EVALUATION OF DOSES DELIVERED DURING CT EXAMINATIONS BY DIFFERENT SCANNERS FOR PURPOSES OF INTERCOMPARISON AND DOSE OPTIMIZATION.

This thesis is submitted to the University of Ghana, Legon in partial fulfillment of the requirement for the award of **MPHIL Medical Physics degree.**

By

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JULY, 2017

DECLARATION

This thesis is the result of a research work carried out by Adam Bashiru in the Department of Medical Physics, University of Ghana, under the supervision of Prof. Cyril Schandorf, Dr. Steven Inkoom and Dr. Francis Hasford.

It is my conviction that, no part of this work has been submitted in part or whole to any other university or institution for the award of a diploma, or degree at any level. All other works and/or researches cited in this work have been duly acknowledged under references.

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ABSTRACT

This research study was aimed at performing dosimetry intercomparison on different CT scanners in the diagnostic radiology departments of Korle-Bu Teaching Hospital (KBTH), Sweden Ghana Medical Center (SGMC) and Global Medical and Imaging Center (GMIC).Using the standard body phantom and integrated ion chamber technique volume computed tomography dose index (CTDIvol) and Dose-Length Product (DLPs) within the phantom were evaluated. The ion chamber technique was applied to two 16 slice Siemens and one Toshiba Aquilion one CT scanners. CTDIvol and DLP values for the standard body polymethyl methacrylate (PMMA) phantom were estimated and comparison made with corresponding console displayed values for accuracy and also to deduce a suitable method for optimization of patients and occupationally exposed worker doses. Effective doses were also calculated. An intra and inter institutional comparison of measured doses and console displayed doses were performed. Chest protocol at Automatic Exposure Control (AEC) was applied during the scanning of the phantom. Estimated CTDIvol values (mGy) were 17mGy, 24mGy and 13.1mGy for SGMC, GMIC and KBTH respectively. These values deviated from the console displayed values by 24.1%, 22.9% and31.3% respectively. Similarly, estimated DLP values (mGy.cm) were 675mGy.cm,944mGy.cm and 419mGy.cm for SGMC, GMIC and KBTH respectively deviating from the console displayed values by 24.1%, 24.2% and 29% respectively. In terms of effective doses (E), the calculated E (mSv) values were 9.45mSv, 13.2mSv and 5.87mSv estimated from the DLPs from SGMC, GMIC and KBTH respectively using K ,the anatomy-specific dose coefficient expressing effective dose normalized to DLP in a standard CT dosimetry phantom of 0.014 mSv mGy-1 cm-1.The estimated doses were compared to other selected international Dose Reference Levels (DRLs) and were within range.

ACKNOWLEDGEMENTS

I enjoyed the support, help and encouragement of many people and institutions within the two year programme and in conducting this research. First and foremost, I would thank the Almighty Allah for His abundant grace, mercies and guidance throughout my study and during this research work. I am very grateful to my supervisors, for their advice and thoughtful suggestions on the work. Their intellectual inspirations and encouragement was instrumental and helpful. My heartfelt gratitude also goes to all the three institutions under this study, Sweden Ghana Medical Centre (SGMC), Korle Bu Teaching Hospital and Global Medical and Imaging Center who were there to assist and provide the needed materials for this research work. Special thanks to Mr. Mark Pokoo-Aikins for his enormous assistance, and to all the medical physicists at Radiological and Medical Sciences Research Institute (RAMSRI) of Ghana Atomic Energy Commission (GAEC) for helping with the procedures involved in the use of the PTW pencil Ion Chamber. To my dearest family I thank you so much for your continuous love and support.

DEDICATIONS

To the Almighty Allah for His love, blessings, protection and direction.

To my adorable wife, whose unending love makes every day a blessing

To my parents, who taught me the importance of honesty and perseverance

To my friends and family, who have all helped bring me to where I am today

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LIST OF ABBREVIATIONS

AAPM	American Association of Physicists in Medicine
ALARA	As Low As Reasonably Achievable
C (unit)	Coulombs
CAT	Computed Axial Tomography
CT	Computed Tomography
CTDI	Computed Tomography Dose Index
CTDI100	Computed Tomgraphy Dose Index measured over 100mm pencil Ion Chamber
CTDIFDA	Computed Tomography Dose Index defined by US Food and Drug Administration
CTDIvol	Volume Computed Tomography Dose Index
CTDIw	Weighted Computed Tomgraphy Dose Index
DLP	Dose – Length Product
DRLs	Dose Reference Levels
EBT	Electronic Beam Tomography
FDA	Food and Drugs Agency
FOV	Field of View
GAEC	Ghana Atomic Energy Commission
GMIC	Global Medical and Imaging Center

HU	Hounsfield Unit
IAEA	International Atomic Energy Agency
ICRP	International Commission on Radiological Protection
ISD	Inter-scan Delay
KBTH	Korle-Bu Teaching Hospital
mAs	milli ampere second
MDA	Multiple Computed Detector Arrays
MDCT	Multi Detector Computed Tomography
mGy	milli Gray
MRI	Magnetic Resonance Imaging
MSAD	Multiple Scan Average Dose
MSCT	Multi-Slice Computed Tomography
NRPB	National Radiological Protection Board
PMMA	polymethyl methacrylate
P1	Peripheral one
P2	Peripheral two
P3	Peripheral three
P4	Peripheral four

RAMSRI	Radiological and Medical Sciences Research Institute
SGMC	Sweden Ghana Medical Center
SSCT	Single Slice Computed Tomography
TLDs	Thermoluminesence Dosimeters
UNSCEAR	United Nations Scientific Committee on the Effects of Atomic Radiation
X-ray	CT X-ray Computed Tomography

LIST OF SYMBOLS AND CONSTANTS

Q represents charges recorded in coulombs

F med represents [Exposure to dose conversion factor] = 0.78 rad/R,

Cf represents [Electrometer/Ion Chamber calibration factor] = 1,

L represents [Ion Chamber length] = 100 mm,

T = Width of one slice or tomographic selection,

N represents [Number of slices or tomographic sections imaged in a single axial scan] = 16,

X represents [Estimated exposure] = Q/m

Pf = Pitch factor used

Mair represents [Mass of air irradiated] = ρair × vair

ρair represents [Density of air at Standard Temperature and Pressure] = 1.293 kg/m3

vair represents [Vol. of irradiated air for single slice] = (slice thickness/100 mm) × vc

vc represents [Vol. of Ion Chamber] = 3.14 cm3 = 3.14 × 10-6 m3

D (Z) Dose measured along the z-axis

E Effective dose

K Anatomy-specific dose coefficient expressing effective dose normalized to DLP in a standard CT dosimetry phantom

CHAPTER ONE

INTRODUCTION

1.1 Background

Computed Tomography (CT) is a non-invasive radiological procedure that uses X-rays to create detailed image slices of the body. It is an important imaging modality used in radiological diagnosis procedure with faster acquisition time compared to other imaging scanners, like Magnetic Resonance Imaging (MRI).

The discovery, development, and application of radiation to provide accurate, timely, and often life-saving medical diagnoses and treatment is without a doubt one of the greatest achievements of the last century. CT scanners are often the most widely used equipment in Radiology Departments, performing wide range radiological examinations. CT scanners come in varying forms with a direct relationship between the numbers of slices and the collective radiation dose. However, the dependence on CT examinations has led to an increase in both the overall number of examinations and the exposure to the individual patients having those examinations (Brenner & Hall, 2007). According to the latest UNSCEAR report, in developed countries with first level of health care, CT contributes up to 6% of the whole number of radiological examinations with 57 CT examinations per 1000 population as against 16 per 1000 worldwide (UNSCEAR, 200). The increased numbers of CT examinations together with relatively high individual doses are the reasons for rapid increase of the collective dose from CT. A report from the United Kingdom states that in a period of ten year (1991 and 2001), CT has more than doubled its contribution and is now responsible for 40% of the total dose to the population from medical X-rays (Hart and Wall,2002). Recently, several generations of multidetector row CT scanners have been introduced in developed countries, resulting in increased contribution to the collective dose, up to 67% in large hospitals in USA (Mettler Jr et al., 2000).

