Sudan University of Science and Technology

College of Graduate Studies and Scientific Research

Assessment of Radiation Dose Received By Patient During CT urography

تقويم الجرعة الاشعاعيه للمريض اثناء تصوير الجهاز البولى بالاشعة المقطعية

A thesis submitted for partial fulfillment for the requirements

of MSc degree in Medical Physics

By:

Aesha Malik Bashir Omar

Supervisor

Dr.Hussein Ahmed Hassan

March 2016

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□بسم الله الرحمن الرحيم

(وَهُوَ اللَّهُ فِي السَّمَاوَاتِ وَفِي الأَرْضِ ۖ يَعْلَمُ سِرَّكُمْ وَجَهْرَكُمْ وَيَعْلَمُ مَا تَكْسِبُونَ)

صدق الله العظيم

سورة الانعام الآية (3)

Dedication

To my parent, to my brothers, to every one presented help me

I Dedicate This Thesis.

Acknowledgements

firstly, I would like to express my gratefulness to Dr. ***Hussein Ahmed*** *for his help and guidance. Without his help this work could never have been accomplished.*

I also would like to thank Military hospital staff for their help.

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Table of Abbreviations

CT	Computed Tomography
CTU	Computed Tomographic Urography
AEC	Automatic Exposure Control
MSCT	Multi-Slice Computed Tomography
SSCT	Single Slice Computed Tomography
ICRP	International Commission Radiation Protection
CTDI	Computed Tomography Dose Index
MSAD	Multi-Scan Average Dose
DLP	Dose Length Product
DAP	Dose Area Product
KUB	Kidney, Ureters and Bladder
IV	Intravenous
ROI	Region Of Interest

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الملخص

يمثل التصوير الاشعاعي دورا اساسيا في طرق التشخيص الطبي ولذلك يقع على عاتق اخصائي و تقني الاشعه تحديد عوامل الجرعه الاشعاعيه و ذلك لتوفير التوازن بين جودة الصوره و جرعة الاشعاع وحفظ المريض من التعرض الاشعاعي.

الهدف الرئيسي من هذه الدراسه تقييم الجرعه الاشعاعيه للمرضى اثناء فحوصات الجهاز البولي بالاشعه المقطعيه.

الجرعة الإشعاعية قدرت بواسطة برنامج الهيئه القومية للوقاية من الاشعاع وبرنامج حساب الجرعات للاشعه المقطعيه.

69,47±19.81هو عامل الجرعه للاشعه المقطعيه الحجمي ملي قراى.الجرعه لوحدة الطول هي 3647.81±950.34 ملسفرت للسنتمتر و الجرعه الفعاله 54.72±14.26 ملسفرت.

وكان احتمال حدوث سرطان للرحم28.03لكل مليون و27.05للبرستات.

الجرعه الاشعاعيه تختلف بحسب الجهاز المستخدم و حجم المريض ونوع البروتوكول المستخدم في هذهالدراسه تعتبر الجرعه الاشعاعيه عاليه مقارنه مع الدراسات السابقه قد تكون نسبه لحجم المريض او البروتوكول المستخدم في المركز.

Abstract

Radiography has major role of diagnostic method in medical field. Urography provides the radiologist with useful detailed information. However, it is the responsibility of radiologist and technologist to determine scanning technique factor that provide balance between image quality and radiation dose and share in keeping patient radiation exposure at lowest as possible.

The objective of this study to assess radiation dose from CT and estimate radiation risk for patient during CTU.The patient radiation dose value from this study 3647.81±950.34mGy/cm (DLP), 69.47±14.81mGy(CTDIvol), Effective dose54.72±14.26mSv and cancer probability per million(28.03) ureters and (27.05)for testiles.

Radiation dose can vary considerably between scanner and between institutions. Clinical dose are reported as the dose to standard dosimetry phantom. However due to large variation in patient size these dose may not estimate accurately the deliver to patient during a particular exam.In this study the radiation dose isconsidered high compared with previous studies, this may be due to scanner, or protocol used. A patient radiation risk for particular exam is proportional to the dose delivered during exam.

Chapter one

Introduction

Introduction:

1.1 Computed Tomography (CT) Imaging:

Johann Radon showed in 1917 that 2-D section images could be reformulated using mathematical transformation of projection data (i.e. using a Radon transform). Projection data are line integrals (summations of image values) recorded across an object at some angle. The link between projection data and x-ray images (maps of the effects of attenuation) was not obvious. However, motivation was high since section x-ray images would have the ability to make high contrast section images of the body by removing interference from overlapping tissues. Later in this chapter we will see how the projection dilemma was resolved. Even with the knowledge of how to make x-ray images into projections, imaging instrumentation and computing power was not able to provide this capability early on, so we had to wait many years for technology to catch up with the theory. By the 1960s several research labs were able to reconstruct x-ray section images from x-ray projections acquired from physical objects, and these successes spurred intensive research into devices that could be used in humans. In the 1970s x-ray computed tomography (CT) was formally introduced for clinical use, which was followed by rapid technological refinement. Since reconstructed images looked like the thinly sliced tissue sections used for microscopic inspection, the term "Tomography", literally meaning a picture of a cut section, was adopted, and early x-ray tomographic imaging systems were called Computed Axial Tomographic or “CAT” scanners. However, common use has dropped this designation in favor of computed tomography or just CT. In 1979

two early researchers in the field, A. M. Cormack and Godfrey Hounsfield, were jointly awarded the Nobel Prize for Computed Tomography.(unenhanced helical computerized tomography 2002).

Computed tomography (CT) is an imaging technique which produces a digital topographic image from diagnostic x-ray. In the early 1970s a major innovation was introduced into diagnostic imaging. This innovation, x-ray computed tomography (CT), is recognized today as the most significant single event in medical imaging since the discovery of x-rays. (unenhanced helical computerized tomography 2002).

Computed Tomography (CT) was invented by a British engineer, Sir Godfrey Hounsfield who also won the Nobel Prize because of his invention. CT was first introduced in the clinical practice in 1972 which was only limited to the brain scan. Prior to that, X-ray planar radiography and fluoroscopy systems were the main contributors of radiation in imaging. (William *et al* 2002).

Computed tomography (CT) is in its fourth decade of clinical use and has proved invaluable as a diagnostic tool for many clinical applications, from cancer diagnosis to trauma to osteoporosis screening. CT was the first imaging modality that made it possible to probe the inner depths of the body, slice by slice. Since 1972, when the first head CT scanner was introduced, CT has matured greatly and gained technological sophistication. Concomitant changes have occurred in the quality of CT images. The first CT scanner, an EMI Mark 1, produced images with 80 X 80 pixel resolution (3-mm pixels), and each pair of slices required approximately 4.5 minutes of scan time and 1.5 minutes of reconstruction time. Because of the long acquisition times required for the early scanners and the constraints of cardiac and respiratory motion, it was originally thought that CT would be practical only for head scans.

CT is one of the many technologies that were made possible by the invention of the computer. The clinical potential of CT became obvious during its early clinical use, and the excitement forever solidified the role of computers in medical imaging. Recent advances in acquisition geometry, detector technology, multiple detector arrays, and x-ray tube design have led to scan times now measured in fractions of a second. Modern computers deliver computational power that allows reconstruction of the image data essentially in real time. (Dahlmanp *et al* 2009). CT has fascinated the world with production of high contrast resolution images for visualizing soft tissues and the ability of producing tomographic and three dimensional (3D) volumetric images (IAEA 2007). Thus, it has changed the perception on medical diagnostic quality and as a result it has improved the quality of healthcare. Now, CT is becoming a common diagnostic tool in many major hospitals in the whole world. It is obvious that CT gives a lot of advantages such as faster scanning procedure, good spatial resolution and good contrast, compared to other modalities. Nowadays, many medical centers choose to send cases like accident and emergency cases, urology, cardiac imaging and pediatric imaging for CT scan as their first option for easy diagnosis of the symptoms. In some countries, sinusitis cases were likely referred to CT compared to the plain radiograph because CT were able to show important structures. Having taken notice of that, the manufacturers are also intense in introducing the latest technologies and applications of their CT due to the high demand of the CT scanners. (Goldman *et al* 2007).

1.2 Computed Tomography X-ray System in Medicine

Computed Tomography (CT) was invented by a British engineer, Sir Godfrey Hounsfield who also won the Nobel Prize because of his invention. CT was first

introduced in the clinical practice in 1972 which was only limited to the brain scan. Prior to that, X-ray planar radiography and fluoroscopy systems were the main contributors of radiation in imaging. (William *et al* 2002) .CT has fascinated the world with production of high contrast resolution images for visualizing soft tissues and the ability of producing tomographic and three dimensional (3D) volumetric images. Thus, it has changed the perception on medical diagnostic quality and as a result it has improved the quality of healthcare. Now, CT is becoming a common diagnostic tool in many major hospitals in the whole world. It is obvious that CT gives a lot of advantages such as faster scanning procedure, good spatial resolution and good contrast, compared to other modalities. Nowadays, many medical centers choose to send cases like accident and emergency cases, urology, cardiac imaging and pediatric imaging for CT scan as their first option for easy diagnosis of the symptoms. In some countries, sinusitis cases were likely referred to CT compared to the plain radiograph because CT were able to show important structures. (Goldman *et al* 2007) .Having taken notice of that, the manufacturers are also intense in introducing the latest technologies and applications of their CT due to the high demand of the CT scanners.

1.3 Radiation Issues in Computed Tomography

The distribution of X-ray in CT is different from planar radiography. In CT, a complete scan consists of thousands of radiation beams projected in circular directions around the object. It is very obvious that CT imparts relatively higher radiation dose than planar radiography. For example, a single routine CT of the chest has been identified to give radiation equivalent dose of 400 planar radiography of the chest. (Jerrold T *et al* 2002).

Council of European Union (1997) has clearly stated in the Council Directive 97/43/EUROTOM (June 1997) that CT produces the radiation as high as that of interventional radiology and radiotherapy:"Member States shall ensure that appropriate radiological equipment, practical techniques and ancillary equipment are used for the medical exposure- of children, - as part of a health screening program,- involving high doses to the patient, such as inter venation radiology, computed tomography or radiotherapy."

