**Sudan University of Science and Technology**

**College of Graduate Studies and Scientific Research**

# Assessment of Radiation Dose for Patients Undergoing Brain and Abdomen Computed Tomography

**تقييم جرعة الاشعة للمريض اثناء فحوصات الدماغ والبطن بالاشعة المقطعية المحوسبة**

*A thesis submitted in partial fulfillment for the requirements*

*Of Master degree in Medical Physics*

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2016

# الآيـة

هُوَ الَّذي جَعَلَ الشَّمْسَ ضِيَاءً وَالقَمَرَ نُوراً وَقَدَّرَهُ
مَنَازِلَ لِتَعْلَمُوا عَدَدَ السِنِينَ وَالحِسَاب مَا خَلَقَ الله ذلكَ إِلاَّ
بِالحقِّ يُفصِّلُ الأياتِ لِقَومٍ يَعْلَمُون" (آية 5)يونس

# Dedication

*To doses of the cup blank to give me a drop of love*

*To those of the fingers to give us a moment of happiness*

*To reap the thorns out of my way for me to pave the way science*

*To heart the great my father.*

*Of whom breastfed me to love and compassion*

*Into a symbol of love and healing balm*

*To the heart as pure whitenessmy parents*

*To my friends, And to all my family*

# Acknowledgements

*First, I would like to express my deepest gratitude to Dr.* ***Ahmed Abukonna****for his support and guidance. Without his help this work could not have been accomplished.*

*I also would like to thank Dr.* ***MOHAMMED ABD ALWHAB.***

*Deep thanks to my friend Finally, I would like to sincerely thank my family for their consistent mental support*

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Abstract

Computed tomography (CT) is an imaging technique which produces a digital topographic image from diagnostic x-ray. It 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.

The aim of this study was to evaluate the level of radiation dose received by the patients during brain and abdomen CT examination. In this study, a total of 128 adult patients undergoing brain and the abdominal CT scanning exams were evaluatedusing CT Dose index and dose length product (DLP)

The result of this study revealed that the mean effective dose for abdomen in hospital (1) and hospital (2) was(64.31 ± 29.8)mSv and (71.61 ± 0.97) mSvrespectively. The mean effective dose for brain in hospital (1) and hospital (2) was (2.96 ± 0.97)mSv, (3.11±0.51)mSv respectively. These values were found to be at standard dose reference level.

Unjustified screening the Abdomen and head should thus be banished. Such policy is unacceptable in young patients who are at a low risk of having an incidental associated disease. Similarity, repeated acquisition should not be performed in circumstances where they do not specifically yield additional information.

**ملخص البحث:**

التصوير المقطعي (CT)، هو عملية التصويربالأشعة السينية التي تنتج صور لمقاطع جسم المريضبجودة عالية، وبالمقارنة مع التشخيص الإشعاعية الأخرى، الاشعه المقطعية ذات جرعات اكبر للمرضى.

هنالك ادراك متزايد بان جودة الصورة في التصوير المقطعي غالبا ما تتجاوز المستوى المسموح للتشخيص وجرعات المرضى هي اعلي من اللازم.

كان الهدف من هذه الدراسة هو تقييم مستوى الجرعة الإشعاعية من قبل المرضى التي وردت خلال فحصي الدماغ والبطن بالتصوير المقطعي . في هذه الدراسة تم تقييم مجموعه 128 مريضا بالغا للدماغ والبطن باستخدام التصوير المقطعي الجرعة مؤشر(CTDI) وجرعة المنتج طول (DLP). كشفت نتائج هذه الدراسة أن الجرعة الفعالة المتوسطة للفحص البطن في المستشفى (1) ومستشفى (2) كان (64.31 ± 29.8) ملي سيفرت و (71.61 ± 0.97) ملي سيفرت على التوالي. كان متوسط الجرعة الفعالة للدماغ في المستشفى (1) ومستشفى(2) (2.96 ± 0.97) ملي سيفرت (3.11 ± 0.51) ملي سيفرت على التوالي. تم العثور على هذه القيم لتكون على مستوى الجرعة المرجعية القياسية. فحص البطن والدماغ غير المبررة ينبغي نفيها. لا يجب اتباع نفس السياسات في المرضى الصغار الذين هم في خطر منخفض من وجود مرض يرتبط به عرضيا. تكرار الفحص لا يجب ان يؤدى في الظروف التي لا تسفر عن تحديد الحصول على معلومات إضافية.

# Chapter one

# Introduction

## 1.1 Introduction

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

Computed Tomography (CT) was invented by a British engineer, Sir Godfrey Hounsfield who also won the Nobel Prize because of Jhis invention. CT was first introduced in the clinical practice in 1972 which was only limited to the brain scan. Prior to that, X-ray planar radiography and fluoroscopy systems were the main contributors of radiation in imaging (Goldman 2007).

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

CT is one of the many technologies that were made possible by the invention of the computer. The clinical potential of CT became obvious during its early clinical use, and the excitement forever solidified the role of computers in medical imaging. Recent advances in acquisition geometry, detector technology, multiple detector arrays, and x-ray tube design have led to scan times now measured in fractions of a second. Modern computers deliver computational power that allows reconstruction of the image data essentially in real time (Jerrold T, J.antony, Edwin, Boone,2002).

CT has fascinated the world with production of high contrast resolution images for visualizing soft tissues and the ability of producing tomographic and three dimensional (3D) volumetric images (IAEA 2007). Thus, it has changed the perception on medical diagnostic quality and as a result it has improved the quality of healthcare. Now, CT is becoming a common diagnostic tool in many major hospitals in the whole world. It is obvious that CT gives a lot of advantages such as faster scanning procedure, good spatial resolution and good contrast, compared to other modalities. Nowadays, many medical centers choose to send cases like accident and emergency cases, urology, cardiac imaging and pediatric imaging for CT scan as their first option for easy diagnosis of the symptoms. In some countries, sinusitis cases were likely referred to CT compared to the plain radiograph because CT were able to show important structures (Zammit-Maepelet al.2003). Having taken notice of that, the manufacturers are also intense in introducing the latest technologies and applications of their CT due to the high demand of the CT scanners. This can be seen in Figure 1.3 where the number of CT scanners installed in Germany has increased linearly (Nagel 2000). Drastic increase happened after 1990 when the helical CT was introduced to the market.

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

The individual risk from radiation associated with a CT scan is quite small compared to the benefits that accurate diagnoses and treatment can proved. Still, unnecessary radiation exposure during medical procedures should be avoided. Unnecessary radiation may be delivered when CT scanner parameters are not appropriate adjusted for the patient size (Anne et al. 2001).

There is no doubt that many patient have benefited from the rapid diagnoses made possible by CT and from its value for monitoring chronic disease. However is increasing concern regarding the risk of this exposure to radiation. It is well established that radiation can be harmful and has both deterministic and stochastic effects. Deterministic effect, such as hair loss, skin burns, and cell death, are dose dependent but do not occur below a threshold of 150-200 mSv.Since the typical estimated dose associated with proper use of CT is in the range of 2-10 mSv, deterministic effects are not normally a concern. Induction of cancer by radiation is a probabilistic (stochastic) effect, not a deterministic effect. That is, higher radiation doses are associated with a higher likelihood of carcinogenesis, but even low doses of radiation could potentially induce carcinogenesis and it is more difficult to assess a safe level of exposure. (Roundset al.2003)

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 et al.2010)

In conventional X-ray procedure, medical personnel can tell if the patient has been overexposed because of the film is overexposed, produce a dark image (ICRP 2006). However, with CT there is no obvious evidence that the patient has been overexposed because the quality of the image may not be compromised. Several recent articles (Kalender et al. 1999, Rehani M, Berry M 2000, Rehani M 2000)stress that is important to use the lowest radiation dose necessary to provide an image from which an accurate diagnosis can be made, and that signification dose reduction can be achieved without compromising clinical efficacy.

The United Nation Scientific Committee on the Effects of Atomic Radiation (UNSCEAR, 2000) has highlighted that the worldwide there about 93 million CT examination performed annually at a rate of about 57 examination per 1000 persons. UNSCEAR also estimated that CT constitutes about 5% of all X-ray examination worldwide will accounting for about 34% of the resultant collective dose. In the countries that were identified as having the highest levels of healthcare, the corresponding figures were 6% and 41% respectively. New advancement of the CT has also led to great increase of the radiation dose to thepatients. The use of multi-slice computed tomography (MSCT) has aggravated the scenario with the increasing of collective dose of CT examinations because the MSCT produces higher dose to the patients compared to single slice CT (SSCT) (Hunold et al. 2003).

## 1.2 Problem of study

Most clinical in Sudan cannot applied diagnostic reference level of does, use different exposure factor and not found the base line of practice. Most of these clinics only take in account the image quality without taking care about patient dose spatially the radiation exposure here is closer to sensitive tissues and in most workers there is a limitation in knowledge of the harm effects of ionizing radiation and the lack of knowledge of the persons working with radiation about the basic principles of radiation protection and physical variables that control the dose and distribution within the patient to give a high quality image in order to minimize the patient dose and increase the image quality and all clinical in Sudan use same protocol(Kv,mAs, pitch) to children and adults patient without take age and body weight in consideration.

### 1.3 Objective:

**General Objective:**

To evaluate the level of radiation dose received by the patients during brain and abdomen CT examination.

**Specific Objective**:

1. Quantify the patient dose in CT examinations.
2. Estimate the patient effective dose.
3. To identify the variation of radiation dose in different centers.
4. To compare the result with the local diagnostic reference level (DRL).

### 1.4 Thesis outline:

This thesis is concerned with the assessment of radiation dose for patients during CT examinations for different CT Modalities:

Accordingly, it is divided into the following chapters:

Chapter one is the introduction to this thesis. This chapter presents the historical background and radiation risks, in addition to study problem, objectives and scope of the work. It also provides an outlines of the thesis.

Chapter two contains the background material for the thesis. This chapter also includes a summary previous work performed in this field.

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

Chapter four presents the results of this study.

Chapter five presents the discussion, conclusion and recommendations of this thesis and presents the suggestions for future work.

# Chapter two

# Theoretical Background

## 2.1 CT Machine

### 2.1.1The CT scanner components

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

1- The Data Acquisition and Transfer system, which encompasses the gantry, the patient's table, the power distribution unit and the data transfer unit.( Nunes,2010)

- The Gantry which is a central opening gantry is a moveable frame that contains the x-ray tube including collimators and filters, detectors, data acquisition system (DAS), rotational components including slip ring systems and all associated electronics such as gantry angulations motors and positioning laser lights. A CT gantry can be angled up to 30 degrees toward a forward or backward position.
- The Table is where the patients is positioned (lie down), and it moves through the gantry. The patient's table and the gantry constitute CT scanner itself.
- The power Distribution unit supplies power to the gantry, the patient's table and the computers of the Computing System, which is localized in a separate room as will be explained next.

2- The computing System(or operator's console) is installed in separate room, making it possible for the operator (technician)to control the acquisition process, introducing patient data and selecting several acquisition parameters such as the kVp , mA values the protocol is going to use (Nunes, 2010). Also there is another operator's console for editing and post-processing is also necessary, so it possible to analyze and review previous exam data, without interfering with the current examinations taking place.

**Figure (2.1) CT scanner**

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

## 2.2 CT Generations

### 2.2.1 First-Generation CT Scanners

The EMI Mark I scanner, the first commercial scanner invented by Hounsfield, was introduced in 1973 (Hounsfield GN., 1973). This scanner acquired data with an x-ray beam collimated to a narrow “pencil” beam directed to a single detector on the other side of the patient; the detector and the beam were aligned in a scanning frame. A single projection was acquired by moving the tube and detector in a straight-line motion (translation) on opposite sides of the patient (Fig 2.2). To acquire the next projection, the frame rotated 1°, then translated in the other direction. This process of translation and rotation was repeated until 180 projections were obtained. The earliest versions required about 4.5 minutes for a single scan and thus were restricted to regions where patient motion could be controlled (the head). Since procedures consisted of a series of scans, procedure time was reduced somewhat by using two detectors so that two parallel sections were acquired in one scan. Although the contrast resolution of internal structures was unprecedented, images had poor spatial resolution (on the order of 3 mm for a field of view of 25 cm and 80 × 80 matrix) and very poor z-axis resolution (□13-mm section thickness).

