25%, as tabulated in Table 5-11. All recent DRLs values were less than the previous DRLs values. Since the aim of developing DRL is to optimize patient protection at the lowest possible radiation dose without jeopardizing the intended diagnostic information. Therefore, the researcher recommended the concerned bodies to use the updated research DRL (41 mGy and 16 mGy for head and chest exposure) proposal at 75th percentile of the mean $CTDI_v$ dose distribution for head and chest. This DRLs were estimated based on the prospective Data. It was related to the actual radiation dose measurements of PMMA head and body phantoms that characterized (mimicked) the adult patient.

Table V-11

Summarized proposed DRL based on retrospective and prospective data analysis for the 75th percentile for CTDIv and DLP for head and body

Phantom Type	Previous DRL(phase -1 result)	Recent DRL (phase 2)	Variation (%)
	$CTDI_v$ (mGy)	**$CTDI_v$ (mGy)**	**$CTDI_v$**
Head	52	41	-26.83
Chest*	17	16	-6.25

*The chest DRL values of the recent study was based on the body phantom exposure result

5.2.5 Comparison with International Data

Comparing the results of this study in the international context (Table 5-12), the CTDIv values for head was lesser in magnitude than the recently published DRLs of EU, USA, Japan, UK and Turkey, resulting in an underestimation of 46.3%, 36.6%, 107.3%, 46.3% and 62.0% respectively across the respective countries. For chest protocol DRLs values of CTDIv, the results of this study were all greater than the recently published DRL values of EU, USA, Japan, UK and Turkey, causing an overestimation of 37.5%, 25%, 6.3%, 25% and 17.5%, respectively across the countries, respectively. These great estimated results variation of this study compared with the international values will be due to variation of exposure factors which is highly related with equipment features as well as patient body sizes and weights. It is clear that diagnostic reference level can be able to developed for standard man or phantom depending on certainly quantifiable as well as reproducible radiation dosage parameters (Japan, 2015) for broadly defined types of equipment (Paulo, 2015). The amounts of radiation dose which differ with the patient size and scanned body part (ICRP, 2000b). But, the standard man was developed based on the population distribution of developed countries. Individual examinations of sampled patients from Addis Ababa (which is grouped in developing country). Patient thickness in developed and developing country countries can vary enormously because of the great differences of living standards which bring great variation in patient size and weight. The differences of patient size and weight for the same protocol can bring differences of examination parameters. That means, fatty patient (mostly seen in developed countries) exposed to higher radiation compared with thinner patient (mostly seen in underdeveloped countries). Therefore, this variation in DRLs is expected even if the same area of anatomy and clinical indications (i.e. examination protocols) have been considered in the research.

Table V-12

Comparison of DRLs for other European records with this study

Study	CTDIv measured at 75th percentile						
	IAEA CRPR(26)	EU (25)	USA (5)	Japan (22)	UK (5)	Turkey (30	This study
Head	47	60	56	85	60	66.4	41
Chest*	9.5	10	12	15	12	11.6	16

5.2.6 Discussions

This study represents the collection of prospective CT dose data related to CT procedures from Addis Ababa's health facilities patient data archive sources to establish regional DRL. It is clear that every increment of X-ray irradiation to a patient will experience certain risk (Mourão A, Aburjaile W, 2019). To manage the risk from medically irradiated patients, the development of DRL by the nation is indispensable. This is useful to identify optimal practice and to enhance professional awareness with respect radiology in patient dose administration (Kanal et al, 2017). When unusually high radiation dose is being delivered to the patient, DRLs is the best option as an alarm to take correction actions by reviewing DRLs values. Because DRL is a form of investigation level (Joseph et al, 2017) to assist patient dose management (Ruiz et al, 2016). On the contrary, under dose radiological procedure do not provide expected useful medical uses or do not yield expected diagnostic information to the patient (European Commission, 2014b). This work describes the first study in Addis Ababa - Ethiopia on adult patient's radiation dose related CT.

Wide dose variations were noticeable among health centers participated in this study area for all anatomical area. The higher dose values perceived in definite anatomical CT procedures were conceivably due to multiple acquisitions of pre-contrast with simultaneous monitoring contrast scans. The technique of protocols variation and use of dose reduction software, even though software of ATCM existed in every CT that be able to reduce the dose of radiation delivered to patient and keeping the noise constant during scanning. Hence, the local clinical radiation protection mechanisms should be improved and encouraged according to the CT technology being used in medical practices. Moreover, those CT procedures practiced in study area should be properly justified and the dose should be reduced as low as reasonably achievable.

Even, when the results obtained was compared with international DRL values, higher values have been recorded among CT scanners for some of the procedures in the study area that can be due to the variation to exam protocols and technical parameters in use.

Hence, further investigation may be required to ensure the optimization of patient-dose in the research area. To accomplish that, accurate dose reduction approaches should be implemented by the concerned bodies.

When the inter-centers comparison of the mean values (Table 5-13) of the study, some specific sites were identified exceeding of the proposed DRLs. For instance, by ignoring the differences among CT models of the study, the variation of $CTDI_v$ from the proposed DRLs were varied from 0 to 7 mGy for head and body CT-scanning, respectively. This implies that different centers deliver varying degrees of patient doses for similar types of procedure. Therefore, the practice should be taken into consideration for further attention for optimization and investigation.

Centers with doses beyond and below the proposed DRLs are encouraged to perform serious research through remedial acts or for justifying the application of unusually greater doses (Kanal et al, 2017). In addition, they need consistently bring renew and evaluate their protocols towards assuring the agreement with the principles of ALARA. From Table 5-7, it is possible to say that the general scan protocols and adjusted scan parameters for Philips, Siemens and GE CT scanners were almost similar in each scanner, however deviation has been realized among the brands of CT scanners assessed for $CTDI_v$ dose distribution. This would be due to exam protocol related issues like using higher image quality requirements. This sight highlights sound to give special attention across CT patient examination doses, which requires actions to bring solutions for these inconsistencies.

Table V-13

Inter-center comparison of participated health facilities with difference from proposed DRLs values

Machine Brand	Phantom Type	CTDI	X = DRL - mean
Philips	Head	$CTDI_v$	7
	Body	$CTDI_v$	4
Siemens	Head	$CTDI_v$	5
	Body	$CTDI_v$	3
GE	Head	$CTDI_v$	6
	Body	$CTDI_v$	0

x=$CTDI_v$, wo=without contrast, w=with contrast and mp=multiphase abdomen

In Table 5-13, positive numeral indicated that the mean $CTDI_v$ values of the health centers lower than the DRL values of the corresponding dose descriptors. Likewise, the zero values indicated the mean values CTDIv are equivalent to the corresponding

dose descriptors. The higher the differences of the mean $CTDI_v$ values of the centers is informing that the local clinical practice of the department should get greater attention to investigate their examination protocols as to optimize dose.

5.2.7 Limitation

DRLs is mostly recommended for representative standard phantoms or patients (typically 70 kg) which does not represent all patients. Individual patient dose whose body size vary from the size of standard phantom may be under/over-estimated. For instance, larger number of patients may not meet the standards, hence there may be discrepancy between the present DRLs and internationally recommended standards. Additionally, DRLs could not apply for individual patients.

Another limitation is that the difference between the real patient size and short ionization chamber (100 mm) (Rehani, 2015) that creates limitation on the clinical relevance of DRL developed based on the standard phantom and ionization chamber. Because the $CTDI_{100}$ report from ion chamber may be insufficiently long to record the contribution of scatter radiations (American Association of Physicists in Medicine, 2010). That means phantom and chamber based DRLs systematically underestimate the cumulative mean dose distributions used.

The user of this DRL should recognize that DRLs cannot be de facto dose limit that should not be exceeded. Bigger body patients may receive higher radiation dose than the DRL values to achieve satisfactory images quality. This research was not considered for the helical mode scanning. But, unforgettable health institutions are using the helical mode of scanning in the region. This research data collection was not used barometers and thermometer, which help the chamber to calibrate itself to the present pressure and thermometer. Therefore, the researcher recommended that the coming researcher should develop methods of resolving all the aforementioned limitations of the research results.

5.2.8 Conclusion

DRLs are definitely the best instrument for continuously optimizing the medical examinations set at 75^{th} percentile of the median dose distribution at a healthcare facility with ultimate goal of obtaining excellent image quality with a results of adequate diagnostic information. The DRLs process makes possible the optimization of patient protection by identifying equipment and examinations for which radiation

dose levels are very high or very low. The 75th percentile $CTDI_v$ dose distribution values of the experimental data based outcome of this research can be considered as the best possible DRLs values compared with the previous DRLs proposed based on the retrospective data for Addis Ababa CT diagnosis procedures of the specified examination protocols. DRLs inspire radiation protection because it gives signals to the health facilities how much radiation doses being used for diagnosis. In this study, wide measured radiation dose variations were noticed among health centers participated. The high dose values perceived in definite anatomical CT-procedures could possibly due to variations of CT brand, protocol, scan length, dose reduction software and exposed sample. In conclusion, the proposed DRLs for examinations can be used as a spring board to expand this work in other regions of the country which cumulatively used to adhere all CT protocols to a national DRLs frame.

5.3 Phase – 3 – Results

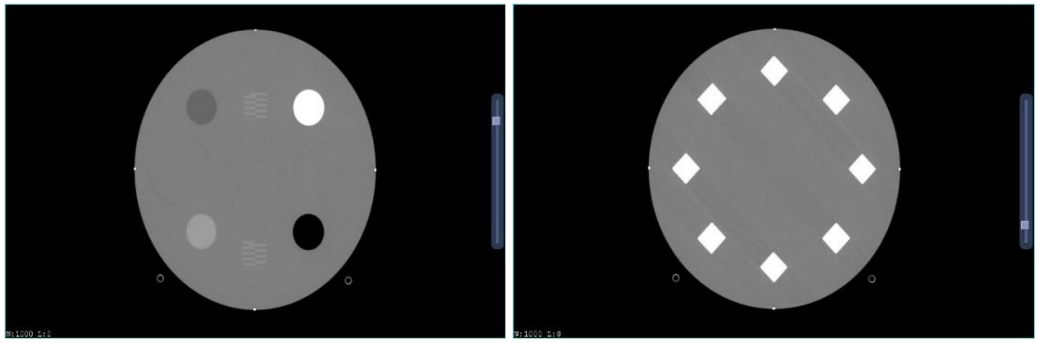
The CT image quality were measured using homogeneous ACR CT phantom according to the quality assurance control recommendations provided by AAPM, IEC and IPEM. ICRU report 54 recommended that statistical decision theories must be implemented in clinical imaging. Hence, imaging performance relied on different parameters such as (a) image contrast, sharpness and noise; (b) the diagnosis nature and patient illness complexity; and (c) the capability of image scientists (radiologists) to understand the image information. The diagnostic consequence is highly related with the capability of the image scientists towards comprehending the required medical image information. The image scientist observes and read the diagnostic medical image subjectively using his eye (Lee et al., 2021). He generates subjective decision based on his understanding. This will lead to aggravate the error frequency of medical diagnosis and varying medical outcome may be recorded for the same signals. Anyhow, the results of this study were illustrated for each HRC and abdomen protocols for each image quality parameters as follows.

5.3.1 Beam Alignment

Figure 5-4 revealed that the CT machine had good alignment at the time of image quality assurance, because the central wires were clearly brightly both top and bottom ramps in module 1 as well as all BBs were brightly seen at 3, 6, 9 and 12 o'clock. The image orientation was also good.

Figure V-4

Assessment of positioning and beam alignment by using the 1 mm diameter steel BBs positioned at the 3, 6, 9 and 12 o'clock on module 1 and module 4 ACR CT accreditation phantom

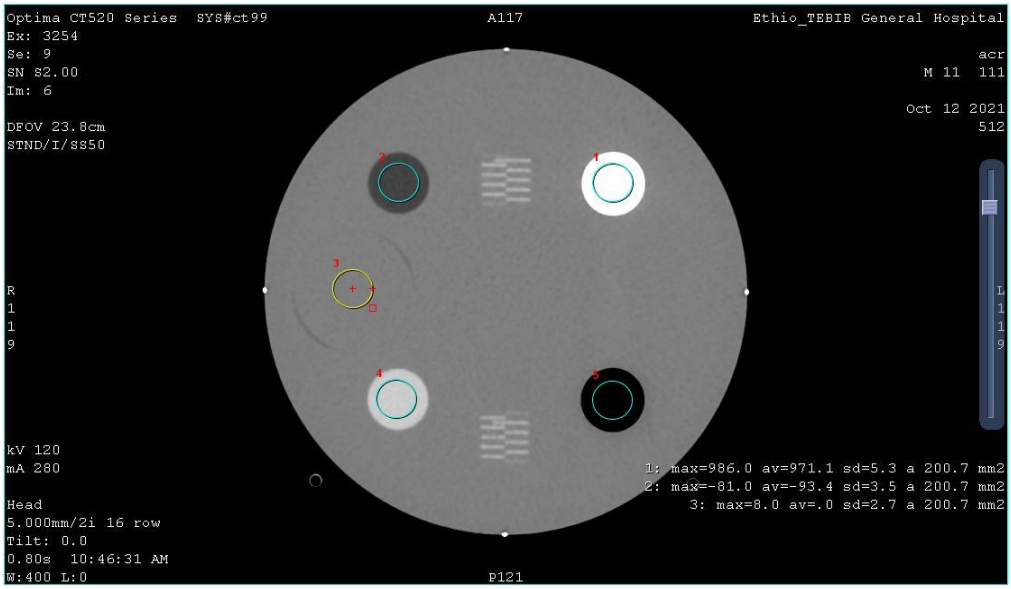


5.3.2 CT Number Linearity

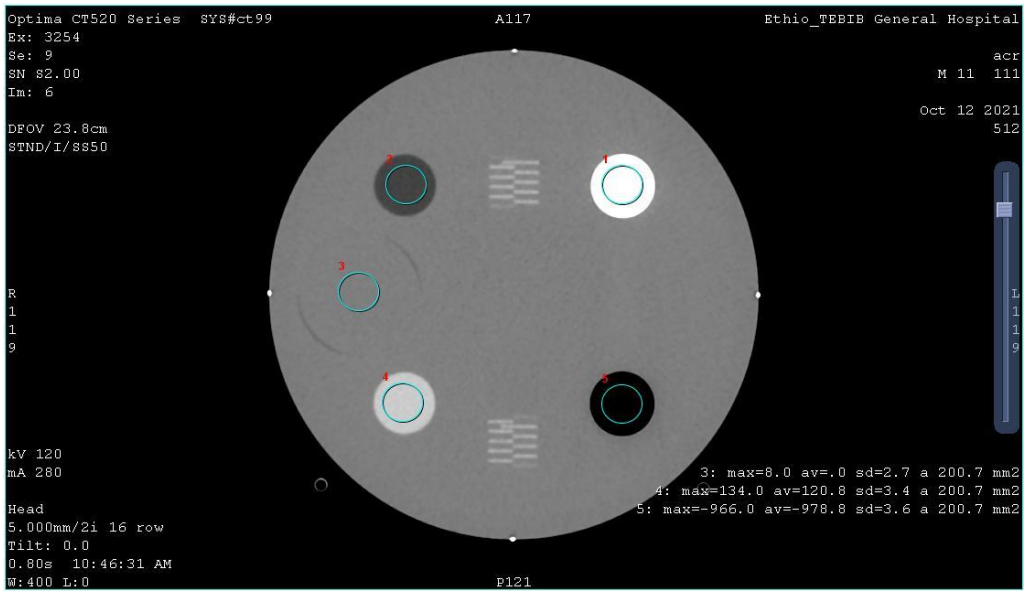
The CT number values were measured for each HRC and abdomen protocols as shown in Figure 5-5. The mean CT number values in the ROI for air, acrylic, water, polyethylene and polyamides were given as displayed on the Figure 5-5. The values for each ROI were within the recommended values and highly correlated with the electron density values of each material.