Recent publications and media reports have brought radiation exposures from CT examinations to the attention of the public. In 2007, Brenner and Hall reported on concerns of radiation dose to patients and the possibility of cancer occurrence. Increasing incidents of radiation over exposure in CT examinations has led to the interest of evaluating radiation doses in CT (UNSCEAR, 2013). Due to this increase in usage and widespread attention, the radiology and medical physics communities in particular have accepted renewed responsibilities in the realm of CT. These include minimizing the amount of radiation exposure while maintaining diagnostic results, ensuring patient exposures do not surpass the aforementioned optimized levels once they are determined, ensuring the appropriate selection of patients undergoing CT examinations, and providing practical yet accurate comparisons between radiation risks and clinical benefits. The measures, if well implemented, would ensure optimization of the usage of CT imaging throughout the medical community.

Computed Tomography Dose Index (CTDI) is defined as the integral of the single scan radiation dose profile along the z-axis that is normalized to the thickness of the imaged section (Shope et al., 1981). With corrections to scan spacing, CTDI can be used to estimate MSAD conveniently and in a standardized and way. CTDI is a volume-averaged measure (AAPM, 2008; Bauhs et al., 2008). CTDI can be measured using a 100-mm long pencil ionization Chamber and denoted $CTDI_{100}$ (AAPM, 2011). In this procedure, the ionization Chamber is placed in the CT head and body phantom and the CTDI is measured in the axial scan mode for an individual rotation of the x-ray source. Measurements of doses are taken at the center and at the periphery of the phantom and combined using a weighted average ($CTDI_w$) to give a single estimate of the radiation dose to the phantom. $CTDI_{vol}$ represents radiation output from the CT scanner to the phantom; $CTDI_{vol}$ measured in the CT head phantom is a reference to head CT and pediatric body CT in some cases and $CTDI_{vol}$ measured in CT body phantom is used as a reference to adult CT in the body (chest,

abdomen and pelvis) and can as well serves as a reference to the pediatric body CT (Shrimpton, 2004).

Due to the proliferation of CT centers with different CT machines and varying scan protocols at their disposal the question that arises is, do patients attending different diagnostic radiology centers and undergoing the same body scanning but assigned to different CT scanners receive the same radiation doses regardless of scanner designation? Measuring, evaluating and intercomparing the radiation doses delivered from the various institutions will ensure that irrespective of the scanner type the delivered doses are within tolerance level.

1.2 Statement of Research Problem

The increased use of CT scanning has led to concern about the magnitude of radiation doses arising from CT examinations. Multi-slice CT scanners, due to their superior scanning speed, are capable of delivering high patient doses unless technical factors are carefully selected by the operator. Clinical applications of CT techniques have continued to increase the dose to patients during recent decades, as CT examinations have come to provide higher quality X-ray imaging with substantial benefits in clinical diagnosis. Notwithstanding the potential benefits to the healthcare of patients using CT, the fundamental concern in radiological protection is the optimization of radiation exposure. CT protocols should be regularly reviewed to ensure that they are optimized to limit patient radiation dose while reliably achieving diagnostic quality images. Radiation doses for commonly performed examinations must be audited and periodically compared with published diagnostic reference levels (DRLs). If DRLs are consistently exceeded, radiation doses should be reviewed to determine whether radiation protection has been optimized. With quiet a number of radiological centres in Ghana and using different machines and operating protocols, ensuring that the delivered doses are within acceptable limits cannot be underestimated.

1.3 Research Objectives

The primary objective of the research study is to perform dosimetry intercomparison on different CT scanners in the diagnostic radiology departments of Korle-Bu Teaching Hospital, Sweden Ghana Medical Center and Global Medical and Imaging Centre using the standard body phantom.

The specific objectives of the study are:

- To determine measured doses (CTDI) in specific locations within the standard body phantom using ionization chamber.
- To perform comparison between measured doses with the ionization chamber and the scanner console displayed values.
- To perform inter-institutional comparison of measured doses
- To deduce a procedure for optimization of patient and occupationally exposed worker doses.

1.4 Relevance and Justification

This research would be useful in the area of patients' optimization and protection in CT scanning procedures in Ghana as a whole and particularly at Korle-Bu Teaching Hospital, Sweden Ghana Medical Centre and Global Medical and Imaging Centre. With concerns over high effective radiation doses to patients coming from CT albeit it representing only a small portion of all radiologic procedures as reported by Brix et al., 2009; Brenner & Hall, 2008; Mettler et al., 2000.,the need for patients' dose optimization lingers on. There is therefore the need to measure the amount of radiation dose given to patients during CT examinations with CT dosimetry measuring device and technique, and compare the measured values to the console displayed values in each center and then intercompare the three centers for the purposes of patients' dose optimization. This study therefore seeks to use the scan protocol for chest examination and at

automatic exposure control at each department to irradiate the phantom for the measurements to enable the evaluation and comparison of the measured and console displayed doses for the purposes of patient and occupationally exposed worker dose optimization. The study will also use known methodologies to measure and audit the results obtained from the selected diagnostic centers to ensure the accuracy in the delivery of radiation dose to patients during CT procedures.

1.5 Scope

The study involve measurements, evaluation and comparison of CT doses from different scanners in three diagnostic radiology centers, namely Korle-Bu Teaching Hospital(KBTH), Sweden Ghana Medical Centre(SGMC) and Global Medical and Imaging Centre(GMIC) using the PTW Ion Chamber and electrometer technique. Korle-Bu Teaching Hospital and Sweden Ghana Medical Centre are located in Accra, Ghana, and Global Medical and Imaging Centre is located in Kumasi. The dose descriptor, to be measured in this study, will be the CTDI on a standard body phantom. The dosimetric measurements were made using cylindrical CT dosimetry body phantom (32-cm in diameter). Dosimeter used for the dose measurements was the PTW ionization chamber and connected electrometer which recorded charges from CT procedures. Sets of measurements were taken at the specified locations (center and periphery) within the phantom and dosimetric estimation and comparison performance in the study. Mathematical equations and formulae from AAPM Report 96,1993 were used in the dose calculations from charges to exposure to dose.

1.6 Structure of the Thesis

This thesis is organized into five separate chapters. Chapter 1 provides the background of the research study, statement of the research problem, objectives, relevance and justification of the

study, scope and delimitation of the entire research work. Chapter 2 presents review of pertinent literature relevant to this study area. Materials and methods employed in obtaining the results of the study are presented in chapter 3, while chapter 4 contains the analyzed findings and discussions. Chapter 5 summarizes the study by providing the conclusions and recommendations.

CHAPTER TWO

LITERATURE REVIEW

2.0 OVERVIEW

This chapter provides a review of relevant literature applicable to this study.

2.1 INTRODUCTION

The first CT scanner (EMI Mark 1) was developed by Godfrey Hounsfield in 1972 at EMI laboratories. Its initial purpose was to eliminate the inherent problems associated with superposition of overlying anatomy in conventional planar radiography when a three-dimensional object is depicted on a two-dimensional medium. The anatomical superposition was eliminated with the introduction of CT by obtaining numerous individual projection images of a subject and then reconstructing the information using mathematical principles established by Radon in 1917 (Herman GT,1980).

Computed tomography (CT) is considered as a major source radiation exposure to the population from diagnostic X-ray examinations and an important tool in diagnostic radiology that provides high quality cross-sectional x-ray images of the body, albeit with relatively large patient doses. The increasing application of this modality has made a substantial impact on both patient care and also on population exposure. In recent times, CT scanners in clinical use have risen steadily over the past 25 years to reach a global total number of about 20,000 units, in 1997 with an associated annual total of some 67 million CT procedures.

The distribution of scanners is far from uniform, however, and there are significant variations in frequency of use between countries, even within the European Union. Practice is reported to have grown worldwide at a compound annual rate of about 4 % over the period 1993-1995, although national trends differ widely. CT already provides in many countries a substantial proportion of

the collective dose from medical X rays, for example around 35% in Germany and 40% in the UK. Notwithstanding the potential benefits to the health care of patient from CT, the of promoting ion of computed tomography (CT) scanners is based upon the he first CT scanner design, a single X-ray source and a single X-ray fundamental concern in radiological protection is the reduction of unnecessary exposures. These are examinations that are either unlikely to be helpful to patient management or involve doses that are not as reasonably practicable in order to meet specified clinical objectives.

Potential scope for improvement in the optimization of protection for patient undergoing CT has already been demonstrated in national surveys; for example, variations by factors of 10-40 have been observed in the typical dose between individual scanners for a given general type of procedure in the UK. Such variations are largely due to differences between hospitals in the local scanning technique employed. The concept of reference doses is recognized as a useful and practical way optimization of patient protection (Shrimpton, et al., 1998)

2.2 CT DOSE MEASUREMENT

Even though it is generally agreed in the medical fraternity that patients undergoing CT examination receive as minimum radiation dose as possible, the argument has always been the suitable method of measuring the radiation dose in CT.

