Sudan University of Science and Technology

College of Graduate Studies

كلية الدراسات العليا

Evaluation of Patient Effective Dose and Organ Dose during Chest Computed Tomography

تقييم الجرعة الإشعاعية الفعالة لأعضاء جسم المريض أثناء التصوير المقطعي للصدر

A Thesis Submitted for partial fulfillment of the requirements of M.Sc degree in medical physics

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2019

الآيـة

قال تعالى:

﴿وَمَا تَوْفِيقِي إِلَّا بِاللَّهِ عَلَيْهِ تَوَكَّلْتُ وَإِلَيْهِ أُنِيبُ﴾

صدق الله العظيم

سورة هودالآية(88)

Dedication

To my parents for their support along the whole year.

To my sister and brother for supporting me spiritually.

To my friends for being there whenever I needed them.

And I can never forget my doctors who supported me more than I can ever ask for and for being there for every consultation, question and self-doubt.

ACKNOWLEDGEMENT

Sure I would like to thank Allah first for making such work possible.

Also I would like to thank my supervisor **Dr. Hussein Ahmad Hassan** for his precious guidance, support and advices which helped me making this work possible.

I would also like to appreciate the great help of all radiological department staff and the students of the College of medical radiological science for help and support.

Abstract

The aim of this study to evaluate of patient effective dose and radiation doses received by critical organs of the patients undergoing chest CT examination.

The study was done on 61 patients (31 males and 30 females) in age range between 16 to 90 years , who came for routine chest multi-detector computed tomography model Toshiba Aquilition (16 slice) and CT Toshiba Alixon (16 slice) in two hospitals, in Khartoum state, namely (Alzaytouna and Yasatpshoron), during the period from September 2018 to February 2019. The parameters collected of the data (KVP, mAS, age, gender, slice thickness, total scan time, pitch, scan length, $CTDI_{vol}$, and DLP), and calculated $CTDI_{vol}$, DLP, E, and organ doses by CT expo version 2.5 software and the data were analyzed using Excel.

The results of this study were the means of critical organ dose for hospital (Y) which were (12.8± 0.04)mSv, (12.4± 0.1) mSv, (6.7 ± 4.6) mSv for Breast, Lung, Thyroid respectively, and the means of critical organ dose for hospital (Z) were (27.8 ± 0.04) mSv, (27.2 ± 0.1) mSv, (38.8± 2.9) mSv, Breast, Lung, Thyroid respectively. The means of effective dose in hospital (Y) and hospital (Z) were (5.3 ± 1.04) mSv, (14.2 ± 1.8) mSv respectively.

From which it could be conclude the radiation dose received by the critical organs of the patients undergoing CT chest examination at hospital (Z) is higher than radiation dose received by the critical organs of the patients undergoing CT chest examination at hospital (Y), and the mean of effective dose for patients at hospital (Z) is higher than the mean of effective dose for patients at hospital (Y).

The effective dose in this study was compared with different reported values from Tanzania, Taiwan, United Kingdom, and EC reference and it was the lowest effective dose.

مستخلص الدراسة

الهدف من هذه الدراسة تقييم الجرعة الإشعاعية الفعالة للمريض والجرعة الاشعاعية المستقبلة بواسطة الاعضاء الحساسة في جسم المريض أثناء التصوير المقطعي للصدر.

أجريت هذه الدراسة على 61 مريضاً (31 ذكراً و 30 أنثى) في المدى العمري ما بين 16 إلى 90 سنة، للمرضى الذين خضعوا للفحص الروتيني للصدر عن طريق الأشعة المقطعية ذات الكواشف المتعددة، خلال الفترة الزمنية ما بين سبتمبر 2018 الى فبراير 2019. في ولاية الخرطوم في مستشفيي (يستبشرون والزيتونة).

الأجهزة المستخدمة في هذه الدراسة تشمل توشيبا16 شريحة موديل (Aquilition) و توشيبا 16 شريحة موديل(Alixon)، البيانات التي جمعت تشمل (نوع وعمر المريض ، عوامل التعريض : الكيلو فولت ،التيار ، عرض الشريحة ، الزمن الكلي للفحص ، طول مقطع المسح ، عوامل لها علاقة بالجرعة الإشعاعية : ($CTDI_{vol}$, DLP) وتم إستخدام هذه البيانات لتقييم الجرعة الإشعاعية الفعالة للمرضي عن طريق معالج البيانات CT Expo 2.5.

أظهرت هذه الدراسة أن متوسط الجرعة الإشعاعية الواصلة للأعضاء الأكثر حساسية للإشعاع في منطقة الصدر : الثدي، الرئة، الغدة الدرقية المتحصل عليها من مستشفي (Y)هي:mSv(±12.8 0.04)،mSv(0.1 ± 12.4) وmSv(4.6 ± 6.7) على التوالي ، ومتوسط الجرعة الإشعاعية الواصلة للأعضاء الأكثر حساسية للإشعاع في منطقة الصدر: الثدي، الرئة، الغدة الدرقية المتحصل عليها من مستشفي (Z)هي:mSv(0.04 ± 27.8)،mSv(0.1 ± 27.2) وmSv(2.9 ± 38.8) على التوالي . ولقد وجد أن متوسط الجرعة الإشعاعية الفعالة للمرضى الذين خضعوا لفحص الصدر بواسطةجهاز الاشعة المقطعية توشيبا 16شريحة من مستشفيي (Y)و(Z)هيmSv(14.2±1.8)وmSv (5.3±1.04) على التوالي .

ولقد وجد في هذه الدراسة أن متوسط الجرعة الإشعاعية الفعالة للمرضى الذين خضعوا لفحص الصدر بواسطة جهاز الأشعة المقطعية في مستشفي (Z)أعلى من الجرعة الإشعاعية الفعالة للمرضى الذين خضعوا لفحص الصدر بواسطة جهاز الأشعة المقطعية في مستشفي(Y).

الجرعة الفعالة في هذه الدراسة قورنت مع نتائج دراسات عالمية من تنزانيا، تايوان، المملكة المتحدة و المفوضية الاوربية وكانت اقل جرعة فعالة.

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List of Abbreviations

CT	Computed Tomography
CAT	Computed Axial Tomography
DAS	Data Acquisition System
MSAD	Multiple Scan Average Dose
MDCT	Multi Detector Computed Tomography
CTDI	Computed Tomography Dose Index
CTDIw	Weighted Computed Tomography dose index
CTDIvol	Volumetric Computed Tomography dose index
DLP	Dose Length Product
D	Absorbed Dose
E	Effective Dose
H_T	Dose Equivalent
Eq	Equivalent Dose
W_R	Radiation Weighting Factor
W_T	Tissue Weighting Factor
Q	Quality Factor
D_T	Organ Dose
K	Karma
Kerma	Kinetic Energy Released per unit Mass
KVp	Kilo Voltage Peak
MAs	Mile Amber Second
MSv	Mile Sievert
MGy	Mile Gray
FDA	Food and Drugs administration
DRL_S	Diagnostic reference levels

ICRP	International Commission on Radiation Protection
$NDRL_S$	National Dose reference levels
EC	European Commission
UK	United Kingdom
Max	Maximum
Min	Minimum

Chapter One

Introduction

Chapter One

Introduction

1.1 Introduction:

Computed Tomography (CT) is an image modality that has played a significant role in detection of human diseases and perform an image guided interventions. Radiation dose in CT is so high and differ from that in conventional x-ray, so when the patient regularly exposed to high radiation during CT examination or for a long time this may result in development of serious health effects for examples: cancer, hereditary effects, depilation and pericardites …etc. (Lois, 2011).

Computed tomography (CT) technology offers high-quality images and valuable diagnostic results, especially for the management of lung and heart diseases. Application of CT exams is increasing in hospitals as they provide more accurate diagnoses and screening of lung and heart diseases, but radiation doses are increased in organs in the scanning field during CT imaging. (Janbabanezhad,2015).

Sensitive organs to ionizing radiation during thorax CT are breast, lung and heart. Breast is superficial and more sensitive than lung or heart to radiation doses in women. This issue demands a re-evaluation of female CT imaging processes and techniques. So the Brest is a sensitive organ and targeting for radiation during CT chest examinations, it may receive a considerable radiation dose from the radiation. So there is substantially increased risk of breast carcinoma in persons who had a history of irradiation either from diagnostic or therapeutic methods. (Angel, 2009).

The thyroid gland is a sensitive organ and targeting for radiation during CT chest examinations, it may receive a considerable radiation dose from the scatter radiation. So there is substantially increased risk of thyroid carcinoma in persons who had a history of irradiation either from diagnostic or therapeutic methods. (Maria, 2017).

Computed Tomography (CT) dosimetry concept is an important principle for dose measurements, and participate effectively in informing both the public and other hospitals personnel about the dose in CT. (Lois, 2011).

Estimation of the effective dose of critical organ from Computed Tomography (CT) chest examination help in providing good information about the dose that affect it and expect the radiation risk which is result commonly in cancer and another health effects.(Maria, 2017).

1.2 Problem of the Study

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

1.3 Objectives of the study

1.3.1 General objective

To evaluate the patient effective dose and organ dose during chest computed tomography.

1.3.2 Specific objectives

To calculate effective dose for patients during CT chest examinations.

To estimate dose for critical organs.

To Compare Effective and organ doses between two hospitals.

To Compare results with international studies.

1.4 Thesis outline

This thesis is concerned with the assessment of radiation dose for patients during CT chest imaging. Accordingly, it is divided into following the chapters:

Chapter one is the introduction to this thesis. This chapter discusses the objectives and scope of work and introduces necessary background. It also provides an outline of the thesis.

Chapter two contains the background for the thesis. Specifically it discusses the dose for all absorbed dose measurements and calculations. This chapter also includes a summary of previous work performed in this field.

Chapter three describes the materials and methods used to measure dose for CT machines and explain in details the methods used for dose calculation.

Chapter four presents the results of this study.

Chapter five presents the discussion, conclusion and recommendations as well as suggestions for future work.