Figure V-5

Illustrates beam alignment and CT number calibration (Image source: researcher archive)



(a)



(b)

As described in Figure 5-5 (a) and (b), the average CT number for, air, acrylic, water, polyethylene and bone were tabulated in Table 5-14. As the values revealed that all recorded quantities were within the acceptable limit. The materials CT number correlation is also equal to 0.9991 as depicted in Figure 5-6.

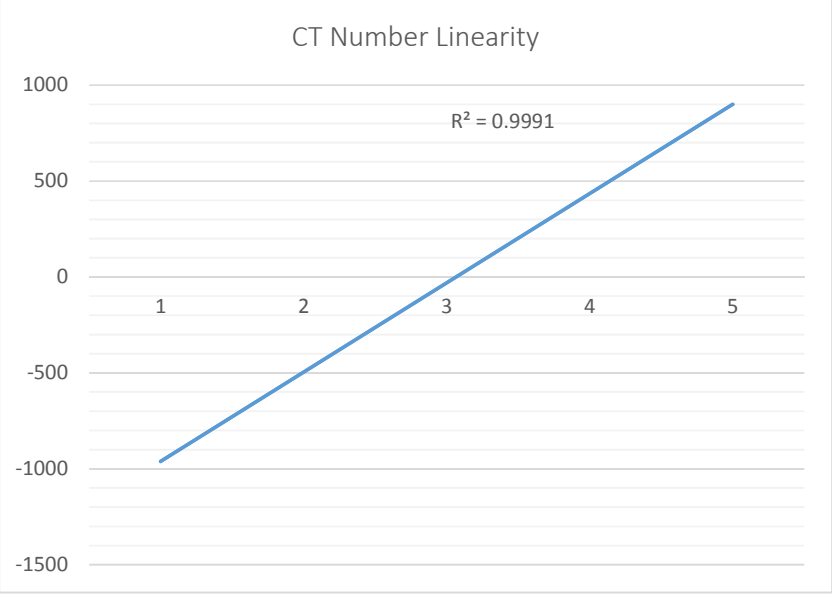
Table V-14

Average CT number values for each material in the ROI

Material	Mean CT# tolerance (HU)	Results obtained (HU)
Air	-1005 to -979	-978.8
Acrylic	+110 to +130	120.8
Water	-7 to +7 (0±5)	0
Polyethylene	-107 to -87	-93.4
Bone	+850 to +970	971.1

Figure V-6

CT number and electron density correlation



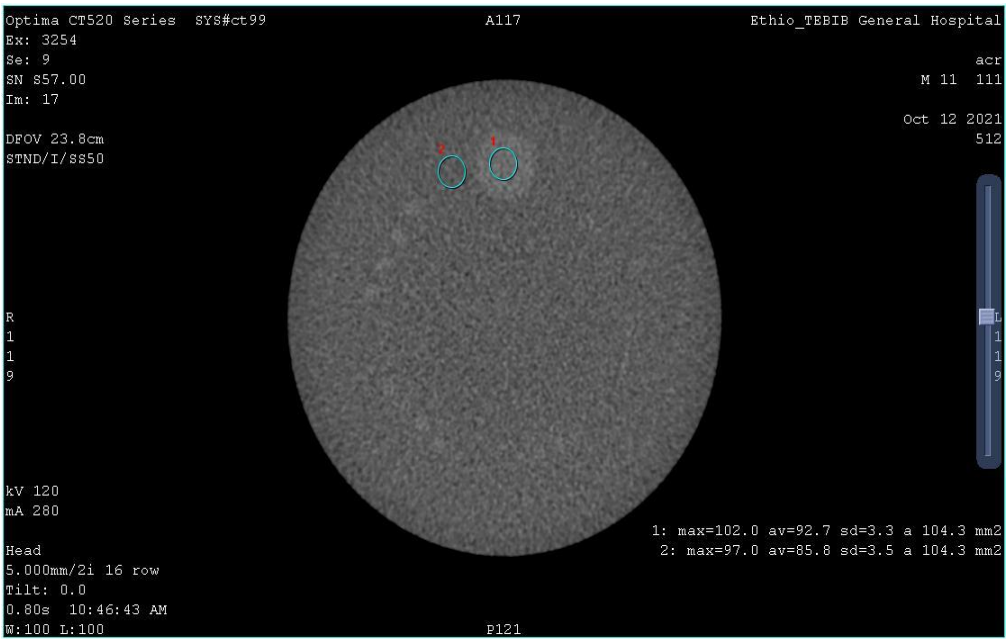
The slice thickness measured according the recommendation given by the ACR CT accreditation phantom for the image analysis was within the acceptable limit. Because, for MDCT scanners which have 64 detector rows may allow a total acquisition width of 32 to 40 mm that is quantified at the isocenter. This kind of image acquisition able to generate slice width changing from 0.5 – 10 mm (IAEA, 2012).

5.3.3 Low Contrast Detectability

Adult head protocol was used to assess the low contrast detectability, see Figure 5-7. Good low contrast resolution was seen as the 6 mm and 5 mm diameter cylinder groups were clearly visible. The figure revealed that there were no shade artifacts and annotation covering. Hence, the machine capability of diagnosing low contrast materials of the patient illness was adequate at the time of assessment.

Figure V-7

Low contrast resolution assessment by using the four cylinder groups. As figure exposed, the 6 mm and 5 mm diameter cylinder groups were clearly visible (Image source: Researcher archive)



The measurement of low contrast resolution on an axial head ACR CT accreditation phantom by using the CNR equation delivered the values of 2.0, as calculated below:

$$CNR = \frac{|92.7 - 85.8|}{\left(\frac{|3.3 + 3.5|}{2}\right)} = \frac{6.9}{3.4} = 2.0$$

5.3.4 CT Number Uniformity and Noise

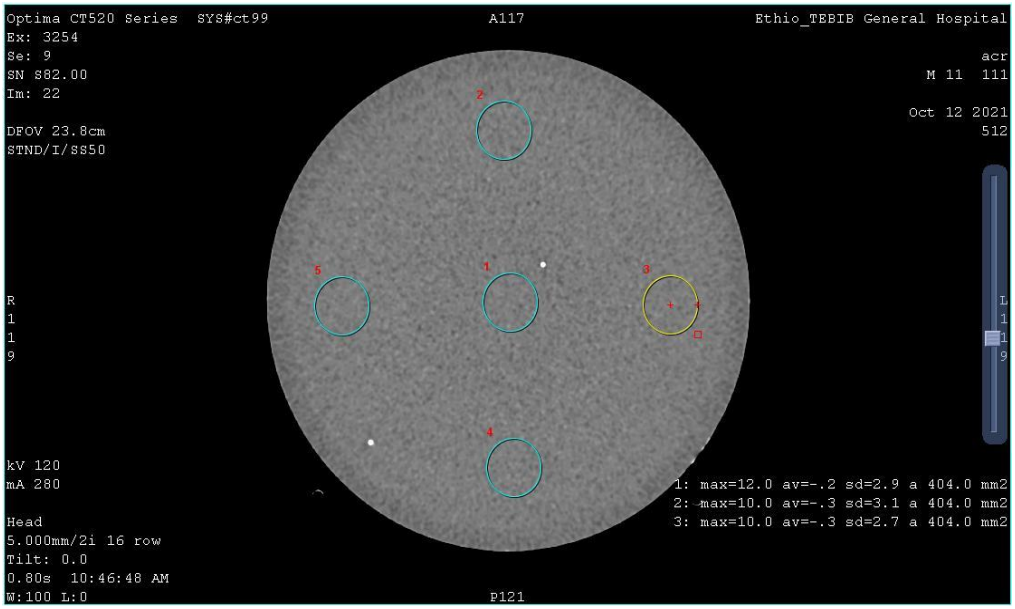
The measurement of CT number at the center and the four edges (i.e. at 12:00, 3:00, 6:00 and 9:00) were -0.2, -0.3, - 0.3, 0.1 and – 0.2 HU, respectively, as shown in Figure

5-8. The center standard deviation value was 2.9/2.7 HU, which should be between -7 and +7 HU (or 0±5) (McCollough et al, 2004). Therefore, the noise value (which is described by STD values of measurements) of image at the time of assessments was within the tolerance value.

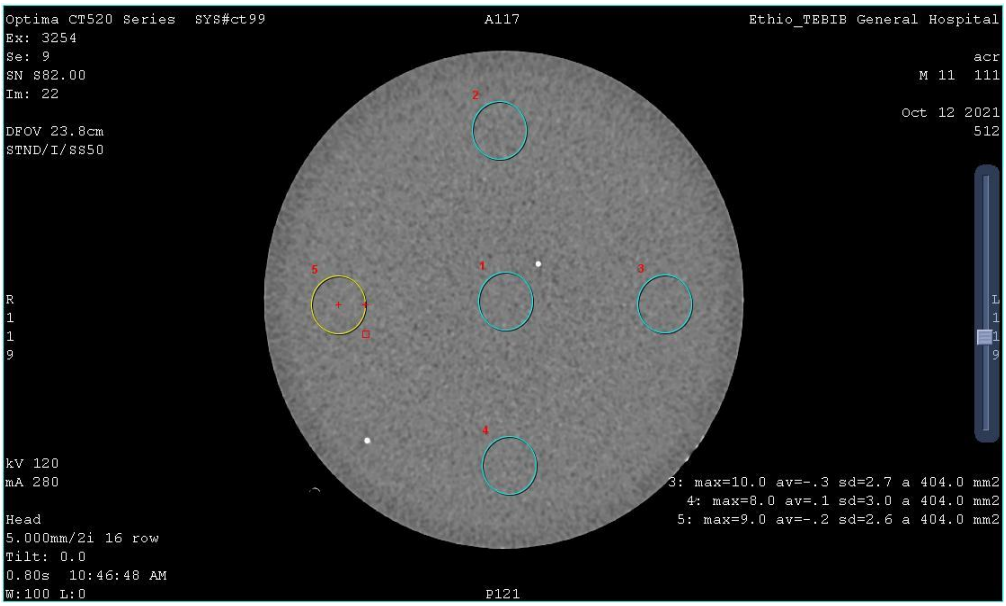
The CT number uniformity value was calculated by computing the absolute value for the center mean CT number minus the edge mean CT number. The result was 0.85 HU that was below 5 HU, which is acceptable value, see Table 4-5 (mentioned in chapter 4 in section 4-11 in sub-section 4.11.1).

Figure V-8

Position selection of center and edges ROI for noise and CT number uniformity measurement. The center ROI used for noise measurement, and center and edge ROIs used for CT number uniformity measurement (Image source: Researcher archive)



(a)

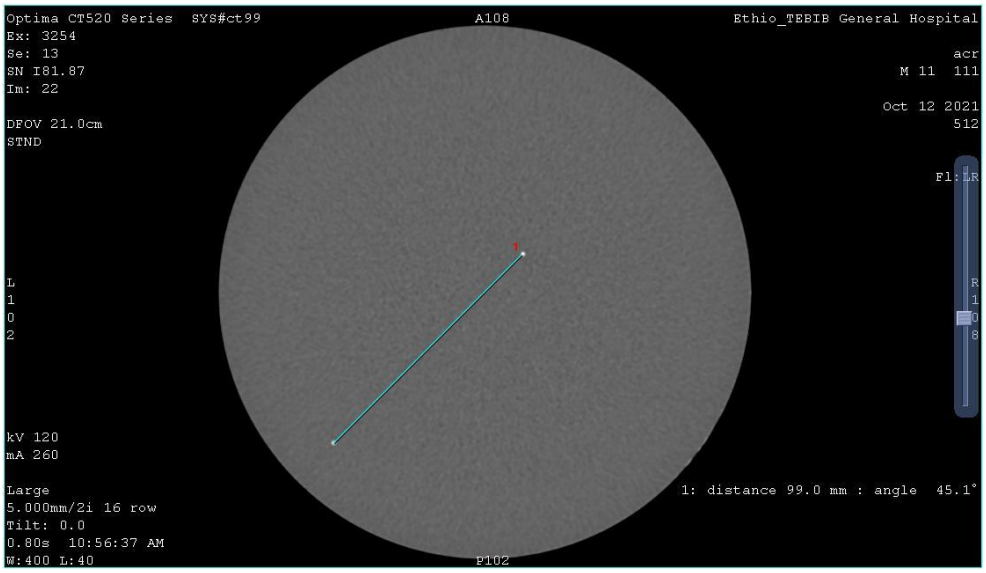


(b)

The in-plane distance accuracy or section sensitivity profile assessments can be done through accurate measuring of the distance of the two very small 0.28 mm steel BBs, see Figure 5-9. The measurement of in-plane distance accuracy for the image in Figure 5-9 shown 99.0 mm which is in acceptable range.

Figure V-9

Illustrates an image of module 3 in-plane distance accuracy and section sensitivity profile assessment were also conducted using the two very small 0.28 mm BBs

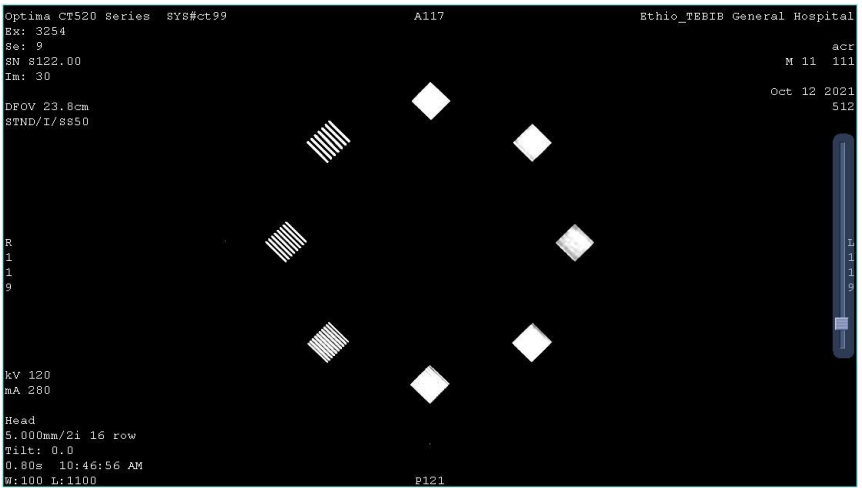


5.3.5 High Contrast Resolution

Towards showing the high contrast frequencies of eight aluminum bar patterns, the HRC protocol was used. Figure 5-10 illustrates good high contrast spatial resolution image with 10 lp/cm at 6th aluminum bar patterns in module 4 of the ACR phantom analysis for HRC protocol. The tolerance level is 5 lp/cm. This implies that the CT scanner is working properly by delivering the images of high contrast materials with adequate diagnostic information.