With the introduction of the spiral CT in the early 1990s and subsequent introduction of four slices CT has revolutionaries modern CT scanning capable of providing high quality diagnostic information. This however, is generally described as being a high dose procedure (Kulama, 2004). Availability of 16 and 64 – slice CT scanners in addition to other models providing 320 slices with large area, have very much altered the clinical use of CT as new clinical applications have evolved, taking into account, the use of vascular and cardiac exams, perfusion imaging and whole body imaging.

A number of factors affect the radiation dose delivered to patients in CT imaging. These include; the radiologist, application specialist and technician who choose the parameters for the tube current, tube potential (kVp) and the differences in scanning parameters (Catalono, et al., 2007). In MSCT examination, the radiation dose delivered to patients are quite high, this therefore call for the need to keep radiation as low as reasonable achievable (ALARA). Thus, extra careful is therefore needed to in reducing radiation dose and maintaining an image quality that is acceptable for diagnosis (Jurik et al., 1997). The radiation dose to the patient will differ depending on the make and model of MSCT scanner in terms of variations in MSCT geometry, filtration and the awareness of the image quality from the CT scanner. This is where the needs are to be balanced between radiation dose and image quality (Tsapaki & Rehani, 2007).

It is necessary for manufacturers, radiologists, technologists and physicists to work together to find a plan to decrease patient dose (ALARA Principle). Advances in the use and development of MSCT scanners have resulted in the ability to provide images of adequate quality, with a resultant low radiation dose to the population (Kalra et al., 2004), however this is not commonly understood in the practical medical imaging field (Tsapaki & Rehani, 2007).

Many CT parameters have been briefly studied, and manufacturers have adopted an auto mA protocol for minimizing the radiation dose whiles keeping image quality constant. It is commonly believed that a change in the kVp is difficult, because any change in the kVp would have major impact on the image quality and dosage (McNitt-Gray & Geffen, 2006). The aim of optimization in diagnostic radiology is to achieve optimal parameters and protocols needed to create high image quality with the lowest possible dose to patients. As a result of this, optimization of radiation dose is necessary for each particular x-ray unit and for each x-ray examination. The optimization procedure requires an evaluation of patient dose and image quality (Mahesh, 2009).

2.2.1 CT DOSE DESCRIPTORS

A number of dosimetric quantities have been developed over the past decades to attempt to best determine the dose delivered during a CT exam, with varying degrees of applicability and success.

2.2.2 Computed Tomography Dose Index (CTDI)

The most widely used dose descriptor for CT is the computed tomography dose index (CTDI). The CTDI represents the average absorbed dose, along the z-axis, from a sequence of contiguous irradiations. The CTDI is measured from one axial CT scan, and is calculated by dividing the integrated absorbed dose by the nominal total beam collimation (Jessen et al., 2000). The CTDI is continually measured in the axial scan mode for a single rotation of the X-ray source, and theoretically approximates the average dose within the central region of a scan volume consisting of multiple, contiguous CT scans [multiple scan average dose (MSAD)] for the case where the scan length is sufficient for the central dose to approach its asymptotic upper limit (Nagel, 2000). The MSAD represents the average dose over a small interval (- I/2, I/2) about the centre of the scan length (z = 0) for a scan interval I, but requires multiple exposures for its direct measurement. The CTDI offered a more convenient yet nominally equivalent method of estimating this value, and required only a single scan acquisition, which in the early days of CT saved a considerable amount of time. Mathematically, CTDI is defined as

$$CTDI = \frac{1}{NT}\int_{-\infty}^{\infty} D(Z)dz \ldots\ldots\ldots\ldots\ldots (1)$$

Where,

D (z) = the absorbed dose profile along the axis of rotation of the scanner (z-axis),

N = the number of tomographic sections imaged in a single axial scan. This is equal to the number of data channels used in a particular scan. The value of N may be less than or equal to the maximum number of data channels available on the system, and

T = the width of the tomographic section along the z-axis imaged by one data channel. In multiple detector-row (multi-slice) CT scanners; several detector elements may be grouped together to form one data channel. In single-detector-row (single-slice) CT, the z-axis collimation (T) is the nominal scan width.

2.2.3 $CTDI_{FDA}$

Owing to variations in the radiation beam geometry, other definitions aimed at making this dosimetric quantity more applicable to the advances in CT imaging have been defined. Amongst them is the Food and Drug Administration (FDA) definition.

$$CTDI_{FDA} = \frac{1}{NT}\int_{-7T}^{+7T} D(D)dz \ldots\ldots\ldots\ldots (2)$$

Where,

N is the number of slices obtained in the scan and T is the nominal slice width.

The limits of integration were changed to ± 7T to normalize CTDI measurements. This was because the previous limits of ±∞ were not physical limits. In addition, it had been noted that the measured dose would depend upon the medium being imaged .The equivalence of the MSAD and the CTDI necessitates that all contributions from the tails of the radiation dose profile be included in the CTDI dose measurement. The exact integration limits required to meet this criterion depend upon the width of the nominal radiation beam and the scattering medium. The FDA introduced the integration limits of ± 7T, where T represented the nominal slice width [United

States FDA Code of Federal Regulations, 1984). Remarkably, the original CT scanner, the EMI Mark I, was a dual-detector row system. Hence, the nominal radiation beam width was equal to twice the nominal slice width (i.e. N x T mm). To account for this, the CTDI value must be normalized to 1/NT as in equation (2) above.

Unluckily, the limits of integration were not equally conveyed in terms of NT, permitting for the potential underestimation of the MSAD by the CTDI. For the technology available circa-1984, the use of NT in the integration limits was deemed unnecessary at the time (Dixon, 2006). The scattering media for CTDI measurements were also standardized by the FDA (United States FDA Code of Federal Regulations, 1984). These comprise of two polymethyl methacrylate (PMMA, e.g. acrylic or Lucite) cylinders of 14-cm length. To estimate dose values for head examinations, a diameter of 16 cm is to be used. To estimate dose values for body examination, a diameter of 32 cm is to be used. These are typically referred to, respectively, as the head and body CTDI phantoms. These changes allowed meaningful comparisons to be made based on measurements taken from several scanners.

2.2.4 $CTDI_{100}$

$CTDI_{100}$ is a dose descriptor that represents the collective multiple scan dose at the center of a 100-mm scan and underestimates the accumulated dose for longer scan lengths. It is therefore lesser than the equilibrium dose or MSAD. The $CTDI_{100}$, like the $CTDI_{FDA}$, requires integration of the radiation dose profile from a single axial scan over specific integration limits. In the case of $CTDI_{100}$, the integration limits are ± 50 mm, which corresponds to the 100 mm length of the commercially available "pencil" ionization Chamber (AAPM, 1990; AAPM, 1993; McNitt-Gray, 2002).

$$CTDI_{100} = \frac{1}{NT}\int_{-50mm}^{+50mm} D(Z)\,dZ \ldots\ldots (3)$$

The use of a single, consistent integration limit avoided the problem of dose overestimation for narrow slice widths (e.g. < 3 mm) (AAPM, 1990). CTDI100 is acquired using a 100-mm long, 3-cm3 active volume CT "pencil" ionization Chamber and the two standard CTDI acrylic phantoms [head (16-cm diameter) and body (32-cm diameter)] (AAPM, 1990, United States FDA Code of Federal Regulations, 1984). The measurement must be performed with a stationary patient table. The pencil chamber of active length ℓ is not really measuring air kerma, but rather the integral of the single rotation dose profile D(z).

2.2.5 Weighted CTDI (CTDIw)

The CTDI varies across the field-of-view. For example, for body CT imaging, the CTDI is typically a factor or two higher at the surface than at the centre of the field of view. The average CTDI across the field-of-view is estimated by the Weighted CTDI (CTDIw) (Jessen et al., 2000, International Electrotechnical Commission, 2002, Leitz et al., 1995), where

$$CTDI_W = \frac{1}{3}CTDI_{100,center} + \frac{2}{3}CTDI_{100,periphery} \ldots\ldots (4)$$

The values of 1/3 and 2/3 approximate the relative areas represented by the centre and periphery values (Leitz et al., 1995).

