Sudan University of Science and Technology

College of Graduate Studies

Estimation of Radiation Hazards of Computed Tomography Dose in Khartoum State

تقدير مخاطر جرعة الأشعة المقطعية المستخدمة بمستشفيات ولاية الخرطوم

M.Sc Degree in Diagnostic Radiological Technology

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Chapter One

Introduction

1.1 **Historical background:**

Since the discovery of X.ray in November, 8, 1895, X.ray have been widely used as the most important and reliable scientific tool for effecting proper diagnosis of diseases and for assessing the results of the treatment given to patients. In1895 F.H. Williams succeeds in taking the first chest X.ray, in 1903 E.A.O pasche builds a collimator for suppressing scattered radiation , in 1913 Gustar Backy develops the scattered radiation grid in Berlin , Germany, Radiation protection for all parri es involved for the first time . In 1930/31 Allesandro Vallebona and Bernard Ziedees des Plantes , thus paring the way For tomography. In 1972 in London , Godfrey N .Hunsfield‘s development of Computed tomography marks the beginning of a new era in diagnostic imaging. In 1974 first CT system from a medical equipment manufacturer (Siemens 2004).

The developments of CT is marked by several developments in scanning Principles detectors as well as the reconstruction mathematics and computers. One major development in CT is the introduction of slip ring scanners that allow

Data to be collected faster than conventional scanners .(Euclid et al 2008).

1.2 **Benefits of CT :**

The benefits of CT scanning are, CT is painless, noninvasive and accurate. Using a spiral CT unit to examine children is faster than older CT scanners, reducing the need for sedation and general anesthesia. Major advantage of CT is that, it is able to image bone , soft tissue and blood vessels all at the some time. CT examination are fast and simple in emergence cases , they can reveal internal injuries and bleeding quickly enough to help save lives . It can be performed if patient have an implanted medical device of any kind(unlike MRI). CT imaging provides real -time imaging , making it a good tool for guiding minimally invasive procedures such as needle biopsies and needle aspiration of many areas of the body , particularly the lung , abdomen , pelvis and bone. Clinical application of 3D imaging range from craniofacial surgical planning and orthopedics to neurosurgery , cardiac surgery ,evaluation of the lungs, and radiation treatment planning (Euclid et al 2008).

1.3 Radiation Hazards from CT:

As in many aspects of medicine, there are both benefits an hazards associated with the use of CT. The main hazards are those associated with abnormal test results for a benign or incidental finding, leading to unneeded, possibly invasive , follow-up tests that may present additional hazards and the increased of cancer induction from X-ray radiation exposure. The probability for absorbed X.rays to induce cancer or heritable mutations leading to genetically associated diseases in offspring is thought to be very small for radiation doses of the magnitude that are associated with CT procedures(Euclid et al 2008). Such estimates of cancer and genetically heritable risk from X-ray exposure have abroad range of statistical uncertainty, and there is some scientific controversy regarding the effects from very low doses and dose rates as discussed below. Under some rare circumstances of prolonged , high-dose exposure, X. rays can cause other adverse health effects, such as skin erythema (reddening), skin tissue injury, and birth defects following in-utero exposure. But at the exposure levels associated with most medical imaging procedures, including most CT procedures, these other adverse effects would not occur (Fred et al 2008).

In the field of radiation protection, it is commonly assumed that the risk for adverse health effects from cancer is proportional to the amount of radiation dose absorbed and the amount of dose depends on the type of x-ray examination. A CT examination with an effective dose of 10 millisieverts (mSv.) may be associated with an increase in the possibility of fatal cancer of approximately 1 chance in

2000. This increase in the possibility of a fatal cancer from radiation can be compared to the natural incidence of fatal cancer in the population, about 1 chance in 5. In other words, for any one person the risk of radiation-induced cancer is much smaller than the natural risk of cancer. Never the less, this small increase in radiation-associated cancer risk for an individual can become a public health concern if large numbers of the population undergo increased numbers of CT screening procedures of uncertain benefit (Fred et al 2008).

It must be noted that there is uncertainty regarding the risk estimates for low levels of radiation exposure as commonly experienced in diagnostic radiology procedures . There are some question whether there is adequate evidence for a risk of cancer induction at low doses. However, this position has not been adopted by most authoritative bodies in the radiation protection and medical areas. The effective doses from diagnostic CT procedures are typically estimated to be in the range of 1 to 10 mSv . This range is not much less than the lowest doses of 5 to 20 mSv received by some of the Japanese survivors of the atomic bombs . These survivors , who are estimated to have experienced doses only slightly larger than those in CT , have demonstrated a small but increased radiation - related excess relative risk for cancer mortality (Fred et al 2008).

Radiation dose from CT procedures varies from patient to patient.

particular radiation dose will depend on the size of the body part examined, the type of procedure, and the type of CT equipment and its operation. Typical values cited for radiation dose should be considered as estimates that cannot be precisely associated with any individual patient, examination, or type of CTsystem (Brenner 2004).

The tremendous advances in computed tomography (CT) technology and applications have increased the clinical utilization of CT, creating concerns about individual and population doses of ionizing radiation. Scanner manufacturers have subsequently implemented several options to appropriately manage or reduce the radiation dose from CT. Modulation of the x-ray tube current during scanning is one effective method of managing the dose. However, the distinctions between the various tube current modulation products are not clear from the product names or descriptions. Depending on the scanner model, the tube current may be modulated according to patient attenuation or a sinusoidal-type function. The modulation may be fully preprogrammed , implemented in near real - time by using a feedback mechanism , or achieved with both preprogramming and a feedback loop. The dose modulation may occur angularly around the patient, along the long axis of the patient, or both. Finally, the system may allow use of one of several algorithms to automatically adjust the current to achieve the desired image quality (Brenner 2004).

Modulation both angularly around the patient and along the z-axis is optimal, but the tube current must be appropriately adapted to patient size for diagnostic image quality to be achieved (Michael 2006).

1.4 CT doses measurement profile:

In conventional radiography, radiation dose decreases continuously from the beam's entrance in to the body to its exit , whereas in CT the dose is distributed more uniformly across the scanning plane because the patient is equally irradiated from all directions. In a head CT examination , for instance ,the dose is uniform across the field of view . In larger objects such as the abdomen , the dose is equally distributed around the periphery of the scanned object and decreases by a factor of only two near the center of the object . Hence,in comparisons between CT and conventional radiography in terms of skin dose are not appropriate. Furthermore, the radiation volume , scattered radiation , divergence of the radiation beam , and limits to the efficiency of beam collimation all contribute to the radiation exposure beyond the boundaries of the scan volume. In the case of the multiple scan acquisition required to image some length of a patient's anatomy , it becomes essential the effect of the radiation dose delivered beyond the boundaries of a single scan , the radiation dose descriptor known as the CT dose index , or CTDI , integrates the radiation dose delivered both within and beyond the scan volume . The average across the field of view to take into account variations in absorbed dose from the periphery to the center of the object results in a dose descriptor known as the weighted CTDI ,or $CTDI_w$.