The increase number of CT scanners installed world wide has led to drastic increase of CT examinations. The contribution of radiation dose from CT examinations to the patients also increases and this has caused anxiety to the radiological communities. In the UK, it has been reported that CT constituted only about 2-3% of all radiological examination but it has contributed 20-30% to the total radiation dose in medical practices. (Galanski *et al* 2000). Until 10 years ago, there was about 35% increase of radiation dose from CT of the abdomen and pelvis in the UK (Rehani M, Berry M 2000) and this increase has made substantial impact on the patient care and, patient and population exposure from medical X-ray (Shrimpton et al 1991) . In the US, although CT comprised approximately 10% of total diagnostic radiological procedures, but it contributes approximately 65% of the effective radiation dose to the total national medical X-ray examinations. Based on the United Nations Scientific Committee on the Effects of Atomic Radiation report (European Commission1999), there was about 20% increase of global collective dose for 5 years 8 period (1985 to 1990 and 1991 to 1996). The number of CT examinations on children is also increasing. It has been reported that 2 – 3 millions of the CT examinations were performed on children every year (Jerrold T *et al* 2002). Noteworthy, children are more sensitive to the radiation than adults as their growth rates are faster. New advancement of the CT has also led to great increase of the

radiation dose to the patients. The use of multi-slice computed tomography (MSCT) has aggravated the scenario with the increasing of collective dose of CT examinations because the MSCT produces higher dose to the patients compared to single slice CT (SSCT)(unenhanced helical computerized tomography 2002) .

1.4 Radiation risk

The individual risk from radiation associated with a CT scan is quite small compared to the benefits that accurate diagnoses and treatment can proved. Still, unnecessary radiation exposure during medical procedures should be avoided. Unnecessary radiation may be delivered when CT scanner parameters are not appropriate adjusted for the patient size (Mettler FA *et al* 2000).

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 burns, 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. (Mettler FA *et al* 2000).

CT was always considered a “high dose” technique, there is growing realization that image quality in CT often exceeds the level needed for confident diagnosis and that patient doses are higher than necessary. (UNSCEAR2000).

In conventional X-ray procedure, medical personnel can tell if the patient has been overexposed because of the film is overexposed, produce a dark image (ICRP 2006). However, with CT there is no obvious evidence that the patient has been overexposed because the quality of the image may not be compromised (JerroldT *et al* 2002). The United Nation Scientific Committee on the Effects of Atomic Radiation (European Commission 1999) has highlighted that the worldwide there about 93 million CT examination performed annually at a rate of about 57 examination per 1000 persons. UNSCEAR also estimated that CT constitutes about 5% of all X-ray examination worldwide will accounting for about 34% of the resultant collective dose. In the countries that were identified as having the highest levels of healthcare, the corresponding figures were 6% and 41% respectively. New advancement of the CT has also led to great increase of the radiation dose to the patients. The use of multi-slice computed tomography (MSCT) has aggravated the scenario with the increasing of collective dose of CT examinations because the MSCT produces higher dose to the patients compared to single slice CT (SSCT) (unenhanced helical computerized tomography 2002).

1.5 Health Effects from Exposure to Ionizing Radiation:

The individual risk from radiation associated with a CT scan is quite small compared to the benefits that accurate diagnosis and treatment can provide. Still, unnecessary radiation exposure during medical procedures should be avoided. Unnecessary radiation may be delivered when CT scanner parameters are not appropriately adjusted for the patient size (Mettler FA *et al* 2000). In conventional X-ray procedures, medical personnel can tell if the patient has been overexposed because of the film is overexposed, producing a dark image [ICRP 2006]. However, with CT there is no obvious evidence that the patient has been overexposed because the quality of the image may not be compromised. Several recent articles (Galanski *et al* 2000,Richar D 2009)stress that it is important to use the lowest radiation dose

necessary to provide an image from which an accurate diagnosis can be made, and that significant dose reductions can be achieved without compromising clinical efficacy.

The United Nation Scientific Committee on the Effects of Atomic Radiation (UNSCEAR 2000) has highlighted that the worldwide there about 93 million CT examinations performed annually at a rate of about 57 examinations per 1000 persons. UNSCEAR also estimated that CT constitutes about 5% of all X-ray examinations worldwide while accounting for about 34% of the resultant collective dose. In the countries that were identified as having the highest levels of healthcare, the corresponding figures were 6% and 41% respectively.

The doses tissues from CT can often approach or exceed levels known to increase the probability of cancer, technologists are responsible for managing the dose in collaboration radiologists and medical physicists, CT examinations are increasing in frequency, newer CT techniques have often increased doses when compared with standard CT, referring physicians and radiologists should make sure that the examination is indicated and many practical possibilities currently exist to manage dose. The most important is reduction in mA.

1.6 CT in Sudan

In Sudan, the numbers of CT scans are increasing rapidly. The first ever CT scan was installed in The Military Hospital in 1981 followed another machine installed in Modern Medical Center, Khartoum in 1984. Since that data, the number of CT scan are more than 42 scanners, varying between 2 slice and 64 slice and one of them are 128 slice and all the rest are spiral CT.

1.7 Urography:

Diagnostic radiography is an acceptance imaging modality for the diagnostic of pathological conditions in human. Intravenous urography (IVU) is a radiographic study of the urinary system disorders. It is useful in the detection of renal and ureteral calculi. IVU has been a major and first choice method for diagnosing urinary system disease, since its emergence in the medical field in 1923.

It provides structural as well as functional information of the urinary tract. Despite the widespread use of advanced imaging modality (e.g ultrasonography, nuclear medicine , CT and MRI) ,intravenous urography examination still has a leading role in imaging urinary tract disorders especially in the developing countries. however ,during the procedure patients are exposed to significant radiation dose. (Keith J *et al* 2010).

1.8 CT Urography

The helical computed tomography (CT), the radiologic evaluation of patients with urologic disease has changed rapidly. Two major approaches to CT urography have been developed. The first approach combines axial CT with timed excretory urography (EU) performed by using conventional radiography, digital radiography, or CT scanned projection radiography (SPR). This approach produces traditional projection urograms, and the timed imaging technique is familiar to radiologists and clinicians. Additional excretory phase CT can be performed when the EU findings are positive or indeterminate. Improved CT SPR processing technology produces radiograph like images, thus eliminating patient transportation between the CT and urography suites or the necessity for a CT suite with a ceiling-mounted x-ray tube and a modified CT tabletop for performance of EU. The second approach to CT

urography combines axial CT with thin-section excretory phase CT. The near - isotropic volume data set enables creation of high-resolution two and three-dimensional reformatted images. However, the increased amount of radiation and the time required for data manipulation are concerns. Further studies evaluating large numbers of patients with various urothelial abnormalities will be necessary to determine the optimal CT urography technique for clinical practice.

1.8.1 Radiation risk in CT urography

One of the most significant potential risk to patient who are examined with diagnostic CT is an increased the probability of cancer due to the radiation exposure from X-ray. This risk is known as a stochastic effect. A stochastic effect is defined as one in which the probability of occurrence rather than the severity of the effect, is proportional to the radiation dose. Stochastic effect has no threshold dose. In general radiation doses from diagnostic x-ray imaging exam are considered to be low and there still question as to whether sufficient evidence exists that establishes the risk of cancer at these low dose never the less, in clinical practice, current radiation protection standard assume that there is no threshold dose below which the risk of cancer induction is zero. This linear non threshold model substantiated in the report by the committee on biological effect of ionizing radiation is used most often in medical applications when considering radiation risk to patients. (Kalender WA 1999).

1.9 problem of the study:

Radiation dose measurements are important in order to assess the possible risks from diagnostic procedures. CTU is one of the most common procedures due to their numerous indications. During these procedures, patients are exposed frequently to radiation.

1.10 Objectives of The Study:

1.10.1 General Objective:

The main objective of the study is to assess optimum patients radiation dose in a CTU in depart of Military Hospital.

1.10.2 Specifics Objective: To

- Measure effective dose (ED).and

- Estimate the radiation risks for patients undergoing CTU procedures.

This thesis is concerned with the assessment of radiation dose for patients during CT urography.

Accordingly, it is divided into the following chapters:

Chapter one is the introduction to this thesis. This chapter presents the historical background and radiation risks, in addition to study problem, objectives and scope of the work. It also provides an outlines of the thesis. Chapter two contains the background material for the thesis. Specifically it reviews the dose for all absorbed dose measurements and calculations. This chapter also includes a summary previous work performed in this field. Chapter three describes the materials and methods that used to measure dose for CT machines and explains in details the methods for calculation and optimization. Chapter four presents the results of this study. Finally Chapter five presents the discussion, conclusion and recommendations of this thesis and presents the suggestions for future work.

Chapter Two

Literature review

2.1 Theoretical background:

2.1.1 The Urinary System:

The urinary system is also called the excretory system of the body because one of its functions is to remove waste products from the blood and eliminate them from the body. The urinary system consists of:

Two kidneys: this organ extracts wastes from the blood; balance body fluids and form urine, Two ureters: this tube conducts urine from the kidneys to the urinary bladder, The urinary bladder: this reservoir receives and stores the urine brought to it by the two ureters, And The urethra: this tube conducts urine from the bladder to the outside of the body for elimination.

The major functions of the urinary system Excretion of wastes, Hormonal production (rennin-angiotensin and erythropoietin) and Acid base balancing

External Anatomy of the kidney:

A pair of reddish brown, bean shaped organ located in the posterior wall of the abdominal region, one in each side of the vertebral column. They usually span between T12 to L3. They are protected at least partially by the last pair of ribs and capped by the adrenal gland. The bean shape of the kidney is medially concave and laterally convex. On the medial concave border is the hilus (small indented area) where blood vessels, nerves & ureters enter and leave the kidney.

Covering and supporting each kidney are three layers of tissue:

Renal capsule – innermost, Adipose capsule – the middle layer composed of fat and Renal fascia – is outer sub-serous membrane, connective tissue layer. (intravenous urography 2001).