CTDIw is a useful indicator of scanner radiation output for a specific kVp and mAs. According to IEC 60601-2-44, CTDIW must use CTDI100 as described above and an f-factor for air (0.87 rad/R or 1.0 mGy/mGy) (Jessen et al., 2000, International Electrotechnical Commission, 2002).

2.2.6 Volume CTDI (CTDIvol)

To represent dose for a specific scan protocol, which usually involves a series of scans, it is essential to take into account any gaps or overlaps between the X-ray beams from consecutive rotations of the X-ray source. This is achieved with the use of a dose descriptor known as the Volume CTDI (CTDIvol.).

$$CTDI_{vol} = \frac{NT}{I}.CTDI_w \text{.....(5)}$$

Where,

I = the table increment per axial scan (mm) (International Electrotechnical Commission, 2002). Since pitch is defined (International Electrotechnical Commission, 2002) as the ratio of the table travel per rotation (I) to the total nominal beam width (N*T) (International Electrotechnical Commission, 2002, McCollough and Zink, 1999):

But

$$\frac{I}{NT} = pitch \text{.....(6)}$$

Therefore, volume CTDI can be expressed as:

$$CTDI_{vol} = \frac{CTDI_w}{Pitch} \text{.....(7)}$$

Whereas CTDIw represents the average absorbed radiation dose over the x and y directions at the centre of the scan from a series of axial scans where the scatter tails are negligible beyond the 100. $CTDI_{vol}$. is therefore pitch corrected $CTDI_w$

2.2.7 Dose-length product (DLP)

Of all the various CTDI measurements discussed above, none of them takes into consideration the number of slices obtained during an actual CT examination. In fact, they are all involved in attempts to determine a normalized dose index for a single slice. Meanwhile, clinical CT examinations always involve acquisition of more than one slice. With a direct relationship between the total number of axial images acquired and the actual dose that a patient will receive from a given CT examination, there was therefore the need to introduce another quantity known as the dose length product (DLP) to account for the total energy delivered.

For a better representation of the overall energy delivered by a given scan protocol, the absorbed dose can be integrated along the scan length to compute the dose-length product (DLP) (Jessen et al., 2000), where DLP (mGy-mm) = CTDIvol. (mGy) · scan length (mm) ……..(8)

The DLP reflects the total energy absorbed (and thus the potential biological effect) attributable to the complete scan acquisition. Thus, an abdomen only CT exam might have the same CTDIvol as an abdomen/pelvis CT exam, but the latter exam would have a greater DLP, proportional to the greater z-extent of the scan volume. In helical CT, data interpolation between two points must be performed for all projection angles. Thus, the images at the very beginning and end of a helical scan require data from z-axis projections beyond the defined "scan" boundaries (i.e. the beginning and end of the anatomic range over which images are desired). This increase in dose-length product due to the additional rotation(s) required for the helical interpolation algorithm is often referred to as over-ranging. For MDCT scanners, the number of additional rotations is strongly pitch dependent, with a typical increase in irradiation length of 1.5 times the total nominal beam width.

2.2.8 Effective Dose Estimation

Effective dose, E, is a dose descriptor that reflects the difference in biological sensitivity common to all modalities utilizing ionizing radiation. It is a single dose parameter that reflects the risk of a non-uniform exposure in terms of an equivalent whole-body exposure (Mullenders, et al., 2009). A broad estimates of effective dose (E) may be derived from values of DLP for an examination using appropriately normalized coefficients designed by European commission.

The International Commission on Radiological Protection (ICRP) has recommended that effective dose should be used for dose assessment for planning, optimization and compliance with dose limits for regulatory purposes(ICRP, 2007). In this study, the effective dose was estimated using values of DLP for CT examination, and normalized conversion factors for adults based on European guidelines for CT (European Commission, 1999), using the mathematical equation

$$E \approx k\,(DLP) \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots \quad (9)$$

Where

k is the anatomy-specific dose coefficient expressing effective dose normalized to DLP in a standard CT dosimetry phantom. Adult k values adopted for chest in this study was 0.014 mSv mGy-1 cm-1.

2.3 CT Dose measurements inter-comparisons

Implementation of an intermittent if not a routine patient-dose audit of CT examinations in diagnostic radiological cannot be under estimated. This will provide updated information as to how CT radiation doses for all examinations are within international diagnostic reference levels and contribute towards establishing a recommended national reference dose (NRD) levels

agreeable with the standard international DRLs. Inter-institutional level inter-comparison also necessary. Due to the complexity of modern day CT machines and concerns associated with a variety of uncertainties in their dose delivery and the implications on patients' radiation exposure. Thus, comprehensive quality assurance (QA) procedures are essential to check delivered doses during CT examinations. For patient related QA, the validation of the CT console displayed dose descriptors (CTDIvol, DLP)and the estimated doses measured with CT dosimetric phantom(s) is important. In addition to the patient and machine related QA procedures performed by the institutions themselves, an audit organized by an independent external body is a fundamental step in any dosimetry QA program. In this study choice of CTDIvol for the inter-comparison works was informed by its ability to:

- provide information about the amount of radiation used to perform the study
- provide a way to track across patients and protocols for QC purposes
- allow for comparison between scanners and facilities.

2.3.1 Detectors

In clinical practice, the main detectors used for dosimetry are ionization chambers and semiconductors. Ionization chambers are commonly used for phantom measurements because of their accuracy and practicality (IAEA Technical Reports Series No. 277, Vienna, Austria 1987). Semiconductor diodes are routinely used for absorbed dose measurements of clinical studies. These measurements make use of the advantages of semiconductors, such as ease of handling and the dose determination in real time (Rikner et al., 1987). Thermoluminescent (TL) dosimeters are widely used for radiation detection in the fields of environmental, industrial and personnel applications, just to mention a few.

2.3.2 Phantoms

In order to evaluate radiation doses resulting from different CT examinations, direct measurements using Physical anthropomorphic phantoms representing adult male and female patients or paediatric patients and ion chambers, dose profiler thermoluminescent dosimeters (TLDs) among others have been used to measure patient doses from CT. These phantoms are usually cut into sections, which contain holes for the position of dosimeters. The PMMA head and body phantom and water-filled phantom are mostly used for CT dosimetric intercomparison studies:

2.3.2.1 *Water (distilled water)*

A water-filled, 30-cm diameter phantom, 50 cm long corresponds to attenuation and absorption of the average-sized adult body. The phantom is designed to be transported empty, and once placed on the table, it can be quickly filled or emptied in 2 minutes with a small pump operating from a room sink as a reservoir. A second water-filled phantom, of 20-cm diameter, could be used to correspond to the attenuation properties of an adult head and pediatric body.

2.3.2.2 *Polymethyl methacrylate (PMMA)*

While dose in a 32-cm diameter PMMA phantom would be approximately 30% lower than the dose in a 30-cm diameter water cylinder, there is a large stock of PMMA phantoms already in use in the field, and such phantoms are assembled contiguously for requisite lengths. Dose in 16-cm diameter PMMA is relatively close to the value in a 20-cm cylinder of water.

2.4 AUTOMATIC EXPOSURE CONTROL (AEC) IN CT

Automatic exposure control is a CT imaging technique that performs automatic modulation of tube current in the x, y plane (angular modulation), or along the scanning direction, z-axis, (longitudinal modulation), or both (combined modulation) (Kalra et al., 2005a). The modification is done with respect to the patient's size, shape, weight and attenuation of body parts being scanned. In its mode of operation, the radiologic technician is required to select the desired image quality level and the system adjust the tube current to obtain the predetermined image quality with improved radiation efficiency to produce the desired image quality, while limiting the radiation dose to the patient (Söderberg, 2008). AEC system can in most cases reduce radiation dose by typically between 10-50 percent whiles maintaining a consistent image quality (Kalender, 2005b). Results from studies have shown that, the use of AEC systems reduced patient dose by about 35%-60% for the body and 18% for the neck, across all sizes of patient, compared with fixed tube current techniques. These dose reductions vary between different studies and depend on the tube current being used for the fixed technique and the size of the patient (Lee et al., 2009; Rizzo et al., 2006).

2.4.1 Principles of AEC System for Different CT Manufacturers

Each manufacturer of CT systems has developed different AEC techniques and application capabilities (Söderberg, 2008). With their main purpose of maintaining image quality, control of patient radiation dose, avoidance of photon starvation artefacts and reduced load on the x-ray tube (Kulama, 2004). There exist a number but for the purpose of this work the discussion is limited to those used by the scanners employed in this study.

