Sudan University of Science and Technology

College of Graduate Studies

Estimation of Effective Radiation Dose During CT K.U.B

تقويم الجرعة الإشعاعية الفعالة أثناء تصوير الكلى والحالب والمثانة بالأشعة المقطعية المحوسبة

A Thesis Submitted in partial Fulfillment for requirements Degree of in

M.Sc. Medical physics

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الآيــــــة

بسم الله الرحمن الرحيم

قَالَ تَعَالَىٰ:

﴿ مُتَّكِـِٔينَ عَلَىٰ فُرُشِۭ بَطَآئِنُهَا مِنۡ إِسۡتَبۡرَقٖۚ وَجَنَى

ٱلۡجَنَّتَيۡنِ دَانٖ ٥٤ ﴾

سورة الرحمن:الاية ٥٤

Dedication

To support me in my life

To My Parents.

To whom support me in my life

To my brothers and my sisters.

To that souls who with me always and push me for continue this work

To my friends.

Acknowledgment

"I humbly thank Allah Almighty, the Merciful and the Beneficent, who gave me health, thoughts and cooperative people to enable me achieve this goal"

I express my deepest thanks to my supervisor

Dr. / Hussein Ahmed Hassan

My thanks to Mr. Mustafa Awad who was help me when I was need to him Also, my deep thanks to my brother Khalid Ahmed who support and help me Also, my thanks to my friend Elameer Ezzaldein who encouraged me to write this work and my thank to the families of the three hospitals in Khartoum state from which the data of this study was collected.

Abstract

CT is an x-ray procedure that generates high quality sectional images of the body, these images provide more detailed information than normal x-ray images, but CT gives patients high radiation exposure compared to conventional x-ray devices.

The the main objective of this study to estimate the effective radition dose during CT KUB CT scan.

In this study the effective radiation dose was estimated for patients underwent KUB CT scan in three hospitals in Khartoum state (November February 2020).

The data was collected form scan of 90 adult patients during KUB CT scan perfomed by different CT scanners (Toshiba 64 slice, Siemens 32 slice, Toshiba 16 slice), the CTDI, DLP were used to estimate effective radiation dose received by patients.

The result of this study showed that the mean effective dose in CT devices (Toshiba 64 slice), (Siemens 32 slice) and (Toshiba 16 slice) (8.23 ± 0.68) mSv, (4.82 ± 1.95) mSv and (4.14 ± 1.33) mSv respectively and total mean effective dose in this study is (5.7 ± 1.32) mSv.

In this study found that the effective radiation dose received by patient from CT scanner 64 slices is higher than CT scanner 32 slices and CT scanner 16 slice, this differencedue to the difference in protocols of scan used.

المستخلص

التصوير المقطعي المحوسب ينتج صور مقطعية عالية الجودة للجسم, وتوفر هذه الصور معلومات اكثر تفصيلا من صور الاشعة السينية العادية لكن استخدام جهاز الاشعة المقطعية يعطي المرضى جرعة اشعاعية عالية مقارنة مع تصوير الاشعة السينية التقليدية.

تهدف هذه الدراسة تقدير الجرعة الاشعاعية الفعالة فى تصوير المقطعى المحوسب للكلى والحالب والمثانة.

في هذه الدراسة تم تقدير الجرعة الاشعاعية الفعالة للمرضى الذين خضعوا لفحص الكلى والحالب والمثانة عن طريق الأشعة المقطعية في ثلاث مستشفيات في ولاية الخرطوم خلال الفترة (نوفمبر_ فبراير 2020).

مجموع المرضى في هذه الدراسة 90 مريض من البالغين الذين خضعوا لفحص الكلى والحالب والمثانة عن طريق الأشعة المقطعية, وذلك بإستخدام انواع مختلفة من اجهزة الأشعة المقطعية وهى (توشيبا 64 شريحة, سيمنث 32 شريحة, توشيبا 16 شريحة), تم جمع البيانات مؤشر الجرعة للاشعة المقطعية ومعامل الجرعة الطولي , ومن ثم تقييم الجرعة الإشعاعية الفعالة للمرضى.

نتائج هذه الدراسة أظهرت أن متوسط الجرعة الفعالة في اجهزة الاشعة المقطعية (توشيبا 64 شريحة), (سمينث 32 شريحة) و (توشيبا 16شريحة) هى (8.23 ± 0.68 mSv), (4.82 ± 1.95) mSv (4.14 ± 1.33) mSv على التوالي, وأن متوسط الجرعة الفعالة الكلية لهذه الدراسة هى

(5.7 ± 1.32) mSv.

لخصت هذه الدراسة أن متوسط الجرعة الاشعاعية الفعالة للمرضى بواسطة جهاز الأشعة المقطعية توشيبا 64 شريحة أعلى من بالجرعة الاشعاعية لجهاز سمينث 32 شريحة و توشيبا 16 شريحة.

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List of abbreviation

ALARA: As Low As Reasonably Achievable.

CT: Computed Tomography.

CTDIvol: Computed Tomography Dose Index Volume.

CTDIw: Computed Tomography Dose Index weighted.

DLP: The Dose Length Product.

DSR: Dynamic Spatial Reconstructor.

EBCT: Electron Beam Computed Tomography

ICRP: International Commission on Radiological Protection.

ICRU: International Commission on Radiation Unit.

KUB: Kidney Ureter Bladder.

kV: Kilovolt.

LET: Linear Energy Transfer.

mAs: milli Ampere second.

MSAD: Multiple Scanned Average Dose.

NCRP: National Committee on Radiation Protection.

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CHAPTER ONE
Introduction

1.1 Introduction

Computed tomography introduced since 1972, computed tomography (CT) has become an integral diagnostic tool of modern medicine. According to national surveys of CT use, it is estimated that approximately 70 million CT examinations are performed per year in the USA, and the use of CT in Europe and Asia has experienced large increases as well. (Menzel el at., 2012)

Medical imaging has experienced significant changes in both technologic and clinical arenas. Innovations have become common in the radiology department, and today we see the introduction of new ideas, methods, and techniques, as well as improvement in existing techniques. The outmost goal of these developments is to acquire f optimum diagnostic information while improving the quality of care forded to patients. One such development, which has come to be known as a revolutionary process in medicine, and particularly in medical imaging, is computed tomography (CT). (Seeram E, 2000)

There are three general types of x-ray examinations: radiography, fluoroscopy, and computed tomography.

Computed tomography uses a rotating x-ray source and detector array. A volume of data is acquired so that fixed images can be reconstructed in any anatomical plane-coronal, sagittal, transverse, or oblique. There are many variations of these three basic types of examinations, but in general, x-ray imaging equipment is similar. (Boshong, 2017)

Figure (1.1): shows computed tomography instrument.

In this research the patient undergoes to CT imaging so it exposed for radiation Unfortunately, patient radiation dose is on the rise, including doses high enough to cause injury. This injury can be deterministic effect and stochastic effect.

