CHAPTER ONE

1.1 INTRODUCTION

CT reconstructs a cross-sectional image of the body from a 'virtual pile of X-ray photographs'. A tomographic image is an image of a slice through the body. The word 'tomography' comes from the Greek: tomos means slice , graphein stands for 'to write'. So, tomography literally means 'writing slices '. Structures and lesions previously impossible to visualise can now be seen with remarkable clarity .   CT is unique from other imaging techniques in that 3D image acquisition of the entire thoracic region- can be performed within a single breath-hold . As a result, chest CT is frequently used as a diagnostic tool (Hricak H, Brenner DJ, Adelstein SJ, 2010) .

As a consequence of the success of medical imaging over the past decades for aid in accurately diagnosing disease or injury and guiding therapy, the collective radiation dose delivered to the U.S. population from medical imaging has increased six-fold since the 1980s . This has resulted in substantial concern from physicians, patients, and regulators. Consequently, radiation dose management and reduction have become one of the most important challenges facing medical imaging providers (Mettler FA Jr., Bhargavan M, Faulkner K,2009) .

In discussion of individual risk during CT, organ dose is a better measure for estimating patient risk than is effective dose because effective dose is intended for estimating radiation exposure of entire populations, not for individual dose estimates .

For chest CT, the most radiosensitive organs are the lungs and glandular breast . Risk of breast cancer may correlate with doses less than 100 mGy, and risk of lung cancer may correlate with doses as low as 100 mGy .

Radiation protection in medicine is based on two guiding principles: (a) the examination or procedure must be medically indicated, and (b) the examination or procedure must use doses that are as low as reasonably achievable—the ALARA principle—without compromising the diagnostic task (Fred A. Mettler, Jr., et al., 2008) .

However, these simple principles can be difficult to apply in clinical practice, because quantification of either the risks or the benefits associated with exposures to ionizing radiation is not always straight forward. Even the standard metrics for describing the amount of radiation dose delivered to a patient and the optimal methods for dose reduction are still a matter of debate .

Although a majority of the publications are related to dose issues arising from medical uses of computed tomography (CT) (probably because CT is responsible for the largest part of collective dose delivered by medical imaging, followed by nuclear medicine and interventional radiology) .

1.2 PROBLEM OF THE STUDY :

Most existing methods for estimating radiation dose to the chest during chest CT are based on measurements or simulations of a single patient model or phantom with unrealistic anatomic features (NCRP, 2009) .

1.3 Objectives :

1.3.1 General Objectives :

❖ The aims of this study is to evaluate the dose -using Monte Carlo - to radiosensitive organs (Chest) in patients of various sizes undergoing routine chest CT examinations .

1.3.2 Specific Objectives :

❖ To investigate the relation between patient size and organ dose to Chest resulting from chest CT examinations.

❖ To find if the patient doses are kept as low as reasonably achievable to ensure that any patient risks are minimized.

1.4 Materials and Methods :

Most existing methods for estimating radiation dose to the chest during chest CT are based on measurements or simulations of a single patient model or phantom with unrealistic anatomic features (e.g., breasts modeled as a homogeneous material located directly anterior to the thoracic region) . It is not known how well these methods serve for estimation of organ doses in an actual patient population, which includes natural variations in patient size and breast composition.

Monte Carlo simulations were used to simulate CT scans by means of modeling of the patients, scanner geometry, and photon transport through the patients .

Using Monte Carlo simulation–based MDCT scanner models, we can overcome the limitations of previous dose estimation tools by incorporating a range of actual anatomic features of patients .

The data used in this study will be collected from diagnostic radiology departments at Khartoum state. Patient models based on CT images of actual men and women were extended to include chest contours for this study.

CHAPTER TWO

THEORITICAL BACKGROUD

2.1 Radiation:

The propagation of energy from a radioactive source to another medium is termed radiation. This transmission of energy can take the form of particulate radiation or electromagnetic radiation (i.e., electromagnetic waves). The various forms of radiation originating from atoms, which include (among others) visible light, X-rays and □-rays, are grouped together under the terms "electromagnetic radiation" or "the electromagnetic Spectrum". Radio waves, which have the longest wavelengths and thus the lowest frequencies and energies of the various types of electromagnetic radiation, are located at tone end of the electromagnetic spectrum, whereas X-rays and □-rays, which have the highest frequencies and energies, are situated at the other end of this spectrum. (James E. Martin,2006).