Chapter Two

Literature Review

Chapter Two

Literature Review

2.1 Theoretical background

2.1.1 Computed Tomography (CT) Imaging:

CT imaging, also referred to as a computed axial tomography (CAT) scan, and involves the use of rotating x-ray equipment, combined with a digital computer, to obtain images of the body. Using CT imaging, cross sectional images of body organs and tissues can be produced. Though there are many other imaging techniques, CT imaging has the unique ability to offer clear images of different types of tissue. CT imaging can provide views of soft tissue, bone, muscle, and blood vessels, without sacrificing clarity. Other imaging techniques are much more limited in the types of images they can provide. To understand the difference between CT imaging and other techniques, consider an x-ray of the head. Using basic x-ray techniques, the bone structures of the skull can be viewed. With magnetic resonance imaging (MRI), blood vessels and soft tissue can be viewed, but clear, detailed images of bony structures cannot be obtained. On the other hand, x-ray angiography can provide a look at the blood vessels of the head, but not soft tissue. CT imaging of the head can provide clear images not only of soft tissue, but also of bones and blood vessels. CT imaging is commonly used for diagnostic purposes. In fact, it is a chief imaging method used in diagnosing a variety of cancers, including those affecting the lungs, pancreas, and liver. Using CT imaging, not only can physicians confirm that tumors exist, but they can also pinpoint their locations, accurately measure the size of tumors, and determine whether or not they've spread to neighboring tissues. In addition to the diagnosis of certain cancers, CT imaging is used for planning and administering radiation cancer treatments, as well as for planning certain types of surgeries. Thanks to its ability to provide clear images of bone, muscle, and blood vessels, CT imaging is a valuable tool for the diagnosis and treatment of musculoskeletal disorders and injuries. It is often used to measure bone mineral density and to detect injuries to internal organs. CT imaging is even used for the diagnosis and treatment of certain vascular diseases that, undetected and untreated, have the potential to cause renal failure, stroke, or death. CT imaging is pain-free. Some individuals, however, may experience some discomfort because of the requirement to lie still for a period of time. The procedure is typically performed by a well trained CT technologist.(Niki Foster,2012).

2.1.2 CT Chest

When a CT scan is directed to the chest, the individual receives the equivalent radiation of 30 to 442 chest X-rays. Recent modeling estimates that use of chest CTs and CT angiography in 2007 alone will lead to an additional 5,300 cases of lung and breast cancer within the next two to three decades. Other modeling suggests that 1 in 150 women who are 20 years old when they undergo CT angiograms of the chest, and 1 in 270 women (total) having the procedure, will subsequently develop cancers of the chest, including breast cancer. When taking a look at the influence of ionizing radiation on breast cancer risk, a number of studies can be found which evaluated the impact on the possibility of malignant tumor development in women exposed to high doses. The incidence of breast cancer in atom bomb survivors, in women subjected to multiple fluoroscopic examinations or treated by radiotherapy was found to be increased.

Accordingly, a high radiation sensitivity of breast tissue was stated. This is also reflected in the new assignments of relative tissue risk factors by the International Commission of Radiation Protection (ICRP): the weight factor for the breast was increased from 0.05 to 0.12 in the newest ICRP report .In diagnostic thoracic computed tomography (CT) the breast is always included in the scan field, but it is rarely or never the organ of interest.

Given the radiation sensitivity of the glandular tissue and the fact that the breast is not the object of interest in imaging. In CT chest, the breast mean glandular dose is up to 5cGy.(Redberg,2009).

2.1.3 Radiation Risk during CT Chest

There is no doubt that many patient have benefited from the rapid Diagnoses made possible by CT and from its value for monitoring chronic disease. However is increasing concern regarding the risk of this Exposure to radiation. It is well established that radiation can be harmful and has both deterministic and stochastic effects.

Deterministic effect, Such as hair loss, skin bums, and cell death, are dose dependent but do not occur below a threshold of 150-200 mSv.

Since the typical estimated Dose associated with proper use of CT is in the range of 2-10 mSv, Deterministic effects are not normally a concern. Induction of cancer by Radiation is a probabilistic (stochastic) effect, not a deterministic effect. That is, higher radiation doses are associated with a higher likelihood of Carcinogenesis, but even low doses of radiation could potentially induce Carcinogenesis and it is more difficult to assess a safe level of exposure.(Redberg, 2009).

2.1.4 CT machine (components)

The general structure of CT equipment can be divided in three principle elements:

-The Data Acquisition and Transfer system, which encompasses the gantry, the patient’s table, the power distribution unit and the data transfer unit.

-The computing System (or operator’s console) is installed in separate room, making it possible for the operator (technician) to control the acquisition process, introducing patient data and selecting several acquisition parameters such as the kVp , mA values the protocol is going to use. Also there is another operator’s console for editing and postprocessing is also necessary, so it possible to analyze and review previous exam data, without interfering with the current examinations taking place.

-The image reconstruction system: receives the X-ray transmission data information from the data transfer unit, in a digital format. This gathered data is then corrected to using reconstruction algorithms and later stored. (Cattin, 2010).

Fig (2.1): CT scanner

2.1.4.1 CT Gantry

the first major component of a CT system is referred to as the scan or imaging system. The gantry is a moveable frame that contains the x-ray tube including collimators and filters, detectors, data acquisition system (DAS), rotational components including slip ring systems and all associated electronics such as gantry angulation motors and positioning laser lights. A CT gantry by can be angled up to 30 degrees toward a forward or backward position. Gantry angulation is determined by the manufacturer and varies among CT systems. The opening through which a patient passes is referred to as the gantry aperture. Gantry aperture diameters generally range from 50-85 cm. Lasers or high intensity lights are included within or mounted on the gantry. The lasers or high intensity lights serve as anatomical positioning guides that reference the center of the axial, coronal, and sigittal planes. (Bushong,1993).

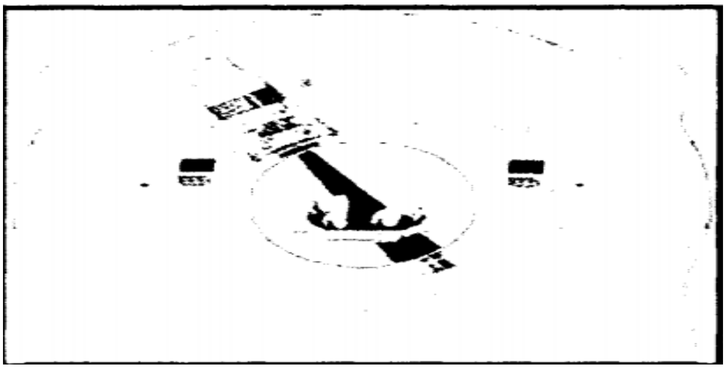


Fig (2.2): CT Gantry

2.1.4.2 X-ray Tube Collimation, Filtration

CT procedures facilitate the use of large exposure factors, (high mA and kVp values) and short exposure times. CT systems produce x-radiation continuously or in short millisecond bursts or pulses. CT x-ray tubes must possess a high heat capacity which is the amount of heat that a tube can store without operational damage to the tube. Modem CT systems utilize x-ray tubes that have a heat capacity of approximately 3.5 to 5million heat units (MHU). Many CT x-ray tubes utilize a combination of oil and air cooling systems to eliminate heat and maintain continuous operational capabilities. ACT x-ray tube anode has a large diameter with a graphite backing. The large diameter backed with graphite allows the anode to absorb and dissipate large amounts of heat. CT tubes utilize a bigger filament than conventional radiography x-ray tubes. The use of a bigger filament increases the size of the effective focal spot decreasing the anode or target angle decreases the size of the effective focal spot. CT tubes employ a target angle approximately between 7 and 10 degrees. The decreased anode or target angle also helps elevate some of the effects caused by the heel effect. CT can compensate any loss of resolution due the use of larger focal spot sizes by employing resolution enhancement algorithms such as bone or sharp algorithms, targeting techniques, and decreasing section thickness.
In CT collimation of the x-ray beam includes tube collimators, a set of pre patient collimators and post-patient or pre-detector collimators. Some CT

systems utilize this type of collimation system while other does not. The tube or source collimators are located in the x-ray tube and determine the section thickness that will be utilized for particular CT scanning procedure. A second set of collimators located directly below the tube collimators maintain the width of the beam as it travels toward the patient. A final set of collimators called post-patient or pre-detector collimators are located below the patient and above the detector. The primary responsibilities of this set of collimators are to insure proper beam width at the detector and reduce the number of scattered photons that may enter a detector.

There are two types of filtration utilized in CT. Mathematical filters such as bone or soft tissue algorithms are included into the CT reconstruction process to enhance resolution of a particular anatomical region of interest. Inherent tube filtration and filters made of aluminum or Teflon are utilized in CT to shape the beam intensity by filter in gout low energy photons that contribute to the production of scatter. Special filters called “bow-tie" filters absorb low energy photons before reaching the patient. Heavy filtration of the x-ray beam results in a more uniform beam. The more uniform the beam, the more accurate the attenuation values or CT numbers are for the scanned anatomical region.(Bushong, 1993).

