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Estimation of Radiation Dose to Eye Lens for the Patients Undergoing Scan of Brain by Computed Tomography

تقدير الجرعة الإشعاعية لعدسة العين للمرضى الذين يخضعون لتصوير الدماغ عن طريق التصوير المقطعي

A thesis submitted in partial fulfillment for the requirements

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الآيـة

قال تعالى : ((أَوَلَيْسَ الَّذِي خَلَقَ السَّمَاوَاتِ وَالْأَرْضَ بِقَادِرٍ

عَلَى أَنْ يَخْلُقَ مِثْلَهُم بَلَى وَهُوَ الْخَلَّاقُ الْعَلِيمُ {81} إِنَّمَا أَمْرُهُ

إِذَا أَرَادَ شَيْئًا أَنْ يَقُولَ لَهُ كُنْ فَيَكُون {82} فَسُبْحَانَ الَّذِي بِيَدِهِ

مَلَكُوتُ كُلِّ شَيْءٍ وَإِلَيْهِ تُرْجَعُونَ {83}))

صدق الله العظيم

سورة يـــس

Dedication

To my family …

To my friends …

And to all the medical physicists in the world…

Acknowledgements

First and foremost , I would like to express my deepest gratitude to Dr. Mohamed Elfadil , without his help this work could not have been accomplished

I also would like to thanks Dr. Suhaib Alameen for his support and helping.

Deep thanks to my family for their consistent mental support finally, I would like to thanks my friend.

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List of abbreviations:

RCIH Royal Care International Hospital

ANH: Alamal National Hospital

ZSH: Zayatona Specialist Hospital

CT: Computed Tomography

E: Effective Dose

CTDI: Computed Tomography Dose Index

CTDIvol: Computed Tomography Dose Index Volume

CTDIW: Computed Tomography Dose Index Weighted

DLP: Dose Length Product

ICRP: International Commission on Radiological Protection

NRPB: National Council of Radiation Protection

CTU: Computed Tomography Urography

EC: European Commission

FBP: Filter Back Projection

PTFE: Poly tetra fluoro ethylene

SSDE: Size-Specific Dose Estimates

K: Conversion Coefficients Factor

APMM: American Association of Physicists in Medicine

Abstract:

The aim of this study to estimation the radiation dose for eye lens for patients underwent the scan head by CT model of Toshiba 64 slices in the (***RCIH, ANH and ZSH).***

The data collected from 51patient in three hospitals, Data were used to estimation and comparing the Equivalent dose of eye lens from different diagnostic procedures in three hospitals with the other study from deferent country. was used CT-Expo software Version 2.5 for dose calculations, the effective dose between the all hospital in this study, the small value of Effective dose in RCIH (3.92 mSv) then ANH (4.64 mSv) and higher dose in ZSH (5.34 mSv) and higher than the study in Tanzania (2.1 mSv), Australia (3.61 mSv) and UK (1.5 mSv), the Equivalent dose of eye lens is small dose in RCIH (82.97 mSv), then ANH (88.61 mSv) and higher dose in ZSH (97.01 mSv),

The mean $CTDI_{VOL}$ values in hospitals is higher than similar study in other country example as (UK (55-65 mGy), Sudan (65.4 mGy) and Tanzania (43 mGy)), except the Australia (86.7 mGy), DLP values was found in this study were higher (1547.5 mGy*cm) than those presented in Tanzania (913 mGy*cm), Australia (1508.4 mGy*cm) and UK (760-930 mGy*cm).

الخلاصة :

الهدف من هذه الدراسه تقييم الجرعه الاشعاعيه لعدسة العين لعدد 51 مريض اثناء تصوير الراس بالاشعة المقطعية بواسطة جهاز الاشعة المقطعية ذو 64 شريحة في كل من مستشفى الامل الوطني ومستشفى الزيتونة التخصصي ومستشفى رويال كير العالمية .

تم جمع البيانات من الثلاثة مستشفيات المذكورة ، و استخدمت البيانات لتقييم جرعة عدسة العين حسب البروتوكول المتبع في كل من المستشفيات الثلاثة ، وقد تم استخدام برنامج (CT-Expo) إصدار 2.5 لحساب الجرعة المكافئة والجرعة الفعالة في هذه الدراسة، وكانت الجرعة الفعالة في هذه الدراسة صغيرة في RCIH (3.92 ملي سيفرت) ثم في ANH (4.64 ملي سيفرت) و عالية في ZSH (5.34 ملي سيفرت) ومتوسطها (4.649 ملي سيفرت) وهي عالية مقارنة بدراسات سابقة في كل من تنزانيا (2.1 ملي سيفرت) و المملكة المتحدة (1.5 ملي سيفرت) واستراليا (3.61 ملي سيفرت) ، والجرعة المكافئة لعدسة العين كانت صغيرة في RCIH (82.97 ملي سيفرت)، ثم في ANH (88.61 ملي سيفرت) وعالية في ZSH (97.01 ملي سيفرت)

متوسط قيم $CTDI_{VOL}$ (82.812 ملي غري) كانت أعلى من دراسات مماثلة أجريت في دول أخرى مثل (المملكة المتحدة (55-65 ملي غري) والسودان (65.4 ملي غري) وتنزانيا(43 ملي غري)) باستثناء أستراليا (86.7 ملي غري) ، و كانت قيم ال DLP في هذه الدراسة هي (1547.5 ملي غري في السنتيميتر) كانت أعلى من تلك التي عرضت في تنزانيا (913 ملي غري في السنتيميتر) و استراليا (1508.4 ملي غري في السنتيميتر) والمملكة المتحدة (760-930 ملي غري في السنتيميتر) .

Chapter One

Introduction

1.1. An overview:

Computed Tomography (CT) is a radiologic modality that provides clinical information in the detection, differentiation, and demarcation of disease. It is the primary diagnostic modality for a variety of presenting problems and is widely accepted as a supplement to other imaging techniques. CT is a form of medical imaging that involves the exposure of patients to ionizing radiation[P. This,lta,2014]. During a CT scan a rotating source passes x-rays through a patient's body to produce several cross-sectional images of a particular area. These two-dimensional images can also be digitally combined to produce a single three-dimensional [Ali,2015].

Computed tomography (CT) developed from an x ray modality that was limited to axial imaging of the brain in neuroradiology into a versatile 3-D whole body imaging modality for a wide range of applications, including oncology, vascular radiology, cardiology, traumatology and interventional radiology. CT is applied for diagnosis and follow-up studies of patients, for planning of radiotherapy, and even for screening of healthy subpopulations with specific risk factors [Dance, et al].

1.2. Problem of Study:

Using CT scanning, patients are exposed to more doses which may result in unintended health effects, During brain CT scans, the lens is irradiated indirectly and/or directly. However, it is difficult to estimate the lens dose during brain CT in individual cases, because the dose varies considerably due to differences in the type of CT scanners and scan settings; the CTDI can be used theoretically to estimate the average dose from multiple scans with table increments. to avoid unnecessary of high dose to the patient need to estimate the equivalent dose.

1.3. Objective:

1.3.1. General Objective:

To estimated equivalent dose in patient CT examination for Toshiba x64 slice using CT. Expo software version 2.5.

