الآيــة

قال تعالى :

(وفى انفسكم افلا تبصرون)

سورة الذاريات : الأية 21

Dedication

To my parent

To my brothers

To my teachers

To my friends

Acknowledgement

My full thanks to GOD in every thing. My great and deep gratitude to my supervisor Dr .HUSSAIN AHMED HASSAN

I offer my regards and blessings to all of those in modern medical center radiologists , technologist, and staff .

To whom helped me I gave them my great thanks **.**

Abbreviations

CT	Computed tomography
Kerma	kinetic energy relies by unit mass
P	The pitch
L	The scan length
1	The table increment
Kv	Kilovoltage
mAs	milli ampere second
mGy	mili Gray
C TDI	Computed tomography dose index
C TDIw	weighted Computed tomography dose index
CTDlv	volume Computed tomography dose index
nCTDlw	normalized weighted computed tomography dose
DLP	Dose Length Product
E	The effective dose
RP	Radiation Protection
SI	Standard International
NRPB	National Radiation Protection Board
IEC	International elcctrotechniqual commission
UK.	United Kingdom
DRL	Diagnostic Reference Level
ACR	American College of Radiologists
EC	European Commission
ICRP	International CommissioRadiologicalProtection
R	Roentgen
X	Exposure
S I P	standard temperature and pressure

Abstract

The use of CT in medical diagnosis delivers radiation dose to patients that are higher than those from other radiological procedures . Lake of optimized protocols could be an additional Source of Increased dose .

The goal of study was estimating radiation dose from CT Scan for the patients at modern medical center in Khartoum .

Details of this study have been taken from 35tests of patients by CT scan .and radiation dose have been calculated from volume CT dose index , Dose length product .

The most important results that the Dlp average is 311.7 ±306.34 mGy cm ,CTDI is 7.60±2.54 mGy. And E=37.41

الملخص

استعمال الاشعه المقطعيه فى مجال التشخيص الطبى يعطى جرعه اشعاعيه عاليه للمرضى اكثر من اى جهاز تشخيص طبى اخر .قله استعمال برتوكولات مثاليه مناسيه ادت الى زياده الجرعه .

هدف الدراسه هو تقدير الجرعه الاشعاعيه من اختبارات التصوير بالاشعه المقطعيه للمرضى فى المركز الطبى الحديث بالخرطوم .

تفاصيل هذه الدراسه اخذت من 35 اختبار للمرضى بواسطه الاشعه المقطعيه .تم حساب الجرعات الاشعاعيه للمرضى من موشر جرعه الاشعه المقطعيه ,volume CT dose Index والجرعه الاشعاعيه الكامله , Dose length product والجرعه الفعاله .Effective dose وحسبت لكل اختبار استعمال عوامل متعلقه بالتعرض الاشعاعى .

ومن اهم النتائج ان متوسط CTDI=7.6048 mGy ، DLP = mGy-cm 311.77 sv E=37.41,

CONTENTS

List of table

list of figure

Chapter One

1.1 Introduction

Computed tomography (CT) is an imaging technique which produces a digital tomographic image from diagnostic X-ray. The basic principle of CT involve digitizing an image received from a slit scan projection of the patient's body and then back-projecting the image through mathematical algorithms. The invention of CT has been credited to Godfrey N. Hounsfield for his work in 1970-71, although preliminary work was done by Oldendrof in 1961 and Alan Cormack in 1963, and all three based their work on the investigations of the Austrian mathematician J.Radon, who proved in 1917 that an image of a three dimensional object could be produced from its mathematical projections. Hounsfield was a research engineer with electro- musical instruments LTD (EMI). The English firm that gain international renown as the Beaties' music publishers during the 1960s. His original scanner was built on a lathe bed and required 9 days to produce a single section image. Hounsfield and Cormack were awarded the 1979 Noble prize in Medicine. Hounsfield's invention parallels Rontgen's discovery of x-ray as the most outstanding contributions to medical imaging. Early CT units were capable of only axial tomography, and the term computerized axial tomography (CAT) scanning become popular acronym. Modem CT units are capable of diverse modes much more complex than simple axial scanning. Since the introduction of helical CT in the early 1990s, the technology and capabilities of CT scanners have changed tremendously helical and spiral CT are equivalent technologies. The introduction of dual-slice systems in 1994 and multislice systems in 1998 (four detector arrays along the z-axis) has further accelerated the implementation of

many new clinical applications. The number of slices, or data channels, acquired per axial rotation has increased, with 16- and 64-slice systems now available (as well as models having 2, 6, 8, 10, 32, and 40 slices). Soon even larger detector arrays and axial coverage per rotation (>4 cm) will be commercially available, with results from a 256-slice scanner having already been published. These tremendous strides in technology have resulted in many changes in the clinical use of CT. these include, but are not limited to, increased use of multiphase exams, vascular and cardiac exams, perfusion imaging, and screening exams (primarily the heart, chest, and colon, but also self-referred "whole body" screening exams). A modem CT includes a gantry, table , x-ray tube, detectors, computer .display console and image storage units [Simpson1991]

1.2 CT Radiation dose and related risks.

As per the United Nations scientific committee UNSCEAR 2000 [Keith J. Marilyn J.2010], CT contributes over 34% of collective dose from diagnostic X ray examinations in the world. This figure is much larger for developed countries, approaching as much as 50% to 70% even though the frequency of CT examinations in these countries is of the order of 5 to 12%. As compared with the previous

UNSCEAR report six years earlier, the collective dose has grown to a factor of about 2.5 [Keith J. Marilyn J.2010]. There is no doubt that many patients have benefited from the rapid diagnoses made possible by CT and from its value for monitoring chronic disease. However, there is increasing concern regarding the risks of this exposure to radiation. It is well established that radiation can be harmful and has both deterministic and stochastic effects. Deterministic effects, 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. [Newsletter for Referring Physicians] 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. [Keith J. Marilyn J.2010]. CT technology in its various guises utilizes many photons in what amounts to a series of exposures rather than the single exposure of conventional projection radiography. The price paid by the patient is a greater radiation dose. There is another important factor at work, namely the loss of self-regulation inherent in conventional radiography afforded by the use of film. Thus, if too high a dose is used to obtain a plain film, the loss of quality (e.g. a black film) will be noticeable; if too high/unnecessarily high, dose is used in CT the result will be admirable, high quality images [P Dawson2004]. All spiral/helical systems are associated with a somewhat higher dose burden than incremental scanners because a larger volume of body is in the event scanned than that which is selected. The reason for this lies in the interpolation technique used to reconstruct the image. To reconstruct the highest slice, a data set of a higher slice yet must be available; similarly, to obtain the lowest slice, a lower slice data set must be available. This factor may contribute up to 10% increase in the effective dose. Radiation dose in CT is of particular importance for children. It is very well known that children are more sensitive and likely to get radiation induced cancer than adults, a study in 2001 in US indicated that exposure factors in CT examination used for children are similar to adults.[chappiel] Basically, risk is determined using either direct measures of dose, such as organ dose, or a weighted

measure of radiation dose taking into account various organ doses and sensitivities (effective dose).

The risk of radiation-induced malignancies from a single CT exposure is difficult to assess. There have been no published prospective studies measuring incidence of cancer among CT exposed people; however, hypothetical cancer, induction rates have been calculated from the long-term follow up of populations exposed to large doses of radiation.