$CTDI_w$ represents the average dose in the scan volume for contiguous CT scan . In the case when there is either a gap or an overlap between sequential scans , $CTDI_w$ must be scaled accordingly , resulting in the dose descriptor volume CTDI, or $CTDI_{vol}$.

$CTDI_{vol}$ represent the average dose within scan volume (relative to standardized CT phantom) and is now required to be displayed on the user interface of the CT scanner (Mannudeep et al 2004).

1.5 The statement of the problem:

CT in the main radiation source to the general population worldwide. There is an increased CT usage owing to the technological advancement. Multiple slice scanners produce better image quality with less scan time and more clinical applications quality is linked to higher radiation dose to patients. In Sudan, few studies was conducted regarding patient dose in CT examinations. In addition, CT dose optimization protocol is not implemented in all departments yet.

1.6 Objectives of the study:

General objective:

To measure patient dose during CT examination for the brain, chest, abdomen and pelvis for dual, sixteen and sixty four slice CT machines.

Specific objective:

To estimate the region dose.

To estimate the radiation hazard

1.7 Organization of the study:

- **Chapter One:** Introduction.
- **Chapter Two:** Literature Review.
- **Chapter Three:** Materials and Methods.
- **Chapter Four:** Results.
- **Chapter Five:** Discussion, Conclusion and Recommendations.

References

Chapter Two

Theoretical Background

2.1 Conventional X. ray Images:

All x-ray imaging is based on the absorption of X. rays as they pass through

the different parts of a patient's body. Depending on the amount absorbed in a

particular tissue such as muscle or lung, a different amount of X. rays will pass

through and exit the body. The amount of X. rays absorbed contributes to

the radiation dose to the patient. During conventional X.ray imaging, the exiting

X. rays interact with a detective device (X.ray film or other image receptor) and

provide a 2-dimensional projection image of the tissues within the patient's body an X.ray produced "photograph" called a "radiograph." The chest X. ray

(Figure 2.1)show is the most common medical imaging examination. During this

examination,an image of the heart, lungs, and other anatomy is recorded on the

film (Herman 2009).

Fig (2.1) Shows the Chest X. ray Image (Herman 2009).

2.2 Computed Tomography (CT):

Although also based on the variable absorption of X. rays by different

tissues, computed tomography (CT) imaging , also known as "CAT scanning"

(Computerized Axial Tomography), provides a different form of imaging known

as cross-sectional imaging . The origin of the word "tomography" is from the

Greek word "tomos" meaning "slice" or "section" and "graphe" meaning

"drawing." CT imaging system produces cross-sectional images or "slices" of
anatomy , like the slices in a loaf of bread . The cross – sectional images
(Figure 2.2) are used for a variety of diagnostic and therapeutic purposes
(Herman 2009).

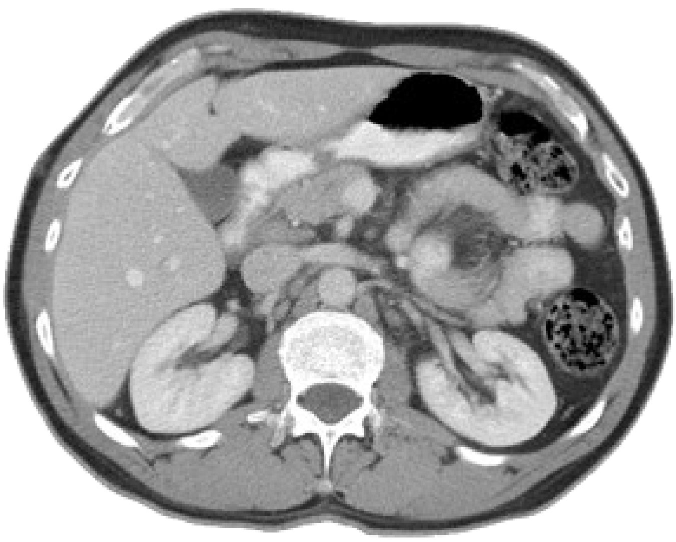


Fig (2.2) Shows the Cross-sectional Image of Abdomen (Herman 2009).

2.3 Generation:

2.3.1 First generation rotate/translate, pencil beam:-

The first generation of CT scanners employed a rotate/translate, pencil beam system. Only two x-ray detectors were used, and they measured the transmission of x-rays through the patient for two different slices. The acquisition of the numerous projections and the multiple rays per projection required that the single detector for each CT slice be physically moved throughout all the necessary positions. This system used parallel ray geometry. Starting at a particular angle, the x-ray tube and detector system translated linearly across the field of view (FOV), acquiring 160 parallel rays across a 24cm. When the x-ray tube/detector system completed its translation, the whole system was rotated slightly, and then another translation was used to acquire the 160 rays in the next projection. This procedure was repeated until 180 projections were acquired at 1-degree intervals. A total of 180 x 160 = 28,800 rays were measured (Jerrold et al 2009).

As the system translates and measures flux from the thickest part of the head to the area adjacent to the head, a huge change in x-ray flux occurred. The early detector systems could not accommodate this large change in signal, and consequently the patient's head was pressed into a flexible membrane surrounded by a water bath. The water bath acted to bolus the x-rays so that the intensity of the x-ray beam outside the patient's head was similar in intensity to that inside the head one advantage of the first-generation CT scanner was that it employed pencil beam geometry-only two detectors measured the transmission of x-rays through the patient. The pencil beam allowed very efficient scatter reduction, because scatter that was deflected away from the pencil ray was not measured by a detector. With regard to scatter rejection, the pencil beam geometry used in first-generation CT scanners was the best (Jerrold et al 2009).

2.3.2 Second generation: rotate/translate narrow fan beam:

The next incremental improvement to the CT scanner was the incorporation of a linear array of 30 detectors. This increased the utilization of the x-ray beam by 30 times, compared with the single detector used per slice in first-generation systems. A relatively narrow fan angle of 10 degrees was used. In principle, a reduction in scan time of about 30-fold would be expected. However, this reduction time was not realized. The shortest scan time with a second-generation scanner was 18 seconds per slice, 15 times faster than with the first-generation system (Jerrold et al 2009).

Although a narrow fan beam provides excellent scatter rejection compared with plain film imaging, it does allow more scattered radiation to be detected than was the case with the pencil beam used in first-generation CT (Jerrold et al 2009).