1.3.2. Specific Objective:

- To estimate volume Computed Tomography Dose Index ($CTDI_{vol}$).
- To measure Dose Length Product (DLP).
- To estimate the equivalent dose received by the lens of the eyes in brain scan by CT of each projection.
- To estimate the effective dose received by the lens in CT brain examinations.
- Compare the present study with the International Recommendations (DRLs) of Lens Dose.

1.4. Thesis outlines:

Chapter one: This chapter is general introduction to the computed tomography and represents the goal of calculate patient dose. The published literature and studies done on the research subject were reviewed in this chapter to know about bases and methods of assessing the patient dose. The objectives of this study were also mentioned in this chapter.

Chapter two: This chapter explores the computed tomography, hardware of CT, dosimetric quantities and units, quantities related to stochastic and deterministic effect.

Chapter three: This chapter describes the materials and methods used in this research to assess the Equivalent dose HT.

Chapter four: This chapter consists of: presentation of the results in tables and discussion of the results.

Chapter five: Introduce the conclusion that had been derived out from the research.

Chapter Two

Theoretical Background

2.1. Benefits of CT:

CT is an important and sometimes life-saving tool for diagnostic medical examinations and guidance of interventional and therapeutic procedures. It allows rapid acquisition of high-resolution three-dimensional images, providing radiologists and other physicians with cross-sectional views of the patient's anatomy. CT can be used to image many types of tissues, such as soft tissues, bones, lungs, and blood vessels. CT examinations are also non-invasive, although a contrast agent is sometimes administered to the patient. As a consequence of the benefits of CT examinations, it has become the gold standard for a variety of clinical indications, such as diagnosing certain cancers, surgical planning, and identifying internal injuries and bleeding in trauma cases[Human, et al, 2006].

Diagnostic importance of CT examinations is outstanding, so the increase of examination frequency is justified. According to the International Commission on Radiological Protection (ICRP) dose limits should not be applied for medical exposures either diagnostic or therapy, because patients have direct benefit from the exposure. However according to the basic principles of radiation protection the medical diagnostic procedures should be optimized and unjustified exposures should be minimized [Ali, 2015]

CT procedures give patients more radiation dose than traditional x-ray imaging modalities. Patients are exposed to more dose which may result in unintended health effects health care providers need to be able to estimate and track the dose these patients Receive from their CT scan [Prins, et al, 2011].

2.2. System of radiation protection:

Radiation protection the principles of radiation protection and safety are those developed by the International Commission on Radiological Protection (ICRP). The principles of justification and optimization apply in all three exposure situations (occupational exposure, medical exposure, and public exposure) whereas the principle of application of dose limits applies only for doses expected to be incurred with certainty as a result of planned exposure other than medical exposure. These principles are defined as follows:

- Justification: Any decision that alters the radiation exposure situation should do more good than harm.
- Optimization of Protection: The likelihood of incurring exposure, the number of people exposed, and the magnitude of their individual doses should all be kept as low as reasonably achievable, taking into account economic and societal factors.
- Dose Limits: The total dose to any individual from regulated sources in planned exposure situations other than medical exposure of patients should not exceed the appropriate limits specified by the ICRP [ICRP 103, 2007].

2.3. Application of safety principles in diagnostic radiology:

Justification of practices: By weighing the diagnostic benefits they produce a against the radiation detriment they might cause, taking into account the benefits and risks of available alternative techniques that do not involve medical radiation exposure.

Optimization of protection and safety: In diagnostic medical exposure, keeping the exposure of patients to the minimum necessary to achieve the required diagnostic objective, taking into account norms of acceptable image quality established by appropriate professional bodies and relevant guidance levels for medical exposure. Dose limits is not applicable in overall medical exposure [Ntly, et al, 2006].

2.4. Geometry and Historical Development:

Computed tomography (CT) is in its fourth decade of clinical use and has proved invaluable as a diagnostic tool for many clinical applications, from cancer diagnosis to trauma to osteoporosis screening. CT was the first imaging modality that made it possible to probe the inner depths of the body, slice by slice. Since 1972, when the first head CT scanner was introduced, CT has matured greatly and gained technological sophistication. Concomitant changes have occurred in the quality of   CT Images. The first CT scanner, an EMI Mark 1, produced images with 80 X 80 pixel resolution (3-mm pixels), and each pair of slices required approximately 4.5 minutes of scan time and 1.5 minutes of reconstruction time [Jerrold,et al, 2002].

2.4.1. First Generation: Rotate/translate, Pencil beam:

The first generation of CT scanners employed a rotate translate, pencil beam system. Only two x-ray detectors were used, and they measured the transmission of x-rays through the patient for two different slices [Jerrold,et al,2002].

2.4.2. Second Generation: Rotate/translate, Narrow Fan Beam:

The next incremental improvement to the CT scanner was the incorporation of a linear array of 30 detectors. This increased the utilization of the x-ray beam by 30 times, compared with the single detector used per slice in first-generation systems. A relatively narrow fan angle of 10 degrees was used. In principle, a reduction in scan time of about 30-fold could be expected. However, this reduction time was not realized, because more data were acquired to improve image quality the shortest scan time with a second-generation scanner was 18 seconds per slice, 15 times faster than with the first-generation system [Jerrold,et al,2002].

2.4.3. Third Generation: Rotate/Rotate, Wide Fan Beam:

The translational motion of first- and second-generation CT scanners was a fundamental impediment to fast scanning. At the end of each translation, the motion of the x-ray tube/detector system had to be stopped, the whole system rotated, and the translational motion restarted. The success of CT as a clinical modality in its infancy gave manufacturers reason to explore more efficient, but more costly approaches to the scanning geometry the motion of third-generation CT is "Rotate/Rotate" referring to the rotation of the x-ray tube and the rotation of the detector array. By elimination of the translational motion the scan time is reduced substantially. The early third-generation scanners could deliver scan times shorter than 5 seconds. Newer systems have scan times of one half second. The evolution from first- to second- and second- to third-generation scanners involved radical improvement with each step. Developments of the fourth- and fifth-generation scanners led not only to some improvements compromises in clinical CT images, compared third-generation scanners but also to some indeed, rotate/rotate canners are still as viable today as they were when they were introduced in 1975. The features of third- and fourth-generation CT should be compared by the reader, because each offers some benefits but also some tradeoffs [Jerrold,et al,2002].

2.4.4. Fourth Generation (Rotate/Stationary):

Third-generation scanners suffered from the significant problem of ring artifacts and in the late 1970s fourth-generation scanners were designed specifically to address these artifacts [Jerrold,et al,2002].

2.4.5. Fifth Generation (Stationary/Stationary):

A novel CT scanner has been developed specifically for cardiac tomographic imaging. This "cine-CT" scanner does not use a conventional x-ray tube; instead, a large arc of tungsten encircles the patient and He's directly opposite to the detector ring [Jerrold,et al,2002].

2.4.6. Sixth Generation: Helical:

Helical CT scanner acquire data while the table is moving; as a result, the x-ray source moves in a helical pattern around the patient being scanned Helical CT scanners use either third- or fourth-generation slip-ring Designs. By avoiding the time required to translate the patient table, the total scan time required to image the patient can be much shorter consequently, helical scanning allows the use of less contrast agent and increases patient throughput.