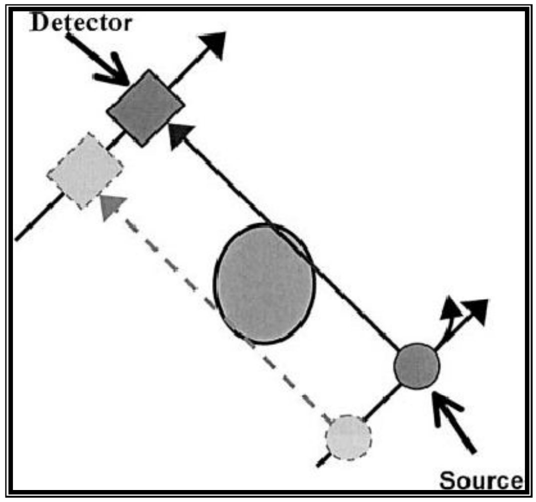
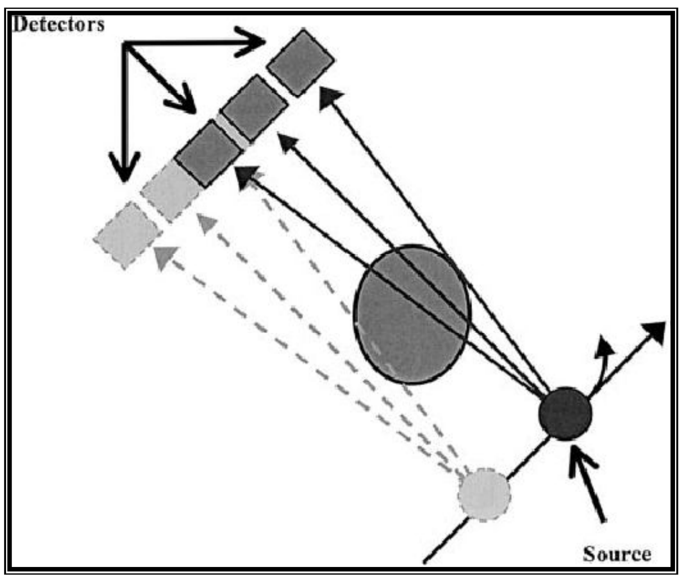



Fig (2.2): Diagram of the first-generation CT scanner, which used a parallel x-ray beam with translate-rotate motion to acquire data. From (Mahesh, 2002).

### 2.2.2 Second-Generation CT Scanners

The main impetus for improvement was in reducing scan time ultimately to the point that regions in the trunk could be imaged. By adding detectors angularly displaced, several projections could be obtained in a single translation. For example, one early design used three detectors each displaced by 1°. Since each detector viewed the x-ray tube at a different angle, a single translation produced three projections. Hence, the system could rotate 3° to the next projection rather than 1° and had to make only 60 translations instead of 180 to acquire a complete section (Fig 2.3). Scan times were reduced by a factor of three. Designs of this type had up to 53 detectors, were ultimately fast enough (tens of seconds) to permit acquisition during a single breath hold, and thus were the first designs to permit scans of the trunk of the body. Because rotating anode tubes could not withstand the wear and tear of rotate-translate motion, this early design required a relatively low output stationary anode x-ray tube. The power limits of stationary anodes for efficient heat dissipation were improved somewhat with the use of asymmetrical focal spots (smaller in the scan plane than in the z-axis direction), but this resulted in higher radiation doses due to poor beam restriction to the scan plane. Nevertheless, these scanners required slower scan speeds to obtain adequate x-ray flux at the detectors when scanning thicker patients or body parts.




Fig(2.3): Diagram of the second-generation CT scanner, which used translate-rotate motion to acquire data

### 2.2.3 Third-Generation CT Scanners

Designers realized that if a pure rotational scanning motion could be used, then it would be possible to use higher-power, rotating anode x-ray tubes and thus improves scan speeds in thicker body parts. One of the first designs to do so was the so-called third generation or rotate-rotate geometry. In these scanners, the x-ray tube is collimated to a wide, fan-shaped x-ray beam and directed toward an arc-shaped row of detectors. During scanning, the tube and detector array rotate around the patient (Fig 2.4), and different projections are obtained during rotation by pulsing the x-ray source or by sampling the detectors at a very high rate. The number of detectors varied from 300 in early versions to over 700 in modern scanners. Since the slam-bang translational motion was replaced with smooth rotational motion, higher-output rotating anode x-ray tubes could be used, greatly reducing scan times. One aspect of this geometry is that rays in a single projection are divergent rather than parallel to each other, as in earlier designs. Beam divergence required some modification of reconstruction algorithms, and sampling considerations required scanning an additional arc of one fan angle beyond 180°, although most scanners rotate 360° for each scan. Nearly all current helical scanners are based on modifications of rotate-rotate designs. Typical scan times are on the order of a few seconds or less, and recent versions are capable of sub second scan times.

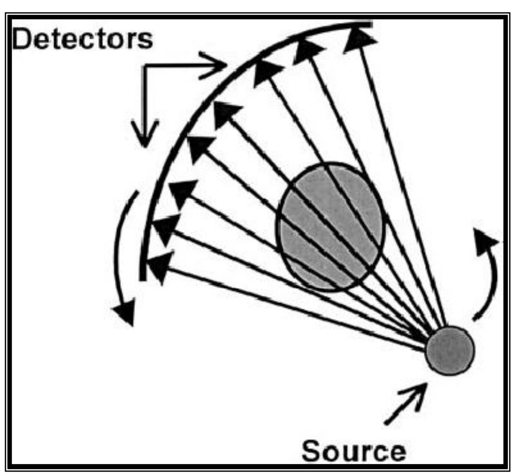


Fig (2.4): Diagram of the third-generation CT scanner, which acquires data by rotating both the x-ray source with wide fan beam geometry and the detectors around the patient. Hence, the geometry is called rotate-rotate motion.

### 2.2.4: Fourth-Generation CT Scanners:

This design evolved nearly simultaneously with third-generation scanners and also eliminated translate-rotate motion. In this case, only the source rotates within a stationary ring of detectors (Fig 2.5). The x-ray tube is positioned to rotate about the patient within the space between the patient and the detector ring. One clever version, which is no longer produced, moved the x-ray tube out of the detector ring and tilted the ring out of the x-ray beam in a wobbling (nutation) motion as the tube rotated. This design permitted a smaller detector ring with fewer detectors for a similar level of performance. Early fourth-generation scanners had some 600 detectors and later versions had up to 4,800. Within the same period, scan times of fourth-generation designs were comparable with those of third-generation scanners. One limitation of fourth-generation designs is less efficient use of detectors, since less than one-fourth are used at any point during scanning. These scanners are also more susceptible to scatter artifacts than third-generation types, since they cannot use anti scatter collimators. CT scanners of this design are no longer commercially available except for special-purpose applications.(Mahesh, 2002)

Until around 1990, CT technology had evolved to deliver scan plane resolutions of 1–2 lp/mm, but z-axis resolution remained poor and interscan delay was problematic due to the stop-start action necessary for table translation and for cable unwinding, which resulted in longer examination times. The z-axis resolution was limited by the choice of section thickness, which ranged from 1 to 10 mm. For thicker sections, the partial volume averaging between different tissues led to partial volume artifacts. These artifacts were reduced to some extent by scanning thinner sections. In addition, even though it was possible to obtain 3D images by stacking thin sections, inaccuracy dominated due to involuntary motion from scan to scan. A typical 3D reconstruction of this era is shown in Figure 2.7 the step like contours could be minimized by overlapping of CT sections at the expense of a significant increase in radiation to the patient. Also, the conventional method of section-by-section acquisition produced misregistration of lesions between sections due to involuntary motion of anatomy in subsequent breath holds between scans. It was soon realized that if multiple sections could be acquired in a single breath hold, a considerable improvement in the ability to image structures in regions susceptible to physiologic motion could result. However, this required some technological advances, which led to the development of helical CT scanners. (Mahesh, 2002)

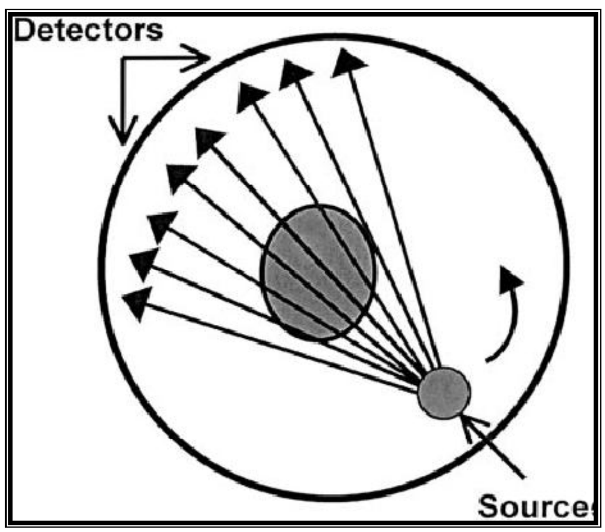



Fig (2.5): Diagram of the fourth-generation CT scanner, which uses a stationary ring of detectors positioned around the patient. Only the x-ray source rotates with wide fan beam geometry, while the detectors are stationary. Hence, the geometry is called rotate-stationary motion. From (Mahesh, 2002)

## 2.3 Principles of Helical CT Scanners

The development of helical or spiral CT around 1990 was a truly revolutionary advancement in CT scanning that finally allowed true 3D image acquisition within a single breath hold. The technique involves the continuous acquisition of projection data through a 3D volume of tissue by continuous rotation of the x-ray tube and detectors and simultaneous translation of the patient through the gantry opening (Fig 2.6) (Kalender, et al, 1990). Three technological developments were required: slip-ring gantry designs, very high power x-ray tubes, and interpolation algorithms to handle the non-coplanar projection data (Beck, 1996).

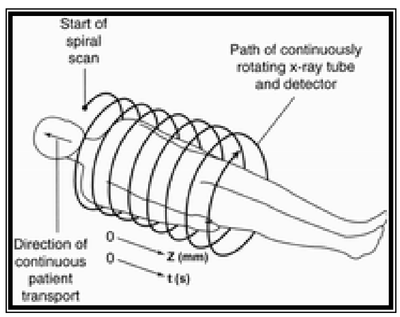



Fig (2.6): Principles of helical CT. As the patient is transported through the gantry, the x-ray tube traces a spiral or helical path around the patient, acquiring data as it rotates. t = time in seconds.(Mahesh, 2002).

## 2.4 Slip-Ring Technology

Slip rings are electromechanical devices consisting of circular electrical conductive rings and brushes that transmit electrical energy across a moving interface. All power and control signals from the stationary parts of the scanner system are communicated to the rotating frame through the slip ring. The slip-ring design consists of sets of parallel conductive rings concentric to the gantry axis that connect to the tube, detectors, and control circuits by sliding contactors (Fig 2.7). These sliding contactors allow the scan frame to rotate continuously with no need to stop between rotations to rewind system cables (Brunnett, et al., 1994). This engineering advancement resulted initially from a desire to reduce interscan delay and improve throughput. However, reduced interscan delay increased the thermal demands on the x-ray tube; hence, tubes with much higher thermal capacities were required to withstand continuous operation over multiple rotations. (Mahesh, 2002)

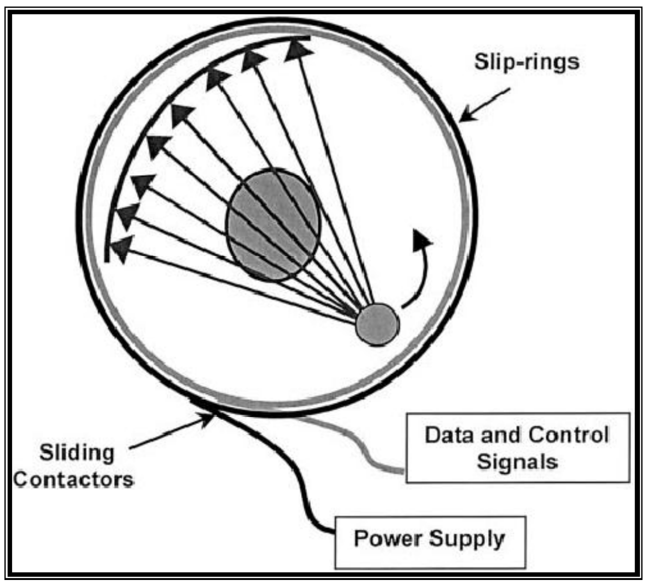



Fig (2.7): Diagram of the slip-ring configuration. Sliding contactors permit continuous rotation of the x-ray tube and detectors while maintaining electrical contact with stationary components.

