DETERMINATION OF COMPUTED TOMOGRAPHY DIAGNOSTIC REFERENCE LEVELS IN NORTH-CENTRAL NIGERIA

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MUHAMMAD KABIR ABDULKADIR

(10435399)

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DECLARATION

This thesis is the result of the research work undertaken by Muhammad Kabir Abdulkadir in the department of Medical Physics, School of Nuclear and Allied studies, University of Ghana, under the supervision of Professor Cyril Schandorf, Professor J.J Fletcher and Mr. Francis Hasford.

……..	...……
MUHAMMAD K. ABDULKADIR	DATE
(STUDENT)	
……..	……
PROFESSOR C. SCHANDORF	DATE
(PRINCIPAL SUPERVISOR)	
……..	……..
PROFESSOR J.J FLETCHER	DATE
(CO-SUPERVISOR)	
…….	…… …….
MR. FRANCIS HASFORD	DATE
(CO-SUPERVISOR)	

DEDICATION

This work is dedicated to God Almighty the most gracious and most merciful for making this programme possible for me; I will always be grateful.

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List of abbreviations

ACR	American College of Radiology
AAPM	American Association of Physicist in Medicine
CT	Computed Tomography
CTDI	Computed Tomography Dose index
CTDIvol	Volume weighted Computed Tomography Index
DLP	Dose Length Product
DRLs	Diagnostic Reference Levels
EC	European Commission
FWHM	full width half maximum
LDRL	Local Diagnostic Reference Levels
RDRL	Regional Diagnostic Reference Levels
IAEA	International Atomic Energy Agency
IEC	International Electro technical Commission
ICRP	International Commission on Radiological Protection
IPEM	Institute of Physicist and Engineers in Medicine
ImPACT	Imaging Performance and Assessment of Computed Tomography

NCRP	National Commission on Radiological protection
NNRA	Nigerian Nuclear Regulatory Authority
NRPB	National Radiological Protection Board
mGy	Milli Gray
mGy.cm	Milli Gray per Centimeter
MSCT	Multislice Computed Tomography
MRI	Magnetic Resonance Imaging
SSCT	Single Slice Computed Tomography
PMMA	Poly methyl methacrylate

ABSTRACT

The aim of this study is to estimate computed tomography (CT) dose levels for common CT examinations in North-Central Nigeria. Dose parameters and scan parameters for the most commonly performed CT examinations (head, chest and abdominal CT scans) were surveyed during a four month period in 4 CT centres with Multislice scanning capabilities (4- 64 slices). Data on CT volume index ($CTDI_{vol}$) and dose length product (DLP) displayed on scanner console was recorded for a minimum of 10 averaged-sized (70±10kg) patients for each facility to estimate the DRLs. The rounded 75th percentile of the distribution was then used to calculate a DRL for each centre and the region by compiling all results from centres surveyed. Data for 226 patients was collected. CT dosimetry software ImPACT CT patient dosimetry calculator, version 1.0.4 with National Radiation Protection Board SR250 data set was used to validate and compare surveyed scanner generated dose values. Estimated regional DRLs for head, chest and abdominal scans are (60mG and 1024 mGy.cm), (10mGy and 407mGy.cm) and (15mGy and 757mGy.cm) for $CTDI_{vol}$ and DLP respectively. Mean effective dose values are 1.7mSv, 5mSv and 11.9mSv for head, chest and abdominal scan respectively. A wide variation of mean doses was observed across the centres, however, DRLs estimates were lower than EC (1999) values but above UK (2003) DRLs except for chest examination, this indicates a need for optimization. Validation result show unity (<10% overall variation) between scanner generated and software calculated dose values.

CHAPTER ONE

1.0 Introduction

Preamble

This chapter presents a background to the study, statement of the problem, significance, objective, scope and limitations of this study.

1.1 Background

Optimizing the protection of patients, and maintaining appropriate good practice is a priority for all diagnostic radiological examinations including computed tomography (CT) examinations. This is because they involve the use of ionizing radiation which is known to have harmful effect on human body, unless all recommended safety and radiation protection principles/measures are strictly adhered to. In CT imaging, an optimized protocol is one that produces the required image information with the lowest possible radiation dose to the patient.

Diagnostic reference levels (DRLs) are optimization tools used as special type of dose constraints above which doses must be reviewed (Friberg, Widmark and Hauge, 2004) and considered above acceptable levels, especially if acceptable image quality can be achieved at lower doses. This will ensure that dose to each patient is kept as low as reasonably achievable for the clinical purpose of the radiologic examination i.e. image of good diagnostic quality.

X-ray computed tomography (CT) was introduced into clinical use in 1973, which over time has successfully become the primary diagnostic modality. Mettler et al, (2009) have

reported increased utilization of computed tomography examination for clinical diagnosis worldwide. Although CT imparts high radiation dose to patients, its benefits can far outweigh the risk if all equipment, personnel and the technical knowhow guiding the proper use of the equipment are well adopted.

Fast scanning speed, isotropic spatial resolution, non-invasive, affordability compared to other modalities such as magnetic resonance imaging, applications in staging, treatment planning and follow up of cancer treatment are some of its unique advantages (Lifang Yu, et al, 2009). However, the increase in the patronage of CT globally as well as in Nigeria have led to concerns about radiation hazard from its use; hence need for proper radiation audit and optimization of CT practice (Ogbole and Obed, 2014). Thus, the first step in optimization should be survey of doses.

In recent time, literatures have described the risk associated with CT, one of which described CT as the highest contributor, contributing almost one half the total radiation exposures from medical use (Mettler et al, 2009). A single routine chest CT has been identified to give radiation equivalent dose of 400 planar radiography of the chest (Rehani and Berry 2000), so therefore radiation doses produced by CT are (considered as high doses) only comparable to that of interventional radiology (EUROTOM, 1997). New advancements in CT such as multi-slice which gives higher doses to the patient have also been reported to have led to further increase in the collective dose of CT examinations (Hunold et al. 2003 and Abdullahi, 2009). As much as 1.5 – 2% of cancer may eventually be caused by the radiation currently used in CT (Banner and Hall, 2007). Dose to tissues in the imaging field although not in the target organ of interest for such procedure are also of great concern, because most of this organs tend to be highly

radiosensitive organs for e.g. lens of the eye in brain scan, the breast in chest scan, uterus, ovaries and testis in abdominal and pelvic scans respectively.

This study adopts the European Commission (EC), American College of Radiology (ACR, resolution-47, 2013) in collaboration with American Association of Physicist in Medicine, International Atomic Energy Agency (BSS) and the International Commission on Radiological Protection; recommended guide lines/procedures to determine DRLs in radio diagnostics. This guideline states that; weighted CT dose index (CTDIw) now replaced by volume weighted CT dose index ($CTDI_{vol}$) and dose length product (DLP), are the appropriate dose quantities for the establishment of DRLs for optimizing patient exposure in CT. Literatures have also suggested that every country including Nigeria should have its own DRLs values. This is because of the differences in practice and technological advancement such as iterative reconstruction, from country to country and the fact that Nigeria is yet to have a guideline for DRL (Ogbole & Obed, 2014).

The International Electro technical Commission (IEC, 2001), specify that the dose descriptors ($CTDI_{vol}$) measured in mGy and (DLP) measured in mGy-cm should be available for display on control consoles of most modern CT scanners. $CTDI_{vol}$ is a measure of the average dose within the scan volume to a standardized phantom and the total amount of radiation delivered to the standard phantom is represented by the DLP. The DRLs for each examination or patient group are set on the basis of distributions of the typical (mean) doses observed in a national or regional survey, commonly by adopting the third quartile values, which are then compared with internationally recommended DRLs to provide investigation levels for unusual practices or abnormally

high doses i.e. doses in top 25%, (Institute of Physics and Engineering in Medicine (IPEM, 2004).

This study determined CT doses received by patients undergoing common CT procedures and contribute data as reference required in order to establish a guideline of optimized CT protocols for clinical usage in Nigeria, since guidelines were basically based on the compilation of publications from many researches and observations associated with radiation hazards (Arif, 2009). The World Health Organization (WHO, 2008) also stated that the global burden of radiation related disease must be based on scientific assessment of health risks related to radiation exposure. This study will also assess how well doses are optimized in comparison to established standards, through a regional survey.

1.2 Statement of the Problem

The need for improvement in the optimization of patient protection through implementation of measures to keep all doses imparted to patients undergoing computed tomography within acceptable ranges for the clinical purpose of each examination has been a topic of global recognition (IPEM, 2004). This was as a result of the higher radiation dose being used to produce images that could otherwise be produced at lesser patient dose without losing any diagnostic information or image quality. However, radiation doses used in CT procedures are associated with or pose a potential risk of radiation induced malignancy (cancer). Therefore reducing radiation dose in CT is of utmost importance particularly in the light of continued increase in the number of CT examinations performed annually (NCRP, 60). Computed Tomography Diagnostic Reference Levels (DRLs), which is the recommended tool in achieving optimization of

doses, is yet to be set or unavailable for computed Tomography practice in Nigeria (Ogbole and Obed 2014). Practices are presently referenced to U.K radiological practice standards. More so, IPEM (2004) recommends that every country should have or set its DRLs, because practices and advancement in technology varies from one country to another and hence one country's DRL cannot be a good representation of another.

The study sought to answer the following research questions: (i) What is the estimated mean and third quartile values of CTDIvol and DLP received by patients undergoing common CT examinations in North-Central Nigeria?; (ii) is there a significant difference among CTDI and DLP values received by patients in other countries and in Nigeria? (iii) Does CTDIvol and DLP values for other countries show better optimization of practice than Nigerian Practice or otherwise

1.3 Objective of the Study

The aim of this study is to provide estimate of CT machine generated CTDIvol and DLP (absorbed dose) delivered to patients undergoing head, chest and abdominal CT examinations in North-central Nigeria for the establishment of a Regional dose reference levels (RDRL).

Other objectives include performing inter-comparison studies of the data collected with recommended DRLs from other countries and also to analyze the trends of these values and undertake optimization studies in order to ascertain whether better optimization is being practiced in Nigeria.

1.4 Significance of Study

The outcome of this study will provide a broad check on the optimization process since dose survey and the determination of DRLs are recognized tools in optimization of all practices involving medical use of ionizing radiation, especially focusing on the use of CT for diagnostic radiology in Nigeria. Identification and correction of potentially unusual practices in the population can be achieved as well as guidance for adherence to dose levels for CT examinations will be provided particularly in line with the growing number of CT centres and CT examinations in the country. Hence radiation protection of patients undergoing CT examination in Nigeria can be improved from the knowledge gained from the research findings.

Findings of this study will also enable appropriate recommendations to be made to hospital authorities and the regulatory authority for the incorporation of DRLs into clinical practice and regulatory control programmes. This study is also expected to trigger the development of the national diagnostic reference levels for Nigeria.

1.5 Scope and limitations of the study

This study was conducted on four selected CT centers with computed tomography that have multislice scanning capability in North-Central region of Nigeria, from September 2014 to February 2015. Only adult patients that weigh 70±10 kg who report for routine CT scans of the head, chest and abdomen were included in this study.

This study is however limited to Multislice CT scanners only, because single slice scanners have become obsolete in Nigerian hospitals. The major limitation of this study is that the researcher could not have access to the required experimental apparatus namely;

ionization chamber and electrometer/dose meter which would have been used to measure computed tomography machine dose output. This would have enabled the researcher compare experimentally collected values with the computed tomography machine generated values in order to ascertain the reliability of the machine generated dose values. However, the work continued because of the IEC (2001) recommendation that guaranteed the use of CT machine generated dose values for the establishment of diagnostic reference levels.

The data collection time frame of four months was not enough to allow the inclusion of larger number of participants, hence limiting the total number of participants that was surveyed, though the number surveyed was enough to set a DRL as specified by the EC, (1999) guideline i.e. at least 10 patients in each examination group was achieved.

Frequent break down of machine was experienced in one of the centres due to uninterruptible power supply (UPS) malfunction.

1.6 **Thesis outline**

Chapter one deals with historical review of CT development and impact in clinical practice, radiation risk and radiation protection issues including the aim, objectives and significance of diagnostic reference levels in optimization. Chapter two provides literature review of relevance to the research topic. Chapter three discuses mainly the materials and methods used in the collection of CT patient and CT dose data and the procedure for setting Diagnostic reference levels (DRLs) as adopted from the EC 1999 guidelines for setting DRLs. This chapter also presented an over view of the statistical

data analysis method adopted for this study. Chapter four presents the results of the dose estimates done according to the above mentioned method in chapter three. The results are then discussed with reasonable scientific explanations and justification in line with the various approaches and methods that have been reviewed in chapter two. And finally, chapter five conclude with a detailed description of the estimated doses as a description of the current practice in this region and hence, it represents the diagnostic reference levels for this region at least before the establishment of Nigerian national DRLs. This is followed by recommendations especially to the relevant authorities.

CHAPTER TWO

Literature review

Preamble

This chapter presents historical preview and trends in CT practice both globally and in Nigeria. It also highlight opinions from published literatures from research works, journal articles, paper presentations, reports and reviews from various stakeholders that have made significant contribution to radiation protection and optimization in CT practice especially in the area of setting diagnostic reference levels. It provides an insight on the need and justification of this study and explains the approach in terms of methodology appropriate for this type of study. Dose quantities and methods of dose quantification in CT are highlighted. Factors that influence dose are also discussed as well as current and future perspectives with regards to dose reduction in CT.

2.1 Computed Tomography Evolution and Radiation

Computed tomography despite being a high dose diagnostic imaging technique that uses X-ray to generate body image, its utilization and application still remain on the rise. This due to its fast scanning or imaging acquisition time and its wide clinical applications even though the introduction of Magnetic Resonance Imaging (MRI) and ultrasound (US) imaging were introduced to complement its use (Garba, 2014). As at the year 1999, (NRPB, 1999) estimates showed that CT covers 4% of all diagnostic examinations in the UK but contributes 40% of the total patient dose. Four years later, CT examinations have grown to 9% and represented 47% of total patient dose. Hence, radiation protection in CT is needed. Protection in CT incorporate care in referrals (justification) to ensure that other

non-ionizing diagnostic modalities (such as MRI and US) are considered first before a patient is referred for CT procedures. The other aspect is that if CT procedure has been justified then care is needed in equipment choice and use (optimization). Lastly reference dose levels must not be exceeded. Table 2.1 shows dose contribution from different radiological examinations.

Table 2.1: Radiation dose contribution from different radiological examinations (Seabourn, 2010)

Examinations	Percentage of diagnostic imaging studies	Percentage of Radiation exposure
Radiography	74%	11%
Nuclear Medicine	5%	26%
Interventional	4%	14%
CT	17%	49%

The first clinical scanners which were dedicated to head imaging only were installed between 1974 and 1976. However, whole body systems became available in 1976 and in the 1980s it became widely available. The use of CT in medical practice in Nigeria dates back to more than 3 decades and ever since it has been experiencing increase in application and utilization (Ogbole and Obed, 2014). Third generation scanner (single-slice) were initially installed, but in keeping with advancement of CT technology, most centres in the country now operate with the sixth generation (multislice) CT scanners, mostly the 16-slice scanners (Garba, 2014). This implies that an increase by 10-30% patient dose may have occurred with the use of multislice CT (ICRP, 87). The Nigeria Nuclear Regulatory Authority (NNRA) report 2009 shows inadequate number of CT machines (30) in the country considering a population of 120 million (Erondu, Okoro,

Ugwu, 2011). For the purpose of this study only multi-slice CT scanners were surveyed, since multi-slice scanner form the majority of the scanners in the country presently.

2.2 Radiation risk associated with computed tomography

A lot of recent articles have described the risk associated with CT, one study describe CT as the highest contributor, contributing almost one half of the total radiation dose from medical use (Mettler et al, 2009). A single routine chest CT has been identified to give radiation equivalent dose of 400 planar radiography of the chest (Rehani and Berry 2000), so therefore radiation doses produced by CT are (considered as high doses) only comparable to that of interventional radiology and radiotherapy (EUROTOM, 1997). New advancements in CT such as multi-slice which gives higher doses to the patient have also been reported to have led to an increase in the collective dose of CT examinations (Hunold et al. 2003 and Abdullahi, 2009). As much as 1.5 – 2% of cancer may eventually be caused by the radiation currently used in CT (Banner and Hall, 2007). Dose to tissues in the field although they are not the target organs of interest for such procedures are also of great concern, because most of this organs tend to be highly radiosensitive organs for e.g. lens of the eye in brain scan, the breast in chest scan, uterus, ovaries and testis in abdominal and pelvic scans respectively. The effect of radiation is classified into stochastic and deterministic effects; radiation effect from doses used in CT can lead to stochastic effect with the probability of life time induction of cancer especially in children and female patients (Yu et al., 2009). Efforts aimed at curbing the incidence or the probability of radiation induced cancer, a life time effect. Radiation exposure in CT has also been of great interest to the international community.