The international commission on radiological protection (ICRP) reports a nominal probability coefficient of 5% per Sv effective dose for the lifetime risk of fatal cancer in a population of all ages and both sexes exposed to radiation at the relatively low doses used in CT examinations. Diederich et al [6]used the same risk coefficient (suggesting that there will be 5 fatal cancers in every 100,000 individuals exposed to 1 mSv) and reports that risks for infants and children would be higher than the risk for adults [ACC2006]. Optimization of CT procedures is also important to secure a dose as low as reasonable achievable for obtaining an adequate image quality. Optimization of CT protocols and the management of scanned patients remains a work in progress. Advances in technology continually change the design and capabilities of CT scanners, even from the same manufacturer and certainly from different manufacturers. Each CT scanner requires unique protocol development to optimize dose savings. That is why working with the medical physicist at any institution is important The international commission on radiation units and measurements (ICRU) states that “to assess the risk from stochastic and deterministic effects from medical x-ray imaging, it is necessary to know the organ or tissue doses, the dose distribution and the age and gender of the patients”. The quantities and units to be used in medical x-ray imaging as well as methods for patient dose calculation and measurements are given in ICRU report 74 The unit is the gray (Gy). CT doses may be calculated from “computed

tomography air-kerma (dose) index" (CTDI) and "air-kerma (dose) - length product" (DLP) measurements . Doses can also be measured directly by placing thermoluminiscent dosimeters (TLDs) or diodes on the patients during the procedures. The dosimetric role of the medical physicist is crucial. Availability of radiation doses to patients during CT permits a direct comparison to be made of the radiation hazards of CT scanning with alternative diagnostic procedures that also use ionizing radiation [Ware, Huda, 1999;]. Since reference dose values are intended to help identify potentially poor dosimetric performance, they may in practice be based, for example, on the results of large-scale surveys which take into account the variations in dose between centres. Such reference dose values should not be applied locally or on an individual patient basis, but rather to the mean doses observed for representative groups of patients. The values are intended to act as thresholds to trigger internal investigations by departments where typical practice is likely to be well away from the optimum and where improvements in dose-reduction are probably most urgently required. Typical levels of dose in excess of a reference dose value should either be thoroughly justified or reduced [A. Jessen,].

There is growing evidence that comparison of dose values with diagnostic reference levels has led to a decrease in patient doses, and therefore the use of this optimization tool should be widely promoted (AJR:190, June 2008]. Although the individual risk estimates are small, the concern about the risks from CT is related to the rapid increase in its use- small individual risks applied to an increasingly large population may create a public health issue some years in the future, on the basis of such risk estimates and data on CT use from 1991 through 1996, it has been estimated that aboutO.4% of all cancers in the United States may be attributable to the radiation by adjusting this estimate for current CT, this estimate might now be in the range of 1.5 to 2.0%.

To our knowledge, few studies have been performed that has investigated effective doses to individual patients by taking into account the individual technique factors, as well as the physical size of the patients undergoing these CT procedures. A comparison of the selected technique factors and the corresponding patient doses will help to determine whether these CT radiation doses to patients are as low as reasonably achievable, as required by the international commission on radiological protection [Ware, Huda, 1999;].

CT in Sudan the number of CT scans are increasing rapidly. The first ever CT scan was installed in The Modem Medical Center, Khartoum in 1984. Since that date, the number of CT scan are more than 100 scanners, two of them are 64 slice and all the rest are spiral CT. (Ware, Huda, 1999;)

1.4 Statement of the problem

The radiation dose must be known to estimate the patient's potential risk from radiation and to weight the risk against the benefits of scanning. In addition, most radiation regulatory agencies require the measurement or estimation of radiation dose to the patient from medical x-ray units. The ICRP has thus warned against CT in report 87 stating that: "the absorbed dose to tissue from CT (10 to 100 mGy) can often approach or exceed the levels known to increase the probability of cancer". As a medical physicist we should focus on the evaluation of CT protocols and assessment of patient dose by radiation dosimerty to help identify potentially poor dosimetric procedure to improve CT to individuals patients and to secure a dose as low as reasonable achievable for obtaining an adequate image quality.

1.3 Objectives

The objectives of the study were to:

- Estimation doses from CT examinations of adult patients and to compare the doses with international standards as provided in DRLs.
- Measure and evaluate the organ and effective dose in adult and pediatric patients during Abdomen CT scanning by CT software.
- Improve the radiological techniques to assure that radiation dose to patient comply to the ALARA principles.
- Provide a protocol for the optimal exposure factors that can give dose without exceeding DRLs with high image quality during the Abdominal scanning during CT examinations.

1.5 thesis outlines

This thesis is concerned with the assessment of doses from CT examinations of adult and paediatric patients and to compare the doses with international standards as provided in DRLs. It is divided into the following chapters: Chapter One: Introduction, Chapter Two: Theoretical background , Chapter Three: Materials and Methods

Chapter Four :results and Chapter five: discussions Conclusion and Recommendation.

Chapter Two

Theoretical Background

2.1 Principles of Computed Tomography

The CT scanner is a device using an X-ray source which can be used to give precise information on the attenuation properties of a thin sectional volume of the body. The basic elements of the CT scanner include the X-ray tube and the detector or detector array located in the gantry and known as the data acquisition system, the image processing system, and the image display system. The X-ray tube rotates around the patient producing a tightly collimated X-ray radiation photon beam. Once attenuated by the patient the attenuated beam strikes the detectors which convert the photon intensity to a digital signal. Multiple profiles of patient attenuation are collected (Brenner,2007).

2.2 Historical Development of CT System

Each change in the fundamental CT tube-detector structure is known as a CT generation. The CT generation has changed from the first introduced in 1972 up to the fifth in more recent years. The generation development has improved acquisition time and image quality (Website: http://www.state.nj.us/dep/rpp).

2.3 .1the first and second generation scanners

The first generation scanner was based on parallel beam geometry and the translate-rotate scanning motion. It used a single highly collimated X-ray beam (pencil beam) and one or two detectors which first translated across the patient collecting transmission readings This was done for 180° around the patient. This first generation scanner took 4.5 to 5.5 minutes to produce a complete scan . The major problem for this generation is patient throughput, motion artifacts caused by the patient, and poor image quality. Second-generation scanners are based, on a small fan beam geometry and translate- rotate motion This method is referred to as rectilinear multiple pencil beam scanning and the path traced by the X-ray tube during the scanning is same as first generation that is 180°. The scan motion and increasing detector numbers reduced the scanning time to 20 to 60 seconds. The first and second generation scanners are no longer in use. They have been replaced by third and fourth generation scanners (Fowlkes 1995).