Deterministic effect of radiation exposure are produced by high radiation doses. Stochastic effects of radiation exposure are the result of low doses delivered over a long period.

Radiation protection guides are based on suspected or observed stochastic effects of radiation and on an assumed linear, non-threshold dose response relationship. (Boshong, 2017)

Therefor the radiation dose must be optimize dose. that refers to a dose that provides enough but not perfect image quality but not with excessive radiation, and is the practical application of ALARA (As Low As Reasonably Achievable) principle. (Tack et al., 2012)

1.2 Problem of study

Due to use CT scanning patients are exposed to more doses which may result in unintended health effects, to avoid unnecessary of high dose to the patient need to estimate the effective dose.

1.3 Objective

1.3.1 General objective

To estimate radiation dose for patients during Kidney, Ureter and Bladder (KUB) in Computed Tomography (CT) in different centers.

1.3.2 Specific objective

- To determine effective dose to patient undergoing CT KUB.
- To evaluate the organ dose in the different centers.
- To Compare effective dose between the different centers.
- To Compare effective dose between this study and previous study.

1.4 thesis Overview

Chapter one is the introduction of the thesis.

Chapter two contains the background material for the thesis and previous studies.

Chapter three describes the materials and a method used in this study.

Chapter four is details the results of this study.

Chapter five presents the discussion, conclusion and recommendations of the thesis and Reference.

CHAPTER TWO

Theoretical Background

2.1 Radiation

Radiation may be classified as electromagnetic or particulate, with electromagnetic radiation. Radiation is classified as ionizing or non-ionizing, depending on its ability to ionize matter:

●non-ionizing radiation cannot ionize matter.

●ionizing radiation can ionize matter**.**

2.1.1 Ionization Radiation

Ionizing radiation is any type of radiation that is capable removing an orbital electron from the atom with which it interacts. (Boshong, 2017)

The major form of ionizing radiation are: Alpha particle (helium nuclei), Beta particle (electron), X and Gamma rays (photons). (Hill, 1979)

Ionization occurs when an x-ray passes close to an orbital electron of an atom and transfers enough energy to the electron to remove it from the atom. The ionizing radiation may interact with and ionize additional atoms. The orbital electron and the atom from which it was separated are called an ion pair. The electron is a negative ion, and the remaining atom is a positive ion. (Bonshong, 2017)

The X-ray or gamma photon transfer energy to the electron and produce the biological effect. (Thayalan, 2001)

The biological effects of ionizing radiation are threefold: short term (e.g. radiation sickness), long term (e.g. leukemia) and genetic (e.g. congenital malformation). (Hill, 1979)

2.2 X-ray production

2.2.1 Discovery of X-rays

X -rays were discovered by W. C. Roentgen, the German physicist in 1895 when he was researching conduction of electricity through gases at law pressure in glass tubes. He noticed that positive electrodes in the tubes gave off invisible rays that caused fluorescent screen (barium platino cyanide screen kept near the tube) to glow and fogged photographic plates. The rays were very penetrating, they passed through black paper and even thicker objects. They were not deflected in a magnetic field. So, Rontgen concluded that they were not charged particles. As their nature was not known he called them X-ray. Later they were shown electromagnetic radiation of very short wavelength (high frequency).

2.2.2 Requirements of X-ray production

The production of X-ray needs the following:

- Electron source (cathode).
- Target to top the electron (anode).
- High voltage supply to accelerate electron.
- Vacuum.
- Tube insert and housing.

Electron can be produced either by ionization in gas or by thermionic emission. The electron source acts as a cathode and the target acts an anode. The high voltage is applied in between the cathode and anode. This voltage accelerates the electrons to a high velocity as result the electron will have high kinetic energy. When the electrons are stopped by the target, the electron kinetic energy is converted into X-ray energy. Thus X-ray are produced. The equipment having all the above requirements is called an X-ray tube. (Thayalan, 2001)

2.3 The principle of CT

X-ray projection, attenuation and acquisition of transmission profiles:

The process of CT image acquisition involves the measurement of X-ray transmission profiles through a patient for a large number of views. A profile from each view is achieved primarily by using a detector arc generally consisting of 800–900 detector elements (dels), referred to as a detector row. By rotation of the X ray tube and detector row around the patient, a large number of views can be obtained. The use of tens or even hundreds of detector rows aligned along the axis of rotation allows even more rapid acquisition. The acquired transmission profiles are used to reconstruct the CT image, composed of a matrix of picture elements (pixels). (Dance et al., 2014)

The values that are assigned to the pixels in a CT image are associated with the attenuation of the corresponding tissue, or, more specifically, to their linear attenuation coefficient μ (m^{-1}). The linear attenuation coefficient depends on the composition of the material, the density of the material and the photon energy, as seen in beer's law:

$$I(x) = I_o e^{-\mu x} \qquad \text{..................................} \quad (1)$$

Where I(x) is the intensity of the attenuated X-ray beam, I_o the un-attenuated X-ray beam and x the thickness of the material. Note that beer's law only describes the attenuation of the primary beam and does not take into account the intensity of scattered radiation that is generated.

As an X-ray beam is transmitted through the patient, different tissues are encountered with different linear attenuation coefficients. If the pathway through the patient ranges from 0 to d, then the intensity of the attenuated X ray beam, transmitted a distance d, can be expressed as

$$I\,(d) \;=\; I_o e^{-\int \mu(x) dx} \qquad \text{……………..} \quad (2)$$

Since a CT image is composed of a matrix of pixels, the scanned patient can also be regarded as being made up of a matrix of different linear attenuation coefficient volume elements (voxels). Figure 2.1 shows a simplified 4 × 4 matrix representing the measurement of transmission along one line. For such a discretization, the equation for the attenuation can be expressed as:

$$I(d) = I_o e^{-\sum_{i=1}^{i=4} \mu i \Delta x} \quad \ldots\ldots\ldots\ldots\ldots (3)$$

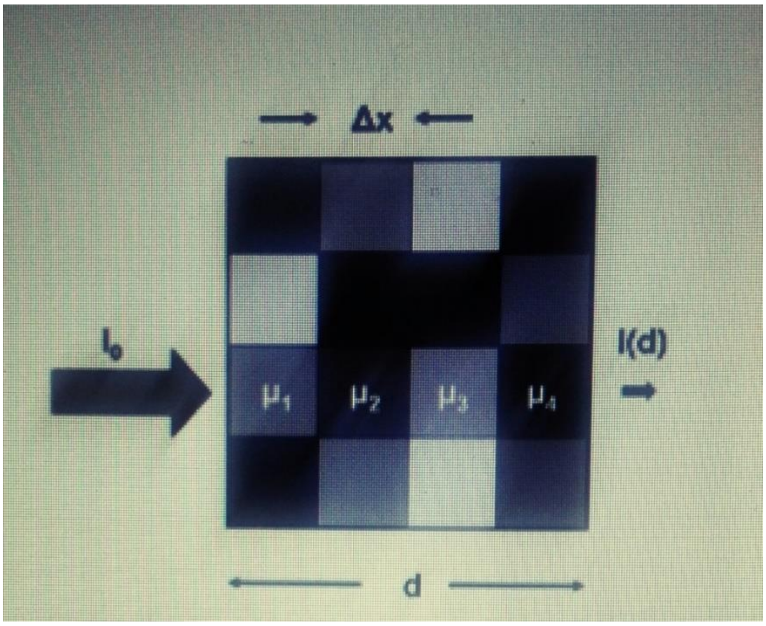

Figure 2.1: The principle of attenuation of an X-ray beam in a simplified 4×4 matrix. Each element in the matrix can, in principle, have a different value of the associated linear attenuation coefficient. (Dance D et al., 2014)