2.3 Radiation protection in CT and role of Diagnostic Reference Levels (DRLs)

The rational for setting national diagnostic reference level (NDRL) as stated in an International Atomic Energy Agency (IAEA) document termed 'Radiation Protection in Patients' emphasized the need for optimization, i.e. to keep all CT doses as low as reasonably achievable within clinical ranges since surveys of CT dose estimates have shown significant variations in practice for the same patient categories and size that have undergone identical types of examinations, hence, examination-specific DRLs were suggested (IAEA,1990). Norway was one of the pioneer countries in establishing national DRLs which were first performed in 1987 and published for six conventional radiological examinations in 1996. Since then other countries followed suit for various radiological procedures (Friberg, Widmark & Hauge, 2008).

DRLs are also intended to improve patient protection by allowing comparison of current practice, comparison of similar examination for similar purpose and requiring similar technique including clinical indications rather than broad categories of examinations (IPEM, 2004).

In CT practice, significant variation in doses were observed in several different surveys and the differences was associated with differences in scanning protocols and scanner related parameters and therefore standardization of protocols and optimization of scan parameters was suggested (Roshan and Paul, 2010). In addition, studies both in Nigeria and abroad have reported intra- and inter radiological centre dose discrepancies for the same diagnostic procedure. A ratio of almost 50 between the hospital with the highest dose and that with the lowest dose for an average size adult in 20 different hospitals

nationwide was reported by the National Radiological Protection Board in the United Kingdom (NRPB, 1990). Several studies on survey of entrance surface dose in different hospitals in Nigeria also show significant intra- and inter radiological centre variation in doses for the same diagnostic procedure. These studies include; Sherifat, and Olarinoye, (2009); Ogundare et al.,(2008); Ogundare et al., (2004); Ogunseyinde et al.,(2002); Ajayi and Akinwumiju, (2000). Similarly, the Food and Drug Administration of the U.S.A (Gray, 1999), conducted a similar national survey and found out that, the ratio of maximum to minimum exposure range from 8.8 to 126.7. (Shrimptom et al, 1986 and Faulkner & Corbelt, 1998) [Table 2.2] also confirmed variations in dose levels for the same X-ray examination up to a factor of 100. This called for standardization and optimization of practice globally and hence the recommendation of dose guidance level by the International Commission on Radiological Protection (ICRP, 1990) and International Atomic Energy Agency (IAEA, 1990). This is to identify abnormal practices and provide a reference to which practices can be compared in order to maximize optimization.

The importance of setting DRLs cannot be over emphasized; however, it is relevant to know that DRLs are not universal but specific to a country because of equipment and personnel training, DRLs established for one country (with different CT practice and technology) may not be wholly relevant to another country's circumstances (RPOP, 2014, Ogbole & Obed, 2014 and Olowokere et al, 2012). Iterative reconstruction which is an advancement in CT technology must also be considered when setting DRL or comparing one practice to another. Establishing DRL alone does not guarantee long term optimization of doses. Doses must be reviewed from time to time since diagnostic

reference levels and achievable doses are dynamic values changing overtime and with changes in technology (NCRP, report 172).

Ogbole & Obed, (2014) and Olowokere et al, (2012) have reported that Nigeria is yet to have a NDRL for CT. They both advocated for studies to survey CT doses across the country in order to standardize and optimize practice in line with the increasing utilization of CT in the country. This will ensure that doses to patients are kept within acceptable levels by curtailing bad practices and identifying centres where optimization process may be required. IPEM, (2004), states that each CT centre should determine its typical levels of dose (CTDIvol and DLP) for each type of examination as the mean values observed for representative samples for each patient group (adult and children of different sizes). Meanwhile only University College Hospital in the southwestern Nigeria have establish and published its CT local (site) DRL and estimated doses for head, cervical, chest, abdomen/pelvic and lumbar CT examinations Ogbole & Obed, (2014). Similarly (Garba, 2013), also estimated CTDI for brain CT scans only. Although reported doses in both studies were higher than EC recommended doses for most type of examination. The paucity of data for the establishment of CT DRL is not limited to Nigeria alone, but also a problem in most developing countries and sub-Sahara Africa. In spite of the above fact, DLP were found to be above and CTDI below proposed DRL (EC guidelines) in a survey of Tanzanian CT doses (Nagile et al., 2006), high DLP values were influenced by large scan length used in Tanzanian hospitals. Inkoom, et al., (2014) reviewed CT doses in Ghana, mean CTDIvol and DLP of head, abdomen and lumbar spine were below the European commission DRLs, while mean DLP of chest and pelvis exceeded the reference levels by 2 and 6% respectively. Suliman et al, (2006) (Table 2.2)

also surveyed and proposed Sudanese DRLs and75th Percentile for most of the procedure were below European DRLs.

Table 2.2: Sudanese CT diagnostic reference levels for head, chest and abdominal examinations (Suliman et al, 2006)

Examination	**Head**	**Chest**	**Abdomen**
Diagnostic reference levels			
CTDIvol(mGy)	65	11.5	11.6
DL(mGy.cm)	758	327	437

Already there are sources of established/reviewed national and international DRLs upon which subsequent surveys can be compared in order to ensure that reference dose levels are strictly adhered to and doses above threshold can be reviewed, these include; European Commission (EC, 1999), United Kingdom (IPEM, 2004), United States of America American College of Radiology)(ACR, 2008). These organizations have worked diligently and have successfully established national diagnostic reference levels in their respective countries through nationwide survey and review from time to time of doses in diagnostic and interventional radiology as well as nuclear medicine (Tables 2.3 and 2.4). They also laid down recommendations and guidelines to be adhered to by individuals, states or centres aspiring to set their guidance levels in order to standardize the process of optimization of practices involving ionizing radiation.

Table 2.3: Diagnostic reference levels in terms of CTDI and DLP surveyed from other countries.

Examination DRL	EC 1999	UK study	Austrailian
Head CT			
CTDIw (mGy)	**60**	**66**	**47**
DLP (mGy-cm)	**1051**	**787**	**525**
Chest CT			
CTDIw(mGy)	**30**	**17**	**9.5**
DLP(mGy-cm)	**650**	**488**	**447**
Abdominal CT			
CTDIw(mGy)	**35**	**19**	**10**
DLP(mGy-cm)	**780**	**472**	**696**

Table 2.4: Diagnostic reference levels in terms of DLP (mGy-cm) from New Zealand compared with recent national studies (Stirling and Cotterill, 2009)

Procedure	**New Zealand 1992-single slice CT**	**European union 1999-single slice CT**	**UK 2003-single slice CT**	**UK 2003-multi slice CT**	**British Columbia 2004-multislice CT**	**New Zealand 2007 multislice CT**
Head	1050	1050	760	930	1300	1300
Sinuses	-	360	-	-	-	290
Routine chest	700	650	430	580	600	690
Chest,abdomen and pelvis	-	-	760	940	-	1400
Abdomen and pelvis	1470	780	510	560	1100	630

However, failure of clinics/centres to implement established guidelines for optimization purpose was reported as a setback in achieving standardized practice (Friberg et al., 2008). National development of local DRLs nationwide in collaboration with the

appropriate regulatory Authority (such as Nigerian Nuclear Regulatory authority (NNRA)) was suggested as a tool to improve response and optimization of practice in a country/state (Olowokere et al. 2012).

Dose area product (DAP), volume computed tomography dose index ($CTDI_{vol}$) & dose length product (DLP) and average glandular dose (AGD) were used for conventional X-ray examinations, Computed Tomography and Mammography respectively in establishing reference levels (EC,1999).

It is reported in literatures that Diagnostic reference level (DRL) should be reported in either regional or national levels (ICRP, 60) in other to represent a particular practice over a wide area. India estimated doses imparted to patients and formulated DRLs through two separate (regional diagnostic reference level) surveys, (Roshan & Paul, 2011 and Saravanakumar et al., 2014). Doses in the studies were found to be below the European Commission doses. They also reported variation in doses from one centre to the other and therefore recommended standardization and optimization of doses. Therefore standardization of scanning protocols and scanning related parameters was advised.

The International Electro technical Commission (IEC 2001), specify that; dose descriptors ($CTDI_{vol}$) measured in mGy and (DLP) measured in mGy-cm should be available for display on control consoles of most modern CT scanners and hence can be used in establishing DRL. Comparison between calculated and displayed dose values has shown reasonable agreement between the two sets of data with the mean ratios for each of the quantity being close to unity: 0.98 for CTDIvol and 0.90 for DLP per sequence and 0.95 for DLP per examination (Schrimpton et al., 2003). Foley et al., (2012), proposed

Irish CT diagnostic reference levels (DRLs) for the first time by surveying radiation doses using CT generated dose value displayed on the console for nine most commonly performed CT examinations. All 34 CT equipment surveyed were multi-slice scanners (2-128slices). Dose descriptors $CTDI_{vol}$ (mGy) and DLP (mGy cm) on a minimum of 10 average-sized patients in each category were recorded to calculate mean local $CTDI_{vol}$ and DLP value (LDRL). After compiling all results, the rounded 75th percentile was used to calculate both LDRL and NDRL. This study shows that CT doses surveyed/currently in practice are 42% lower than previously recommended values in (EU, 1999; 2004). The reduction in dose was associated with the recent advancement in CT technology, since single-slice CT are no more in use and are not included in this particular study compared to the reference survey to which it was compared. The use of weighted CT dose index ($CTDI_w$) used in previous studies has been superseded by volume weighted CT dose index ($CTDI_{vol}$).

Friberg et al. (2008) revealed minor (non-significant) effect in accommodating non-standard sized (weight) patients on the setting of DRLs just as it is argued for the acceptance of all breast thicknesses in mammography DRLs. The study described the definition of a standard-sized patient (70±5kg) by the European Commission (EC) as narrow and difficult to deal with in a typical Norwegian hospital, where 20% of the population of both men and women are overweight as reported from statistics in Norway data. Hence, broader weight group (55-90kg) was accommodated to represent a standard-size patient for this population (Norway) by the national guidance on DRL. This shows that the definition of standard sized patient stated above cannot be universally applicable,

since body mass index and weight in all categories (age group and sex) vary significantly from one country to another.

For the purpose of this study the EC description is adhered to because available data shows that average body mass index (BMI) for Nigeria is not significantly different from British BMI (Chartsbin 2011; Ogbole and Obed, 2014).

Justification of practice is another important aspect of radiation protection in CT, since CT is a high dose procedure, a series of clinical factors must be considered, these include collecting adequate patient clinical information including records of previous investigations and in certain cases prior investigation of the patient by alternative imaging technique might be required first before considering or requesting for CT examination. Optimization should also cover; performing CT examination by only trained and qualified personnel (radiologist, radiographer), adherence to standard examination protocols with effective supervision and adoption of quality criteria as a check on the routine performance of the entire imaging process. These quality/image criteria test include the following: uniformity and linearity, CT number accuracy, resolution, contrast, z-axis sensitivity, alignment, imaging performance (noise), irradiated slice thickness, couch travel accuracy and gantry tilt.

2.4 Image criteria or diagnostic requirements in computed tomography

Image criteria as described by Shrimpton et al. (2003) refer to characteristic features of imaged anatomical structures that are defined in the region of examination with specific degree of visibility. They also defined the Degree of visibility as: Visualization - organs

and structures detectable in the volume of investigation; Critical reproduction – the structures to the specific indication are discriminated to a level essential for diagnosis.

Image criteria in CT are basically categorized into two namely: anatomical and physical image criteria. Anatomical image criteria may be defined in terms of visualization and critical reproduction of anatomic features. Evaluation of image quality based on image criteria takes into account both the anatomy of the area under examination and the contrast between different tissues which is essential for detection of pathological changes (shrimptom et al 2003). While the physical image criteria are measurable by objective means such as noise, low contrast resolution, spatial resolution, linearity, uniformity and stability of CT numbers, slice thickness and dose sentence not complete? Hence, for optimization of practice and doses to be achieved, the inter-play between achieving the above mentioned image criteria for any CT examination and dose must be carefully understood by CT operators, since optimization in CT requires the use of the smallest dose of radiation to produce the best quality diagnostic image (EC, 1999).

2.5 Referral criteria and the most commonly performed CT examination in Nigeria

Most countries have established DRL based on selection of the most frequently performed examination and patient category but some studies have also determined doses from special CT examination/procedures such as CT pulmonary angiography and high resolution CT (Foley et al. 2012). The most commonly performed examinations can be determined by prior survey of the most commonly performed procedure/examination amongst the participating centres (Foley et al. 2012; Erondu et al., 2011), in their work ‘′patterns of CT referrals among physicians in the south-south of Nigeria’’ found that

brain CT is the most frequently requested with intracranial hematoma as the most common finding representing 63.78% of the data collected, 8.88% thoracic mass for chest CT, 13.56% abdomen, 5.56% spine, 3.78% neck, 1.76% sinuses, 1.55% pelvis and 1.11% orbit CT examinations. They also found out that majority of all scans taken within this period (2 years) were referred by Neurosurgeons accounting for 43.1% of the total data, followed by 33.33% from general practitioners, Paediatricians (2.68%), Psychiatrics (4.44%), ENT (4.00%) and the remaining percentage accounts for referrals from other health care specialist. The authors of this work concluded that CT referral in this country is low as compared to other developed countries such as UK, Europe and Australia. They attributed lack of awareness of diagnostic capabilities and clinical benefits of CT among physicians, high cost of CT examinations, and poor access to health insurance in Nigeria. Pattern and frequency of CT scanning practice amongst children aged 4-14years in Nigeria also shows that brain scan was most commonly performed examination accounting for 88.8% of the total with convulsion being the most common indication for brain scans. Next in frequency are abdominal CT (4.9%) and the remaining percentage (6.3%) represents frequency of other examinations (Anas and Muhammad 2013). Head CT, Abdominal CT and Chest CT were selected after a wide survey as the most commonly performed CT examinations in Australian and Indian studies (ARPANSA 2008; Roshan & Paul 2011).

2.6 Computed Tomography Dosimetry

2.6.1 Dose distribution in Computed Tomography

The spatial dose distribution of CT fundamentally differs from conventional (projection) X-ray. Applied dose in CT decrease exponentially in accordance with Beer Lamberts law,

Since tissue attenuate the corresponding X-ray beam, the dose decreases along the axis from the tube adjacent side towards the opposite side (y-axis). When CT is compared to conventional PA of the same skull, the dose distribution here is almost homogenous inside the skull because the object (skull) is X-rayed from all sides.