2.3.2 Third and fourth generation scanners

Third generation scanners employ rotate-rotate configuration based on a fan beam geometry with no translation and complete rotation of the X-ray tube and detectors. The X-ray tube is coupled to a curved detector array that subtends an arc of about 30° to 40° from the apex of the fan located at the X-ray tube. The fan beam geometry rotates continuously around the patient for 360°. The minimum scan time is 1 to 4 seconds. The fourth generation scanners are based on fan beam geometry and complete rotation of the X-ray tube around a stationary 360° ring of detectors . The number of detectors in such a scanner varies from about 300 to 4000. The scan time ranges from 2 to 8 seconds

2.3 .3 Spiral CT scanners

CT scanner discussed above are known as “conventional” scanners. These scanners acquire successive slices and are characterized by the table being stationary during X-ray acquisition. In spiral CT however, the scanning mode has the X-ray tube continuously rotating together with a continuous linear translation of the patient through the gantry aperture in order to achieve volumetric data acquisition. The introduction of slip ring technology makes the spiral CT possible . The slip ring technology consists of multiple sets of parallel rings and electrical components that rotate without constraint by cables. Data are assimilated by fixed brushes contacting multiple parallel rings. Voltage can also be supplied to the X-ray tube. Slip ring technology is used in both the recent third and fourth generation scanners to allow the continuous rotation of the X-ray tube through 360° around the patient and data collection continuously forming a spiral pattern. As the X-ray tube rotates the detector array records views taken by the X-ray tube at different angles. These views are interpolated before being reconstructed using filtered backprojection .For spiral CT, the pitch factor was introduced, which is defined as table feed per 360° rotation divided by the slice thickness (Radiol 2001).

2.4 X-ray And Matters

To better appreciate patient dose it is useful to review the process of the X-ray from production to patient interaction. Thus, this section will review the X-ray characteristics, its spectrum after filtration and the X-ray attenuation concept. (Williams and wilkins 2012).

2.4.1 X-Ray Characteristics And Spectrum

X-rays are produced through two processes namely characteristic K-radiation emission and, bremsstrahlung production. The latter process produces more

radiation than the former. The intensity of radiation is proportional to the product of the number of photons and their energy. The total spectrum can be simply represented by the equation below:

1(E) = CZ(Emax-E) (2.1)

Where

I (E) is the intensity at energy

E C is constant

Z is the atomic number of the target

Thus, the resultant X-ray spectrum depends on the target material, energy applied and filtration. The effect of filtration will be discussed below (Hart,Wall2004).

2.4.2 Filtration X-ray beam

filtration in CT may be provided by typical additional filters such as 4 to 6 mm Aluminium, 0.5 mm Copper and a beam shaper. The filtration in CT scanners may serve the two main purposes; of changing X-ray beam energy and changing intensity over the X-ray fan beam.

Filtration thus hardens the beam by absorbing the “soft” (low energy) radiation so that a more homogenous beam with better penetration is utilised. A specially designed filter for CT known as bow tie or beam shapers used before the patient to compensate the attenuation across the patient thus provides uniform x-ray beam to reach the detector. These may reduce the intensity of the beam away from the centre of the scan field. Both filtration techniques serve to also reduce patient exposure (Matthews,Brennan 2009).

2.4.3 Attenuation and Half Value Layer

The attenuation of X-ray photons occurs when absorbed or scattered with in a medium. Linear attenuation is defined by the Lambert-Beer Law.

$I=i_0 e^{-\mu x}$ *(2.2)*

Where

i= the transmitted photon intensity

1_0 = incident photon intensity

x = the thickness of the material (cm)

H=the linear attenuation coefficient (cm^{-1})

The factors that affect the linear attenuation coefficient are the energy of the beam, material characteristics including atomic number, density and the number of electrons per gram of the material. Increasing the energy will increase the number of transmitted photons and decrease the attenuation, while, increasing the material density, atomic number and electrons per gram will increase the attenuation. Another fundamental attenuation coefficient is the mass coefficient, which is obtained from the linear coefficient by dividing by the density p. The mass attenuation coefficient (µm) or (µ/p) is independent of the density.

The attenuation coefficient is defined as the linear attenuation coefficient, µ and the dimensions of per cm. If the thickness is in g/cm^2 then the absorption coefficient is known as the mass attenuation coefficient, µ m with its dimension cm^2/g. The relationship between µm and µ. can be seen in the equation below

$\mu(cm^{-1})=\mu m(cmg^{-1})*p(g.cm^{-3})$ (2.3)

where p is the density of the absorber.

Half value layer is the thickness of absorber required to reduce the intensity of a radiation beam by a factor of two **Thus,**

HVL =ln2/µ **(2.4)** **(Aroua,Besancon 2004).**

2.4.4 Contrast media in CT examination

The contrast media used in a CT examination can be either positive or negative (Baert & Sartor 2001). Positive contrast media can be iodinated or a diluted barium suspension, and may be administered orally, rectally or intravenously. Negative contrast media, such as gas or water, may also be used in the examination. Both kinds of contrast

may be used together to optimize the detection of abnormalities. CT examinations can be carried out with or without contrast enhancement based on the clinical situation (Slone 2000).

Pre-contrast series (without contrast enhancement) are carried out to demonstrate the presence of stone, or calcification, or as a standard baseline before contrast enhancement. Contrast series are preferred in most cases to demonstrate different phases of blood intake in any organ or abnormal structures. Administration of contrast media is contra indicated however, for patient allergies to contrast media such as iodine. The timing of image acquisition after contrast administration is crucial to achieve the examination objectives. If arterial and venous phases are carried out, it is known as a biphasic series. Further, if delay

series are taken after those two series, it is known as a triphasic series. With current CT technology, scanning after contrast media injection can be initiated automatically once the appropriate density, for example around 50 Hounsfield Unit (HU) (Siemens 2000). Is selected. A Hounsfield unit is a quantitative scale for describing radiopacity of image area or image density. For example, liver enhances about 40 HU within 30 seconds with 2.5 ml/seconds rate and volume 125 ml bolus. Although, the use of contrast media has many advantages in demonstrating abnormalities, it also increases the number of scan series, thus increasing scan volume, which directly increases the patient radiation dose (Seeram 2000)

2.5 CT examination protocols

Current CT scanners have preset protocols installed in their system, These protocols allow the user to automatically select appropriate parameters to be used for CT examinations. Scan parameters include kV, mAs, slice thickness and table speed, and can also be manipulated manually after consideration of patient size, the organ involved, and patient condition. Occasionally the user may choose specific organs to be examined which require different parameters to the original preset parameters, eg: routine abdominal CT examinations include the upper abdomen and pelvis and is normally scanned with the patient requested to suspend end-expiration to reduce internal pressure and motion. X-ray tube potentials normally range from 120 to 140 kV while, mAs can typically vary from 210 to 330. These two scan parameters are vital for dose optimization with patient size. The selection of slice collimation may be from 5 to 8 mm with a pitch of 1 to 1.6. For individual organs such as the pancreas or kidneys, slice thickness may range from 2 to 5 mm to allow detection of small lesions (Seeram 2000).

The applied protocols from different centres affect the dose measured amongst scanners. To provide best practice the American College of Radiologists and the European Commission (European Commission 1998) have provided guidelines on CT protocols .The information includes appropriate protocols for parameters, image interpretation and quality assurances ((ACR 2001).

2.6 General definitions of exposure and dose

Radiation dose to a patient is better appreciated by an understanding of the relationships between radiation exposure and dose. Many dosimetry definitions had been described in international publications by the International Commission on Radiological Protection (ICRP 1991). Thus in this section a review of the terms, radiation exposure.

absorbed dose and effective dose and their relationships is discussed.