The advent of helical scanning has introduced many different considerations for data acquisition. In order to produce reconstructions of planar sections of the patient, the raw data from the helical data seared interpolated to approximate the acquisition of planar reconstruction data the speed of the table motion relative to the rotation of the CT gantry is a very important consideration and the pitch is the parameter that describes this relationship[J. M. B. Jerrold,lta2002].

2.4.7. Seven Generation: Multiple Detector Array:

X-ray tubes designed for CT have impressive heat storage and cooling capabilities although the instantaneous production of x-rays is constrained by the physics governing x-ray production. An approach to overcoming x-ray tube output limitations is to make better use of the x-rays that are produced by the x-ray tube. When multiple detector arrays are used the collimator spacing is wider and therefore more of the x-rays that are produced by the x-ray tube are used in producing image data. With conventional, single detector array scanners, opening up the collimator increases the slice thickness which is good for improving the utilization of the x-ray beam but reduces spatial resolution in the slice thickness dimension. With the introduction of multiple detector arrays, the slice thickness is determined by the detector size and not by the collimator. This represents a major shift in CT technology. The flexibility of CT acquisition protocols and increased efficiency resulting from multiple detector array CT scanners allows better patient imaging; however, the number of parameters involved in the CT acquisition protocol is increased as well. Also with multiple detector arrays, the notion of helical pitch needs to be redefined [Jerrold,et al,2002].

2.5. Gantry and table:

The gantry contains all the system components that are required to record transmission profiles of the patient. Since transmission profiles have to be recorded at different angles, these components are mounted on a support within the gantry that can be rotated. The x- ray tube with high voltage generator and tube cooling system, the collimator, the beam shaping filters, the detector arc and the data acquisition system are all mounted on this support. The engineering of these components is complex, since they need to be able to withstand the strong centrifugal force that occurs during the fast rotation of the gantry. Forces of several tens of g arise for rotation times of the order of 0.25 s. electrical power is generally supplied to the rotating gantry by means of slip ring contacts. Recorded projection profiles are generally transmitted from the gantry to a computer by means of wireless communication technologies. The design and engineering of the table, as with the gantry, are critical to allowing accurate acquisition of data at high rotational speeds. The table must also be able to withstand heavy weights without bending. The position of the patient on the table can be head first or feet first, and supine or prone; this position is usually recorded with the scan data [Dance,et al].

2.5.1. The X- ray tube and generator:

Owing to the high X - ray flux required for CT, the X ray tube uses a tungsten anode designed to withstand and dissipate high heat loads. With long continuous acquisition cycles, a forced cooling system using oil or water circulated through a heat exchanger is often used [Dance,et al].

2.5.2. Collimation and filtration:

After transmission through the patient, the x-ray beam is collimated to confine the transmission measurement to a slice with a thickness of a few millimeters. Collimation also serves to reduce scattered radiation to less than 1% of the primary beam intensity.

The height of the collimator defines the thickness of the CT slice. This height, when combined with the area of a single picture element (pixel) in the display, defines the three-dimensional volume element (voxel) in the patient corresponding to the two-dimensional pixel of the display. A voxel encompassing a boundary between two tissue structures (e.g., muscle and bone) yields an attenuation coefficient for the pixel that is intermediate between the values for the two structures. This "partial-volume artifact" may be reduced by narrowing the collimator to yield thinner slices. However, this approach reduces the number of x rays incident upon the detector. With fewer x rays interacting in the detector, the resulting signals are subject to greater statistical fluctuation and yield a noisier image in the final display[F. Edition].

2.5.3. Detectors:

The essential physical characteristics of CT detectors are a good detection efficiency and a fast response with little afterglow. Currently, solid state detectors are used, as they have detection efficiency close to 100% compared with high pressure, xenon filled ionization chambers that were used previously and that had a detection efficiency of about 70%. Solid state detectors are generally scintillators, meaning that the X rays interacting with the detector generate light. This light is converted to an electrical signal, by photodiodes that are attached to the back of the scintillator, which should have good transparency to ensure optimal detection. Typically, an ant scatter grid is mounted at the front of the detector, which consists of small strips of highly attenuating material (e.g. tungsten) aligned along the longitudinal (z) axis of the CT scanner. Detector row consists of thousands of dells that are separated by septa designed to prevent light generated in one Del from being detected by neighboring dells. These septa and the strips of the ant scatter grid should be as small as possible since they reduce the effective area of the detector and thus reduce the detection of x- rays [Dance,et al].

2.6. Dosimetric Quantities and Units:

2.6.1. Basic Dosimetric Quantities:

2.6.1.1. Particle Number N:

The particle Number N is the number of particles that are emitted transferred, or received Unit: 1[IAEA Dosimetry].

2.6.1.2. Radiant Energy R:

The Radiant Energy R is the energy (excluding rest energy) of particles that are emitted, transferred, or received.

Unit: j [ICRU Report 51].

2.6.1.3. Fluence Φ:

The Fluence ϕ, is the quotient dN by dA, where dN is the number of particles incident on of cross-sectional area a sphere.

$$\Phi = \frac{dN}{dA} \dots \dots \dots \dots \dots \dots \dots \dots .. 1.2$$

The unit of particle Fluence is m^{-2} [Podgorsak].

2.6.1.4. Energy fluence ψ:

The Energy Fluence ψ, is the quotient of d E by dA, where d E is the radiant energy incident on a sphere of cross-sectional area dA [Podgorsak] .

$$\Psi = \frac{dE}{dA} \dots \dots \dots \dots \dots \dots .. 2.2.$$

2.6.1.5. Kerma K:

The Kerma K is an acronym for kinetic energy released per unit mass. Kerma is defined as the mean energy transferred from the indirectly ionizing radiation to charged particles (electrons) in the medium $d\bar{E}$ per unit mass dm:

$$K = \frac{d\bar{E}}{dm} \dots \dots \dots \dots \dots .. 3.2$$

The unit of kerma is joule per kilogram (J/kg). The name for the unit of kerma is the gray (Gy), where 1 Gy = 1 J/kg [Podgorsak] .

2.6.1.6 Energy imparted:

The mean energy imparted to the matter in volume equals the radiant energy, R_{in} of all those charged and uncharged ionizing particles which enter the volume minus the radiant energy,

R_{out} of all those charged And uncharged ionizing particles, which leave the volume plus the sum ΣQ, of all changes of the rest energy of nuclei and elementary particles which occur in the volume, thus[IAEA Dosimetry]:

$$E = \sum R_{in} - R_{out} + Q \ldots\ldots\ldots\ldots .4.2.$$

2.6.1.7. Absorbed Dose D:

The Absorbed Dose D, is the quotient of dĒ by dm, where dĒ is the mean energy imparted by ionizing radiation to matter of mass dm thus

$$D = \frac{d\bar{E}}{dm} \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots .5.2$$

Unit: J/ Kg

The special name for the unit of absorbed dose is gray (Gy) [ICRU Report 51].