2.4 CT Generation

2.4.1 First Generation: Rotate/translate, pencil beam

The first CT scanners used a rotate/translate system with a single X-ray beam called a pencil beam. The pencil beam used parallel beam geometry. The parallel beam geometry is defined by a set of parallel rays that generates a projection profile. Highly collimated x-ray beam and one or two detectors first translate across the patient to collect transmission reading. After one transition, the tube and detector rotate by 1 degree and translate again to collect reading from a different direction. This repeated foe 180 degree around the patient.

Frist generation CT scanner took at least 4.5 to 5.5 minutes to produce a complete scan of the patient. A major drawback of first generation CT scanner was the amount of time it took to acquire the images and to reconstruct the image using the computer.

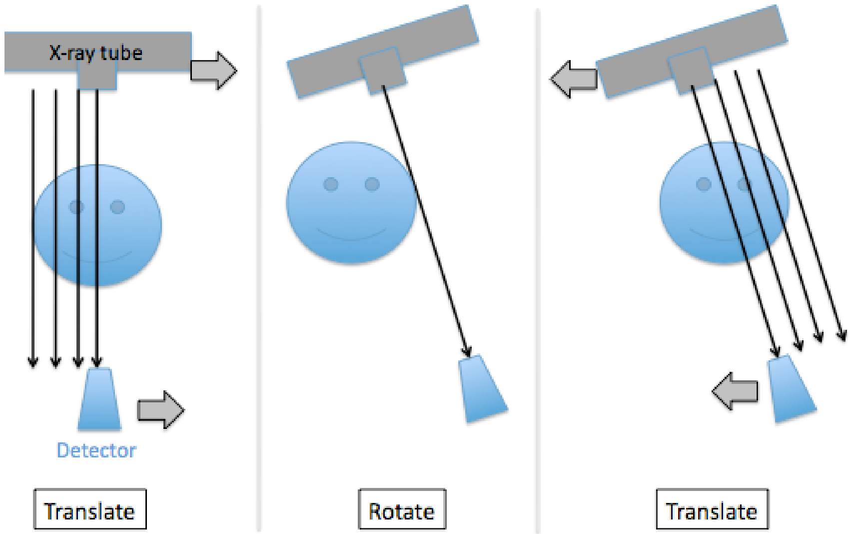

Figure 2.2: first generation scanners used translation and rotation. The (middle), and translate back across (right).

2.4.2 Second Generation: rotate/translate, Narrow Fan Beam

Second generation scanner were based on the translate-rotate principle of first generation scanners with a few fundamental differences such as a linear detector array about 30 detector connected to the x-ray tube and multiple beams. The result is beam geometry that describes a small fan whose apex originates at the x-ray tube. In second generation scanner, the fan beam translates across the patient to collect a set of transmission readings. After one translation, the tube and detector array rotate by larger increments and translate again. this process is repeated for 180 degree and it took shorter scan time than first generation that range from 20 second to 3.5 minutes.

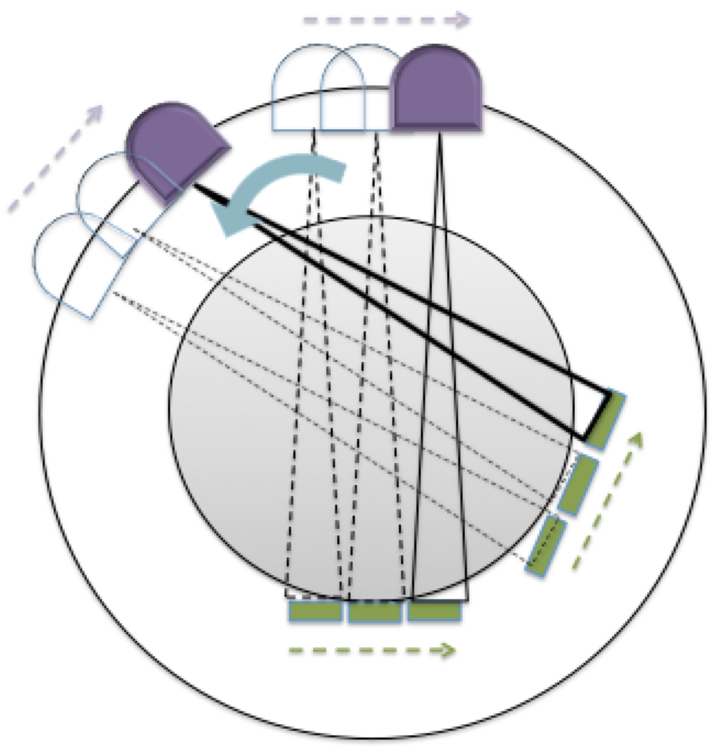

Figure 2.3: Second generation CT scanner Rotation-translation of a narrow fan beam.

2.4.3 Third Generation: Rotate/Rotate, Wide Fan Beam

The third generation CT scanners were based on a fan beam geometry that rotates continuously around the patient for 360 degrees. The x-ray tube is coupled to a curved detector array that subtends an arc of 30 to 40 degrees or greater from the apex of the fan. As the x-ray tube and detector rotate, projection profiles are collected and a view is obtained for every fixed point of the tube and detector. The path traced by the tube describes a circle rather than the first and second generation and the time scan shorter than pervious units (generally within a few seconds).