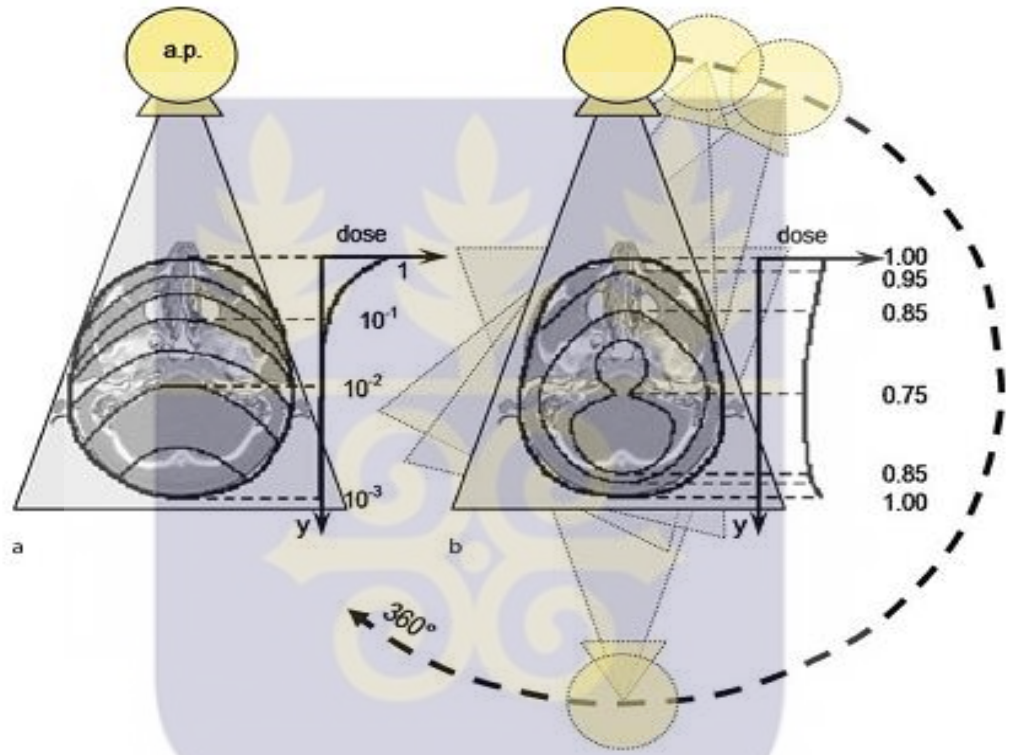

Figure 2.1: Spatial dose distribution in projection radiography (a) and 360º CT (b)

Image source: Buzug (2008)

Also a difference in simultaneously irradiated volume can be observed if the CT acquisition is limited by means of collimation to one single slice; the patient is primarily just irradiated in this respective layer with a thickness of only few millimeters. It is observed that dose profile for one single slice given a nominal slice thickness say 10mm is not restricted to the collimated area but dose is applied to the patient even outside the

slice demarcated by collimation, this is as a result of scattered radiation being produced in the respective layer. This makes it apparent that measured dose profile does not conform to an ideal rectangular function. Typically, the nominal thickness of a layer is within the range of the full width at half maximum (FWHM) of the dose profile.

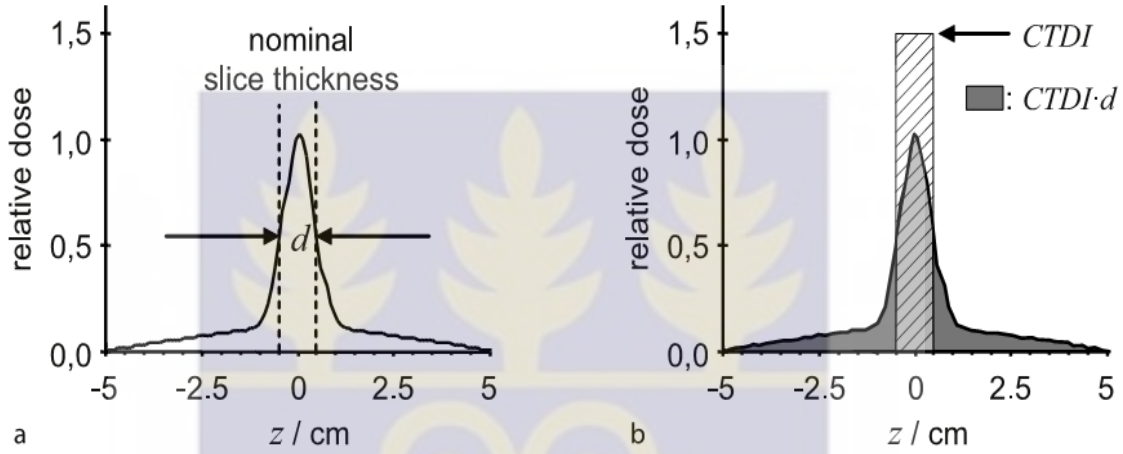

Figure 2.2: (a) Single slice dose profile for a nominal slice thickness d and (b) How Computed tomography dose index (CTDI) is obtained through the area of dose profile for a certain slice thickness.

Image source: Buzug (2008)

2.6.2 Dose efficiency in Multislice Computed Tomography (MSCT)

Dose efficiency refers to the fractions of X-rays that reach the detector and that is actually captured and contributes to image formation. Dose efficiency consists of two components namely; geometric efficiency which refers to the fractions of X-rays that exit the patient and eventually enters the active detector areas (Goldman, 2008).

Absorption efficiency refers to the fraction of X-rays that enters active detector areas and that are actually absorbed (captured) to the stray rays that could not be captured. Absorption efficiencies are similar for all SSCT and MSCT scanner that have solid state

detectors. But two aspects of MSCT reduce its geometric dose efficiency relative to that of SSCT.

The first is the dividers between individual detector elements along the x-axis, which create dead space that, do not exist in SSCT. Depending on the detector design and element size, dead space associated with the dividers can represent up to 20% of the detector surface area. This implies that up to 20 % x-rays exiting the patient will strike dead space and not contribute to image formation.

X-ray beam width is the second factor that reduces MSCT geometric efficiency. In SSCT, the beam width is taken to be the z-axis dose profile width measured at the isocenter between profile points corresponding to 50% of the maximum intensity (referred to as the FWHM). A collimator is designed such that the profile FWHM corresponds to the desired slice thickness (Goldman, 2008). That is, if MSCT detectors configured to acquire four 2.5cm slices are irradiated with 10mm wide X-ray beam, as specified for SSCT, outer 2 slices will receive lower intensity and yield higher image noise. To compensate, MSCT beams are widened to use only inner non-penumbra regions. Penumbra regions that were partially used in SSCT are discarded in MSCT, leading to reduced dose efficiency.

2.7 Factors affecting Dose in Computed Tomography

According to literature, there are many factors discovered to affect dose in CT practice. This include optimization of dose, that involves the inter-play between diagnostic quality of the CT image; radiation dose to the patient; and choice of scanning parameters or technique (Shrimpton et al., 2003). Equipment specification and design also plays

important role in dose reduction. The following are factors which determine the amount of dose a patient receives while undergoing CT scan.

Detectors: Two dose-relevant characteristics of a detector are quantum detection efficiency and geometric efficiency, which together describe the effectiveness of the detector in converting incident photon energy into signals. Detectors with high quantum detection, rapid response and low afterglow are preferable (e.g. Gadoliniumoxysulphide) compared to xenon gas detectors (Hsieh, 2006). This is because they require less radiation to form the image and hence less dose to the patient.

Collimators: Collimators positioned between X-ray sources prevent unnecessary radiation dose to the patient. Pre-patient collimators are positioned between X-ray source and patient to define X-ray beam coverage and avoid unnecessary radiation dose to the patient. With increased width of detector collimation, the geometric efficiency increases and the doses utilization is improved. Post-patient collimators are those located between the patient and the detectors mostly in front of the detectors to reject scatter radiation which improves image quality but sacrifices dose efficiency (Yu et al. 2009).

Beam shaping filters: X-ray beam-shaping filter is an important consideration for dose performance of an X-ray system. The X-ray filter is a physical object that attenuates and hardens the beam spectra so that the X-ray beam is hard enough to efficiently penetrate the patient. Filters are design to be of special shape (bowtie) in order to reduce the incident X-ray intensity in the peripheral region (McCollough, Primak, Saba et al, 2007), so that radiation dose to the patient especially the skin dose is reduced. Many filters are

available, and the appropriate for the clinical application most be selected for optimization.

Scan range: A scan range is a technique factor that is directly proportional to the radiation dose delivered to the patient. Scan length should cover only area of clinical significant in order to avoid unnecessary irradiation of body part that will not contribute any diagnostic information to the clinical purpose of such examination. Dose length product (DLP) is dependent on the length of the body part imaged, the higher the length the higher the DLP value and hence the higher integral dose. Many studies have identified the use of higher scan length as a major factor why they have higher DRL values (Roshan & Paul, (2010), Ogbole & Obed (2014) and Muhogora & Rehani (2014)).

X-ray tube current (mA): This determines the quantity of electrons that will be used to produce X-rays and consequently the amount of exposure. The tube current for a particular CT model increases proportionately with dose (EUR 16262, 1996). A patient with more body width requires an increase in the tube current to achieve adequate image quality. This is why size specific DRLs is encouraged (i.e. reduce tube current for thin patients and children). Nowadays tube current modulation have been employed in most scanners to ensure that dose is distributed according to weigh or size based, so that thicker parts of the body gets more and exposure to thinner parts is minimized (Yu, et al, 2009).

Tube kilovoltage (kV): Higher kV should be used especially when examining regions with higher absorption instead of higher mA values; when higher kV is used, beam hardening is achieved with higher X-ray absorption and penetration and the low energy

components of the X-rays that contribute more dose to the patient are reduced (Hottler, 2007).

Pitch: in helical CT, pitch has two terminologies depending on whether single slice or multi-slice. In single-slice helical CT, pitch is defined as the distance in millimeters that the table travels or moves in one complete rotation of the X-ray tube, divided by the nominal scan width (millimeters), this is termed detector pitch. Here, increasing the pitch by increasing the table speed can decrease both the radiation dose to the patient and scanning time, but at the cost of image resolution (Lewis, 2005). Beam pitch is the term used in multi-slice CT and is defined as table distance travelled in one 360* gantry rotation divided by total thickness of all simultaneously acquired slices. (Mehadevappa et al 2001) showed that radiation dose to a phantom was identical for varying pitch selections on a particular scanner; Somatom Plus 4 VZ multi-slice helical CT system. This was due to an automatic proportionate increase in tube current when pitch selection is increased. An advantage of increasing pitch in spiral multislice CT is to reduce scanning time, not to reduce dose (IAEA, 2013), hence it is appropriate to select pitch factor values that provide a balance between the image quality, scan time requirement and concerns for patient exposure.

Scan time: It is advantageous to select a scan time as short as possible particularly in abdominal or chest studies where heart movement and peristalsis may degrade image quality. Radiation dose to the patient increases with increasing scan time. Selecting larger slice thickness decreases dose because it takes less time to cover the scan length on the other hand selecting smaller cut means increase scan time and increase dose.

Slice or section thickness: Thinner slices are associated with higher patient dose to radiation but gives greater spatial resolution. When the slice thickness is reduced, the size or volume of the individual tissue volume element called voxel is reduced and smaller voxels will capture or absorb less total radiation or number of X-ray photons required to form the image. When the number of photons per voxel is reduced, image noise increases because of the statistical nature of photon interactions (IAEA, 2013). Therefore when slice thickness is reduced, dose has to be increased (mAs usually) in order to maintain the same level of noise as when a larger slice thickness is selected

Patient size or thickness: this determines the volume of investigation or imaging volume, which is the whole volume of the region under examination. The extent of the volume of investigation depends on the clinical needs. With regards to dose, the greater the volume, the higher the integral dose to the patient unless an inter-slice distance or pitch factor is used (EUR, 1996).

Access or availability of dose reducing software and algorithms: The different types of the new or modern iterative reconstruction algorithm and automatic exposure control software will reduce radiation dose to patient because of their capacity to reduce image noise and hence maximization of protection of patient. CT systems with any of this software or algorithm incorporated tend to optimize dose than system that lack this facility.

2.8 Dose Quantities in CT

Radiation dose in CT can be quantified in a variety of ways; scanner radiation output, organ dose and effective dose are several of the more common dose descriptors. The scanner radiation output is currently represented by the volume CT dose index (CTDIvol), dose length product (DLP), which describes the radiation output in relation to two standardized tissue equivalent acrylic phantoms (head and body CTDI phantoms of 16 and 32cm respectively) [figure 2.3]. The SI units are mGy and mGy.cm respectively. CTDIvol and DLP are the recommended dosimetric quantities in CT (EC, 2000).

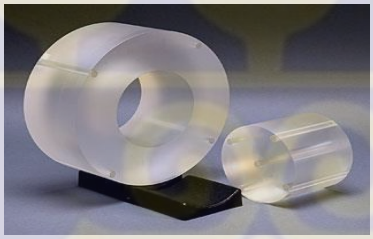

Figure 2.3: CT 32cm and 16cm acrylic phantoms, adopted from Google images (2014)

2.8.1 Computed Tomography Dose Index (CTDI)

CTDI represents an important CT-specific dose quantity, which relates the total amount of dose to an ideal rectangular dose profile along the z-axis, CTDI is calculated as follows:

$$\text{CTDI} = \frac{1}{d}\int_{-\infty}^{\infty} D(z)dz \quad \text{.....Eqn 1}$$

Or

$$CTDI = \frac{1}{TN}\int_{-\infty}^{\infty} D(z)dz \quad \text{.....Eqn 2}$$

Where,

d= the nominal slice thickness

D (z) = radiation dose along the Z-axis

Similarly, for the second equation representing multiple-detector CT;

N = is number of tomographic sections imaged in a single axial scan.

This is equal to the number of data channels used in particular scan. The value of N may be less than or equal to the maximum number of data channels available on the system (AAPM, 2008).

T = the width of the tomographic section along the z-axis imaged by one data channel.

In multiple-detector (multi-slice) CT scanners, several detector elements may be grouped to form a data channel. In a single-detector row (single slice) CT, the z-axis collimation (T) is the nominal scan width.

CT dose index 100 ($CTDI_{100}$)

*CTDI*100 represents the accumulated multiple scan dose at the center of a 100-mm scan and underestimates the accumulated dose for longer scan lengths. It is thus smaller than the equilibrium dose or the MSAD. The $CTDI_{100}$ require integration of the radiation dose profile from 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.

$$CTDI_{100} = \frac{1}{NT}\int_{-50mm}^{50mm} D(z)dz \qquad \text{.....Eqn. 3}$$

Weighted CT Dose Index (CTDIw)

This is the weighted average of the $CTDI_{100}$ measured at the center and the periphery points in the phantom. This represent absorb dose average in a scan plane or single cross section. This can be calculated from CTDI100 as follows (Figure 2.4):

$$\boldsymbol{CTDIw} = \left(\frac{2}{3}CTDI_{100}(periphery) + \frac{1}{3}CTDI_{100}(center)\right) \quad \text{.....Eqn. 4}$$

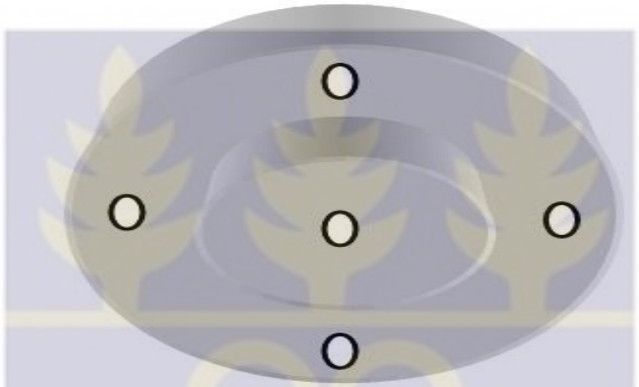

Figure 2.4: PMMA head/body phantom, with the holes (white) representing the periphery and the central measurement points.

Volume CT Dose Index (CTDI vol)

The CTDI vol provides a single CT dose parameter, based on a directly and easily measured quantity, which represents the average dose within the scan volume for a standardized (CTDI) phantom. It represent a specific scan protocol which is usually made of a series of axial scans taking into account the gaps or overlap between X-ray beams from consecutive rotation of the x ray tube. These can be deduced from CTDIw as follows (Figure 2.5):

$$CTDI_{vol} = \frac{N \times T}{I} \times CTDI_{w} \quad \text{.....Eqn. 5}$$

Where l = table increment per axial in mm.

Since pitch is defined as the ratio of table travel per rotation (l) to the total nominal beam width (N×T). Pitch = $\frac{l}{N \times T}$ ……...Eqn. 6

Then, the CTDI vol can be express as

$$CTDI_{vol} = \frac{1}{pitch} \times CTDIw \quad \text{……..Eqn.7}$$

CTDI w represents the average absorbed radiation dose over the *x* and *y* directions at the center of the scan from a series of axial scans where the scatter tails are negligible beyond the 100-mm integration limit. CTDIvol represents the average absorbed radiation dose over the x, y and z axis.

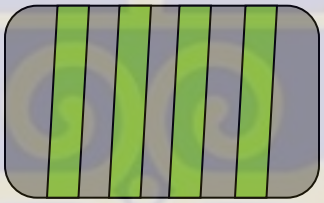

Figure 2.5: Average dose over scanned volume CTDIvol

2.8.2 Dose length product (DLP)

To better represent 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), where

DLP (mGy-cm) = *CTDI* (mGy) x *scan length* (cm)…… Eqn. 8

The DLP reflects the total energy absorbed (and thus the potential biological effect) attributable to the complete scan acquisition.