2.1.1 Photon:

If the smallest unit of an element is considered to be its atoms, the photon is the smallest unit of electromagnetic radiation. Photons have no mass.

2.1.2 Common features of electromagnetic radiation:

It propagates in a straight line, it travels at the speed of light (nearly 300,000 km/s), it transfers energy to the medium through which it passes, and the amount of energy transferred correlates positively with the frequency and negatively with the wavelength of the radiation.

The energy of the radiation decreases as it passes through a material, due to absorption and scattering, and this decrease in energy is negatively correlated with the square of the distance traveled through the material.

Electromagnetic radiation can also be subdivided into ionizing and non ionizing radiations. Non ionizing radiations have wavelengths of 10^3−7 m. Non ionizing radiations have energies of <12 electron volts (eV); 12 eV is considered to be the lowest energy that an ionizing radiation can possess. (James E. Martin,2006).

2.2 X-ray :

x-rays are produced when highly energetic electrons interact with matter, converting some of their kinetic energy into electromagnetic radiation. A device that produces x-rays in the diagnostic energy range typically contains an electron source, an evacuated path for electron acceleration, a target electrode, and an external power source to provide a high voltage (potential difference) to accelerate the electrons. Specifically, the x-ray tube insert contains the electron source and target within an evacuated glass or metal envelope; the tube housing provides protective radiation shielding and cools the x-ray tube insert; the x-ray generator supplies the voltage to accelerate the electrons; x-ray beam filters at the tube port shape the x-ray energy spectrum; and collimators define the size and shape of the x-ray field incident on the patient. The generator also permits control of the x-ray beam characteristics through the selection of voltage, current, and exposure time. These components work in concert to create a beam of x-ray photons of well defined intensity, penetrability, and spatial distribution .
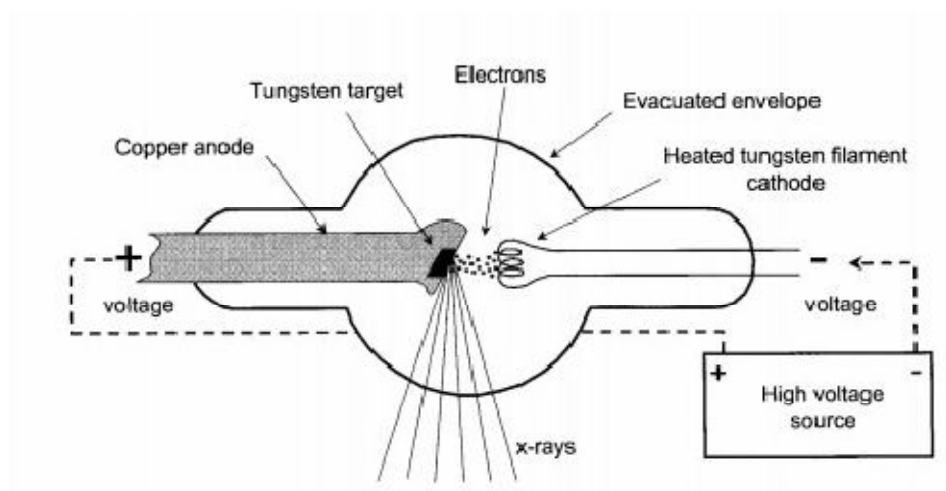
2.3 Production Of X-ray :

x-rays are created from the conversion of kinetic energy of electrons into electromagnetic radiation when they are decelerated by interaction with a target material. A simplified diagram of an x-ray tube (Fig. 2-1) illustrates these components. For diagnostic radiology, a large electric potential difference (the SI unit of potential difference is the volt, V) of (20 to 150 kV) is applied between two electrodes (the cathode and the anode) in the vacuum. The cathode is the source of

electrons, and the anode, with a positive potential with respect to the cathode, is the target of electrons. As electrons from the cathode travel to the anode, they are accelerated by the voltage between the electrodes and attain kinetic energies equal to the product of the electrical charge and potential difference . A common unit of energy is the electron volt (eV), equal to the energy attained by an electron accelerated across a potential difference of 1V. Thus, the kinetic energy of an electron accelerated by a potential difference of 50 kV is 50 keV. One eV is a very small amount of energy, as there are $6.24 * 10^{18}$ eV/j.