Figure V-10

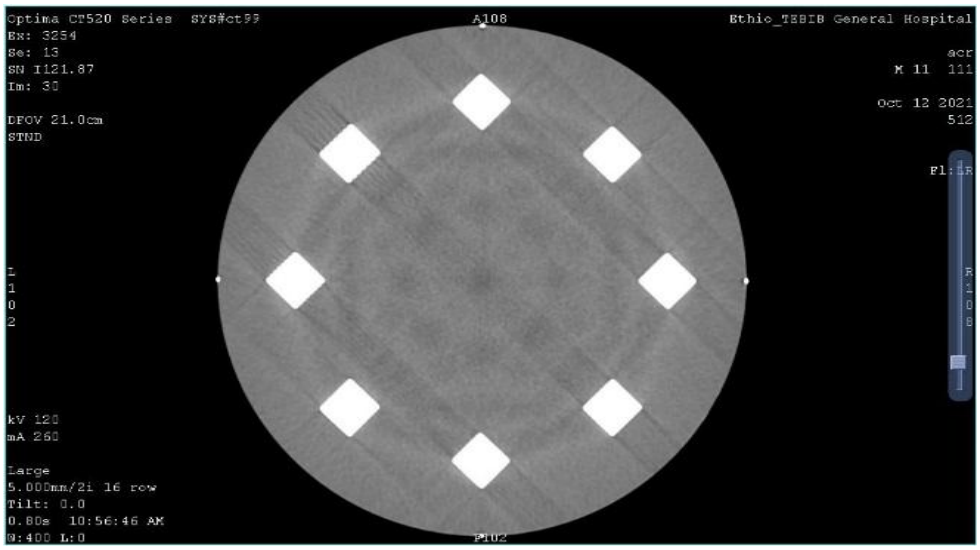
The spatial frequencies of the eight aluminum bar patterns in High contrast resolution



Streak artifacts shown in one of the radiology department at soft tissue window and level values, Figure 5-11. As tabulated in Table 7-6 above, the recommended acceptable level of image clinical quality assessment is no artifacts that have the potential to compromise diagnostic confidence. While, the recommended achievable level of medical image quality assessment is no visible artifacts. In this study, the CT image quality assessment in most radiology department revealed within an acceptable and achievable range.

Figure V-11

Streak artifacts shown at soft tissue window and level values



5.3.6 Discussion

The WHO defined quality assurance program (QAP) in diagnostic radiography as "diagnostic image must be adequately great excellence by giving sufficient clinical info at minimum cost and patient dose to radiation" (IAEA, 2012). The QAP is detailed in the book clearly. Hence, any regulatory body required the compliance the licensee to achieve sufficient and adequate image quality intended for the diagnostic purpose. For that purpose, the regulatory body need well documented periodic and regular QA and optimization of an exam protocols with regard to radiation dose and image quality. Hence, any CT facility should insure quality services to the patient by monitoring radiation dose and image quality through adapting and exercising regular QA culture.

The diagnostic consequence is highly related with the capability of the image scientists towards comprehending the required medical image information. The image scientist observes and read the diagnostic medical image subjectively using his eye (Lee et al., 2021). He generates subjective decision based on his understanding. This will lead to aggravate the error frequency of medical diagnosis and varying medical outcome may be recorded for the same signals. As the most common operator errors described in section 7.5, individual errors is the main sources for the inconsistency of diagnostic information for the same signals. This quantitative analysis of image quality is highly sensitive to subjective medical image assessment (Lee et al., 2021). ACR CT phantom image analysis required consistent criteria.

The phantom and scanner alignments were calibrated and the result revealed that all the four BBs were clearly visible as well as the central wires were symmetrically positioned, as shown in Figure 7-13 above.

Besides, HU is commonly applied in medical diagnosis. Many paper review revealed that measurement of HU can be applied for clinical diagnosis for characterization and differentiation of tissue as well as for determining the threshold levels of malign and benign tumors in adrenal glands and others (Roa, Andersen, & Martinsen, 2015). The results of this research revealed that HU measurement variation for the same ACR CT phantom among the different CT modalities. Hence, the medical professionals must be mindful for this variation when they are using HU for diagnosis aimed for characterization and differentiation of tissue as well as for determining the threshold levels of malign and benign tumors in adrenal glands and others tissues.

The spatial resolution can be expressed as the capability of detecting images of small objects that have high contrast subject with excellent spatial information in the CT scanner. Spatial resolution mostly signifies to objects that have high contrast nature like iodine enhanced vessels or bones. It is relied on size of detector elements, reconstruction algorithm, display FOV, geometry of the CT scan, size of focus and reconstruction matrix (Roa et al., 2015). The result of this research disclosed that spatial resolution determined in different CT modalities had no significant differences. Most CT scanner were clearly viewing the high contrast materials in the ACR CT phantom within an acceptable range, Figure 7-17.

The most ACR CT phantom image quality was acceptable for the test parameter used. The CT number linearity was acceptable. The linearity test is due to the electron density proportionality of water, acrylic, air, Polyethene and polyamides. Their correlation amount was 0.9991 (the tolerance value is 0.99). The image noise and high contrast (spatial) resolution tests were all in the acceptable range except low contrast detectability.

According to this research outcome, further image quality tests must be conducted periodically in each facility in order to assure consistent image quality using consistent IQ criteria for each examination at each health center. Greater attention should be also given to image quality QA in order to include as parts of their total QA program.

5.3.7 Limitation

The limitation of this research were on the image quality data collected. That were obtained from different CT modalities. Even though the researcher used the same ACR CT phantom for all scanner examination, tube collimation, voltage and reconstruction algorithms were varied keeping other scanning parameters were the same. These variations will consequence on the results of image quality. This image quality data will not be used to compare among CT modalities. Rather, it will slightly show the image quality criteria considered at each health facility based on the CT vendor technical specification. The next researcher is advised to use different image quality phantom for the different CT modality based on the vendor image quality phantom recommendation.

5.3.8 Conclusion

Assessments of IQ is highly essential in medical diagnosis at feasibly regulated radiation doses using the principles of ALARA. The evaluation of image quality is the main elements in CT quality control program. In this research, MDCTs (delivers significantly high patient radiation doses requiring wise optimization of the technical parameters by the operator (IAEA, 2012)) were scanned towards understanding the clinical image quality factors using multimodality ACR CT accreditation solid phantom. The phantom contained four modules made from water equivalent materials. This research was obtained the images of each module based on routine examination protocols of head and abdomen to determine the acceptance of image quality with respect to uniformity noise, low and high contrast resolutions. Currently, image quality classification is increasing with the emerging CT technology. Hence, image quality pedals using standard ACR CT accreditation phantom is highly essential to generate consistent and adequate diagnostic medical information as well as towards blocking defective medical equipment from clinical application. Regular QC program used to understand the equipment's current performance. The equipment performance of the participated CT facility to this study was generally concluded satisfactory. The image quality tests of was also acceptable except the slight drifting of low contrast detectably. This variation may be due to variation in the CT scanners.

CHAPTER VI

CONCLUSIONS AND RECOMMENDATIONS

6.1 Conclusion

This research introduces applicable CT diagnostic reference level (DRL) projecting towards patient dose optimization. DRL is adhered with radiation protection because it gives signals to the health facilities how much radiation doses being used for diagnosis. Dissemination of information is highly required before any kinds of regulation and standardization are enacted. Hence, this research outcome was organized and disseminated to medical professionals and any other stakeholders through Addis Ababa health bureau.

To the best of researcher's information, the outcome of the current CT DRL is the first time CT dose audit for common CT examination in Addis Ababa, Ethiopia. Phantom based CT radiation dose quantification was successfully performed according to European guidelines for quality assurance in CT, IAEA, ACR and ICRP recommendations. Because indirect dose evaluation by using phantom is encouraged in any CT facility. The summarized and established DRLs obtained from this research presented in Table 5-5. CT facilities in Addis Ababa can apply this proposed DRLs in order to reduce the collective effective dose burden of the patient significantly. This will decrease the associated risks of cancer.

Assessments of IQ is highly essential in medical diagnosis at feasibly regulated radiation doses using the principles of ALARA. The evaluation of image quality is the main elements in CT quality control program. In this research, MDCTs (delivers significantly high patient radiation doses requiring wise optimization of the technical parameters by the operator (IAEA, 2012) were scanned towards understanding the clinical image quality factors using multimodality ACR CT accreditation solid phantom. The phantom contained four modules made from water equivalent materials. This research was obtained the images of each module based on routine examination protocols of head and abdomen to determine the acceptance of image quality with respect to uniformity, noise, low and high contrast resolutions. Currently, image quality classification is increasing with the emerging CT technology. Hence, image quality pedals using standard ACR CT accreditation phantom is highly essential to

generate consistent and adequate diagnostic medical information as well as towards blocking defective medical equipment from clinical application. Regular QC program used to understand the equipment's current performance. The equipment performance of the participated CT facility to this study was generally concluded satisfactory. The image quality tests of was also acceptable except the slight drifting of low contrast detectably. This variation may be due to variation in the CT scanners

6.2 Recommendations According to Findings

6.2.1 Recommendations to Healthcare Management Bodies

Although the estimated DRLs values of $CTDI_v$ and DLP for the commonest CT examinations all fell within acceptable international records. The researcher was only proposed DRLs for five common examinations. Hence, the health institutions are requested to customize the outcome of this research to its local DRLs; and also requested to develop local DRLs for the rest of CT procedures those are not included in this research. As a consequence, the health center management body need to ensure the radiological protection of the patients by managing the advisory dose levels set by scientific bodies and if it is exceeded consistently prompt local review of the local DRLs is required by the management bodies. Refresher training related to the practical application of DRLs too should be arranged for the medical professional to encourage their capability of optimizing patient dose. Proper and periodic radiation monitoring exercise (i.e. quality assurance program) should be cultured to control the possible rise in patient dose. Furthermore, medical professional who are working in radiology department should be provided with personal monitoring service (thermo-luminescent dosimeter) and personal protective equipment.

6.2.2 Recommendation to Health Bureau of Addis Ababa

Addis Ababa health bureau need to develop mechanisms to implement the legislations, regulations and bylaws through serving guidelines, protocols and reference levels that striving dose optimization to the medical diagnosis. The demand of CT modality in medical diagnosis continues to increases alarmingly throughout the country. For instance, according to ERPA data base archive report, Addis Ababa has 35 functioning CT scanners, 92 within the country. However, reference levels were not in placed to monitor the patient dose. This tells the need of developing dose optimization strategies in line with the demand of medical diagnosis using CT modality. The researcher found that some health centers in Addis Ababa applied adult scanning protocols for pediatric

scans that increases children's risks of cancer. When these health centers requested not to use adult scanning protocols for pediatric scans, they replied they don't have pediatric scanning protocols installed into the CT settings. Hence, the bureau need to find solution for those health facilities.

In another health institution, the CT scan mode selection tab were unable to change from one mode to the other (i.e. from sequential to spiral or vice versa) or the operators do not know the how to operate to change scan mode. Such kinds of problems can affect patient dose. Hence, the health bureau should notice the risks of radiation and exercise its authority towards patient protection via developing investigation levels based on research facts.

6.3 General Recommendations for Further Research

This is the first studies in Addis Ababa that drown conclusive local DRLs to Addis Ababa based on detailed analysis of huge data gathered in three phases of data collection during the research period. The main limitation of this research was that the established regional DRLs focused only limited CT procedures. Therefore, the future work should be conducted by including all types of CT examinations. It was also limited to the CT practiced for adult patients, pediatric patients were not considered. And, the research was restricted in Addis Ababa only. Therefore, future study should package gathering of data from all over the country to produce data that uses to develop national DRL. The efforts of keeping patient dose consistent with ALARA should also expanded to other imaging modalities like conventional radiography, fluoroscopy, mammography, dental radiography, nuclear medicine, MRI and other practices based on their priority by considering both adult and pediatric patients.

The heterogeneity of methodologies that has been publicized in writings for data collection to develop DRLs creates the comparison of results a problematic task. This problem is highly pronounced to pediatrics DRLs caused by large variety of patient characteristics. The potential confusion as a result of units applied in writings by different authors to report dose values. For instance, to define CTDI values, authors may report the quantities in cGy, mGy, or Gy. And to define DLP values, authors may present cGy.cm, mGy.cm, Gy.cm, rad.cm. In addition, some authors report CT doses in $CTDI_w$ instead of $CTDI_v$ that was created confusion for comparison of results. Obtaining dose values for all anatomical regions of CT procedures were also the very challenging problem to generate comparison among results data.

6.4 Tips of Dose Optimization Strategies

Dose optimization and dose reduction are different in their meaning. Dose optimization is limiting the radiation dose received from X-ray medical imaging that is highly required without jeopardizing the importance of diagnosis; however, dose reduction is minimizing the amount of radiation dose to significantly low. Most of the time low doses can produce inadequate images diagnostically that consequences in repeated scan of the patient. However, the application of significantly low doses of radiation to diagnose patient contradict the goal of lowering patient radiation dose. Therefore, when lowering the radiation dose to the patient diagnostically, there should follow the radiological protection principle of optimization. Consequently, the risk-to-benefit ratio of CT procedures can be exploited via optimization of doses. All-inclusive legislation, guidelines, procedures and protocols should be developed by the concerned body which are highly important at the national level for quality improvement and control of cost (Li, 2015) in the process of radiological protection principle. Conversely, enormous recommended guidelines and protocols relied on lacking, misinterpreted or misleading scientific confirmation which has not been assessed in the suitable settings (Li, 2015). Hence, research based evidences health care decision making are highly essential to develop dose optimization strategies which requires cooperative participation of many stakeholders such as healthcare professionals, patients and organizations (Li, 2015). Especially, the role of medical professionals to implement dose optimization strategies are indispensable to reduce the entailing risks of damage to health incurred due to medical diagnosis using CT. This requires careful follow-up and responsiveness of radiological protection of CT procedures by medical management bodies.

Hence, to be consistent with ALARA principle, CT examinations should be justified in advance of its conducts. The advisory role of radiologists or radiology professionals are highly credible in the decision of stating medical professionals. CT should the last diagnostic alternative next to MRI, ultrasound, etc. Radiology professionals should use every approach of setting CT parameters towards minimizing patient dose. Some of the approaches of optimizing patient dose are minimizing scan range, reducing number of repeated scans, using alternative imaging modality and using developed reference levels like DRLs. The clinical indication and imaging protocols of perfusion CT should be certainly decided due to its high doses of radiation (Goo, 2012).

6.5 Optimization Strategies via Rules

Legislation is a binding rule that legitimates legal forces and effects. It should be enforced, recognized and prescribed by a governing authority. When rule is broken, criminal punishment or civil liability will have subjected to an individual. According to European Union, there existed three types of legislations including acts (statutes), regulations and bylaws. Statute is general governing rules requiring public consensus before going to regulation. Regulations and bylaws are the functioning details of statutes (Li, 2015). Countries require radiation protection legislations and radiation protection policies governing for the application of radiation in medical imaging. These legal frameworks will deliver the overall standards of CT scanners operational life cycle such as installation, operation, and decommissioning, for instance ERPA is empowered to discharge these three CT operational life cycle through proclamation No.1025/2017. The legal acts should also inculcate the radiological protection of optimization by giving special emphasis to CT procedures to care medically exposing patients. Hence, Ethiopia, specifically Addis Ababa health bureau need to develop supplemental documents to bylaws to optimize implementation of radiation protection policies that is highly essential to the practical applications of the proposed DRLs at the health centers. Then, best practices of patient protection will be cultured through time at institutional levels to certify that CT scanners are conducted consistent with the ALARA principle.