2.8.3 Effective dose (E)

It is important to recognize that the potential biological effects from radiation depend not only on radiation dose to a tissue or organ, but also on the biological sensitivity of the tissue or organ irradiated. The dose quantity that reflects the biological sensitivity of different tissues is the effective dose (E).

Effective dose can be estimated by the following means:

Effective dose (mSv) = DLP × CF …….. Eqn. 9

Where CF is conversion factor, calculated from National Radiological Protection Board (NRPB) Monte Carlo organ coefficients also represented as 'k' (Jones & Shrimpton, 1991),(Table 2.5) the value of k is dependent only on the region of the body being scanned. Based on this, Effective dose can be calculated from DLP displayed on most CT systems.

Table 2.5(a): Normalized values of effective dose- length product to a standard 16cm diameter phantom. (Stirling and Cottril, 2009)

Region of body	Effective dose per DLP (mSv/mGy.cm) by age				
	0 years old	1 year old	5year old	10 years old	Adult
Head	0.011	0.0067	0.004	0.0032	0.0021
Head and neck	0.013	0.0085	0.0042	0.0042	0.0031

Table 2.5(b): Amended normalized values of effective dose- length product to a standard 32cm diameter phantom. (Stirling and Cottril, 2009)

Region of body	Effective dose per DLP (mSv/mGy.cm) by age				
	0 years old	1 year old	5year old	10 years old	Adult
Chest	0.078	0.052	0.036	0.026	0.014
Abdomen and Pelvis	0.098	0.06	0.04	0.03	0.015
Trunk	0.088	0.056	0.038	0.028	0.015

The effective dose values can also be estimated from any of the available software developed by imPACT scan group which also make use of the NRPB-S250 Monte Carlo data sets (Lewis et al., 1997).

2.9 Dose reduction strategies in Computed Tomography

Improving dose efficiency in CT is related to CT system components, operator's choice of technique and manufacturer's role.

- **Technique measures of dose reduction**

The following gives a brief outline of technique measures if properly incorporated into practice, will reduce patient dose in CT, as listed in ICRP 87 publication: Limit the scan volume, reduce mAs values, use of automatic exposure control by adapting the scanning parameters to the patient cross section (with this 10-50% reduction in dose can be achieved without any loss in image quality). Use of spiral CT with a pitch > 1, shielding of superficial organs such as thyroid, breast, eye lens and gonads particularly in children and young adults, selecting separate factors for children, adequate selection of image reconstruction parameters, record of dose/exposure factors. Employing noise control strategies in image reconstruction and data processing by selecting optimal data processing and image reconstruction methods can generate images with lower noise levels. Image-based filtering technique is usually used to reduce image noise while still maintaining high-contrast resolution (Bai et al., 2009).

- **Examination-specific dose reduction techniques**

This is driven by the rapid technological advancement in CT with a wide range of clinical applications. CT capabilities has extended to cater for dedicated imaging of various body parts with different physiological function, the objectives of which is greatly to improve temporal and spatial resolution, noise reduction and dose reduction.

Dual-source CT (DSCT) technology for example was introduced in 2006 to provide the higher temporal resolution (83ms) and a much higher demand for radiation dose in cardiac CT imaging than non-cardiac CT imaging. The high resolution is achieved by simultaneously acquiring data from X-ray sources (tubes). Thus allowing cardiac scans at higher heart rate and hence noise reduction is achieved without the use of β-blockers to stabilize patients. Chinaiyan, (2014) evaluated the dose performance of DSCT and 64-slice CT scanners, and find out that DSCT make a large difference in terms of radiation exposure. It substantially reduces radiation exposure by up to 61%. This reduction of radiation is primarily due to the use of more aggressive ECG-pulsing window width and increased pitch especially for patients with higher heart rate, the use of cardiac bowtie filter and 3-dimensional adaptive noise reduction filter also contribute to the total dose reduction in DSCT (Bai et al, 2009).

Dual energy CT technology allows the use of low and high-tube potential data acquisition from a dual energy scan. Dose reduction is achieved using dual energy CT by allowing the creation of virtual pre-contrast images from a post contrast dual energy scan in CT examinations that will otherwise involve repeated scans thus the pre-contrast scan is avoided and hence dose reduction is achieved (Johnson et al., 2007).

CT perfusion examinations are examinations that require long scanning time long enough sometimes such that skin injury (deterministic effect) is possible, making radiation dose from such examination to be higher than routine CT scans (van der Molen & Geleijins 2007). In order to achieve dose reduction, lower technique factors (kV and mAs) which are undermined by increased noise and high artifact level are selected. With the use of recently proposed filtering techniques [highly constrained back projection local reconstruction (HYPR-LR) and multiband filtering (MBF)] that exploit spatial-temporal relationships of perfusion scans, image or quantum noise is reduced significantly to nearly the same level with routine dose images (Lui et al.,2009).

Interventional CT or CT fluoroscopy provides an effective image guidance tool for percutaneous interventional procedures (Daly et al., 1999). But are high dose scans because it involves long scan time and repeated scans. Dose reduction can be achieved by lowering tube potential, tube current and scan time, using or increasing larger slice thickness and limiting scan range to only body area of interest. Also recently proposed dose reduction methods in CT fluoroscopy which are lacking in older scanners include, an intermittent mode (or quick check mode) which delivers lower radiation dose than the older continuous mode (Carlson et al., 2001). The X-ray beam is turned-off during needle insertion which is followed by a very short CT scan to check for needle position.

- **Advanced/modern technology aided dose reduction techniques**

Other dose reduction methods which are undergoing trial have shown great prospect and are expected to make impact in clinical application of CT in the nearest future include; Iterative reconstruction which has been put into use in positron emission tomography (PET) and single photon emission computed tomography (SPECT) scanners are beginning to receive attention in conventional CT scanners. Lui (2014) in a review have shown that the new model-based iterative reconstruction algorithm has greater potential in reducing radiation dose in modern CT scans and improves spatial resolution and contrast resolution. It also reduces some artifacts when compared with earlier adaptive statistical iterative reconstruction (ASIR) and traditional filtered back projection (FBP) technique. The advantages of iterative reconstruction in dose reduction over existing or conventional filtered back projection technique include; the use of more accurate noise models which are based on photon statistics and incorporates other physical effects such as beam spectrum, noise, beam hardening effect, scatter and incomplete data sampling (Lui, 2014).

Individualizing scanning technique is another method which involves taking all patient-specific factors into consideration. Automatic exposure technique (AEC), design of different beam-shaping filters for different patient groups and clinical applications (Toth et al., 2005).

The use of photon-counting detectors by operation of pulse mode and current mode also promises to reduce dose in CT (Knoll, 2000). Photon-counting detectors are now being introduced into medical imaging systems and have been shown to have negligible levels

on electronic noise. The low noise performance enables new scanning techniques and ultimately reduces radiation dose to the patient (Levinson, 2014). Photon-counting can reduce signal to noise ratio (SNR) due to the elimination of electronic noise, and the benefit of this improved SNR can be used directly to reduce dose. Skappler et al.,(2014) also investigated CT dose using photoconductive (PC) detectors made of cadmium telluride (CdTe) and found out that in addition to dose reduction through efficient X-ray detection, PC detectors provide better contrast enhancement and enables new material-identification techniques for clinical CT.

CHAPTER THREE

Material and Methods

3.0 Introduction

This chapter describes the materials, method, organization, analysis and interpretation of data obtained, all aimed at answering the research question; what is the dose in $CTDI_{vol}$, DLP and mean effective doses received by patients undergoing CT scan of the head, chest and abdomen? The EC (1999) recommended methodology and approach was adopted. Doses from this study were compared with already established DRLs from other countries in order to assess CT practice in the study population. The data was then validated with the imPACT CT dosimetry calculator.

3.1 Materials

- **Selection of CT scanners:** Only multi-slice CT scanners that display dose description parameters ($CTDI_{vol}$ and DLP) were selected, with regards to International Electrotechnical Commission (IEC, 2002) requirement. Out of the six states that make up the region (North-central), a total of 4 CT scanners were selected randomly. Two CT scanners were selected from one state (Kwara state), because both were found to be functioning and having met the selection criteria, one site (CT scanner) each was then selected from two other states (Nasarawa state and FCT Abuja). Other available sites in this region could not participate in the study because of equipment breakdown while another site was yet to be commissioned as at the time of this study. The study sites comprised of 3 government or public centres and 1 privately owned centre. Majority (2 of 4) of scanners surveyed in this study were General Electric Health care (GE) scanners.

This is in line with Ugwu et al., (2009) findings; that majority of CT scanners in the country (Nigeria) are GE scanner products.

Dose data and scan parameter data were recorded during scan protocol selection and at the end of each examination, this is to ensure that dose data recorded is verified with the dose report that the machine provides at the end of an examination.

Table 3.1: Details of computed tomography scanners surveyed

Centre	**Centre type**	**Number of Scanners**	**Scanner name/model**	**Specifications**	**Number of CT scan per week**
A	Public	1	General Electric Light speed Delight	64 slice 40mm (64×0.625)	20
B	Public	1	General Electric Bright speed Excel	4 slice 20mm (4×5)	15
C	Private	1	Siemens Emtion 6 (Germany)	6 slice 18mm (6×3)	25
D	Public	1	Philip Brilliance	16 slice 12mm (16×0.75)	30

To ensure that all CT scanners were functioning to their optimal level and hence to guarantee validity and reliability of machine output, all the centres were found to have been practicing daily quality control test on their various machines in the form of tube warm up and air calibration.

- **A mobile weighing scale:** was used to acquire patient body weight in kilogram, in order to ensure that only normal sized adults were accommodated for this study. It was ensured that the weighing scale was in good working condition and always at the zero mark before it was used to weigh participants.
- **CT Dosimetry Software:** A commercially available CT dosimetry software ImPACT CT patient dosimetry calculator, version 1.0.4, from London, England, was used to validate and compare the dose values generated by the various CT scanners surveyed in order to ascertain the reliability of machine generated dose values. The impact scan dose evaluator with National Radiological Protection Board SR250 Monte Carlo data set is a system that models the conditions of exposure on a mathematical phantom for a range of common makes of CT scanners.

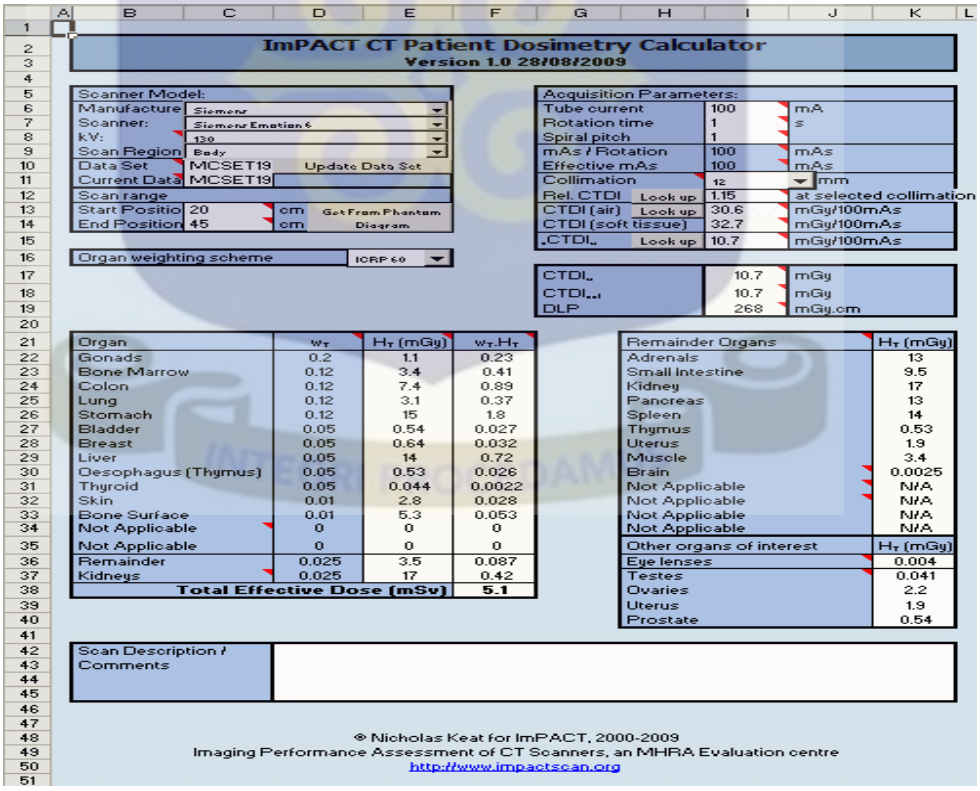

Figure 3.1: imPACT dosimetry calculator data package excel spread sheet

The software package provides values of CT dose index; these include CT dose index in air, weighted CT dose index, $CTDI_{vol}$, and the corresponding value of DLP when scanner specific information and scan parameter information are entered. The software also calculate effective dose using the National Radiation Protection Board (NRPB) organ dose coefficients and International Commission on Radiological Protection (ICRP) 103 organ weighing factors

3.2 Research methodology

A prospective quantitative research method was adopted to estimate radiation dose received by patients undergoing CT scan of the head, chest and abdomen. Numerical data was obtained from CT machine generated dose values and applied scan parameters used to scan standard sized patients in this part of Nigeria. There was no adjustment to the scan protocols adopted by all centres prior to this study. This was to ensure that the study reflects the practice in all the centres even before this study.

3.3 Site selection

North-central sub region of Nigeria was selected because this type of study had never been conducted there, there was availability of CT centres in the region, and it is the region where the researcher worked and resided. This region is one of the six geo-political zones of Nigeria comprising of six states and the Federal capital territory. It also provides a good representation of all the Nigerian ethnic/tribal population.

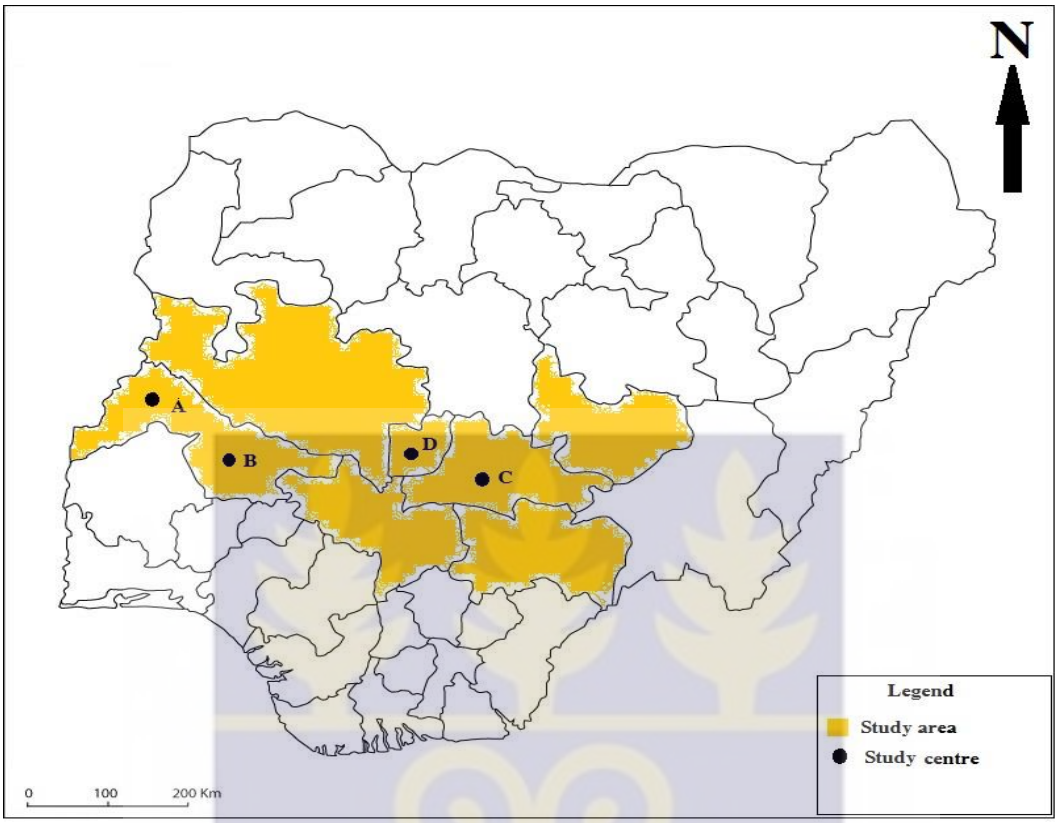

Figure 3.2: Map of Nigeria showing study area (North-Central Nigeria shaded in yellow colour) and study sites denoted by alphabets A-D. Adopted from world Google map

3.4 Study population

This comprise of patients referred for CT examination of the head, chest and abdomen

3.5 Sample size

A purposive sampling method was adopted as prescribed by the EC (1999), which recommends that: determination of DRL should be done using a minimum sample of 10 standard-sized (70±10kg) patients in each type of procedure or body part to be examined. A total of 226 patient's data with distribution 88(38.9%) brain, 60 (26.6%) chest and 78 (34.5%) abdominal scan were collected and surveyed from the 4 participating centres.