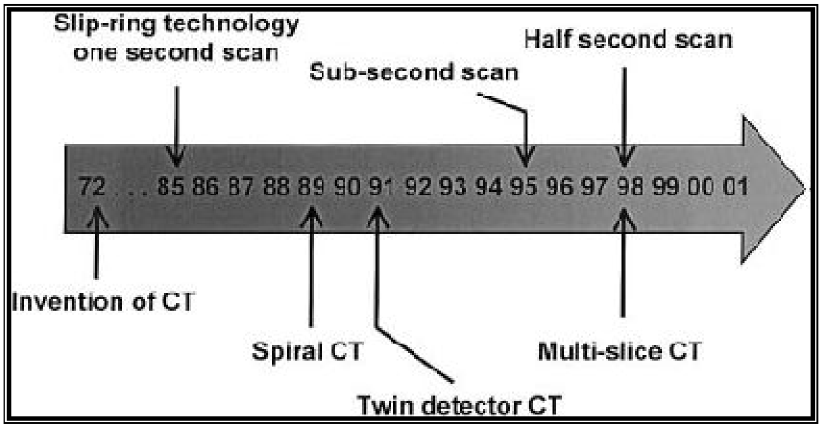



Fig (2.8): Time line of the key technological developments in CT. From (Mahesh,2002).

### 2.5 Capabilities of Single-Row Detector Helical CT

With the advent of helical CT, considerable progress was made on the road toward 3D radiography. An example of a 3D reconstruction from single-row detector helical scanning is shown in Fig (2.9).Complete organs could be scanned in about 30–40 seconds; artifacts due to patient motion and tissue misregistration due to involuntary motion were virtually eliminated. It became possible to generate sections in any arbitrary plane through the scanned volume. Significant improvements in z-axis resolution were achieved due to improved sampling, since sections could be reconstructed at fine intervals less than the section width along the z axis. Near-isotropic resolution could be obtained with the thinnest (□1 mm) section widths at a pitch of 1, but this could be done only over relatively short lengths due to tube and breath-hold limitations (Kalender 1995), (Levy, 1995). Higher-power tubes capable of longer continuous operation coupled with faster rotation speeds could scan greater lengths with higher resolution. The practical limit on such brute force approaches, however, became the length of time a sick patient could reliably suspend breathing. This turns out to be no more than 30 seconds. Even though the z-axis resolution for helical CT images far exceeds that of conventional CT images, the type of interpolation algorithm and the pitch still affect the overall image quality. The section sensitivity profiles of helical CT images are different compared with those of conventional CT images, which are influenced by the type of interpolation algorithm and the selected pitch.

## 2.6 Multiple-Row Detector Helical CT

Continued scanner development on the road to a 3D radiograph called for further progress, but single-row detector helical scanners had reached their limits. An obvious improvement would be to make more efficient use of the x rays that are produced by the tube while improving z-axis spatial resolution; this led to the development of multiple-row detector arrays. The principal difference between single- and multiple-row detector helical scanners is illustrated in Figure (2.9). The basic idea actually dates to the very first EMI Mark I scanner, which had two parallel detectors and acquired two sections simultaneously.

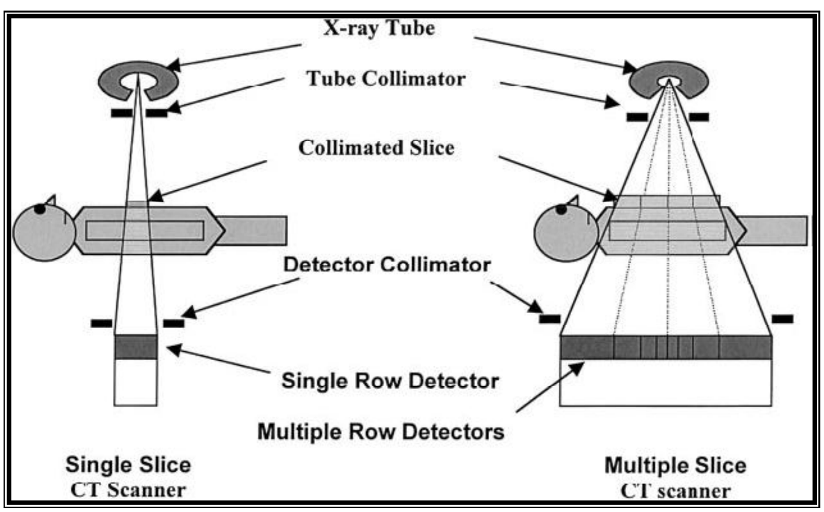



Fig (2.9): Diagram shows the difference between single-row detector and multiple-row detector CT designs. The multiple-row detector array shown is asymmetrical and represents that of one particular manufacturer.

The first helical scanner to use this idea, the CT Twin was launched in 1992. (Mahesh, 2002).This design was so superior to single-row detector designs that all scanner manufacturers went back to the drawing board. By late 1998, all major CT manufacturers launched multiple-row detector CT scanners capable of acquiring at least four sections per rotation. The arrangement of detectors along the z axis and the widths of the available sections vary between the systems. Fig (2.10) illustrates different multiple-row detector array configurations from several manufacturers.

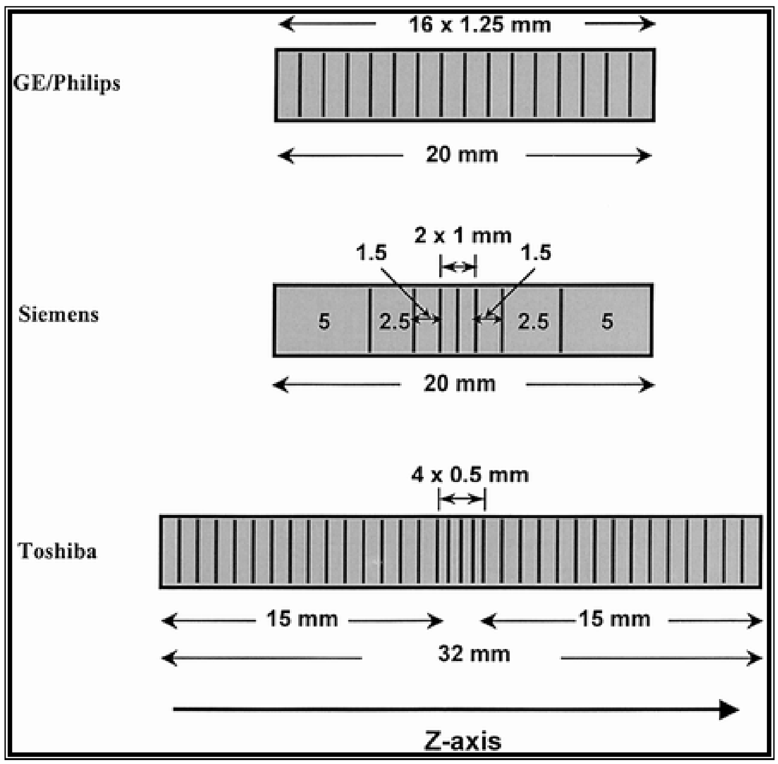



Fig (2-10): Various detector array designs used in multiple-row detector CT scanners.

In single-row detector helical CT designs, scan volume can be increased with an increased pitch at the expense of poorer z-axis resolution, whereas z-axis resolution can be preserved in multiple-row detector designs. For example, if a 10-mm collimation were divided into four 2.5-mm detectors, the same scan length could be obtained in the same time but with a z-axis resolution improved from 10 mm to 2.5 mm. In another example, a multiple-row detector scanner with four 5-mm detectors and a beam width of 20 mm reduces the scan time by a factor of 4–15 seconds for the same z-axis resolution (Mahesh, 2002). By increasing the number of CT scanner detector rows, data acquisition capability dramatically increases while greatly improving the efficiency of x-ray tubes. Further developments in scanner rotational speeds and tube outputs have made isotropic resolution a practical possibility with even better improvements on the horizon. Current multiple-row detector scanners can scan large 40-cm volume lengths in less than 30 seconds with near-isotropic resolution and image quality that could not be envisioned at the time of Hounsfield's invention.

MDCT systems are CT scanners with a detector array consisting of more than a single row of detectors. The "multi-detector-row" nature of MDCT scanners refers to the use of multiple detector arrays (rows) in the longitudinal direction (that is, along the length of the patient lying on the patient table). MDCT scanners utilize third generation CT geometry in which the arc of detectors and the x-ray tube rotate together. All MDCT scanners use a slip-ring gantry, allowing helical acquisition at rotation speeds as fast as 0.33 second for a full rotation of 360 degrees of the X-ray tube around the patient. A scanner with two rows of detectors (Mahesh, 2002) had already been on the market since 1992 and MDCT scanners with four detector rows were introduced in 1998 by several manufacturers. The primary advantage of these scanners is the ability to scan more than one slice simultaneously and hence more efficiently use the radiation delivered from the X-ray tube (Fig.2.11). The time required to scan a certain volume could thus be reduced considerably.

The number of slices, or data channels, acquired per axial rotation continues to increase, with 64-detector systems now common (Flohr et al., 2005a;

Flohr et al., 2005b). It is likely that in the coming years even larger arrays of detectors having longitudinal coverage per rotation > 4 cm will be commercially available. Preliminary results from a 256-detector scanner (12.8 cm longitudinal coverage at the center of rotation) have already been published (Mori et al., 2004). Further, an MDCT system with two x-ray sources is now commercially available, signaling continued evolution of CT technology and applications (Flohr et al., 2006).

MDCT scanners can also be used to cover a specific anatomic volume with thinner slices. This considerably improves the spatial resolution in the longitudinal direction without the drawback of extended scan times. Improved resolution in the longitudinal direction is of great value in multiplanar reformatting (MPR, perpendicular or oblique to the trans axial plane) and in 3-dimensional (3D) representations. Spiral scanning is the most common scan acquisition mode in MDCT, since the total scan time can be reduced most efficiently by continuous data acquisition and overlapping data sets and this allows improved multi-planar reconstruction (MPR) and 3D image quality to be reconstructed without additional radiation dose to the patient.

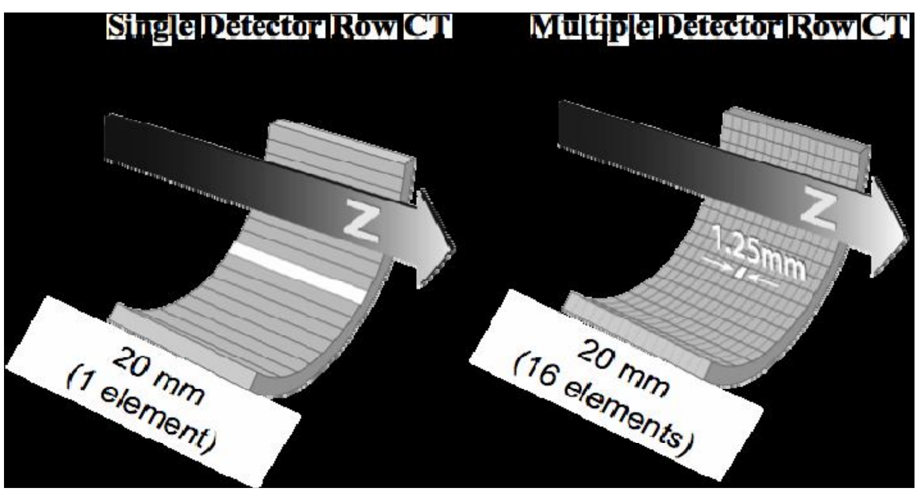



Fig (2.11): single CT detector versus Multi slice CT detector. From (ICRP 32/219,2006).