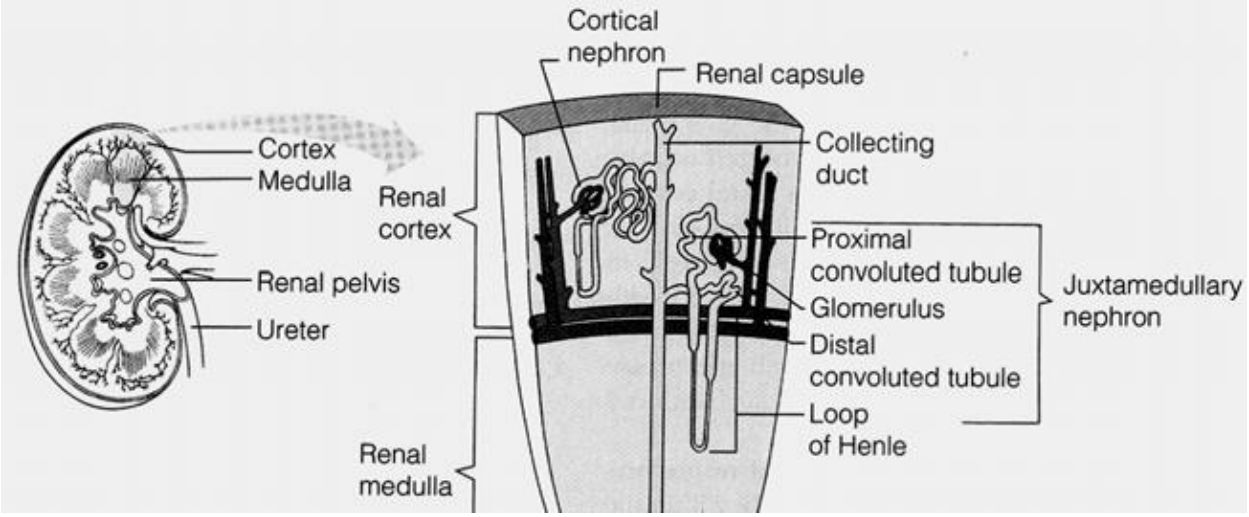

Figure: (2.1) Internal structure of the kidney (source: Grollman Sigmund, (1969), The human body it's structure and physiology, London, The Macmillan company, 2nd ed)

Internal Anatomy of the kidney:

A sagital section of the kidney reveals three distinct regions called pelvis, medulla and cortex (from inside out). The renal pelvis is the large collecting space with in the kidney formed from the expanded upper portion of the ureters. The pelvis branch to two cavities, these are 2-3 major calyces and 8 to 18 minor calyces.

The Renal medulla is the middle portion of the kidney. It consists of 8 to 18 renal pyramids, which are longitudinally striped, one cone shaped area. The base of

each pyramid is adjacent to the outer cortex. The apex of each renal pyramid ends in papilla, which opens to a minor calyx. Pyramids contain tubules and collecting ducts of the nephron. Tubules involved in transportation and re-absorption of filtered materials. The renal cortex is the outermost portion of the kidney. It is divided in to two region the outer cortical and the inner juxtamedullary region. The cortical tissue that penetrates between pyramids forms ***Renal Columns***. The renal columns composed of mainly collecting tubules.

2.1.2 The CT scanner 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 and the power distribution unit and the data transfer unit.

The Gantry which is a central opening 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 angulations motors and positioning laser lights. A CT gantry can be angled up to 30 degrees toward a forward or backward position.

The Table is where the patients is positioned (lie down), and it moves through the gantry. The patient's table and the gantry constitute CT scanner itself, The power Distribution unit supplies power to the gantry, the patient's table and the computers of the Computing System, which is localized in a separate room as will be explained next.

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 post-processing is also necessary, so it possible to analyze and review previous exam data, without interfering with the current examinations taking place.(Healthcare.philips)

Figure (2.2) CT scanner (Healthcare. philips)

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. (Healthcare.philips)

The gantry includes the x-ray tube, the detector array, the high-voltage generator, the patient support couch, and the mechanical support for each. These subsystems receive electronic commands from the operating console and transmit data to the computer for image production and post-processing tasks. *X-ray Tube.* X-ray tubes used in multi-slice helical CT imaging have special requirements. Multi-slice helical CT places a considerable thermal demand on the x-ray tube. The x-ray tube can be energized up to 60 s continuously.

Although some x-ray tubes operate at relatively low tube current, for many, the instantaneous power capacity must be high. High-speed rotors are used in most for the best heat dissipation. Experience has shown that x-ray tube failure is a principal cause of CT imaging system malfunction and is the principal limitation on sequential imaging frequency. Focal-spot size is also important in most designs even

though the CT image is not based on principles of direct projection imaging. CT imaging systems designed for high spatial resolution imaging incorporate x-ray tubes with a small focal spot. Multi-slice helical CT x-ray tubes are very large. They have an anode heat storage capacity of 8 MHU or more. They have anode-cooling rates of approximately 1 MHU per minute because the anode disc has a larger diameter, and it is thicker, resulting in much greater mass.

The limiting characteristics are focal-spot design and heat dissipation. The small focal spot must be especially robust in design. Manufacturers design focal-spot cooling algorithms to predict the focal-spot thermal state and to adjust the mA setting accordingly. One company has produced a revolutionary x-ray tube in which the whole insert rotates in a bath of oil during an exposure. The beam of electrons is deflected onto the anode in a process similar to that seen in a cathode ray tube. The result is that it can withstand up to 30 million heat units and cools at a rate of 5 million heat units per minute .*Detector Array.* Multi-slice helical CT imaging systems have multiple detectors in an array that numbers up to tens of thousands**.** Previously, gas-filled detectors were used, but now, all are scintillation, solid state detectors. Sodium iodide (NaI) was the crystal used in the earliest imaging systems. This was quickly replaced by bismuth germanate (Bi4Ge3O12 or BGO) and cesium iodide (CsI). Cadmium tungstate (CdWO4) and special ceramics are the current crystals of choice. The concentration of scintillation detectors is an important characteristic of a CT imaging system that affects the spatial resolution of the system. Scintillation detectors have high x-ray detection efficiency. Approximately 90% of the x-rays incident on the detector are absorbed, and this contributes to the output signal. It is now possible to pack the detectors so that the space between them is nil. Consequently, overall detection efficiency approaches 90%. The efficiency of the x-ray detector array reduces patient radiation doses, allows faster imaging time, and improves image quality by increasing signal-to-noise ratio. Detector array

design is especially critical for multi-slice helical CT. *Collimation.* Collimation is required during multi-slice helical CT imaging for precisely the same reasons as in conventional radiography. Proper collimation reduces patient radiation dose by restricting the volume of tissue irradiated. Even more important is the fact that it improves image contrast by limiting scatter radiation. In radiography, only one collimator is mounted on the x-ray tube housing. In multi-slice helical CT imaging, two collimators are used .One collimator is mounted on the x-ray tube housing or adjacent to it. This collimator limits the area of the patient that intercepts the useful beam and thereby determines the patient radiation dose. This pre-patient collimator usually consists of several sections, so a nearly parallel x-ray beam results. when properly coupled with the pre-patient collimator, defines the slice thickness, also called the *sensitivity profile*. The pre-detector collimator reduces scatter radiation that reaches the detector array, thereby improving image contrast. The pre-detector collimator restricts the x-ray beam viewed by the detector array. This collimator reduces the scatter radiation incident on the detector array and, *High-Voltage Generator.* All multi-slice helical CT imaging systems operate on high-frequency power. A high-frequency generator is small because the high-voltage step-up transformer is small, so it can be mounted on the rotating gantry. The design constraints placed on the high-voltage generator are the same as those for the x-ray tube. In a properly designed multi-slice helical CT imaging system, the two should be matched to maximum capacity. Approximately 50 kW power is necessary. *Patient Positioning and the Support Couch.* In addition to supporting the patient comfortably, the patient couch must be constructed of low-Z material, such as carbon fiber, so it does not interfere with x-ray beam transmission and patient imaging. It should be smoothly and accurately motor driven to allow precise patient positioning that is unaffected by the weight of the patient. When patient couch positioning is not exact, the same tissue can be imaged twice, thus doubling the radiation dose or it can

be missed altogether. The patient couch is indexed automatically, so the operator does not have to enter the examination room between imaging sequences. Such a feature reduces the examination time required for each patient.

The computer is a unique subsystem of the CT imaging system. Depending on the image format, as many as 250,000 equations must be solved simultaneously; thus, a large computing capacity is required. At the heart of the computer used in CT are the microprocessor and the primary memory. These determine the time between the end of imaging and the appearance of an image—the reconstruction time. The Many CT imaging systems use an array processor instead of a microprocessor for image reconstruction. The array processor does many calculations simultaneously and hence is significantly faster than the microprocessor. (Healthcare.philips).

2.1.3 scan Parameters:

In order to properly calculate and compare doses, it is imperative to have a standardized nomenclature to ensure that all data is comparative .Without this, it will be difficult tore produce measurements, and to develop consistent protocols. When performing a CT examination, a number of parameters are defined by the operator. The thesis will cover the parameters deemed important for correct, uniform dosimetry: tube current, tube voltage, rotation time, total scan length, slice thickness and pitch. Automatic exposure control (AEC) and iterative reconstruction will be briefly covered, as their impact on dose and image quality is more of a qualitative influence than a quantitative one. (Healthcare.philips).

2.1.3.1 Tube current:

The tube current [mA] influences the number of photons exiting the X-ray tube, as it determines the number of electrons leaving the cathode. The tube current is directly proportional to radiation dose, and as such is a prime parameter in

adjusting the dose. Instead of tube current is sometimes used the tube-current-time-product [mAs], which is the tube current multiplied with the scan time. (Rihar D2009).

2.1.3.2 Tube Voltage:

The tube voltage [kV] determines the voltage across the anode and cathode of the X-ray tube, and therefore the acceleration of the electrodes across the interior vacuum. This determines the kinetic energy of the electrodes when they reach the anode, and therefore the number of interactions they can initiate before being absorbed. As a consequence, an increase in tube voltage will increase the dose, all other factors kept constant; however, the increase is not directly proportional as was the case with current. Voltage determines the energy of the electrons, and therefore the energy distribution of the incident X-rays. It is rarely adjusted from the customary value of 120 kV. Certain examinations use a different voltage, but seldom outside the range of 80 to 140 kV. (Richar D 2009).

2.1.3.3 Rotation Time:

The rotation time of the gantry [s] has decreased greatly over the last few decades; with modern scanners, which having a rotation time in the area of 0.4 seconds. The main consequence of the decreased rotation time is an increase in the noise and a reduction in absorbed dose. To avoid the noise, it is customary to increase the tube current accordingly. (Richar D 2009).

2.1.4 Examination parameters:

2.1.4.1 Total Scan Length:

It is apparent that the total scan length [cm] influence the absorbed dose, as an increase in scan length will expose a larger part of the patient to radiation. Therefore, it is imperative that scan length is to be limited to cover just the diagnostically relevant part of the patient; otherwise, an unnecessary increase in dose

will be seen (ICRP, 2000). This is relatively easy with SSCT; however, the situation is more complicated for MSCT. At the initiation of the scan, the X-ray tube will be activated the moment the first row of detectors reach the diagnostic area. The X-ray beam will irradiate the entire detector-array, but only the first row of detectors will be acquiring image data. The remaining detector rows will not acquire data, but the area will still be irradiated. This is called over scan, and a small degree of over scan is required for correct reconstruction. As the table moves, more rows of detectors are entering the diagnostic area, contributing to the image. At the reverse end of the patient, the same scenario occurs, and a noteworthy part of the dose is absorbed in the patient outside the diagnostic area. (Richar D2009).