More than ten samples were surveyed for most of the procedure/examination in order to increase the statistical relevance of data. Only adults (15years and above) that weighed between 70 ±10 kg and agreed to participate in the study were included in this study.

3.6 Participant selection

Inclusion criteria

- Only standard-sized adult male and female patients whose weight were within 70±10 kg (EC, 1999), and were referred for only routine examinations of head, chest and abdominal scan.
- Multi-slice CT with capability to display dose parameters on its console
- Nigerian nuclear regulatory authority (NNRA) accredited CT centers

Exclusion criteria

- Adult patients whose body weights were above or below recommended standard-size (70± 10 kg).
- Patient referred for special CT examinations such as CT urography, perfusion studies, CT angiography, etc.
- Critically ill patients were exempted
- Single slice CT (SSCT)

3.7 Data collection

Data was collected by researcher with the assistance of qualified CT radiographers (research assistant) who were trained on how to take the data. Data collected were of the existing scanning protocol at each unit, no modification in order to reflect the existing practice at various centres. Data recorded include individual patient demographic information (age, sex and weight), CT scan parameters [tube voltage (kV), tube current

(mA), pitch, tube rotation time, field of view (FOV), scan length, number of slices and scan mode] and CT dose parameters ($CTDI_{vol}$ and DLP).

3.8 Ethical considerations

There was adherence to confidentiality principle by ensuring that patient and Centre name was not included in the data booklet. Permission was sought from the centres to allow the use of their facilities for this study. A template of the clearance or permit is presented in appendix I.

3.9 Analysis of data

Statistical analysis was performed using Microsoft excel 2010 version. Data obtained was entered into excel spread sheet (appendix a, b, c and d). The analysis of results employed the use of descriptive and inferential methods of data analysis. Quantitative variables were expressed by descriptive analysis to summarize and show variability of the data for the study in mean, range and standard deviation.

$CTDI_{vol}$ and DLP data from each site were averaged and the rounded 75^{th} percentile was used to calculate a DRL for each site and for the region by compiling all the results from all centres. Comparison was made between the estimated doses and data from recommended standards (countries) where there are existing established DRLs. Statistically significant results of scan parameter between study scan parameters and European scan parameters were determined using percentage. Percentage coefficient of variation was used to validate or compare surveyed data (scanner generated data) with imPact CT dosimetry software generated dose values.

CHAPTER FOUR

Results and Discussion

Introduction

A total of 226 patient's data was collected, out of which 69.1% were male and 30.9% were female, they had a mean weight of 69.8kg. Eighty-eight brain scans (38.9% of data), 60 chest scans (26.5% of data) and 78 abdominal scans (34.5% of data) were surveyed from the 4 participating centres (one private and three public healthcare centres). All scanners surveyed had multislice capabilities ranging from 4 to 64 slices (Table 3.1). All the centres were able to survey the minimum recommended number (at least 10) of patients recommended for each examination within a four month data collection period, though others reported more than 10 for majority of the examinations. Patient's age ranged from 16 – 90 years, since hospital age classification in Nigeria considered sixteen years of age as an adult (Garba, 2014). All the centres adopted axial/sequential mode for their CT brain scans so therefore this study surveyed sequential scan mode for its brain CT scan.

Summary of statistical distribution of the study data is presented in Table 4.1. Head scans (N=88) were requested more than any other examination in all the centres while chest scans (N=60) are the least requested examinations. Brain scan also presents higher $CTDI_{vol}$ and DLP dose values. This is because the diameter of the head is smaller which allows radiation dose to be distributed in smaller volume compared to the larger diameter and volume in other body regions (chest and abdomen) as well as tube current (mA) used for head scan is higher than that used for the body scans.

Table 4.1: Summary of mean, range (maximum and the minimum) and 75th percentile for brain, chest and abdominal CT scan in $CTDI_{vol}$ (mGy) and DLP (mGycm) for this study

Examination		N	Mean	Range	75th percentile
Head					
	CTDIvol	88	52.2(±10)	25-69	60
	DLP		841.5(±324)	320- 1968	1024
Chest					
	CTDIvol	60	8.8(±4)	2-23	10
	DLP		333(±173)	72-1048	407
Abdomen					
	CTDIvol	78	12(±5)	4-23	15
	DLP		590(±260)	218-1149	757

N is the number of participants.

4.1 Establishing the diagnostic reference levels (DRLs)

Details of the descriptive statistics from the surveyed examinations in dose quantities CTDIvol (mGy), DLP (mGycm) and mean effective dose (mSv) are presented in Figures 4.1, 4.2 & 4.3. The mean local DRLs in $CTDI_{vol}$ and DLP for each CT examination were calculated for each centre, this was also used to compare doses across CT centres. The individual centres are denoted by the alphabets A, B, C and D in order to avoid mentioning centre names for confidentiality purpose.

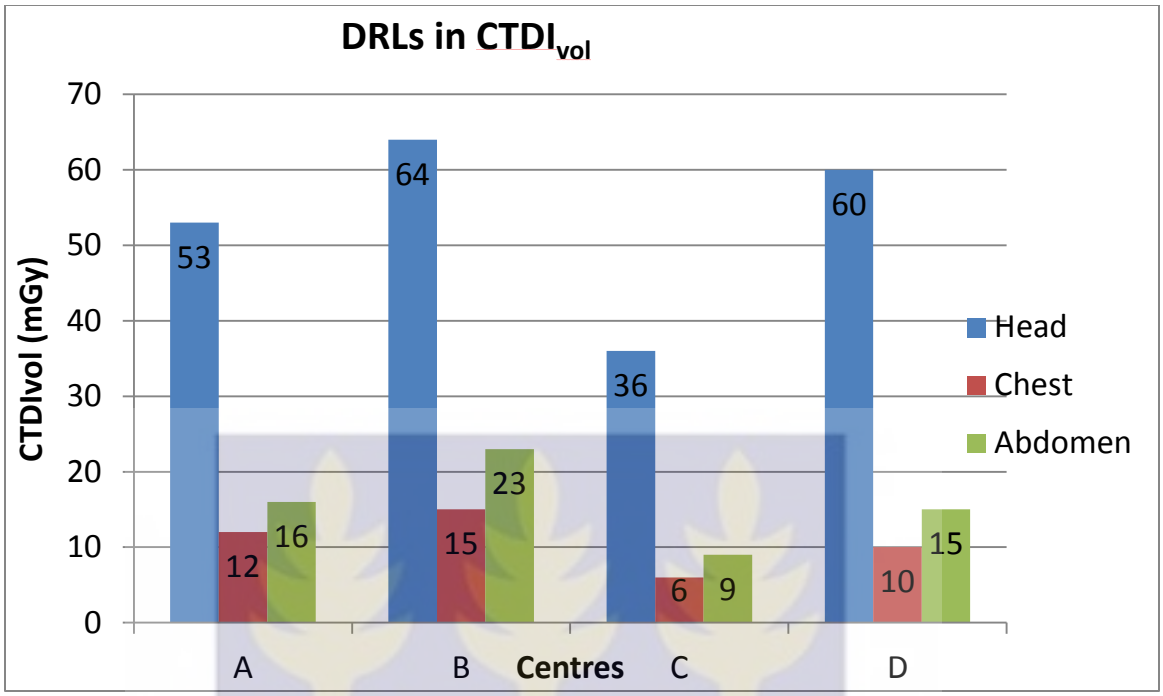

Figure 4.1: $CTDI_{vol}$ for CT centres under study

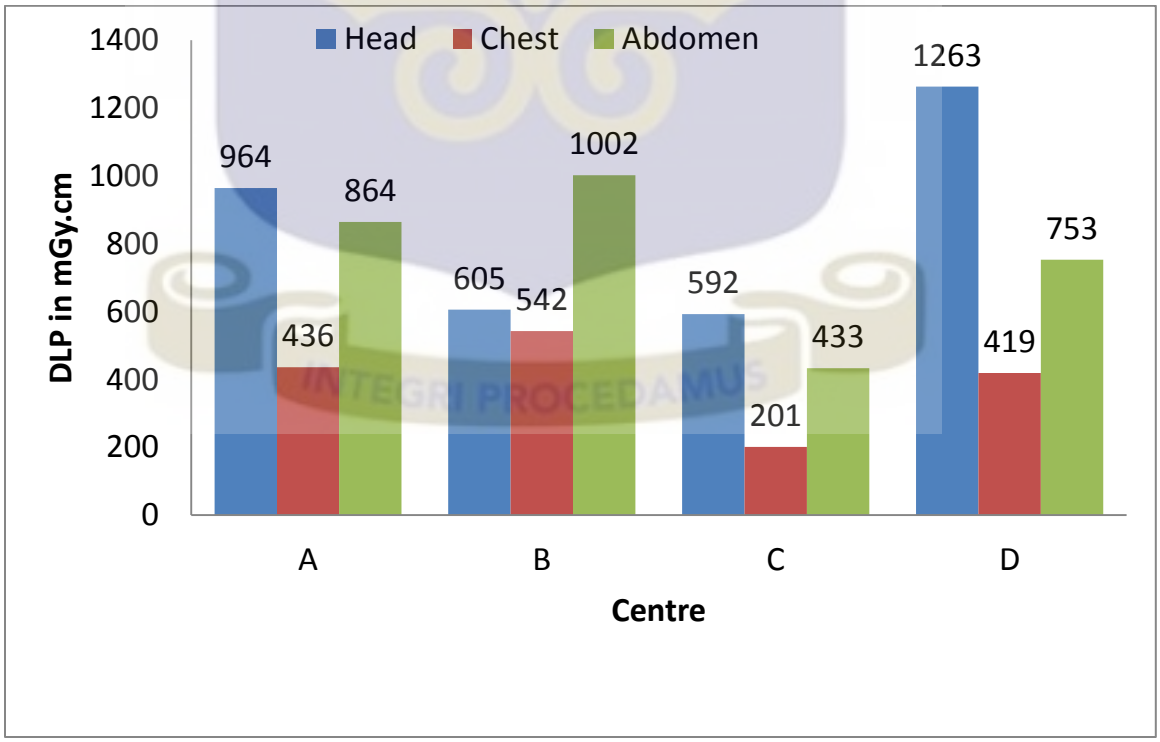

Figure 4.2: DLP for CT centres surveyed

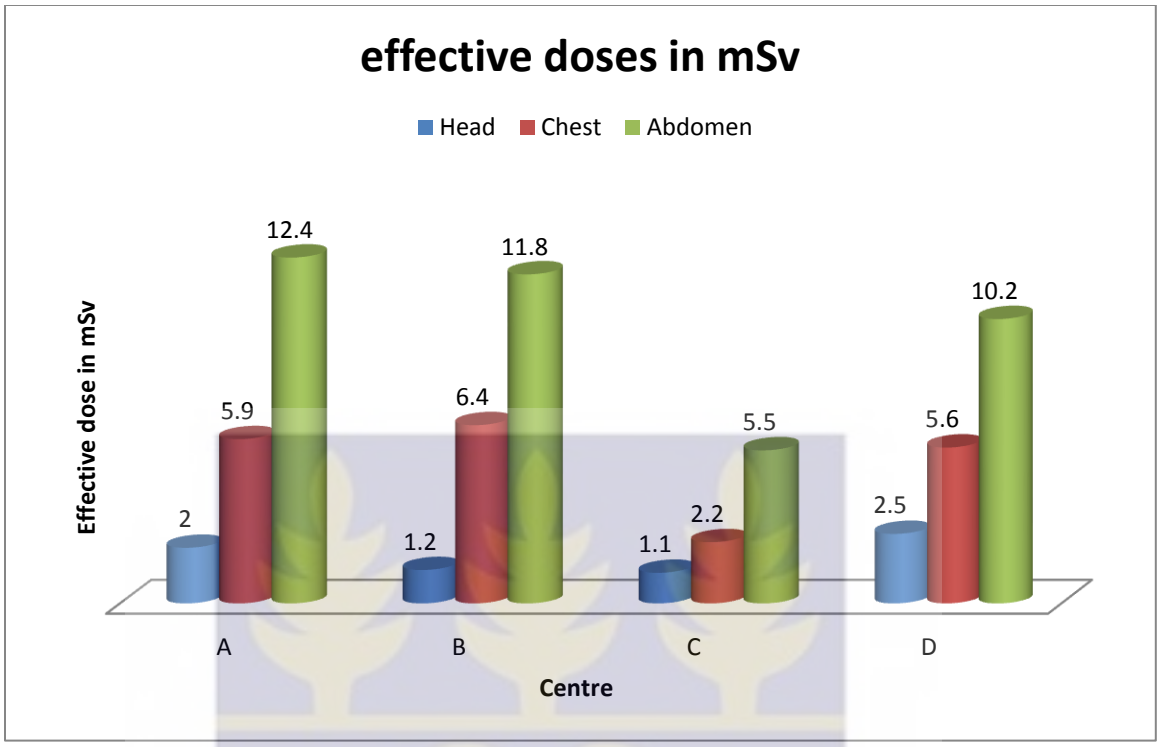

Figure 4.3: Mean effective dose across centres for head, chest and abdominal CT scan.

The effective dose was calculated using the adult normalized values of effective dose per dose-length product (DLP) over various body regions as shown in Tables 2.5 a & b. Effective dose is an expression or meaningful way of comparing doses from practices since it describes overall risk to organs from a radiation dose. This is represented as a distribution of known radiation doses to individual organs converted into an equivalent uniform whole body radiation. Variation in effective dose values was observed in this study (Figure 4.3). This is mainly due to the fact that considerable variation was observed in the individual/ local centre dose-length product values employed for the examinations (Figure 4.2).

Estimated regional diagnostic reference levels (DRLs) for Head, Chest and Abdominal CT examinations from this study expressed in $CTDI_{vol}$ (mGy) and DLP (mGycm) is

presented in Table 4.2. These values represent 75th percentiles of the mean values for a particular examination from all the centres surveyed. The 75th percentiles imply that 75 percent of the centres surveyed operate at or below the dose values presented for all the categories of examination surveyed. These values represent values with which CT practice in this region can be compared to recommended standards, other countries and regional surveys. Centres with dose values mainly within the remaining 25th percentile i.e. above 75th percentile for a particular examination are considered as unusually high doses, which should be considered for downward review in order to achieve optimization.

Table 4.2: Estimated regional diagnostic reference levels

Diagnostic reference levels	Examination		
	Head	Chest	Abdomen
CTDIvol (mGy)	60	10	15
DLP (mGy.cm)	1024	407	757

4.2 Comparison of diagnostic reference levels (DRLs) and effective dose

Table 4.3 shows the values of diagnostic reference levels from this study compared to that from United Kingdom which surveyed only multislice computed tomography CT scanners and European commission diagnostic reference levels which surveyed mainly single slice CT scanners (Shimpton et al., 2005; European Commission; 1999). As shown, diagnostic reference levels in North-Central Nigeria is less than EC 1999 DRLs but higher than the more recent UK diagnostic reference levels for head and abdominal CT scan.

Table 4.3: Comparison of diagnostic reference levels (DRLs)

Examination		Mean value	75th percentile value	European DRL	U.K Study2003
Head					
	CTDIvol(mGy)	52	60	60	65
	DLP(mGy.cm)	841	1024	1050	930
Chest					
	CTDIvol(mGy)	8.8	10	30	14
	DLP(mGy.cm)	333	407	650	580
Abdomen					
	CTDIvol(mGy)	12	15	35	14
	DLP(mGy.cm)	590	757	780	560

Comparison of dose values in terms of dose quantities $CTDI_{vol}$ and DLP from this study with two available literatures of CT dose survey estimates from Nigeria (Table 4.4). The results show that one of the study (Garba, 2014) only estimated CT doses for head scans and quantified CT doses using the older $CTDI_w$, but his dose values in DLP (789 mGycm) tends to be lower compared to that of this study (1024 mGycm). Higher values from this study can be attributed to the fact that this study accommodated larger number of participants as well as larger number of different CT machine manufacturers. The CTs have different inherent machine design and technological approach to dose reduction. The other study (Ogbole & Obed, 2014) is a survey result from one CT scanner (GE 16-slice), which presents or report higher dose values in all the examination categories compared to values from this study.