## 2.7 CT imaging protocol

The technique used in CT-scanners share most of its characteristics with conventional X-ray imaging, and the prime differences are seen in projection, detection and acquisition as presented in Figure 2.12 below.

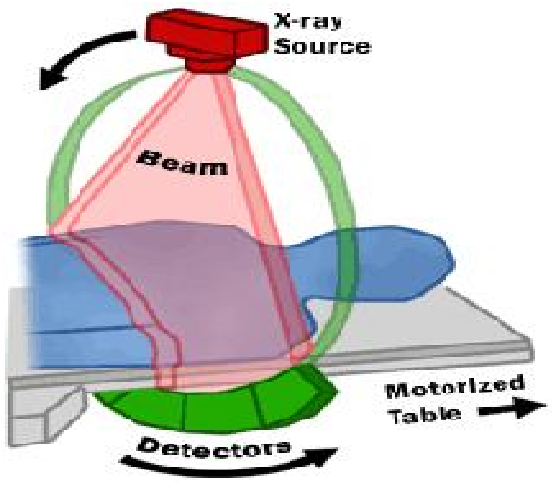



Figure (2.12) Simple overview of a third generation CT-imaging system

### 2.7.1 Parameters

In order to properly calculate and compare doses, it is imperative to have a standardized nomenclature to ensure that all data is comparative (Kalra, M. K ,et al 2006). Without this, it will be difficult toreproduce measurements, and to develop consistent protocols. When performing a CT examination, a number of parameters are defined by the operator. The thesis will cover the parameters deemed important for correct, uniform dosimetry: tube current, tube voltage, rotation time, total scan length, slice thickness and pitch. Automatic exposure control (AEC) and iterative reconstruction will be briefly covered, as their impact on dose and image quality is more of a qualitative influence than a quantitative one.

#### 2.7.1.1 Tube current

The tube current [mA] influences the number of photons exiting the X-ray tube, as it determines the number of electrons leaving the cathode. The tube current is directly proportional to radiation dose, and as such is a prime parameter in adjusting the dose. Instead of tube current is sometimes used the tube-current-time-product [mAs], which is the tube current multiplied with the scan time.

#### 2.7.1.2 Tube Voltage

The tube voltage [kV] determines the voltage across the anode and cathode of the X-ray tube, and therefore the acceleration of the electrodes across the interior vacuum. This determines the kinetic energy of the electrodes when they reach the anode, and therefore the number of interactions they can initiate before being absorbed. As a consequence, an increase in tube voltage will increase the dose, all other factors kept constant; however, the increase is not directly proportional as was the case with current. Voltage determines the energy of the electrons, and therefore the energy distribution of the incident X-rays. It is rarely adjusted from the customary value of 120 kV. Certain examinations use a different voltage, but seldom outside the range of 80 to 140 kV (Kalra, M. K ,et al 2006).

#### 2.7.1.3 Rotation Time

The rotation time of the gantry [s] has decreased greatly over the last few decades, with modern scanners having a rotation time in the area of 0.4 seconds. The main consequence of the decreased rotation time is an increase in the noise and a reduction in absorbed dose. To avoid the noise, it is customary to increase the tube current accordingly (M. K., Maher,et al 2004).

#### 2.7.1.4 Total Scan Length

It is apparent that the total scan length [cm] influence the absorbed dose, as an increase in scan length will expose a larger part of the patient to radiation. Therefore, it is imperative that scan length is to be limited to cover just the diagnostically relevant part of the patient; otherwise, an unnecessary increase in dose will be seen (ICRP, 2000). This is relatively easy with SSCT; however, the situation is more complicated for MSCT. At the initiation of the scan, the X-ray tube will be activated the moment the first row of detectors reach the diagnostic area. The X-ray beam will irradiate the entire detector-array, but only the first row of detectors will be acquiring image data. The remaining detector rows will not acquire data, but the area will still be irradiated. This is called over scan, and a small degree of over scan is required for correct reconstruction. As the table moves, more rows of detectors are entering the diagnostic area, contributing to the image. At the reverse end of the patient, the same scenario occurs, and a noteworthy part of the dose is absorbed in the patient outside the diagnostic area (M. K., Maher,et al 2004).

#### 2.7.1.5 Slice Thickness

In SSCT, with only a single row of detectors, the slice thickness [cm] is determined by simple collimation. The maximum slice thickness is limited by the width of the individual detector element (typically 10 mm (M. K., Maher,et al 2004)), and by collimating the beam, this thickness can be decreased. In other words, the width of the beam is equal to slice thickness. In MSCT, the width of each individual detector element in the longitudinal direction determines the minimum slice thickness, and by merging multiple adjacent detector elements during detection, one can increase the slice thickness. This has a significant impact on image quality, as thin slices have better spatial resolution compared to thick slices, but lower SNR. To address the decrease in SNR, it is necessary to increase for instance the tube current, resulting in a significant increase in dose to the patient (Kalender,et al,2005).

### 2.7.1.6 Pitch

With the prevalence of helical MSCT, it is necessary to incorporate the incremental movement of the table, in relation to the irradiated area. This is defined as pitch, being the increment of the table per rotation, divided by the width of the beam. In Figure 2.13 below, a 4-slice MSCT is rotated twice around the patient, resulting in the acquisition of eight slices in pairs of two (indicated by color). The slices are in reality at an incline, as the patient is moving during exposure.

pitch = table feed per rotation/collimation

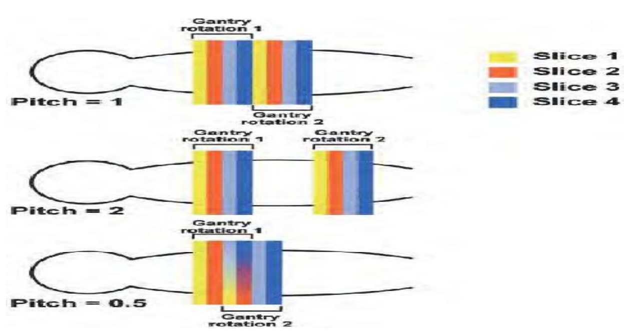



Figure (2.13) the effect of pitch on irradiated area, with a overlap for pitch < 1

(23

#### 2.7.1.7 Automatic Exposure Control

Technological advances lead to the development of a technique where the tube current is modulated in real-time, in order to minimize the dose while retaining image quality. This technique, Automatic Exposure Control(AEC) varies the tube current during exposure. The variance is relative to patient thickness, optimized to achieve dose distribution defined by a desirable image quality. It is possible to achieve a significant reduction in dose based on which type of AEC is used: either the exposure varies within a single slice, i.e. in the image plane of the slice, or it is modulated in the longitudinal direction of the patient. It is also possible to combine these two types of AEC.

### 2.7.2 Protocols

Brain and abdomen.All of the above parameters are defined in protocols, which are a basic set of parameters that are hereafter modified in order to accommodate the individual patient. Protocols serve as basic guidelines for a specific examination on a specific scanner.

## 2.8 CT Dose equivalent and unit

### 2.8.1 Radiation dose units

The specific units of measurement for radiation dose commonly referred to as effective dose (mSv). Other radiation dose measurement units include; Rad, Rem, Rontgen, and Sievert. Because different tissues and organs have varying in sensitivity to radiation exposure, the actual effective dose to different parts of the body for X-ray procedure varies. The term effective dose is used when referring to the dose averaged over the entire body. The effective dose accounts for the relative sensitivities of different tissues exposed. More importantly, it allows for qualification of risk and comparison to more familiar sources of exposure that range from natural background radiation to radiographic medical procedure. As with other medical procedures, X-rays are safe when used with care. Radiologists and X-ray technologists have been trained to use the minimum amount of radiation that is necessary to obtain the needed results. The decision to have an X-ray examination is a medical one, based on the likelihood of benefit from the examination and the potential risk from radiation (ICRP 1990, ICRP 1991).

### 2.8.2 Effective dose

Effective dose is becoming a very useful radiation quantity for expressing relative risk to humans, both patients and other personnel. It is actually a simple and very logical concept. It takes into account the specific organs and areas of the body that are exposed. The point is that all parts of the body and organs are not equally sensitive to the possible adverse effects of radiation, such as cancer induction and mutations (Perry Sprawls.org, Online).

For the purpose of determining effective dose, the different areas and organs have been assigned tissue weighting factor (WT) values. For a specific organ or body area the effective dose is:

$$\text{Effective Dose (Gy)} = \text{Absorbed Dose (Gy)} \times WT \qquad (2.1)$$

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

It is generally assumed that the exposure to natural background radiation is somewhat uniformly distributed over the body. Since the tissue weighting factor for the total body has the value of one (1), the effective dose is equal to the absorbed dose. This is assumed to be 300 mrad in the illustration.

Let's look at an illustration. If the dose to the breast, MGD, is 300 mrad for two views, the effective dose is 45 mrad because the tissue weighting factor for the breast is 0.15.

What this means is that the radiation received from one mammography procedure is less than the typical background exposure for a period of two months.

**Table: 2.1** Tissue Weighting Factors (UNSCEAR 2008):

| Weighting factors for different organs | | | |
|---|---|---|---|
| Organs | Tissue weighting factors | | |
| | ICRP30(I36) 1979 | ICRP60(I3) 1991 | ICRP103(I6) 2008 |
| Gonads | 0.25 | 0.20 | 0.08 |
| Red Bone Marrow | 0.12 | 0.12 | 0.12 |
| Colon | - | 0.12 | 0.12 |
| Lung | 0.12 | 0.12 | 0.12 |
| Stomach | - | 0.12 | 0.12 |
| Breasts | 0.15 | 0.05 | 0.12 |
| Bladder | - | 0.05 | 0.04 |
| Liver | - | 0.05 | 0.04 |
| Oesophagus | - | 0.05 | 0.04 |
| Thyroid | 0.03 | 0.05 | 0.04 |
| Skin | - | 0.01 | 0.01 |

| | | | |
|---|---|---|---|
| Bone surface | 0.03 | 0.01 | 0.01 |
| Salivary glands | - | - | 0.01 |
| Brain | - | - | 0.01 |
| Remainder of body | 0.30 | 0.05 | 0.12 |

### 2.9 CT dose measurements

Although CT presents only a small percentage of radiology examinations, it results in a significant portion of the effective radiation dose from medical procedures; (I) with the increasing use of CT for screening procedures, (II) and advances in scanner technology, they tend for increasing numbers of procedures performed with this modality may increase. Although CT is clearly providing many clinical benefits, the motivation to understand radiation dose in general as well as the specific concepts related to CT grows with prevalence of this modality (ImPACT 2007, Jones et al. 1993).

#### 2.9.1 CT parameters that influence the radiation dose

The radiation exposure to the patients undergoing CT examinations is determined by two factors: equipment-related factors, .e. the design of the scanner with respect to dose efficiency, and applications-related factors, i.e. the way in which the radiologist and X-ray technologist makes use of the scanner (Nagel 2007). In this chapter the features and parameters influencing patient dose are outlined. First, however, a brief introduction on the dose descriptors applicable to CT is given (Nagel 2007).

#### 2.9.2 CT dose descriptors

The dose qualities used in this projection radiography are not applicable to CT for three reasons (ImPACT 2007, Jones et al. 1993):

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

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

Third, the situation with CT is further complicated by the circumstances in which-unlike in conventional projection radiography-the volume to be imaged is not irradiated simultaneously. This often leads to confusion about what dose from a complete series of e.g. 15 slices might be compared with the dose from a single slice (ImPACT 2007, Jones et al. 1993).