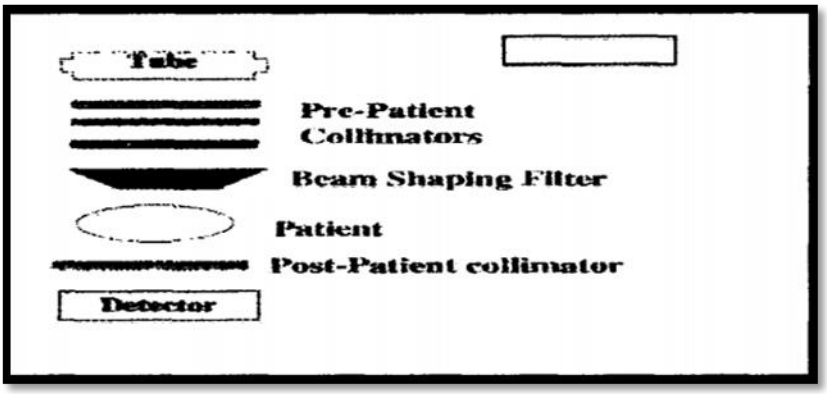


Fig (2.3): Image acquisitions in CT

2.1.4.3 Detectors

When the x-ray beam travels through the patient, it is attenuated by the anatomical Structures it passes through. The image receptors that are utilized in CT are referred to as detectors. The CT process essentially relies on collecting attenuated photon energy and converting it to an electrical signal, which will then be converted to a digital signal for computer reconstruction. The two types of detectors utilized in CT systems are scintillation or solid state and xenon gas detectors. Scintillation detectors utilize a crystal that fluoresces when struck by an x-ray photon which produces light energy. A photodiode is attached to the scintillation portion o f the detector. The photodiode transforms the light energy into electrical or analog energy. The most frequently used scintillation crystals are made o f Bismuth Germinate (Bi4Ge3012) and Cadmium Tungstate (CdW 04).

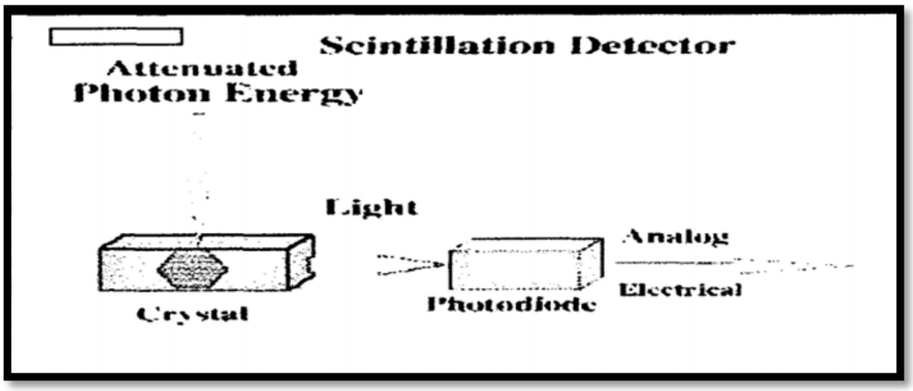


Fig (2.4): Scintillation detector image acquisition in CT

The second type of detector utilized for CT imaging system is a gas detector. The gas detector is usually constructed utilizing a chamber made of a ceramic material with long thin ionization plates usually made of Tungsten submersed in Xenon gas. The long thin tungsten plates act as electron collection plates. When attenuated photons interact with the charged plates and the xenon gas ionization occurs. The ionization of ions produces an electrical current. Xenon gas is the element o f choice because of its ability to remain stable under extreme amounts of pressure. The term detector refers to a single element or a single type of detector used in a CT system. The term detector array is used to describe the total number of detectors that a CT system utilizes for collecting attenuated information. (Wolbarst, Cattin,1993).

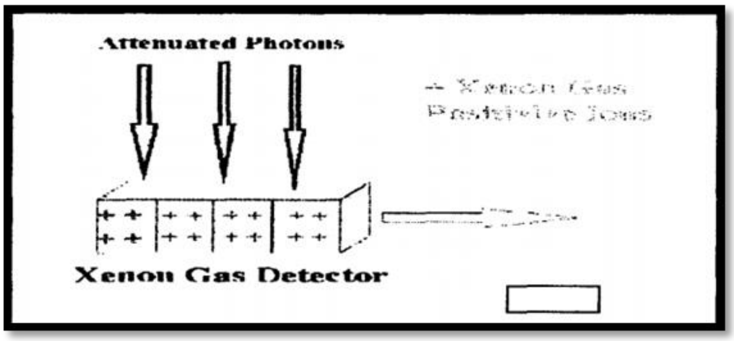


Fig (2.5): Xenon Gas detectors

The path that an x-ray beam travels from the tube to a single detector is referred to as a ray. After the x-ray beam passes through the object being scanned, the detector samples the beams intensity. The detector reads each ray and measures the resultant beam attenuation. The attenuation measurement of each ray is termed a ray sum. A complete set of ray sums is referred to as a view or projection. It takes many views to create a computed tomography image. The more photons collected, the stronger and more accurate the detector signal. This is essential for accurate image reconstruction. The dynamic range determines the ability of a detector to detect and differentiate wide range of x-ray intensities. Dynamic range o f a detector describes the range of x-ray exposures at the detector to which the system can respond without saturation and produce satisfactory gray-scale images. CT systems have the ability to respond to 1,000,000 x-ray intensities at approximately 1,100 views per second. (Morgan,1983)

2.1.4.4 Data Acquisition System (DAS)

Once the detector generates the analog or electrical signal it is directed to the data acquisition system (DAS). The analog signal generated by the detector is a weak signal and must be amplified to further be analyzed. Amplifying the electrical signal is one of the tasks performed by the data acquisition system (DAS). (Seeram, 1994).

The DAS is located in the gantry right after or above the detector system. In some modem CT scanning systems the signal amplification occurs within the detector itself. Before the projection or raw data, which is currently in the form of an electrical or analog signal, goes to the computer it must be converted to digital information. The computer does not "understand" analog signals therefore; the information must be converted to digital information. This task is accomplished by an analog to digital converter which is an essential component of the DAS. The digital signal is transferred to an array processor. The array processor solves the statistical information using algorithmic calculations essential for mathematical reconstruction of a CT image. An array processor is a specialized high speed computer designed to execute mathematical algorithms for the purpose of reconstruction. The array processor solves reconstruction mathematics faster than a standard microprocessor. It is important to note that special algorithms may require several seconds to several minutes for a standard microprocessor to compute. Recently, processors that compute CT reconstruction mathematics faster than an array processor have been utilized to solve reconstruction mathematics essential to the development o f CT fluoroscopy. The term image or reconstruction generator is used to describe this type of computer.(Berland,1987).

2.1.4.5CT Patient Table or Couch

The final component o f the scan or imaging system is the patient table or couch. CT tables or couches should be made with a material that will not cause artifacts when scanned. Many CT tables or couches are made o f a carbon fiber material. Various attachments are available for different types of scanning procedures. Attachments for direct coronal scanning and therapy planning are commonly used in many CT departments. (Bushong,1993).

2.1.4.6 Power Distribution Unit

The power Distribution unit supplies power to the gantry, the patient's table and the computers of the Computing System. (Bushong, 1993).

2.1.5 CT generations

2.1.5.1 First Generation: Rotate/translate, Pencil beam

CT scanners represent a marriage of diverse technologies, including computer hardware, motor control systems, x-ray detectors, sophisticated reconstruction algorithms, and x-ray tube/generator systems. 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. The acquisition of the numerous projections and the multiple rays per projection required that the single detector for each CT slice be physically moved throughout all the necessary positions. (Bushberg et al., 2003)

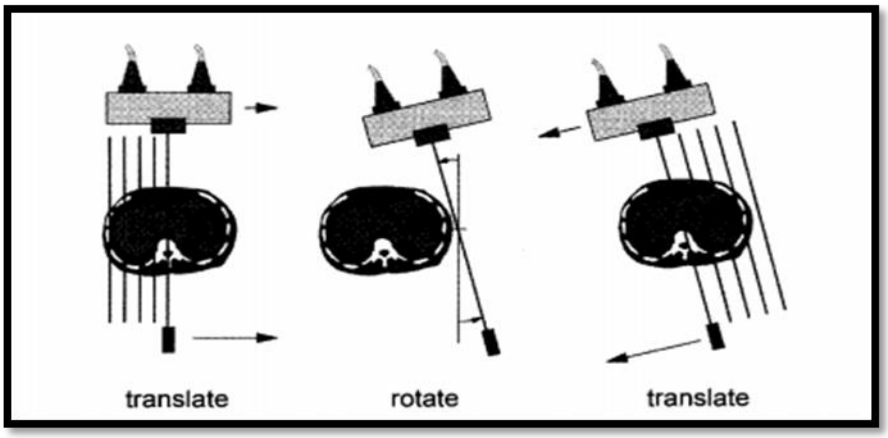
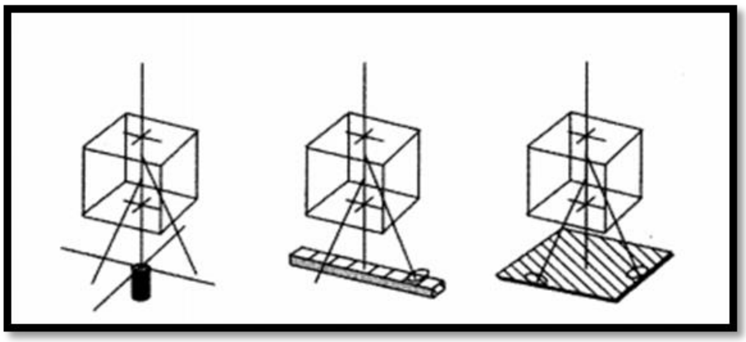


Fig (2.6):First-generation (rotate/translate) computed tomography CT. The x-ray tube and a single detector translate across the field of view, producing a series of parallel rays. The system then rotates slightly and translates back across the field of view, producing ray measurements at a different angle. This processes repeated at 1-degree intervals over 180 degrees, resulting in the complete CT data set.(Bushberg et al., 2003)

2.1.5.2 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 30times, 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. Incorporating an array of detectors, instead of just two, required the use of an arrow fan beam of radiation. Although a narrow fan beam provides excellent scatter rejection compared with plain film imaging, it does allow more scattered radiation to be detected than was the case with the pencil beam used in first-generation CT.(Bushberg et al., 2003)



Pencil beam   Fan beam   Open beam

Fig (2.7): The difference between pencil beam, fan beam, and open beam geometry in terms of scatter detection.(Bushberg et al., 2003)

2.1.5.3 Third Generation: Rotate/Rotate, Wide Fan Beam

The translational motion of first- and second-generation CT scanners was a principal obstacle to fast scanning. At the end of each translation, the movement of the x-ray tube system had to be hinder, the whole system rotated, and the translational movement restarted. The attaining prosperity of

CT as a clinical modality in its infancy gave craftsman reason to explore more capable, but more costly approaches to the scanning geometry the motion of third-generation CT is "Rotate/Rotate" referring to the motion of the x-ray tube and the movement of the detector array. By elimination of the translational motion the scan time is decreased virtually. The early third-generation scanners could deliver scan times shorter than 5 seconds. Newer systems have scan times of one half second. The development from first- to second- and second- to third-generation scanners included radical enhancement 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 scanners 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.(Bushberg et al., 2003)

2.1.5.4 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.