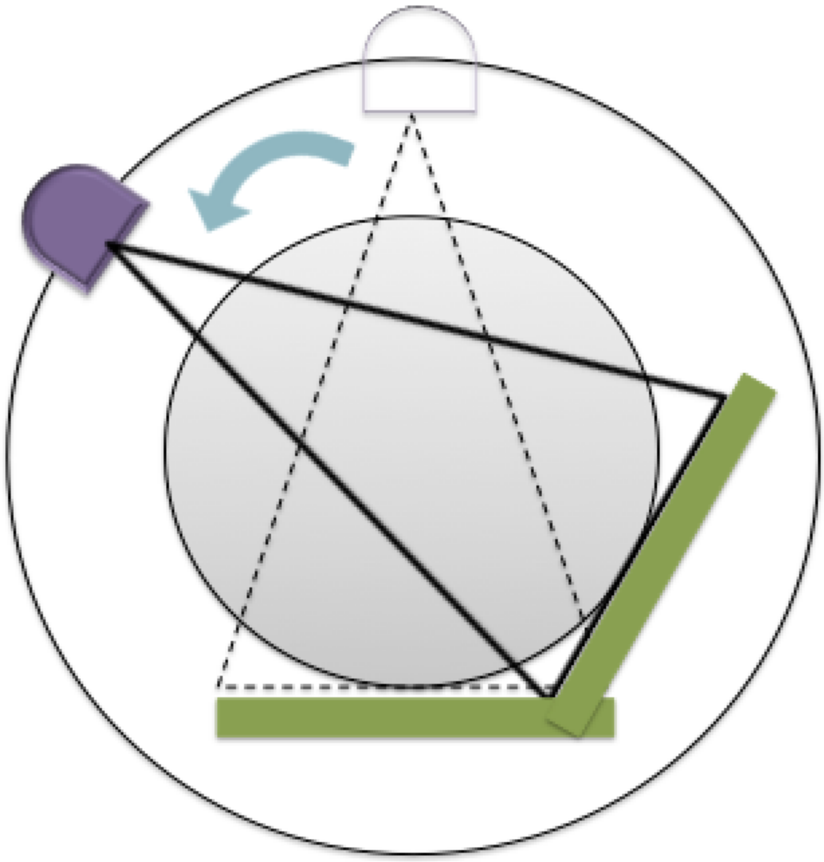

Figure 2.4: Third generation CT scanner. Rotation of a wide fan beam.

2.4.4 Fourth Generation Scanners: Rotate/Stationary

Fourth generation CT scanners feature two type of beam geometries: a rotating fan beam within a circular detector array, and rotating fan beam outside a nutating detector ring.

The rotating beam within a circular detector array: the x-ray is positioned within a stationary, circular detector array, the beam geometry describes a wide fan, the tube moves from point to point within the circle and the scan times are very short and vary from scanner to scanner, depending on the manufacturer.

The rotating fan beam outside a nutating detector ring: scanners with this type of scanning motion eliminate the poor geometry of other schemes, the tube rotates inside its detector ring, near the objet. However, nutate-rotate system are not currently manufactured.

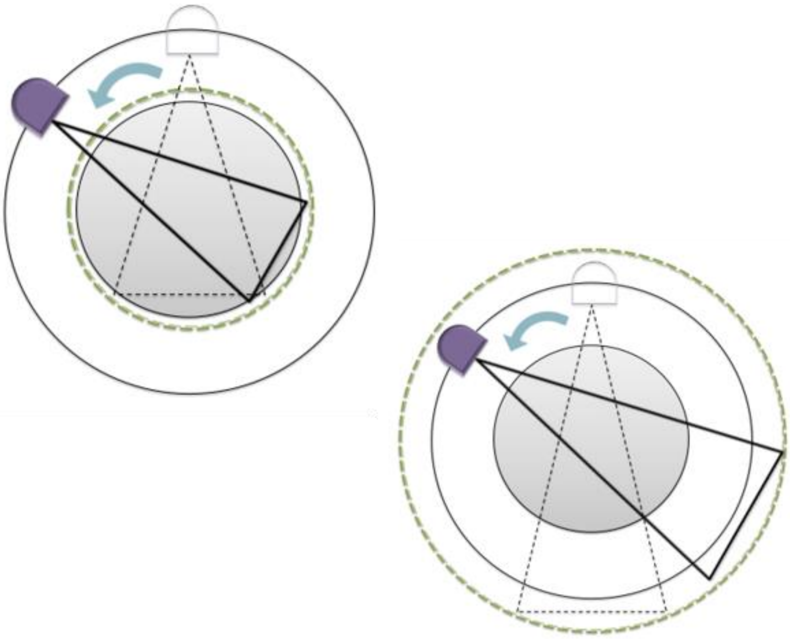

Figure 2.5: fourth generation CT scanner. Rotation fixed with closed detector ring. The detector can be located either inside or outside of the rotation axis of the x-ray tube.

2.4.5 Fifth Generation Scanners: Stationary/Stationary

Fifth generation scanners are classified as high speed CT scanners because they can acquire scan data in milliseconds. Two such scanners are the electron beam CT scanner (EBCT) and dynamic spatial reconstructor scanner. In electron beam CT scanner, the data acquisition geometry is a fan beam of x-ray produced by beam of electron that scans several stationary tungsten target rings. The fan beam passes through the patient and X-ray transmission reading are collected for image reconstruction. The dynamic spatial reconstructor scanner (DSR) is highly specialized fifth generation, high speed scanner capable of producing dynamic three dimensional (3D) images of volumes of the patient. (Seeram, 2000)

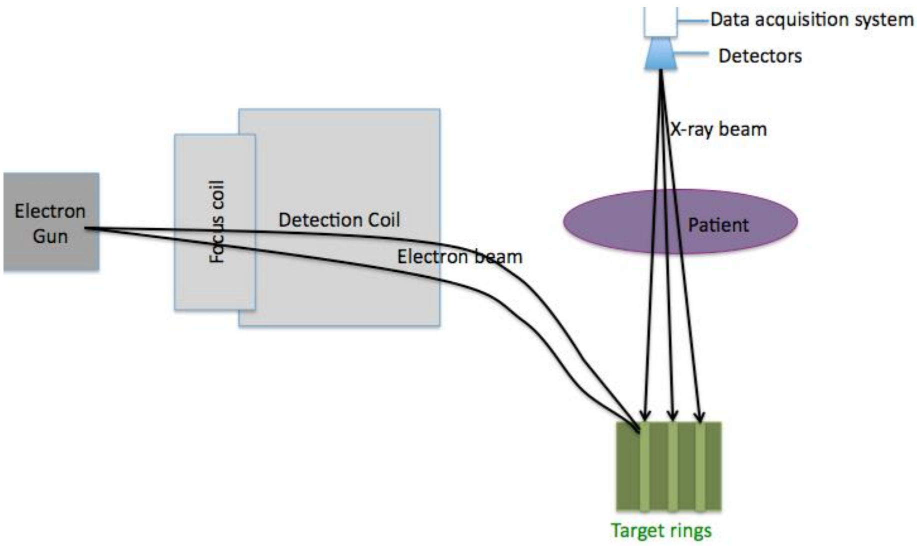

Figure 2.6: fifth generation scanner (electron beam scanner).the electron beam is directed around the target rings, allowing for all stationary instrumentation.

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 increase 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 relationship.