REFERENCES:

A. Brink and L. Miller. (2015). U.S. national diagnostic reference levels : Closing the. Radiology.Rsna.Org, 277(1), 3–6.

A. Saravanakumar et al. (2014). Establishment of diagnostic reference levels in computed tomography for select procedures in Pudhuchery, India. Journal of Medical Physics, 39(1), 50–55. doi:10.4103/0971-6203.125509

A. Tsalafoutas and V. Koukourakis. (2010). Patient dose considerations in computed tomography examinations. World Journal of Radiology, 2(7), 262–268. doi:10.4329/wjr.v2.i7.262

Abebe A and Catriona W. (2015). Ethiopia: improving health system efficiency. Human resources for health reforms.

ACR. (2004). Instruction manual for testing the ACR CT phantom.

Albngali, A., Shearer, A., van der Putten, W., Tuohy, B., Colgan, N. (2018). CT Output Dose Performance-Conventional Approach versus the Dose Equilibrium Method. International Journal of Medical Physics, Clinical Engineering and Radiation Oncology, 07(01), 15–26. doi:10.4236/ijmpcero.2018.71002

Ali, W. M., Elawad, R. M., & Ibrahim, M. A. A. (2016). Establishment of Dose Reference Levels for Nuclear Medicine in Sudan. Open Journal of Radiology, 06(04), 258–263. doi:10.4236/ojrad.2016.64034

Alsafi, K. G. (2016). Radiation Protection in X-Ray Computed Tomography : Literature Review. Int J Radiol Imaging Technol, 2(3).

American Association of Physicists in Medicine. (2008). The Measurement, Reporting, and Management of Radiation Dose in CT. Report of AAPM Task Group 23: CT Dosimetry Diagnostic Imaging Council CT Committee.

American Association of Physicists in Medicine. (2010). Comprehensive methodology for the evaluation of radiation dose in X-ray computed tomography. A new measurement paradigm based on a unified theory for axial, helical, fan-beam, and cone-beam scanning with orwithout longitudinal translation of the patient t. American Association of Physicists in Medicine.

American association of pysicists in medicine. (2011). AAPM Report No. 204, Size-specific dose estimates (SSDE) in pediatric and adult body CT examinations.

American College of Radiology. (2018). ACR – AAPM – SPR PRACTICE PARAMETER FOR DIAGNOSTIC REFERENCE LEVELS AND ACHIEVABLE DOSES IN MEDICAL X-RAY IMAGING (resolution 40), 1076.

ARPANSA. (2008). Safety guide: Radiation protection in diagnostic and interventional radiology. Radiation protectiob series publication No. 14.1. Australian radiation

protection and nuclear safety agency.

Ataç G, Parmaksız A, İnal T, et al. (2015). Patient doses from CT examinations in Turkey. Diagn Interv Radiol, 1–7. doi:10.5152/dir.2015.14306

Aynalem A, (2014). Health institutions and services. Lesson 12.

Bevelacqua, J. J. (2009). Contemporary Health Physics, Problems and Solutions (Second, Up). Weinheim: Wiley-VCH Verlag GmbH & Co. KGaA.

Brant W and Helms C. (2012). Fundamentals of diagnostic radiology. (C. A. H. William E. Brant, Ed.) (4th editio). Philadelphia, Baltimore, New York, London, Buenos, Hong Kong, Sydney, Tokyo: Lippincott Williams & Wilkins, Wolters Kluwer business.

C. Schmidt. (2012). CT Scans: Balancing health risks and medical benefits. Environmental Health Perspectives, 120(3), A118-121.

Campeloa M, S. M. and T. R. (2016). CTDI versus New AAPM Metrics to assess Doses in CT : a case study. Braz. J. Rad. Sci., 1–15.

Commission, E. (1999). Guidance on diagnostic reference levels (DRLs) for medical exposures.

Commonwealth of Australia. (2015). The 2011 – 2013 national diagnostic reference level service report. ARPANSA Technical Report Series No. 171.

Conference of radiation control program directors. (2015). Part F (Diagnostic x-rays and imaging systems in the healing arts). Suggested State Regulations for Control of Radiation, Vol 1, Ionizing Radiation. Frankfort. SSRCR, I(April), 1–64.

Costello J, C. N., & Tucker J, et al. (2013). CT Radiation Dose : Current Controversies and Dose Reduction Strategies. Medical Physics and Informatics, 201(6).

Council, M. (2004). Diagnostic Reference Levels Position Paper. Medical Council, 1998(September).

Cynthia. (2010). Diagnostic Reference Levels. American College of Radiology, (November), 1–6.

Cynthia McCollough, Theresa Branham, Vince Herlihy, Mythreyi Bhargavan, Lavonne Robbins, Krista Bush, Michael McNitt-Gray, J. Thomas Payne, Tom Ruckdeschel, Doug Pfeiffer, Dianna Cody, and R. Z. (2011). Diagnostic Reference Levels From the ACR CT Accreditation Program. Journal of the American College of Radiology, 8(11), 795–803. doi:10.1016/j.jacr.2011.03.014

Dendy, P. P., & Heaton, B. (2001). Physics for diagnostic radiology. Series in medical physics and biomedical engineering. (and K.-H. N. Webster JG, Tabakov S, Ed.), Medical Physics (Third edit, Vol. 28). New York: CRS press. doi:10.1118/1.1350686

Deserno TM. (2011). Biomedical image processing. New York: Springer.

Di Zhang. (2012). Reduction of radiation dose to radiosensitive organs and its tradeoff with image quality in Computed Tomography. University of California.

Ekpo E et al. (2018). Diagnostic reference levels for common computed tomography (CT) examinations : Results from the first Nigerian nationwide dose survey. Journal of Radiological Protection Accepted Manuscript. doi:10.1088/1361-6498/aaaaf8

Elnour, H., Hassan, H. A., Mustafa, A., Osman, H., Alamri, S., & Yasen, A. (2017). Assessment of iimage quality parameters for computed tomography in Sudan. Open Journal of Radiology, 7, 75–84. doi:10.4236/ojrad.2017.71009

EMoH. (2010). Health Sector Development Program: 2010/11 -2014/15. Federal Democratic Republic of Ethiopia Ministry of Health (Vol. IV).

Enriquez, M. (2015). The thing about " PITCH !" In CT registry review seminar (pp. 1–27).

Erena eta al. (2017). City profile: Addis Ababa. Report prepared in the SES (Social Inclusion and Energy Management for Informal Urban Settlements) project, funded by the Erasmus+ Program of the European Union. Retrieved from http://moodle.donau-uni.ac.at/ses/

European Commission. (1996). European guidleines on quality criteria for computed tomography. Report EUR 16262 EN, Office for Official Publications of the European Communities, Luxembourg.

European Commission. (1997). COUNCIL DIRECTIVE 97/43/EURATOM of 30 June 1997 on health protection of individuals against the dangers of ionizing radiation in relation to medical exposure , and repealing Directive 84/466/Euratom. Official Journal of the European Communities, (L180/22), 22–27.

European Commission. (2008). Radiation Protection N°154. European Guidance on Estimating Population Doses from Medical X-Ray Procedures. Final Report of Contract TREN/04/NUCL/S07.39241.

European Commission. (2014a). Council Directive 2013/59/Euratom of 5 December 2013 laying down basic safety standards for protection against the dangers arising from exposure to ionising radiation, and repealing Directives 89/618/Euratom, 90/641/Euratom, 96/29/Euratom, 97/43/Euratom a. Off J Eur Commun L13, (December 2003), 1–73. doi:doi:10.3000/19770677.L_2013.124.eng

European Commission. (2014b). RadiationProtection No. 180 - Diagnostic Reference Levels in Thirty-six European Countries (Part 2/2).

European Commission. (2018). European guidelines on diagnostic reference levels for paediatric imaging. Radiation protection No 185, ISSN 2315 - 2826, Directorate-General for Energy Directorate D — Nuclear Energy, Safety and ITER Unit D3 — Radiation Protection and Nuclear Safety. Lux. doi:10.2833/003998

F. Ebrahimzadeh, S. M. M. G. A. M. H. R. K. (2015). Diagnostic reference levels (DRLs) for routine X-ray examinations in Lorestan province, Iran. International Journal of Radiation Research, 13(1), 85–90. doi:10.1002/smj.882

Federal Negarit Gazette. Radiation and Nuclear Protection Proclamation No.1025 /2017 (2017).

FMoH. (2010). Health Sector Development Programme IV (2010/11 – 2014/15). Federal Democratic Republic of Ethiopia, Ministry of Health (Vol. IV). https://doi.org/10.1093/toxsci/kft290

Gammex. (2020). CT ACR 464 Phantom. Sun Nuclear Corporation, 1–2.

Gerber, T. C., & Gibbons, R. J. (2010). Weighing the Risks and Benefits of Cardiac Imaging With Ionizing Radiation. JCMG, 3(5), 528–535. doi:10.1016/j.jcmg.2010.03.003

Gevernment of Ehiopia. Health policy of the transitional government of Ethiopia (1993).

Goo, H. W. (2012). CT Radiation Dose Optimization and Estimation : an Update for Radiologists. Korean Journal of Radiology, 13(1), 1–11.

Government of Ethiopia. (1993). Health Policy of Ethiopia.

Grainger & Allison. (2015). Diagnostic radiology: A textbook of medical imaging. (J. H. G. and C. M. S.-P. Andreas Adam, Adrian K. Dixon, Ed.) (6th editio). Edinburgh, London, New York, Oxford, Philadelphia, St Louis, Sydney and Toronto: Elsevier Limited.

Graingr and Allison. (2015). Diagnostic radiology, a textbook of medical imaging. (J. G. and C. S. A. Adam, AK. Dixon, Ed.) (6th editio). Edinburgh, London, New York, Oxford, Philadelphia, St Louis, Sydney and Toronto: Elsevier Limited.

H.Cember and E. Johnson. (2009). Introduction to health physics (Fourth edi). New York: The McGraw-Hill Companies.

Health Canada. (2016). Canadian Computed Tomography Survey - National Diagnostic Reference Levels.

Hendee, W. R., & Ritenour, E. R. (2002). Medical Imaging Physics. Medical Physics (Fourth edi, Vol. 30). New York. doi:10.1118/1.1563664

IAEA. (2001). Radiological Protection of Patients in Diagnostic and Interventional Radiology , Nuclear Medicine and Radiotherapy. In proceedings of an international con- ference held in Malaga, Spain, 26–30 March 2001 / organized by the International Atomic Energy Agency...[et al.] (pp. 26–30).

IAEA. (2002). Radiological Protection of Patients In General Diagnostic Radiology (Radiography), 1–4. Retrieved from https://www-pub.iaea.org/MTCD/publications/PDF/csp_007c/PDF.../CSPS-7-P-CD.pdf

IAEA. (2007). Dosimetry in diagnostic radiology: an international code of practice. International atomic energy agency (Vol. 457).

IAEA. (2009). Dose reduction in CT while maintaining diagnostic Confidence: A feasibility/demonstration study, IAEA-TECDOC-1621. Austria-vienna.

IAEA. (2011). Radiation Protection and Safety of Radiation Sources : International Basic Safety Standards. General Safety Requirements Part 3, No. GSR Part 3. (Interim ed, Vol.

3). Vienna: International Atomic Energy Agency. doi:STI/PUB/1578

IAEA. (2012). Quality Assurance Programme for Computed Tomography: Diagnostic and Therapy Applications IAEA humane health series No. 19. Vienna: International Atomic Energy Agency. Retrieved from http://www.iaea.org/Publications/index.html

ICRP. (1982a). Protection of the patient in diagnostic radiology. ICRP publication 34, volume 9 No. 2/3.

ICRP. (1982b). Protections of the patient in diagnostic radiology, a report of committee 3 of the international commicssion on radiological protection. On ICRP publication 34 (ICRP publication 34).

ICRP. (2000a). ICRP Publication 87: Managing patient dose in computed tomography. Annals of the ICRP, 30(4), 7–45. doi:10.1016/S0146-6453(01)00049-5

ICRP. (2000b). Managing patient dose in computed tomography. ICRP publication 87, Ann. ICRP 30(4).

ICRP. (2007). Annals of the ICRP-the 2007 Recommendations of the International Commission on Radiological Protection. Publication 103. (J. VALENTIN, Ed.). Elsevier on behalf of the international commission on radiological protection.

ICRP. (2015). Radiological Protection in Cone Beam Computed Tomography (CBCT). ICRP Publication 129. Ann. ICRP 44(1) (Vol. 44).

ICRP. (2016). Diagnostic Reference Levels in Medical Imaging. Annals of the ICRP, 131. Retrieved from http://journals.sagepub.com/doi/pdf/10.1177/0146645317717209

ICRP. (2017a). Diagnostic Reference Levels in Medical Imaging. ICRP Publication 135. Ann. ICRP 46 (1). doi:10.1594/esi2018/ESI-0018

ICRP. (2017b). Diagnostic reference levels in medical imaging. ICRP Publication 135. Ann. ICRP 46(1) (Vol. 46).

IPSM. (1992). National Protocol for Patient Dose Measurements in Diagnostic Radiology. London.

J-F. Lecomte, S. Solomon, J. Takala, T. Jung, P. Strand, C. Murith, S. Kiselev, W. Zhuo, F. Shannoun, A. J. (2015). Annals o f the ICRP (Vol. 44).

Japan. (2015). Diagnostic Reference Levels Based on Latest Surveys in Japan — Japan DRLs 2015. Medical Exposure Research Information Network (J-RIME), 1–17.

JD. Enderle, Bronzino, J. (2012). Introduction to biomedical engineering (3rd editio). Burlington: Academic press.

Jemal. (2016). Evaluation of the level of norms and associated radiological hazards & risks from mining activities of Kenticha tantalum mines in Ethiopia.

Joseph et al. (2017). Establishment of local diagnostic reference levels (drls) for radiography examinations in north eastern Nigeria. Science World Journal, 12(4), 51–57.

Kanal et al. (2017). U . S . Diagnostic Reference Levels and achievable Doses for 10 adult cT

examinations. Radiology, 284(1).

Khaled ME. and Sandra AO. (2014). Introduction to diagnostic radiology. (Khaled ME. and Sandra AO, Ed.). New York, Chicago, San Francisco, Athens, London, Madrid, Mexico City, New Delhi, San Juan, Singapore, Sydney and Toronto: McGraw-Hill Education.

Khelassi-Toutaoui N, Merad A, Tsapaki V, et al. (2020). Adult CT examinations in Algeria: Towards updating national diagnostic reference levels. Radiation Protection Dosimetry, 1–8. doi:10.1093/rpd/ncaa116

Korir, G. K., Wambani, J. S., Korir, I. K., Tries, M. A., & Boen, P. K. (2016). National diagnostic reference level initiative for computed tomography examinations in Kenya. Oxford University Press., 168(2), 242–252.

Korir, G. K., Wambani, J. S., Yuko-Jowi, C. A., Korir, I. K., Tries, M., & Kidali, M. M. (2014). Establishing diagnostic reference levels for interventional procedures in Kenya. Radiography, 20(2), 148–152. doi:10.1016/j.radi.2013.10.007

Kwadzo, J. N. (2016). Developing a framework of quality in radiographic service delivery in Ghana Developing a Framework of Quality in Radiographic Service Delivery in Ghana A thesis submitted in partial fulfilment of the requirements of for the degree of Doctor of Philosophy.