It is never possible to have a large number of detectors in perfect balance with each other, and this was especially true 25 years ago. Each detector and its associated electronics have a certain amount of drift, causing the signal levels from each detector to shift over time.

The rotate/rotate geometry of third-generation scanners leads to a situation in which each detector is responsible for the data corresponding to a ring in the image. Detectors toward the center of the detector array provide data in the reconstructed image in a ring that is small in diameter, and more peripheral detectors contribute to larger diameter rings. Fourth-generation CT scanners were designed to overcome the problem of ring artifacts. With fourth-generation scanners, the detectors are removed from the rotating gantry and are placed in a stationary 360-degree ring around the patient requiring many more detectors. Modern fourth-generation CT systems use about 4,800 individual detectors. Because the x-ray tube rotates and the detectors are stationary, fourth-generation CT is said to use a rotate/stationary geometry.(Bushberg et al., 2003)

2.1.5.5 Fifth Generation (Stationary/Stationary)

A new CT scanner has been promoted specifically for cardiac tomographic imaging. This "cine-CT" scanner does not use a imitative x-ray tube; instead, a large arc of tungsten encircles the patient and He's directly positioned in the other side to the detector ring.(Bushberg et al., 2003)

2.1.5.6 Sixth Generation: Helical

Third-generation and fourth-generation CT geometries solved the mechanical inertia limitations involved in acquisition of the individual projection data by eliminating the translation motion used in first- and second-generation scanners.

However, the gantry had been stopped after each slice was acquired, because the detectors (in third-generation scanners) and the x-ray tube (in third- and fourth-generation machines) had to be connected by wires to the stationary scanner electronics. Helical CT acquire data while the table is moving; as a result, the x-ray source moves in a helical pattern around the patient being scanned.(Bushberg et al., 2003)

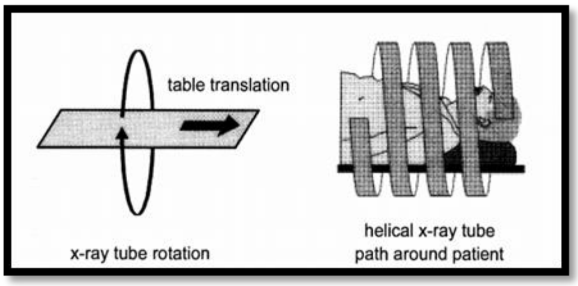


Fig (2.8): helical computed tomographic scanners.

the x-ray tube rotates around the patient while the patient and the table are translated through the gantry.(Bushberg et al., 2003)

2.1.5.7 Seven Generation: Multiple Detector Array

X-ray tubes designed for CT have effective heat storage and cooling capacity although the immediate production of x-rays is limited 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. The collimator spacing is wider when multiple detector arrays are utilized 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 decreases spatial resolution in the slice thickness dimension. The slice thickness is not determined by the collimator but by the detector size. This represents a significant shift in CT technology.

The elasticity of CT acquirement protocols and the more efficiency ensuing from multiple detector array CT scanners provides for better patient imaging, however, the number of parameters involved in the CT acquisition protocol is increased as well.(Bushberg et al., 2003)

2.1.6 Chest CT scan

CT scanning of the chest uses special equipment to obtain multiple cross-sectional images o f the organs and tissues of the chest. CT produces images that are far more detailed than a conventional chest x-ray. CT is especially useful because it can simultaneously show many different types of tissue including the lungs, heart, bones, soft tissues, muscle and blood vessels(Glassberg,2013). Using a variety o f techniques, includingadjusting the radiation dose based on patient size and new software technology, the amount o f radiation needed to perform a chest CT scan can be significantly reduced. A low-dose chest CT produces images o f sufficient image quality to detect many lung diseases and abnormalities using up to 65 percent less ionizing radiation than a conventional chest CT scan. This is especially true for detecting and following lung cancer. Other diseases, such as the detection of pulmonary embolism and interstitial lung disease may not be appropriate for low-dose chest CT. Radiologist will decide the proper settings to be used for your scan depending on your medical problems and what information is needed from the CT scan. If your child is to have a CT scan, the proper low-dose pediatric settings should be used (American college,2012). To produce high-quality scans at a lower radiation dose, low-dose CT scanning uses a variety of techniques, including:

- Dose modulation, in which radiation dosage is continuously adjusted to the patient's size at each location as the patient moves through the scanner.
- "Noise management" software to filter out unnecessary data.
- The use of shields (this method depends on the type of CT scannerbeing used).
- External shields made out of bismuth may be placed on the patient.
- The x-ray tube may be turned off during part of its rotation.
- Lower peak voltage settings. (American college,2012).

2.1.7 Radiation Quantities and Units

Several forms of ionizing radiation are used in medical imaging. Even though the risk is low, if there is a risk at all, it is appropriate to manage the radiation delivered to patients being imaged and to use only sufficient radiation to produce the necessary image quality. The question we begin with is: How much radiation is delivered to a patient's body? As w e are about to see, that is not always an easy question to answer. There are several factors contributing to the complexity. They include the many quantities that can be used to express the amount of radiation, the different units that are used, and the generally uneven distribution of the radiation within the patient's body. (Perry Sprawls,1993).

2.1.7.1 Radiation Units

Throughout the course of history there have been many different systems of units developed to express the values of the various physical quantities. In more recent times the metric system has gradually replaced some of the other more traditional or classic systems, this is also true for the units used for many of our radiation quantities. (Perry Sprawls,1993).

2.1.7.2 Radiation Quantities

There are many different physical quantities that can be used to express the amount of radiation delivered to a human body. Generally, there are advantages and applications as well as disadvantages and limitations for each of the quantities. There are two types of radiation quantities: those that express the concentration of radiation at some point, or to a specific tissue or organ, and there are also quantities that express the total radiation delivered to a body. I will be considering each of these quantities in much more detail. (Perry Sprawls,1993).

2.1.7.2.1 Exposure

Exposure is the quantity most commonly used to express the amount of radiation delivered to a point. The conventional unit for exposure is the roentgen (R). And the SI unit is the coulomb per kilogram of air (C/kg):

$$\mathbf{1\ R = 2.58 \times 10^{-4}\ C/kg.}$$

$$\mathbf{1\ C/kg = 3876\ R.}$$

The specific effect used to measure exposure is the ionization in air produced bv the radiation. Exposure is generally measured by placing a small volume of air at the point of measurement and then measuring the amount of ionization produced within the air.

Exposure is a quantity of radiation concentration. For specific photon energy, exposure is proportional to photon concentration or flounce, (PerrySprawls,1993).

1 R = 2.58 x 1 O'4 C/kg.

2.1.7.2.2 Absorbed Dose

Absorbed dose is the quantity that expresses the concentration of radiation energy absorbed at a specific point within the body tissue. Since an x-ray beam is attenuated by absorption as it passes through the body, all tissues within the beam will not absorb the same dose. The absorbed dose will be much greater for the tissues near the entrance surface than for those deeper within the body. Absorbed dose is defined as the quantity of radiation energy absorbed (ε) per unit mass of tissue (m):

$$D = \frac{d\varepsilon}{dm}$$

The conventional unit for absorbed dose is the rad. which is equivalent to 100 ergs of absorbed energy per g of tissue. The SI unit is the gray (Gy), which is equivalent to the absorption of 1 J of radiation energy' per kg of tissue. (Perry Sprawls,1993). The relationship between the two units is:

1 rad = 100 erg/g = 0.01 J/kg = 0.01 Gy.

1 Gy= 100 rad.

2.1.8 Radiation protection quantities

The absorbed dose is the basic physical dosimetry quantity, but it is not entirely satisfactory for radiation protection purposes because the effectiveness in damaging human tissue differs for different types of ionizing radiation. In addition to the physical quantities, other dose related quantities have been introduced to account not only for the physical effects but also for the biological effects of radiation upon tissues. These quantities are organ dose, equivalent dose, effective dose, committed dose and collective dose.(Podgorsak, 2005).

2.1.8.1 Organ dose

Organ dose (D_T) is defined as mean dose in a specified tissue or organ T of the human body, given by:

$$D_T = \frac{\varepsilon_T}{m_T}$$

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.(Podgorsak, 2005).

Quality Factor

Quality factor (Q)is accounts for the different biological effects produced by the different types of ionizing radiation such as the diagnostic x ray used in CT where it equals (1).(Lois, 2011).

2.1.8.2 Dose Equivalent

Dose equivalent (H) is the quantity commonly used to express the biological impact of radiation on persons receiving occupational or environmental exposures. Personnel exposure in a clinical facility is often determined and recorded as a dose equivalent.

Dose equivalent is proportional to the absorbed dose (D). The quality factor (Q). And other modifying factors (N) of the specific type of radiation. Most radiations encountered in diagnostic procedures (x-ray? gamma, and beta) have quality and modifying factor values of l. Therefore, the dose equivalent is numerically equal to the absorbed dose. Some radiation types consisting of large (relative to electrons) particles have quality factor values greater than 1. For example, alpha particles have a quality factor value of approximately 20, the conventional unit for dose equivalent is the rem, and the SI unit is the sievert (Sv). When the quality factor is (1), the different relationships between dose equivalent (H) and absorbed dose (D) are:

$$\mathbf{H(rem) = D(rad).}$$

$$\mathbf{H(Sv) = D(Gv).}$$

Dose equivalent values can be converted from one system of units to the other by: (Perry Sprawls,1993).

$$\mathbf{1\ Sv = 100\ rem.}$$

2.1.8.3Equivalent Dose

Equivalent dose (Eq)is a similar dose to the dose equivalent but a newer one and it equals the product of the absorbed dose and the radiation weighting factor (W_R), it's also measured in the same units of rem and Sv. (Lois, 2011).

$$Eq = D \times W_R$$

Another definition is: the attempts to account the particular effects to the patient's tissues that have absorbed the radiation dose. (Lois, 2011).