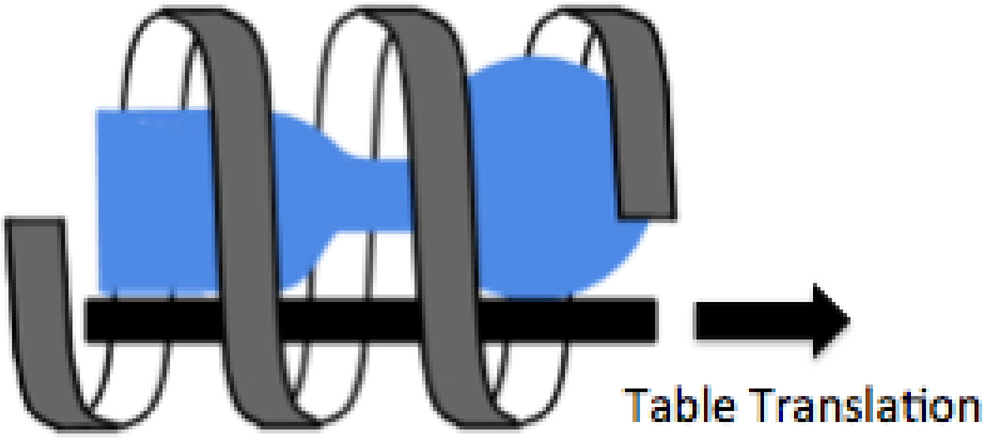

Figure 2.7: Sixth generation CT scanner (Helical CT). X-ray source and detector array rotate continuously as patient table is moved progressively through the scanner.

2.4.7 Seventh 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**. (**Bushberg et al., 2002)

2.5 Components of CT scan

2.5.1 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., 2014)

2.5.2 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.

2.5.3 Collimation and filtration

The X-ray beam should be collimated to the desired dimensions. The beam width in the longitudinal axis is generally small; therefore, the collimated X-ray beam is often referred to as a fan beam. In the plane perpendicular to the table motion, also known as the x–y or axial plane, the beam is shaped to reduce the dynamic range of the signal that is recorded by the detectors. Beam shaping (bowtie) filters are used to achieve the desired gradient, with one of a number of mounted bowtie filters moved into the X ray beam during acquisition. (Dance et al., 2014)

2.5.4 Detector

The essential physical characteristics of CT detectors are a good detection efficiency and a fast response with little afterglow. currently, solid statedetectors1 are used, as they have a 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 antiscatter 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, forming a 1-D antiscatter grid.

A detector row consists of thousands of dells that are separated by septa designed to prevent light generated in one del from being detected byneighbouringdels. These septa and the strips of the antiscatter 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., 2014)

2.6 Dosimetric Quantities

2.6.1 Fluence

The fluence, Φ, is the quotient of dN by d where dN is the number of particle incident on a sphere of cross-sectional area dA thus

$\Phi = \frac{dN}{dA}$ ……………………… (4)

Unit of fluence is m-2

2.6.2 Energy fluence

The energy fluence , is the quotient of dR by d , where dR is the radiant energy incident on a sphere of cross-sectional area dA , thus

$\Psi = \frac{dR}{dA}$……………………. (5)

The unit of energy fluence is J m-2. (Menzel el at., 2012)

2.6.3 Kerma

This nonstochastic quantity is relevant only for fields of indirectly ionizing radiations (photons or neutrons) or for any ionizing radiation source distributed within the absorbing medium. (Attix, 1986)

Kerma is defined for indirectly ionizing radiation-uncharged particles such as photons and neutrons and is related to the first step of transfer of energy from these particles to matter, in which uncharged particles transmit kinetic energy to secondary charged particles.

The kinetic energy transferred to the secondary particles is not necessarily spent in the volume dV, where they were liberated. The kerma definition is constrained to the energy the secondary particles receive at the moment of liberation.

kerma is the acronym for kinetic energy released per unit mass. It is defined as

$K = \frac{d\bar{E}tr}{dm}$…………….. (6)

Where the quantity dEtr is the expectation value of the energy transferred from indirectly ionizing radiation to charged particles in the elemental volume dV of mass dm. the SI unit of kerma is joules per kilogram (j/kg), which is given the special name Gray(Gy). (Dance et al., 2014)

2.6.4 Absorbed Dose

Owing to the penetrating character of ionizing radiation, when a large volume is irradiated, energy can be imparted to the matter in a specific volume by radiation that comes from other regions, sometimes very far from the volume of interest. As the absorbed dose includes all the contributions that impart energy in the volume of interest, it is hardly possible to correlate absorbed dose and the fluence of the incident radiation. In fact, knowledge of the radiation fluence in the volume of interest, including scattered radiation, is a necessary condition for calculation of the absorbed dose.

Absorbed dose (D), a physical non-stochastic quantity, is defined simply as the ratio

D= $d\varepsilon/dm$……………………….. (7)

The unit of absorbed dose is joule per kilogram (J/kg). The name for the unit of absorbed dose is the gray (Gy). (Dance D, 2014)

2.7 Radiation quantities

2.7.1 Organ dose

The organ dose is define as the mean dose DT in a specified tissue or organ T of the human body, given by:

$D_T = \frac{\varepsilon_T}{m_T}$ … … … … (8)

Where:

m_T Is the mass of the organ or tissue under consideration.

ε_T Is the total energy imparted by radiation to that tissue or organ.

2.7.2 Equivalent Dose

The biological detriment (harm) to an organ depends not only on the physical average dose received by the organ but also on the pattern of the dose distribution that results from the radiation type and energy. For the same dose to the organ, or neutron radiation will cause greater harm compared with gamma rays or electrons.

The organ dose is multiplied by a radiation weighting factor WR toaccount for the effectiveness of the given radiation in inducing health effects; the resulting quantity is called the equivalent dos H_T.The equivalent dose is defined as:

$H_T = W_R D$ ……………………….. (9)

Where D is the absorbed dose and $W_R$is the radiation weighting factor.

The SI unit of equivalent dose is J/kg and its name is the Sievert (Sv) The organ dose D is a measure of the energy absorption per unit mass averaged over the organ, while the equivalent dose HT is a measure of the consequent biological harm (detriment) to the organ or tissue.(Podgorsak,2005)

2.7.3 Effective dose

The effective dose, E, is defined in ICRP 60 and ICRU 51.It

is the sum over all the organs and tissues of the body of the product of the equivalent dose, H_T, to the organ or tissue and a tissue weighting factor, W_T, for that organ or tissue, thus:

$E = \sum_T \quad W_T H_T \ldots \ldots \ldots$………… (10)

The tissue weighting factor, wT for organ or tissue T represents the 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).

The sum over all the organs and tissues of the body of the tissue weighting factors, T is unity. (Technical Reports Series no. 457,2007)

The National Committee on Radiation Protection (NCRP) has identified various tissues and organs and the relative radio sensitivity of each show table below. (Boshong, 2017)

Table 2.1: Tissue weighting factors for different organs

Tissue	Tissue Weighting Factor (WT)
Gonad	0.20
Colon	0.12
Lung	0.12
Stomach	0.12
Bladder	0.05
Breast	0.05
Esophagus	0.05
Liver	0.05
Skin	0.01

2.8 CT parameter

2.8.1 X-Ray Tube Potential (kV)

There are generally three or four discrete tube potential (kV) settings available on a CT scanner, typically between 80 and 140 kV. One manufacturer has recently made a 70 kV setting available. Lower kilovoltage settings will therefore result in lower patient dose at the same mAs, but because fewer X-ray photons reach the detectors this will lead to higher image noise levels. However, lowering the kilovoltage will also increase the image contrast, particularly for materials with a high atomic number (Z), such as iodine.