2.3.3 Third generation rotate / rotate, wide fan beam:

The number of detectors used in third-generation scanners was increased substantially, and the angle of the fan beam was increased so that the detector array formed an arc wide enough to allow the x-ray beam to interrogate the entire patient. However, spanning the dimensions of the patient with an entire row of detectors eliminated the need for translational motion. The multiple detectors in the detector array capture the same number of ray measurements in one instant as was required by a complete translation in the earlier scanner systems. The mechanically joined x-ray tube and detector array rotate together around the patient without translation. The motion of third-generation CT is "rotate/rotate," referring to the rotation of the x-ray tube and the rotation of the detector array. The early third-generation scanners could deliver scan times shorter than 5 seconds. Newer systems have scan times of one half second (Jerrold et al 2009).

The evolution from first to second and second to third-generation scanners involved radical improvement with each step (Jerrold et al 2009).

2.3.4 Fourth generation: rotate / stationary

It is never possible to have a large number of detectors in perfect balance with each other, and this was especially true 25 years ago. Each detector and its associated electronics has a certain amount of drift, causing the signal levels from each detector to shift over time. Detectors toward the center of the detector array provide data in the reconstructed image in a ring that is small in diameter, and more peripheral detectors contribute to larger diameter rings (Jerrold et al 2009).

Fourth-generation CT scanners were designed to overcome the problem of ring artifacts. With fourth-generation scanners, the detectors are removed from the rotating gantry and are placed in a stationary 360-degree ring around the patient, requiring many more detectors. Modern fourth-generation CT systems use about 4,800 individual detectors. Because the x-ray tube rotates and the detectors are stationary, fourth-generation CT is said to use a rotate/stationary geometry (Jerrold et al 2009).

During acquisition with a fourth-generation scanner, the divergent x-ray beam emerging from the x-ray tube forms a fan-shaped x-ray beam. However, the data are processed for fan beam reconstruction with each detector as the vertex of a fan, the rays acquired by each detector being fanned out to different positions of the x-ray source, whereas fourth-generation uses a detector fan. The third-generation fan data are acquired by the detector array simultaneously, in one instant of time. The fourth-generation fan beam data are acquired by a single detector over the period of time that is required for the x-ray tube to rotate through the arc angle of the fan. With fourth-generation geometry, each detector acts as its own reference detector (Jerrold et al 2009).

2.3.5 Fifth generation: stationary / stationary

A novel CT scanner has been developed specifically for cardiac tomographic imaging. This "cine-CT" scanner does not use a conventional x-ray tube; instead, a large arc of tungsten encircles the patient and lies directly opposite to the detector ring. X-rays are produced from the focal track as a high-energy electron beam strikes the tungsten. There are no moving parts to this scanner gantry. The electron beam is produced in a cone-like structure behind the gantry and is electronically steered around the patient so that it strikes the annular tungsten target. Cine-CT systems, also called electron beam scanners, are marketed primarily to cardiologists. They are capable of 50-msec scan times and can produce fast-frame-rate CT movies of the beating heart (Jerrold et al 2009).

2.3.6 Sixth generation: helical / spiral CT

Helical (spiral) scanners were made possible by advances in slip ring connection technology. slip ring consist of brushes that fit into grooves to permit the current and voltage to the x-ray tube to be supplied while the tube is in continuous rotation around the gantry .

2.3.7 Seventh Generation: Multiple Detector Array

X-ray tubes designed for CT have impressive heat storage and cooling capabilities, although the instantaneous production of x-rays (i.e., x-rays per milliampere-second [mAs]) is constrained by the physics governing x-ray production. An approach to overcoming x-ray tube output limitations is to make better use of the x-rays that are produced by the x-ray tube. When multiple detector arrays are used , the collimator spacing is wider and therefore more of the x-rays that are produced by the x-ray tube are used in producing image data. With conventional, single detector array scanners, opening up the collimator increases the slice thickness, which is good for improving the utilization of the x-ray beam but reduces spatial resolution in the slice thickness dimension. With the introduction of multiple detector arrays, the slice thickness is determined by the detector size and not by the collimator. This represents a major shift in CT technology (Jerrold et al 2009).

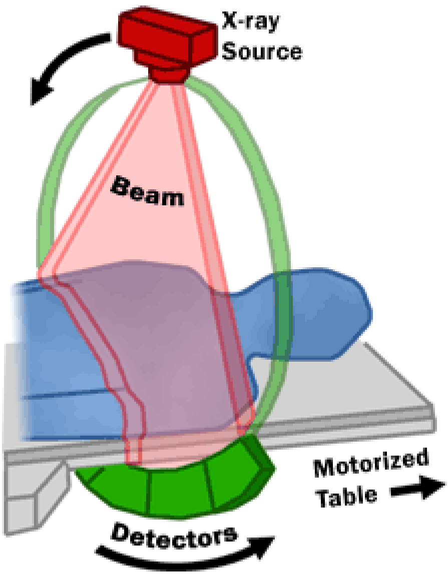
The flexibility of CT acquisition protocols and increased efficiency resulting from multiple detector array CT scanners allows better patient imaging; however, the number of parameters involved in the CT acquisition protocol has increased as well. Also with multiple detector arrays, the notion of helical pitch needs to be redefined (Jerrold et al 2009).

2.4 How a CT system works:

A motorized table moves the patient (Figure 2.3) through a circular opening in the CT imaging system. (Herman 2009).

Fig(2.3) Shows the Patient in CT Imaging System(Herman 2009).

As the patient passes through the CT imaging system, a source of X. rays rotates around the inside of the circular opening. A single rotation takes about 1 second. The X.ray source produces a narrow, fan-shaped beam of X.rays used to irradiate a section of the patient's body (Figure 2.4). The thickness of the fan beam may be as small as 1 millimeter or as large as 10 millimeters. In typical examinations there are several phases; each made up of 10 to 50 rotations of the X.ray tube around the patient in coordination with the table moving through the circular opening. The patient may receive an injection of a "contrast material" to facilitate visualization of vascular structure (Herman 2009).



Fig(2.4) Shows the CT Fan Beam(Herman 2009).

Detectors on the exit side of the patient record the X. rays exiting the Section of the patient's body being irradiated as an X.ray "snapshot" at one position (angle) of the source of X. rays. Many different "snapshots" (angles) are collected during one complete rotation (Herman 2009).

The data are sent to a computer to reconstruct all of the individual "snapshots" into a cross-sectional image (slice) of the internal organs and tissues for each complete rotation of the source of x. rays (Herman 2009).

2.5 Advances in Technology and Clinical Practice:

Today most CT systems are capable of "spiral" (also called "helical")

scanning as well as scanning in the formerly more conventional "axial" mode.

Inaddition, many CT systems are capable of imaging multiple slices

simultaneously. Such advances allow relatively larger volumes of anatomy to be

imaged in relatively less time. Another advancement in the technology is electron

beam CT, also known as EBCT. Although the principle of creating cross-sectional

images is the same as for conventional CT, whether single- or multi-slice, the

EBCT scanner does no t require any moving parts to generate the individual

"snapshots." As a result , the EBCT scanner allows a quicker image acquisition

than conventional CT scanners (Herman 2009).