2.1.4.2 Slice Thickness:

In SSCT, with only a single row of detectors, the slice thickness [cm] is determined by simple collimation. The maximum slice thickness is limited by the width of the individual detector element (typically 10 mm), and by collimating the beam, this thickness can be decreased. In other words, the width of the beam is equal to slice thickness. In MSCT, the width of each individual detector element in the longitudinal direction determines the minimum slice thickness, and by merging multiple adjacent detector elements during detection, one can increase the slice thickness. This has a significant impact on image quality, as thin slices have better spatial resolution compared to thick slices, but lower SNR. To address the decrease in SNR, it is necessary to increase for instance the tube current, resulting in a significant increase in dose to the patient. (Richar D 2009) .

2.1.4.3 Pitch:

With the prevalence of helical MSCT, it is necessary to incorporate the incremental movement of the table, in relation to the irradiated area. This is defined as pitch, being the increment of the table per rotation, divided by the width of the

beam. In Figure 2.3 below, a 4-slice MSCT is rotated twice around the patient, resulting in the acquisition of eight slices in pairs of two (indicated by color). The slices are in reality at an incline, as the patient is moving during exposure. (Richar D 2009).

pitch = table feed per rotation/collimation

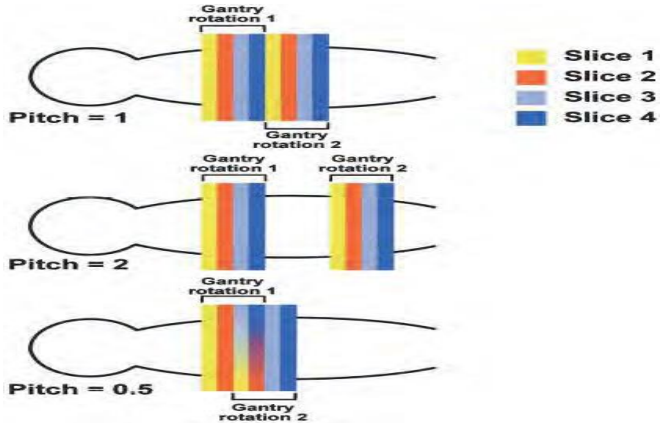

Figure (2.3) the effect of pitch on irradiated area, with a overlap for pitch < 1 (Healthcare.philips)

2.1.5 Radiation dose units:

The specific units of measurement for radiation dose commonly referred to as effective dose (mSv). Other radiation dose measurement units include; Rad, Rem, Rontgen, and Sievert. Because different tissues and organs have varying in sensitivity to radiation exposure, the actual effective dose to different parts of the body for X-ray procedure varies. The term effective dose is used when referring to the dose averaged over the entire body. The effective dose accounts for the relative sensitivities of different tissues exposed. More importantly, it allows for qualification of risk and comparison to more familiar sources of exposure that range

from natural background radiation to radiographic medical procedure. As with other medical procedures, X-rays are safe when used with care. Radiologists and X-ray technologists have been trained to use the minimum amount of radiation that is necessary to obtain the needed results. The decision to have an X-ray examination is a medical one, based on the likelihood of benefit from the examination and the potential risk from radiation. (Healthcare.philips).

During the early days of radiological experience there was no precise unit of radiation does that was suitable either for radiation protection or for radiation therapy. For purposes the radiation protection, a common "dosimeter" was a piece of dental film with a paper clip attached. A daily exposure great enough to just produce a detectable shadow was considered a maximum permissible dose. For greater does and for therapy purposes the dose unit was frequently the "skin erythema unit". Because of the great energy dependence of the dose units could be biologically meaningful or useful either in quantitative study of the biological effects of radiation or for radiation protection purposes. Furthermore, since the fraction of the energy in a radiation field that is absorbed by the body is dependent, it is necessary to distinguish between radiation exposure and radiation absorbed dose. (ICRP 60: 1990 recommendation)

2.1.5.1 Absorbed dose:

Absorbed dose is a non-stochastic quantity, defined as the expectation value of the energy imparted to matter, ε, per unit mass of tissue at the point of interest dm.

$$D = d\varepsilon/dm \qquad (2.1)$$

Radiation damage depends on the absorption of energy from the radiation and is approximately proportional to the concentration of absorbed energy in tissue.

2.1.5.2 Gray:

The basic unit of radiation dose called the gray (Gy) and is defined as: one gray is an absorbed radiation dose of one joule per kilogram. The gray is universal applicable to all types of ionizing radiation dosimetry.

2.1.5.3 Rad:

Before the universal absorption of the SI units, radiation dose was measured by a unit called the rad (Radiation Absorbed Dose). One rad is an absorbed radiation dose of 100 ergs per gram.

$$1 \text{ rad} = 100 \text{ergs/g} \qquad (2.2)$$

Since 1 J=10^7 ergs, and since 1kg=1000g, 1Gy=100 rads.

Although the gray is the newer unit, and will eventually replace the rad.

2.1.5.4 kerma:

Kerma is a non-stochasitc quantity, defined as the expectation value of the energy transferred (εtr) by uncharged particles (e.g. photons or neutrons) to charge particles per unit mass at the point of interest dm

$$K = d\varepsilon \, tr/dm \qquad (2.3)$$

Kerma has been defined as, and is an acronym for, the sum of the kinetic energies of all those primary charged particles released by uncharged particles (here photons) per unit mass (Kinetic Energy Released per unit mass) the unit of kerma is grey (Gy), where 1 Gy=1j kg-1.

In a photon field, the kerma at the point of interest is expressed as

$$K = \int_{E=0}^{Emax} \Psi(E)\frac{\mu tr}{p} \quad (2.4)$$

Where Ψ(E) is the distribution of photon energy fluence and$\frac{\mu tr}{p}$ is the mass energy –transfer. Photon energy fluence is defined as the product photon fluence and energy E.

Kerma is greater than absorbed dose by a factor of 1/ (1-g). this relation is valid only for irradiation in the condition of charged particle equilibrium i,e when the number and energies of charged particles leaving is equal to the number and energies of particles entering this volume.

$$D=(1-g)K \quad (2.5)$$

The factor g represent the average fraction of the kinetic energy of secondary charged particles (produced in all types of interactions) that is subsequently lost in radiative (photon emitting) energy-loss processes as the particles slow to rest in the medium. (D Tack 2007)

2.1.5.5 Exposure:

Exposure is a radiation quantity referring to the intensity of radiation for external radiation of any give energy flux, the absorbed to any point with in an organism depends on the types and the energy of radiation, the depth within the organism of the point at which the absorbed dose is required, and elementary constitution of the absorbing medium at this point. The exposure unit is a measure of photon flux, and is related to the amount of energy transferred from the X-ray field to a unit mass of air. One exposure unit is defined as that quantity of x-or gamma radiation that produces in air, ions carrying 1 coulomb of charge (of either sign) per Kg air.

1x unit= 1c/Kg air.

The exposure unit is based on ionization of air because of the relative ease with which radiation induced ionization can be measured. The exposure unit bay be converted in to more fundamental unit of energy absorption per unit mass of air by using the charge on a single ion is 1.6x10-19 coulombs and that the average energy dissipated in the production of a single ion pair in air is 34ev.(Healthcare.philips).

Therefore:

1x unit= $\frac{c}{\text{Kg air}}$x $\frac{1\text{ ion}}{1.6\text{x}10-19}$ x 34 eV/ion x 1.6x10-19 j/ev x 1Gy/j/Kg=34Gy (in air).

2.1.5.6 Equivalent dose:

Equal doses of all types of ionizing radiation are not equally harmful. Alpha particles produce greater harm than do beta particles, gamma rays and x rays for a given absorbed does. To account for this difference, radiation does is expressed as equivalent does. The equivalent does (HT) is a measure of the radiation does to tissue where an attempt has been made to allow for the different relative Biological effects of different types of ionizing radiation. Equivalent does is therefore a less fundamental quantity than radiation absorbed does, but is more biologically significant. Equivalent does has units of Sieverts (Sv). Another unit, rontgen equivalent man (REM or rem), is still in common use in the US, although regulatory and advisory bodies are encouraging transition to Sieverts (100 Rontgen equivalent man=100 REM= 1 sievert). (Healthcare.philips)

Equivalent does (HT) is calculated by multiplying the absorbed does to the organ or tissue (DT) with the radiation weighting factor, WR. This factor is selected for the type and energy of the radiation incident on the body, or in the case of sources

within the body, emitted by the source. The value of WR is 1 for x-rays, gamma rays and beta particles, but higher for protons, neutrons, alpha particles etc.).

$$H_{T,R} = W_R \times D_{T,R} \qquad (2.6)$$

Where $H_{T,R}$ = equivalent dose to tissue T from radiation R

$D_{T,R}$ = absorbed dose D (in grays) to tissue T from radiation R

The dose in Sv is equal to "absorbed dose" multiplied by a "radiation weighting factor" (WR – see Table 2.1 below). Prior to 1990, this weight factor was referred to as Quality Factor (QF).

Radiation Weight Factors of x-ray is one according to ICRP 60: 1990 recommendations.

2.1.5.7 Effective dose:

Effective dose equivalent (Now replaced by Effective Dose) is used to compare radiation doses on different body parts on an equivalent basis because radiation does not affect different parts in the same way. The effective dose is the sum of weighted equivalent doses in all the organs and tissues of the body.).

Effective dose = sum of [organ doses x tissue weighting factor]. (Healthcare.philips).

The effective dose (E) to an individual is found by calculating a weighted average of the equivalent dose (H) to different body tissues, with the weighting factors (W) designed to reflect the different radio-sensitivities of the tissues:

$$E = \sum_i H_i * W_i \qquad (2.7)$$

The unit for effective dose is the sievert (Sv).

Table (2.1) represent relative sensitivity of organs for developing cancer.

Tissue or Organ	Tissue Weighting Factor (WT)
Gonads (testes or ovaries)	0.20
Red bone marrow	0.12
Lung, stomach and colon	0.12
Thyroid gland, breast, bladder, Esophagus , Liver and Remainder**	0.05
Skin, Bone surfaces	0.01
Whole body	1.00

One sievert is a large dose. The effects of being exposed to large doses of radiation at one time (a cute exposure) vary with the dose. Here are some.