Historically, a tube potential of 120 kV has been most commonly used in most routine adult scanning protocols. The relationship between dose and tube potential is not linear, but dose is approximately related to the square of the kV.

2.8.2 X-Ray Tube Current (mA) and Gantry Rotation Time (s)

The patient absorbed dose, DP is proportional to both the X-ray tube current (mA) and the gantry rotation time (s), and so these are considered together, as the tube current-gantry rotation time product (mAs).

Note that this is the mAs per rotation, and the total exposure time must also be taken into account for studies with multiple rotations

DPL α mAs…………….. (11)

The tube-current time product (mAs) directly affects the dose in a proportional relationship, and the image noise by an inverse square root relationship.

'Effective mAs' takes into account the pitch used in helical scanning, and is the effective mAs per length of patient.

2.8.3 Pitch

The pitch is a parameter that is applicable in helical (spiral) scan mode. The standard definition of pitch in CT is given in Eq :

Pitch=Table translation per rotation (mm) (I) ÷z axis- x-ray beam width (N×T) (mm)………………………………………… (12)

For a given collimation, the pitch is determined by the table speed. The advantage of using a higher pitch is that the scan is completed in a shorter time. It is often stated that increasing the pitch can be used as a method of dose reduction, as, for a fixed tube current, radiation dose is inversely proportional to pitch, due to the shorter time of radiation exposure over the given volume.

2.8.4 Scan Length (cm)

On a CT scanner the user defines the limits of the volume to be imaged. The planned scan length is defined by the start and end positions selected by the user. On some scanners the start and end are defined as the centre of the z-axis collimation, so even in sequential (axial) scan mode the irradiated scan length is longer than the planned scan length. On other scanners the start position is defined by the trailing edge of the X-ray beam, and the end position by the leading edge. In this case, in sequential scan mode the planned scan length is equal to the irradiated scan length.

The difference between these two alternative ways of defining the scan length becomes more significant for bigger z-collimations and for short scan lengths.

Scan length does not affect the absorbed dose, it is affect the total energy absorbed, the Dose Length Product and the effective dose, and therefore the risk calculated to the patient. (Tack et al., 2012)

2.9 Quantities of CT Dosimetry

Unlike a conventional diagnostic X-ray where the surface entrance radiation dose is the highest, and decreases through the patient, in CT the radiation dose is more uniformly distributed throughout a scanned object or patient since it is irradiated from all angles.

The radiation dose distribution from a CT scanner is a complex pattern determined in the scan plane by the nature of the X-ray fan beam, passing through a shaped filter, and irradiating all angles around a patient. Along the z-axis (patient axis) this distribution is determined by the spacing of the axial scans, or the spirl pitch in helical scanning.

This presents particular challenges for identifying suitable dose parameters to describe the nature of the radiation dose to a patient. The absorbed dose descriptor widely used in CT is the volume computed tomography dose index (CTDIvol) (mGy) calculated from measurements in standard phantoms. The total amount of absorbed dose from a CT examination can be characterised by taking into account the CTDIvol and the physical length of the examination, and this product is described as the DLP (mGy.cm).

Any scan parameter that affects the CTDIvol will affect DLP in the same way. Radiation dose is measured in order to obtain some information about the effect on the patient, and in order to do this the effective dose, E, Sievert, (Sv) is defined, as a measure of the risk of cancer induction in the patient from the effects of the radiation. In CT the effective dose can be estimated by the product of the CTDIvol value and the exposure length to obtain the DLP from which the related radiation risk, as measured by E, can be calculated using tabulated factors that depend upon the radiation sensitivity of the organs covered in the scan.

2.9.1 Computed Tomography Dose Index (CTDI)

The general form of the CTDI, whether measured in phantoms or air, consists of three components; the dose integral along z-axis (D(z)), the integration limits (±L/2) (where L is length of the detector active volume) and the nominal beam width (N * T), where N is the number of simultaneously acquired data (or image) slices and T is the nominal data acquisition (or slice) width:

$$CTDI = \frac{1}{N \times T} \int_{-l/2}^{+l/2} D(z)dz \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots \textit{(13)}$$

2.9.2 CTDI100

It is generally known as the CTDI100 when measured with the 100 mm pencil ion chamber, giving an integration distance of 100 mm.

$$CTDI_{100} = \frac{1}{NT} \int_{-50}^{50} D(z)dz \ldots\ldots\ldots\ldots\ldots (14)$$

2.9.3 CTDIw

It is given as the weighted average of the dose at the central position and the peripheral positions. It is weighted by one-third of the central position to two-thirds of the peripheral position .The aim is to represent the average dose across the whole of the phantom cross-section

$$CTDI_W = \frac{1}{3} CTDI_{100,center} + \frac{2}{3} CTDI_{100,peripheral} \cdots\cdots\cdots\cdots\cdots\cdots (15)$$

Where

CTDIC = CTDI100 measured in the central phantom position CTDIP = CTDI100 measured in the periphery phantom positions The concept of the CTDIw is to represent the average dose in the central slice region of a scanned volume of length 100 mm, as though the phantom were scanned with a pitch of 1 or contiguous axial slices, and can be interpreted as the multiple scanned average dose (MSAD).

2.9.4 Computed Tomography Dose Index (CTDIvol)

CTDIvol is a value of radiation dose that represents the absorbed dose (energy imparted per unit mass, generally quoted in milliGray (mGy)) to the central slice region of a scanned volume.

To give an indicator dose for volumes that are scanned with non-contiguous slices, or with a pitch P not equal to one, a correction factor is applied to the CTDIw to give the CTDIvol (mGy).

$$CTDI_{vol} = \frac{1}{pitch} \times CTDI_W \ldots\ldots\ldots\ldots\ldots\ldots \text{(16)}$$

2.9.5 The Dose Length Product (DLP)

While the CTDIvol is a measure of the absorbed dose at the central slice region of a 100 mm scanned volume, some consideration needs to be given for the extent of the patient receiving this dose.

The DLP takes into account the length of patient scanned. It is a value representing the total amount of radiation dose imparted. It is the CTDIvol multiplied by the scanned length (L), in units of mGy.cm.

DLP (mGy-cm) = *CTDI*vol (mGy) x *scan length* (cm)…… (17)

The CTDIvol can be used to compare the absorbed dose for specific protocols, however the DLP considers all aspects of the protocol and so can be used to determine the radiation risk. (Tack et al., 2012)

2.10 Effective Dose in CT

The effective dose is a “dose” parameter that reflects the risk of a nonuniform exposure in terms of a whole body exposure. It is a concept used to normalize partial body irradiations relative to whole body irradiations to enable comparisons of risk.

The calculation of effective dose requires knowledge of the dose to specific sensitive organs within the body, which are typically obtained from Monte Carlo modeling of absorbed organ doses within mathematical anthropomorphic phantoms, and recently also voxel phantoms based on real humans. Effective dose is expressed in the units of milli Sieverts (mSv).