Radiation Weighting Factor

Radiation weighting factor (W_R) is extrapolates the risk of partial body exposure to patients, and it differs according to the type of radiation because it varies in its degree of ionizing and biological effects.(Lois,2011).

Table (2.1): Radiation weighting factors in publication ICRP 60 and Q in publication ICRP 60:

Radiation type and Energy	**Radiation weighting factor W_R**	**Q**
Photons (All energies)	**1**	**1**
Electrons (All energies)	**1**	**1**
Neutrons		**-**
< 10 keV	5	-
10 keV to 100 keV	10	-
> 100 keV to 2 MeV	20	-
>2 MeV to 20 MeV	10	-
>20 MeV	5	-
Protons > 2 MeV	**5**	**1**
Alpha particles, fission fragments and heavy nuclei	**20**	**20**

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2.1.8.4Effective dose

Effective dose is becoming a very useful radiation quantity for expressing relative risk to humans, both patients and other personnel. It is actually a simple and very logical concept. It takes into account the specific organs and areas of the body that are exposed. The point is that all parts o f the body and organs are not equally sensitive to the possible adverse effects o f radiation, such as cancer induction and mutations, For the purpose o f determining effective dose, the different areas and organs have been assigned tissue weighting factor (WT) values. For a specific organ or body area the effective dose is:

$$\textbf{Effective Dose (Sv) = Absorbed Dose (Gy) x } \mathbf{W_T}$$

If more than one area has been exposed, then the total body effective dose is just the sum o f the effective doses for each exposed area. It is a simple as that. Now let's see why effective dose is such a useful quantity. There is often a need to compare the amount o f radiation received by patients for different types of x-ray procedures, for example, a chest radiograph and a CT scan. The effective dose is the most appropriate quantity for doing this. Also, by using effective dose it is possible to put the radiation received from diagnostic procedures into perspective with other exposures, especially natural background radiation. (Perry Sprawls,1993).

Table (2.2): Weighting Factors as defined in ICRP 26, 60 and 103.

Tissue or Organ	Weighting factors (ICRP 26)	Tissue Weighting Factors (ICRP 60)	Tissue Weighting Factors (ICRP 103)
Bladder	-	0.05	0.04
Bone	0.03	0.01	0.01
Brain	-	-	0.01
Breasts	**0.15**	**0.05**	**0.12**
Colon	-	0.12	0.12
Esophagus	-	0.05	0.04
Liver	-	0.05	0.04
Lungs	0.12	0.12	0.12
Ovaries/testes	0.25	0.2	0.08
Red marrow	0.12	0.12	0.12
Remainder tissues	0.3	0.05	0.12
Salivary glands	-	-	0.01
Skin	-	0.01	0.01
Stomach	-	0.12	0.12
Thyroid	0.03	0.05	0.04
Total	**1.00**	**1.00**	**1.00**

2.1.9 Quantities for CT Dosimetry

The dose qualities used in projection radiography are not applicable to CT for three reasons**:**

First: The dose distribution inside the patient is different from that of conventional radiologram, in case of CT, as a consequence of scanning procedure that equally irradiates from all direction.

Second: Scan procedure using narrow beams along the longitudinal z-axis of the patient.

Third: the situation with CT, unlike with conventional projection radiography is further complicated by the circumstances in which the value to be imaged is not irradiated simultaneity. This often led to confusion about what dose from a complete series. As a consequence, dedicated dose quantities that account for these peculiarities are needed the computed tomography dose index (CTDI) and the dose–length product (DLP) and effective dose. (Lois, 2011).

2.1.9.1 Multiple Scan Average Doses (MSAD)

For a series of multiple scans with constant separation, the multiple scan average doses (MSAD) is an indication of the magnitude of the dose along the length of the scanned volume at a particular radial depth it is usually calculated from multiple scans. Measurements are made from the central slice; which has the total radiation dose, plus several points in the peripheral slices (scatter overlap or tails). It will increase if the slices overlapped and will decrease if there were a gap between the slices. (Lois,2011).

2.1.9.2 Computed tomography dose index (CTDI)

Another measurement for radiation dose in CT is the Computed tomographic dose index (CTDI) which is what manufacturers now report to FDA and prospective customers regarding the dose delivered by their machines. The CTDI can only be calculated if slices are contiguous, that is, there are no overlapping or gapped slices. If there is slice overlap or gaps, the CTDI is multiplied by the ratio of slice thickness to slice increment. This would technically be the MSAD, because the CTDI conditions would no longer exist. (Lois, 2011).

The CTDI unit: is milli gray (mGy) is derived from the dose distribution a long a line which is parallel to the axis of rotation for the scanner (z-axis and which is recorded for a single rotation of X-ray source.

The CTDI is the equivalent of the dose value inside the irradiated slice (beam) that would result if the absorbed radiation dose profile were entirely concentrated to a rectangular of width equal to the nominal beam width N.hcol, with N being the number of independent (i.e. non-overlapping) slices that are acquired simultaneously.

The corresponding mathematical definition of CTDI therefore describes the summation of all dose contributions along the z-axis:

$$\mathbf{CTDI} = \frac{1}{N.hcol}\int_{-\infty}^{+\infty} D(z)dz$$

Where: *D* (*z*) is the radiation dose profile along the z-axis, N.hcol is the nominal value of the total collimation (beam width) that is used for data acquisition. CTDI is therefore equal to the area of the dose profile (the dose-profile integral) divided by nominal beam width.(Lois, 2011).

2.1.9.3 Weighted Computed tomography dose index ($CTDI_w$)

The weighted CTDI ($CTDI_W$) is used for approximating the average dose over a single slice and is defined by the following equation, separately for the head and body phantoms:

$$\mathbf{CTDI_W = 1/3CTDI_C + 2/3CTDI_P}$$

The dose for body scans are not uniform across the scan field of view, the dose at the periphery of the slice is higher than the central dose. The $CTDI_w$ adjusts for this by providing a weighted average of measurements at center and the peripheral slice locations. (Lois, 2011).

2.1.9.4Volume Computed tomography dose index ($CTDI_{vol}$)

Provides measurements of the exposure per slice of tissue for the whole scan length, under the condition of no gaps or overlapping in the scan slices.(Lois, 2011).

$CTDI_{vol}$ affected by pitch as shown in table (2.3) there is direct relation between $CTDI_{vol}$ and pitch.

$$\mathbf{CTDI_{VOL} = CTDI_W \times \frac{NT}{I} = \frac{CTDIw}{Pitch}}$$

Where NT is the total nominal collimation width and **I** is table travel per rotation during a helical scan (pitch factor = $\frac{I}{NT}$).

***Table (2.3): shows the effect of the pitch over the CTDI$_{vol.}$ All other factors were held constant at 120kVp, 300 mA, 1s and 10mm. results are from a single-detector row CT scanner*.**(Lois, 2011).

Pitch	**CTDI$_{vol}$ in head phantom (mGy)**	**CTDI$_{vol}$ in body phantom (mGy)**
0.5	80	36
0.75	53	24
1.0	40	18
1.5	27	12
2.0	20	9

2.1.9.5 Dose length product (DLP)

Dose length product (DLP) is a measure of CT radiation exposure. It is related to CTDI$_{vol}$, but CTDI$_{vol}$ represents the dose through a slice of an appropriate scan. DLP accounts for the length of radiation output along the long axis of the patient. It Provides a measurement of the total amount of exposure for series of scans. (Lois, 2011).

It can be calculated if the length of the irradiated volume (Scan length) and CTDI$_{vol}$ are known by using the following equation:

$$\mathbf{DLP = CTDI_{vol} \times Scan\ length.}$$

That equation conclude if the scan length is increased, this will also increase the DLP. (Lois, 2011).

Although the DLP more closely reflects the radiation dose for a specific CT examination, its value is affected by variances in patient anatomy. Therefore, the $CTDI_{vol}$ is more useful tool for comparing radiation doses among different protocols. (Lois, 2011).

2.1.9.6 Effective dose in CT

Effective dose is a biological dose, determines how dangerous an individual's exposure to radiation can be and it is used commonly in radiation protection. (Lois, 2011).

It can be estimated by following equation:

$$\mathbf{E = DLP \times W_T}$$

or $$\mathbf{E = DLP \times K}$$

E ≡ Effective dose. DLP ≡ Dose length product.

W_T ≡ Tissue Weighting Factor. K ≡ K factor.

(K) Factor is called kerma which stands for kinetic energy released to matter and it is a measurement of the amount of energy transferred from photons to electrons per unit mass (matter) at a certain position. Measured by the (SI) unit Gray.

The amount of the K factor differs according to the organ and some examples are shown in the table (2.4) below: (Dr Dan et al, 2018).

Table (2.4): Shows the amount of the K factor of different organs.

The Organ	K Factor(Gy)
Head	0.0021
Neck	0.0059
Chest	0.014
Abdomen and pelvis	0.015

2.1.10 Factors that influence the amount of radiation dose

Although the scanner design is of some importance, surveys on CT practice have regularly shown that the way how the scanner is used has the largest impact on the doses applied in a CT examination. (Hans Dieter, 2007)

2.1.10.1Beam filtration

Which is well known to reduce the portions of radiation spectrum with no or less contribution in the image formation. The size of the filter differs according to the scanner type and the dose values and may range from 18mm Al equivalent (in old scanners) to 5-6mm Al equivalent in newer ones and may be more or less due to manufacturers differences. The use of additional filtration impairs primary contrast and increase noise due to the reduction of beam's intensity which effects the detectably of small or low-contrast details and is compensated by for example increasing tube current-time production. (Hans Dieter, 2007).