2.6 Radiation Dose:

When ionizing radiation penetrates the human body or an object, it deposits energy . The energy absorbed from exposure to radiation is called a dose. Radiation dose quantities are described in three ways: absorbed, equivalent, and effective.

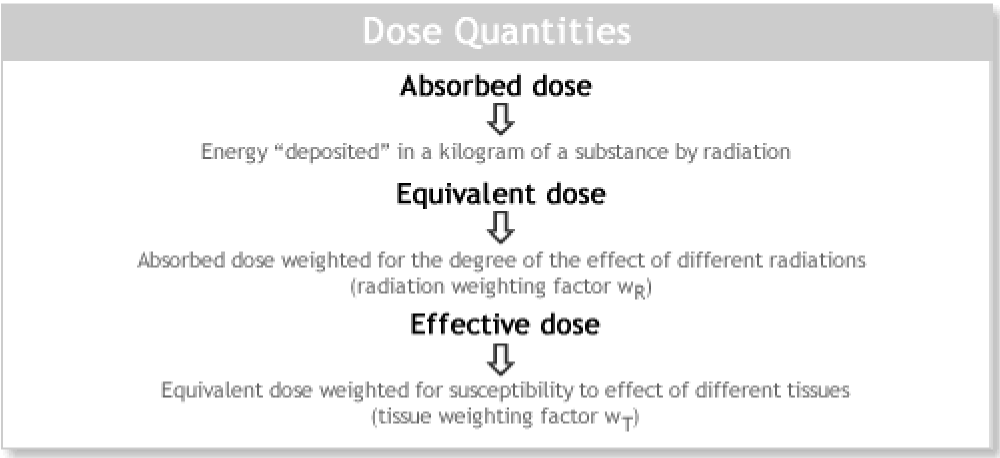


Fig (2.5) Shows the **Dose Quantities**(Smith et al 2009).

2.6.1 Absorbed dose:

The amount of energy deposited in a substance (e.g., human tissue), is called the absorbed dose. The absorbed dose is measured in a unit called the gray (Gy). A dose of one gray is equivalent to a unit of energy (joule) deposited in a kilogram of a substance (Smith et al 2009).

2.6.2 Equivalent dose:

When radiation is absorbed in living matter, a biological effect may be observed. However, equal absorbed doses will not necessarily produce equal biological effects. The effect depends on the type of radiation (e.g., alpha,beta, gamma, etc) and the tissue or organ receiving the radiation. For example, 1 Gy of alpha radiation is more harmful to tissue than 1 Gy of beta radiation.

Aradiation weighting factor (wR) is used to equate different types of radiation with different biological effectiveness. This weighted absorbed quantity is called the equivalent dose and is expressed in a measure called the sievert (Sv). This means that 1 Sv of alpha radiation will have the same biological effect as 1 Sv of beta radiation(Smith et al 2009).

Because doses to workers and the public are so low, most reporting and dose measurements use the terms millisievert (mSv) and microsievert (µSv) which are 1/1000 and 1/1000000 of a sievert respectively. These smaller units of the sievert are more convenient to use in occupational and public settings(Smith et al 2009).

To obtain the equivalent dose, the absorbed dose is multiplied by a specified radiation weighting factor (wR). The equivalent dose provides a single unit which accounts for the degree of harm of different types of radiation (Smith et al 2009).

2.6.3 Effective dose:

Different tissues and organs have different radiation sensitivities. For example, bone marrow is much more radiosensitive than muscle or nerve tissue. To obtain an indication of how exposure can affect overall health, the equivalent dose can be multiplied by a factor related to the risk for a particular tissue or organ. This multiplication provides the effective dose absorbed by the body. The unit used for effective dose is also the sievert(Smith et al 2009).

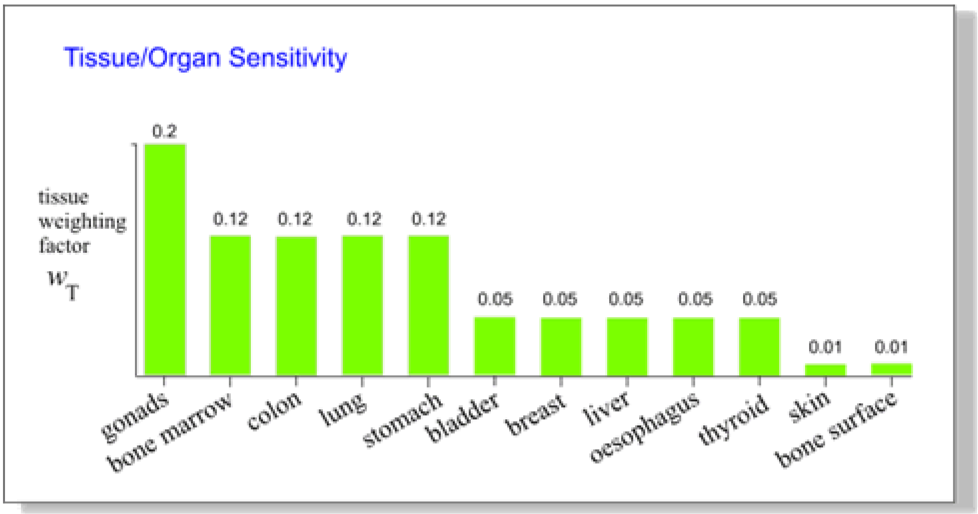


Fig (2.6) Shows the **Effective dose**(Smith et al 2009).

As a simple example, if someone's stomach and bladder are exposed separately to radiation, and the equivalent doses to the tissues are 100 and 70 mSv respectively, the effective dose is : (100 mSv x 0.12) + (70 x 0.05) = 15.5 mSv. The risk of harmful effects from this radiation is equal to 15.5 mSv received uniformly through the whole body (Smith et al 2009).

2.7 RADView CT Dosimeter:

The RADView™ CT dosimeter badge is a1.2″X2.4″ plastic radiation dosimeter that provides visual measurement of absorbed radiation dose to patients who are undergoing CT scan . The RADView™ is radiologically equivalent to soft tissue, which means that it is effectively transparent to x.ray at the kVp setting used for CT scan, but nevertheless measures the surface dose from x.ray treatment. The RADView™ dosimetery process employs a self developing radiation sensitive film that is calibrated for radiation dose (D.Tack et.al 2009).

Fig (2.7) Shows RADView CT Dosimeter(D.Tack et.al 2009).

If the circle is darker than the surrounding box, the dose is higher than that threshold. If the circle is lighter than box, the dose is less than the indicated threshold (D.Tack et.al 2009).

2.8 CT parameter:

2.8.1 Tube potential (kVp):

The relationship between the dose and the tube potential is not straight and linear one tube potential is usually modified only through the (kVp) setting. These kVp values differ from one manufacture to another, as well from one CT scanner to another, and vary from 80kVp to 140kVp(ICRU 1979).