Table 4.4**:** Mean CTDI (mGy) and DLP (mGycm) for this study and two other Nigerian studies.

Examinations	This study	Garba (2014)	Ogbole & Obed (2014)
Head			
CTDI $_{vol}$	**60**	$CTDI_w$ **76**	**73.5**
DLP	**1024**	**789**	**1898**
Chest			
CTDI $_{vol}$	**10**	**N/A**	**22.7**
DLP	**407**		**1189**
Abdominal			
CTDI $_{vol}$	**15**	**NA**	**37.9**
DLP	**757**		**1902**

Table 4.5**:** Mean effective doses from this study compared to average and range of effective dose values for computed Tomography examinations compiled from reported literatures (Mettler et. al., 2008).

Examination	Average Effective Dose (mSv)	Values reported in literatures (mSv)	Mean effective Dose from this study (mSv)
Head	**2**	**0.9-4.0**	**1.7**
Chest	**7**	**4.0-18.0**	**5.0**
Abdomen	**8**	**3.5-25**	**11.9**

Mean effective dose value estimates from this study (1.7mSv, 5.0mSv and 11.9mSv) fall within range of values reported in literatures (Mettler et. al., 2008) for all the examination

groups (head, chest and abdomen). The values are away from the lower and upper limits of the reported values. However estimated mean effective dose values vary slightly when compared to the reported average effective dose values (2mSv, 7mSv and 8mSv).

4.3 Scan/exposure parameter results and comparison study

Mean values of scan/exposure parameters per centre are presented in Table 4.6, these parameters are measures of comparing scan protocols or practice across centres. The manipulation and choice of these parameters directly influence dose delivered, parameters surveyed include; Tube kilovoltage (kV), Tube current in milliamperage (mA), Time (s), and Pitch, Scan length (cm), Number of slices and Slice thickness (mm). This study observed a direct proportional variation in dose (CTDIvol, DLP & mean effective dose) with regards to tube current mA choice, an increase in tube current (mA) showed proportionate or correspondent increase in all dose quantities.

Table 4.6: Mean values of scan/exposure parameters per centre

Centre	Examination	kV	mA	Tube rot. Time	Pitch	Scan length	No. of slices	Slice thickness
A	Head	120	338	0.8		19	36	5
	Chest	120	162	0.7	0.98	38	64	5
	Abdomen	120	190	1	0.98	51	92	5
B	Head	120	200	1		10	37	2.5, 5
	Chest	120	100	1	0.75	34	64	5
	Abdomen	120	200	1	0.75	44	70	5
C	Head	110	130	1.5		16		6
	Chest	110	79	1	0.85	33		10, 5
	Abdomen	130	104	1	0.85/1.5	45		10
D	Head	120	401	1		19	54	5
	Chest	120	150	1	0.935	40	88	5
	Abdomen	120	220	1	0.935	45	103	5

Most of the centres maintained a common tube voltage value for all the three categories of examination except centre C, but no significant variation or effect on dose was noticed. Variations in pitch value are likely to affect dose values, such that the dose is reduced by half if pitch is doubled and other factors remained unchanged. Number of cuts or slices requested for an examination also influence dose. An increase in dose was observed for a study that had more cuts or slices compared to a similar examination or scan with lesser number of cuts. Centre **C**, showed better optimization of scan parameters as shown in Figure 4.2, it uses the lowest tube current (mA) and the highest pitch value and hence it delivers the lowest dose to patients as shown in the dose estimates presented above (Figure 4.1 To 4.3) in the dose results.

Figure 4.4 shows the values of scan parameter (scan length in cm) for Head, Chest and Abdominal CT scans respectively. Scan length from this study (15.7, 36 and 46) compared to that from United Kingdom (12.7, 39.3, and 41) and National Radiological Protection Board (14.6, 25, and 40) (UK 2003; NRPB 2005). Scan length used for Head and abdominal CT scan in North-Central Nigeria are higher than scan length in the two comparing references except for chests CT scans.

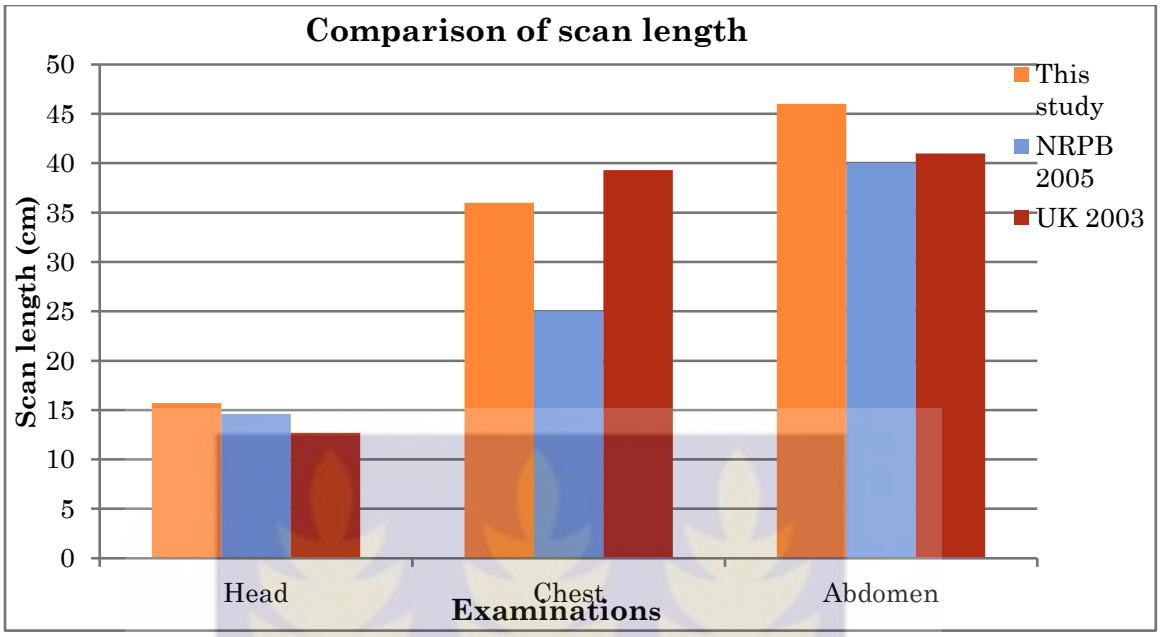

Figure 4.4: comparison scan length

4.4 Validation results (Scanner generated values and Dosimetry software)

Table, 4.7 (a, b, & c) shows comparison of dose in $CTDI_{vol}$/DLP for head, chest and abdominal examinations in percentage (%) coefficient of variation between scanner generated and dosimetry software calculator package generated values across the study centres.

Table 4.7a: Centre A with GE light speed VCT, 64 slice CT scanner

Examination	Dose quantity	% Coefficient of variation	Overall average variation
Head	$CTDI_{vol}$ DLP	11 1	4.9
Chest	$CTDI_{vol}$ DLP	6 7	
Abdomen	$CTDI_{vol}$ DLP	0.5 4	

Table 4.7b: Centre C with Siemens Emotion 6, 6 slice CT scanner

Examination	Dose quantity	% Coefficient of variation	Overall average variation
Head	$CTDI_{vol}$ DLP	5 7	5.8
Chest	$CTDI_{vol}$ DLP	5 11	
Abdomen	$CTDI_{vol}$ DLP	2 5	

Table 4.7c: Centre D with Philips Brilliance 16 slice CT scanner

Examination	Dose quantity	% Coefficient of variation	Overall average variation
Head	$CTDI_{vol}$ DLP	1 0.1	6.5
Chest	$CTDI_{vol}$ DLP	11 9	
Abdomen	$CTDI_{vol}$ DLP	9 9	

4.5 Discussion

This study represents the collection of estimated local Computed Tomography DRLs in North-Central region of Nigeria and it revealed enormous variations between the different centres in reported local DRLs and individual patient doses. The reasons for these variations in line with many DRL studies (Saravankumar et al., 2014, Roshan and Paul, 2014, Olarinoye & Sherifat 2010), are mainly attributed to different exposure parameters and radiographic technique. This shows a huge optimization potential among almost all the centres and standardization of practice is also lacking. Considerable reduction in dose delivered and contribution by CT can be achieved if optimization is adopted by ensuring that examination protocols exceeding the study's 75th percentiles are adjusted and then reviewed.

A comparison of DRLs or estimated 75th percentile values from this survey, EC 1999 and UK 2003 studies as shown in (Table 4.3); 75th percentile values in EC 1999 tends to have higher values compared to that of this study. This can be attributed to the fact that most of the scanners surveyed in EC 1999 are single slice CT scanners which are associated to higher dose delivery compared with technologically advanced or modern multislice CT scanners (Goldman, 2008), which were surveyed in our study. UK 2003 study is a better means of comparing with this study since the values were obtained from a survey of multislice CT scanners. However, result of the comparison revealed a need for optimization of doses in North-Central Nigeria, since 75th percentile values of the $CTDI_{vol}$ and DLP from this survey (60mGy, 1024mGy.cm: 10mGy, 407mGy.cm and 15mGy, 754mGy.cm) are higher for all the examinations except for chest scan compared

to the UK DRLs (65mGy, 930mGy.cm: 14mGy, 580mGy.cm and 14mGy, 560mGy.cm) for head, chest and abdominal CT scans respectively.

Scan parameters such as kV, mA, pitch factor, scan time, scan length and slice thickness influence dose. kV, mA and scan time have directly proportional relationship with absorbed dose. All the centres adopt different parameters (kV mA and scan time) for different examination types. Amongst all the scanners/centres studied, centre C (Siemens electronics) optimized its practice the most, as it recorded the least $CTDI_{vol}$, DLP and mean effective dose values for all examination categories. This is because the scanner adopted or used tube current (mA) and scan length less than other scanners. This scanner also used the highest pitch value especially for abdominal CT scans. This implies that they avoided overlap of adjacent slices in helical scans and hence absorbed dose is reduced. This is in line with Chinnaiyan et al., (2014) findings.

Scan length is the most compared exposure parameter in CT scan because dose length product (DLP), which describes the dose delivered in a scan volume is directly depended on the length of the scanned body region. Although, image quality criteria or subjective request of varying scan lengths by radiologist and lack of standardization of protocol may be the cause of variation in the selection or coverage of area scanned. A properly selected scan length should only include areas of diagnostic significance and exclude areas not indicated or of no diagnostic benefits (Shrimpton et al., 2003). Comparison of this study scan length with that of UK 2003 and NRPB 2005 (Figure 10) shows that in all examination categories (head, chest & abdomen) surveyed, scan length can still be reduced by 12.3%, 14.4% and 11.5% for head, Chest and Abdomen respectively without losing any area of diagnostic benefit. If these reductions in scan length are applied

throughout the centres then a significant reduction in absorbed dose per patient volume scanned can be achieved. This study also reported that head scans are the most requested CT scans in line with Erondu et al. (2011) and Foley et al. (2012) findings.

Comparison of head, chest and abdominal examinations dose values generated by scanners and dosimetry software package are presented in (Table 4.6 a, b, & c). With the exception of centre B (GE, Bright speed Hct, 4 slice CT scanner) whose model was not found on the software, typical variation from the mean value according to the scanner generated and software calculated values per examination, per scanner as well as overall average were below 10%. The variations were least observed in the head scan dose values across all the scanners. This may be due to the fact that head scans were acquired using an axial scan and not helical scan that depend on a lot of complex but adjustable manufacturer depended parameters (pitch, filters) and various dose reduction softwares. This unity between scanner generated and ImPACT software calculated dose values is in line with International Electro technical Commission (2001) recommendation on the use of scanner generated parameters in the setting of diagnostic reference levels. Walter et al., (2008) have also shown that the ImPACT spread sheet or data package agree with other commercially available dosimetry software packages with only a variation of approximately 5 percent. Therefore, reliability in the use of scanner generated dose values and the use of ImPACT dosimetry software with NRPB data set is once again justified.

CHAPTER FIVE

Conclusion and Recommendations

5.1 Conclusion

This study provided estimates of computed tomography dose parameters for common computed tomography examinations in the North-Central sub-region of Nigeria. The CT examinations are description of the current practice in this region and hence, may represent the diagnostic reference levels for the region at least before the establishment of Nigerian national diagnostic reference levels. A means of comparing CT practice and data for dose audit are provided. From the analysis of the results, the number of head scans is more than the rest of the examinations in all the centres studied followed by abdomen and the least is chest. Brain scan also presented higher $CTDI_{vol}$ and DLP values due to smaller diameter of the head allowing radiation to be distributed in a smaller volume compared to other parts of the body considered in the study. Also, the abdomen had the highest effective dose, because longer scan lengths is required to cover its volume, this is followed by the chest and the least is the head. Variation in effective doses was observed across centres and was attributed mainly to variations in individual or local centre DLP values. Similarly reported CT doses and applied scan parameters such as scan length in this study are slightly above recommended standards (UK DRLs). This indicates the need for optimization. The reasons for higher DRLs were because longer scan lengths, higher mA and lower pitch values are being used. The need for standardization of practice is also required as a result of observed variation of scan parameters for the same examination across centres.

Manipulation of scan parameters such as kV, mA, scan time and scan length influence absorbed dose in CT examinations, so operator's knowledge about diagnostic reference levels and experience in the selection exposure parameters play important role in CT dose optimization.

This study suggests Standardization of practice across centres as a means to increase optimization of doses because of observed variations in exposure parameters selection for the same examination.

5.2 Recommendations

To Hospital Authorities;

- There is need to increase practitioners (Radiographers, Radiologist, Medical Physicist and Hospital Authorities) awareness about the significance, adherence and application of diagnostic reference levels in the optimization of computed tomography practice. .

To Regulatory Authorities;

- This study suggests a nationwide survey of CT doses to have a better picture of computed tomography practice and optimization in Nigeria.
- The Nigeria Nuclear Regulatory Authority should incorporate DRLs into its regulatory control programmes and or adopt stringent measures to enforce compliance by hospital authorities and must ensure that practices are reviewed from time to time.

To Scientific Community;

- Experimental measurement of computed tomography machine dose output to compare Computed Tomography generated dose values.