Lee, K. B., Nam, K. C., Jang, J. S., & Kim, H. C. (2021). Feasibility of the Quantitative Assessment Method for CT Quality Control in Phantom Image Evaluation. Appl. Sci., 11(3570). Retrieved from https://doi.org/10.3390/app 11083570

Li, A. (2015). The Formulation of a Comprehensive Strategy for Computed Tomography Radiation Dose Optimization in Ontario. University of Toronto ©.

McCollough, C. H. (2010). Diagnostic Reference Levels - Image Wisely. American College of Radiology, (November), 1–6. Retrieved from http://www.imagewisely.org/Imaging-Modalities/Computed-Tomography/Medical-Physicists/Articles/Diagnostic-Reference-Levels

McCollough et al. (2004). The phantom portion of the American College of Radiology (ACR) Computed Tomography (CT) accreditation program: Practical tips, artifact examples, and pitfalls to avoid. Medical Physics, 31(9), 2423–2442. doi:10.1118/1.1769632

McCollough et al. (2009). Strategies for Reducing Radiation Dose in CT. Radiol Clin North Am., 47(1), 27–40. doi:10.1016/j.rcl.2008.10.006.Strategies

MF. Reiser et al. (2009). Medical radiology, diagnostic imaging. (M. K. G. and K. H. ALB. Leuven, Ed.) (3rd editio). Springer.

MF. Reiser et al. (2012). Radiation dose from multidetector CT. (M. K. and P. G. D. Tack, Ed.) (2nd editio). Heidelberg, New York, Dordrecht and London: Springer.

MK Abdulkadir. (2015). Determination of computed tomography diagnostic reference levels in north-central Nigeria. In partial fulfilment of the requirements for the award of master

of philosophy degree in medical physics. University of Ghana, Legon.

Moifo, B., Tapouh, J. R. M., Guena, M. N., Ndah, T. N., Samba, R. N., & Simo, A. (2017). Diagnostic Reference Levels of Adults CT-Scan Imaging in Cameroon: A Pilot Study of Four Commonest CT-Protocols in Five Radiology Departments. Open Journal of Medical Imaging, 07(01), 1–8. doi:10.4236/ojmi.2017.71001

Monnehan G, Silue K, Djagouri K, et al. (2017). Determination of diagnostic reference levels in Côte d ' Ivoire : Case of the adult skull computed tomography in the district of Abidjan. International Journal of Science and Research (IJSR), 6(8), 2015–2018. doi:10.21275/ART20175383

Mourão A, Aburjaile W, and S. F. (2019). Dosimetry and Protocol Optimization of Computed Tomography Scans using Adult Chest Phantoms. Int J Radiol Imaging Technol, 5(1), 1–6. doi:10.23937/2572-3235.1510050

Muhogora W, Ahmed A, Beganovic A, et al. (2009). Patient doses in CT examinations in 18 countries: initial results from international atomic energy agency projects. Radiation Protection Dosimetry, 1–9.

Muhogora, W. E., Ngatunga, J. B., Mboya, G., Lema, U. S., Kalambo, C., & Jusabani, A. (n.d.). Patient doses during radiography and CT examinations in Tanzania : Preliminary results. Radiography and CT patient doses in Tanzania, 1–6.

Mulyadin et al. (2018). Study of image quality from CT scanner multi-detector by using Americans college of radiology (ACR) phantom. IOP Conf. Series: Journal of Physics: Conf. Series, 979(012080).

NRPB. (2020). National Diagnostic Reference Levels (NDRLs) from 19 August 2019 [Internet]. Retrieved from https://www.gov.uk/government/publications/diagnostic-radiology-national-diagnostic-reference-levels-ndrls/ndrl

Nyathi, M., & Shivambu, G. I. (2019). Local diagnostic reference levels for common computed tomography procedures at a tertiary Hospital in South Africa. Iranian Journal of Medical Physics, 16(5), 1–6.

Nyathi, T. (2012). Dose optimization in diagnostic radiology (Doctoral dissertation). University of the Witwatersrand.

Øberg, M., & Andersen, C. E. (2011). Patient Doses for CT Examinations in Denmark by In collaboration with Patient Doses for CT Examinations in Denmark, (July).

Paulo, G. do N. N. (2015). Optimisation and Establishment of Diagnostic Reference Levels in paediatric plain radiography. Retrieved from https://estudogeral.sib.uc.pt/handle/10316/29455

Rad-aid. (2007). Ethiopia Country Report.

Rehani, M. M. (2015). Commentary limitations of diagnostic reference level (DRL) and introduction of acceptable quality dose (AQD). BJR Radiol, (November 2014), 11–13.

doi:10.1259/bjr.20140344

Reiser et al. (2012). Radiation dose from multidetector CT. Medical radiology, diagnostic imaging. (and G. P. Tack D, Kalra MK, Ed.) (Second edi). Berlin, Heidelberg: Springer.

Richard. (2009). Reviewing Ethiopia's Health System Development The Health Situation in Ethiopia. International Medical Community, 52(4), 279–286.

Roa, A. M. A., Andersen, H. K., & Martinsen, A. C. T. (2015). CT image quality over time : comparison of image quality for six different CT scanners over a six-year period. Journal of Applied Clinical Medical Physics, 16(2), 350–365.

Ruiz et al. (2016). Diagnostic reference levels and complexity indices in interventional radiology: a national programme. European Radiology, Springer, 26(12), 4268–4276. doi:10.1007/s00330-016-4334-2

S.w. Yoon, J. K. and S. K. L. (2018). Second Korean national CT diagnostic reference levels in 14 CT protocols. European Radiology, 1–13.

S J Foley et al. (2012). Establishment of CT diagnostic reference levels in Ireland. British Journal of Radiology, 85(1018), 1390–1397. doi:10.1259/bjr/15839549

Salama, D. H., Mahdaly, G., Gilley, D., Rehani, M. M., Shawki, M., Vassileva, J., & Salama, A. (2017). Establishing national diagnostic reference levels (DRLs) for computed tomography in Egypt. Physica Medica, 39, 16–24. doi:10.1016/j.ejmp.2017.05.050

Saravanakumar A, Govindarajan K, et al. (2016). Establishment of CT diagnostic reference levels in select procedures in South India. International Journal of Radiation Research, 14(4), 1–8. doi:10.18869/acadpub.ijrr.14.4.341

Semghouli S. et al. (2017). Establishment of a diagnostic reference level for brain ct procedures in Moroccan Hospitals. J. Adv. Res, 5(5), 319–324. doi:10.21474/IJAR01/

Siemens healthcare. (2015). AAPM computed tomography radiation dose education slides. AAPM working group on standardization of CT nomenclature and protocols.

Souza ABB et al. (2009). Evaluation of entrance surface air kerma rate and clinical images quality in chest radiography. Radiol Bras, 42(5), 315–319.

Thakur, Y., Mclaughlin, P. D., & Mayo, J. R. (2013). Strategies for Radiation Dose Optimization. Curr Radiol Rep, 1, 1–10. doi:10.1007/s40134-013-0007-y

Thulani T. (2012). Dose optimization in diagnostic radiology. University of the Witwatersrand.

Tsapaki V, Aldrich J, Sharma R, et al, et al. (2006). Dose Reduction in CT while Maintaining Diagnostic Confidence: Diagnostic Reference Levels at Routine Head, Chest, and Abdominal CT—IAEA-coordinated Research Project. Radiology, 240(3), 828–834.

UN-HABITAT. (2003). Cities and citizens series, urban inequities report: Addis Ababa. Addis Ababa.

United Nations. (2016). Department of Economic and Social Affairs, Population Division. The

World's Cities in 2016, data booklet, ISBN 978-92-1-151549-7. https://doi.org/10.18356/8519891f-en

United Nations. (2017). Department of Economic and Social Affairs, Population Division (2017). World Population Prospects: The 2017 Revision, Key Findings and Advance Tables. Working Paper No. ESA/P/WP/248 (Vol. 248).

UNSCEAR-vol II. (2008). Sources and Effects of Ionizing Radiation. United Nations Scientific Committee on the Effects of Atomic Radiation. UNSCEAR 2008, report to the General Assembly with Scientific Annexes (Vol. II, Scient). Vienna, Austria: International Atomic Energy Agency.

UNSCEAR. (2000a). Report of the United Nations Scientific Committee on the Effects of Atomic Radiation to the General Assembly. New York.

UNSCEAR. (2000b). United Nations Scientific Committee on the Effects of Atomic Radiation. Souces an effects of ionizing radiation. UNSCEAR 2000 Report to the General Assembly, with Scientific Annexes (Vol. I). New York: UNITED NATIONS. Retrieved from http://www.unscear.org/docs/publications/2000/UNSCEAR_2000_Report_Vol.I.pdf

UNSCEAR. (2008). Sources, Effects and Risks of Ionizing Radiation. United Nations Scientific Committee on the Effects of Atomic Radiation, Volume I: UNSCEAR 2008 Report to the General Assembly with Scientific Annexes. United Nations publication (Vol. I). https://doi.org/10.2307/3577647

UNSCEAR. (2010). Sources and Effects of Ionizing Radiation - Annex A: Medical Radiation Exposures. UNSCEAR 2008 Report to the General Assembly with Scientific Annexes. United Nations publication (Vol. I).

Verdun, F. R., Racine, D., Ott, J. G., Tapiovaara, M. J., Toroi, P., & Bochud, F. O. (2015). Physica Medica Image quality in CT : From physical measurements to model observers. Physica Medica, 31(8), 823–843. doi:10.1016/j.ejmp.2015.08.007

Virtanen et al. (2020). Underlying elements of image quality assessmnet: preference and terminology for communicating image quality characteristics. University of Helsinki. https://doi.org/10.1037/aca0000312

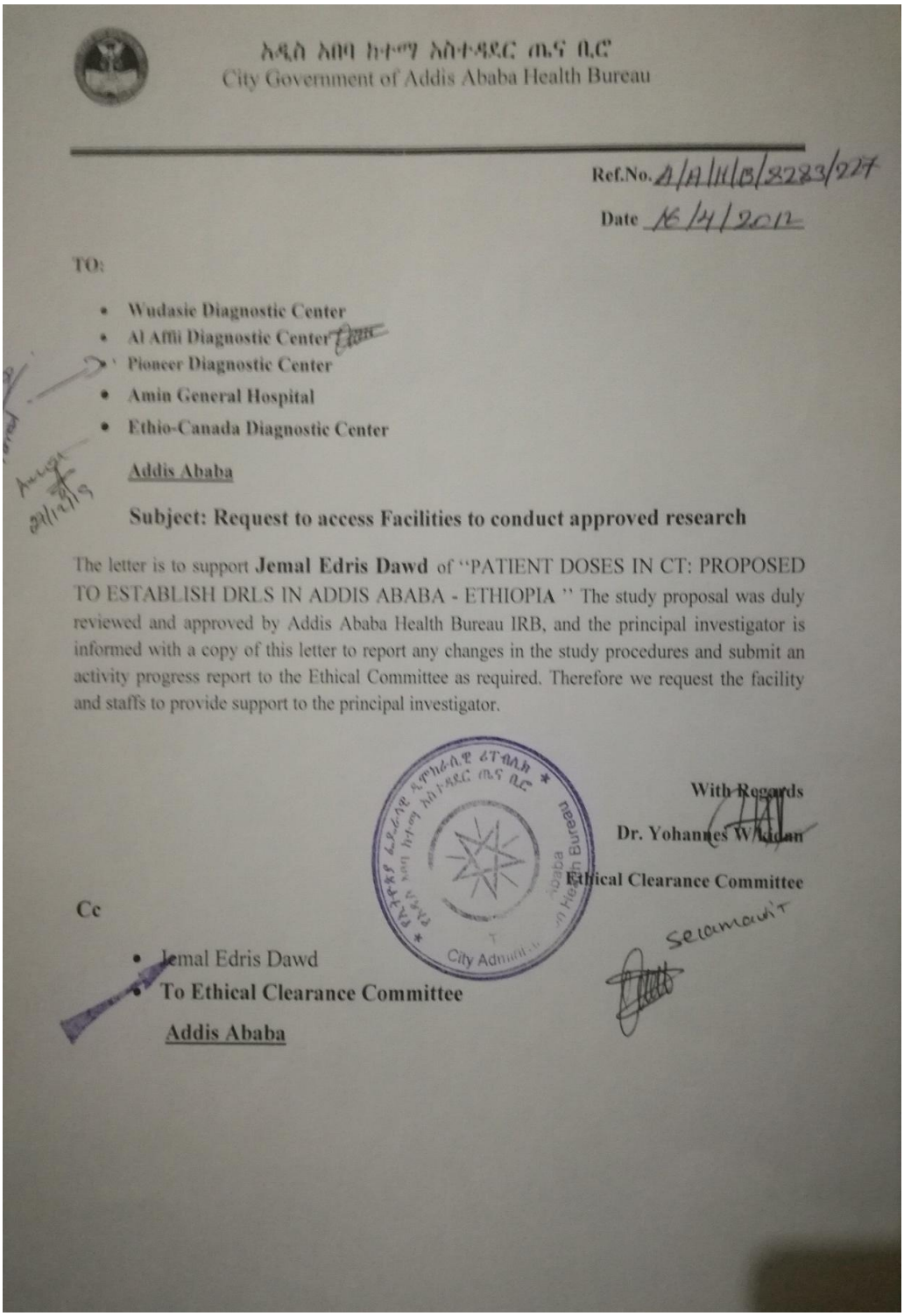
WHO. (2017). Primary health care systems (primasys): case study from Ethiopia, abridged version. Geneva: World health organization.

World Bank. (2015). Addis Ababa, Ethiopia - Enhancing Urban Resilience (City Strength: Resilient Cities Program).

Yurt et al. (2020). Local study of diagnostic reference levels for computed tomography examinations of adult patients in izmir, Turkey. Radiation Protection Dosimetry, (September), 1–6. doi:10.1093/rpd/ncaa121

APPENDICES

APPENDIX I: ETHICAL APPROVAL FROM ADDIS ABABA HEALTH BUREAU IN ETHIOPIA



አዲስ አበባ ከተማ አስተዳደር ጤና ቢሮ
City Government of Addis Ababa Health Bureau

Ref.No. አ/አ/ጤ/ቢ/8283/17

Date ሰ/4/2012

TO:

- Wudasie Diagnostic Center
- Al Affii Diagnostic Center
- Pioneer Diagnostic Center
- Amin General Hospital
- Ethio-Canada Diagnostic Center

Addis Ababa

Subject: Request to access Facilities to conduct approved research

The letter is to support **Jemal Edris Dawd** of "PATIENT DOSES IN CT: PROPOSED TO ESTABLISH DRLS IN ADDIS ABABA - ETHIOPIA " The study proposal was duly reviewed and approved by Addis Ababa Health Bureau IRB, and the principal investigator is informed with a copy of this letter to report any changes in the study procedures and submit an activity progress report to the Ethical Committee as required. Therefore we request the facility and staffs to provide support to the principal investigator.