2.6.2. Quantities for CT dosimetry:

2.6.2.1. Computed Tomography Dose Index CTDI:

The CTDI is the primary dose measurement concept in CT, Where

$$CTDI = \frac{1}{N} \int_{-\infty}^{\infty} D\,(z)dz \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots ..6.2$$

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

Where: N is the number of tomographic section imaged in a single axial. This is equal to the number of data channels used in a particular scan T= the width of the tomographic section along the z-axis imaged by one data channel. In multiple-detector-row (multiline) CT scanner, several detector elements may be grouped together to form one data channel. In single-detector-row (single-slice) CT, the z-axis collimation (T) is the nominal scan width CTDI represents the average absorbed dose, along the z-axis forma series of contiguous irradiations. It is measured form one axial CT scan (one rotation of the x-ray tube), and is calculated by dividing the integrated absorbed dose by the nominal total beam collimation. The CTDI is always measured in the axial scan mode for a single rotation of the x-ray source, and theoretically estimates the average dose within the central region of scan volume consisting of multiple, contiguous CT scans [Multiple Scan Average Dose (MSAD)] for the case where the scan length is sufficient for the central dose to approach its asymptotic upper limit. The MSAD represents the average dose over a small interval (-1/2, 1/2) about center of the scan length (z=0) for scan interval 1, but requires multiple exposure for its direct measurement. The CTDI offered a more convenient yet nominally equivalent method of estimating this value, and required only a single-scan acquisition, which in the early days of CT,saved a consider- able amount of time[Beck,1993].

2.6.2.2. $CTDI_{FAD}$:

Theoretically, the equivalence of the MSAD and the CTDI requires that all contributions from the tails of the radiation dose profile be included in the CTDI dose measurement. The exact integration limits required to meet this criterion depend upon the width of the nominal radiation beam and the scattering medium. To standardize CTDI measurements (infinity is not a likely measurement parameter), the FDA introduced the integration limits of ±7T, where to represent the nominal slice width Interestingly, the original CT scanner, the EMI Mark I, was a dual detector -row system.Hence, the nominal radiation beam width was equal to twice the nominal slice width (i.e., NxTmm).To account for this, the CTDI value must be normalized to 1/NT:

As described in equ below [Beck,1993].

$$CTDI_{FAD} = \frac{1}{NT}\int_{-7T}^{7T} D(z)dz \dots\dots\dots\dots\dots\dots\dots\dots\dots 7.2.$$

2.6.2.3. $CTDI_{100}$:

CTD_{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 $CTDI_{100}$, like the $CTDI_{FAD}$ requires integration of the radiation dose profile froma 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 as described in equ below [Beck,1993].

$$CTDI_{100} = \frac{1}{NT}\int_{-50}^{50} D(z)dz \ldots\ldots\ldots\ldots\ldots .8.2.$$

2.6.2.4. Weighted$CTDI_W$:

The CTDI varies across the field of view (FOV). For example, for body CTimaging, the CTDI is typically a factor or two higher at the surface than at the center of the FOV. The average CTDI across the FOV is estimated by the Weighted CTDI ($CTDI_W$),

Where

$$CTDI_W = {}^{1}/_{3}\, CTDI_{100,center} + {}^{2}/_{3}\, CTDI_{100,edge}$$

The values of 1/3 and 2/3 approximate the relative areas represented by the center and edge values. CTD_W is a useful indicator of scanner radiation output for a specific Kv_p and mAs [Beck,1993].

2.6.2.5. Volume $CTDI_{VOL}$:

To represent dose for a specific scan protocol, which almost always involves series of scans, it is essential to take into account any gaps or overlaps between the x-ray beams from consecutive rotations of the X-ray source. This is accomplished with use of a dose descriptor known as the Volume $CTDI_W$ ($CTDI_{VOL}$),

Where

$$CTDI_{VOL} = \frac{NT}{I} \times CTDI_W \ldots\ldots..9.2$$

Where: I = the table increment per axial scan (mm) Since the pitch is define as the ratio of the table travel per rotation (I) to the total nominal beam width (N×T)

$$pitch = \frac{I}{NT} \ldots\ldots\ldots\ldots\ldots\ldots 10.2$$

Thus, the volume CTDI can expressed as

$$\mathrm{C}CTDI_{vol} = \frac{1}{pitch} \times CTDI_W \ldots\ldots\ldots.11.2$$

Whereas $CTDI_W$ 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, $CTDI_{vol}$ represents the average absorbed radiation dose over the *x*, *y*, and *z* directions. 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. The SI units are milligray (mGy)[Beck,1993].

2.6.2.6. 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

$$DLP \text{ (mGy-cm)} = CTDI\text{vol (mGy) x } scan\ length \text{ (cm)}.........12.2$$

The DLP reflects the total energy absorbed (and thus the potential biological effect) attributable to the complete scan acquisition. Thus, an abdomen-only CT exam might have the same $CTDI_{vol}$ as an abdomen/pelvis CT exam, but the latter exam would have a greater DLP, proportional to the greater *z*-extent of the scan volume[Beck,1993].

2.6.3. Quantities Related To Stochastic and Deterministic Effect:

2.6.3.1. Organ and Tissue Dose D_T:

The mean absorbed dose in a specified tissue or organ. It is equal to the ratio of the energy imparted, $\bar{E}_T$, to the tissue or organ to the mass, M_T, of the tissue or organ, thus

$$D_T = \frac{E_T}{m_T} \ldots\ldots\ldots\ldots 13.2$$

The mean absorbed dose in a specified tissue or organ is sometimes simply referred to as the organ dose [IAEA Dosimetry].

2.6.3.2. Equivalent dose H_T:

The equivalent dose, H_T, to an organ or tissue, T, is defined for a single type of radiation, R, it is the product of a radiation weighting factorW_R, for radiation R and the organ dose, D_T, thus:

$$H_T = W_R D_T \ldots\ldots..14.2$$

Unit: J/Kg.

The special name for the unit of equivalent dose is Sievert (Sv) The radiation weighting factor,W_R, allows for differences in the relative biological effectiveness of the incident radiation in producing stochastic effects at low doses in tissue or organ, T. For x-ray energies use in diagnostic radiology, W_R is taken to be unity[ICRU Report 51].

Table 2.1: Tissue Weighting Factors W_T

Tissue	Weighting Factors
Gonads	0.08
Breast	0.12
Red bone marrow	0.12
Lung	0.12
Thyroid	0.04
Bone surface	0.01
Colon	0.12
Stomach	0.12
Bladder	0.04
Esophagus	0.04
Liver	0.04
Brain	0.01
Kidney	-
Salivary glands	0.01
Skin	0.01
Remainder	0.12

The remainder is composed of the following additional tissue and organs: adipose tissue, adrenals, connective tissue, extra thoracic airways, gall bladder, heart wall, kidney, lymphatic nodes, muscle, pancreas, prostate, small intestine wall, spleen, thymus and uterus/cervix.

2.6.3.3. Effective dose E:

The effective dose, E, is defined for the sum over all the organ and tissue of the body of the product of the equivalent dose, H_R, to the organ or tissue and at Tissue weighting factor, W_T, for that organ or tissue, thus:

$$E = \sum W_T H_T \ldots \ldots .. 15.2$$

The tissue weighting factor, W_T, for organ or tissue T represents the relative contribution relative contribution of that organ or tissue to the total detriment arising from stochastic effects for uniform irradiation of the whole body.