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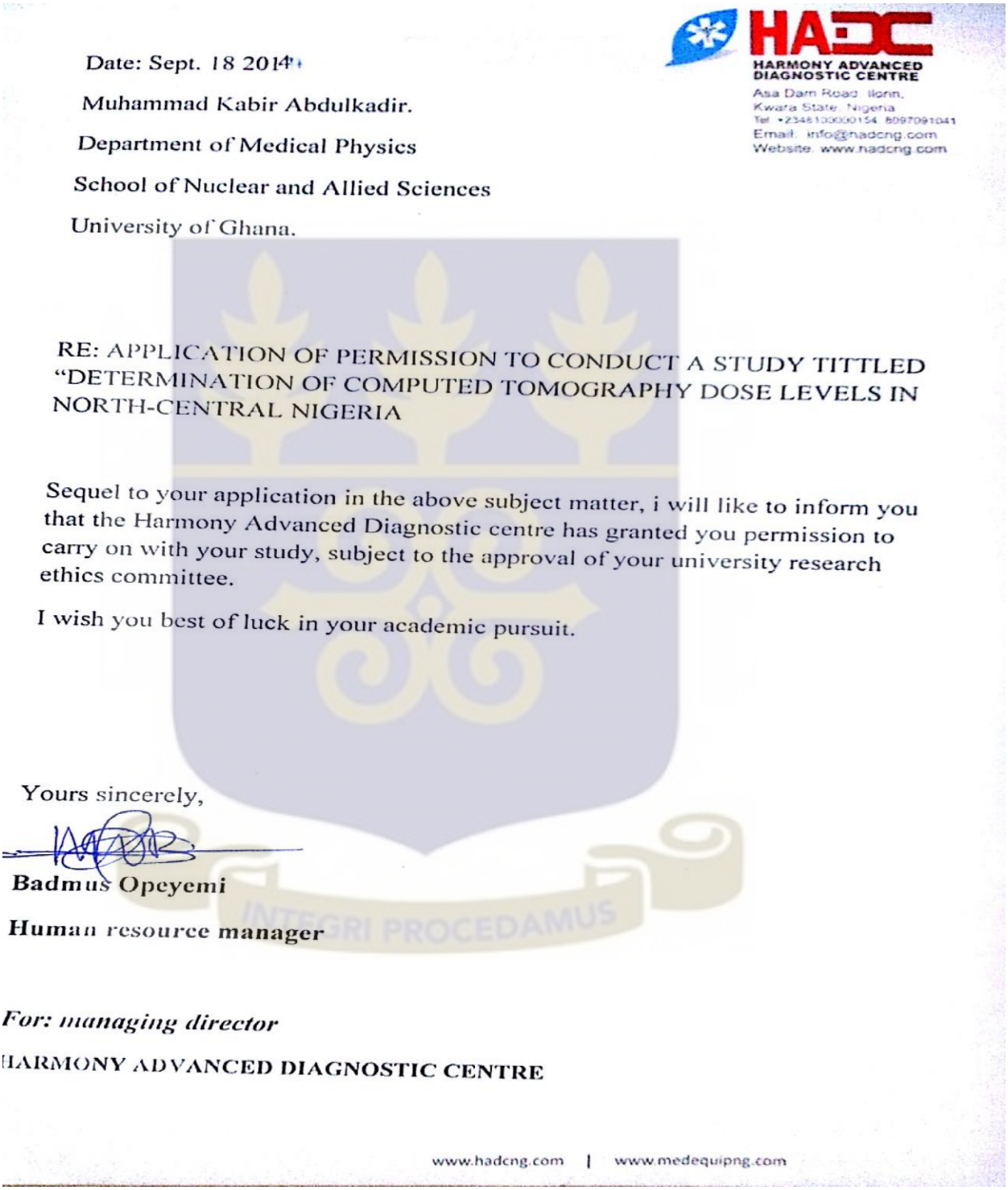
APPENDICES

APPENDIX M: ETHICAL CLEARANCE FROM STUDY SITES



HARMONY ADVANCED DIAGNOSTIC CENTRE
Asa Dam Road, Ilorin,
Kwara State, Nigeria
Tel: +2348100000154, 8097091041
Email: info@hadcng.com
Website: www.hadcng.com

Date: Sept. 18 2014

Muhammad Kabir Abdulkadir.

Department of Medical Physics

School of Nuclear and Allied Sciences

University of Ghana.

RE: APPLICATION OF PERMISSION TO CONDUCT A STUDY TITTLED "DETERMINATION OF COMPUTED TOMOGRAPHY DOSE LEVELS IN NORTH-CENTRAL NIGERIA

Sequel to your application in the above subject matter, i will like to inform you that the Harmony Advanced Diagnostic centre has granted you permission to carry on with your study, subject to the approval of your university research ethics committee.

I wish you best of luck in your academic pursuit.

Yours sincerely,

Badmus Opeyemi

Human resource manager

For: managing director

HARMONY ADVANCED DIAGNOSTIC CENTRE

www.hadcng.com | www.medequipng.com

APPENDIX A: CENTRE A (BRAIN)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	SCANNER	SCAN MODE	KV	mA	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	TUBE SPEED	FOV	CTDIvol	DLP	E(mSv)
1	1630	34	M	73	MULTISLIC	AXIAL	120	350	5	36	18	0.8	32	51.7	940	1.97
2	1632	70	M	65	MULTISLIC	AXIAL	120	310	5	32	15.8	0.8	26	54.7	869	1.82
3	1639	68	M	68	MULTISLIC	AXIAL	120	300	5	32	16.1	0.8	25	53.2	866	1.81
4	1636	51	F	66	MULTISLIC	AXIAL	120	290	5	60	30	0.8	25	51.7	1554	3.2
5	1648	33	M	74	MULTISLIC	AXIAL	120	310	5	35	19	0.8	25	54	1052	2.2
6	1651	57	M	67	MULTISLIC	AXIAL	120	310	5	36	18	0.8	25	54	975	2.04
7	1653	70	F	65	MULTISLIC	AXIAL	120	280	5	28	16	0.8	25	50	828	1.73
8	1654	25	M	70	MULTISLIC	AXIAL	120	340	5	33	18	0.8	28	60	1107	2.32
9	1655	32	F	69	MULTISLIC	AXIAL	120	300	5	36	18	0.8	25	53.7	965	2.02
10	1656	80	M	71	MULTISLIC	AXIAL	120	270	5	36	19	0.8	25	47	898	1.88
11	1658	57	M	66	MULTISLIC	AXIAL	120	300	5	36	18	0.8	25	53	965	2.02
12	1659	85	M	65	MULTISLIC	AXIAL	120	280	5	36	19	0.8	25	49	939	1.97
13	1662	37	M	66	MULTISLIC	AXIAL	120	280	5	36	18	0.8	25	49	923	1.93
14	1663	50	F	61	MULTISLIC	AXIAL	120	290	5	32	16	0.8	25	52	850	1.78
15	1665	64	M	72	MULTISLIC	AXIAL	120	280	5	36	18	0.8	25	50	933	1.95
16	1672	47	F	69	MULTISLIC	AXIAL	120	300	5	36	18	0.8	25	52	951	1.99
17	1675	46	M	75	MULTISLIC	AXIAL	120	280	5	36	19	0.8	25	49	927	1.94
18	1676	65	M	70	MULTISLIC	AXIAL	120	300	5	36	16	0.8	25	53	861	1.8
19	1677	34	M	68	MULTISLIC	AXIAL	120	280	5	36	19	0.8	25	49	927	1.94
20	1679	61	M	76	MULTISLIC	AXIAL	120	280	5	36	19	0.8	25	50	949	1.99
21	1707	50	M	79	MULTISLIC	AXIAL	120	300	5	44	22	0.8	25	53	1185	2.48
22	1685	31	M	70	MULTISLIC	AXIAL	120	290	5	36	18	0.8	25	52	961	2.01
23	1686	82	M	72	MULTISLIC	AXIAL	120	280	5	36	19	0.8	25	49	942	1.97
24	1687	64	M	69	MULTISLIC	AXIAL	120	280	5	36	19	0.8	25	50	942	1.97
25	1689	41	F	65	MULTISLIC	AXIAL	120	270	5	36	18	0.8	25	47	874	1.83
26	1692	85	F	80	MULTISLIC	AXIAL	120	300	5	36	18	0.8	25	53	956	2
27	1698	65	F	68	MULTISLIC	AXIAL	120	280	5	36	18	0.8	25	50	921	1.93
28	1700	78	M	73	MULTISLIC	AXIAL	120	270	5	36	20	0.8	25	48	953	2

APPENDIX B: CENTRE A (CHEST)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	TUBE SPEED	PITCH	FOV	CTDIvol	DLP	E(mSv)
1	1628	70	M	65	120	140	MULTISLICE	HELICAL	5	46	38	0.7	0.98	36	9	336	4.7
2	1632	60	M	68	120	160	MULTISLICE	HELICAL	5	47	34	0.7	0.98	36	10	351	4.91
3	1640	47	M	72	120	140	MULTISLICE	HELICAL	5	68	38	0.7	0.98	36	8	325	4.55
4	1656	67	M	70	120	150	MULTISLICE	HELICAL	5	67	34	0.7	0.98	36	10	335	4.69
5	1659	16	M	75	120	160	MULTISLICE	HELICAL	5	48	35	0.7	0.98	32	8	372	5.2
6	1697	41	F	71	120	160	MULTISLICE	HELICAL	5	52	35	0.7	0.98	36	11	379	5.3
7	1783	47	M	69	120	255	MULTISLICE	HELICAL	5	91	49	0.7	0.98	46	21	1048	14.67
8	1803	71	F	71	120	230	MULTISLICE	HELICAL	5	71	40	0.7	0.98	36	15	608	8.51
9	1849	41	F	68	120	150	MULTISLICE	HELICAL	5	69	38	0.7	0.98	38	8	307	4.29
10	1870	53	M	72	120	84	MULTISLICE	HELICAL	5	88	48	0.7	0.98	32	4	178	2.49
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				APPENDIX C: CENTRE A (ABDOMEN)													
S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	SCAN TIME	PITCH	FOV	CTDIvol	DLP	E(mSv)
1	1631	51	M	70	120	145	MULTISLIC	HELICAL	5	93	50	0.5	0.93	36	6.82	345	5.17
2	1633	68	M	70	120	350	MULTISLIC	HELICAL	5	32	54	0.5		36	16	866	12.99
3	1634	18	M	65	120	110	MULTISLIC	HELICAL	5	90	48	0.5		36	5	263	3.94
4	1637	52	M	69	120	150	MULTISLIC	HELICAL	5	87	47	0.5		36	7	345	5.17
5	1645	37	M	71	120	110	MULTISLIC	HELICAL	5	86	47	0.5		36	5	236	3.54
6	1646	28	F	68	120	270	MULTISLIC	HELICAL	5	101	54	0.5		36	13	757	111.35
7	1649	33	F	74	120	170	MULTISLIC	HELICAL	5	87	47	0.5		36	8	390	5.85
8	1650	59	F	80	120	170	MULTISLIC	HELICAL	5	89	49	0.5		36	8	385	5.77
9	1664	50	F	71	120	380	MULTISLIC	HELICAL	5	100	54	0.5		47	18	973	14.59
10	1666	35	F	65	120	400	MULTISLIC	HELICAL	5	95	51	0.5		39	19	996	14.94
11	1668	44	M	71	120	150	MULTISLIC	HELICAL	5	93	51	0.5		37	7	385	5.77
12	1670	65	M	72	120	150	MULTISLIC	HELICAL	5	92	50	0.5		37	7	364	5.46
13	1673	74	F	66	120	190	MULTISLIC	HELICAL	5	78	43	0.5		38	9	390	5.85
14	1674	47	M	69	120	110	MULTISLIC	HELICAL	5	93	55	0.5		36	5	287	4.3
15	1678	47	M	64	120	220	MULTISLIC	HELICAL	5	103	56	0.5		36	9	495	7.42
16	1680	52	F	75	120	370	MULTISLIC	HELICAL	5	89	50	0.5		43	17	863	12.94
17	1682	29	F	68	120	150	MULTISLIC	HELICAL	5	95	52	0.5		36	7	375	5.62
18	1683	35	M	67	120	220	MULTISLIC	HELICAL	5	101	54	0.5		36	10	549	8.23
19	1688	32	F	77	120	130	MULTISLIC	HELICAL	5	98	53	0.5		36	6	315	4.72
20	1691	65	F	71	120	270	MULTISLIC	HELICAL	5	96	52	0.5		41	13	690	10.35
21	1693	38	F	69	120	150	MULTISLIC	HELICAL	5	89	48	0.5		36	7	335	5.02
22	1694	34	F	61	120	440	MULTISLIC	HELICAL	5	100	54	0.5		36	20	1088	16.32
23	1695	86	F	64	120	270	MULTISLIC	HELICAL	5	96	54	0.5		36	13	700	10.5
24	1699	46	F	78	120	350	MULTISLIC	HELICAL	5	105	57	0.5		42	16	930	13.95
25	1702	52	M	69	120	220	MULTISLIC	HELICAL	5	101	42	0.5		44	10	597	8.95
26	1703	71	F	64	120	250	MULTISLIC	HELICAL	5	84	48	0.5		49	12	609	9.13
27	1706	70	M	73	120	380	MULTISLIC	HELICAL	5	86	47	0.5		38	18	851	12.76
28	1710	60	M	75	120	380	MULTISLIC	HELICAL	5	112	60	0.5		39	18	1111	16.66
29	1712	58	M	70	120	400	MULTISLIC	HELICAL	5	97	53	0.5		36	19	1012	15.18

APPENDIX D: CENTRE B (BRAIN)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	SCAN TIME	FOV	CTDIvol	DLP	E(mSv)
1		65	M	74	120	200	MULTISLIC	AXIAL	2.5, 5	38	9	1	25	61	570	1.19
2		25	M	61	120	200	MULTISLIC	AXIAL	2.5, 5	38	9	1	25	64	596	1.25
3		37	M	78	120	200	MULTISLIC	AXIAL	2.5, 5	38	8	1	25	56	457	0.95
4		72	M	65	120	200	MULTISLIC	AXIAL	2.5, 5	36	9	1	25	62	576	1.2
5		85	M	62	120	200	MULTISLIC	AXIAL	2.5, 5	38	9	1	25	60	553	1.16
6		90	M	61	120	200	MULTISLIC	AXIAL	2.5, 5	36	9	1	25	65	603	1.26
7		64	F	79	120	200	MULTISLIC	AXIAL	2.5, 5	34	9	1	25	64	594	1.24
8		60	F	68	120	200	MULTISLIC	AXIAL	2.5, 5	38	10	1	25	64	647	0.77
9		47	F	64	120	200	MULTISLIC	AXIAL	2.5, 5	34	9	1	25	66	606	1.27
10		32	F	64	120	200	MULTISLIC	AXIAL	2.5, 5	38	8	1	25	56	457	0.95
11		32	M	70	120	200	MULTISLIC	AXIAL	2.5, 5	43	18	1	25	69	1242	2.6
12		50	M	67	120	200	MULTISLIC	AXIAL	2.5, 5	44	9	1	25	58	524	1.1
13		42	M	69	120	200	MULTISLIC	AXIAL	2.5, 5	38	9	1	25	67	606	1.27
14		32	M	75	120	200	MULTISLIC	AXIAL	2.5, 5	34	9	1	25	64	588	1.23
15		50	M	75	120	200	MULTISLIC	AXIAL	2.5, 5	34	9	1	25	63	575	1.2
16		80	M	73	120	200	MULTISLIC	AXIAL	2.5, 5	40	10	1	25	61	626	1.31
17		27	F	67	120	200	MULTISLIC	AXIAL	2.5, 5	38	9	1	25	62	571	1.19
18		60	F	63	120	200	MULTISLIC	AXIAL	2.5, 5	38	5	1	25	64	331	1.69
19		20	M	77	120	200	MULTISLIC	AXIAL	2.5, 5	40	7	1	25	59	431	0.9
20		20	F	70	120	200	MULTISLIC	AXIAL	2.5, 5	38	9	1	25	54	497	1.04