Other studies says that in order to maintain the contrast to noise ratio for constant image quality the standard beam filtration must be increased, while new surveys on CT revealed that scanners with comparable age with largely differing beam filtration operate at almost similar dose levels.(Hans Dieter, 2007).

2.1.10.2 Beam collimator

Beam collimator used for defining the thickness of the slice to be imaged and they are two of them; **primary collimation** which is close to the x-ray tube and it affects the dose profile along with its distance from the focal spot, and the size and shape of the focal spot. The penumbral effects are increased with the decreased width of the collimator that's why the collimator is wider than the activated detectors arrays. (Hans Dieter, 2007).

The second type is **post-patient collimator** which is close to the detectors array and used to remove scattered radiation, and this collimator may be further narrowed in single and dual scanners to improve the shape of the slice profile. (Hans Dieter, 2007).

Newer scanners are equipped with means that automatically adapt the mAs settings to the individual size and shape and they are called **Automatic dose control.** (Hans Dieter, 2007).

2.1.10.3 Dose Display

With the dose display (CTDI-DLP), dose is not saved per scan, but feedback is provided that may help to achieve this goal, e.g. by comparison of the displayed dose values with dose recommendations. In addition, changes in scan parameter settings and their implications for patient exposure are made immediately obvious. Thus, the dose display can be used for purposes of dose optimization. Finally, $CTDI_{vol}$ can be used as a fair estimate for the dose to organs that are entirely located in the scan range. (Hans Dieter, 2007).

The interpretation of the dose values displayed at the scanner's console needs special attention in the following situations, first; Many dose recommendations are given in terms of weighted CTDI ($CTDI_w$); in order to allow for comparisons, the pitch correction involved in $CTDI_{vol}$ must be reverted by multiplying $CTDI_{vol}$ with the pitch factor and secondly; Up to now, the dose values for examinations carried out in body scanning mode are always based on body-CTDI regardless of patient size. In pediatric CT examinations, the displayed figures should be multiplied by 2 for children and by 3 for infants in order to give a realistic estimate of patient dose. (Hans Dieter, 2007).

2.1.10.4 Tube current-time product

Which is also known as mAs, and as in conventional radiology it has a linear relation with the dose. And it affects the noise of the image as well.[5]

Often the mAs is used as a surrogate for the patient dose which is misleading because there are other almost 6 factors that can affect it, only $CTDI_{vol}$ and DLP can be used for this purpose. (Hans Dieter, 2007).

Even though, AEC should be used to reduce the tube current according to the patient's size, body part thickness and the dose requirements for each type of exam. (Hans Dieter, 2007).

2.1.10.5 Slice thickness

It is known to have an inverse relationship with the dose and image noise but a linear relation with spatial resolution and small details viewing. And the use of weather thick or thin slice thickness is determined according to the patient's condition and the part needs to be seen. (Hans Dieter, 2007).

2.1.10.6 The pitch

In MSCT scanners make use of spiral interpolation thus the slice profile stays the same with changes in the pitch unlike in SSCT scanners, instead image noise changes with the pitch unless the tube current is adopted using automatic adaptation of mAs. (Hans Dieter, 2007).

2.1.10.7 Object diameter

Although is not a parameter to be selected from the operating console but still needs to be considered in the context. (Hans Dieter, 2007).

Considerable reduction in mAs settings are appropriate whenever thin or young patients are examined to avoid unnecessary over-exposure. The mAs is adopted by the operator unless AEC is available. (Hans Dieter, 2007).

2.1.10.8 Scan Length

The local dose, i.e. CTDI, is almost independent of the length of the scanned body section. The same does not hold, however, for the integral dose quantities, i.e. dose-length product and effective dose. Both increase in proportion to the length of the body section. Therefore, limiting the scan length according to the clinical needs is essential. So, it should be as short as possible especially in multi-phase scans and follow up and selected individually according to the scan to be done. (Hans Dieter, 2007).

2.2 Previous studies

This section presents other related researchers who conducted studies similar to the proponents that will also greatly help in progress of the study. And it will also help the understanding of the proposition.

Mehnati. P. et al, on study research under title "**Reducing Radiation Doses in Female Breast and Lung during CT Examinations of Thorax**". The study intended to compare the female breast and lung doses using split and standard protocols in chest CT scanning. The breast and lung doses were measured by the sliced chest, and breast female phantoms were used. CT exams were performed using a single-slice (SS) - and a 16 multi slice (MS) - CT scanner at 100 kVp and 120 kVp. Two different protocols, including standard and split protocols, were selected for scanning. The application of a split scan technique instead of standard protocol has a considerable potential to reduce breast and lung doses in SS- and MS- CT scanners. If split scanning protocol is associated with an optimum kV and MSCT, the maximum dose decline will be provided. (Mehnatiet al, 2015).

Abdurrahman Salahudeen Abdullateef in Sudan, (2018) had a study under the title **"Estimation of Radiation Dose during Chest Computed Tomography Examinations"**. The aim of the study was to evaluate the radiation doses from Chest Computed Tomography examinations. The result of the study revealed that the values from CT Chest with contrast exam was $CTDI_{vol}$ 8.69± 3.620 mGy, DLP 702.54±344.37 mGy/cm and Effective Dose (ED) was 9.84± 4.82 mGy, from HRCT study; $CTDI_{vol}$ was 12.48±3.732 mGy, DLP 387.92±121.48 mGy/cm and ED5.43±1.70 mGy and from Pulmonary angiography procedure was $CTDI_{vol}$13.53±3.20 mGy, DLP 986.73±276.12 mGy/cm and ED 13.81±3.86 mGy.(David Pltten, 2005).

Hurwitz et al, on study research under titled **"Radiation Dose to the Female Breast from 16MDCT Body Protocols".** The study intended to determine the radiation dose to the female breast from current 16-MDCT body examinations. Metal oxide semiconductor field effect transistor (MOSFET) detectors were placed in four quadrants of the breast of a female-configured anthropomorphic phantom to determine radiation dose to the breast. Imaging was performed on a 16-MDCT scanner using current clinical protocols designed to assess pulmonary embolus (PE) (140 kVp, 380 mA, 0.8-sec rotation), appendicitis (140 kVp, 340 mA, 0.5-sec rotation) and renal calculus (140 kVp, 160 mA, 0.5-sec rotation). Radiation dose to the breast ranged from 4 to 6 cGy for the PE protocol and 1-2 cGy in the inferior aspect of the right breast and lateral aspect of the left breast for the appendicitis protocol. The renal calculus protocol yielded less than 150 cGy absorbed breast dose. Current clinical chest and abdomen protocols result in variable radiation doses to the breast. The magnitude o f exposure may have implications for imaging strategies. (Hurwitz et al, 2006).

Lee et al, worked on study research under title **"Breast Cancer Risk Estimates Increased with Repeated Prior CT "**. The purpose of that study was to estimate increased with repeated CT imagines. They collected CT dose information from 1,656 patients who Underwent CT examinations that exposed the breast to radiation and, using a new automated computational method, estimated the patients' effective radiation dose and the amount of radiation absorbed by the breast. The researchers then estimated the women imaging-related risk of breast cancer and compared it to their underlying risk of developing breast cancer. Each woman's 10-year imaging- related risk of developing breast cancer, beginning 10 years after her exposure to imaging and based on her age at exposure, was estimated using die breast-specific radiation data and Statistical risk model. They found that young women receiving several chest and or cardiac CTs had the greatest increased risk of developing breast cancer at approximately 20 percent. To lower imaging-related risk of developing breast cancer, imaging providers should analyze the radiation doses associated with each exam, reduce the use o f multi-phase protocols and employ dose-reduction software wherever possible to minimize exposures. (Lee et al, 2012).

Livingstone et al, studied **"Radiation doses during chest examinations using dose modulation techniques in multislice CT scanner".** The purpose of this study was to evaluate the radiation dose and image quality using a manual protocol and dose modulation techniques in a 6-slice CT scanner. Two hundred and twenty-one patients who underwent contrast-enhanced CT of the chest were included in the study. For the manual protocol settings, constant tube potential (kV) and tube current-time product (mAs) o f 140 kV and 120 mAs, respectively, were used. The angular and z-axis dose modulation techniques utilized a constant tube potential of 140 kV; mAs values were automatically selected by the machine. Effective doses were calculated using dose length product (DLP) values and the image quality was assessed using the signal-to noise (SNR) ratio values. Mean effective doses using manual protocol for patients of weights 40-60 kg, 61-80 kg, and 81 kg and above were 8.58mSv, 8.54 mSv, and 9.07 mSv, respectively. Mean effective doses using z-axis dose modulation for patients of weights 40-60 kg, 61-80 kg, and 81 kg and above were 4.95 mSv, 6.87 mSv, and 10.24 mSv, respectively. The SNR at the region o f the liver for patients of body weight of 40- 60 kg was 5.1 H, 6.2 H, and 8.8 H for manual, angular, and z-axis dose modulation, respectively. Dose reduction o f up to 15% was achieved using angular dose modulation and o f up to 42% using z-axis dose modulation, with acceptable diagnostic image quality compared to the manual protocol. (Livingstone et al, 2010).

Smith-Bindman et al, studied **"radiation dose associated with common CT examinations and the associated lifetime attributable risk of cancer"**. They sought to estimate the radiation dose associated with common CT studies in clinical practice and quantify the potential cancer risk associated with these examinations. They conducted a retrospective cross sectional study describing radiation dose associated with the 11 most common types o f diagnostic ct studies performed on 1119 consecutive adult patients at 4 San Francisco Bay Area in situation California. They estimated life time attributable risks o f cancer by study types from these measured doses. Radiation doses varied significantly between the different types of CT studies. The overall median effective doses were 31 mSv for a multiphase abdomen and pelvis CT scan. Within each type of CT study effective dose varied significantly within across institutions, with a mean 13-fold variation between the highest and lowest dose for each study type. The estimated number o f CT scans that will lead to the development o f a cancer varied widely depending on the specific type of CT examination and the patient's age and sex. An estimated 1 in 270 women who underwent CT coronary angiography at age 40 years will develop cancer from that CT scan (1 in 1180 men). For 20 year old patients, the risks were approximately doubled, and for 60 year old patients, they were approximately 50% lower. Radiation doses from commonly performed diagnostic examinations are higher and more variable than generally quoted, highlighting the need for greater standardization across institutions. (Smith-Bindman et al, 2009).