2.4.2 Siemens - CARE Dose 4D

Siemens use a combined tube current modulation system called CARE Dose 4D (Söderberg, 2008).The system works with automatic tube current modulation of the patient's size and shape

together with real time, online, controlled tube current modulation during each tube rotation (Siemens, 2004). Based on a single CT localizer radiograph, "topogram", anterior-posterior or lateral attenuation profile (size, anatomic shape and attenuation at each position) along the patient's long axis (z-axis) is measured in the direction of the projection and estimated for the perpendicular direction with a mathematical algorithm (Söderberg, 2008. The CARE Dose 4D has adequate image noise according to Siemens modulation, which differs depending on the patient's size and shape. The operator can choose the level of tube current, which can be selected according to the patient's size, using 'weak', 'average' or 'strong' settings to control the amount of mA supplied. The CARE Dose 4D modulations are able to provide a lower tube current to keep image noise consistent regardless of the patient size (Keat, 2005).

2.4.3 Toshiba - Sure Exposure 3D

The Toshiba CT scanners use a combine modulation system called Sure Exposure 3D. The Sure Exposure system gives the operator two ways of setting the required image quality: standard deviation (SD) of CT numbers or image quality level (Söderberg, 2008). These methods are based on measurements of SD of pixel values measured in a patient-equivalent water phantom (McCollough et al., 2006). With the sure Exposure 3D system, the user specify the SD value for the HU image noise as well as the maximum and minimum tube current. The image noise intensifies when the tube current is low leading to very poor images and very high tube current cause high radiation exposure with minimal noise level (Söderberg, 2008). The image acquisition process is done by acquiring a frontal and a lateral localizer radiograph, known as "scanogram" of the patient. The scanogram is then used to map the selected image quality with respect to the tube current values. The sure exposure 3D makes use of the frontal and lateral diameters and the detector intensities to determines the oscillating tube current modulation during each gantry rotation (Söderberg, 2008).

2.5 OPTIMIZATION OF PROTECTION IN CT

CT examination is a "high dose" procedure. A series of clinical factors play a special part.These include availability of adequate clinical information, including the records of previous imaging investigations. In certain applications prior investigation of the patient by alternative imaging techniques might be required. An additional training in radiation protection is required for radiologists. Once a CT examination has been clinically justified, the subsequent imaging process must be optimized. Optimal use of ionizing radiation involves the interplay of the imaging process , the diagnostic quality of the CT image , the radiation dose to the patient and staff and the choice of radiological technique. CT examinations should be performed under the responsibility of a radiologist according to the national regulations. Standard examination protocols should be available. These with effective supervision may aid radiation protection by terminating the examination when the clinical requirement has not been satisfied. Quality Criteria can be adopted by radiologists, radiographers, and medical physicists as a check on the routine performance of the entire imaging process

2,5.1 PATIENT DOSE PROTECTION

Nedical imaging procedures should be appropriate and conducted at the lowest radiation dose consistent with acquisition of the desired information. Any discourse of risks related to radiation dose from medical imaging procedures should be accompanied by acknowledgement of the benefits of the procedures

2.5.1.1Lead Shields

The proper use of lead shielding on patients can be confusing because of differing reports in literature. For some time now, it has been common practice to shield all patients of reproductive age receiving ionizing radiation from radiography(Staskiewiez et al.,2006). As CT became common and radiographers cross-trained into roles as CT technologists, they have had to revise how they shield patients. Methods of shielding patients during CT scanning differ from shielding in radiography. In conventional radiography, the x-ray tube remains stationary, which makes it

easier to position lead shielding accurately. The technologist places shields between the patient and the x-ray tube to protect the patient from leakage radiation and misdirected primary radiation, or the technologist places the shield over radiosensitive organs to protect them from scatter radiation(McCollough et al.,2007). Proper use of shielding can reduce patient dose, but if the shielding is placed over the anatomy of interest erroneously, a repeat examination will be required and will result in increased patient dose(Minnigh & Gallet,2009).Therefore, shielding in CT should be approached differently than it is for shielding in conventional radiography and fluoroscopy. During a CT scan, the x-ray tube and detectors rotate 360° around the patient. Placing a shield only on one side of patients does not fully shield them. The shielding should be wrapped around the patient. A patient dose savings of 5% to 78%

can be achieved when lead shielding is placed carefully just outside of the area of interest(Iball & Brettle,2011).Incorrect use of shielding in CT has the potential to increase patient dose more than just requiring a repeat exposure because it is covering important anatomy. The automatic kV selection and the current modulation programs use localizer information. If the lead apron is in the localizer image, the dose optimization programs falsely can identify the patient as being larger or more attenuating, such as more muscular. Therefore, the current modulation and kV selection algorithms might select higher technical factors than are required, which would result in higher dose to the patient. When the scan area of interest is close to the patient's gonads, technologists can acquire the localizer image before placing the lead apron around the patient to prevent the current modulation and automatic kV selection from selecting incorrect higher techniques. This approach still includes the risk of placing the lead apron in the active scan area, which forces higher current in that area, resulting in unnecessary dose to the patient or obscuring the anatomy of interest. It has been suggested that shielding in CT actually provides little dose savings, especially when the shielding is far from the scan field of view. For example, wrapping a shield around the waist of a patient having a head scan provides more reassurance to the patient than

exposure reduction(McCollough et al.,2007). The scatter produced within the head reaches the gonads via internal scatter. A shield wrapped around the patient cannot intercept it.

2.5.1.2 Bismuth Shields

Bismuth is a chemical element with the symbol Bi and atomic number 83. It has one more proton than lead, which has an atomic number of 82. According to the AAPM, Bismuth shields are easy to use and have been shown to reduce dose to anterior organs in CT scanning. owever, there are several disadvantages associated with the use of bismuth shields, especially when used with automatic exposure control or tube current modulation. Other techniques exist that can provide the same level of anterior dose reduction at equivalent or superior image quality that do not have these disadvantages. The AAPM recommends that these alternatives to bismuth shielding be carefully considered and implemented when possible(AAPM,2016). The use of organ-specific current modulation is a better alternative to the use of bismuth shielding. The current is reduced automatically as the x-ray tube passes anteriorly over the patient, thus reducing breast dose. As the x-ray tube passes posteriorly, no attenuation of the beam occurs, as with shielding(Vollmar & Kalender,2008).

2.5.2 STAFF DOSE PROTECTION

2.5.2.1 ***Leaded Apparel***

Most often, CT technologists are not in the room during scanning, which is a best practice because of the high amount of Compton scatter inside the room. One exception is during interventional procedures, such as biopsies. During these examinations and procedures, personnel are in the active scanning room, and leaded garments should be worn to protect against scatter radiation.

2.5.2.1 Structural Shielding

Because staff members usually remain in the control room during CT scans, their primary protection comes from the structural shielding in the walls. Regulations vary from state to state, but shielding design should be performed by a qualified medical physicist during a room's design phase. This should be followed by conducting a safety survey after construction and before the equipment is put to clinical use. Shielding designs take into account the workload, use of adjacent areas, and how often adjacent areas are occupied. For example, if a waiting room is located next to the CT scanner, the area is considered uncontrolled and must be shielded to keep the doses below 1 mSv per year. The physicist calculates the amount of shielding required to reduce the exposure in the adjacent areas to acceptable dose levels using guidelines set out by NCRP Report No. 147.(NCRP,2004) After installation of the shielding and CT scanner, the physicist conducts a safety survey to verify that the appropriate amount of shielding was installed, including lead-equivalent windows. Technologists should not assume that lead in the walls stops all radiation, only that it lowers doses to acceptable levels. Technologists are considered occupationally exposed workers and, therefore, are permitted an annual dose of 50 mSv or less. The shielding in the walls is designed to prevent technologists from reaching close to that limit, but Compton scatter radiation still passes through the walls.(NCRP,2004) A common misconception is the belief that the control room must have a door or other solid structure separating it from the scan room. The scattered photons do not bend around corners; therefore, it is acceptable to have an opening between the control room and the scan room. Still, technologists should ensure that shielded barriers are between them and the CT equipment during active scanning.

CHAPTER 3

MATERIALS AND METHODS

This chapter presents a detailed account of the method employed in the dose estimation in this study, inter institutional dose comparison, as well as procedure for the data analysis.