With Regards

Dr. Yohannes W/Amlak

Ethical Clearance Committee

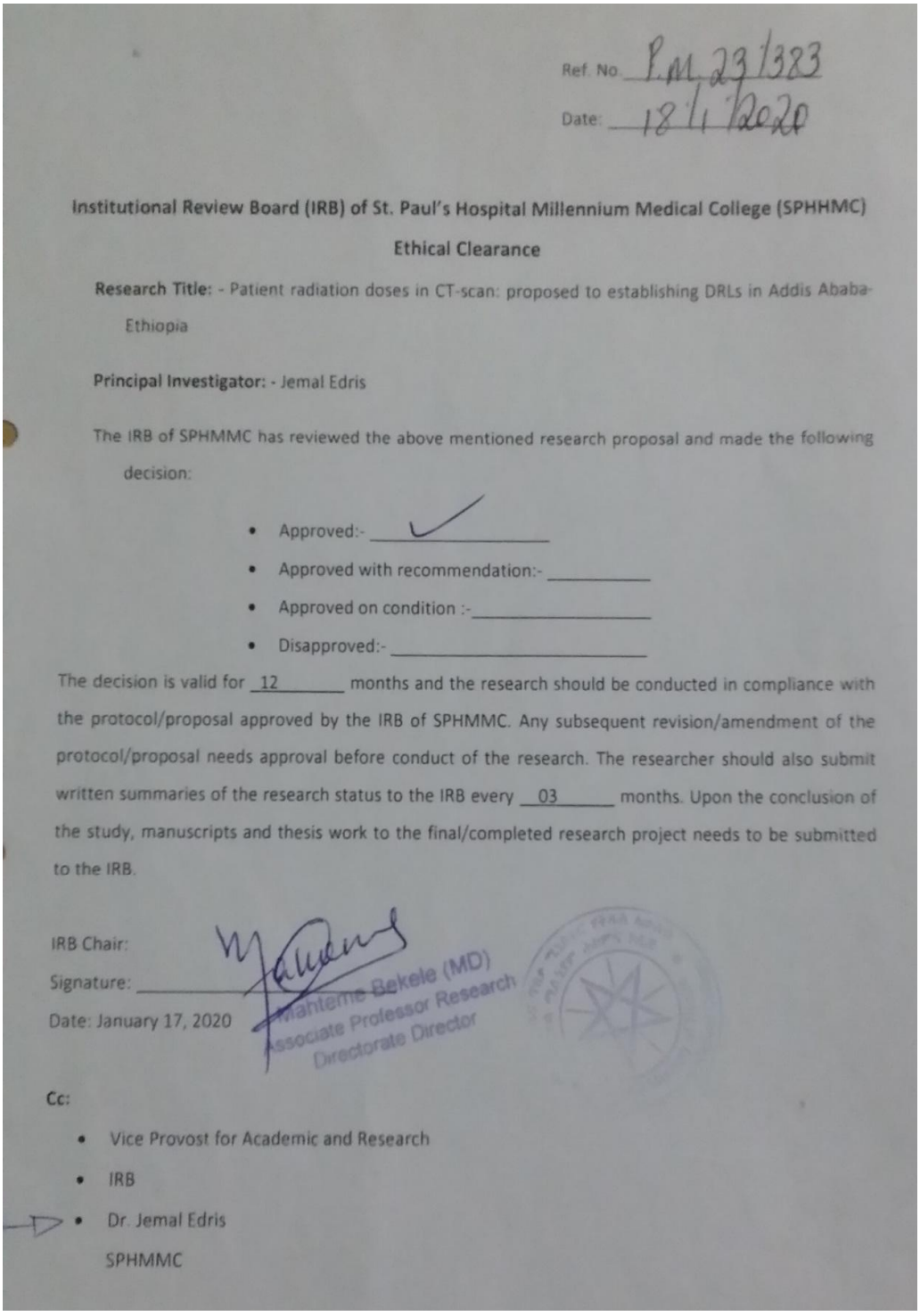
Cc

- Jemal Edris Dawd
- To Ethical Clearance Committee

Addis Ababa

APPENDIX II: ETHICAL APPROVAL RECEIVED FROM REFERRAL HOSPITALS

E.g. St. Paul's Hospital Millennium Medical College.



Ref. No P.M 23/383

Date: 18/4/2020

Institutional Review Board (IRB) of St. Paul's Hospital Millennium Medical College (SPHHMC)

Ethical Clearance

Research Title: - Patient radiation doses in CT-scan: proposed to establishing DRLs in Addis Ababa-Ethiopia

Principal Investigator: - Jemal Edris

The IRB of SPHMMC has reviewed the above mentioned research proposal and made the following decision:

- Approved:- ✓
- Approved with recommendation:- _____
- Approved on condition :-_____
- Disapproved:- _____

The decision is valid for 12 months and the research should be conducted in compliance with the protocol/proposal approved by the IRB of SPHMMC. Any subsequent revision/amendment of the protocol/proposal needs approval before conduct of the research. The researcher should also submit written summaries of the research status to the IRB every 03 months. Upon the conclusion of the study, manuscripts and thesis work to the final/completed research project needs to be submitted to the IRB.

IRB Chair:

Signature: _____

Date: January 17, 2020

Mahteme Bekele (MD)
Associate Professor Research
Directorate Director

Cc:

- Vice Provost for Academic and Research
- IRB
- Dr. Jemal Edris
  SPHMMC

APPENDIX III: ETHICAL APPROVAL RECEIVED FROM NEAR EAST UNIVERSITY

APPENDIX IV: ETHICAL APPROVAL FROM ETHIOPIAN RADIATION PROTECTION AUTHORITY

የኢትዮጵያ ፌደራላዊ ዲሞክራሲያዊ ሪፐብሊክ
The Federal Democratic Republic of Ethiopia
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Ethiopian Radiation Protection Authority

ኢ. ጨ. መ. ባ
ERPA

ቁጥር
Ref. No ERPA/30/8006/2011
ቀን
Date 04 SEP 2019

TO WHOM IT MAY CONCERN

Dear Sir,

In partial fulfilment of the requirement for the Ph.D degree, Mr. Jemal Edris Dawd is pursuing assistance. He is expected to complete a research program in Biomedical Engineering related to Diagnostic radiology. He has expressed his interest to collect data from Hospitals and diagnostic centres in Addis Ababa for his Dissertation work.

The title of his dissertation is: "Establishing diagnostic reference levels for patient radiation doses in CT-scan procedures in Addis Ababa – Ethiopia".

Therefore, the Ethiopian Radiation Protection Authority is kindly requesting your esteemed organization to cooperate and offer the necessary assistances.

Thank you for your kind cooperation.

Yours faithfully,

The Federal Democratic Republic of Ethiopia
Ethiopian Radiation Protection Authority
Solomon Getachew
Director General

ለላቀ የጨረራ መከላከልና ደህንነት መረጋገጥ እንተጋለን!
Striving for Excellence in Radiation Safety and Security!

☎ +251 114 70 55 79
+251 114 70 55 80
+251 114 70 55 82
20486 ኮድ 1000 አዲስ አበባ
20486 Code 1000 Addis Ababa
Fax:- +251 114 70 55 78/+251 114 70 55 84
E-mail:- info@erpa.gov.et
ድረ ገፅ /Website:-www.erpa.gov.et

APPENDIX V: DATE COLLECTION BOOKLET-1

NEAR EAST UNIVERSITY, GRADUATE SCHOOL OF APPLIED SCIENCES,

DEPARTMENT OF BIOMEDICAL ENGINEERING

COMPUTED TOMOGRAPHY PATIENT DOSE, DATA COLLECTION BOOKLET-1

Name of Health Center: _____ Date of Examination:_____

Manufacturer:_____Model:_____ Serial No.:_____Date of Manufacture:_____

Tyep: Single slice [] Multislice [] Projection: Axial [] Helical [] CT No:_____

Types of Exam: Head [] Chest [] Abdomen [] pulvice [] C-spine []

Patient	EXAMINATION PARAMETERS													
	Sex	Age	Contrast	kVp(v)	mAs(mAs)	Scane_t(s)	Rotation_t(s)	Slice_T(mm)	N	NT(mm)	L(mm)	pitch	$CTDI_V$(mGy)	**DLP(mGy.cm)**
1														
2														
3														
4														
5														
6														
7														
8														
9														
10														
11														

Where, Scan_t=scanning time, Rotation_t=gantry rotation time, T=nominal slice thickness (mm), N=number of slice images, L=scan length, NT=nominal scan width, $CTDI_V$=volume computed tomography dose index and DLP=dose length product.

APPENDIX VI: CALIBRATION CERTIFICATE REGARDING USB-MULTIMETER AND ACCESSORY

IBA Dosimetry GmbH

Service Calibration Laboratory

Kalibrierschein / MTK nach § 14 MPBetreibV
Calibration certificate

Kalibrierscheinnummer:
Calibration certificate number: MM-001413

MagicMaX Typ *MagicMaX Type*	**MagicMaX** Universal	
Gegenstand *Object*	**USB-Multimeter und Zubehör** USB-Multimeter and accessory	
Hersteller *Manufacturer*	**IBA Dosimetry GmbH**	
Serien-Nr. *Serial number*	**G19-0860**	
	Typ *Type*	Serien-Nr. *Serial number*
Detektor/en *Detector/s*	**XR-Detector**	**R19-0766**
	DCT10-MM	**2640**
	Current Probe	**18380061**
Auftragsnummer *Order No.*	**743887**	
Anzahl der Seiten des Kalibrierscheines *Number of pages of the certificate*	**4**	

Dieser Kalibrierschein dokumentiert die Rückführung auf nationale Normale zur Darstellung der Einheiten in Übereinstimmung mit dem Internationalen Einheitensystem (SI). Die für die Kalibrierung verwendeten Referenznormale, werden regelmäßig überprüft und sind rückführbar auf die Normale der Physikalisch Technischen Bundesanstalt (PTB) in Braunschweig kalibriert.

Für die Einhaltung einer angemessenen Frist zur Wiederholung der Kalibrierung ist der Benutzer verantwortlich. Die Nachprüffrist nach 1.6 der Anlage 2 MPBetreibV beträgt 5 Jahre.

This calibration certificate documents the traceability to national standards, which realize the units of measurement according to the International System of Units (SI).
The reference norms used for the calibration are those norms specified by the National Metrology Institute of Physics in Braunschweig (Physikalisch-Technische Bundesanstalt), Germany.

The user is obliged to have the object recalibrated at appropriate intervals.

Für den oben aufgeführten Kalibriergegenstand empfehlen wir alle zwei Jahre eine Werkskalibrierung.
For the a.m device we recommend a factory Calibration at our works every two years.

Dieser Kalibrierschein darf nur vollständig und unverändert weiterverbreitet werden. Auszüge oder Änderungen bedürfen der schriftlichen Genehmigung von IBA Dosimetry GmbH. Kalibrierscheine ohne Unterschrift haben keine Gültigkeit. In der Bundesrepublik Deutschland gilt dieser Kalibrierschein als Nachweis für die erfolgreiche Durchführung der messtechnischen Kontrolle (MTK) nach § 14 MPBetreibV.

This calibration certificate must not be reproduced other than in full except with the written permission of IBA Dosimetry GmbH. Calibration certificates without signature are not valid. In Germany this calibration certificate is a proof for the correct implementation of "messtechnische Kontrolle" (MTK) according to the German law § 14 MPBetreibV.

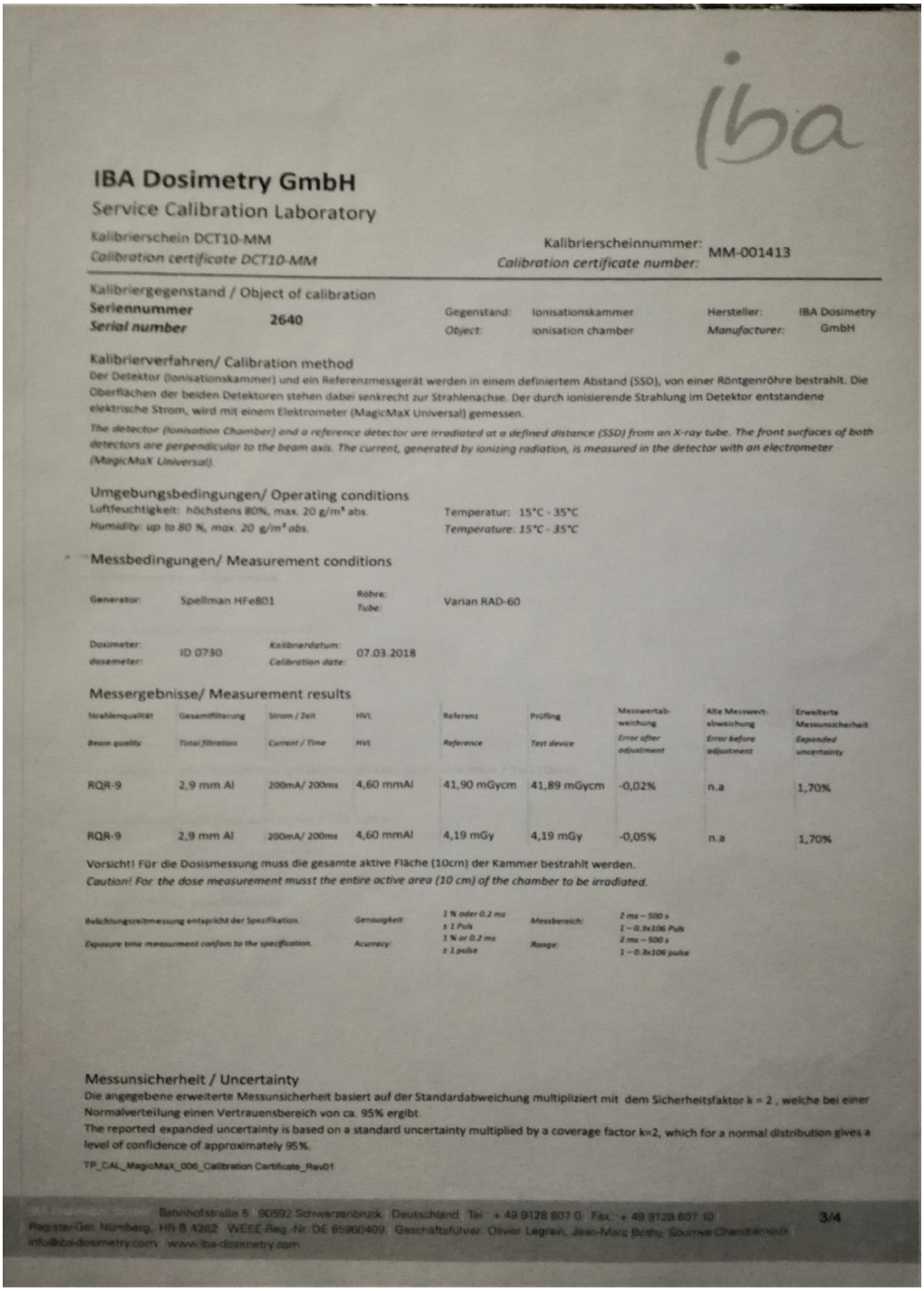
Kalibrierdatum
Calibration date
19. Juni 2019
19 June 2019

Bearbeiter
Person in charge
Slim Najah

TP_CAL_MagicMaX_006_Calibration Certificate_Rev02

IBA Dosimetry GmbH Bahnhofstraße 5 | 90592 Schwarzenbruck | Deutschland | Tel.: + 49 9128 607 0 | Fax.: + 49 9128 607 10 |
Register-Ger. Nürnberg, HR B 4262 | WEEE-Reg.-Nr. DE 65960409 | Geschäftsführer: Olivier Legrain, Jean-Marc Bothy, Soumya Chandramouli |
info@iba-dosimetry.com | www.iba-dosimetry.com

APPENDIX VII: CALIBRATION CERTIFICATE REGARDING IONIZATION CHAMBER



IBA Dosimetry GmbH

Service Calibration Laboratory

Kalibrierschein DCT10-MM
Calibration certificate DCT10-MM

Kalibrierscheinnummer: MM-001413
Calibration certificate number:

Kalibriergegenstand / Object of calibration

Seriennummer **2640**
Serial number

Gegenstand: Ionisationskammer — Hersteller: IBA Dosimetry GmbH
Object: ionisation chamber — *Manufacturer:*

Kalibrierverfahren/ Calibration method

Der Detektor (Ionisationskammer) und ein Referenzmessgerät werden in einem definiertem Abstand (SSD), von einer Röntgenröhre bestrahlt. Die Oberflächen der beiden Detektoren stehen dabei senkrecht zur Strahlenachse. Der durch ionisierende Strahlung im Detektor entstandene elektrische Strom, wird mit einem Elektrometer (MagicMaX Universal) gemessen.