As a consequence, dedicated dose quantities that account for these peculiarities are needed. The 'Computed Tomography Dose Index (CTDI)', which is a measure of the local dose, and the Dose Length Product (DLP), representing the integral radiation exposure associated with a CT examination. Fortunately, a bridge exists that enables to compare CT with radiation exposure from the other modalities and sources; this can be achieved by the effective dose (E). So there are three dose descriptors in all, which everyone dealing with CT should be familiar with (Nagel 2007).

#### 2.9.2.1 Computed tomography dose index (CTDI)

The 'Computed Tomography Dose Index (CTDI)' is the fundamental CT dose descriptor. By making use of this quantity, the first two peculiarities of CT scanning are taken into account: The CTDI (unit: Milligray (mGy)) is derived from the dose distribution a long a line which is parallel to the axis of rotation for the scanner (=z axis) and which is recorded for a single rotation of X-ray source. (Fig.2.14) illustrates the meaning of the term: CTDI is the equivalent of the dose value inside the irradiated slice (beam), that would result if the absorbed radiation dose profile were entirely concentrated to a rectangular of width equal to the nominal beam width with N being the number of independent (i.e. non-overlapping) slices that are acquired simultaneously. Accordingly, all dose contributions from outside the nominal beam width, i.e. the areas under the tails of the dose profile, are added to the area inside the slice (Nagel 2007).

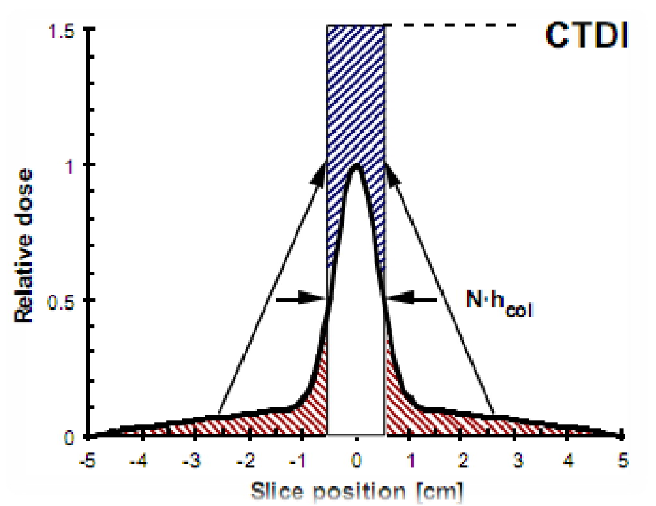



**Figure: 2.14:** Illustration of term ‘Computed Tomography Dose Index (CTDI)’: is the equivalent of the dose value inside the irradiated slice (beam) that would result if the absorbed radiation dose profile were entirely concentrated to a rectangular of width equal to the nominal beam width N.hcol, with N being the number of independent (i.e. non-overlapping) slices that are acquired simultaneously (Nagel 2007).

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

$$\mathrm{CTDI} = 1 \div \mathrm{N.hcol}.\int_{-\infty}^{+\infty} \mathrm{D(z).dz} \tag{2.2}$$

Where D(z) is the value of the dose at a given location, z, and N.hcol is the nominal value of the total collimation (beam width) that is used for data acquisition. CTDI is therefore equal to the area of the dose profile (the 'dose-profile integral') divided by the nominal beam width. In practice, the dose profile is accumulated in a range of -50 mm to +50 mm relative to the centre of the beam, i.e. over a distance of 100mm.

The relevancy of CTDI becomes obvious from the total dose profile of a scan series with e.g. n=15 subsequent rotations (Fig.2.15).The average level of the total dose profile, which is called 'Multiple Scans Average Dose (MSAD)' (Shope 1981), is higher than the peak value of each single dose profile. This increase results from the tails of the single dose profiles. Obviously MSAD and CTDI are exactly equal of the table feed (TF) is equal to the nominal beam width N.hcol, i.e. if the pitch factor

$$\mathrm{P} = \frac{\mathrm{TF}}{\mathrm{N.hcol}} \tag{2.3}$$

is equal to 1. In general (i.e. if the pitch factor is not equal to 1, Fig.2.16), the relationship between CTDI and MSAD is given by:

$$MSAD = {}^{1}\!/_{P.CTDI} \tag{2.4}$$

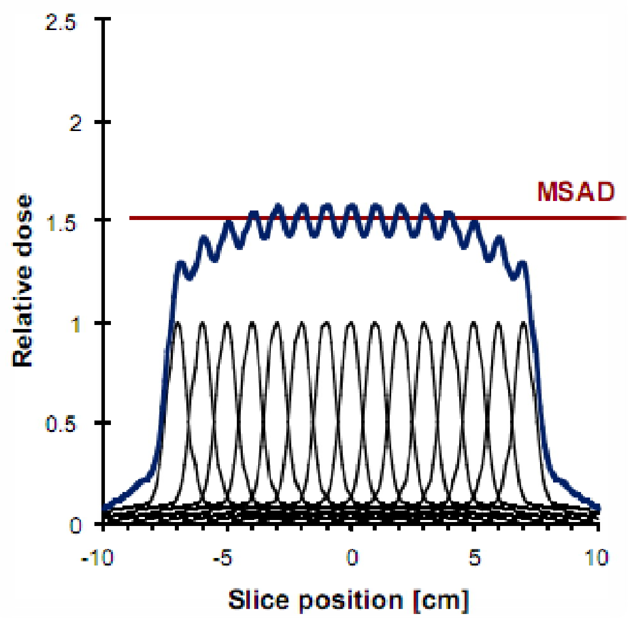



**Figure: 2.15:** The average level of the total dose profile, which is called 'Multiple Scans Average Dose (MSAD)'- (Shope 1981), is higher than the peak value of each single dose profile. This increase results from the tails of the single dose profiles (Nagel 2007).

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

$$CTDI_W = \frac{1}{3}CTDI_{100_c} + \frac{2}{3}CTDI_{100_P} (2.5)$$

If pitch-related effects on radiation exposure are taken into account at level of local dose (i.e. CTDI) already, a quantity named volume CTDI (CTDIvol)' is defined [IEC 2001]:

$$\text{CTDIvol} = \text{CTDIw/P} \qquad (2.6)$$

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

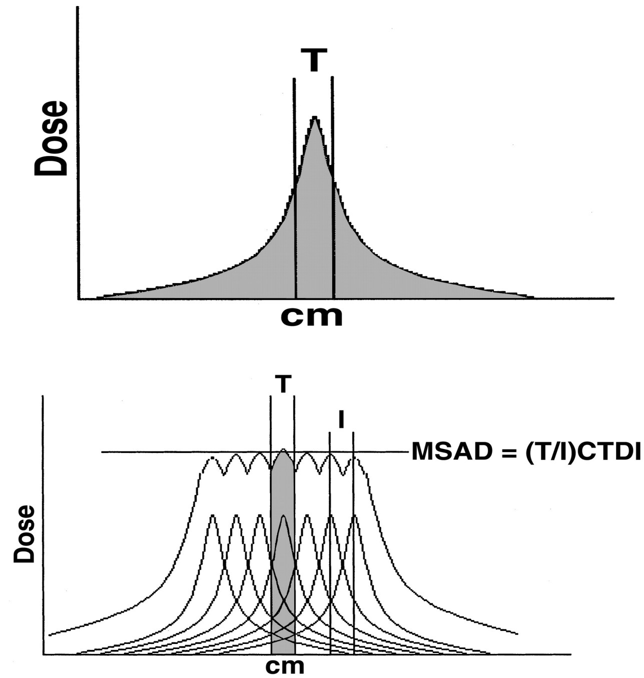



**Figure: 2.16: (1)** Schematic illustrates the profile of radiation dose delivered during a single CT scan. The CTDI equals the shaded area under the curve divided by the section thickness (*T*). (**2)** Schematic illustrates the profile of radiation dose delivered during multiple CT scans. *T* represents section thickness, and *I* represent the interval between sections. The MSAD includes the contributions of neighboring sections to the dose of the section of interest (D.Tack 2007).

### 2.9.2.2 Dose length product (DLP) unit (mGy)

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

$$DLP = CTDI_{vol} \cdot scan\ length \quad (2.7)$$

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

$$L = n*TF + N.hcol \quad (2.8)$$

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

$$L = n \cdot TF = T/_{t_{rot}} \cdot p \cdot N \cdot h_{col} \quad (2.9)$$

Where T is the total scan time, trot is the rotation time, and p is the pitch factor. While in sequential scanning the scan length L is equal to the range from the begin of the first slice till the end of the last, the (gross) scan length for spiral scanning not only comprises the (net) length of the imaged body section but also includes the additional rotations at the begin and the end of the scan ('over-ranging') that are required for data interpolation [European Commission 1999]. If an examination consists of several sequential scan series or spiral scans, the dose-length product of the complete examination (DLP exam) is the sum of the dose-length products of each single series or spiral scan:

$$DLP_{exam} = \sum_i DLP_i \quad (2.10)$$

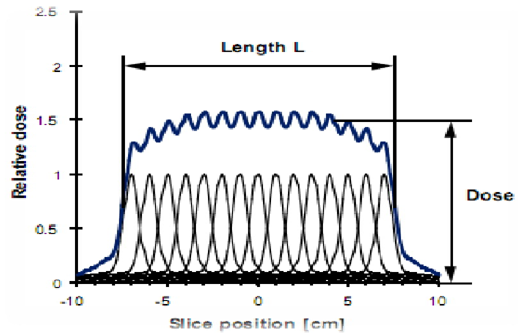



**Figure 2.17:** Dose length product (DLP) in CT (Total dose profile of a scan series with n=15 sub-sequent rotations. The dose-length product (DLP) is the product of the height (dose, i.e. CTDIvol) and the width (scan length L) of the total dose profile and is equal to the area under the curve (Nagel 2007).

### 2.9.3 DLP and Effective Dose

CTDI and DLP are CT specific dose descriptors that do not allow for comparisons with radiation exposure from other sources, projection radiography, nuclear medicine or natural background radiation. The only common denominator to achieve this goal is the (Effective Dose). With effective dose, the organ doses from a partial radiation of the body are converted into an equivalent uniform dose to the entire body. An effective Dose E unit (millisevert, mSv) according to ICRP 60 (ImPACT 2007) is defined as the weighted average of organ dose values HT for a number of specific organs:

$$E = \sum iWi* \qquad (2.11)$$

## 2.10 Previous studies

RT , Sodickson A , 2009 ,evaluated the cumulative radiation exposure and cancer risk estimate in emergency department patients undergoing repeat or multiple CT in order to define a conservative estimate of the number of patients undergoing repeat or multiple emergency department CT studies and to quantify their cumulative CT radiation doses and lifetime attributable risk of developing cancer. They found in conclusion a small proportion (1.9%) of emergency department patients undergoing CT of the neck, chest, abdomen, or pelvis have high cumulative rates of multiple or repeat imaging. Collectively, this patient subgroup may have a heightened risk of developing cancer from cumulative CT radiation exposure.

Numerous studies have suggested that, although CT is not the most commonly performed radiologic examination, it is the largest source of radiation dose. (Nagel et al. 1989) found that, although CT represents only about 4% of all radiologic examinations, it is responsible for up to 35% of collective radiation dose to the population from radiologic examinations. In related National Cancer Institute report, data suggested that the use of CT in adults and children has increased approximately 7 folds in the past 10 years. In large U.S hospitals, CT represents 10% of diagnostic procedures and accounts for approximately65% of the for all medical effective radiation dose examinations.