Unit: J/Kg.

The special name for the unit of effective dose is Sievert (Sv)[IAEA Dosimetry].

Table 2.2: Radiation Weighting Factors WR

Radiation	Weighting Factors
Photons all energies	1
Electrons and muons, all energies	1
Neutrons	
< 10 Kev	2.5
10 – 100 Kev	2.5 to 10
100 – 2 Mev	10 to 20
2 – 20 Mev	7 to 17.5
>20Mev	5 to 7
Protons, energy > 2 Mev	2
Alpha particles, fission fragment, heavy nuclei	20

2.7 Previous studies

N.N. Jibiri et al (2014) Estimation of radiation dose to the lens of eyes of Patients undergoing cranial computed tomography, Used The Entrance Surface Dose (ESD) to the lens of eyes of 26 patients who had cranial CT procedures at a University Teaching Hospital in Ile-Ife, Nigeria has been determined in order to assess the level of radiation protection compliance and optimization of radiation safety at the hospital. The Results shows indicate that the doses to the patients ranged between 17.13 mGy and 51.98 mGy within the period under study. The average doses obtained for the pediatric patients (1.5-18 yrs), young adults (19-49 yrs) and adults (≥50 yrs) were 31.14 ±11.02 mGy, 41.81±12.60 mGy and 31.97 ± 11.31 mGy respectively. The mean dose obtained in this study was lower than threshold for lens damage, therefore the dose recorded in this study is clinically safe. In a teaching Hospital in Osun state, Nigeria.

M-Michele et al (2011) Eye lens radiation exposure and repeated head CT scans: A problem to keep in mind. The Objectives to deterministic character of radiation-induced cataract is being called into question, raising the possibility of a risk in patients, especially children, exposed to ionizing radiation in case of repeated head CT-scans. This study aims to estimate the eye lens doses of a pediatric population exposed to repeated head CTs and to assess the feasibility of an epidemiological study. And used Children treated for a cholesteatoma, who had had at least one CT-scan of the middle ear before their tenth birthday, were included. Radiation exposure has been assessed from medical records and telephone interviews. The results show of the 39 subjects contacted, 32 accepted to participate. A total of 76 CT-scans were retrieved from medical records. At the time of the interview (mean age: 16 years), the mean number of CT per child was 3. Cumulative mean effective and eye lens doses were 1.7mSv and 168 mGy, respectively.

S. Suzuki et al (2010) Lens Exposure during Brain Scans Using Multidetector Row CT Scanners: Methods for Estimation of Lens Dose, and used 8 types of multidetector row CT scanners, both axial and helical scans were obtained for the head part of a human-shaped phantom by using normal clinical settings with the orbitomeatal line as the baseline. We measured the doses on both eyelids by using an RPLGD during whole-brain scans including the orbit with the starting point at the level of the inferior orbital rim. To assess the effect of the starting points on the lens doses, we measured the lens doses by using 2 other starting points for scanning (the orbitomeatal line and the superior orbital rim). The result show of the CTDIvols and the lens doses during whole-brain CT including the orbit were 50.9–113.3 mGy and 42.6–103.5 mGy, respectively. The ratios of lens dose to CTDIvol were 80.6%–103.4%. The lens doses decreased as the starting points were set more superiorly. The lens doses during scans from the superior orbital rim were 11.8%–20.9% of the doses during the scans from the inferior orbital rim.

Chapter Three

Materials and Methods

3.1. Materials:

3.1.1 patients:

In this study collected the data of 51 patients from three hospitals (RCIH, ANH and ZSH) and containing the 34 male and 17 female, the Age is ranged from 19 to 95 years, Data were used to assess the doses of eye lens underwent head CT examinations. The local ethics committees of all participating institutions approved the study protocol.

3.1.2 CT equipment-specific information:

All the hospital in this study was used CT 64 slice model TOSHIBA with Specifications:

- TOSHIBA scanner Aquition (model TSX-101A, the input 3-200 V 50/60 H_Z, Max input power 100 KvA)

- The X-Ray High Voltage Generator (model CXG-012A, output (120 kv 600 mAs) (135 kv 530 mAs), Max input power 90 KvA)

- CT Scanner Gantry model CGGT- 021A, input 3-200v 50/60H_Z , max input power 25 KvA

3.2. Methods:

3.2.1 CT protocol:

In **RCIH** the protocol included the Kvp 120 kv ,mAs is ranged from 150 to 225 mA, slice thickness from3 to 5 mm, collimation from 19 to 32 mm and scan length from 11 to 23 cm.

In **ANH** the Kvp 120 kV, mAs is 225 mA, slice thickness from3 to 5 mm, collimation is 32 mm and scan length from 14 to 24 cm.

In **ZSH** the Kvp 120 kV, mAs is 225 mA, slice thickness is 5 mm, collimation is 32 mm and scan length from 14 to 24 cm.

3.2.2 Dosemetric calculations:

CT Expo software was used to calculate common CT dose descriptors: (i) CT weighted dose index ($CTDI_W$) and volume dose index ($CTDI_{vol}$) provides an indication of the average absorbed dose in the scanned region, (ii) CT dose –length product (DLP) the integrated absorbed dose along a line parallel to the axis of rotation for the complete CT examination, and (iii) Equivalent dose HT (E): a method for estimation and comparing the Equivalent dose of lens of eyes from different diagnostic procedures in three hospitals in Khartoum. In this study was used CT-Expo Version 2.5 software tool for dose calculations and CT-Expo tools—based on Monte Carlo data published by the Research Center for Environment and Health in Germany—for dose calculation. Dose estimation is done based on mathematical phantoms for adult (ADAM and EVA). The software allows the calculations of the CT dose descriptors ($CTDI_{vol}$ and DLP), Equivalent dose in accordance with new recommendations of the international commission for radiological protection ICRP 103 [Suliman II, lta,2014].

3.3 Place and time of study

This study was performed at three radiology department in Royal Care International Hospital (RCIH), Alzaytona Specialized Hospital ZSH and Alamal Hospital (ANH), during the period from (June to December 2016).

3.4 Data analysis

All dose parameters will have registered from Data collection sheet, then used as input to the Microsoft excel and SPSS software for analysis.

Chapter Four

Results

4.1. Results:

The results are presented for dose measurements performed in three CT units and 51 patients. Doses were estimated in terms of CTDIvol, DLP and E.