2.1.5.8 occupational exposure:

Although CT presents only a small percentage of radiology examinations, it results in a significant portion of the effective radiation dose from medical procedures; (I) with the increasing use of CT for screening procedures, (II) and advances in scanner technology, they tend for increasing numbers of procedures performed with this modality may increase. Although CT is clearly providing many clinical benefits, the motivation to understand radiation dose in general as well as the specific concepts related to CT grows with prevalence of this modality. (Healthcare.philips).

2.1.6 CT dose descriptors:

The radiation exposure to the patients undergoing CT examinations is determined by two factors: equipment-related factors, .e. the design of the scanner with respect to dose efficiency, and applications-related factors, i.e. the way in which the radiologist and X-ray technologist makes use of the scanner. In this chapter the features and parameters influencing patient dose are outlined. First, however, a brief introduction on the dose descriptors applicable to CT is given.

The dose qualities used in this projection radiography are not applicable to CT for three reasons. (Stewart 10^{th} edition).

First, the dose distribution inside the patient is completely different from that of a conventional radiography where the dose decreases continuously from entrance of the X-ray beam to its exit, with the ratio of between 100 and 1000 to 1. In the case of CT, as a consequence of the scanning procedure that equally irradiates the patient from all directions; the dose is almost equally distribution in the scanning plane. A dose comparison of CT with conventional projection radiography in term of skin dose therefore does not make any sense.

Second, the scan procedure using narrow beams along the longitudinal z-axis of the patient implies that a significant portion of the radiation energy is deposited outside the nominal beam width. This is mainly due to penumbra effects and scattered radiation produced inside the beam.

Third, the situation with CT is further complicated by the circumstances in which-unlike in conventional projection radiography-the volume to be imaged is not irradiated simultaneously. This often leads to confusion about what dose from a complete series of e.g. 15 slices might be compared with the dose from a single slice.

As a consequence, dedicated dose quantities that account for these peculiarities are needed. The ‘Computed Tomography Dose Index (CTDI)’, which is a measure of the local dose, and the Dose Length Product (DLP), representing the integral radiation exposure associated with a CT examination. this can be achieved by the effective dose (E). So there are three dose descriptors in all.

2.1.6.1 Computed tomography dose index (CTDI):

The ‘Computed Tomography Dose Index (CTDI)’ is the fundamental CT dose descriptor. By making use of this quantity, the first two peculiarities of CT scanning are taken into account: The CTDI (unit: Milligray (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. (Fig.2.5) illustrates the meaning of the term: 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 with N being the number of independent (i.e. non-overlapping) slices that are acquired simultaneously. Accordingly, all dose contributions from outside the nominal beam width, i.e. the areas under the tails of the dose profile, are added to the area inside the slice. (Goldman *et al* 2007).

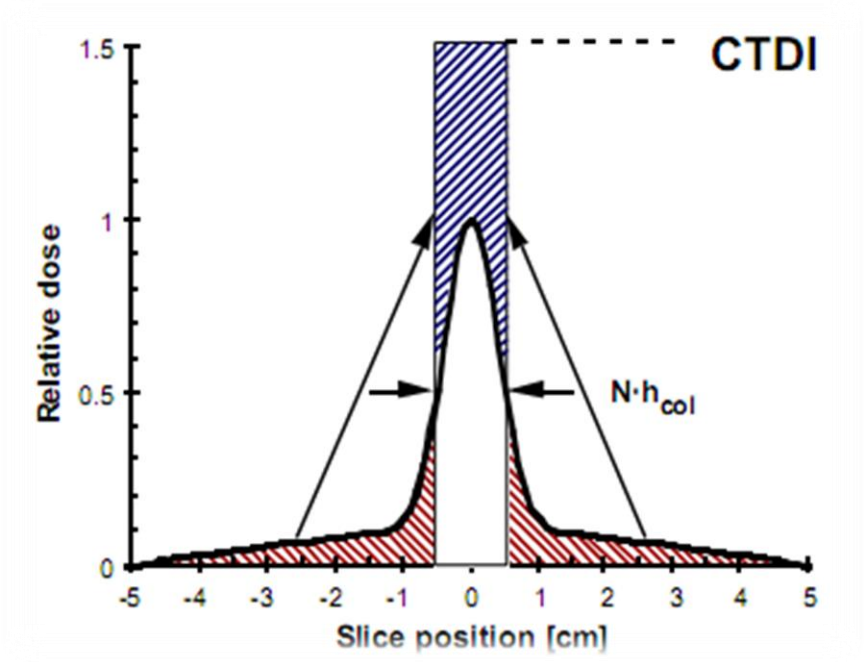

Figure: (2.4): Illustration of term 'Computed Tomography Dose Index (CTDI)': is the equivalent of the dose value inside the irradiated slice (beam) that would result if the absorbed radiation dose profile were entirely concentrated to a rectangular of width equal to the nominal beam width N.hcol, with N being the number of independent (i.e. non-overlapping) slices that are acquired simultaneously (Goldman *et al* 2007).

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

$$\mathrm{CTDI} = 1 \div \mathrm{N.hcol.} \int_{-\infty}^{+\infty} \mathrm{D(z).dz} \qquad (2.8)$$

Where D(z) is the value of the dose at a given location, z, and 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 the nominal beam width. In practice, the dose profile is accumulated in a range of -50 mm to +50 mm relative to the center of the beam, i.e. over a distance of 100mm.

The relevancy of CTDI becomes obvious from the total dose profile of a scan series with e.g. n=15 subsequent rotations (Fig.2.6).The average level of the total dose profile, which is called 'Multiple Scans Average Dose (MSAD), is higher than the peak value of each single dose profile. This increase results from the tails of the single dose profiles. Obviously MSAD and CTDI are exactly equal of the table feed (TF) is equal to the nominal beam width N.hcol, i.e. if the pitch factor

$$P = \frac{TF}{N.hcol} \tag{2.9}$$

is equal to 1. In general (i.e. if the pitch factor is not equal to 1, Fig.2.3), the relationship between CTDI and MSAD is given by:

$$MSAD = {}^{1}/_{P.CTDI} \tag{2.10}$$

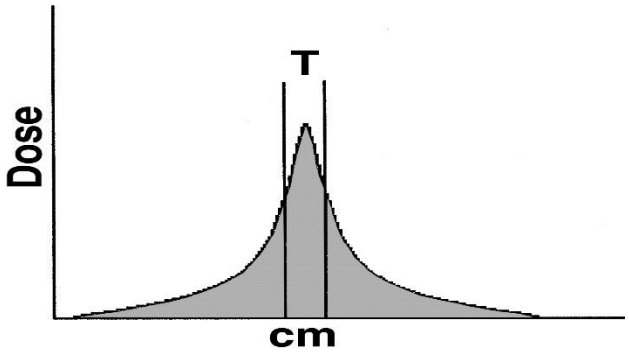

Figure: (2.5): The average level of the total dose profile, which is called 'Multiple Scans Average Dose (MSAD)'- (Shope 1981), is higher than the peak value of each single dose profile. This increase results from the tails of the single dose profiles (Goldman *et al* 2007).

Each pair of CTDI (central and peripheral) can be combined into a single are named weighted CTDI (CTDIw):

$$CTDI_W = \frac{1}{3}CTDI_{100_c} + \frac{2}{3}CTDI_{100_P} \tag{2.11}$$

If pitch-related effects on radiation exposure are taken into account at level of local dose (i.e. CTDI) already, a quantity named volume CTDI (CTDIvol)' is defined:

$$CTDIvol = CTDIw/P \quad (2.12)$$

So CTDIvol is the pitch-corrected CTDIw. Apart from the integration length, which is limited to 100 mm, CTDIvol is practically identical to MSAD based on CTDIw (i.e.MSADw). Since averaging includes both the cross section and the scan length, CTDIvol therefore represents the average dose for a given scan volume. CTDIvol is used as the dose quantity that is displayed at the operator's console of newer scanners. (Goldman et al 2007).

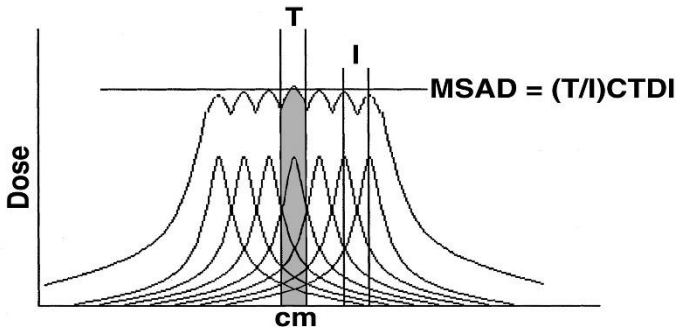

Figure: (2.6): (1) **Schematic illustrates the profile of radiation dose delivered during a single CT scan. The CTDI equals the shaded area under the curve divided by the section thickness (*T*). (**2) **Schematic illustrates the profile of radiation dose delivered during multiple CT scans. *T* represents section thickness, and *I* represent the interval between sections. The MSAD includes the contributions of neighboring sections to the dose of the section of interest (Stewart 10th edition).**

2.1.6.2 Dose length product (DLP) unit (mGy):

DLP = CTDIw**.** L(mGy-cm). DLP takes both the 'intensity') represented by CTDIvol) and the extension (represented by scan length L) of an irradiation into account:

$$DLP = CTDI_{vol} \cdot scan\ length \quad \textbf{(}2.13\textbf{)}$$

So DLP increases with number of slices (correctly: with length of irradiated body section), while the dose (i.e. CTDIvol) remains the same regardless of the number of slices or length, respectively. The area of the total dose profile of the scan series represents the DLP. DLP is the equivalent of the dose-area product (DAP) in projection radiography, a quantity that also combines both aspects (intensity and extension) of patient exposure. In sequential scanning, the scan length is determined by the beam width N.hcol and number of the table feed (TF):

$$L = n*TF + N.hcol \quad \textbf{(}2.14\textbf{)}$$

While in spiral scanning the scan length only depends on the number (n) of rotations and the table feed (TF):

$$L = n \cdot TF = {}^{T}\!/_{t_{rot}} \cdot p \cdot N \cdot h_{col} \quad \textbf{(}2.15\textbf{)}$$