(Aldrich et al. 2007) conducted a study to compare the dose length product (DLP) and effective radiation dose to the patients from CT examinations. They compared data from 1070 CT examinations and concluded that considerable variation existed in the dose length product and patients radiation dose for specific examination. This study called attention to the need to optimize the effective dose to the patient and conduct more research to determine which additional efforts are needed to minimize patient exposure. Optimizing technical factors for examinations can help reduce patient radiation dose, thereby reducing risks. A pivotal study by (Lee 2001) assessed awareness levels among patients, emergency department physicians and the radiologists concerning radiation dose and the risks involved with CT scans. Lee and colleagues concluded that patients were not given information about the risks, benefits and radiation dose from a CT scan. Regardless of their experience levels, few of the participants in the study (including the emergency department physicians and the radiologists) were able to provide accurate estimates of CT radiation doses. This study underscores the prevalent lack of attention to the issue lifetime cumulative radiation doses. This must be become a central issue so that risk can be studied and monitored. One disadvantage to communicating instinct of cumulative radiation dose would be the natural instinct of some patients to defer or cancel the examination. Professionals should highlight the benefits of the examination when discussing risks with the patient. Physicians improve their understanding of radiation risks from medical imaging examinations.

(Alice B, et al. 2009) quantified retrospectively the effect of systematic use of tube current modulation for neuroradiology CT protocols on patient dose and image quality. The authors evaluated effect of dose modulation on four types of neuroradiologic CT studies: brain CT performed without contrast, material (unenhanced CT) in adult patients, unenhanced brain CT in pediatric patients, adult cervical spine CT, and adult cervical and intracranial CT angiography. For each type of CT study, three of 100 consecutive studies were reviewed: 100 studies performed without dose modulation, 100 studies performed with z-axis dose modulation, and 100 studies performed with x-y-z-axis dose modulation. For each examination, the weighted volume CT dose index (CTDIvol) and dose length product (DLP) were recorded and noise was measured. Each study was also reviewed for image quality. Continuous variables (CTDIvol, DLP, noise) were compared by using t test and categorical variables (image quality) were compared by using Wilcoxon rank-sum test.For unenhanced CT of adult brains,theCTDIvol and DLP, respectively, were reduced by 60.9% and 60.3%, respectively, by using z-axis dose modulation and by 50.4% and 22.4% by using x-y-z-axis dose modulation. Significant dose reductions ($P < 0.001$) were also observed for pediatric unenhanced brain CT, cervical spine CT, and adult cervical and intracranial CT angiography performed with each dose modulation technique. Image quality and noise were unaffected by use of either dose modulation technique (P < 0.05). Use ofdose modulation techniques for neuroradiology CT examinations afford significant dose reduction while image quality is maintained.

For unenhanced CT of adult brains, the CTDIvol and DLP, respectively, were reduced by 60.9% and 60.3%, respectively, by using z-axis dose modulation and by 50.4% and 22.4% by using x-y-z–axis dose modulation. Significant dose reductions (P < .001) were also observed for pediatric unenhanced brain CT, cervical spine CT, and adult cervical and intracranial CT angiography performed with each dose modulation technique. Image quality and noise were unaffected by the use of either dose modulation technique (P > .05).

Finally, a unique study conductedin Sudan regarding patient dose in CT (M A Aziz 2007). The study assessed the radiation doses for patients undergoing routine CT examinations in four centers in Khartoum state for various CT examinations of head, neck, abdomen, pelvis and chest. CTDIvol, DLP and effective dose were calculated using CT-exposure software. The mean CTDIw, CTDIvol DLP and effective dose were found to be 32.6 mGy, 26.5 mGy, 454 mGy and 3.3 mSv respectively.

# Chapter Three

# Material and Method

## 3.1 Material:

### 3.1.1 Subject:

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

| Hospital | Brain | Abdomen | Total |
|---|---|---|---|
| Hospital 1 | 45 | 24 | 69 |
| Hospital 2 | 35 | 24 | 59 |
| Total | 80 | 48 | 128 |

### 3.1.2 CT machines

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

**Table 3.2 CT machine**

| Hospital | manufacture | Model | installation | FAD | Detected Type |
|---|---|---|---|---|---|
| Hospital 1 | Toshiba | Aquilion 64 | 2009 | 70 | 64slice |
| Hospital 1 | Toshiba | Aquilion 64 | 2009 | 70 | 64slice |

## 3.2Method

### 3.2.1 Technique used

Data were collected using a sheet for all patients in order to maintain consistency of the information from display. A data collection sheet was designed to evaluate the patient doses and the radiation related factor. The collected data included , sex, and age; tube voltage and tube current–time product settings; pitch; section thickness; and number of sections, In addition, we also recorded all scanning parameters, as well as the CT dose descriptors CT weighted dose index (in millisievert) and dose-length product (in millisievert-centimeters).All these factors have a direct influence on radiation dose. The entire hospital was passed successfully the extensive quality control tests performed by Sudan atomic energy commission and met the criteria of this study.

### 3.2.2 Analysis of data

All dose parameters were registered down and from the display monitor in 64 slice CT scan and they use in calculation for the effective dose using conversion factor to the brain and abdomen, then used as input to the statistical software (SPSS) and Microsoft excel for analysis.

# CHAPTER FOUR

## Result

Table 4.1 show Patient exposure parameters during CT procedures: Mean±sd deviation and the range in the parenthesis at constant kVp =120.

**Descriptive Statistics (Brain)**

| | N | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|---|
| CTDI | 80 | 28.50 | 154.60 | 75.9200 | 15.44742 |
| Effective Dose | 80 | 1.09 | 7.21 | 3.0282 | .79849 |

Table 4.2 show CTDI and Effective dose for patient in two hospitals:

**Group Statistics (Brain)**

| | Hospital | N | Mean | Std. Deviation | Std. Error Mean |
|---|---|---|---|---|---|
| CTDI | Hospital 1 | 45 | 75.4444 | 19.39502 | 2.89124 |
| | Hospital 2 | 35 | 76.5314 | 8.18283 | 1.38315 |
| Effective Dose | Hospital 1 | 45 | 2.9635 | .96690 | .14414 |
| | Hospital 2 | 35 | 3.1115 | .50882 | .08601 |

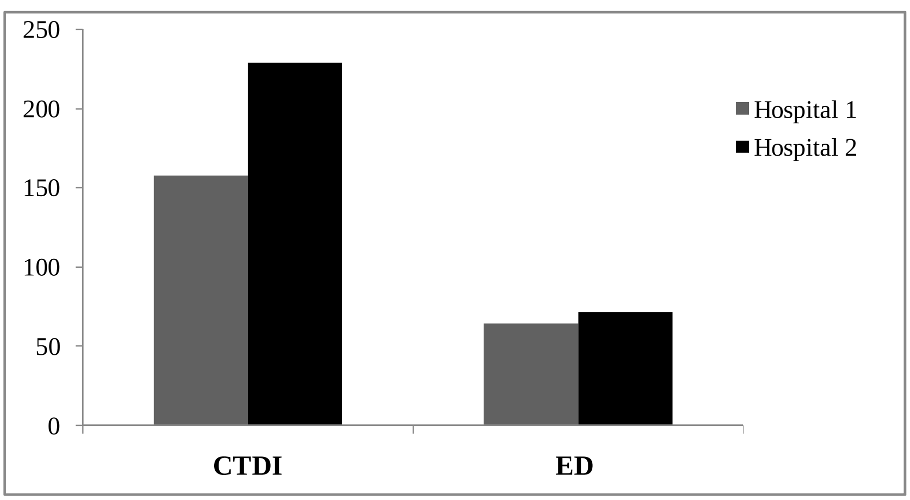



Figure 4-1Show the difference between Effective Dose and CTDIvol in two hospitals during Brain CT scan.