Table 4.1: show statistical parameters of patients and dose information's in RCIH:

RCIH	*Mean*	*Median*	*STD*	*Min*	*Max*
Age	52.53	52	18.059	23	80
KVp	120	120	0	120	120
MAs	216.18	225	24.908	150	225
S	4.76	5	0.664	3	5
NO. S	675.76	647	100.8	581	983
COLL	30.47	32	4.317	19	32
S.L	16.06	16	2.817	11	23
Pitch	0.47	0	0.514	0	1
CTDI vol	76.18	81	10.02	50	81
DLP	1459.24	1504	313.493	778	2090

Table 4.2: ***show statistical parameters of patients and dose information's in ANH:***

ANH	Mean	Median	STD	Min	Max
Age	47.36	43	21.331	19	80
KVp	120	120	0	120	120
MAs	225	225	0	225	225
S	4.76	5	0.664	3	5
NO. S	695.14	673.5	84.688	612	947
COLL	32	32	0	32	32
S.L	16.79	16	2.455	14	24
Pitch	54.29	37.5	38.08	22	150
CTDIvol	80.29	81	4.681	77	95
DLP	1562.5	1540	197.846	1302	2073

Table 4.3: ***show statistical parameters of patients and dose information's in ZSH:***

ZSH	Mean	Median	STD	Min	Max
Age	61.3	64	21.969	27	95
KVp	120	120	0	120	120
MAs	225	225	0	225	225
S	5	5	0	5	5
NO. S	752.1	707	102.531	618	978
COLL	32	32	0	32	32
S.L	17.6	16	2.927	14	24
Pitch	92.6	62.5	79.198	27	330
CTDIvol	79.55	81	2.502	72	81
DLP	1620.35	1544	234.791	1309	2150

Table 4.4: Shows mean CTDIw (mGy), CTDIvol (mGy), DLP (mGy.cm), equivalent dose of eye lens and effective dose (mSv) calculated using CT-Expo software in RCIH.

RCIH	***CTDIw (mGy)***	***CTDIvol (mGy)***	***DLP (mGy*cm)***	***Eff D mSv***	***Eq D Eye mSv***
Mean	43.312	80.112	1420.925917	3.924	82.965
Median	44.800	84.000	1468.824417	3.900	87.000
Std. Deviation	4.4865	11.6287	330.5626451	1.2498	13.1610
Minimum	31.4	49.0	606.3073	1.2	47.9
Maximum	45.1	86.6	1798.9010	5.6	91.1

Table 4.5: Shows mean CTDIw (mGy), CTDIvol (mGy), DLP (mGy.cm), equivalent dose of eye lens and effective dose (mSv) calculated using CT-Expo software in ANH***:***

ANH	***CTDIw (mGy)***	***CTDIvol (mGy)***	***DLP (mGy*cm)***	***Eff D mSv***	***Eye lenses (mSv)***
Mean	44.900	85.286	1598.565764	4.636	88.607
Median	44.800	86.600	1590.069690	4.500	89.300
Std. Deviation	.2660	2.8852	169.5556031	1.3299	2.6863
Minimum	44.6	78.9	1358.6695	3.1	82.8
Maximum	45.3	89.1	2034.1683	8.7	92.8

Table 4.6: Shows mean CTDIw (mGy), CTDIvol (mGy), DLP (mGy.cm) and effective dose (mSv) calculated using CT-Expo software in ZSH.

ZSH	***CTDIw (mGy)***	***CTDIvol (mGy)***	***DLP (mGy*cm)***	***Eff D mSv***	***H Eye lenses mSv***
Mean	45.067	83.439	1627.316682	5.344	87.017
Median	45.100	82.750	1528.378971	4.600	86.600
Std. Deviation	.2114	2.7032	249.2177867	2.3078	2.5096
Minimum	44.6	78.9	1276.5230	3.1	83.5
Maximum	45.3	89.1	2165.8250	10.2	92.9

Figure 4.1 show comparison of effective dose in CT head scans between three hospitals:

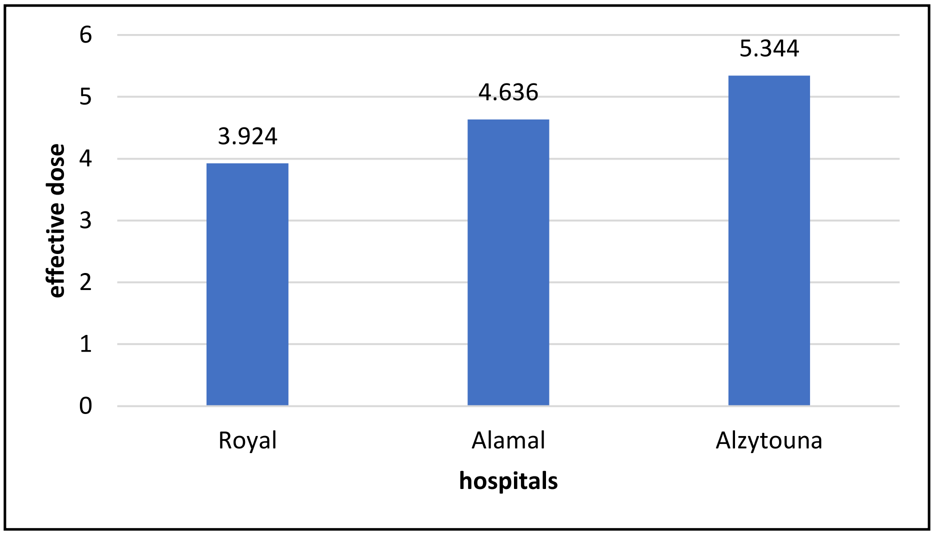

Figure 4.2 show comparison of equivalent dose of lens between three hospitals:

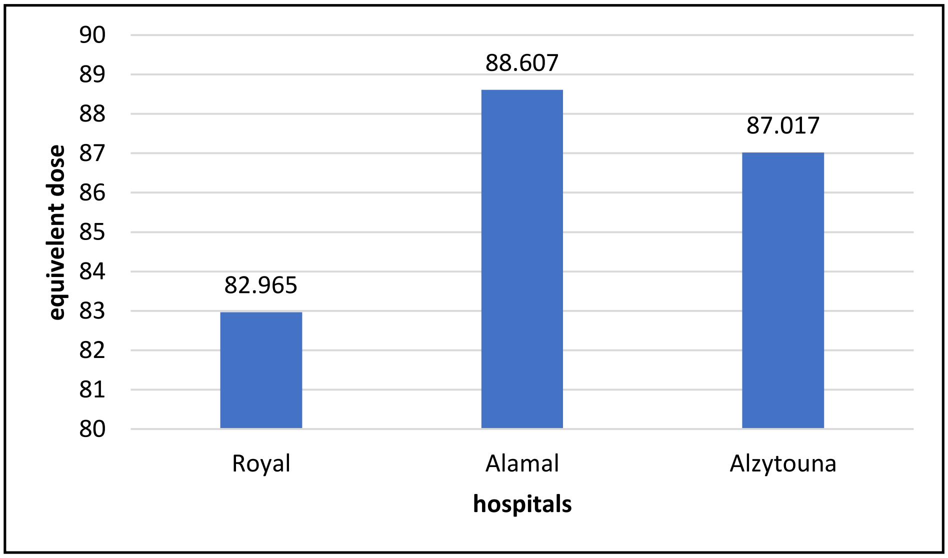

Figure 4.3 show comparison of CTDIw and CTDIvol between three hospitals:

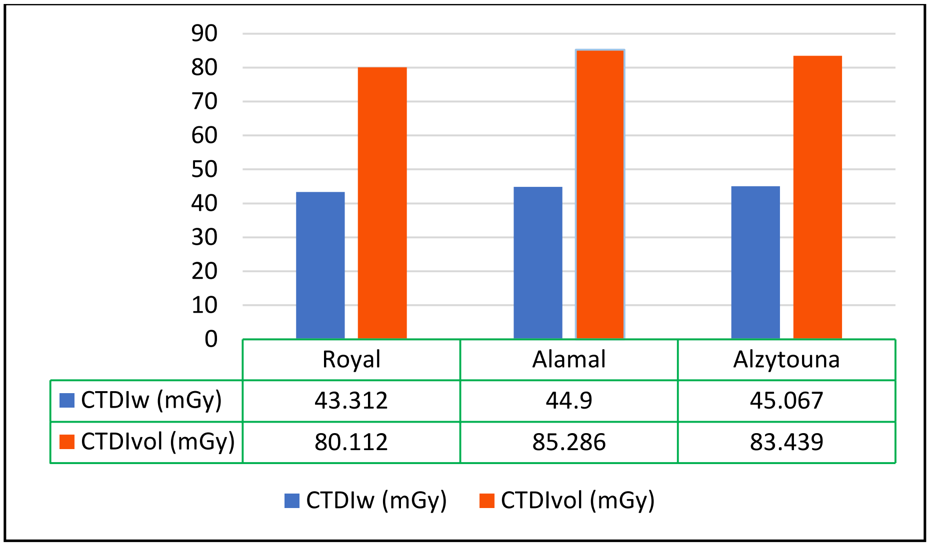

Figure 4.4 show comparison of DLP between the three hospitals:

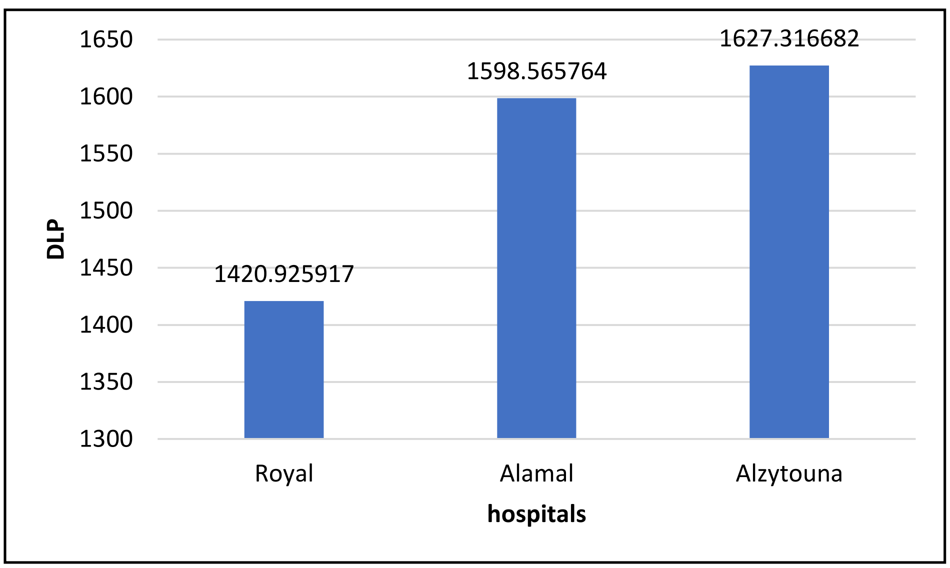

Table 4.7: Shows mean CTDIw (mGy), CTDIvol (mGy), DLP (mGy.cm), equivalent dose of eye lens and effective dose (mSv) calculated using CT-Expo software for all hospitals.

All hospital	***CTDIw (mGy)***	***CTDIvol (mGy)***	***DLP (mGy*cm)***	***Eff D mSv***	***Eq Eye lenses***
Mean	44.410	82.812	1547.497175	4.649	86.065
Median	44.900	84.200	1535.846400	4.300	87.700
Std. Deviation	2.7209	7.3779	273.9177232	1.8038	8.2189
Minimum	31.4	49.0	606.3073	1.2	47.9
Maximum	45.3	89.1	2165.8250	10.2	92.9

Table 4.8: Comparison of CTDIvol, DLP and E Dose obtained in the present study with previously published data for routine CT examinations.

Doses	This study	Tanzania	Australia	UK	Sudan	Canada
$CTDI_{vol}$ (mGy)						
Head	82.812±7.3	43	86.7	55-65	65.4	-
DLP (mGy.cm)						
Head	1547.5±273.9	913	1508.4	760-930	758	930-1300
E (mSv)						
Head	4.649±1.2	2.1	3.61	1.5	1.6	-

Chapter Five

Discussion, Conclusion and Recommendation

5.1 Discussion:

Doses in this study are expressed in terms of CTDIvol, DLP, equivalent dose of eye lens and E. They provide an indication of the average absorbed dose in the scanned region (CTDIvol), the integrated absorbed dose along a line parallel to the axis of rotation for the complete CT examination (DLP), a method for comparing patient doses from different diagnostic procedures (E) and equivalent dose of Eye lens (H) [Brenner, et al,2007] [Kim, et al ,2011].

From (Table 4.4) as shown The mean $CTDI_W$ ranged from 31.4 mGy to 45.1 mGy, $CTDI_{vol}$ ranged from 49 mGy to 86.6 mGy, DLP ranged from 606 mGy to 1798.9 mGy, Effective dose ranged from 1.2 to 5.6 mSv and dose of eye lens ranged from 47.9 to 91.1 mSv (in RCIH).

From The (Table 4.5) as shown The mean $CTDI_W$ ranged from 44.6 mGy to 45.3 mGy, $CTDI_{vol}$ ranged from 78.9 mGy to 89.1 mGy, DLP ranged from 1358 mGy to 2034 mGy, Effective dose ranged from 3.1 to 8.7 mSv and dose of eye lens ranged from 82.8 to 92.8 mSv (in ANH).

From The (Table 4.6) as shown The mean $CTDI_W$ ranged from 44.6 mGy to 45.3 mGy, $CTDI_{vol}$ ranged from 78.9 mGy to 89.1 mGy , DLP ranged from 1276.5 mGy to 2165.8 mGy, effective dose ranged from 3.1 to 10.2 mSv (in brain) and dose of eye lens ranged from 83.5 to 92.9 mSv (in ZSH) .

From The (Table 4.7) as shown The mean $CTDI_W$ ranged from 31.4 mGy to 45.3 mGy, $CTDI_{vol}$ ranged from 49 mGy to 89.1 mGy , DLP ranged from 606 mGy to 2165.8 mGy, Effective dose ranged from 1.2 to 10.2 mSv (in brain) and dose of eye lens ranged from 47.9 to 92.9 mSv (for All hospitals) .

From the (Figure 4.1) is shown the comparison of the effective dose between the all hospital in this study, the small Effective dose founded in RCIH (3.92 mSv) after that ANH (4.64 mSv) and higher dose in ZSH (5.34 mSv), from (Figure 4.2) the Equivalent dose of eye lens the small dose founded in RCIH (82.97 mSv), after that ANH (88.61 mSv) and higher dose in ZSH (97.01 mSv), from (Figure 4.3) the $CTDI_W$ and CTD_{vol} founded the $CTDI_W$ higher was in ZSH (45.067 mGy) then in ANH (44.9 mGy) and lowers at RCIH (43.3 mGy) and CTD_{vol} higher was in ANH (85.29 mGy) then in ZSH (83.44 mGy) and lowers at Royal Care International Hospital (80.11 mGy) and from (Figure 4.4) the DLP higher was in ZSH (1627.3 mGy*cm) then in ANH (1598.56 mGy*cm) and lowers at RCIH (1420.9 mGy*cm).