3.1 MATERIALS

CT radiation dose inter-comparison was carried out for three health centers equipped with CT machines (Siemens – Emotion 16 slices and Toshiba- Aquilion one 640 slices) at SGMC, GMIC and KBTH respectively. The dose measurements were done using a pencil shaped ion chamber with integrated electrometer and the body PMMA dosimetry phantom. Technical characteristics of the CT scanners are presented in Table 3.1.

Table 3.1: Technical characteristics of the CT scanners

Institution	Manufacturer	Model	Date of Installation	FAD (cm)	Detector Type
SGMC	Siemens	Emotion	2011	70	16 row
GMIC	Siemens	Emotion	2013	70	16 row
KBTH	Toshiba	Aquilion	2012	70	640 row

3.1.1 The CT Scanners

- The Aquilion One (Toshiba Medical System) CT scanner is a 320-detector row with 640-slice reconstruction, and has a maximum beam width of 160 mm as shown in Figure in 3.1. The CT scanner is equipped with a gantry and has a rotation speed of 0.275 seconds. It has a generator with a power rating capacity of 100 kW, covering 16 cm and thinnest slices at 500 microns (0.5 mm). The system can accommodate larger patients with its 78 cm bore and fast rotation, including bariatric and patients with high heart rates. The scanner has automatic exposure feature (Sure Exposure 3D) and can also be operated manually.

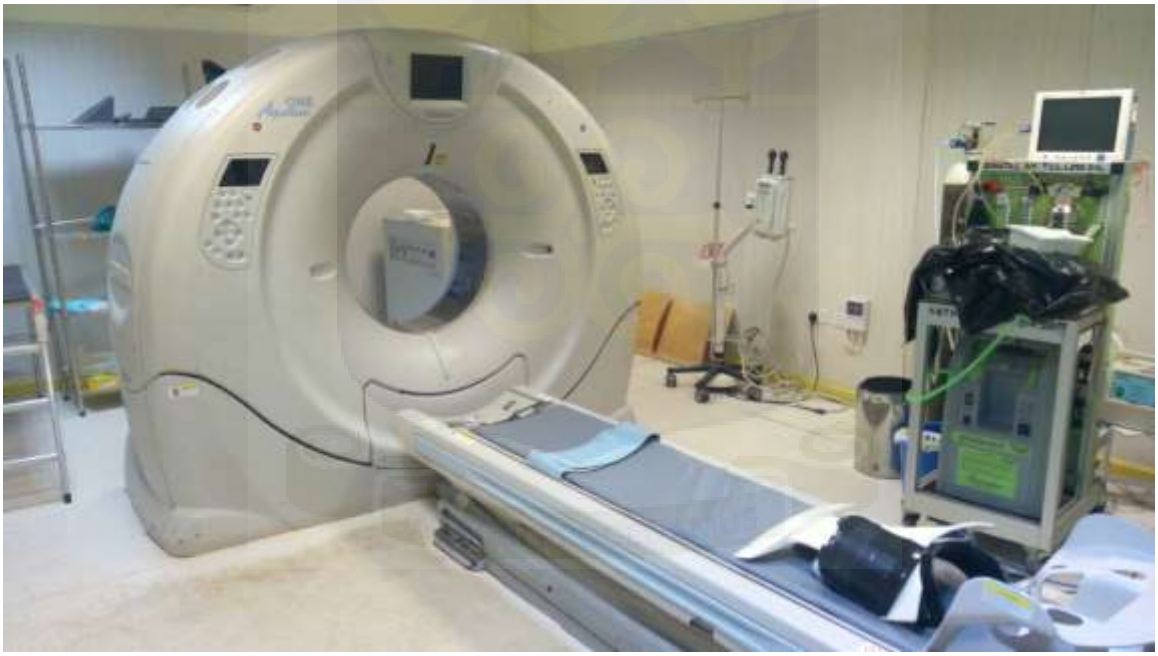

Figure 3.1: Toshiba Aquilion One, CT Scanner [Field work, 2017]

- The CT scanners used at SGMS and GMIC is the Siemens Somatom Emotion CT scanner (Siemens Healthcare, Forchheim, Germany) and (Siemens DE, Muenchen, Germany) with a 16 channel detector configuration a small focal spot capable of making a 16 × 1.2 mm multi-slice imaging as shown in Figures 3.2-3.3.

The scanner has tube potentials of 80 kVp, 110 kVp and 130 kVp and equipped with up to date CareDose4D as its automatic exposure control feature and can also be operated by manually selection of exposure parameters.

Figure 3.2: Siemens CT Scanner Used at SGMC [Field work, 2017]

Figure 3.3 Siemens CT Scanner used at GMIC [Field work, 2017]

3.1.2 The Body Phantom

A cylindrical CT dosimetry body phantom (32-cm in diameter) made from acrylic Polymethyl Methacrylate (PMMA) material was used to mimics an adult thorax region for dose measurements from the three CT scanners. It has five 1 cm holes within it for insertion of an Ion Chamber. The holes are located at the center and 1-cm depth at the peripheral of the phantom, specifically at 12 –, 3 –, 6 – and 9 – O'clock positions. Figure 3.4 shows a picture of the body phantom used in this study.

Figure 3. 4 Picture of the CT body phantom[Field work 2017]

3.1.3 PTW Ion Chamber and Electrometer

The Ion Chamber is a CT chamber type (TM30009) manufactured by PTW in Freiburg, Germany, is a vented cylinder chamber that is made for the measurements of photon radiation in computed tomography. The Ion Chamber is a pencil type chamber used for taking measurements within a CT body or head phantom or sometimes in air. The pencil Ion Chamber is 100mm in length, sensitive along its entire length and shows a homogeneous response over the whole length. It was calibrated against an electrometer, PTW-DIADOS E, type (T11035). The Ion Chamber was used together with the electrometer for the measurement of CTDI values. The electrometer records charges in Coulomb when used with the Ion Chamber. Figure 3.5 below shows a picture of the CT Ion Chamber with an electrometer.

Figure 3.5: Picture of Ion Chamber with integrated Electrometer [Field work 2017]

3.2 METHODOLOGY

CT dosimetry technique using the standard CT body phantom and the PTW Ion Chamber with electrometer (PTW Diados E, type (T11035) was implemented for determining CTDI values. The protocol involved parameters for CT thorax (chest) examination that were routinely used in each institution. The dose measurement procedure for the study begun by positioning the CT body phantom on the CT couch. The phantom was positioned at the isocenter of the CT scanner and the long axis of the phantom was aligned with the z-axis of the scanner with the aid of the

lasers installed in the room. All measurements were done by irradiating the phantom using Automatic Exposure Control (AEC).

3.2.1. Dose Measurements with the PTW Ion Chamber

Dose measurements were done on the CT scanners at the facilities using the body phantom and pencil shaped ion Chamber (model, TM30009) connected to DIADOSE electrometer (PTW Freiburg, Germany) via a cable connection. In the dose measurement, the body phantom was placed on the CT couch and centered at the center of the scanner with the long axis of the phantom aligned with the *z*-axis of the scanner.

Initially, a topogram image of the body phantom was obtained for selection of the required scan volume. The horizontal and vertical lasers in the CT room were switched on to ensure proper alignment of the phantom with ion chamber connected on the couch. The ion chamber was placed in the center hole of the phantom and a scout view image used to select the volume or slice to be imaged. The Dosimeter readings was initially set to zero and exposure in axial mode at thorax CT scan technique used to scan the phantom. Measurements were repeated similarly at the peripheral holes of the phantom at 12, 3, 6 and 9 O'clock positions respectively. Charges (in Nano coulombs) were measured in the central and peripheral holes by changing the ion chamber position from one hole to the other as shown in Figure 3.6. In line with radiation protection principle, the CT room interlock was closed up during the time of exposure for radiation protection purpose. The electrometer readings were recorded in coulomb (C), but there was no need to employ equation (3.1) for temperature and pressure correction since the average temperature of 25.6^{o}C and 100.26 mmHg recorded in the experimental rooms were within the a range specified by the Ion Chamber .The recorded charges were converted into exposure (rad)

using equation 3.2. The CTDI derivatives in (mGy) were calculated using equations, 3.3, and 3.4 respectively.