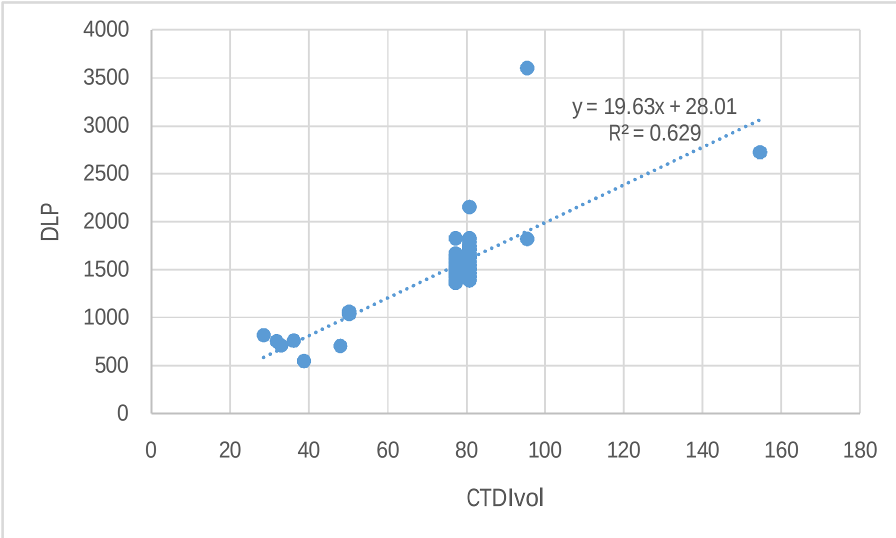



**Figure 4-2Show the relationship between CTDIvol and DLP in hospital 1 during Brain CT scan.**

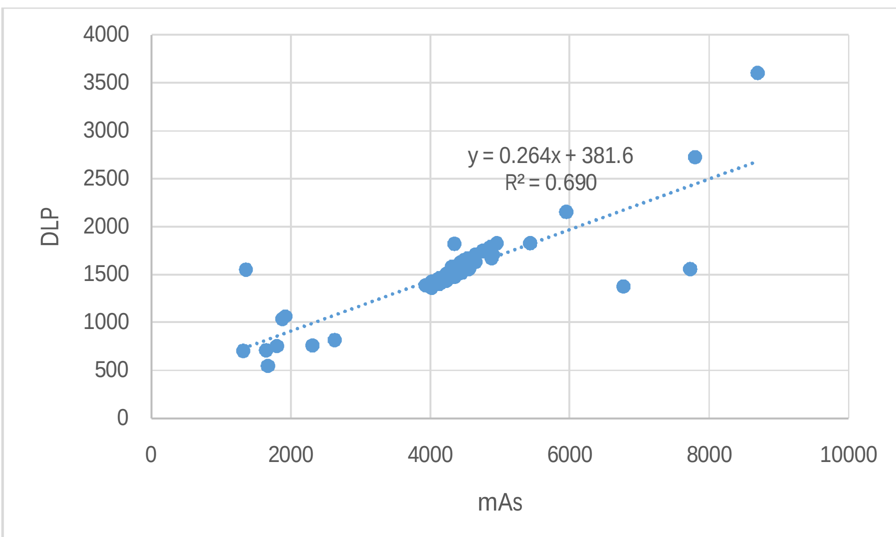



**Figure 4-3Show the relationship between mAs and DLP in hospital 1 during Brain CT scan.**

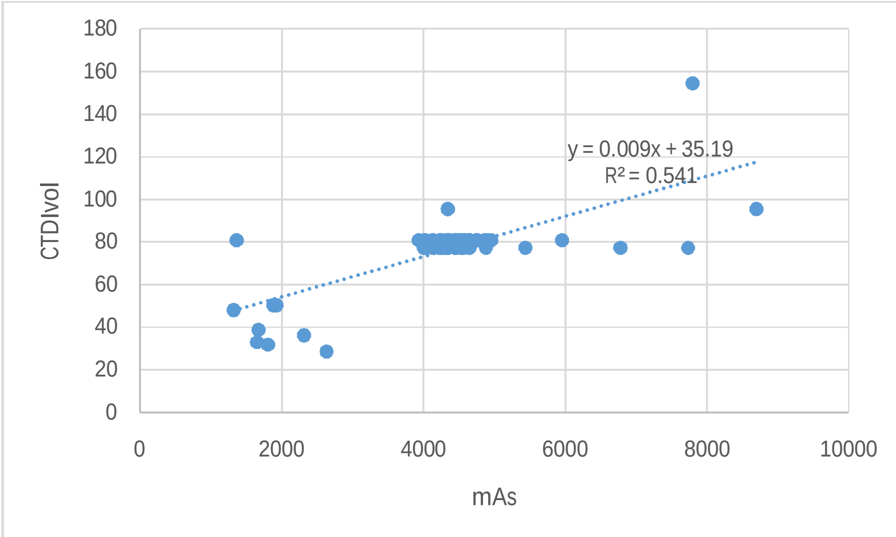



**Figure 4-4Show the relationship between mAs and CTDIvol in hospital 1 during Brain CT scan.**

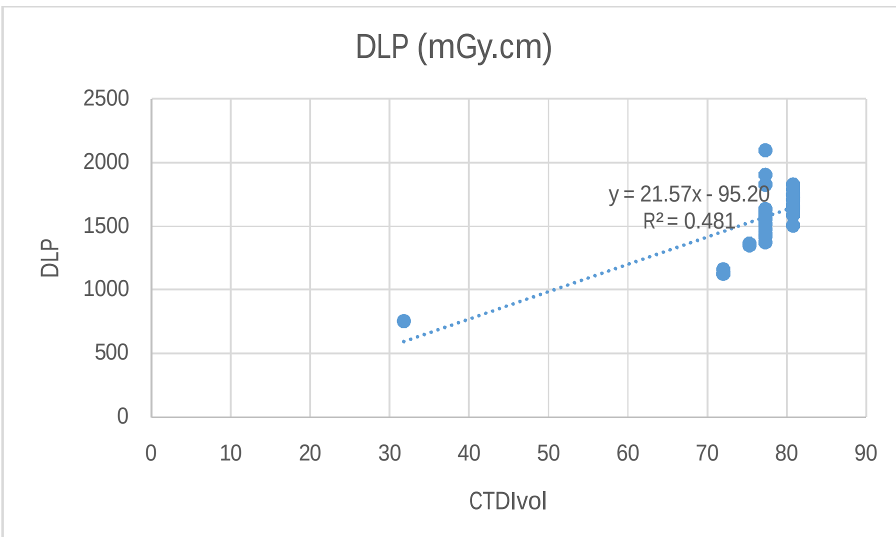



**Figure 4-5Show the relationship between CTDIvol and DLP in hospital 2 during Brain CT scan.**

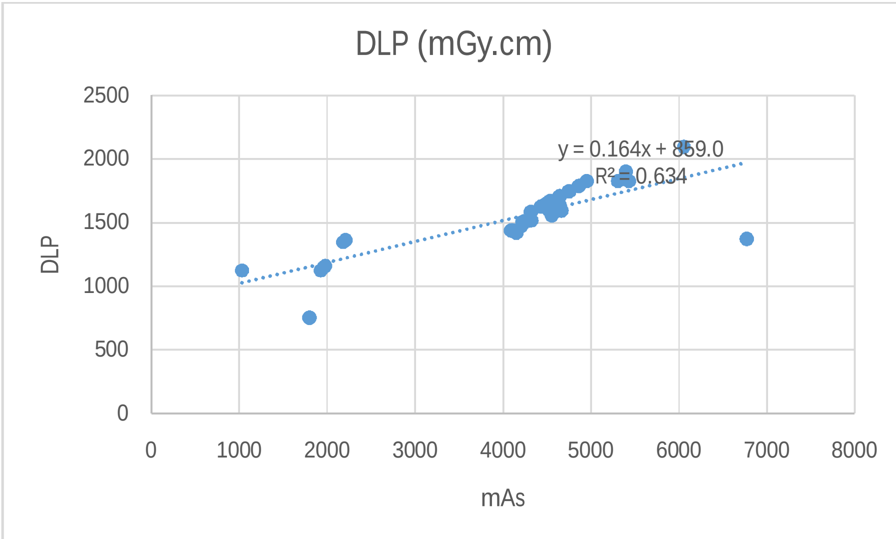



**Figure 4-6Show the relationship between mAs and DLP in hospital 2 during Brain CT scan.**

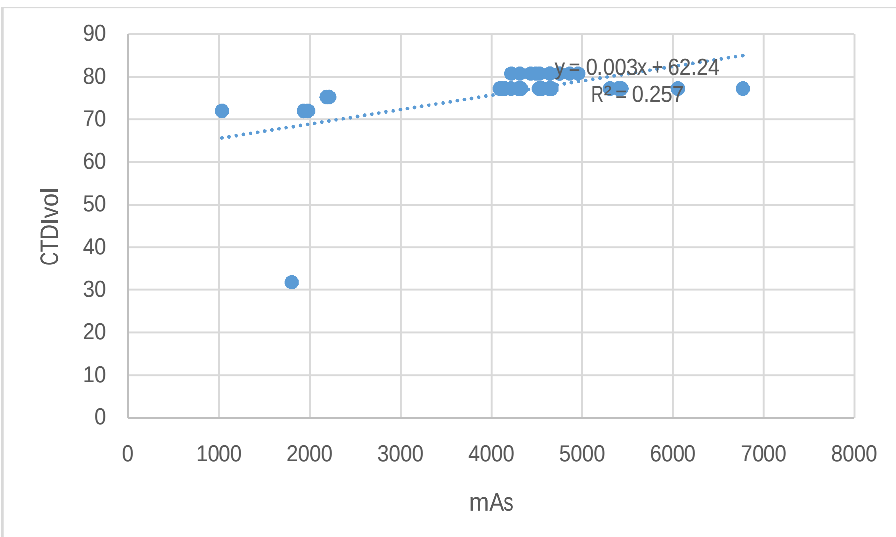



**Figure 4-7Show the relationship between mAs and CTDIvol in hospital 2 during Brain CT scan**

Table 4.3 show Patient exposure parameters during CT procedures: Mean±sd deviation and the range in the parenthesis at constant kVp =120.

**Descriptive Statistics (Abdomen)**

| | N | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|---|
| CTDI | 48 | 10.30 | 599.70 | 193.1542 | 165.24696 |
| Effective Dose | 48 | 7.20 | 119.95 | 67.9613 | 26.69254 |

Table: 4.4 Show CTDI and Effective dose for patient in two hospitals:

**Group Statistics(Abdomen)**

| | Hospital | N | Mean | Std. Deviation | Std. Error Mean |
|---|---|---|---|---|---|
| CTDI | Hospital 1 | 24 | 157.5833 | 132.28014 | 27.00157 |
| | Hospital 2 | 24 | 228.7250 | 188.84314 | 38.54744 |
| Effective Dose | Hospital 1 | 24 | 64.3146 | 29.79568 | 6.08202 |
| | Hospital 2 | 24 | 71.6080 | 23.24705 | 4.74528 |

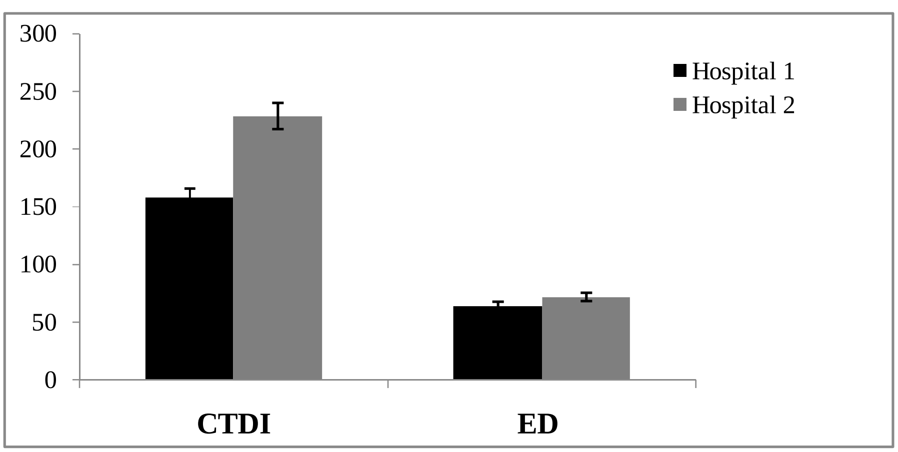



Figure 4-2Show the difference between Effective Dose and CTDIvol in two hospitals during Abdomen CT scan