2.8.2 Tube current – Time product (mAs):

As in convential radiography, a straight linear relationship exists between the mAs and the dose. The setting for mAs should be adapted to the characteristics of the scanner unit, the patient's size, and the dose requirements for the each type of examination (ICRU 1979).

2.8.3 Pitch:

Defined as tube distance traveled in one 360 rotation /total collimated width of x-ray beam (ICRU 1979).

2.9 Radiation dose unit:

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 sensitivity to radiation exposure , the actual dose to different parts of the body for an 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 tissue 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 technologist hav e been trained to use the minimum amount of radiations necessary to obtain the needed results(D.Tack et.al 2009).

The decision to have an x-ray exam is a medical one , based on the like hood of benefit from the exam and the potential risk from radiation (D.Tack et.al 2009).

2.10 Methods of measurement in CT:

In CT the dose is distributed more uniformly across the scanning plane because the patient is equally irradiated from direction .

2.10.1 Computed Tomography Dose Index (CTDI):

Is the primary dose measurement concept in CT.

$$CTDI = \frac{1}{NT}\int D(Z)\, dz \quad ....(2.1)$$

Where:

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

N = the number of slice in a single axial scan.

T = the width of the tomographic section along the z-axis imaged by one data channel.

CTDI represent the average absorbed dose , along z-axis , from a series of rotation (Jessenk et al 2000). The CTDI is always measured in the axial scan mode(Jessenk et al 2000).

2.10.2 Weighted $CTDI_w$:

The CTDI varies across the field of view (FOW) for example, for body CT Imaging , the CTDI is typically a factor or two higher at the surface than the center of the FOV. The average CTDI across the FOV is estimated by the weighted CTDI ($CTDI_w$) (Jessenk et al 2000).

Where:

$$CTDI_w = {}_{1/3}\,CTDI_{100,center} + {}_{2/3}\,CTDI_{100,edge} \text{ --------------(2.2)}$$

$CTDI_{100}$ = represents the accumulated multiple scan dose at the center of 100 mm scan and underestimates the accumulated dose for longer scan length.

2.10.3 Volume $CTDI_{vol}$:

To represent dose for a specific scan protocol , which almost always involves a series of scans , it is essential to take in to a count any gaps or overlaps between the X.ray beam from consecutive rotation of the X.ray source This is accomplished with use of a dose descriptor known as the volume $CTDI_w$ ($CTDI_{vol}$).

Where:

$$CTDI_{vol} = N*T/I * CTDI_w \text{------------------}(2.3)$$

And:

I = the table increment per axial scan (mm).

Since pitch is defined as the ratio of the table travel per rotation (I) to the total nominal beam width (N*T).

$$pitch = I/(N*T) \text{--------------------}(2.4).$$

Thus:

$CTDI_{vol}$ can be expressed as

$$CTDI_{vol} = 1/pitch * CTDI_w \text{------------}(2.5)$$

2.10.4 Dose Length Product (DLP):

To better represent the overall energy delivered by given scan protocol, the absorbed dose can be integrated along the scan length to compute the **Dose Length Product (DLP)** (Jessenk et al 2000).

Where:

$$DLP\ (mGy.cm) = CTDI_{vol}\ (mGy) * \text{scan length (cm)------}(2.6)$$

2.11 Radiation Health Effects:

The word “safe” means different things to different people. For many, the idea of being safe is the absence of risk or harm. However, the reality is that there is a level of risk in almost everything we do. For example, speed limits on roads are set to maximize safety. Never the less, accidents occur even when drivers are obeying the speed limit. Despite the risks, we make a conscious decision to drive.

Similar conscious decisions are made when radiation is used. Radiation exposure carries a health risk. Knowing what the risks are helps the CNSC and other regulatory bodies set dose limits and regulations that limit exposure to an acceptable or tolerable risk (some may even say a safe limit) (Shapiro 2002).

One significant advantage with radiation is that more is known about the health risksc associatedc with it than with any other chemical or otherwise toxic agent. Since the early twentieth century, radiation effects have been studied in depth in both the laboratory and among human populations . Since the establishment of the United Nations Scientific Committee on the Effects of Atomic radiation (UNSCEAR) in 1955 , the mandate of the Committee has been to undertake broad assessments of the sources of ionizing radiation and its effects on human health and the environment t. Those assessments provide the scientific foundation used in formulating international standards for the protection of the general public and workers against ionizing radiation . The UNSCEAR 2010 report (PDF, source : UNSCEAR Web site) consolidates and summarizes , in simple terms, the Committee's detailed understanding of the low-dose radiation effects on health (Shapiro 2002).

2.12 Previous Studies:

Study“Measurement of Pediatric Radiation Dose in Computed Tomography Examination” by A Alsadeg (2009) The assessment of radiation dose to pediatric patient undergoing CT brain , abdomen and chest investigated . In this study variation in doses were observed , the radiation dose is higher in Al –Ribat university hospital than in El-Nilein diagnostic centre , and in general the mean values of doses are higher for CT brain and lower for abdomen and chest compare to other studies . Different data in request form were responsible for these variations. The main contributor for this high dose was the use of different techniques and use for adult protocol , which justify the important of use child protocol . In addition the study has shown a great need referring criteria , continuous training of staff in radiation protection concepts especially for pediatric.

Study “Optimization of Radiation Dose in Abdomen Using Computerized Tomography ” by A.M.Elnour (2009) include 83 patient the DLP was 277.25 mGy.cm and $CTDI_{vol}$ was 9.7. mGy . Optimization could be achieved through optimal study , body region of interest being scanned , and patient size.

Study “Optimization of Radiation Dose in Multislices Computerized Tomography scan Examination” by A.M.Abdelrazig include 130 patient the DLP was2344.4 mGy.cm and $CTDI_{vol}$ was 178.3 mGy.

Study “Radiation Dose Associated With Common Computed Tomography Examinations and the Associated Lifetime Attributable Risk of Cancer” by Rebecca Smith-Bindman, et al, (2009) titled concluded Radiation doses varied significantly between the different types of CT studies. The overall median effective doses ranged from 2 millisieverts (mSv) for a routine head CT scan to 31 mSv for a multiphase abdomen and pelvis CT scan. Within each type of CT study, effective dose varied significantly within and across institutions, with a mean 13-fold variation between the highest and lowest dose for each study type. The estimated number of CT scans that will lead to the development of a cancer varied widely depending on the specific type of CT examination and the patient's age and sex. Radiation doses from commonly performed diagnostic CT examinations are higher and more variable than generally quoted, highlighting the need for greater standardization across institutions.

study“The Control Of Radiation Exposure From CT Scans”by Biswita C. Mozumdar, (2003) concluded that Computed tomography is a popular diagnostic tool in medicine. The widespread use of CT involves considerable radiation exposure to scan subjects. The radiation burden has come under increased scrutiny in recent years. The use of CT as a screening technique provides an additional dimension to the controversy. The article explores conflicting views with respect to radiation exposure from computed tomography. Recent advances in scan application and technology that offer scope for dose reduction are discussed.