Where T is the total scan time, trot is the rotation time, and p is the pitch factor. While in sequential scanning the scan length L is equal to the range from the begin of the first slice till the end of the last, the (gross) scan length for spiral scanning not only comprises the (net) length of the imaged body section but also includes the additional rotations at the begin and the end of the scan ('over-ranging') that are

required for data interpolation [European Commission 1999]. If an examination consists of several sequential scan series or spiral scans, the dose-length product of the complete examination (DLP exam) is the sum of the dose-length products of each single series or spiral scan:

$$DLP_{exam} = \sum_{i} DLP_i \quad (2.16)$$

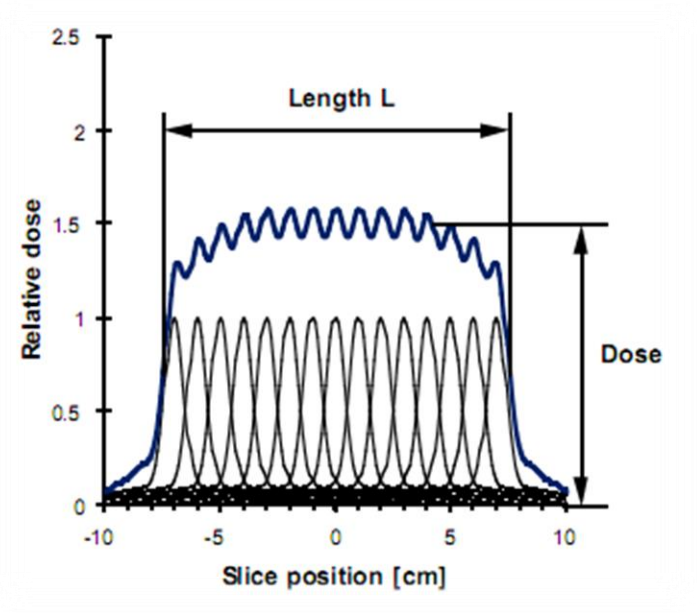

Figure (2.7): Dose length product (DLP) in CT (Total dose profile of a scan series with n=15 sub-sequent rotations. The dose-length product (DLP) is the product of the height (dose, i.e. CTDIvol) and the width (scan length L) of the total dose profile and is equal to the area under the curve (Jerrold T *et al* 2002).

2.6.3 DLP and Effective Dose:

CTDI and DLP are CT specific dose descriptors that do not allow for

comparisons with radiation exposure from other sources, projection radiography, nuclear medicine or natural background radiation. The only common denominator to achieve this goal is the (Effective Dose). With effective dose, the organ doses from a partial radiation of the body are converted into an equivalent uniform dose to the entire body. An effective Dose E unit (millisevert, mSv) according to ICRP 60 (30) is defined as the weighted average of organ dose values HT for a number of specific organs:

$$E = \sum Hi * Wi \qquad (2.17)$$

2.1.7 CT protocols:

Most CTU protocols are tri-phasic examinations that include non-contrast, enhanced, and delayed images.1-3 Non contrast images extending from the top of the kidneys through the bladder are obtained to evaluate for calculi, fat-containing lesions and parenchymal calcifications, and to provide baseline attenuation for assessment of lesion enhancement. Intravenous contrast is administered and, following a 90- to 100-sec delay, scanning of the abdomen and pelvis is performed during the nephrographic phase. Homogeneous enhancement of the kidneys during the nephrographic phase optimizes small renal mass detection. The final acquisition is during the excretory phase after a 12- to 15-min delay, when there is opacification and distention of the collecting systems, ureters, and bladder. Excretory images allow for evaluation of the urothelium. While diagnostic unenhanced and nephro-graphic acquisitions are relatively easy to obtain, optimal opacification and distention of the ureters during the excretory phase can be more problematic. Suboptimal distention of the ureters and intermittent peristaltic waves may result in limited visualization of one or more segments. A variety of techniques have been proposed to improve visualization, including oral and IV hydration, diuretic

administration, use of a compression belt, prone positioning, and a log-rolling prior to the excretory acquisition.

As an alternative to the tri-phasic single bolus CTU technique, the split contrast bolus technique has been designed whereby the contrast is given as 2 boluses before a single enhanced-scan is acquired.

The aimed cumulative effect is that the earlier smaller bolus provides excretory information, while the second, larger bolus provides information on vascular anatomy and the renal parenchyma. The benefit of this protocol is the elimination of an additional acquisition, resulting in a decreased radiation dose; the disadvantage is less reliable opacification of the ureters. While the ideal protocol is not agreed upon, many centers hydrate patients prior to contrast administration (oral and/or IV route) and inject furosemide IV at contrast injection. (Richar D2009).

2.1.8 CT KUB:

In medicine, **KUB** refers to a diagnostic medical imaging technique of the abdomen and stands for **K**idneys, **U**reters, and **B**ladder. A KUB is a plain frontal supine radiograph of the abdomen. It is often supplemented by an upright PA view of the chest (to rule out air under the diaphragm or thoracic etiologies presenting as abdominal complaints) and a standing view of the abdomen (to differentiate obstruction from ileus by examining gastrointestinal air/water levels).

Uses Despite its name, a KUB is not typically used to investigate pathology of the kidneys, ureters, or bladder, since these structures are difficult to assess (for example, the kidneys may not be visible due to overlying bowel gas.) In order to assess these structures radiographically, a technique called an intravenous

pyelogram was historically utilized, and today at many institutions CT urography is the technique of choice.

KUB is typically used to investigate gastrointestinal conditions such as a bowel obstruction and gallstones, and can detect the presence of kidney stones. The KUB is often used to diagnose constipation as stool can be seen readily. The KUB is also used to assess positioning of indwelling devices such as ureteric stents and nasogastric tubes. KUB is also done as a scout film for other procedures such as barium enemas. The KUB does not necessarily include the diaphragm. The projection includes the entire urinary system, from the pubic symphysis to the superior aspects of the kidneys. The anteroposterior (AP) abdomen projection, in contrast, does include the bilateral diaphragm. If the patient is large, more than one film loaded in the Bucky in a "landscape" direction may be used for each projection. This is done to ensure that the majority of bowel can be reviewed. (Le.ac.uk).

2.1.9 CTU technique:

Imaging of the upper urinary tract has traditionally been the purview of intravenous (IV) urography, but over the last decade, computed tomography urography (CTU) has become the modality of choice in imaging the urinary tract. With few exceptions, most notably that of the unenhanced CT performed for acute flank pain and stone disease, many urological symptoms and conditions are now investigated with CTU. Continuing improvements in the spatial resolution and speed of newer CT scanners, combined with advanced multi-planar and volume-rendered image reconstruction, have made CTU a comprehensive examination whereby the kidneys and upper collecting system, ureters, and urinary bladder can be evaluated in one setting.

Indications for CTU continue to evolve. Conditions commonly referred for CTU include urinary calculus disease, hematuria, flank and abdominal pain, suspected renal or urothelial neoplasm, a variety of inflammatory conditions, and congenital anomalies of the kidneys and ureters. Experience with patients who have undergone cystectomy with urinary diversion, most often for treatment of bladder carcinoma, continues to increase, and CTU is commonly used for surveillance of the urothelium in at-risk patients. Currently, CT urographic evaluation of the urinary bladder generally is not considered accurate enough to exclude small superficial urothelial tumors, and cystoscopy is indicated for complete bladder evaluation. The ability to biopsy and resect lesions are added benefits of cystoscopy.

The American Urological Association Best Practices Policy guidelines recommend IV or CT urography as the initial imaging test for patients with asymptomatic microscopic hematuria.1 Likewise, the American College of Radiology rated CTU as the most appropriate imaging procedure in the evaluation of hematuria. Furthermore, extra-urinary findings, some of them clinically important, can be found in a percentage of patients undergoing CTU. Contraindications to CTU are generally limited to those patients who cannot receive iodinated contrast because of renal insufficiency, prior severe reaction, or pregnancy.

2.1.10 Image Quality:

The image quality of conventional radiographs is expressed in terms of spatial resolution, contrast resolution, and noise. These characteristics are relatively easy to describe but somewhat difficult to measure and express quantitatively. Because CT images are composed of discrete pixel values, image quality is somewhat easier to characterize and quantitate. A number of methods are available for measuring CT image quality, and five principal characteristics are numerically assigned. These

include spatial resolution, contrast resolution, noise, linearity, and uniformity. (Richar D 2009).

2.1.10.1Spatial Resolution:

If one images a regular geometric structure that has a sharp interface, the image at the interface will be somewhat blurred. The image is somewhat blurred owing to limitations of the CT imaging system; the expected sharp edge of CT values is replaced with a smoothed range of CT values across the interface. Spatial resolution is a function of pixel size. CT imaging systems allow reconstruction of images after imaging followed by post-processing tasks; this is a powerful way to affect spatial resolution. Thinner slice thicknesses also allow better spatial resolution. Anatomy that does not lie totally within a slice thickness may not be resolved, an artifact called *partial volume.* Therefore, voxel size in CT also affects CT spatial resolution. The design of pre-patient and pre-detector collimation affects the level of scatter radiation and influences spatial resolution by affecting the contrast of the system.

The ability of the CT imaging system to reproduce with accuracy a high-contrast edge is expressed mathematically as the edge response function (ERF**)**. The measured ERF can be transformed into another mathematical expression called the modulation transfer function (MTF). The MTF and its graphic representation are most often cited to express the spatial resolution of a CT imaging system. Although the MTF is a rather complicated mathematical formulation, its meaning is not too difficult to represent. Then consider the series of bar patterns that are imaged by CT. (Richar D 2009).

The spatial frequency for CT imaging systems is expressed often as line pairs per centimeter (lp/cm) instead of line pair per millimeter (lp/mm. The loss in faithful reproduction with increasing spatial frequency occurs because of limitations of the imaging system. Characteristics of the CT imaging system that contribute to such

image degradation include collimation, detector size and concentration, mechanical and electrical gantry control, and the reconstruction algorithm. In simplistic terms, the MTF is the ratio of the image to the object as a function of spatial frequency. If the image faithfully represents the object, the MTF of the CT imaging system would have a value of 1. If the image were simply blank and contained no information whatsoever about the object, the MTF would be equal to zero.

2.1.10.2Contrast Resolution:

The ability to distinguish one soft tissue from another without regard for size or shape is called contrast resolution. This is an area in which multi-slice helical CT excels. The absorption of x-rays in tissue is characterized by the x-ray linear attenuation coefficient. This coefficient, as we have seen, is a function of x-ray energy and the atomic number of the tissue. In CT, absorption of x-rays by the patient is determined also by the mass density of the body part.

Although these differences are measurable, they are not imaged well on conventional radiography. The CT imaging system is able to amplify these differences in subject contrast so the image contrast is high. The range of CT numbers for these tissues is approximately − 100, 50, and 1000, respectively. This amplified contrast scale allows CT to better resolve adjacent structures that are similar in composition. The contrast resolution provided by CT is considerably better than that available in radiography principally because of the scatter radiation rejection of the pre-patient and pre-detector collimators. The ability to image low-contrast objects with CT is limited by the size and uniformity of the object and by the noise of the system. (Richar D 2009).