Dosimetry is the process of determining radiation dose. It includes the description of the type of radiation and the amount of energy it may deposit in some medium. Radiation from an X-ray generator consists of a beam of photons with a variety of energies; however, it can be manipulated to produce a more monoenergetic beam by filtration. The two distinct terms generally used in dosimetry are exposure and absorbed dose.

2.6.1 Kerma

Is the quotient of dE_{tr} by dm, where dE_{tr} is the sum of the initial kinetic energies of all the charged particles liberated by uncharged particles in a mass dm of material, thus

$K = dE_{tr} \setminus dm$ (2-5)

Where. dE_{tr} is tiie sum of the initial kinetic energies of all charged ionizing particles liberated by uncharged ionizing particles in a material of mass dM. The SI unit of kerma is the joule per kilogram ($J.kg^{-1}$), and special unit is gray (Gy) ((ICRP 1991)..

2.6.2 Exposure

Refers to a measure of the radiation needed to ionize air as a medium. It is measured in Roentgens (R) and defined as the quantity of X-rays that produces 2.580 x $I0^{-4}$c of charge collected per unit mass (kilograms) of air at standard temperature and pressure (STP). Equation for exposure (X) is,

$X = dQ \setminus dm$ (2-6)

Where dQ is the absolute value of the total charge of the ions of the one sign produced in air

when all of the electrons liberated by photons in a volume element of air having a mass dm are completely stopped in air ((ICRP 1991).

2.6.3 Absorbed Dose

Absorbed dose is also referred to as absorbed radiation dose or radiation dose. It describes the amount of energy absorbed per unit mass (Johns & Cunningham 1983).The absorbed dose in an organ or tissue. Dy. is the absorbed dose averaged over the volume of a

tissue or organ T rather than at a point. It is measured in grays (Gy) where 1 Gy = I J.kg"'.

The equation for absorbed dose. D, is

D = d E \ dm (2-7)

Where dE is the mean energy imparted by ionizing radiation to matter in a volume element and dM is the mass of matter in the volume element. The energy can be averaged over any defined volume, the average dose being equal to the total energy imparted in the volume

divided by the mass in the volume. The absorbed dose also does not reflect the relative radiosensitivity or risk of detriment to those tissues being irradiated (Johns & Cunningham 1983).

2.6.4 Equivalent Dose

The effect of different types of radiation on tissue is given by the radiation weighting factor (W_R). and for X- ray photon is equivalent to 1 (ICRP 1991). If the absorbed dose (D_T) is multiplied by radiation weighting factor, it is known as the equivalent dose (H_T), which is expressed in Siverts (Sv).

$Hi = W_R.D_T$ (2-8)

The probability of the occurrence of radiation induced stochastic effects is dependent on absorbed dose (ICRP 1977; 1991). Stochastic effects are related to

the ability of ionizing radiation to damage Deoxyribonucleic Acid (DNA) in human cells that can lead to

detrimental effects such as a malignant tumor or the possibility of heredity defects.

2.6.5 Effective Dose

The relationship between the probability of stochastic effects and equivalent dose depends on the particular organ and tissue irradiated (ICRP 1991). Thus, to indicate the combination of the different doses to several different tissues in a way which is likely to correlate well with the total of the stochastic effects, the tissue-weighted factor, T_w is

introduced in (Table 2-1). This factor represents the relative contribution of organ or tissue to the total detriment due to effects resulting from uiform irradiation to the whole body (ICRP 1991)

Table 2.1: Tissue weighting factor (ICRP 1991)

Tissue or organ	Tissue weighting factor W_r
Gonads	0.2
Bone marrow(red)	0.12
Colon	0.12
Lung	0.12
Stomach	0.12
Bladder	0.05
Liver	0.05
Breast	0.05

The effective dose is the sum of the weighted equivalent doses in all the tissues and organs of the body. It is given by the expression,

$$E_r = \sum_T Wr.Hr \qquad (2\text{-}9)$$

where H_T is the equivalent dose in tissue or organ T and W_r is the weighting factor for the tissue T. The unit of effective dose is the sievert (Sv) (ICRP1991).

2.6.6 Specific Dose unit for CT

The currently accepted appropriate descriptor for CT dose is the Computed Tomography Dose Index (CTDI). This is a local dose descriptor of dose output for scanners measured in air at the centre of rotation The CTDI is defined as the radiation dose, normalized to beam width, measured from 100 mm length of a pencil ionisation chamber.

$CTDI_{100}=1/NT\sum_{+5cm}^{-5cm} D(Z)dz$ mGY (2.10)

where D(z) absorbed dose relative to location along the z axis; N is the number of acquired sections per scan (or the number of data channels used during acquisition) and T is the nominal width of each acquired section (product of NT is also known as beam collimation).

Specific imaging protocols also include the pitch as a factor, thus in consideration of that factor, another descriptor has been created. The unit is $CTDI_{vol}$ or $CTDI_w$,. It is defined as

$$CTDI_{vol} = CTDI_w . NT/I \qquad \text{mGY} \qquad (2.11)$$

where N and T are defined in equation 2.10 and represent the total collimated width of the X-ray beam and I is the table increment per rotation for helical scan or spacing between acquisition for axial scans. Pitch is one of the parameters in spiral

CT. Pitch is defined as table distance travelled in one 360° rotation over total collimated width of the X-ray beam, while in conventional CT it is defined as table increment over slice thickness. Pitch can be calculated by the equation 2.12.

Pitch = I/NT (2.12)

In describing the exposure distribution along the z axis another descriptor known as *Dose Length Product (DLP)* (European Commission 1998) is used as an integral dose quantity. This DLP is created as estimated effective dose value without taking account of tissue weighting factor. The DLP is expressed in units of Gy.cm and given in the equation below,

DLP = CTDI $_{vol}$***. Scan lenght mGy. cm (2.13)*** ***(***European Commission 1998)

2.6.7 Diagnostic Reference Levels

In optimization of radiation in medical exposure, the ICRP publication 60 recommended as noted:-"Consideration should be given to the use of dose constraints, or investigation levels, selected by appropriate professional or regulatory agency, for application in some common diagnostic procedures. They should be applied with flexibility to allow higher doses where indicated by sound clinical judgement" (ICRP 1991). Then, the ICRP publication 73 introduced the first "diagnostic reference level" (DRL). It explained the concept of reference levels and expanded in more detail the concept of DRL (ICRP 1996). Thus in the context of optimisation of radiation protection of patient in diagnostic radiology including CT, the introduction of DRL is appropriate. The DRL is defined as a dose level set for standard procedures and for groups of standard-size patients or a standard phantom for broadly defined types of equipment (CEU 1997; European Commission 1999; ICRP 1996). The selection of DRL is decided by professional medical bodies, using third quartile dose values on the observed distribution for patients, and specifically for a country or region (ICRP 2002).The purpose of DRL

is to advise local authorities as a quality assurance tool in identifying individual centers that are consistently unusually high in their dose values levels against clinical doses that need to be reviewed (IAEA 2002; ICRP 2002).