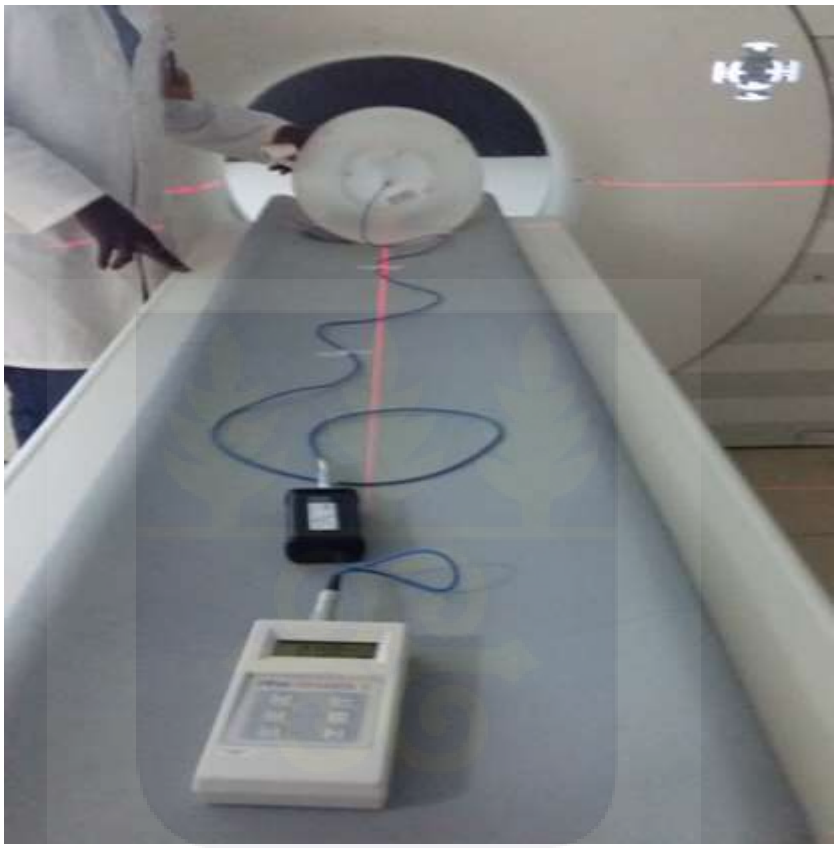

Figure 3. 6 Experimental set-up for measurement of CTDI [Field work 2017]

$$K_{T,P} = \frac{273.2+T}{273.2+20} X \frac{101.3}{P} \quad \text{.....3.1}$$

Where,

T – Temperature measured in the study room

P – Pressure in the room

$K_{T,P}$ – correction for pressure and temperature

$$X_{(rad)} = \frac{Q}{m_{air}}\left(C/_{Kg}\right) = \frac{Q}{m_{air}} \cdot \frac{1}{2.58\times10^{-4}}\,(R).f_{med}(rad/_{R}) \ldots\ldots\ldots\ldots\ldots (3.2)$$

$$CTDI_{100} = \frac{X_{(rad)}.C_f.L_{(mm)}}{N.T_{(mm)}} \ldots\ldots\ldots\ldots\ldots (3.3)$$

$$CTDI_w = \frac{1}{3}CTDI_{100centre} + \frac{2}{3}CTDI_{periphery} \ldots\ldots\ldots\ldots\ldots (3.4)$$

$$CTDI_{vol} = \frac{CTDI_w}{pitch} \ldots\ldots\ldots\ldots (3.5)$$

$$Pitch = \frac{I}{N.T} \ldots\ldots\ldots (3.6)$$

Where, Q represents charges recorded in coulombs

f_{med} represents [Exposure to dose conversion factor] = 0.78 rad/R,

C_f, represents [Electrometer/Ion Chamber calibration factor] = 1,

L represents [Ion Chamber length] = 100 mm,=1cm

T = Width of one slice or tomographic selection,

N represents [Number of slices or tomographic sections imaged in a single axial scan]

X represents [Estimated exposure] = Q/m

m_{air} represents [Mass of air irradiated] = $\rho_{air} \times v_{air}$

ρ_{air} represents [Density of air at Standard Temperature and Pressure] = 1.293 kg/m^3

v_{air} represents [Vol. of irradiated air for single slice] = (slice thickness/100 mm) × v_c

v_c represents [Vol. of Ion Chamber] = 3.14 cm^3 = 3.14 × 10-6 m^3

3.2.2 Effective Dose Estimation

Effective dose, E, is a dose descriptor that reflects the difference in biological sensitivity common to all modalities utilizing ionizing radiation. It is a single dose parameter that reflects the risk of a non-uniform exposure in terms of an equivalent whole-body exposure (Mullenders, et al., 2009). Broad estimates of effective dose (E) may be derived from values of DLP for an examination using appropriately normalized coefficients designed by European commission.

The International Commission on Radiological Protection (ICRP) has recommended that effective dose should be used for dose assessment for planning, optimization and compliance with dose limits for regulatory purposes (ICRP, 2007). In this study, the effective dose was estimated using values of DLP for CT examination, and normalized conversion factors for adults based on European guidelines for CT (European Commission, 1999), using the mathematical equation

$$E \approx k\,(DLP) \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots \quad (3.4)$$

Where

k is the anatomy-specific dose coefficient expressing effective dose normalized to DLP in a standard CT dosimetry phantom. Adult k values adopted for chest in this study was 0.014 mSv mGy-1 cm-1.

CHAPTER 4

4.0 RESULTS AND DISCUSSION

This chapter outlines and discusses the results of dose measurements in three CT facilities using acrylic PMMA dosimetric body phantoms for institutional comparison of radiation doses. Presentation of the summarized data and the analysis are shown below.

4.1 RESULTS

4.1.1 Measurements of CTDI with the Ion Chamber Technique

Tables 4.1 – 4.3 show the charges recorded during the body phantom irradiation with the use of the Ion Chamber in the study and the subsequent calculated exposure, $CTDI_{100}$ values using mathematical expressions. Average Charge, Exposure, and $CTDI_{100}$, are represented in the Tables.

Table 4. 1 CTDI values for body Phantom at AEC using ion chamber.
Institution: SGMC

Position	charge Q(C)	X(C/Kg)	Exposure X(rad)	$CTDI_{100}$ (rad)
Central(C)	5.84E-11	2.88E-04	8.97E-01	1.12E+00
Periphery (P1)	1.02E-10	5.02E-04	1.57E+00	1.96E+00
Periphery (P2)	9.68E-11	4.77E-04	1.49E+00	1.86E+00
Periphery (P3)	1.02E-10	5.03E-04	1.57E+00	1.96E+00
Periphery (P4)	1.10E-10	5.41E-04	1.69E+00	2.11E+00

$CTDI_{100c}$ (rad)	$CTDI_{100p}$ (rad)	$CTDI_w$ (mGy)	Estimated $CTDI_{vol}$ (mGy)	CT Console $CTDI_{vol}$ (mGy)
1.12E+00	1.58E+00	1.69E+01	2.11E+01	1.7E+01

Table 4. 2: CTDI values for body Phantom at AEC using ion chamber.

Institution: GMIC

Position	charge		Exposure	
	Q(C)	X(C/Kg)	X(rad)	$CTDI_{100}$ (rad)
Central(C)	9.08E-11	4.47E-04	1.35E-00	1.69E+00
Periphery (P1)	1.48E-10	7.29E-04	2.20E+00	2.76E+00
Periphery (P2)	1.46E-10	7.17E-04	2.17E+00	2.71E+00
Periphery (P3)	1.41E-10	6.93E-04	2.09E+00	2.62 E+00
Periphery (P4)	1.46E-10	7.17E-04	2.17E+00	2.71E+00

$CTDI_{100c\ (rad)}$	$CTDI_{100p\ (rad)}$	$CTDI_w$ (mGy)	Estimated $CTDI_{vol}$ (mGy)	CT Console $CTDI_{vol}$ (mGy)
1.69E+00	2.16E+00	2.36E+01	2.95E+01	2.4E+01

Table 4. 3: CTDI values for body Phantom at AEC using ion chamber.