Although effective dose calculations require specific knowledge about individual scanner characteristics, a reasonable estimate of effective dose, independent of scanner type, can be achieved using the relationship:

Effective Dose = k • DLP ………………………… (18)

Where k is a weighting factor (mSv × mGy-1 × cm-1) which depends only upon body regions. (Rehani et al., 2000)

Table 2.2: Head, neck, abdomen, or pelvis values of k (European Commission, 2000; Geleijns et al., 1994):

Region of body	k (mSv • mGy-1 • cm-1)
Head	0.0023
Neck	0.0054
Abdomen	0.015
Pelvis	0.019

2.11 System of Radiological Protection

2.11.1 Justification

No practice or source within practice should be authorized unless it produces sufficient benefits to the exposed individuals or society, to offset the radiation harm that it might cause. That is unless the practice is justified, taking into account social, economic and other relevant face.

2.11.2 Optimization

In relation to exposure from any particular source within practice protection and safety shall be optimized in order to keep the magnitude of individual doses, the number of people exposed, and the likelihood of incurring exposures as low as reasonably achievable. Economic and social factors being taken into account, within the restriction that the doses to individuals delivered by the source shall be subjected to dose constraints.

2.11.3 Dose limitation

The normal exposure of individuals shall be restricted so that
neither the total effective dose nor the total equivalent dose to relevant organs or tissues, caused by the possible combination of exposures from authorized practices, the limit on effective dose represents the level above which the risk of stochastic effects due to radiation is considered to be unacceptable.(Alhaj, 2018)

2.12 Previous Study

ZIDAN, Mojahed Mohammed Ahmed (2015) was evaluated patient radiation dose during CT KUB. Were a total of 65 adult patient undergoing CT KUB.

The radiation dose was measured in three hospitals in Khartoum state. using different CT modalities in this study, the mean of dose for 64 slice was (924.5±81.12 mGy.cm). The mean of dose for 16 slice (689.1 ±92.5 mGy.cm). The mean of dose in 4 slice was (401.96± 108.3 mGy.cm). The dose is median than that reported in previous studies.

The radiation dose higher in Al-zaytona hospital than Al-Fisal and Khartoum hospital. MSCT scanners 64 slice exposed patients to a higher dose than MSCT scanners16 slice, and MSCT scanners16 slice exposed patients to a higher dose than MSCT scanners 4 slice (ZIDAN, 2015).

ALI, Bashir Mohammed Ahmed (2016) evaluated of radiation dose for computed tomography of Kidney Ureter and Bladder Protocol. A total of 50 patients in this study were examined in Modern Medical Center. The objective of the study estimate patient radiation dose during CT K.U.B and estimate radiation risk for patient. The patient radiation dose value from this study the average DLP (292.29±29.91mGy.cm), the average CTDIvol (8.2±1.94mGy.cm), the average effective dose for pelvis region (5.55±0.57mSv) and the average effective dose for abdominal region (4.38±0.46mSv) . The average estimating radiation risk for organ the liver (for abdomen), the stomach (for abdomen), the testicle (for pelvis), the ovaries (for pelvis), and the uterus (for pelvis) were (13.15), (34.64), (11.27), (11.49), (71.68) Cancer probability per million respectively (ALI, 2016).

MOHAMMED, Moawia Ibrahim Younis (2016) was estimated Radiation dose during CT KUB . The aim of this study compare ED with ED ICRP recommend (8 mSv) and estimate radiation dose in CT KUB procedure. This study included 50 patients (25 male, 25 female) all undergoing CT KUB in Modern Medical Center.

The main effective dose by this study 4.35 mSv±2 mSv , so that The patients received acceptable dose less than international effective dose (ICRB). (MOHAMMED,2016)

CHAPTER THREE
Material and Method

3.1 Materials

3.1.1 Machines

This study intended to compare the radiation doses from different imaging modalities in different CT scan in three hospitals A, B and C.

Table 3.1: Demonstrate the hospitals and the types of machines, and number of slices used in these hospitals:

Hospitals	Manufacture	Number of slices
Hospital A	Siemens	32
Hospital B	Toshiba	64
Hospital C	Toshiba	16

3.1.2 Patient Data:

CT scans 90 of patients are used to measure the radiation dose during KUB CT in three hospitals in Khartoum state during November to February 2020.

Table 3.2: Demonstrate patients population of the study classified per type of machine:

Machines	Patient population
Thirty two slice	30
Sixty four slice	30
Sixteen slice	30
Total	90

3.2 Methods

3.2.1 Data Collection

The data were collected using a sheet for all patients and collected information from display. A data collected sheet was designed to evaluate the patient doses, it included demographic information (sex and age), scan parameters (KV, mA, scan time, number of image,slice thickness, pitch and scan length), and dosimetric information (CTDI, and DLP).

3.2.2 Dosimetric calculations

CT Expo software was used to calculate common CT dose descriptors: (i) CT weighted dose index (CTDIw), (ii) Volume dose index (CTDIv) provides an indication of the average absorbed dose in the scanned region, (iii) CT dose –length product (DLP) the integrated absorbed dose along a line parallel to the axis of rotation for complete CT examination, (vi) effective dose (E): a method for comparing patient doses from different diagnostic procedures (Effective dose) and (iiv) organ dose.

This all parameters will input to the CT-Expo Version 2.5 software. It is tool used 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).

3.2.3 Data analyses:

All dose parameters registered from the display monitor in three Centers CT/room and they input to the statistical software Excel for analyses.

CHAPTER FOUR

Results

4.1 Results

The results of this study are presented for dose measurements performed in three hospitals for 90 patients under went KUB CT examinations, the doses were estimated in terms of $CTDI_{vol}$, DLP and E, the tables below describe the results in detail.

Table (4.1): summaries the characteristic performance parameters for the CT systems and console displayed form The CT scanner in hospital A (Siemens 32 slice):

	Age (y)	**kV (kv)**	**mAs (mA.s)**	**Scan length (cm)**	**Pitch**	**CTDIvol (mGy)**	**DLP (mGy.cm)**
Mean	44.2	130	75.03	44.39	1.00	7.91	385.9
Median	43	130	73.5	45.4	1.00	7.55	360.5
STD	18.01	0	25.45	4.21	0.00	2.76	155.33
Max	78	130	131	52.2	1.00	14.08	765
Min	21	130	40	37.9	1.00	4.29	195

Table (4.2): summaries the characteristic performance parameters for the CT systems and console displayed form The CT scanner in hospital B (Toshiba 64 slice)

	Age (y)	**kV (kv)**	**mAs (mA.s)**	**Scan length (cm)**	**Pitch**	**CTDIvol (mGy)**	**DLP (mGy.cm)**
Mean	45.13	120	150	45.2	2	12.96	710.28
Median	44.5	120	150	45	2	12.7	698.3
STD	14.34	0.00	0.00	3.44	0	1.69	106.97
Max	75.00	120	150	51	2	21.9	1234
Min	23.00	120	150	36.5	2	12.2	568.7