Table 2.1 Demonstrations the main technical factors and dose data for head CT

Country	No. of patient	Milliampere-second	Thickness (mm)	CTDIw (mGy)	DLP (mGy.cm)
United kingdom	10	270	4.5 or 9	51	720
Poland	51	250	5 or 10	19	527
Thailand	36	260	7.5	43	386

Note: data are mean values , a peak voltage 120 kVp was used

Table 2.2 Demonstrations the main technical factors and dose data for chest CT

Country	No. of patient	Milliampere-second	Thickness (mm)	pitch	CTDIw (mGy)	DLP (mGy.cm)
Canada	43	268	2	1	8.8	294
Greece	50	180	7	1	19.5	540
India	50	90	2.5	6	12.3	355
Poland	38	225	9.5	1.5	14.2	447
Thailand	32	90	8	6	7.2	247
United kingdom	30	93	1.5	1	6.4	203

Note: data are mean values, a peak voltage120 kVp was used and 140 for India

Country	No. of patient	Milliampere-second	Thickness (mm)	pitch	CTDIw (mGy)	DLP (mGy.cm)
Canada	43	267	5	1.6	14.4	696
Greece	54	180	7	1	19.5	740
India	52	133	2.5	5	12.6	459
Poland	54	250	9	1.5	15.8	550
Thailand	66	120	8	5.6	9.5	402
United kingdom	25	122	0.75	1	9.5	446

Table 2.3 Demonstrations the main technical factors and dose data for abdomen

Note: data are mean values a peak voltage 120 kVp was used

Table 2.1, Table 2.2 and Table 2.3 presents the patient doses from different countries. The doses values and dose parameters (exposure factors, slice thickness, pitch) showed wide variations. Therefore, these studies show wide differences in terms of doses. This may attributed to the equipment and interexaminers variability, suggesting that patient dose optimizations methods have not been accomplished yet. In general, there are some factors that have a direct influence on radiation dose, such as the x-ray beam energy (kilovolt peak), tube current (in milliamperes), rotation or exposure time, section thickness, object thickness or attenuation, pitch and/or spacing, dose reduction techniques such as tube current variation or modulation, and distance from the x-ray tube to isocenter. In addition, there are some factors that have an indirect effect on radiation dose—those factors that have a direct influence on image quality, but no direct effect on radiation dose; for example, the reconstruction filter. Choices of these parameters may influence an operator to change settings that do directly influence radiation dose.

The term diagnostic reference level or reference value sets an investigation level to identify unusually high radiation doses or exposure levels for common diagnostic medical X-ray imaging procedures .

Reference levels are based on actual patient doses for specific procedures measured at a number of representative clinical facilities. The levels are set at approximately the 75th percentile of these measured data, meaning that the procedures are performed at most institutions with doses at or below the reference level. Consequently, reference levels are suggested action levels at which a facility should review its methods and determine if acceptable image quality can be achieved at lower doses. The goal of this guideline is to provide guidance and advice to physicians and medical physicists on the establishment and implementation of reference levels in the practice of diagnostic medical X-ray imaging. The goal in medical imaging is to obtain image quality consistent with the medical imaging task. Diagnostic reference levels are used to manage the radiation dose to the patient. The medical radiation exposure must be controlled, avoiding unnecessary radiation that does not contribute to the clinical objective of the procedure. By the same token, a dose significantly lower than the reference level may also be cause for concern, since it may indicate that adequate image quality is not being achieved. The specific purpose of the reference level is to provide a benchmark for comparison, not to define a maximum or minimum exposure limit.

Table 2.4 Demonstrations the proposed European commission reference levels for some CT examination

Examination	CTDIw(mGy)	DLP(mGy.cm)
Head	60	1050
Chest	30	650
Abdomen	35	780

Chapter Three

Materials and Methods

3.1 Introduction:

This study intended to compare the radiation doses from different CT detectors i.e dual slice, 16 slice and 64 slice during brain, chest and abdomen examination.

3.2 CT equipment:

CT scanner that participated in this study is helical CT scanners in three hospitals. The three scanners were displayed volume Computed Tomography Dose Index ($CTDI_{vol}$) and Dose Length Product (DLP). The data were collected from each CT scanner (manufacture, model, year of installation, Focal Axial Distance (FAD) and detector type.

In general, solid-state detectors are more dose efficient than gas detector. Each detector design had its specific advantage and drawbacks: Separating strips and decreasing sensitivity. All 16-slice scanners introduced in 2001now made use of the same hybrid design, with 16 smaller central detectors, accompanied by a number of larger detectors at both sides (Siemens 2004).

3.3 CT machines:

Table 3.1 Demonstrations the demonstrates CT machine

Slice No.	Manufacture	Model	year of installation	FAD	Detector type

Dual slice	Siemens	Somatom emotion Duo	2006	70cm	2 rows
16 slice	Siemens	Sensation 16	2004	70cm	16 rows
64 slice	Toshiba	Aquilion	2010	70 cm	64 rows

Fig (3.1) Shows the CT machine Siemens_Emotion_Duo_2-Slice

Fig(3.2) Shows the CT machine Siemens-Sensation-16-Slice

Fig(3.3) Shows the CT machine Toshiba_Aquilion_64-Slice

3.4 Patient:

A total of 108 patients (50 male and 58female) examined in this study: as illustrated in Table 3.1

Table 3.2 Demonstrations the number of patient in the different multi detector.

Number of slice	Abdomen	Chest	Brain	total
Dual slice	12	11	13	36
16 slice	12	12	13	37
64 slice	12	11	12	35
Total	36	34	38	108

3.5 Patient preparation, protocol and technique:

3.5.1 Patient preparation:

All metallic objects should be removed.

Use sedation or anesthesia (no motion during scan).

Empty stomach if anesthesia or contrast media indicated.

3.5.2 Patient position:

Patient supine and head first (CT brain) or feet first (chest and abdomen).

Positioning the four light lines (sagittal, coronal and two transverse “internal-external”) to put the part that to be exam in x.ray field.

Head rest on head holder (CT brain).

3.5.3 Technique:

For brain scan by “dual-16 slice” 8-10mm slice thickness “64 slice” 3-5 mm slice thickness for axial mode ,for spiral mode multi detector 0.5 mm slice thickness. Axial cuts from the base of the skull to vertex parallel to the radiographic base line.

For chest scan by “dual-16 slice” 5-10mm slice thickness “64 slice” 2 mm slice thickness for high resolution CT chest 5 mm slice thickness for plain CT chest. Axial cuts from the apex of the lung to base of the lung.

For abdomen scan by “dual-16 and 64 slice” 5mm slice thickness .Axial cuts all the abdomen.