Institution: KBTH

Position	charge		Exposure	
	Q(C)	X(C/Kg)	X(rad)	$CTDI_{100}$ (rad)
Central(C)	1.12E-09	5.49E-03	1.66E+01	5.19E-01
Periphery (P1)	2.15E-09	1.06E-02	3.20E+01	1.00E+00
Periphery (P2)	2.08E-09	1.02E-02	3.09E+01	9.66E-01
Periphery (P3)	2.29E-09	1.13E-02	3.41E+01	1.07E+00
Periphery (P4)	2.36E-09	1.16E-02	3.52E+01	1.10E+00

$CTDI_{100c\ (rad)}$	$CTDI_{100p\ (rad)}$	$CTDI_w$ (mGy)	Estimated $CTDI_{vol}$ (mGy)	CT Console $CTDI_{vol}$ (mGy)
5.19E-01	8.26E-01	8.61E+00	1.72E+01	1.31E+01

4.1.2: Estimated CTDI $_{(w, vol)}$ and DLP for the Body Phantom

The results of CTDIvol, CTDIw and DLP values obtained from measurements made with the body phantoms using an ion chamber and the console displayed values from the CT scanners are presented in Table 4.4 below.

Table 4.4: Estimated CTDI $_{(w, vol)}$ and DLP for the Body Phantom

Institution	CTDI (mGy)				DLP (mGy.cm)	
	Estimated Value		Console Value		Estimated Value	Console Value
	$CTDI_w$	$CTDI_{vol}$	$CTDI_w$	$CTDI_{vol}$	DLP	DLP
SGMC	16.9	21.1	N/A	17	675	544
GMIC	23.6	29.5	N/A	24	944	760
KBTH	8.6	17.2	N/A	13.1	419	325

Note: N/A means no available data

4.1.3 Deviation of CTDIvol and DLP for Estimated and Console Displayed Values

Dose deviation between estimated and console displayed values for the CT centers are presented in Table 4.5 below

Table 4.5: Deviation of CTDIvol and DLP for Estimated and Console Displayed Values Compared.

Institution	$CTDI_{vol}$ (mGy)			DLP (mGy.cm)		
	Estimated Value	Console Value	Deviation (%)	Estimated Value	Console value	Deviation (%)
SGMC	21.1	17	24.1	675	544	24.1
GMIC	29.5	24	22.9	944	760	24.2
KBTH	17.2	13.1	31.3	419	325	29.0

4.1.4: Calculated Effective Doses from the three facilities Using DLPs and Adult k Value adopted for Chest.In this Study K was 0.014 mSv mGy-1 cm-1.

Table 4.6**:** Calculated Effective Doses from the three facilities Using DLPs and Adult k Value adopted for Chest.

Institution	DLP(mGy.cm)	Effective dose(mSv)
SGMC	675	9.45
GMIC	944	13.2
KBTH	419	5.87

Note that K is the anatomy-specific dose coefficient expressing effective dose normalized to DLP in a standard CT dosimetry phantom.

4.2 DISCUSSION

The choice of CTDIvol for the work was informed by its ability to:

- provide information about the amount of radiation used to perform the study
- provide a way to track across patients and protocols for QC purposes
- allow for comparison between scanners and facilities.

Analysis of the results for the CT body phantom examination using the estimated CTDIvol and console displayed CTDIvol values show a minimum CTDIvol deviation of 22.9% recorded at GMIC. The maximum deviation of 31.3% was recorded at KBTH.

In comparing the facilities using estimated and console displayed DLPs, the minimum DLP deviation was 24.1% recorded at SGMC while the maximum DLP of 29.0% was again recorded at KBTH.

Considering the deviation in CTDIvol among the facilities there exist variations of 4.9% between SGMC and GMIC but 23.0% between SGMC and KBTH and 26.8% between GMIC and KBTH.

This trend is replicated in the DLP deviations between facilities SGMC and GMIC with 0.004% deviation, but a 16% deviation when comparing SGMC and GMIC to KBTH. This is because the scanners employed at SGMC and GMIC are of the same make but differs from that at KBTH as shown in table 3.1. Moreover, the number of slices employed in CT examination contributes to the radiation dose delivered by a scanner. With no difference in the number of slices between the scanners at SGMC and GMIC (16 slices) but different from that of KBTH (640 slices), the reason is not farfetched since the 640 slices will deliver more dose than the 16 slices.

Using k the anatomy-specific dose coefficient expressing effective dose normalized to DLP in a standard CT dosimetry phantom of 0.014 mSv mGy-1 cm-1, effective doses were calculated and presented in table 4.6 above.

Referring to the said table, the minimum effective dose was 5.87mSv and maximum 13.2mSv calculated using DLPs from KBTH and GMIC respectively. At SGMC it was found to be 9.45mSv.

The estimated CTDI values for the CT body phantom from this work can be compared with research study by Inkoom et al. in 2014. In their study on Adult Medical X-Ray Dose Assessments for Computed Tomography Procedures in Ghana, they researched into adult computed tomography (CT) examinations and dose assessments for head, chest, abdomen, lumbar spine and pelvis at six CT facilities in Ghana (with approval of the management of the participating hospitals). The dosimetric parameters estimated were volume computerized tomography dose index (CTDIvol), dose length product (DLP), and effective dose (E) for the stochastic radiation risk of a non-uniform exposure in terms of whole body exposure.

Further comparing the results of this study to other published works, Descamps et al. in 2012, estimated percentage deviations between measured and console displayed doses for new generation CT scanners. The findings from their work showed that measured doses (CTDIvol) for CT examinations could be as much as 32 – 35% higher or lower than console displayed doses.

Also, Anim-Sampong et al. (2016) compared measured dosimetric parameters of the

latest 640-Slice Aquilion ONE CT scanner with established DRLs for adult and pediatric head, chest and abdominal examinations as a quality assurance test in order to recommend appropriate radiological safety solutions if differences existed. Results from their study showed that volume weighted CT dose index (CTDIvol) and dose length products (DLPs) were generally lower than the ICRP and other internationally recommended DRLs.

Comparison of the findings (CTDIvol, DLP and E) from this study with some international diagnostic reference levels can also be seen in Table 4.7 -4.9

Table 4.7: Comparison of CTDIvol values (mGy) from this study with some international diagnostic reference levels

Examination	This study	CTDIvol Europe2004	CTDIvol UK 2003	CTDIvol Sweden 2002	CTDIvol Netherlands 2008
chest	13.1-24	10	13	20	10

Table 4.8: Comparison of DLP values (mGy. cm) with International Adult Diagnostic Reference Levels.

Examination	This study	DLP Europe2004	DLP UK 2003	DLP Sweden 2002	DLP Netherlands 2008	DLP EC 1999
chest	419-675	430	580	600	350	650

Table 4.9: Comparison of Effective Dose E(mSv) with published values

Examination	This study	Clarke et al 2000	Tsapaki et al 2001	Hart & Wall 2002	Olerud M 2003	UNSCEAR 2008
chest	5.9-13.2	3.8-9.2	10.9	8.0	11.5	9.7

CHAPTER 5

CONCLUSION AND RECOMMENDATION

5.0 Overview

This chapter presents the conclusion of the study and the recommendations made with regards to the findings.

5.1 CONCLUSION

This study has successfully estimated CTDIvol and DLPs as well as effective doses .

For all three CT facilities estimated CTDIvol and console displayed CTDIvol values show a discrepancy as result of the different protocols used and AEC design functionality . The range of the CTDIvol were found to be within the values obtained by other researchers cited in the literature.

Even though SGMC and GMIC operated a Siemens CT machine there was a considerable deviation in CTDIvol ,DLP and E but the deviation was on the increase when compared to KBTH which operated a Toshiba Aquilion One machine.

The values obtained for CTDIvol, DLP and effectives dose were within acceptable diagnostic reference levels. Further optimization is possible when accesses to the techniques factors used are made available from the facilities where the research were conducted.

5.2 RECOMMENDATION

The following recommendations are made to individual parties involved in the use, optimizing of radiation dose and protection of patients.

5.2.1 To the research community

Further works should be done to capture most of, or if possible all, CT facilities in Ghana. Other dosimeters should be employed in further studies to validate the findings from this work .In conducting further research, data should be collected with AEC activated and AEC deactivated and compared.

5.2.2 To the participating health institutions

Considering that AEC reduces dose to patients by factoring size without compromising image quality, i.e., AEC systems is a preselected image quality index, also referred to as a reference or target image quality index. The index is stored in the CT scanner by the manufacturer before shipping. When a radiologic technologist acquires a scan, the unit adjusts milliam-perage so that the exposure approximates that used to create the reference image, the use of AEC should be encouraged at all centers. However, operators of the CT machines as well as the medical physicists should be given periodic on job training regarding how to vary the technique factors at AEC according to the manufacturers' recommendations.

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