The DRL quantities should be easily measured on a simple standard phantom or representative patient for diagnostic radiology. The dose quantities suggested for CT examinations are: (1) CTDIW per slice (mGy) and (2) DLP per exam (mGy.cm) (European Commission 1998). These two quantities provide a useful indication of the relative scanner X-ray output in CT reflecting both the technique factors selected for each examination and overall scope of an examination for a given type of procedure and patient group (Alice B. Smith, William P. Dillon,)

2.6.8 Determination of DRL

The diagnostic reference levels (DRLs) can be derived from the observed distributions of patient doses in a certain area, conducted over a certain period of time. DRLs then are not an individual measure but derived from a representative sample of dose indicators associated with a standard patient size. The DRL must be set at approximately the level of the third quartile in the dose distribution. The third quartile value is chosen as an appropriate investigation level on the grounds that if 75% of X-ray departments can operate satisfactorily below this dose level, then the remaining 25% should be made aware of their considerably less than optimal performance and hence should be encouraged to alter their radiographic equipment or techniques to bring their doses in line with the majority. At the same time adherence to the Diagnostic Requirements for each scan series will ensure that diagnostic effectiveness does not suffer from any dose reduction (European Commission 1998). The DRL in general is the rounded third quartile distribution of mean values of CTDIW and DLP measured for a particular examination on a patient group (library.usyd.edu.au/bitst).

The ideal standard patients' size as recorded by EC (European Commission 1996) is 20 cm AP trunk thickness and 70 kg weight, with an average weight, that is 70 ± 3 kg. Similarly, the UK has adopted the criteria that the mean weight of the sample should lie in the range 65 to 75 kg for the mean dose to be indicative of the typical dose to an average (70 kg) patient (Hart et al. 2002a). However, if there is no average patient available, the measurement period of all patients and the average of the dose result can be calculated as the outcome for a standards patient (Hart et al. 2002a).

Chapter Three

Materials And Methods

3.1 Material

Design of the study is analytical case control study

Area and duration of the study:

This study has been conducted in radiology department of Modern Medical Center in July 2004

CT scanner:

GE Dual slice , Installation date 2003 ,with slice thicknes range 2,3,5,7,and 10 mm,KV up to 120-140 and MAS 20MA,40MA ,50MA to 400 MA

Population Of study :

A total of 36 patients (male and female). Underwent CT study for Abdomen ,for different clinical problems was selected and recorded their Anthropometrical data (Age ,sex,..) with CT protocol technical data (kv, mAs ,DlP ,CTDI) and slice thickness.

CT protocol

All patients underwent CT study for abdomen for different clinical problems with contrast and without contrast CT examinations include the upper abdomen and pelvis , slice thickness (5-10mm) , kv 120 ,, mAs 200-250, scan time 15-25 sec and pitch 1 to 1.5 scanned with the patient suspend breath at end-expiration to reduce internal pressure and motion.

3.2 method

Dose calculation

A conversion factor (E_{DLP}) method has been used by European Commission (European Commission 1998) for a general anatomic region. This value is the normalized effective dose per dose length product over various body regions such as head, neck, chest, abdomen and pelvis for different sizes of patient.

$CTDI_{vol}$ and distance are used to estimate DLP and then multiplied with the conversion factor value.

to estimate effective dose (E) using equation below.

$$\mathbf{E=E_{DLP}.DLP}$$

Where E_{DLP} is the tissue weight factor

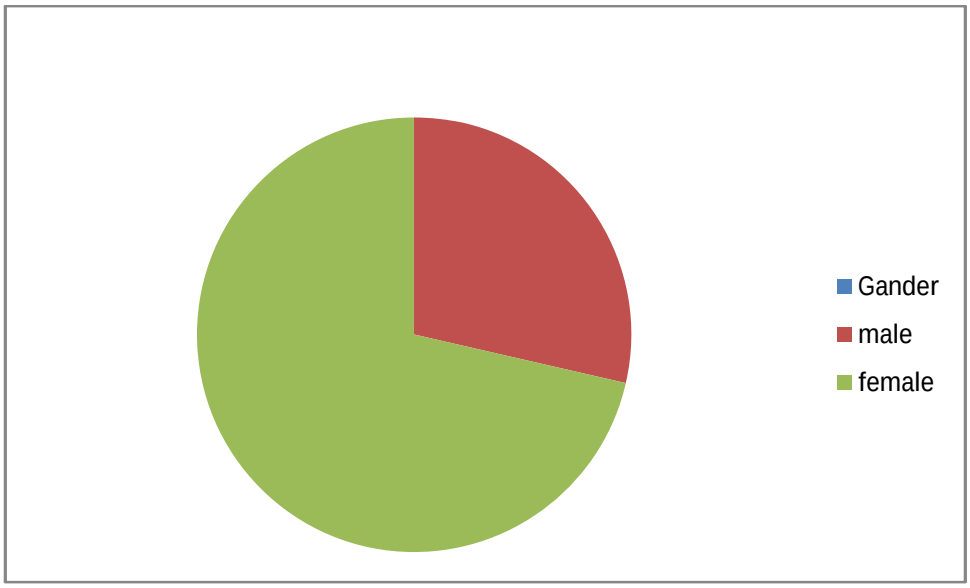
Chapter four

Result

The following tables and figure presented data:

Table4.1 gender distribution:

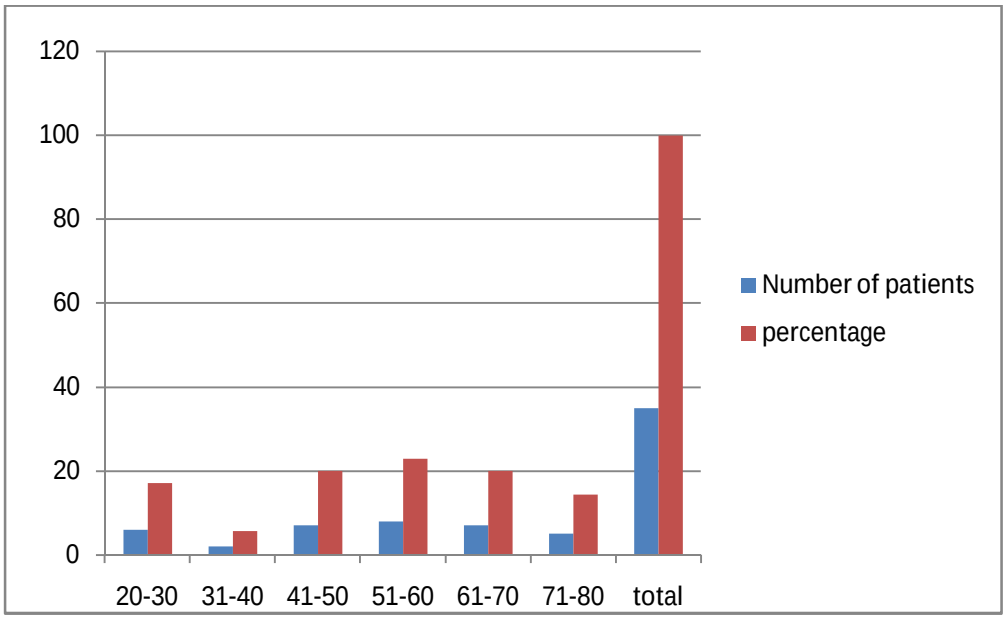
Gender	Number	percentage
Male	**10**	**28.58%**
Female	**25**	**71.42%**
Total	**35**	**100%**



Figures 4.1: gender distribution

Table 4.2 age distribution

Age	20-30	31-40	41-50	51-60	61-70	71-80	Total
Number	**6**	**2**	**7**	**8**	**7**	**5**	**35**
percentage	**17.14%**	**5.7%**	**20%**	**22.8%**	**20%**	**14.2%**	**100%**