Table (4.3): summaries the characteristic performance parameters for the CT systems and console displayed form The CT scanner in hospital C (Toshiba 16 slice)

	Age (y)	**kV (kv)**	**mAs (mA.s)**	**Scan length (cm)**	**Pitch**	**CTDIvol (mGy)**	**DLP (mGy.cm)**
Mean	49.56	120	42.12	43.04	5	6.368	299.85
Median	55	120	37	42	5	5.3	275.6
STD	16.96	0	7.03	4.28	0	1.468151	78.28
Max	84	120	52	54.5	5	10.4	565.5
Min	22	120	37	36.5	5	4.7	189.4

Table (4.4): show the estimation of mean $CTDI_W$,$CTDI_{vol}$,***DLP and effective dose (E) calculated by software expo2.5 were used data collection form three canters:***

Centers	**CTDIw (mGy)**	**CTDIvol (mGy)**	**DLP (mGy.cm)**	**E (mSv)**
Hospital A	8.39± 2.95	8.02±2.82	398.14±160	4.82±1.95
Hospital B	18.23 ±0.00	12.53 ±5.42	657.79±38.07	8.23 ±0.68
Hospital C	6.65 ±1.50	6.36±1.44	319.74±81	4.14±1.33

Table (4.5) Compare the $CTDI_{VOL}$, DLP ***and effective dose (E) between the three canters:***

Centers	**$CTDI_{VOL}$ (mGy)**	**DLP (mGy.cm)**	**Effective dose (E) (mSv)**
Hospital A	8.02±2.82	398.14±160	4.82±1.95
Hospital B	12.5 ±5.42	657.79±38.07	8.23 ±0.68
Hospital C	6.36±1.44	319.74±81	4.14±1.33

Table (4.6) showed the organ dose estimated from The CT scanner in hospital A by CT Expo 2.5 during this study:

Value	ED(liver)	ED(stomach)	ED(uterus)	ED(ovaries)	ED(Testicles)
Mean	6.21	7.54	6.73	4.99	5.17
Max	13.41	15.76	20.25	15.04	20.29
Min	1.44	1.98	0.00	0.00	0.00

Table (4.7) showed the organ dose estimated from The CT scanner in hospital B by CT Expo 2.5 during this study:

Value	ED(liver)	ED(stomach)	ED(uterus)	ED(ovaries)	ED(Testicles)
Mean	11.06	13.18	10.13	7.52	9.55
Max	16.64	17.13	21.03	15.61	18.68
Min	7.21	9.65	0.00	0.00	0.00

Table (4.8) showed the organ dose estimated from The CT scanner hospital C by CT Expo 2.5 during this study:

Value	ED(liver)	ED(stomach)	ED(uterus)	ED(ovaries)	ED(Testicles)
Mean	5.29	6.33	5.49	4.08	4.50
Max	14.41	14.59	17.23	12.71	12.77
Min	1.64	2.26	0.00	0.00	0.00

CHAPTER FIVE

Discussion, Conclusion, and Recommendation

5.1 Discussion

Patient dose in this study were measured in different CT manufacture, modalities and different protocols. The dose presented in term of $CTDI_{vol,}$ DLP and Effective dose (E) The highest dose ($CTDI_{vol}$, DLP and E) were recorded in hospital B and lower were recorded in hospital C The variable dose in three centers refer to the center used different manufacture, different modalities and different protocols.

We show that highest organ dose was recorded in hospital B and lowest organ dose was recorded in hospital C that refer to hospital B used high mAs.

Table (5.1) Comparison of CTDIvol, DLP and E values obtained in the present survey with previously published data for KUB CT examinations.

Doses	This study	Tanzania	Australia	UK	Sudan	Canada
$CTDI_{vol}$	11.09	19.0	26.6	13.5	11.6	12.0
DLP	458.6	908	1052.8	505	264	585
E	5.7	17	18.4	7.1	4	6.72

Considering the entire sample, hospitals achieved mean $CTDI_{VOL}$ values lower than similar study in other country example as (UK, Sudan, Canada, Australia and Tanzania).

The DLP values was found in this study were lower than those presented in other studies performed Tanzania, Canada, UK and Australia excepted in Sudan. Our result were lower this is mainly related to decreased scan coverage and tube-current exposure time used.

The mean value to estimated Effective dose E, in this study was lower than the corresponding values presented in UK, Tanzania, Australia, and Canada excepted in Sudan.

5.2 Conclusion

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

In this study, high effective dose and large variations of radiation dose to various organs were observed. Different manufacture, different modalities and different scanning protocols used among hospitals responsible for these variations. The mean Effective dose in this study were mostly comparable to and lower than reported values from the United Kingdom, Tanzania, Australia, and Canada but higher than Sudan. The large observed variations of Effective dose and organ doses among hospitals and relatively high effective dose and organ doses in hospital B.

5.3 Recommendation

- The high effective dose call for the need to optimize CT scanning protocols.
- This can be achieved through optimal selection of scanning parameters based on indication of study, body region of interest being scanned, and patient size.
- In addition, further studies should be done to investigate the potential for using radio protective materials to protect superficial radiosensitive organs.

Reference

Alhaj A. Estimation of effective Radiation dose during Brain Computed Tomography: Sudan of science and technology, 2018.

Attix F. introduction to Radiological physics and Radiation Dosimetry. Weinheim: Federal Republic of Germany,1986.

Bushberg J, Seibert A, Leidholdt E, The Essential Physics For Medical Imaging. second edition, 2015.

Bushing S. Radiologic Science for Technologists (physics, Radiology, and protection). Eleventh edition. St. Louis Missouri: Elsevier, 2017.

Dance D, Charistofides S, Maidment A, Mclean I, Ng K. Radiology Physics: A hand book for teachers and students. Vienna: International Atomic Energy Agency (IAEA), 2014.

Hill D. Principles of Diagnostic X-ray Apparatus. London: The macmillan press LTD, 1979.

Technical Reports Series no. 457 Dosimetry in Diagnostic Radiology: An international code of practice. Vienna: International Atomic Energy Agency (IAEA), 2007.

Thayalan K. Basic Radiological physics. First edition. India: Jaypee brothers Medical publisher (P) Ltd, 2001.

Moorin R, Gibson D. Radiation dosimetry assessment of routine CT scanning protocols, Australia ;295.

Menzel H et al. Radiation dose and image Quality Assessment in Computed Tomography. Journal of ICRU 2012; 12: 37.

Podgorsak E B. Radiation Oncology physics: A Hand book for teachers and students. Vienna: International Atomic Energy Agency (IAEA), 2005.

Rehani M, Kalra M, McCollough C, Nagel H. Managing patient Dose in Multi-Detector Computed Tomography (MDCT).Germany, 2000.

Seeram E. Computed Tomography physical principle, clinical Applications, and Quality Control. Second edition. Canada, 2000.

Tack D, Reser M, Hricak H, Knauth H. Radiation Dose from Multi detector CT. Second edition. Berlin, 2012.