2.1.10.3 Noise:

If a homogeneous medium such as water is imaged, each pixel should have a value of zero. Of course, this never occurs because the contrast resolution of the system is not perfect; therefore, the CT numbers may average zero, but a range of

values greater than or less than zero exists. This variation in CT numbers above or below the average value is the noise of the system. If all pixel values were equal, the noise would be zero. Noise is the percentage standard deviation of a large number of pixels obtained from a water bath image. It should be clearly understood that noise depends on many factors: kVp and filtration, Pixel size, Slice thickness, Detector efficiency and Patient dose.

Ultimately, the patient radiation dose, the number of x-rays used by the detector to produce the image, controls noise. Noise appears on the image as graininess. Low-noise images appear very smooth to the eye, and high-noise images appear spotty or blotchy. Noise should be evaluated daily through imaging of a 20-cm-diameter water bath. All CT imaging systems have the ability to identify an ROI on the digital image and to compute the mean and standard deviation of the CT numbers in that ROI. When the radiologic technologist measures noise, the ROI must encompass at least 100 pixels. Such noise measurements should include five determinations—four on the periphery and one in the center. (Richar D2009).

2.1.10.4 Linearity:

Computed tomography imaging systems must be calibrated frequently so that water is consistently represented by CT number zero and other tissues by the appropriate CT numbers. A check calibration that can be made daily uses the five-pin performance test object of the American Association of Physicists in Medicine

(AAPM) .Each of the five pins is made of a different plastic material that has known physical and x-ray attenuation properties and is positioned in a water bath. After this test object is imaged, the CT number for each pin should be recorded and its mean value and standard deviation plotted. The plot of CT number versus linear attenuation coefficient should be a straight line that passes through CT number 0 for water. (Richar D 2009).

2.1.10.5Uniformity

When a uniform object such as a water bath is imaged, each pixel should have the same value because each pixel represents precisely the same object. Furthermore, if the CT imaging system is properly adjusted, that value should be zero. Because the CT imaging system is an extremely complicated electronic mechanical device, however, such precision is not consistently possible. The CT value for water may drift from day to day or even from hour to hour. At any time that a water bath is imaged, the pixel values should be constant in all regions of the reconstructed image. Such a characteristic is called spatial uniformity.

Spatial uniformity can be tested easily with an internal software package that allows the plotting of CT numbers along any axis of the image as a histogram or as a line graph. (Richar D 2009).

2.2 Previous studies:

Various studies were published in the last recent years, in 2009 Dahlman p et al, were seeking to optimize of computed tomography urography protocol, 1997 to 2008: effects on radiation dose. Since CT urography began to replace excretory urography as the primary imaging technique in uroradiology, the collective radiation dose to patients has increased. Examined the changes in the CT urography protocol for investigating suspected urinary tract malignancy between the years 1997 and 2008, and how these changes have influenced the mean effective dose. The study was based on 102 patients(mean age 66.1 +/- 14.8 years, range 31-89 years; 30female,72 male) divided into five groups (groupsA-E) corresponding to the time points at which changes were made to the CT urography protocol. the mean effective doses were estimated using the impact CT patient dosimetry calculator.(Dahlman p *et al* 2009).

The number of scan phases at CT urography was reduced from four to three in 1999, resulting in a reduction of the mean effective dose from 29.9/22.5 (femal[f]/male[m]) mSv (group a)to 26.1/18.9(f/m) mSv (group b). in 2001, mAs setting were adapted to patient size, and the mean effective dose was reduced to 16.8/12.0(f/m)mSv(group c). in 2005, scans were performed with a multidetector-raw CT equipped with automatic tube current modulation in the x-and y-axis (CAREDose), since the effective mAs was also lowered in the unenhanced and excretory phase, yet the mean effective dose increased to 18.2/13.1(f/m)mSv (group d), since the effective mAs had to be increased in the corticomedullary phase to maintain image quality . in 2008, as tube current modulation in the x-y- and z-axis was introduced (CARE Dose 4D), the mean effective dose was reduced to 11.7/8.8 (f/m) mSv (group e). this study shows that the individual mean effective dose to patient undergoing CT urography has decreased by 60%, from 29.9/22.5 (f/m)mSv in 1997 to 11.7/8.8(f/m) mSv in 2008.((Hamm M *et al* 2005).

In2007 Eikefjord EN et al , were measured and compared the effective radiation dose in patients undergoing unenhanced MDCT and excretory urography for acute flank pain , and to explore technical and practical factors affecting the effective dose .119 patients with acute flank pain were included. All patients were examined using both MDCT and excretory urography. CT involved one acquisition from the upper kidney margin to the symphsis pubis. The only protocol variation was in the tube current (mAs) , which was made according to patient body mass . the excretory urography protocol consisted of three images , with more when supplementary images were needed . effective radiation doses were computer-simulated using dosimetry programs for CT and conventional radiography ,based on nerwegian radiological protection board dose data sets . mean and SDs of measured patient dose were calculated and compared . further analyses of dose variation in

body mass categories (body mass index) were conducted, as were analyses concerning the number of images taken. The mean effective dose were 7.7 mSv with MDCT and 3.63 mSv with excretory urography .the effective dose varied both in and between techniques but could be predicated . radiation risk decreased significantly with increased patient weight . the average effective dose with MDCT was more than double that with excretory urography .however , the appropriate dose could be strongly predicated by the patient's body mass index and by procedure. An optimum low- dose protocol should be considered before initiating unenhanced MDCT for ureteral colic in order to minimize the radiation –induced cancer risk and to secure adequate image quality . (Eikefjord *et al* 2007).

In 2003 Nawfel et al , measured and compared patient radiation dose from computed tomographic urography and conventional urography and to compare these doses with dose estimates determined from phantom measurements .

Patient skin dose were determined by placing a thermoluminescent dosimeter (TLD) strip (six TLD chips) on the abdomen of eight patient s examined with CT urography and 11 patients examined conventional urography . CT urography group consisted of two women six men (mean age 55.5 years), and conventional urography group consisted of six women and five men (mean age, 58,9 years) . CT urography protocol included three volumetric acquisitions of the abdomen and pelvis . conventional urography protocol consisted of acquisition of several image involving full nephrontomography and blique projections, mean and SD of measured patient doses were compared with corresponding calculated doses and with dose measured on a Lucite pelvic- torso phantom . correlation coefficient (R2) was calculated to compare measured and calculate skin dose for conventional urography examination , and two – tailed p value significant test was use to evaluate

variation in effective dose with patient size . radiation risk was calculated from effective dose estimates . (Nawfel *et al* 2004).

Mean patient skin doss for CT urography measured with TLD strips and calculated from phantom data (CT dose index) were 56.3 mGy +/- 11.5 and 54.6 mGy +/- 4.1 , respectively . mean patient skin doses for conventional urography measured with TL Dstrips and calculated as entrance skin dose were 151 mGy +/- 90 and 145mGy +/- 76 , respectively .correlation coefficient between measured and calculated skin doses for conventional urography examinations was 0.95 . mean effective dose estimates for CT urography and conventional urography were 14.8 mSv +/- 90.0 and 9.7 mSv +/- 3.0 , respectively . mean effective doses estimated for the pelvic-torso phantom were 15.9 mSv (CT urography and 7.8 mSv(conventional urography).(Liu W *et al* 2000).

Chapter Three

Material and Methods

The data used in this study were collected from center radiology unit in military hospital Omdurman, Sudan .data of the technical parameters used in CT procedures was taken during december,2015-janeury2016.

3.1 Patients Data:

A total of 26 patients referred to alselh medical hospital in the period of the study within two months.

3.2 Material and Method for CTU:

3.2.1 Machine Used:

TOSHIBA Multi-Slice 64 Helical.

3.2.2 Patient Preparation :

The patient should be fasting at least 8 hours before the exam and should be well hydrated and full bladder just before the exam.

3.2.3 Patient positioning:

Patient lying in supine position, arms positioned comfortably above the head, lower legs supported.

3.2.4 The Protocol Used in CTU:

Initial scan without IV contrast, patient should suspended respiration in expiration during scan. Start with scout view to localize start and end position (from

diaphragmatic dome to symphysis pubic). Start scan area using slice thickness 7mm, this sequence as KUB. After that injected IV contrast 100 ml 300mgI/ml with delay time 17min then repeat the previous scan KUB this sequence is CTU.

3.3 CT Dose Measurements:

The patient dose estimate from CT examination using the monte carlo technique requires measurements of CTDI and conversion coefficient. In theory the CTDI, which is measure of the dose from single- slice irradiation, is defined as integral along a line parallel to the axis of rotation (z) of the dose profile,D(z), divided by the nominal slice thickness,(t). In this study dose quantity was obtained (CTDI,DLP) from displayed at the operator's console. CTDI and DLP do not include patient specifics such as size and organ radiosensitivity.

Chapter Four

Results

A total of 26 procedures were performed over 2 months: patient demographic data and patient radiation dose value presented in table (4.1) and table (4.2) respectively. patient demographic data in figures (4.3). figure (4:1) show relationship between(weight and effective dose),and figure (4:2) show relationship between (BMI and DLP).

Parameter used in CT machine TOSHIBA multi slice 64 helical, kvp and mAs were constant for all patients 120, 150 respectively, slice thickness 7mm, and rotation time was 0.5 sec and region of scan include abdomen and pelvic these parameter produced radiation value in table (4:2).