APPENDIX E: CENTRE B (CHEST)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	SCAN TIME	PITCH	FOV	CTDIvol	DLP	E(mSv)
1		50	M	71	120	100	MULTISLIC	HELICAL	5	64	34	1	0.75	50	11.68	399	5.58
2		43	M	79	120	100	MULTISLIC	HELICAL	5	75	35	1	0.75	50	23	805	11.27
3		25	F	74	120	100	MULTISLIC	HELICAL	5	56	33.6	1	0.75	50	9.2	310	4.34
4		37	F	72	120	100	MULTISLIC	HELICAL	5	60	35.3	1	0.75	50	10.53	372.4	5.2
5		61	M	66	120	100	MULTISLIC	HELICAL	5	64	35.5	1	0.75	50	11.02	390	5.46
6		55	M	78	120	100	MULTISLIC	HELICAL	5	82	36.9	1	0.75	50	15.73	650	9.1
7		72	M	77	120	100	MULTISLIC	HELICAL	5	74	34.2	1	0.75	50	14.85	506	7.08
8		49	F	69	120	100	MULTISLIC	HELICAL	5	68	34.8	1	0.75	50	12.1	417	5.8
9		66	M	69	120	100	MULTISLIC	HELICAL	5	62	34	1	0.75	50	11.7	380	5.32
10		45	F	71	120	100	MULTISLIC	HELICAL	5	60	35.2	1	0.75	50	10.9	383	5.34
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				APPENDIX F: CENTRE B (ABDOMEN)													
S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	SCAN TIME	PITCH	FOV	CTDIvol	DLP	E(mSv)
1		64	M	65	120	200	MULTISLIC	HELICAL	5	45	45	1	0.75	50	20	933	13.9
2		50	F	74	120	200	MULTISLIC	HELICAL	5	86	42	1	0.75	50	23	975	14.4
3		24	F	62	120	200	MULTISLIC	HELICAL	5	95	49	1	0.75	50	23	1149	17
4		16	M	60	120	200	MULTISLIC	HELICAL	5	81	42	1	0.75	50	10	418	6.27
5		78	F	66	120	200	MULTISLIC	HELICAL	5	80	42	1	0.75	50	23	1000	15
6		45	M	73	120	200	MULTISLIC	HELICAL	5	14	45	1	0.75	50	9	429	6.43
7		70	F	69	120	200	MULTISLIC	HELICAL	5	88	46	1	0.75	50	21	993	14.89
8		27	F	70	120	200	MULTISLIC	HELICAL	5	83	43	1	0.75	50	23	1010	15.15
9		56	M	77	120	200	MULTISLIC	HELICAL	5	65	43	1	0.75	50	11	485	7.2
10		40	M	75	120	200	MULTISLIC	HELICAL	5	71	44	1	0.75	50	13	528	7.92
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APPENDIX G: CENTRE C (BRAIN)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	TUBE ROT. TIME	SCAN LENGTH	FOV	CTDIvol	DLP	E(mSv)
1		70	M	80	110	130	MULTISLIC	AXIAL	6	1.5	20	25	33	658	1.38
2		54	M	75	110	130	MULTISLIC	AXIAL	6	1.5	18	25	36	658	1.38
3		19	F	60	110	130	MULTISLIC	AXIAL	6	1.5	16	25	25	320	0.67
4		56	F	80	110	130	MULTISLIC	AXIAL	6	1.5	16	25	36	592	1.24
5		66	F	75	110	140	MULTISLIC	AXIAL	6	1.5	16	25	26	409	0.85
6		50	F	80	110	130	MULTISLIC	AXIAL	6	1.5	14	25	36	527	1.1
7		56	M	68	110	130	MULTISLIC	AXIAL	6	1.5	16	25	36	592	1.24
8		60	M	78	110	130	MULTISLIC	AXIAL	6	1.5	17	25	36	658	1.38
9		43	F	70	110	130	MULTISLIC	AXIAL	6	1.5	16	25	36	592	1.24
10		48	M	75	110	130	MULTISLIC	AXIAL	6	1.5	17	25	34	542	1.13
11		48	M	80	110	130	MULTISLIC	AXIAL	6	1.5	16	25	32	592	1.24
12		46	M	75	110	130	MULTISLIC	AXIAL	6	1.5	16	25	36	527	1.1
13		64	M	80	110	130	MULTISLIC	AXIAL	6	1.5	14	25	36	658	1.38
14		39	F	70	110	130	MULTISLIC	AXIAL	6	1.5	18	25	36	537	1.12
15		73	M	80	110	130	MULTISLIC	AXIAL	6	1.5	14	25	36	527	1.1
16		60	F	80	110	130	MULTISLIC	AXIAL	6	1.5	14	25	36	527	1.1
17		55	F	65	110	130	MULTISLIC	AXIAL	6	1.5	14	25	36	592	1.24
18		29	M	65	110	130	MULTISLIC	AXIAL	6	1.5	16	25	36	527	1.1
19		37	F	80	110	130	MULTISLIC	AXIAL	6	1.5	14	25	36	525	1.1

APPENDIX H: CENTRE C (CHEST)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	TUBE ROT. TIME	SCAN LENGTH	PICTH	FOV	CTDIvol	DLP	E(mSv)
1		46	F	65	110	84	MULTISLIC	HELICAL	6	1	33	0.8333	29	5	186	2.6
2		41	F	80	110	99	MULTISLIC	HELICAL	10	1	31	0.8333	32	7	222	3.1
3		49	M	80	110	73	MULTISLIC	HELICAL	10	1	38	0.8333	34	6	232	3.24
4		50	M	75	110	58	MULTISLIC	HELICAL	10	1	37	0.8333	32	4	155	2.17
5		32	F	70	110	90	MULTISLIC	HELICAL	10	1	27	0.8333	32	6	174	2.43
6		55	M	80	110	79	MULTISLIC	HELICAL	10	1	33	0.8333	32	6	184	2.57
7		51	M	70	110	37	MULTISLIC	HELICAL	5	1	34	0.8333	30	3	106	1.48
8		38	F	75	110	50	MULTISLIC	HELICAL	5	1	30	0.8333	32	3	107	1.49
9		25	M	80	110	53	MULTISLIC	HELICAL	10	1	31	0.8333	32	4	120	1.68
10		44	F	75	110	68	MULTISLIC	HELICAL	10	1	30	0.8333	27	6	173	2.42
11		55	F	60	110	29	MULTISLIC	HELICAL	10	1	30	0.8333	28	2	72	1
12		57	M	75	110	60	MULTISLIC	HELICAL	10	1	33	0.8333	32	6	204	2.85
13		67	F	80	110	50	MULTISLIC	HELICAL	10	1	33	0.8333	32	3	118	1.65
14		63	M	80	110	61	MULTISLIC	HELICAL	10	1	33	0.8333	32	4	145	2.03
15		51	M	75	110	76	MULTISLIC	HELICAL	10	1	30	0.8333	27	6	193	2.7
16		67	M	70	110	91	MULTISLIC	HELICAL	10	1	33	0.8333	30	7	256	3.58
17		54	M	75	110	76	MULTISLIC	HELICAL	10	1	28	0.8333	25	6	132	1.84
18		62	M	80	110	42	MULTISLIC	HELICAL	10	1	38	0.8333	34	3	134	1.87
19		60	M	75	110	37	MULTISLIC	HELICAL	10	1	39	0.8333	17	3	103	1.44
20		80	M	80	110	83	MULTISLIC	HELICAL	10	1	37	0.8333	32	7	250	3.5

APPENDIX I: CENTRE C (ABDOMEN)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	TUBE ROTATION TIME	SCAN LENGTH	PITCH	FOV	CTDIvol	DLP	E(mSv)
1		43	M	75	110	97	MULTISLIC	HELICAL	10	1	48	1.5	17	7	260	3.9
2		26	F	65	110	88	MULTISLIC	HELICAL	10	1	49	1.5	22	6	310	4.65
3		18	F	75	110	114	MULTISLIC	HELICAL	10	1	39	1.5	17	8	321	4.81
4		32	F	70	110	91	MULTISLIC	HELICAL	10	1	47	1.5	21	6	310	4.65
5		43	F	75	110	126	MULTISLIC	HELICAL	10	1	37	1.5	17	9	347	5.2
6		36	F	60	110	95	MULTISLIC	HELICAL	10	1	49	1.5	21	7	335	5.02
7		46	F	80	110	73	MULTISLIC	HELICAL	10	1	42	1.5	33	8	338	5.07
8		56	M	80	110	93	MULTISLIC	HELICAL	10	1	58	0.85	44	10	588	8.82
9		48	M	75	110	147	MULTISLIC	HELICAL	10	1	50	0.85	39	16	803	12.04
10		32	F	70	110	124	MULTISLIC	HELICAL	10	1	46	0.85	35	13	622	9.33
11		31	F	75	110	58	MULTISLIC	HELICAL	10	1	41	0.85	32	6	266	3.99
12		26	M	80	110	71	MULTISLIC	HELICAL	10	1	42	0.85	33	8	331	4.96
13		38	M	75	110	60	MULTISLIC	HELICAL	10	1	48	0.85	37	6	317	4.75
14		65	M	75	110	73	MULTISLIC	HELICAL	10	1	42	1.5	18	5	218	3.27
15		36	F	80	110	75	MULTISLIC	HELICAL	10	1	52	0.85	40	8	462	6.93
16		36	F	60	110	50	MULTISLIC	HELICAL	10	1	48	0.85	37	5	264	3.96
17		53	M	75	110	66	MULTISLIC	HELICAL	10	1	47	1.5	20	4	220	3.3
18		23	F	70	110	49	MULTISLIC	HELICAL	10	1	42	0.85	33	5	230	3.45
19		30	M	75	110	56	MULTISLIC	HELICAL	10	1	41	0.85	33	6	254	3.81
20		31	F	80	110	104	MULTISLIC	HELICAL	10	1	47	0.85	37	12	583	8.74

APPENDIX J: CENTRE D (BRAIN)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	SCAN TIME	FOV	CTDIvol	DLP	E(mSv)
1	5245	34	F	71	120	401	MULTISLIC	AXIAL	5	34	18.3	1	23.6	60.9	1118	2.34
2	5239	68	F	74	120	401	MULTISLIC	AXIAL	5	37	19.8	1	23.3	60.9	1210	2.5
3	5281	32	F	71	120	401	MULTISLIC	AXIAL	5	32	19.8	1	18.8	43.3	854	1.79
4	5234	43	M	72	120	401	MULTISLIC	AXIAL	5	36	19.3	1	20.3	60.9	1179	2.47
5	5232	30	M	69	120	401	MULTISLIC	AXIAL	5	35	18.8	1	19.9	60.9	1149	2.41
6	5230	82	M	67	120	401	MULTISLIC	AXIAL	5	32	17.3	1	20.7	60.9	1058	2.22
7	5299	25	M	73	120	401	MULTISLIC	AXIAL	5	39	20.8	1	20.2	60.9	1271	2.66
8	5225	30	M	65	120	401	MULTISLIC	AXIAL	5	38	20.3	1	20.1	60.9	1240	2.6
9	5222	24	M	65	120	401	MULTISLIC	AXIAL	5	33	17.8	1	19.9	60.9	1085	2.27
10	5221	77	F	74	120	401	MULTISLIC	AXIAL	5	35	18.1	1	20.1	60.9	1103	2.31
11	5214	52	M	70	120	401	MULTISLIC	AXIAL	5	40	21.3	1	20.5	60.9	1301	2.73
12	5212	42	F	69	120	401	MULTISLIC	AXIAL	5	35	18.8	1	19.9	60.9	1149	2.41
13	5209	30	M	76	120	401	MULTISLIC	AXIAL	5	62	18.8	1	21.2	60.9	1968	4.132
14	5223	27	M	68	120	401	MULTISLIC	AXIAL	5	34	18.3	1	20.4	60.9	1118	2.34
15	5221	54	M	76	120	401	MULTISLIC	AXIAL	5	33	17.8	1	20.3	60.9	1089	2.28
16	5219	28	F	77	120	401	MULTISLIC	AXIAL	5	61	31.8	1	25.9	60.9	1938	4.06
17	5218	38	M	64	120	401	MULTISLIC	AXIAL	5	118	25.4	1	23.7	60.9	1545	3.24
18	5216	62	F	68	120	401	MULTISLIC	AXIAL	5	30	16.3	1	20	60.9	997	2.09
19	5217	32	M	70	90	250	MULTISLIC	AXIAL	3	58	16.1	1	20	17.1	532	1.11
20	5215	58	M	71	120	401	MULTISLIC	AXIAL	5	33	17.8	1	21.2	60.9	1088	2.28

APPENDIX K: CENTRE D (CHEST)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	SCAN TIME	PITCH	FOV	CTDIvol	DLP	E(mSv)
1	5151	44	M	71	120	150	MULTISLIC	HELICAL	3	211	37.9	1	0.9375	30.5	10	379	5.306
2	5148	26	M	75	120	150	MULTISLIC	HELICAL	5	65	38.5	1	0.9375	30.2	10	385	5.39
3	5131	44	F	68	120	150	MULTISLIC	HELICAL	5	63	37.3	1	0.9375	30.9	10	373	5.22
4	5118	54	M	70	120	150	MULTISLIC	HELICAL	5	71	41.6	1	0.9375	30.8	10	416	5.82
5	5109	55	F	70	120	150	MULTISLIC	HELICAL	5	65	38.4	1	0.9375	31	10	384	5.37
6	5270	40	F	69	120	150	MULTISLIC	HELICAL	5	66	39.1	1	0.9375	27.2	10	391	5.47
7	5269	47	M	70	120	150	MULTISLIC	HELICAL	5	77	44.7	1	0.9375	32.7	10	447	6.25
8	5214	35	F	71	120	150	MULTISLIC	HELICAL	5	65	38.4	1	0.9375	33.6	10	384	5.37
9	5210	34	F	67	120	150	MULTISLIC	HELICAL	5	55	33.1	1	0.9375	37.7	10	331	4.63
10	5109	54	F	66	120	150	MULTISLIC	HELICAL	5	70	41	1	0.9375	31	10	410	5.74
11	5107	60	F	74	120	150	MULTISLIC	HELICAL	5	73	42.6	1	0.9375	35	10	426	5.9
12	5552	57	M	67	120	150	MULTISLIC	HELICAL	3	277	46.1	1	0.9375	35	10	461	6.45
13	1690	41	M	72	120	150	MULTISLIC	HELICAL	5	70	41	1	0.9375	31.6	10	410	5.74
14	5601	50	M	73	120	150	MULTISLIC	HELICAL	5	75	37.7	1	0.9375	31.2	10	377	5.2
15	5591	67	M	70	120	150	MULTISLIC	HELICAL	5	72	41.7	1	0.9375	32	10	417	5.83
16	5563	62	M	66	120	150	MULTISLIC	HELICAL	5	80	44.7	1	0.9375	30.9	10	447	6.25
17	5544	52	F	68	120	150	MULTISLIC	HELICAL	5	74	41.6	1	0.9375	29.9	10	416	5.82
18	5516	54	F	66	120	150	MULTISLIC	HELICAL	5	78	45.3	1	0.9375	25	10	453	6.34
19	5086	67	F	76	120	150	MULTISLIC	HELICAL	5	67	39.4	1	0.9375	25	10	394	5.51
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APPENDIX L: CENTRE D (ABDOMEN)

S/N	CT NUMBER	AGE	SEX	WEIGHT (KG)	kV	mA	SCANNER	SCAN MODE	SLICE THICKNESS	NUMBER OF SLICES	SCAN LENGTH	SCAN TIME	PITCH	FOV	CTDIvol	DLP	E(mSv)
1	5219	51	M	67	120	220	MULTISLIC	HELICAL	5	73	40.5	1	0.9375	29.8	15	628	9.34
2	5184	65	F	71	120	220	MULTISLIC	HELICAL	5	80	45	1	0.9375	34.3	15	677	10.15
3	5213	55	M	72	120	220	MULTISLIC	HELICAL	5	71	40.6	1	0.9375	30	15	610	9.15
4	5211	45	M	70	120	220	MULTISLIC	HELICAL	5	75	43.7	1	0.9375	30	15	656	9.84
5	5208	35	M	75	120	220	MULTISLIC	HELICAL	5	75	42.7	1	0.9375	35.9	15	641	9.6
6	5252	34	F	69	120	220	MULTISLIC	HELICAL	5	77	42.7	1	0.9375	26.9	15	652	9.78
7	5200	46	M	72	120	220	MULTISLIC	HELICAL	5	79	44.8	1	0.9375	29.8	15	672	10.08
8	5188	65	F	68	120	220	MULTISLIC	HELICAL	5	79	49.5	1	0.9375	27.6	15	743	11.14
9	5209	28	F	76	120	220	MULTISLIC	HELICAL	5	93	50.4	1	0.9375	35	15	757	11.13
10	5149	94	M	70	120	220	MULTISLIC	HELICAL	5	82	34.6	1	0.9375	33.8	15	519	7.78
11	5143	38	F	70	120	220	MULTISLIC	HELICAL	5	90	49.5	1	0.9375	35	15	743	11.14
12	5118	38	F	66	120	220	MULTISLIC	HELICAL	5	90	50.4	1	0.9375	36.6	15	757	11.13
13	5111	32	F	73	120	220	MULTISLIC	HELICAL	5	88	49.4	1	0.9375	36.6	15	741	11.11
14	5112	64	F	67	120	220	MULTISLIC	HELICAL	5	87	48.9	1	0.9375	29.9	15	734	11.01
15	5104	67	F	65	120	220	MULTISLIC	HELICAL	5	77	43.7	1	0.9375	34.8	15	656	9.84
16	5097	24	M	70	120	220	MULTISLIC	HELICAL	5	100	55	1	0.9375	35	15	835	12.52
17	5090	54	M	70	120	220	MULTISLIC	HELICAL	5	103	55.6	1	0.9375	35.7		858	12.87
18	5262	48	F	72	120	220	MULTISLIC	HELICAL	5	67	24.1	1	0.9375	35.3	15.5	374	5.61
19	5080	64	M	75	120	220	MULTISLIC	HELICAL	5	91	50.2	1	0.9375	35	15.2	765	11.47
20	5262	50	M	78	120	220	MULTISLIC	HELICAL	5	87	42.2	1	0.9375	32.9	15	634	9.51