All multi detector CT scanner scan with 0.5 mm then reconstruct the images according to the selected protocol “2mm, 3mm, 5mm .etc”

When the patient lies in correct position we use spiral technique , the advantage of spiral technique was , short scan time and low dose to the patient. The low dose in spiral technique depend on some factors (mAs , KV , pitch slice thickness).

3.6 Cancer Risk Estimation:

The risk (R_T) of developing cancer in a particular organ (T) following CT procedure after irradiation was estimated by multiplying the mean organ equivalent (H_T) dose with the risk coefficients (f_T) obtained from ICRP (Kosucu et.al 2004).

$$RT = HT \; fT \quad \text{-------------------------}(3.1)$$

The overall lifetime mortality risk (R) per procedure resulting from cancer/heritable was determined by multiplying the effective dose (E)by the risk factor (f).

3.7 Interpretation:

Data were collected using a sheet for all patients in order to maintain consistency of the information from display (Appendix).

A data collection sheet was designed to evaluate the patient doses and the radiation related factor. The collected data included patient height, weight, sex, BMI and age; tube voltage and tube current–time product settings; pitch; section thickness; and number of sections. Clinical indications are important factor in patient dose during CT. In addition, we also recorded all scanning parameters, as well as the CT dose descriptors CT dose index volume (in milligrays) and dose-length product (in milligray-centimeters). All these factors that have a direct influence on radiation dose. The entire hospitals were passed successfully the extensive quality control tests performed by Sudan atomic energy commission and met the criteria of this study.

Chapter Four

Results

4.1 Results:

In this study, a total of 108 patients were examined in three Hospitals in Khartoum state over 4 months.

Table 4.1 describes patient's doses from Dual slice, Table 4.2 describes patients doses 16 slice and Table 4.3 describes patients doses 64 slice. Table 4.4 demonstration the compare between the type of CT detectors .

4.2 Dual Slice:-

Table 4.1 Demonstrations the Average patients dose values Dual slice (range in the parenthesis)

CT parameter	Abdomen (12 patient) Average ± sd min-max	Chest (11 patient) Average ± sd min-max	Brain (13 patient) Average ± sd min-max
Tube potential (kVp)	113.1±7.8 (110-130)	130±0 (130-130)	130±0 (130-130)
Current (mA)	60.9±21.7 (27-94)	80.7±42.7 (37-156)	244.6±40.9 (120-260)
Slice thickness(mm)	9.2±1.9 (5-10)	8.5±1.8 (4-10)	5.5±2.5 (3-8)
Pitch	2±0 (1.9-2)	1.6±0.5 (1-2)	1±0 (1-1)

Total slice number	44±17 (19-73)	35±9 (26-56)	13±2 (11-15)
$CTDI_W$ (mGy)	8.7±2.7 (3.6-12.7)	12.9±5.1 (5.2-21)	55.7±9.3 (27.4-59.2)
DLP (mGy.cm)	164±56 (84-241)	238.8±116.2 (96-466)	331.1±161 (126-521)
$CTDI_{vol}$ (mGy)	4.4±1.3 (1.8-6.3)	8.7±4.6 (4-16.9)	55.7±9.3 (27.4-59.2)

Fig (4.1) shows the relation between CTDIw and current for abdomen for Dual slice

Fig (4.2) Shows the relation between DLP and current for abdomen for Dual slice

Fig (4.3) Shows the relation between CTDIw and current for Chest for Dual slice

Fig (4.4) Shows the relation between DLP and current for Chest for Dual slice

4.3 Sixteen Slice:

Table 4.2 Demonstrations the demonstrates Average patients dose for 16 slice (range in the parenthesis)

CT parameter	Abdomen (12 patient) Average ± sd min-max	Chest (12 patient) Average ± sd min-max	Brain (13 patient) Average ± sd min-max
Tube potential (KVp)	120±0 (120-120)	120±0 (120-120)	120±0 (120-120)
Current (mA)	82.7±28.3 (50-138)	80±28.4 (49-151)	218.5±130.5 (100-360)
Slice thickness (mm)	3±1.5 (2-5)	7.5±1.2 (6-10)	3.5±3.1 (0,6-9)
Pitch	0.8±0 (0.8-0.8)	1.1±0 (1.1-1.1)	0.8±0.2 (0.6-1.1)
Total slice Number	310±194 (10-461)	28±10 (8-46)	121±101 (13-267)
$CTDI_W$ (mGy)	4.4±1.5 (2.6-7.2)	6.3±2.2 (3.9-11.9)	35±24 (11.7-80.6)
DLP (mGy.cm)	270.6±100 (155-470)	209.6±83.3 (135-436)	670±380 (242-1306)
$CTDI_{vol}$ (mGy)	5.8±2 (4-10)	5.6±2 (3.5-10.3)	45.4±24.8 (21.3-80.6)

Fig(4.5) Shows the relation between CTDIw and current for abdomen for 16 slice

Fig (4.6) shows the relation between DLP and current for abdomen for 16 slices

Fig (4.7) Shows the relation between CTDIw and current for chest for 16 slice

Fig (4.8) shows the relation between DLP and current for chest 16 slice

4.4 Sixty-four Slice:

Table4.3 Demonstrations the Average patients dose values for 64 slice (range in the parenthesis)

CT parameter	Abdomen (12 patient) Average ± sd min-max	Chest (11 patient) Average ± sd min-max	Brain (12 patient) Average ± sd min-max
Tube potential (KVp)	120±0 (120-120)	120±0 (120-120)	118.3±5.8 (100-120)
Current (mA)	147.9±7.2 (125-150)	110±28.3 (60-150)	218.8±21.7 (150-225)
Slice thickness (mm)	6.3±3.4 (0.5-10)	6.2±3.5 (0.5-10)	6.2±2.7 (4-14)
Pitch	1.4±0.6 (0.5-2)	1.6±0.5 (0.8-2)	0.9±0.1 (0.8-1.1)
Total slice Number	1244±929 (632-3639)	1938±1052 (777-3706)	365±341 (24-755)
$CTDI_W$ (mGy)	85.3±116.8 (12.2-434)	37.7±19.7 (16.6-76.2)	67.4±18.9 (31.8-103.3)
DLP (mGy.cm)	1444±954.3 (510.8-3136.5)	1121.2±872.1 (203.2-2873.1)	1160.8±435.7 (464.8-1669.9)
$CTDI_{vol}$ (mGy)	75.7±93.9 (12.2-246)	25.2±15 (8.3-58.4)	73.4±21.1 (31.8-96.4)

Table 4.4 Demonstrations the comparison between the three hospitals (range in the parenthesis)