Figures 4.2 the distribution of different age groups in the study population

Table 4.3 the CTDI and DLP value and Slice thickness

CTDI	PLP	Slice thickness
7.6048±2.549mGy	**311.77±306.34 m Gy cm**	**7.88 mm**

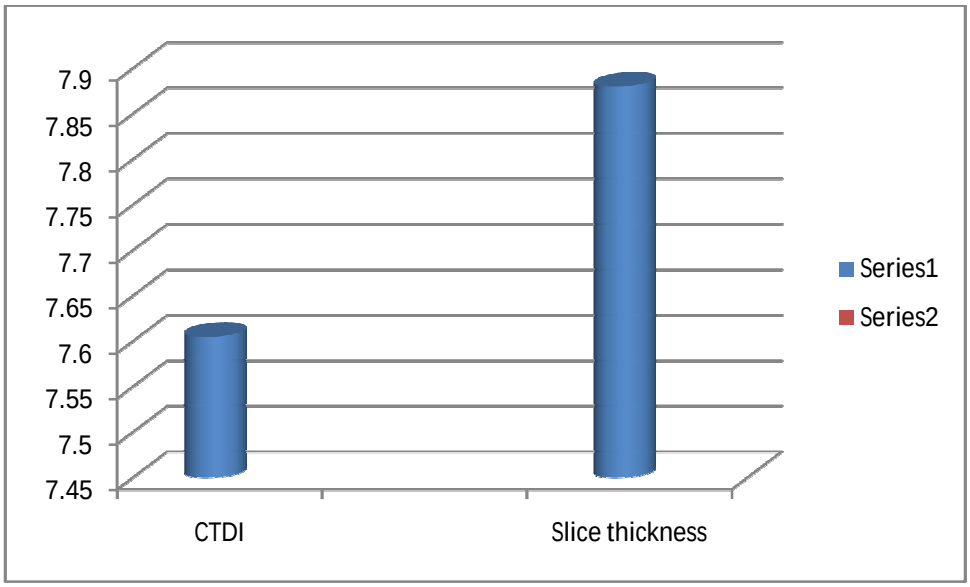


Figure 4.3 shows the coloration between CTDI and slice thickness

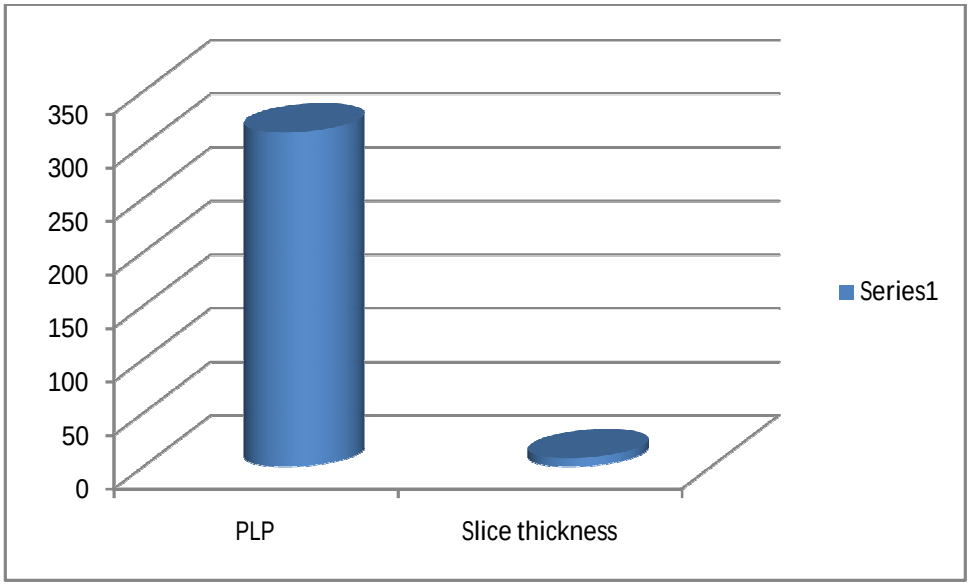


Figure 4.4 shows the coloration between the DLP and slice thickness

Table 4.4 DLP and E_{DLD} and effective dose

DLP	E_{DLP}	E
311.77	**0.12**	**37.412**

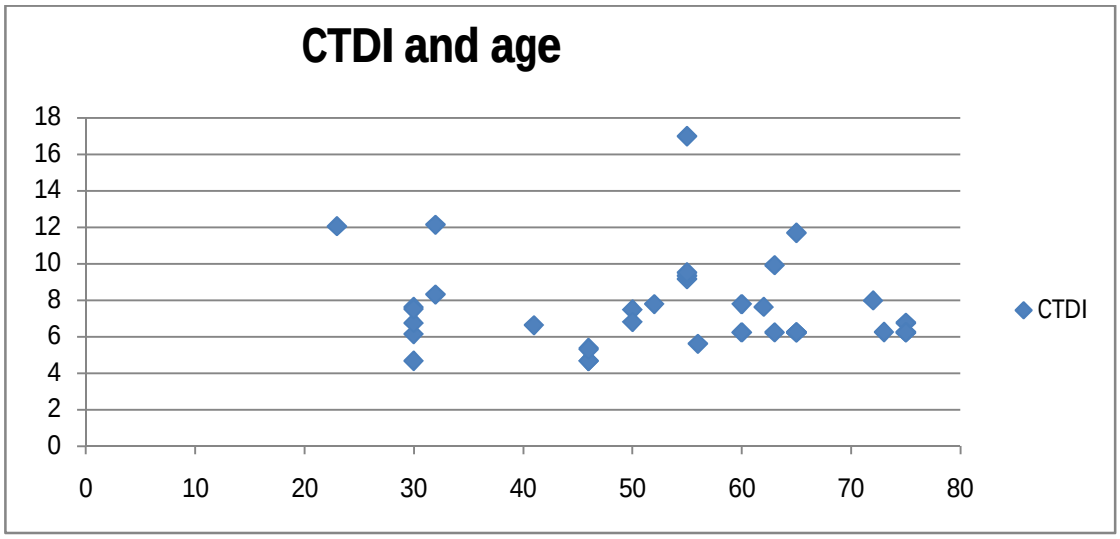


Figure (4.5) showed the CTDI and age

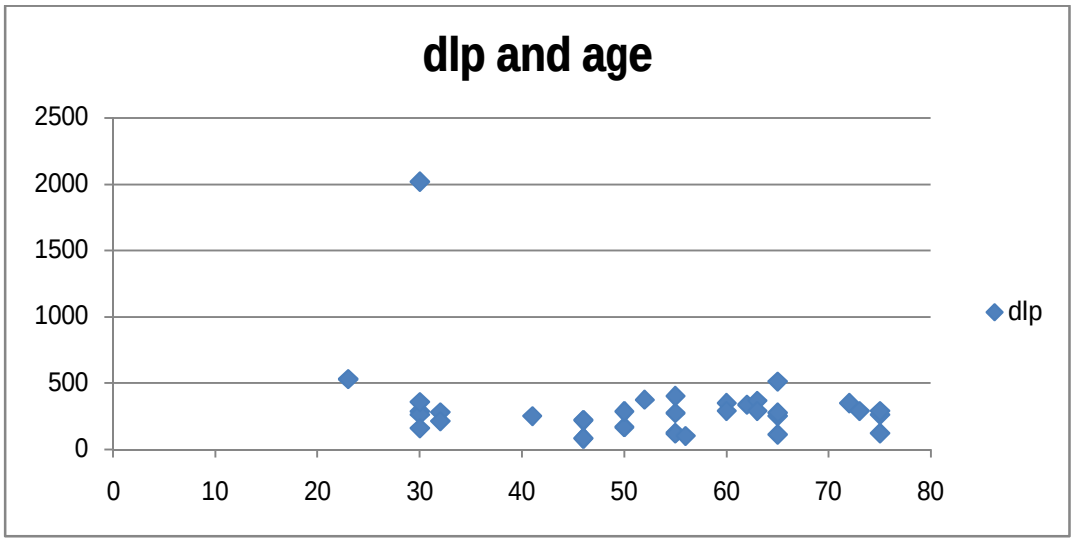


Figure (4.6) showed the DLP and age

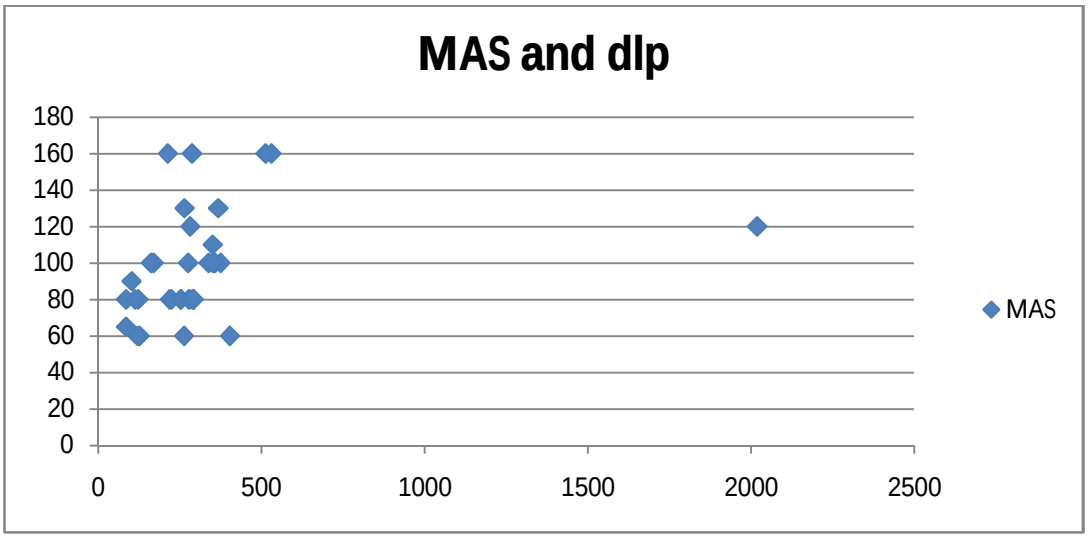


Figure (4.7) showed the MAS and DLP

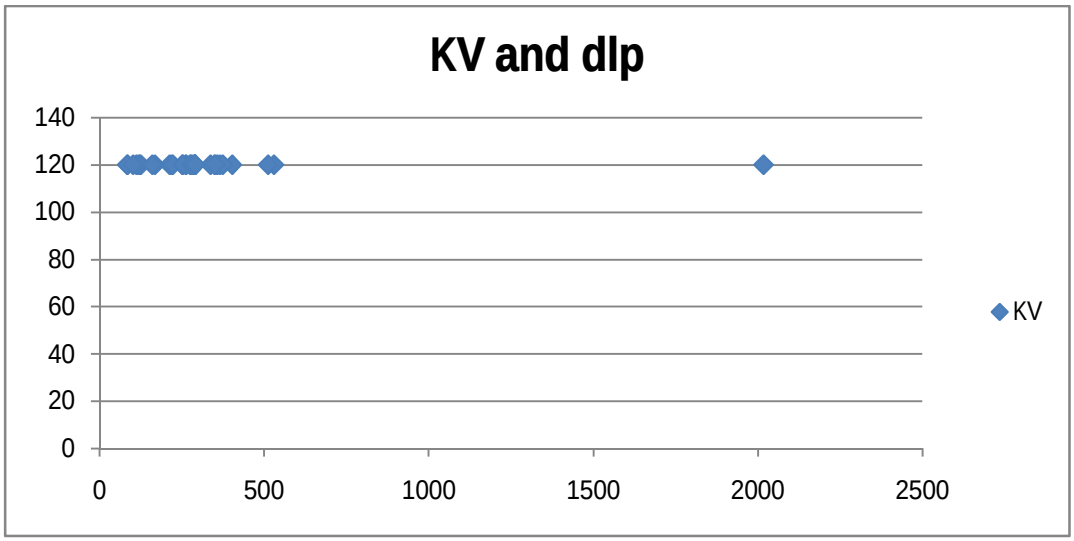


Figure (4.8) showed the KV and DLP

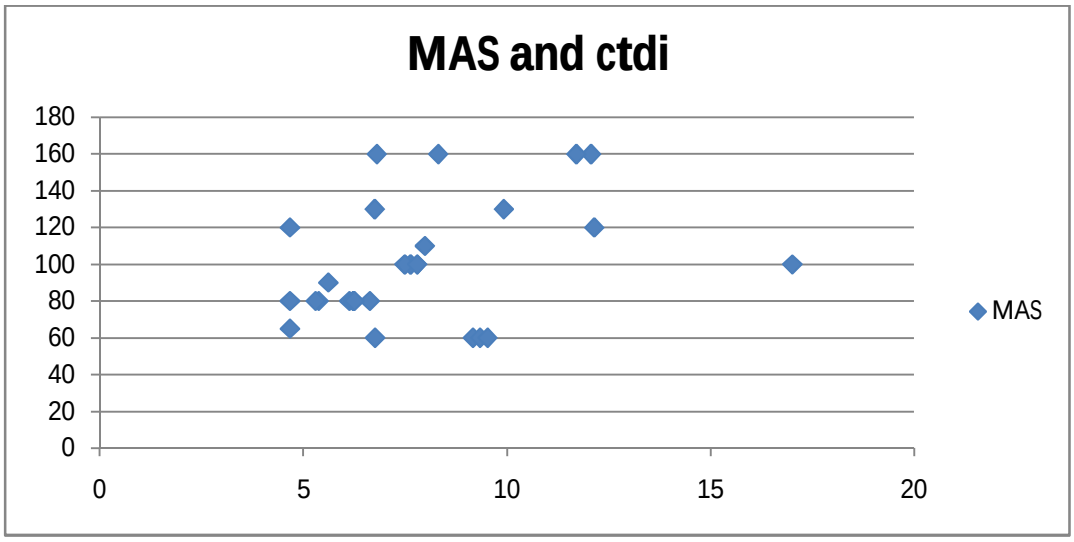


Figure (4.9) showed the MAS and CTDI

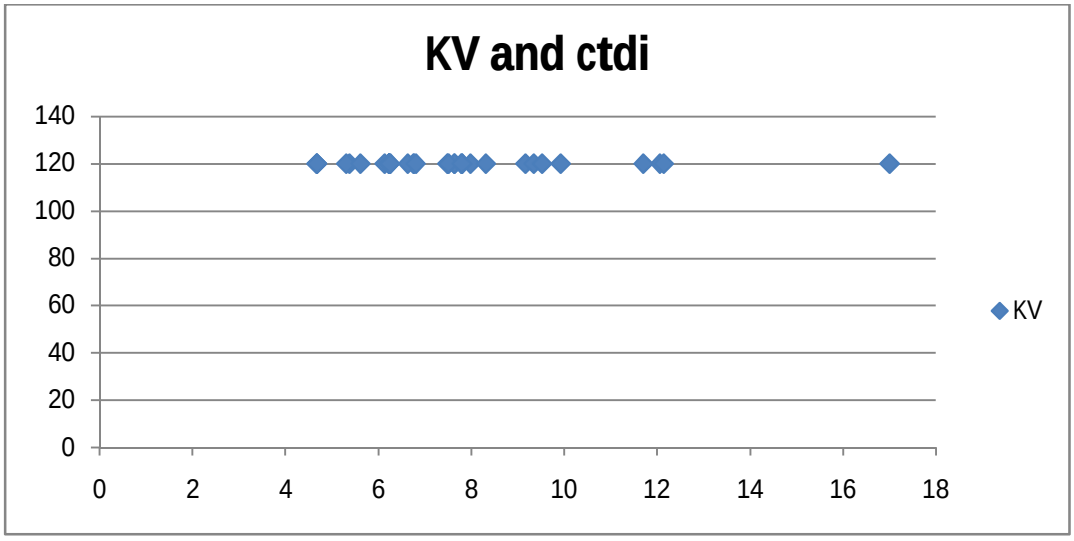


Figure (4.10) showed the KV and CTDI

Chapter Five

Discussion conclusion and recommendations

5.1Discussion:

Radiation dose is measure of how much energy is absorbed when something or someone is exposed to X-rays .this is important because it is this absorption of energy that can cause damage to a person.There are two commonly used ways of quantifying the radiation dose in CT scan procedures: CT dose index (CTDI) and dose length product (DLP)and effective dose .CT dose index and dose length product is easier to measure and is used in national surveys of CT scan equipment.

Effective dose is complicated to calculate but the value can be related directly to radiation –associated risk .

It is important to note that pediatric patients are more sensitive to radiation up to four times higher than the adults . We found in this study that the CTDI is about 7.604857mGy and the DLP 311.777 mGy cm and E=37.41 sv in table 4.3 found that CTDI and DLP inversion proportional with slice thickness . and not coloration between age and Dlp ,CTDI .the dose of CT is related to clinical Applaction

the radiation dose for this study were higher than the most reference levels for other CT application.