5.2 Conclusion:

The assessment of radiation doses of patients undergoing CT examinations in Sudan was investigated. $CTDI_{vol}$, DLP , effective dose and eye lens equivalent dose.

In this study, high effective dose and large variations of radiation dose were observed. Different scanning protocols used among hospitals responsible for these variations. The mean Effective dose in this study were mostly comparable to and slightly higher than reported values from the United Kingdom, Tanzania, Australia, and Sudan. The main contributor for this difference was the use of a larger scan length in Sudan than that used in some of these countries.

comparison of the effective doses and equivalent doses between the hospitals the higher dose was in Alzytouna then in Alamal hospital and lowers at Royal Care International Hospital.

Considering the entire sample, hospitals achieved mean $CTDI_{VOL}$ values higher than similar study in other country example as (UK, Sudan and Tanzania),except the Australia, when compare the all $CTDI_{vol}$ value in all patient was found lower than Australia (Table 4.8). It is important to note all hospitals were using Toshiba CT generally having higher output per tube current as results less filtration compared to scanners from other manufactures.

The DLP values was found in this study were higher than those presented in other studies. Notably the current doses values were the highest among all. This is mainly related to increased scan coverage and tube-current exposure time used.

The mean value to estimated Effective dose E, in this study was higher than the corresponding values presented in UK, Tanzania and Sudan [R. E. Moorin,lta,vol 295].

5.3 Recommendations:

- The large observed variations of Effective dose and organ doses among hospitals and relatively high effective dose and organ doses in Sudan hospitals call for the need to optimize CT scanning protocols.
- Optimal selection of scanning parameters based on indication of study, body region of interest being scanned, and patient size.
- further studies should be done to investigate the potential for using radio protective materials to protect superficial radiosensitive organs.

5.4 Reference:

G. F. Acquah, B. Schiestl, A. Y. Cofie, and J. O. Nkansah, “Radiation dose reduction without degrading image quality during computed tomography examinations : Dosimetry and quality control study Technical Report Abstract,” 2014.

M. H. Ali, “School of Medical Radiation Sciences Trends in CT Abdominal Doses in Malaysian Practices,” no. March, 2005.

P. Lin, T. Beck, C. Borras, G. Cohen, and R. Jucius, AAPM Report No. 39 - Specification and acceptance testing of computed tomography scanners, no. 39. 1993

M. Beeres, M. Römer, B. Bodelle, C. Lee, T. Gruber-rouh, E. Mbalisike, J. M. Kerl, J. L. Wichmann, B. Schulz, T. J. Vogl, and R. W. Bauer, “Chest-abdomen-pelvis CT for staging in cancer patients : dose effectiveness and image quality using automated attenuation-based tube potential selectionek,” pp. 1–8, 2014

D. J. Brenner and E. J. Hall, “Computed Tomography — An Increasing Source of Radiation Exposure,” pp. 2277–2284, 2007

D. R. Dance, S. Christofides, I. D. Mclean, and K. H. Ng, “Diagnostic Radiology Physics.”

F. Edition, MEDICAL MEDICAL Fourth Edition.

H. Human and F. Group, “Computed Tomography Radiation Safety Issues in Ontario Table of Contents,” Appraisal, 2006.

IAEA,“<Dosimetryindiagnosticradiology-Aninternationalcodeofpractice-140906224911-Phpapp01.Pdf>,” vol. TECHNICAL

G. R. Iball, D. S. Brettle, and A. C. Moore, “Assessment of tube current modulation in pelvic CT,” vol. 79, no. January, pp. 62–70, 2006.

“ICRP Publication 103 The 2007 Recommendations of the International Commission on Radiological Protection,” 2007.

ICRU, “ICRU Report 51 : Quantities and units in radiation protection dosimetry.” pp. 1–14, 1993.

J. M. B. Jerrold T. Bushberg, J. Anthony Seibert, Edwin M. Leidholdt, Jr, “The Essential Physics For Medical Imaging second edition-20150929-085027787.” 2002.

S. Kim, H. Song, E. Samei, F. Yin, and T. T. Yoshizumi, “Computed tomography dose index and dose length product for cone-beam CT : Monte Carlo simulations of a commercial system,” vol. 12, no. 2, pp. 84–95, 2011

C. T. Martinsen, H. K. Saether, D. R. Olsen, P. A. Wolff, and P. Skaane, “Improved image quality of low-dose thoracic CT examinations with a new postprocessing software.,” J. Appl. Clin. Med. Phys., vol. 11, no. 3, p. 3242, 2010

R. E. Moorin, R. Forsyth, D. J. Gibson, and R. Fox, “Radiation dosimetry assessment of routine CT scanning protocols used in Western Australia,” vol. 295

O. I. Ntly and S. By, “IAEA- safety reports series No. 39: Applying radiation safety standards in diagnostic radiology and interventional procedures using x-rays-sti/pub/1206 - Vienna (2006),”

A.Ploussi, V. Tsitsia, I. Arapakis, S. Kordolaimi, S. Argentos, I. Seimenis, and E. P. Efstathopoulos, “Evaluation of Radiation Exposure Reduction using Hybrid Iterative Reconstruction Algorithm in Chest-Abdomen-Pelvis CT,” pp. 40–44, 2013.

E. B. Podgorsak, “Radiation Oncology Physics : A Handbook for Teachers and Students.”

L. Sadri, H. R. Khosravi, and S. Setayeshi, “Assessment and evaluation of patient doses in adult common CT examinations towards establishing national diagnostic reference levels,” Int. J. Radiat. Res., vol. 11, no. 4, pp. 245–252, 2013.

R. D.Prins, R. H. Thornton, C. R. Schmidtlein, B. Quinn, H. Ching, and L. T. Dauer, “Estimating radiation effective doses from whole body computed tomography scans based on U . S . soldier patient height and weight,” BMC Med. Imaging, vol. 11, no. 1, p. 20, 2011

Suliman, S. E. Abdalla, N. a Ahmed, M. a Galal, and I. Salih, “Survey of computed tomography technique and radiation dose in Sudanese hospitals.,” Eur. J. Radiol., vol. 80, no. 3, pp. e544–51, 2011.

Suliman II, H. M. Khamis, T. H. Ombada, K. Alzimami, M. Alkhorayef, and a Sulieman, “Radiation Exposure during Paediatric Ct in Sudan: Ct Dose, Organ and Effective Doses,” Radiat Prot Dosim., pp. 1–6, 2014.

P. This, P. Parameters, and T. Standards, “ACR – ASER – SCBT-MR – SPR PRACTICE PARAMETER FOR THE PERFORMANCE OF PEDIATRIC COMPUTED TOMOGRAPHY (CT),” vol. 1076, no. Revised 2008, pp. 1–18, 2014.