Table (4.1) the average value and range for patient demographic data for patients

patient	Age (year)	Height(cm)	Weight(kg)	BMI
M	53.21±21.04 (23-78)	162.4±41.59 (151-178)	63.07±17.73 (53-77)	23.79±8.41 (15.78-36.37)
F	46.18±16.3 (35-63)	160.4±32.09 (150-173)	69.91±23.09 (55-85)	29.12±12.18 (15.18-46.31)
Total	50.88±16.95 (23-78)	161.62±32.09 (150-178)	65.96±38.39 (53-85)	26.44±9.8 (15.8-46.31)

Table (4.2) patients radiation dose value

Value	Average	min	Max
DLP(mGy/cm)	3647.81±950.39	3064.2	4335.5
CTDI(mGy)	69.47±14.81	65.7	91.2
ED(mSv)	54.72±14.26	45.96	97.24

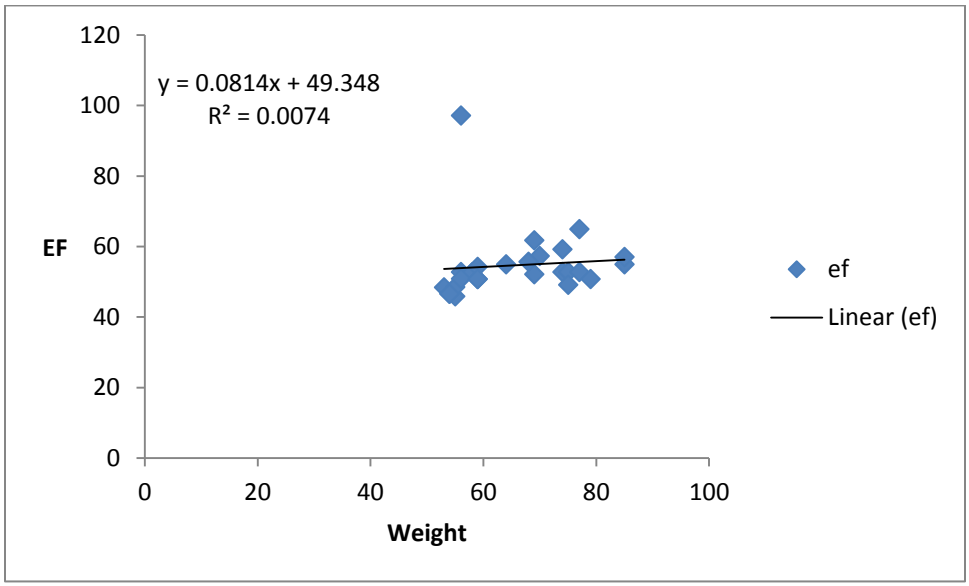

Figure (4:1) the relationship between Weight and EF dose

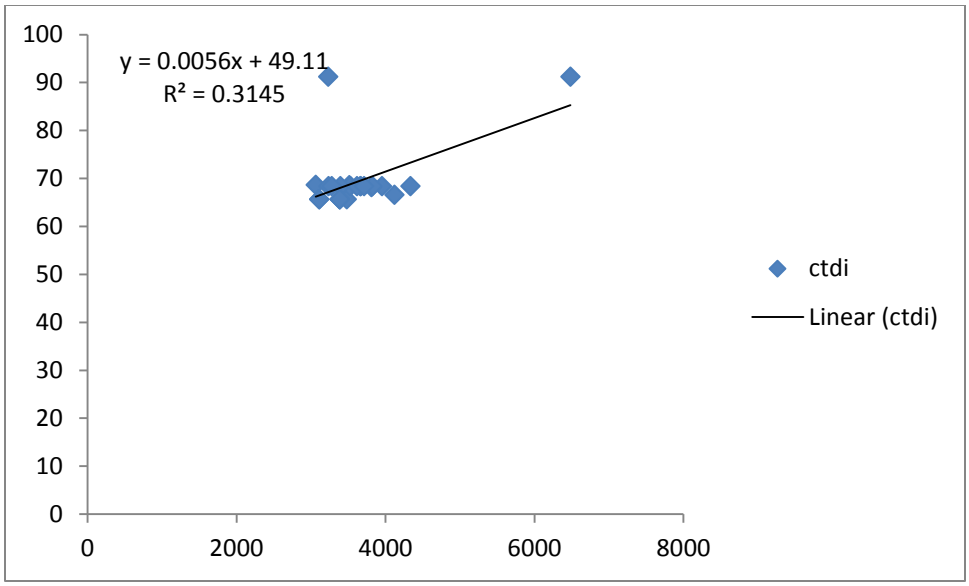

Figure:4.2 The relation between CTDI and DLP

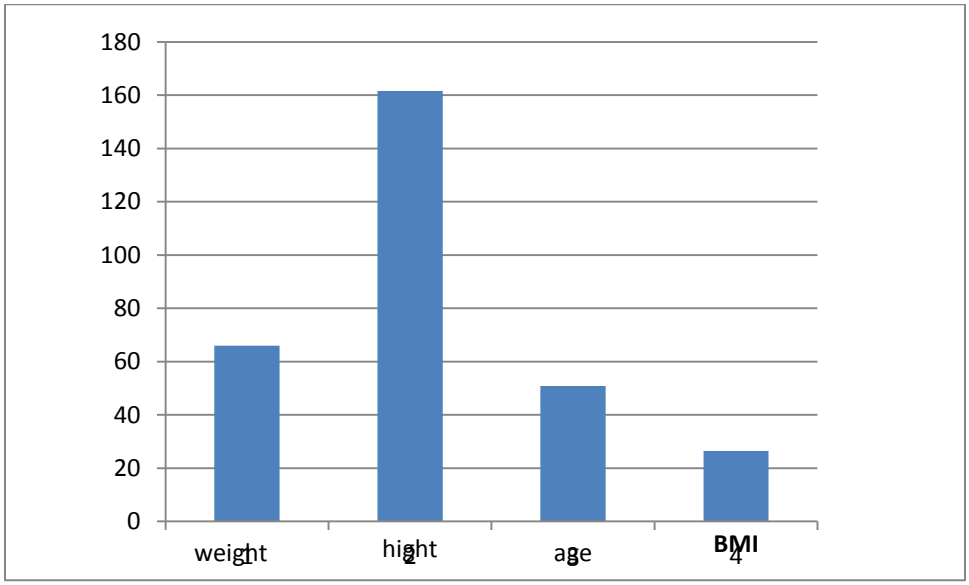

Figure 4:3 the average value for data patient.

Chapter Five

Discussion, Conclusion and Recommendation

5.1 Discussion:

CT scanning has been recognized as a high radiation dose modality, when compared to other diagnostic X-ray techniques, since its launch into clinical practice more than 30 years ago over that time, as scanner technology has developed and its use has became more widespread, concerns over patient radiation doses from CT have grown, the introduction of multi-slice scanners has focused further attention on this issue, and it is generally believed that it will lead to higher patient doses.

CTU provides the radiologist and technologist with useful detailed information. However is the responsibility of them to determine scanning technique factor to provide balance between image quality and radiation dose and share keeping patient radiation exposure at lowest as possible.

The main objectives of this study are to optimize radiation dose to patient and image quality during the CTU.

Patient's demographic data were include 26 patients with in average age(50.88±16.95) height(161.61±32.09), weight (65.96±38.39)and BMI 26.44±9.8)are presented in table (4:1).

Exposure parameter for patient used CT machine TOSIBA Multi-slice 64 helical, mAs and kVp are constant 120and150 respectively. and pitch with 1and, slice thickness was 7mm and rotation time was 0.05 sec and region of scan include the abdomen and pelvic these parameter produced these radiation value represented in table 4:2 which shows the value of DLP average(3647.81±950.34)

and range between (3064.2-4335.5),CTDI average was (69.47±14.81) and range between (65.7-91.2).these value to estimate the organ equivalent dose using software provided by national protection broad and using impact CT patient dosimetry calculator to calculate the total effective dose .the risk of cancer in a particular organ was estimate by multiplying the mean organ equivalent dose with the risk coefficients (f) obtain from (ICRP) the result were the effective dose 54.72 mSv.

Table 5:1 show the previous studies results during CTU.

Author	No of patient	effective dos(mSv)
Dahlman P,et al (2009)	102	29.9-22.5
Eikefjord EN, et al (2007)	119	7.7
Richard D,Nawfel, et al (2004)	8	14.8±9
Hamm M, et al (2002)	109	3.1-4.3
Liu W et al (2000)	60	2.8

5.2 Conclusions:

CT urography is evolving and is a promising diagnostic examination that allows comprehensive evaluation of the urinary tracts. CT urography can combine the superb diagnostic capabilities of EU and CT and is becoming the primary imaging study for evaluation of patients with hematuria and other genitourinary conditions. Important remaining issues to be addressed include cost effectiveness and patient radiation exposure. Further studies evaluating large numbers of patients with urothelial abnormalities are necessary to further evaluate the role of CT urography in clinical practice. CTU had been considered of the good image test for evaluation of urinary tract system. In this study measurement of radiation doses from computed tomography, and evaluate protocols used in CTU image procedure.

Radiation dose can vary considerably between scanner and between institutions. Clinical dose are reported as the dose to standard dosimetry phantom. However due to large variation in patient size these dose may not estimate accurately the deliver to patient during a particular exam.

In this study the radiation dose is considered high compared with previous studies.

5.3Recommendations:

- Continuous training is important for improving the techniques and protocols used in CTU.
- A written protocol for CTU procedures is required in order to get rid of inter-operator dose and technique variations.
- Maintain accurate operating conditions (balance between image quality and radiation exposure) in order to reduce dose while maintaining diagnostic image quality.
- Exposure parameters must base on patient weight and anatomic region of interest.
- It is importance to evaluate the risk of cancer.
- Avoided repetition test without clinical justification.
- Considerate ALARA principle.
- Procedure examination CTU by practitioners and justified by physician and radiologist.

5.4Suggestion for future work:

A survey is highly recommendation in order to establish a national diagnostic reference level of children for CTU.

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Appendix:

Data Sheet

weight	length	age	ctdi	dlp	bmi	ef
74	156	35	68.4	3521	35.1	52.815
85	156	49	68.4	3665.4	46.31	54.981
75	170	33	68.4	3521.7	33.09	52.8255
56	172	65	68.4	3521	20.5	52.815
64	153	52	68.4	3665.4	26.8	54.981
55	162	38	68.7	3064.2	18.7	45.963
74	169	49	68.4	3952.5	31.7	59.2875
77	165	40	68.4	4335.5	32.4	65.0325
53	178	23	91.2	3232	35.93	48.48
85	150	37	68.2	3809.1	15.78	57.1365
58	156	37	68.6	3512.8	21.56	52.692
54	151	65	65.7	3110.1	6.86	46.6515
59	158	78	65.7	3386.4	19.31	50.796
69	160	43	65.7	3478.5	29.75	52.1775
70	170	60	68.4	3825	28.82	57.375
55	160	50	68.4	3234.6	18.91	48.519
77	163	42	68.4	3521	36.37	52.815
56	159	40	68.4	3394.1	19.72	50.9115
59	156	63	68.4	3617.4	22.31	54.261
75	169	53	68.4	3277.7	33.28	49.1655
79	173	60	65.7	3386	36.07	50.79
69	159	42	66.6	4119.3	29.94	61.7895

68	175	56	68.4	3713.1	26.42	55.6965
54	151	65	65.7	3110.1	19.31	46.6515
59	158	78	65.7	3386.4	22.03	50.796
56	153	70	91.2	6482.8	20.49	97.242
65.96154	161.6154	50.88462	69.47308	3647.812	26.44077	54.71717
53				3064.2	6.86	45.963
85				4335.5	46.31	97.242
38.39						1422.647