The detector (Ionisation Chamber) and a reference detector are irradiated at a defined distance (SSD) from an X-ray tube. The front surfaces of both detectors are perpendicular to the beam axis. The current, generated by ionizing radiation, is measured in the detector with an electrometer (MagicMaX Universal).

Umgebungsbedingungen/ Operating conditions

Luftfeuchtigkeit: höchstens 80%, max. 20 g/m³ abs. — Temperatur: 15°C - 35°C
Humidity: up to 80 %, max. 20 g/m³ abs. — *Temperature: 15°C - 35°C*

Messbedingungen/ Measurement conditions

Generator: Spellman HFe801 — Röhre / Tube: Varian RAD-60

Dosimeter / dosemeter: ID 0730 — Kalibrierdatum / Calibration date: 07.03.2018

Messergebnisse/ Measurement results

Strahlenqualität / Beam quality	Gesamtfilterung / Total filtration	Strom / Zeit / Current / Time	HVL	Referenz / Reference	Prüfling / Test device	Messwertabweichung / Error after adjustment	Alte Messwertabweichung / Error before adjustment	Erweiterte Messunsicherheit / Expanded uncertainty
RQR-9	2,9 mm Al	200mA/ 200ms	4,60 mmAl	41,90 mGycm	41,89 mGycm	-0,02%	n.a	1,70%
RQR-9	2,9 mm Al	200mA/ 200ms	4,60 mmAl	4,19 mGy	4,19 mGy	-0,05%	n.a	1,70%

Vorsicht! Für die Dosismessung muss die gesamte aktive Fläche (10cm) der Kammer bestrahlt werden.
Caution! For the dose measurement musst the entire active area (10 cm) of the chamber to be irradiated.

Belichtungszeitmessung entspricht der Spezifikation. / *Exposure time measurment confom to the specification.* — Genauigkeit / Acurracy: 1 % oder 0.2 ms ± 1 Puls / 1 % or 0.2 ms ± 1 pulse — Messbereich / Range: 2 ms – 500 s, 1 – 0.3x106 Puls / pulse

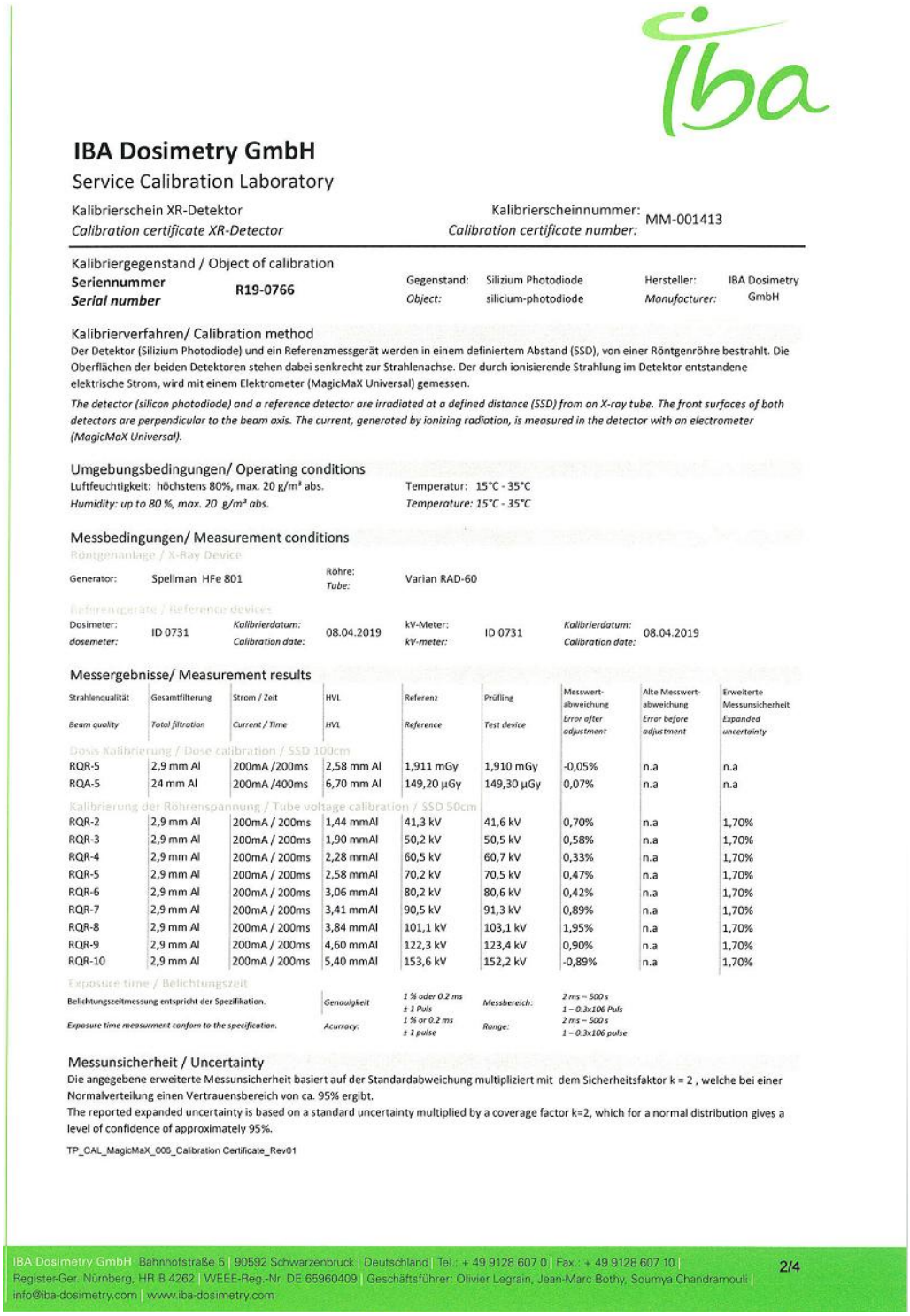
Messunsicherheit / Uncertainty

Die angegebene erweiterte Messunsicherheit basiert auf der Standardabweichung multipliziert mit dem Sicherheitsfaktor k = 2 , welche bei einer Normalverteilung einen Vertrauensbereich von ca. 95% ergibt.
The reported expanded uncertainty is based on a standard uncertainty multiplied by a coverage factor k=2, which for a normal distribution gives a level of confidence of approximately 95%.

TP_CAL_MagicMaX_006_Calibration Certificate_Rev01

Bahnhofstraße 5 90592 Schwarzenbruck Deutschland Tel: + 49 9128 607 0 Fax: + 49 9128 607 10 — 3/4
Register-Ger. Nürnberg, HR B 4262 WEEE-Reg.-Nr. DE 65960409 Geschäftsführer: Olivier Legrain, Jean-Marc Bothy, Soumya Chandramouli
info@iba-dosimetry.com www.iba-dosimetry.com

APPENDIX VIII: CALIBRATION CERTIFICATE SILICIUM-PHOTODIODE



IBA Dosimetry GmbH

Service Calibration Laboratory

Kalibrierschein XR-Detektor
Calibration certificate XR-Detector

Kalibrierscheinnummer: MM-001413
Calibration certificate number:

Kalibriergegenstand / Object of calibration

Seriennummer **R19-0766**
Serial number

Gegenstand: Silizium Photodiode
Object: silicium-photodiode

Hersteller: IBA Dosimetry GmbH
Manufacturer:

Kalibrierverfahren/ Calibration method

Der Detektor (Silizium Photodiode) und ein Referenzmessgerät werden in einem definiertem Abstand (SSD), von einer Röntgenröhre bestrahlt. Die Oberflächen der beiden Detektoren stehen dabei senkrecht zur Strahlenachse. Der durch ionisierende Strahlung im Detektor entstandene elektrische Strom, wird mit einem Elektrometer (MagicMaX Universal) gemessen.

The detector (silicon photodiode) and a reference detector are irradiated at a defined distance (SSD) from an X-ray tube. The front surfaces of both detectors are perpendicular to the beam axis. The current, generated by ionizing radiation, is measured in the detector with an electrometer (MagicMaX Universal).

Umgebungsbedingungen/ Operating conditions

Luftfeuchtigkeit: höchstens 80%, max. 20 g/m³ abs.
Humidity: up to 80 %, max. 20 g/m³ abs.

Temperatur: 15°C - 35°C
Temperature: 15°C - 35°C

Messbedingungen/ Measurement conditions

Röntgenanlage / X-Ray Device

Generator: Spellman HFe 801

Röhre: / *Tube:* Varian RAD-60

Referenzgeräte / Reference devices

Dosimeter: / *dosemeter:* ID 0731
Kalibrierdatum: / Calibration date: 08.04.2019

kV-Meter: / *kV-meter:* ID 0731
Kalibrierdatum: / Calibration date: 08.04.2019

Messergebnisse/ Measurement results

Strahlenqualität / *Beam quality*	Gesamtfilterung / *Total filtration*	Strom / Zeit / *Current / Time*	HVL / *HVL*	Referenz / *Reference*	Prüfling / *Test device*	Messwert-abweichung / *Error after adjustment*	Alte Messwert-abweichung / *Error before adjustment*	Erweiterte Messunsicherheit / *Expanded uncertainty*
Dosis Kalibrierung / Dose calibration / SSD 100cm								
RQR-5	2,9 mm Al	200mA /200ms	2,58 mm Al	1,911 mGy	1,910 mGy	-0,05%	n.a	n.a
RQA-5	24 mm Al	200mA /400ms	6,70 mm Al	149,20 µGy	149,30 µGy	0,07%	n.a	n.a
Kalibrierung der Röhrenspannung / Tube voltage calibration / SSD 50cm								
RQR-2	2,9 mm Al	200mA / 200ms	1,44 mmAl	41,3 kV	41,6 kV	0,70%	n.a	1,70%
RQR-3	2,9 mm Al	200mA / 200ms	1,90 mmAl	50,2 kV	50,5 kV	0,58%	n.a	1,70%
RQR-4	2,9 mm Al	200mA / 200ms	2,28 mmAl	60,5 kV	60,7 kV	0,33%	n.a	1,70%
RQR-5	2,9 mm Al	200mA / 200ms	2,58 mmAl	70,2 kV	70,5 kV	0,47%	n.a	1,70%
RQR-6	2,9 mm Al	200mA / 200ms	3,06 mmAl	80,2 kV	80,6 kV	0,42%	n.a	1,70%
RQR-7	2,9 mm Al	200mA / 200ms	3,41 mmAl	90,5 kV	91,3 kV	0,89%	n.a	1,70%
RQR-8	2,9 mm Al	200mA / 200ms	3,84 mmAl	101,1 kV	103,1 kV	1,95%	n.a	1,70%
RQR-9	2,9 mm Al	200mA / 200ms	4,60 mmAl	122,3 kV	123,4 kV	0,90%	n.a	1,70%
RQR-10	2,9 mm Al	200mA / 200ms	5,40 mmAl	153,6 kV	152,2 kV	-0,89%	n.a	1,70%

Exposure time / Belichtungszeit

Belichtungszeitmessung entspricht der Spezifikation. / *Exposure time measurment confom to the specification.*

Genauigkeit / *Acurracy:* 1 % oder 0.2 ms ± 1 Puls / *1 % or 0.2 ms ± 1 pulse*

Messbereich: / *Range:* 2 ms – 500 s / 1 – 0.3x106 Puls / *2 ms – 500 s / 1 – 0.3x106 pulse*

Messunsicherheit / Uncertainty

Die angegebene erweiterte Messunsicherheit basiert auf der Standardabweichung multipliziert mit dem Sicherheitsfaktor k = 2 , welche bei einer Normalverteilung einen Vertrauensbereich von ca. 95% ergibt.
The reported expanded uncertainty is based on a standard uncertainty multiplied by a coverage factor k=2, which for a normal distribution gives a level of confidence of approximately 95%.

TP_CAL_MagicMaX_006_Calibration Certificate_Rev01

IBA Dosimetry GmbH Bahnhofstraße 5 | 90592 Schwarzenbruck | Deutschland | Tel.: + 49 9128 607 0 | Fax.: + 49 9128 607 10 | Register-Ger. Nürnberg, HR B 4262 | WEEE-Reg.-Nr. DE 65960409 | Geschäftsführer: Olivier Legrain, Jean-Marc Bothy, Soumya Chandramouli | info@iba-dosimetry.com | www.iba-dosimetry.com

2/4

APPENDIX IX: DATE COLLECTION BOOKLET-2.1

NEAR EAST UNIVERSITY, GRADUATE SCHOOL OF APPLIED SCIENCES,

DEPARTMENT OF BIOMEDICAL ENGINEERING

PHANTOM BASED CT DOSE, DATA COLLECTION BOOKLET-2.1

Name of Health Center:_____Date of Data Collection:_____

Manufacturer:_____Model:_____Serial No.:_____Date of Manufacture:_____

Type of machine: Single slice [ ] Multislice [ ] Projection: Axial [ ] Helical [ ]

Types of Exam: Head phantom [ ] Body phantom [ ] No. of Detector (N):_____No. of Phases:_____

Dose Location		Tem (^oC)	Press (hPa)	EXAM PLANE PARAMETERS									
				kVp (v)	mA(mAs)	Scan_t(s)	T(mm)	N	L(mm)	Table increment(mm)	NT(mm)	Pitch	Remark
Head phantom	C												
	P12												
	P3												
	P6												
	P9												
Body phantom	C												
	p12												
	p3												
	P6												
	P9												

Where, Scan_t=scanning time, T=nominal slice thickness (mm), N=number of slice images, L=scan length, NT=nominal scan width; C=center; P12=periphery a 12:00; P3=periphery a :00; P6=periphery a 6:00; P9=periphery a 9:00

APPENDIX X: DATE COLLECTION BOOKLET-2.2

NEAR EAST UNIVERSITY, GRADUATE SCHOOL OF APPLIED SCIENCES,

DEPARTMENT OF BIOMEDICAL ENGINEERING

PHANTOM BASED CT DOSE, DATA COLLECTION BOOKLET-2.2

Name of Health Center:_____Date of Data Collection:_____

Manufacturer:_____Model:_____Serial No.:_____Date of Manufacture:_____

Type of machine: Single slice [ ] Multislice [ ] Projection: Axial [ ] Helical [ ]

Types of Exam: Head phantom [ ] Body phantom [ ] No. of Detector (N):_____No. of Phases:_____