5.2 Conclusion:

The radiation dose for this study were higher than the most reference levels for other CT applications in some country CT scan procedure imposes t he high radiation dose for the patients .No protection during CT scan of abdomen procedure. Establishing DRLs in CT radiology is essential for radiation dose reduction.

5.3 Recommendation:

- QC program are essential to evaluate the patients dose and machine accurate performance.
- National survey is important to evaluate the level of dose in CT scan of abdomen.
- Staff training in radiation protection aspects is essential.
- CT scan examination for any patient must justify.
- Future studies must consider patient index or performed on abdomen phantom to develop a proper CT protocol balancing between radiation dose and image quality

References

AAPM Report 204 , size specific Dose Estimates in Pediatric and Adult Body CT Examinations.

American College of Radiology ,ACR practice Guideline for Daignostic Reference Level in Medical X-RAY Imaging ,'2002 (revised 2008).

.

Author: Lawrence M .Davis, MD Editor :William C. Shiel Jr., MD,FACP, FACR.

Abdelrahman Mohamed elnour, Optimization of Radiation Dose in Abdominal using Computerized Tomography, Sudan University of Science and Technology,College of Graduate Studies

Alice B. Smith, William P. Dillon, Radiation Dose Reduction Strategy for CT Protocols: Successful Implementation in Neuroradiology Section. Br J Radiol 71:1296-1301

Besancon A ., Buchillier-Decka I., et al ., "Adult Reference Levels in Dignostic and Interventional Radiology for Temporary Use in Switzerland "Radiation Protection Dosimetry July 20,2004.

British Journal of Radiology (BJR) 74 (2001), 836-840 (C) 2001

Brix G.," Notice of Diagnostic Reference Values for Radiology and Nuclear Medicine Studies " Federal Office for Radiation Protection , ed Germany ;2003 (available in German only).

Dan E. Ware, MD Walter Huda, PhD, Radiation Effective Doses to Patients Undergoing Abdominal CT Examinations , Radiology 1999; 210:645-

650.
David J. Brenner, and Eric J. Hall ,Computed Tomography — An Increasing Source of Radiation Exposure, n engl j med 357;22 www.nejm.org november 29, 2007