Hospital	Dose	Abdomen	Chest	Brain

Dual slice	$CTDI_W$ (mGy)	8.7±2.7 (3.6-12.7)	12.9±5.1 (5.2-21)	55.7±9.3 (27.4-59.2)
	DLP (mGy.cm)	164±56 (84-241)	238.8±116.2 (96-466)	331.1±161 (126-521)
	$CTDI_{vol}$ (mGy)	4.4±1.3 (1.8-6.3)	8.7±4.6 (4-16.9)	55.7±9.3 (27.4-59.2)
16 slice	$CTDI_W$ (mGy)	4.4±1.5 (2.6-7.2)	6.3±2.2 (3.9-11.9)	35±24 (11.7-80.6)
	DLP (mGy.cm)	270.6±100 (155-470)	209.6±83.3 (135-436)	670±380 (242-1306)
	$CTDI_{vol}$ (mGy)	5.8±2 (4-10)	5.6±2 (3,5-10.3)	45.4±24.8 (21.3-80.6)
64 slice	$CTDI_W$ (mGy)	85.3±116.8 (12.2-434)	37.7±19.7 (16.6-76.2)	67.4±18.9 (31.8-103.3)
	DLP (mGy.cm)	1444±954.3 (510.8-3136.5)	1121.2±872.1 (203.2-2873.1)	1160.8±435.7 (464.8-1669.9)
	$CTDI_{vol}$ (mGy)	75.7±93.9 (12.2-246)	25.2±15 (8.3-58.4)	73.4±21.1 (31.8-96.4)

Fig (4.9) Shows the Comparison between patient doses (mGy.cm) in the three hospitals for the abdomen, chest and brain.

Chapter Five

Discussion, Conclusions and Recommendations

5.1 Discussion:

CT scanning has been recognized as 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 dose 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.

Measure and estimates the impact of irradiating by CT machine during the CT examined is important in order to quantify the dose for further dose optimization.

Radiation doses from CT vary widely, and they could be reduced significantly if strategies for minimizing exposure were more widely followed. in the current work we measured and estimated radiation exposure from different CT scans in 3 hospitals.

In this study, a total of 108 patients suffer from brain, chest and the abdominal CT scanning exams were examined. Patient CT dose in this study were measured and estimated in different CT technologies, Table 3.1 shows that dual-slice (Siemens Somatom Emotion) system and 16-slice (Siemens Sensation) system and sixty four slice (Toshiba).

The results of this study show wide variation in patient dose among different detectors number in terms of DLP and $CTDI_{vol}$ for dual-slice for abdomen examination the $CTDI_W$ and DLP and $CTDI_{vol}$ shown in Table 4.1 the $CTDI_W$ less than the other studies in terms of DLP , $CTDI_{vol}$. Chest examination the $CTDI_W$ and DLP and $CTDI_{vol}$ is less than the reviewed study. Table 4.3 show the data for 64-slice , for abdomen examination the $CTDI_W$ and DLP and $CTDI_{vol}$ is greater than review study. For chest examination the $CTDI_W$ and DLP and $CTDI_{vol}$ is greater than the review study. For brain examination the $CTDI_W$ and DLP and $CTDI_{vol}$ is greater than the previous studies. The mean dose values for

CT brain in16-slice were; DLP 670±380mGy.cm, $CTDI_{vol}$ 45.4±24.8mGy, while in dual-slice the average dose values were; DLP is 331.1±161mGy.cm, $CTDI_{vol}$ 55.7±9.3mGy. While the average dose in 64-slice was; DLP is 1160.8±435.7mGy.cm, $CTDI_{vol}$ 73.4±21.1mGy. For CT abdomen the averaged dose values in 16-slice were; DLP 270.6±100mGy.cm, $CTDI_{vol}$ 5.8±2mGy, while in dual-slice the average dose values were; DLP is 164±56mGy.cm, $CTDI_{vol}$ 4.4±1.3mGy. While in 64-slice the average dose values were; DLP is 1444±954.3mGy.cm, $CTDI_{vol}$ 75.7±93.9mGy. For CT Chest the averaged dose values in 16-slice were; DLP 209.6±83.3mGy.cm, $CTDI_{vol}$ 5.6±2mGy, while in dual-slice the average dose values were; DLP is 238.8±116.2mGy.cm, $CTDI_{vol}$ 8.7±4.6mGy. While in64-slice the average dose values were; DLP is 1121.2±872.1mGy.cm, $CTDI_{vol}$ 25.2±15mGy. Dual slice scanner delivered the least radiation dose while 16 and 64 slice scanners delivered the highest radiation dose. CT dose optimisation protocol is not implemented in all departments.

5.2 Conclusions:

In this study the radiation dose is higher in 64-slice than the 16-slice is higher than the dual-slice . Radiation dose from CT procedures varies from patient to patient. A particular radiation dose will depend on the size of the body part examined, the type of procedure, and the type of CT equipment and its operation. Typical values cited for radiation dose should be considered as measure and estimates that cannot be precisely associated with any individual patient, examination, or type of CT system. The actual dose from a procedure could be two or three times larger or smaller than the estimates. The assessment of radiation dose to pediatric patient undergoing CT brain , abdomen and chest investigated. In this study variation in doses were observed. The main contributor for this high dose was the use for different techniques, which justify the important of use radiation dose optimization technique and technologists training. Dual slice scanner delivered the least radiation dose while 16 and 64 slice scanners delivered the highest radiation dose. CT dose optimization protocol is not implemented in all departments.

5.3 Recommendations:

- Continuous education is highly recommended specially for 64 slice CT machines.
- Requests for CT scanning must be generated only by qualified medical practitioners and justified by both the referring doctor and the radiologist.
- Further studies should be done in order to optimize the radiation dose to establish national diagnostic reference level in Khartoum State.

5.4 Suggestion for future studies:

Future studies should be done in order to optimize the radiation dose to establish national diagnostic reference level in Sudan.

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Appendix:

WORKSHEETS

Estimation of Radiation Hazards of Computed Tomography Dose in Khartoum State

-

Subject No……………………….. Gender…………..Male/Female……………….Hospital…………

Age……………………………….. Weight………………………………….Height…………………………..

Indication…………………………………………………………………………………………………….

Interested organ……………………………………………………………………………………………

Diagnosis……………………………………………………………………………………………………

1. Phase	Value	10. FOV (mm)	Value

2. Tube Potential (kVp)		11. Start anatomic Level End anatomic Level	
3. Milliampere (mA) per rotation		12. Start Couch Level (mm) End Couch Level (mm)	
4. Time for one rotation (s) Or mAs per rotation		Original protocol (Routine) or changes was made	
5. Slice thickness (mm)		**Dosimetry**	
6. Slice Beam collimation (total detector number (N) and beam collimation (h))		Displayed CTDIw	
7. Table feed (T mm)		DLP	
8. Pitch		$CTDI_{Vol}$	
9. Total Slice number			

Machine Information Sheet:-

CT Machine Manufacture…………………………………………………….

Model…………………………………………………………………….

Year of Installation………………………………………………………

Focal Axial Distance (FAD)…………………………………………………

Detector Type…………………………………………………………….

Dose Display: Yes [ ], No [ ].