Dose Location		Tem (°C)	Press (hPa)	C_f	Control console				Phantom Based Dose Estimated						phantom vs console doses comparison			
					$CTDI_R$	$CTDI_V$	DLP_R	DLP	$CTDI_R$	$CTDI_{100}$	$CTDI_{ave}$	$CTDI_w$	$CTDI_V$	DLP	$CTDI_{ph-cc}$	DLP_{ph-cc}	CTDIv % deviation	DLP % deviation
Head	C																	
	P12																	
	P3																	
	P6																	
	P9																	
Body	C																	
	p12																	
	p3																	
	P6																	
	P9																	

Where, $CTDI_v$=volume computed tomography dose index and DLP=dose length product, $CTDI_R$=CTDI values obtained from measurement (mGy), DLP_R=DLP recorded from measurement (mGy.cm), C_f=calibration factor of the ionization chamber (from calibration certificate), T= nominal slice thickness (mm), N=number of slice images, subscript ph-c=difference between phantom and console readings

APPENDIX XI: DATE COLLECTION BOOKLET-4

NEAR EAST UNIVERSITY, GRADUATE SCHOOL OF APPLIED SCIENCES

DEPARTMENT OF BIOMEDICAL ENGINEERING

ACR CT PHANTOM FOR IQ ANALYSIS, DATE COLLECTION BOOKLET-4

Name of Institution: _____Date of Data Collection: _____
CT Manufacturer: _____Model: _____Serial No.: _____Date of Manufacture: _____
Type of machine: Single slice [ ] Multi-slice [ ] Projection: Axial [ ] Helical [ ]
No. of Detector (N): _____No. of Phases: _____
ACR Phantom: Model: Serial No.: Date of Manufacture:

Module – 1						
Mean HU value						
Protocol	**Acrylic**	**PTFE**	**Polyethene**	**Polyamide**	**Air**	**water**
HRC						
Ref. CT# value	**140±15**	**100±15**	**-75±15**	**900±60**	**-1000±60**	**0±5**
Module 2						
Number of Low Contrast Objects - Module 2						
Protocol	**Group I**	**Group II**	**Group III**	**Group IV**	**Group V**	**Remark**
Abdomen						
References	**6 mm**	**5 mm**	**4 mm**	**3 mm**	**2 mm**	
	4 BBS	4 BBs	4 BBs	4 BBs	4 BBs	
Module – 3						
CT number uniformity, Image noise and Artifacts						
HU mean in ROI - Module – 3						
Protocol	**Center**	**12:00**	**3:00**	**6:00**	**9:00**	**mean**
Abdomen						

NEAR EAST UNIVERSITY, GRADUATE SCHOOL OF APPLIED SCIENCES

DEPARTMENT OF BIOMEDICAL ENGINEERING

ACR CT PHANTOM FOR IQ ANALYSIS, DATE COLLECTION BOOKLET-4

Name of Institution: _____Date of Data Collection: _____

CT Manufacturer: _____Model: _____Serial No.: _____Date of Manufacture: _____

Type of machine: Single slice [ ] Multi-slice [ ] Projection: Axial [ ] Helical [ ]

No. of Detector (N): _____No. of Phases: _____

ACR Phantom: Model: Serial No.: Date of Manufacture:

Image Noise measured in HU STD in ROI - Module – 3									
Protocol	**Center**	**Remarks**							
Abdomen									
Artifacts and CT number Uniformity - Module – 3									
Protocol	**Artifact**	**CT Number Uniformity**							
		Center CT #	**D/ce from Ref.**	**edge CT#**	**|CT# (Center - edge)|**	**D/ce from Ref.**			
Abdomen									
Reference	**No Artifacts**	**0±5**				**≤ 5 HU**			
Module – 4									
Number of Distinguishable High Contrast Patterns									
Protocol	**Ramp I**	**Ramp II**	**Ramp III**	**Ramp IV**	**Ramp V**	**Ramp VI**	**Ramp VII**	**Ramp VIII**	**Reference**
HRC									5 lp/cm
Abdomen									6 lp/cm

NB: Perform clinical scans covering the CT number accuracy of the ACR accreditation phantom. Any auto mA features must be disabled. Use a mAs value appropriate for an average size patient. At minimum, the scan conducted must include: (1) adult head (average); and (2) adult abdomen (about 70kg)

APPENDIX XII: DATE COLLECTION BOOKLET-3

NEAR EAST UNIVERSITY, GRADUATE SCHOOL OF APPLIED SCIENCES,

DEPARTMENT OF BIOMEDICAL ENGINEERING

ACR CT PHANTOM FOR IQ ANALYSIS, DATE COLLECTION BOOKLET-3

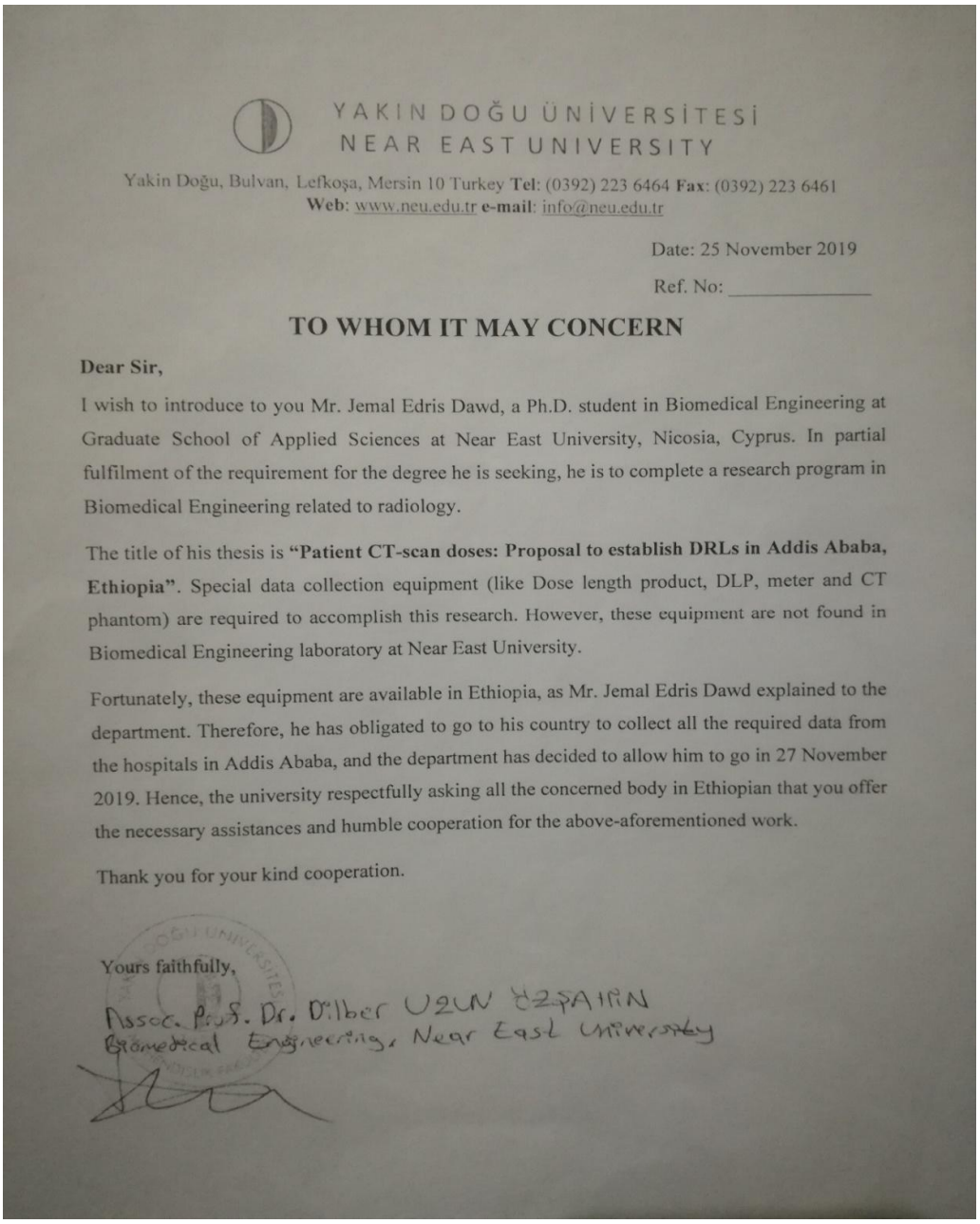
Name of Health Center:_____Date of Data Collection:_____
Manufacturer:_____Model:_____Serial No.:_____Date of Manufacture:_____
Type of machine: Single slice [ ] Multislice [ ] Projection: Axial [ ] Helical [ ]
No. of Detector (N): _____No. of Phases: _____
ACR Phantom: Model: Serial No.: Date of Manufacture:

Protocol	kVp	mA(mAs)	Scan_t (s)	T(mm)	N	Table feed (mm)	Pitch	scan FOV(cm, name)	Display FOV(cm)	Rec. algorithm	Remark
Brain											
Body											
HRC											
Abdomen											

Where, HRC=high resolution chest, scan_t=scanning time, N= Number of Data Channel, T=slice thickness, FOV=field of view, Rec.algorithm=reconstruction algorithm

APPENDIX XIII: PERMISSION LETTER BY GRADUATE SCHOOL OF APPLIED SCIENCES

Support letter written 'To Whom it may Concern' was granted from Graduate School of Applied Sciences at Near East University to request assistance for the research from concerned bodies.



YAKIN DOĞU ÜNİVERSİTESİ
NEAR EAST UNIVERSITY

Yakin Doğu, Bulvan, Lefkoşa, Mersin 10 Turkey **Tel**: (0392) 223 6464 **Fax**: (0392) 223 6461
Web: www.neu.edu.tr **e-mail**: info@neu.edu.tr

Date: 25 November 2019

Ref. No: _____

TO WHOM IT MAY CONCERN

Dear Sir,

I wish to introduce to you Mr. Jemal Edris Dawd, a Ph.D. student in Biomedical Engineering at Graduate School of Applied Sciences at Near East University, Nicosia, Cyprus. In partial fulfilment of the requirement for the degree he is seeking, he is to complete a research program in Biomedical Engineering related to radiology.

The title of his thesis is **"Patient CT-scan doses: Proposal to establish DRLs in Addis Ababa, Ethiopia"**. Special data collection equipment (like Dose length product, DLP, meter and CT phantom) are required to accomplish this research. However, these equipment are not found in Biomedical Engineering laboratory at Near East University.

Fortunately, these equipment are available in Ethiopia, as Mr. Jemal Edris Dawd explained to the department. Therefore, he has obligated to go to his country to collect all the required data from the hospitals in Addis Ababa, and the department has decided to allow him to go in 27 November 2019. Hence, the university respectfully asking all the concerned body in Ethiopian that you offer the necessary assistances and humble cooperation for the above-aforementioned work.

Thank you for your kind cooperation.

Yours faithfully,

Assoc. Prof. Dr. Dilber UZUN HASAIRIN
Biomedical Engineering, Near East University

APPENDIX XIV: TURNTIN SIMILARITY REPORT

APPENDIX XV: CURRICULUM VITAE

PERSONAL INFORMATION

Full Name – Jemal Edris Dawd
Nationality: Ethiopian
Date of Birth: October 22, 1978
Marital status: Married

EDUCATIONAL BACKGROUND

Degree	Institutions	Year of Graduation
M.Phil.	University of Ghana, Department of Medical Physics	2016
B. Edu.	Bahir Dar University, Department of Chemistry	2003

WORK Experience

Year	Place	Enrolment
Feb, 2017 – until now	Ethiopian Radiation Protection Authority	Radiation Research Team leader
Feb, 2014 – Feb, 2017	Ethiopian Radiation Protection Authority	Radiation Safety - Authorization Team Leader
Jan, 2011 – Feb, 2014	Ethiopian Radiation Protection Authority	Radiation Safety_ Notification and Information Senior Officer II
Oct, 2011 – Jan, 2011	Ethiopian Radiation Protection Authority	Radiation Safety_ Notification and Information Senior Officer I
Mar, 2010 – Oct, 2010	Ethiopian Radiation Protection Authority	Radiation Protection Senior Officer
Feb, 2010 – Mar, 2010	Ethiopian Radiation Protection Authority	Radiation Protection Officer V
Nov, 2003 – Feb, 2010	Ministry of Education	Teacher
Sept, 2006 – Oct, 2007	Ministry of Education	Head of the Department

FOREIGN LANGUAGES

English, fluent spoken and written

Short-term Training

- African Regional Workshop on Radiation Safety Information Management System (RASIMS) for English Speaking Countries – on week training from 30 May – 03 June 2011.
- Regional (AFRA) Training Course on Physical Protection of Radioactive Sources – one-week training from 23 – 27 May 2011, organized by IAEA.
- Regional (AFRA) Training Course for Regulators on Authorization and Inspection of Radiation Sources – Three-week training from 04 – 29 October 2010, organized by IAEA.
- Genie-2000 Gamma Spectroscopy Software Training Course – One-week training from 08 – 11 November 2011, organized by CANBERRA PACKARD central Europe GMBH.
- Physical Protection and Security Management of Radioactive Sources – one-week training from 13 -15 March 2012, 0rganized by U.S. Department of Energy.

- Quality management System Development and Implementation Based on ISO 9001:2008 – on week training from 07 – 11 March 2011 which was organized by Quality and Standard Authority of Ethiopia.
- Effective and sustainable regulatory control of radiation sources from March 4 – 8, 2013, organized by IAEA in collaboration with ERPA.
- Regional Training on Internal Dosimetry – one-week training from 26 – 30 August 2013, organized by IAEA.
- Balanced Scorecard(BSC) – one-week training from October 22 – 25, 2012, organized by Ethiopian Civil Service University

Publications in International Refereed Journals (in coverage of SSci/Sci-expanded):

- Jemal ED, Dilber UO, Ilker O. A review of diagnostic reference levels in Computed Tomography. Current Medical Imaging, 2021:17( ). Advance online publication: https://dx.doi.org/102174/1573405617666210913093839
- Jemal ED, Dilber UO, Ilker O. Phantom diagnostic reference levels. Journal of computer assisted tomography, 2021.

Online Publications:

- "Estimation of External Gamma dose and Annual Effective Dose of NORMs from mining activities of Kenticha Tantalum mines in Ethiopia": available in **www.globalscientificjournal.com**

Theses:

M.Phil.

- Jemal ED (2016). Evaluation of the Level of Norms and Associated Radiological Hazards & Risks from Mining Activities of Kenticha Tantalum Mines in Ethiopia. Partly Published Master Thesis, University of Ghana, Department of Medical Physics, and Graduate School of Nuclear and Allied Sciences, Accra, Ghana. Available in **http://ugspace.ug.edu.gh**

Courses Given:

High school:

- General chemistry (English)

Undertaken Projects:

- 2013 – 2017 – project counterpart- (RAF/9/053) Strengthening Technical Capabilities for Patient and Occupational Radiation Protection in Member States, International Atomic Energy Agency (IAEA), and Vienna, Austria.

HOBBY

- Reading, environment and earth especial green areas, Travel, Handiwork especially planting, playing with family

REFERENCE

- Prof. Dr. Aysa Gunay Kibarer
  - E-mail: aysegunay.kibarer@neu.edu.tr
  - Phone: +903922236464/5299
- Assoc. Prof. Dr. Dilber Uzun Ozshin
  - E-mail: dilberuzun.ozsahin@neu.edu.tr
  - Phone: +905338341513
- Assist. Prof. Dr. Ilker Ozsahin
  - E-mail: ilkerozsahin@windowslive.com
  - Phone: +905338767925

I hereby certify that all the information given above is true to the best of my knowledge.