Abdurrahman Mohamed Alnour works this study in Sudan (2015) under the title of **“Establishment of Diagnostic Reference Level for CT Scan in Sudan"**. The aim of this study was to establish a level of exposure to acceptable according to international standards without prejudice to reduce the presence of medical imaging and desired information from the tests and that called DRL_S. In this study used statistical method to represent the results and data in order to calculate DRL_S of the total statistical data, the study showed follows: First in brain imaging the DRL_S are (1209 mGy /cm).Second, a CT scan of the chest, the DRL_S are (650mGy/cm). Thirdly, CT scan of the abdomen and pelvis the study showed that DRL_S are (978 mGy / cm). (Abdelrahman et al, 2015).

Chapter Three

Materials and Methods

Chapter Three

Materials & Methods

3.1 Materials

This study intended to evaluate the patient effective dose and organ dose during chest CT procedures. The data used in this study were collected from two hospitals in Khartoum state: Hospital (Z), Hospital (Y), during the period from September to November 2018.

3.1.1 Patient data

Table (3.1): patient population of the study classified per hospital and type of examination.

Hospital	Exam	Male	Female	Total
Z	Chest	15	15	30
Y	Chest	16	15	31

3.1.2 CT machines

Two CT machines were used to collect data during this study. These machines are installed in two private radiological departments. All quality control tests were performed to the machines prior to any data collection. The tests were earned out by experts from Sudan Atomic Energy Commission (SAEC). All the data were within acceptable range.

Table (3.2): CT machines

Hospital	Manufacturer	Model	Installation	Detected type
Z	Toshiba	Acquilion 16	2018	(16-Slice)
Y	Toshiba	Alixon 16	2010	(16-Slice)

3.1.3 Data collection

Data were collected using a data sheet for all patients in order to maintain consistency of the information displayed during CT examinations (Appendix). All CT machines are equipped with CT dosimetry unit. A data collection sheet was designed to evaluate the patient doses and the radiation related factor. The collected data included:(Gender, age, tube potential(mA), tube current-time product settings(mAs), pitch, slice thickness and total slice number), In addition, all scanning parameters were recorded as well as the CT Dose Index volume ($CTDI_{vol}$) in (mSv) and dose-length product (DLP) in (mSv.cm).All these factors have a direct influence on radiation dose. The entire hospital was passed successfully the extensive quality control tests performed by Sudan Atomic Energy Commission to meet the criteria o f this study.

3.2 Methods

3.2.1 Dosimetric calculations

CT Expo software was used to calculate common CT dose descriptors: (i) CT weighted dose index ($CTDI_w$), (ii) volume dose index ($CTDI_{vol}$) 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 the complete CT examination, (vi) effective dose (E): a method for comparing patient doses from different diagnostic procedures (Effective dose) and (iiv) organ dose

3.2.2 CT-Expo V 2.5 software

The CT-Expo Version 2.5 software tool was 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 were used. Dose estimation was done based on mathematical phantoms for adult (ADAM and EVA).

3.2.3 Data Analysis

The data in this study were analyzed using the Excel**.**

Chapter four

Results

Chapter four

Results

4.1 Results:

The results of this study were presented for dose measurements performed in two hospitals, hospital (Y) with CT scanner Toshiba model Alixon (16-slice) versus hospital (Z) with CT scanner Toshiba model Acquilion (16-slice), and 61 CT examinations in patients, doses were estimated in terms of $CTDI_{vol}$, DLP and E, the tables below describe the results in details.

Table (4.1): summarises the characteristic performance parameters for the CT systems and console displayed form The CT scanner Toshiba model Alixon (16-slice) hospital (Y).

	Age (y)	kVp (kv)	MAs (mA.s)	Scan length (cm)	Pitch	$CTDI_{vol}$ (mSv)	DLP (mSv.cm)
Mean	59.7	120	75	32.5	1.5	8.6	313.8
Median	65	120	75	32	1.5	8.5	309.9
STD	17.6	0	0	3.0	0	0.8	41.3
Max	84	120	75	41	1.5	11.6	448.9
Min	21	120	75	28	1.5	8.1	267

Table (4.2): summarises the characteristic performance parameters for the CT systems and console displayed form The CT scanner Toshiba model Acquilion (16-slice) hospital (Z).

	Age (y)	kVp (kv)	mAs (mA.s)	Scan length (cm)	Pitch	$CTDI_{vol}$ (mSv)	DLP (mSv.cm)
Mean	52.8	120	125	36.96	1	17.4	700.8
Median	52	120	125	36	1	17.4	690.2
STD	18.6	0	0	3.1	0	0.02	54.3
Max	83	120	125	46	1	17.4	855.4
Min	20	120	125	31	1	17.3	594.6

Table (4.3): shows the estimation of mean $CTDI_w$, $CTDI_{vol,}$ DLP and effective dose (E) calculated by software expo2.5 using data collection form CT scanner Toshiba Alixon (16-slice) hospital (Y).

	$CTDI_w$ (mSv)	$CTDI_{vol}$ (mSv)	DLP (mSv.cm)	E (mSv)
Mean	10.4	8	283	5.3
Median	10.4	8	279.1	5.6
STD	0	0	24.1	1.03
Max	10.4	8	351.3	7.3
Min	10.4	8	247	3.8

Table (4.4): shows the estimation of mean $CTDI_w$, $CTDI_{vol,}$ DLP and effective dose calculated by software expo 2.5 using data collection form CT scanner Toshiba Acquilion (16-slice) hospital (Z).

	$CTDI_w$ (mSv)	$CTDI_{vol}$ (mSv)	DLP (mSv.cm)	E (mSv)
Mean	14.6	17.3	741.3	14.2
Median	14.6	17.3	724.5	14.3
STD	0	0	54	1.8
Max	14.6	17.3	897.8	16.2
Min	14.6	17.3	637.8	11.6

Table (4.5) summarises the characteristic statistics parameters for the hospital (Y) from The CT scanner Toshiba model Alixon (16-slice), for male:

	Age (y)	kVp (kv)	mAs (mA.s)	Pitch	$CTDI_{vol}$ (mSv)	DLP (mSv.cm)	E (mSv)
Mean	60.4	120	75	1.5	8	288	4.4
Median	65	120	75	1.5	8	291	4.5
STD	18.8	0	0	0	0	26	0.5
Min	21	120	75	1.5	8	263	3.8
Max	84	120	75	1.5	8	351	5.6

Table (4.6) summarises the characteristic statistics parameters for the hospital (Y) from The CT scanner Toshiba model Alixon (16-slice), for female:

	Age (y)	**kVp (kv)**	**mAs (mA.s)**	**Pitch (cm)**	**$CTDI_{vol}$ (mSv)**	**DLP (mSv.cm)**	**E (mSv)**
Mean	58.9	120	75	1.5	8	278	6.2
Median	70	120	75	1.5	8	263	5.8
STD	16.8	0	0	0	0	21.5	0.5
Min	28	120	75	1.5	8	247	5.6
Max	80	120	75	1.5	8	327	7.3

Table (4.7) the comparison of (E) between male and female from Toshiba CT scanner model Alixon (16-slice) for the hospital (Y):

E	**Gender**		**Total**
	Female	Male	
3.8	0	6	6
4.5	0	2	2
4.6	0	3	3
4.7	0	2	2
4.9	0	1	1
5.0	0	1	1
5.6	1	1	2
5.8	7	0	7
6.2	2	0	2
6.7	3	0	3
6.8	1	0	1
7.3	1	0	1
Total	**15**	**16**	**31**

Table (4.8) summarises the characteristic statistics parameters for the hospital (Z) from The CT scanner Toshiba model Alixon (16-slice), for male:

	Age (y)	**kVp (kv)**	**mAs (mA.s)**	**Pitch**	**$CTDI_{vol}$ (mSv)**	**DLP (mSv.cm)**	**E (mSv)**
Mean	49.9	120	125	0.8	17.3	779	12.5
Median	50	120	125	0.8	17.3	777	12.6
STD	18.7	0	0	0	0	47.2	0.7
Min	20	120	125	0.8	17.3	725	11.6
Max	82	120	125	0.8	17.3	898	14.1

Table (4.9) summarises the characteristic statistics parameters for the hospital (Z) from The CT scanner Toshiba model Alixon (16-slice), for female:

	Age (y)	**kVp (kv)**	**mAs (mA.s)**	**Pitch (cm)**	**$CTDI_{vol}$ (mSv)**	**DLP (mSv.cm)**	**E (mSv)**
Mean	55.7	120	125	0.8	17.3	704	15.8
Median	60	120	125	0.8	17.3	725	16.2
STD	18.7	0	0	0	0	27.9	0.6
Min	25	120	125	0.8	17.3	638	14.4
Max	83	120	125	0.8	17.3	725	16.2

Table (4.10) the comparison of (E) between male and female from Toshiba CT scanner model Acquilion (16-slice) for the hospital (Z):

E	**Gender**		**Total**
	Female	Male	
11.6	0	3	3
12.0	0	2	2
12.3	0	1	1
12.6	0	3	3
12.8	0	2	2
12.9	0	2	2
13.2	0	1	1
14.1	0	1	1
14.4	1	0	1
15.2	3	0	3
15.5	1	0	1
15.9	2	0	2
16.2	8	0	8
Total	**15**	**15**	**30**

Table (4.11) compares the *$CTDI_{vol}$, DLP* ***and effective dose (E) between the two hospitals:***

Hospitals	**$CTDI_{VOL}$ (mSv)**	**DLP (mSv.cm)**	**Effective dose (E) (mSv)**
Y	8	283	5.8
Z	17.3	741	14.2
Mean	**12.7**	**512**	**10**