European Guidelines on Quality Criteria for Computed Tomography, EUR162, May 1999. http://www.drs.dk/guidelines/ct/quality/htmlindex.htm

European Journal ofRadiology ,June 2004.

Hidajat N, Maurer J, Schroder RJ et al. Relationship between physical dose quantities and patient dose in CT. BJR. 1999; 72:556-561.

Hart D, Jones DG, Wall BF. NRPB Software SR262: Normalized Organ Doses for Medical X-Ray Examinations Calculated Using Monte Carlo Techniques. National Radiological Protection Board, Didcot, Oxon (1994).

Hart D, Jones DG, Wall BF. NRPB Report 262: Estimation of Effective Dose in Diagnostic Radiology from Entrance Surface Dose and Dose-Area Product Measurements. National Radiological Protection Board, Chilton, Oxon, UK (1994).
28]Intemational Atomic Energy Agency (IAEA). Protection of patients in diagnostic and intervention radiology, nuclear medicine and radiotherapy. Proceeding of International Conference, Malaga, Spain, 26-30 March 2001; Vienna
International Commission of Radiological Protection. 1990 Recommendations of the International Commission of Radiological Protection.(Report 60)

Intemational Commission of Radiological Protection. Draft Recommendations of the International Commission of Radiological Protection, available online athttp://www.icrp.org/docs/ICRP_Recs_02_276_06_web_cons_5_June.pdf. Accessed on 11.04.07

Jones DG, Shrimpton PC. Normalised organ doses for X-ray computed tomography calculated using Monte Carlo techniques. NRPB-SR250. National Radiological Protection Board (NRPB).

Julian Simpson , Computed Tomography ,General Practitioner 1991

K. A. Jessen P. C. Shrimpton ,Dosimetry for optimisation of patient protection in computed tomography 1991

Keith J. Strauss Marilyn J. Goske *'Image Gently.* Ten Steps You Can Take to Optimize Image Quality and Lower CT Dose for Pediatric Patients,A/i? 2010; 194:868-873.

library.usyd.edu.au/bitstream/

Shrimpton PC., Wall B.F.,Hart D," Diagnostic Medical Exposure in the UK January 1991.

Walter Huda ,Awais Vance ,Patient Radiation Doses from Adult and Pediatric, American Roentgen Ray Society, CTAJR2007; 188:540

WWW.emedicinehealth .com/ ct scan