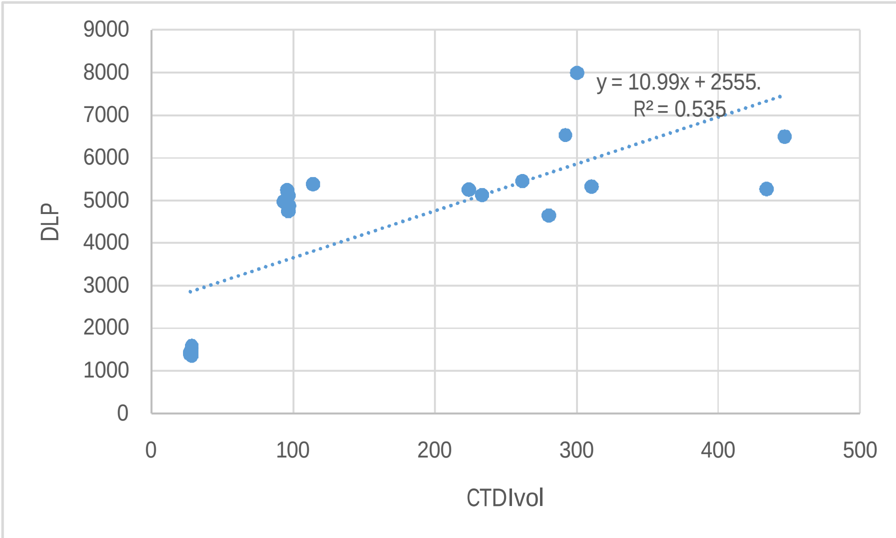



**Figure 4-9 Show the relationship between CTDIvol and DLP in hospital1 during Abdomen CT scan.**

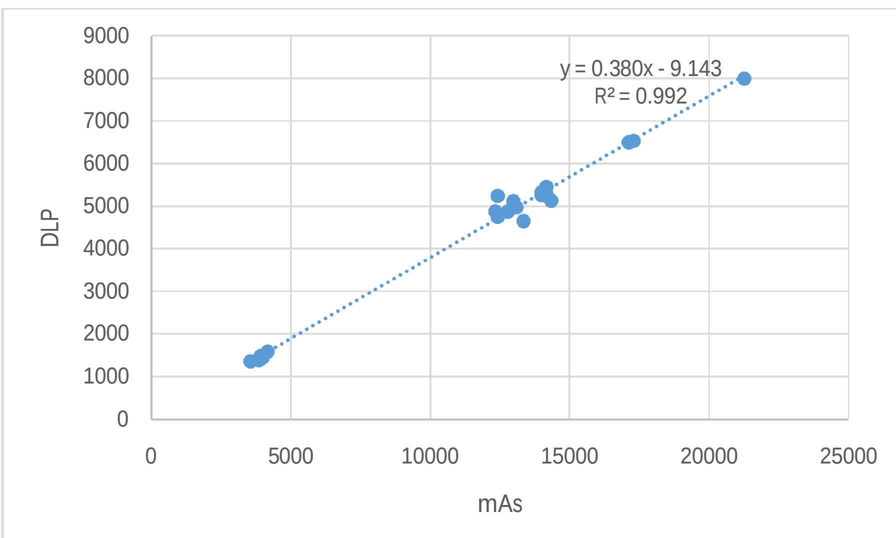



**Figure 4-10 Show the relationship between mAs and DLP in hospital1 during Abdomen CT scan.**

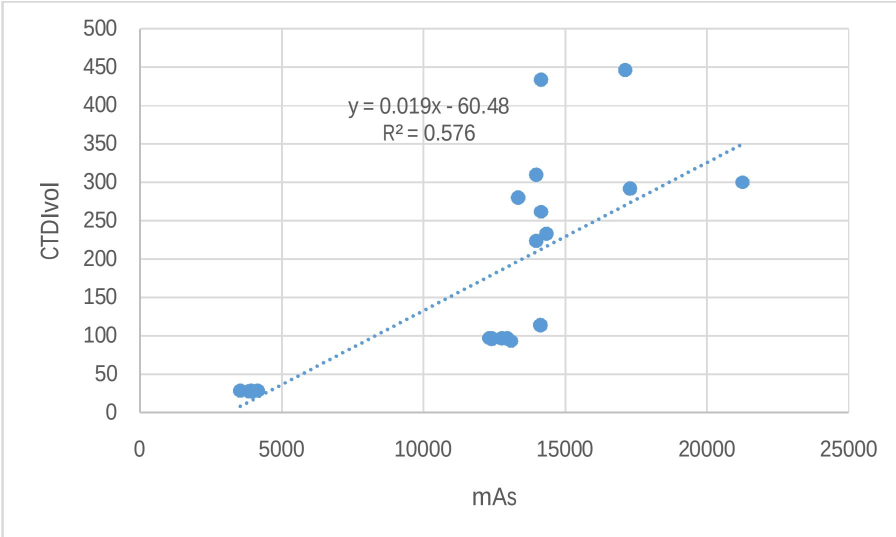



**Figure 4-11Show the relationship between mAs and CTDIvol in hospital1 during Abdomen CT scan.**

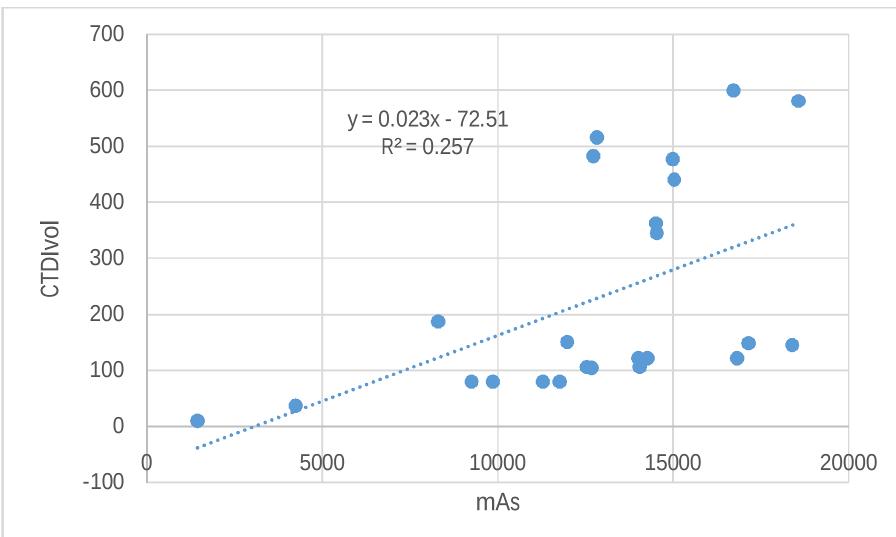



**Figure 4-12 Show the relationship between mAs and CTDIvol in hospital2 during Abdomen CT scan.**

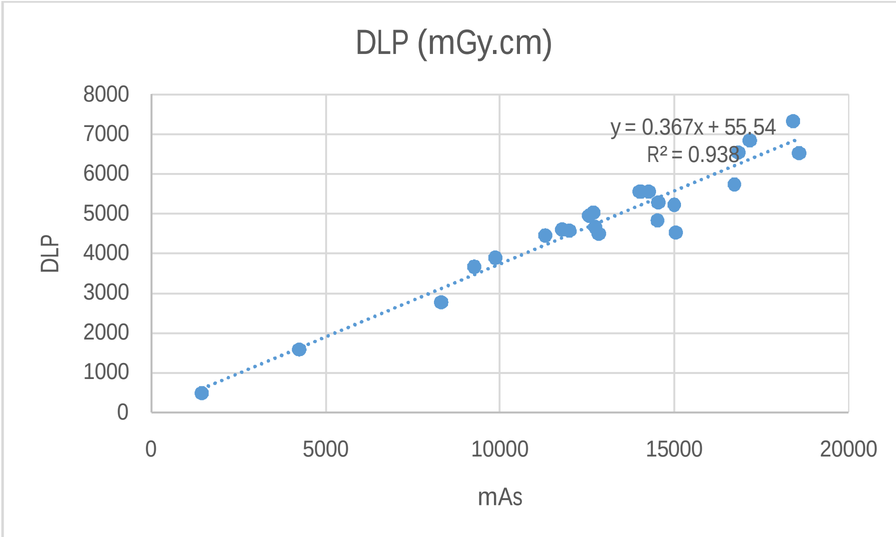



**Figure 4-13 Show the relationship between mAs and DLP in hospital2 during Abdomen CT scan.**

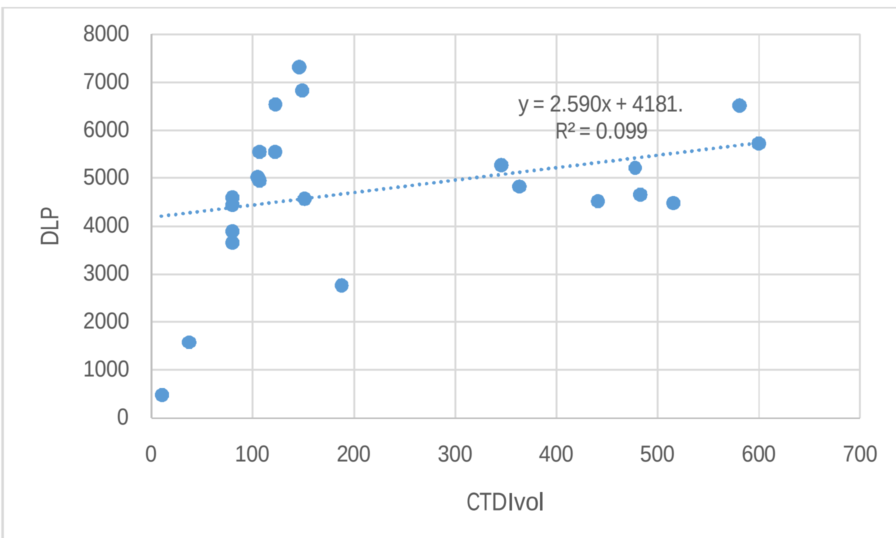



**Figure 4-14 Show the relationship between CTDIvol and DLP in hospital2 during Abdomen CT scan.**

**Table: 4.5:** Comparison of patient dose during CT with previous studies**:**

| Author | No. of pts | Exam | Machine model | Pitch | kVp | mAs | Slice th | Dose mSv | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | | | | | | | | CTDIvol | DLP | Effective dose |
| Ali Abdelrazig | 31 | Chest, Abd, Brain & s | Toshiba Sensation aquilion 64 | 1.5 | 120 | 242.8 | 5.5 | 178.3 | 2344.4 | 20.05 |
| A.M Nour | 83 | Abd | Siemens Somatom emotion | 0.75-1 | 80-120 | 42-243 | 24 | 18.87 mGy | 865.3 mGy.cm | 13.5 |
| Entisar Omer | 51 | CTU | Siemens Somatom emotion duo | | 110 - 130 | 37-111 | | 25.1-10.95 | 85-425 | 1.29-6.37 |
| European Commission1999 | | Routine Abd, Pelvis & Liver | | | | | | | 780 | 11.7 |
| I.I.Suliman | 445 | Head, chest, abd& pelvis | Toshiba Somatom sensation 16 | NA | 120 | 41±17 | 5.8±1 | 65 | 507.3 | 11.3 |
| In this study | 80<br>48 | Head<br>abdomen | Toshiba 64 | N.A | 120 | | N.A | 75.95<br>134.525 | 1518.7<br>4530.7 | 3.02<br>67.96 |

# CHAPTER FIVE

## Discussion, conclusion and recommendations

### 5.1 Discussion

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

In this study, a total of 128 adult patients undergoing brain and the abdominal CT scanning exams were evaluated, exposure factor and dose estimate factors (CTDI and DLP) were collected from 64 slice, multidetector CT (MDCT) scanners.

The patients were scanned with kVp (120), mAs range (1034-12832.66), and the pitch was (0.8) or less than 1. These parameters produced these radiation values represented in Table (4.2 and 4.4) which showed the values of DLP average (4530.7mGy.cm), $CTDI_{vol}$ average (193.1mGy) for CT abdomen and DLP average (1518.7mGy.cm), $CTD_{Ivol}$average (75.95mGy) for CT brain, these values were found to be at standard dose reference level (Mettler, et.al , 2008)

In CT examination, patients are exposed to high radiation dose. Therefore, the used of ordinary dose values (CTDI, or DLP) will provide less information regarding the radiation risks. Effective dose is the unit of choice in this situation (partial exposure) and furthermore, comparisons between different procedures are possible with different imaging modalities. In this study, the mean effective dose for hospital 1 was 2.963 ± 0.966 mSv and 64.31±29.795mSv for the brain and abdomen respectively. The mean effective dose for hospital 2 was 3.1115± 0.50882 mSv and 71.60 ± 23.24 mSv for brain and abdomen respectively.

The result of the study showed that the radiation dose to patients in two hospitals, the effective dose and CTDIvol of CT brain and abdomen examination in hospital 1 was higher in hospital 2. These variations may be due to patient clinical indication, CT system modality and image acquisition parameters.

DRLs can be used to verify the practices for typical examinations for group of standardized patients in order to ensure that the dose should not be exceeded in normal practice without adequate justification [ICRP 1991]. Nevertheless, the available data is still not enough to establish national reference levels, but this could be a baseline for further studies concerning dose optimization.

CT examinations of the whole body are justified by the ability to detect alternative/or additional diagnoses. However, since the body contains sensitive organs, the radiation dose delivered to the patients becomes a particular concern, especially in young patients and in those with chronic diseases who undergo repeated CT studies.

Radiation dose from head CT scans may vary considerably as a result of inherent differences in equipment and because of variations in exposure technique and scanning protocol. Previous studies where systemic changes in scanning parameters were analyzed with respect to resulting image quality have reported dose reductions of up to 40% in CT scans of the head without loss of relevant information or diagnostic image quality [Alice B 2008, Cohnen, et al. 2000].

Unjustified screening the Abdomen and head, should thus be banished. Such policy is unacceptable in young patients who are at a low risk of having an incidental associated disease. Similarity, repeated acquisition should not be performed in circumstances where they do not specifically yield additional information.

Automatic Exposure Control (AEC) devices that are nowadays available in modern equipment module the tube current as function of the table position along the z-axis and of the image quality requested by the radiologist. Such increase it in obese and overweight patients, tending to maintain the image quality constant. Therefore, radiologists using these devices should think on terms of image quality and not of the tube current. Mulkens et al. 2005 showed that systems based on both angular and z-axis modulation reduce the mean tube current by 20% – 68% when applied to the standard MDCT protocols at constant tube current. With such systems, these authors also showed a good correlation between the mean effective tube current and patient's body mass index (BMI), with an adaptation in obese and overweight patients leading to reference tube current level being exceeded.

## 5.2 Conclusion

The radiation dose was measured in two hospitals using similar CT modalities.Theradiation dose higher in hospital 2 than hospital 1. 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 estimates that cannot be precisely associated with any individual patient, examination, or type of CT system. The main dose variations in the same Ct unit could be attributed to the different techniques, which justify the important of use radiation dose optimization technique and technologists training. Dose reduction strategies must be well understood and properly used.

### 5.3 Recommendations

CT operators must optimize the patient dose for patient to reduce patient cancer risks. Should be uses the best strategies available for reducing radiation dose to allow for mAs reduction in relation to the patient’s size and weight , adapted tube current based on patient size (such as weight with fixed tube current scanning).

(i) Implementation of automatic exposure control systems by the manufacturers.
(ii) Achieve optimization through; the design of dose efficient equipment, the optimization of scan protocol and improvement of referring criteria.

The radiologists and CT technologists must be trained to adapt CT scanning techniques based on clinical indications and to assess associated radiation doses with different scanning parameters.

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