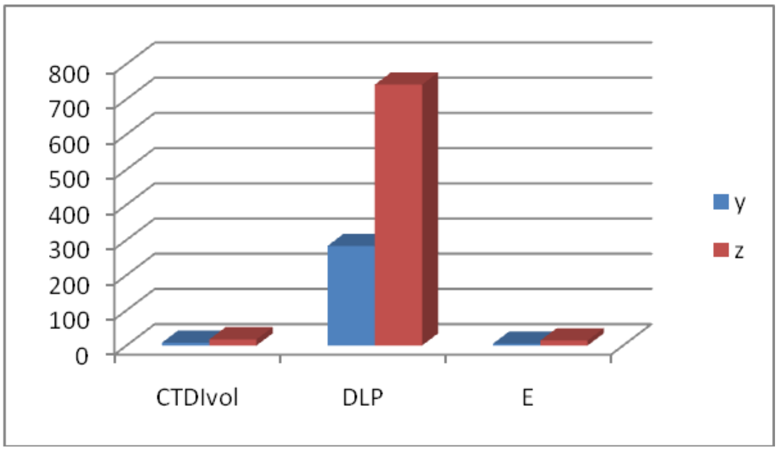

Fig (4.1): Compares the *$CTDI_{vol}$, DLP* ***and effective dose (E) between the two hospitals.***

Table (4.12) showed the organ dose for Breast, Lung, and Thyroid estimated from the hospital (Y) by CT Expo 2.5 software during this study:

	Breast	**Lung**	**Thyroid**
Mean	12.8	12.4	6.7
Median	12.8	12.3	4.9
STD	0.03	0.14	4.6
Min	12.8	12	1.3
Max	12.8	12.6	19.2

Table (4.13) showed the organ dose for Breast, Lung and Thyroid estimated from the hospital (Z) by CT Expo 2.5 software during this study:

	Breast	**Lung**	**Thyroid**
Mean	27.8	27.2	38.8
Median	27.9	27.3	39.7
STD	0.04	0.09	2.9
Min	27.7	27.4	42.5
Max	27.9	27.01	33.3

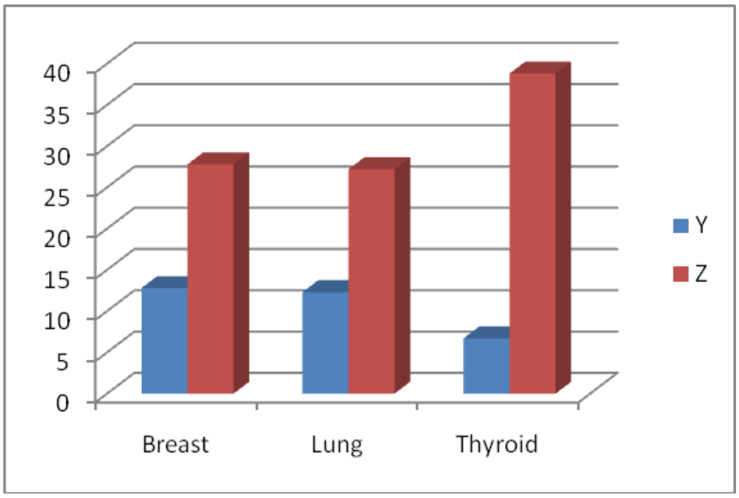

Figure (4.2): Compares the organ dose between the two hospitals

Table (5.1) shows the mean $CTDI_{vol}$, DLP and E as compared with other countries and EC reference dose:

	This study	Tanzania	Taiwan	U K	E C
$CTDI_{vol}$	**12.7**	17	20	11.5	30
DLP	**512**	783	455	850	650
E	**10**	13	8.4	14.9	11.7

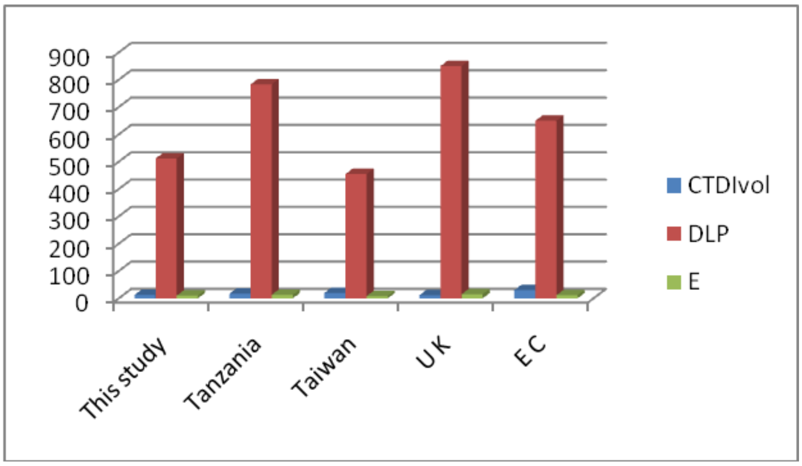

Figure (5.1): comparison between the mean of $CTDI_{VOL}$, DLP and E in this study and other countries and EC reference dose.

Chapter Five

Discussion, Conclusion and Recommendations

Chapter Five

Discussion, Conclusion & Recommendations

5.1 Discussions:

In this study doses were expressed in terms of $CTDI_{vol,}$ DLP, E and organ dose. This an indication of the average absorbed dose in the scanned region ($CTDI_{vol}$), the integrated absorbed dose along a line parallel to the axis of radiation for the complete CT examination (DLP) and effective dose and compared the effective dose between the two hospitals, the results are discussed in details.

Tables (4.1) and (4.2) presented parameters for the CT systems and console displayed form The CT scanner for two hospitals (Y) and (Z) respectively. The results showed that the two hospitals used the same kVp (120 kv), different mAs (75 and 125) and different pitch (1.5and 0.8).

Tables (4.3) and (4.4) presented the estimation of (mean, median, std, min and max) and $CTDI_{vol}$, DLP, E calculated by software CT Expo β verge 2.5 using data collection from CT scanner Toshiba (16 slice) model Alixon and Acquilion. The results showed the two hospitals used different protocol and showed that the mean $CTDI_{vol}$ (8 and 17.3) mSv, DLP (283 and 741) mSv.cm and Effective dose (5.8 and 14.2) mSv from (Y) and (Z) respectively. The effective doses at hospital (Z) were higher than at hospital (Y) because hospital (Z) used higher mAs (125) than hospital (Y) (75).

For the two hospitals in this study when we compared the effective dose between them we found that the effective doses for males (3.8 to 5.6) mSv were lower than females (5.6 to 7.3) mSv, this result was presented in table (4.7) for hospital (y). While the effective doses for males (11.6 to 14.1) mSv and females (14.4 to 16.2) mSv in table (4.10) for hospital (Z) and this difference of effective doses for males and females may be due to difference in composition of tissue and weight.

Table (4.11) which compared the $CTDI_{VOL}$, DLP and effective dose (E) between the two hospitals, were (8 and 17.3) mSv, (283 and 741) mSv.cm and (5.8 and 14.2) mSv for hospitals (Y) and (Z) respectively, and showed the mean of $CTDI_{VOL}$, DLP and effective dose (E) which were 12.7 mSv, 512 mSv.cm and10 mSv, and this result was presented in figure (4.1).

Tables (4.12) and (4.13): presented the organ dose for Breast, Lung and Thyroid which was estimated from the hospital (Y) and hospital (Z) by CT Expo 2.5 software during this study and showed no difference between male and female doses for some organ such as lung and thyroid in both hospitals. However, the dose for the breast in hospital (Z) was higher than hospital (Y) for the Brest.

The organ doses of the critical organs for two hospital (Y) and (Z) respectively, were (12.8 and 27.8) mSv for breast, (12.4 and 27.2) mSv for lung and (6.7 and 38.8) mSv for thyroid, this result presented in figure (4.1).

Table (5.1) were compared the mean $CTDI_{vol}$, DLP, and E in this study with Tanzania, Taiwan, United Kingdom and European Commission reference and we found that the mean of $CTDI_{vol}$, DLP, and E, was lower than Tanzania, United Kingdom and European Commission reference and higher than Taiwan.

5.2 Conclusion:

The general objective in this study was to evaluate of patient effective dose and the radiation dose that received by selected radiosensitive organs of the patients undergoing chest CT examination using CT Toshiba 16 slice in two hospitals and we compare between them. The organs doses in hospital (Z) higher than organs doses in hospital (Y). The mean effective doses to patients undergoing CT chest examination at hospital (Z) higher than mean effective doses to patients undergoing CT chest examination at hospital (Y), because the hospital (Z) used higher mAs than hospital (Y).

The effective dose in this study were compared with different reported values from the Tanzania, Taiwan, United Kingdom, and EC reference and showed that it was lower than Tanzania, United Kingdom and European Commission reference and higher than Taiwan. This difference in effective doses may be due to the difference in protocols.

5.3 Recommendations:

Because the large observed variations of Effective dose and organ doses in the study hospitals call for the need to optimize CT scanning protocols.

-I recommended optimal selection of scanning parameters based on indication of study, body region of interest being scanned, and patient size.

- I recommended

-I recommended further studies should be done to investigate the potential for using radio protective materials to protect superficial radiosensitive organs.

-I recommended use this study to prepare the reference dose level by Sudan Atomic Energy Council.

- I recommended this patient was examined in hospital Z not rebated the CT scan during this year because it was resaved high doses approx trisect the annual dose.

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Appendix

Sudan University of Science and Technology

College of Graduate Studies

Evaluation of the patient Effective Dose and critical organs dose during Chest Computed Tomography

Data Sheet

Hospital Name: ………………………………………..

Date: …/…/.......

Patient info.			Exam	Scan parameters										Console displayed dose	
NO	Sex M/F	Age (y)		kV	mAs	Slice thickness (mm)	pitch	Speed	Total scan time (s)	Number of slice	Table movement		Scan length (cm)	CTDIvol (mGy)	DLP (mGy.cm)
											Start	End			