On impact with the target, the kinetic energy of the electrons is converted to other forms of energy. The vast majority of interactions are *collisional,* whereby energy exchanges with electrons in the target give rise to heat. A small fraction of the accelerated electrons comes within the proximity of an atomic nucleus and is influenced by its positive electric field.

Electrical (Coulombic) forces attract and decelerate an electron and change its direction, causing a loss of kinetic energy, which is emitted as an x-ray photon of equal energy (i.e., bremsstrahlung radiation).

■■FIGURE 2-1 Minimum requirements for x-ray production include a source and target of electrons, an evacuated envelope, and connection of the electrodes to a high-voltage source.

The amount of energy lost by the electron and thus the energy of the resulting x-ray are determined by the distance between the incident electron and the target nucleus, since the Coulombic force is proportional to the inverse of the square of the distance. At relatively large distances from the nucleus, the Coulombic attraction is weak; these encounters produce low x-ray energies (Fig. 2-2, electron no. 3). At closer interaction distances, the force acting on the electron increases, causing a greater deceleration; these encounters produce higher x-ray energies (see Fig. 2-2, electron no. 2).
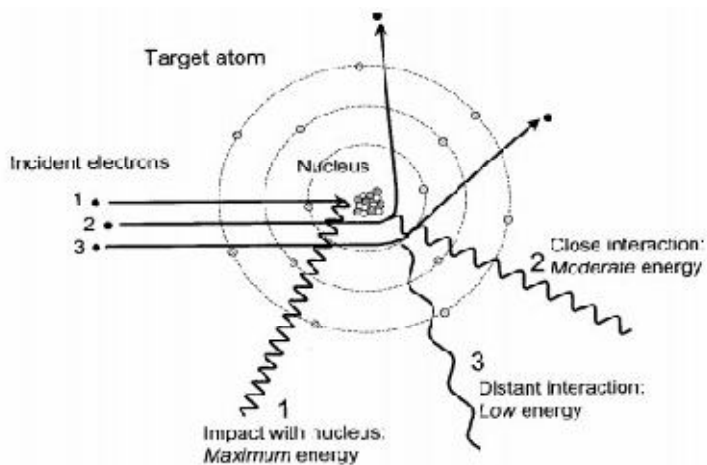
■■FIGURE 2-2 Bremsstrahlung radiation arises from energetic electron interactions with an atomic nucleus of the target material.

 In a "close" approach, the positive nucleus attracts the negative electron, causing deceleration and redirection, resulting in a loss of kinetic energy that is converted to an x-ray. The x-ray energy depends on the interaction distance between the electron and the nucleus; it decreases as the distance increases.

A nearly direct impact of an electron with the target nucleus results in loss of nearly all of the electron's kinetic energy (see Fig. 2-2, electron no. 1). In this rare situation, the highest x-ray energies are produced. The probability of electron interactions that result in production of x-ray energy E is dependent on the radial interaction distance, r, from the nucleus, which defines a circumference, 2pr. With increasing distance from the nucleus, the circumference increases, and therefore the probability of interaction increases, but the x-ray energy decreases. Conversely, as the interaction distance, r, decreases, the x-ray energy increases because of greater electron deceleration, but the probability of interaction decreases. For the closest electron-atomic nuclei interactions,

the highest x-ray energies are produced. However, the probability of such an interaction is very small, and the number of x-rays produced is correspondingly small.

The number of x-rays produced decreases linearly with energy up to the maximal x-ray energy, which is equal to the energy of the incident electrons. A bremsstrahlung spectrum is the probability distribution of x-ray photons as a function of photon energy (keV). The unfiltered bremsstrahlung spectrum (Fig. 2-3A) shows an inverse linear relationship between the number and the energy of the x-rays produced, with the highest x-ray energy determined by the peak voltage (kV) applied across the x-ray tube.

A typical filtered bremsstrahlung spectrum (Fig. 2-3B) has no x-rays below about 10 keV; the numbers increase to a maximum at about one third to one half the maximal x-ray energy and then decrease to zero as the x-ray energy increases to the maximal x-ray energy. Filtration in this context refers to the removal of x-rays by attenuation in materials that are inherent in the x-ray tube (e.g., the glass window of the tube insert), as well as by materials that are purposefully placed in

the beam, such as thin aluminum and copper sheets, to remove lower energy x-rays and adjust the spectrum for optimal low-dose imaging.

Major factors that affect x-ray production efficiency include the atomic number of the target material and the kinetic energy of the incident electrons.
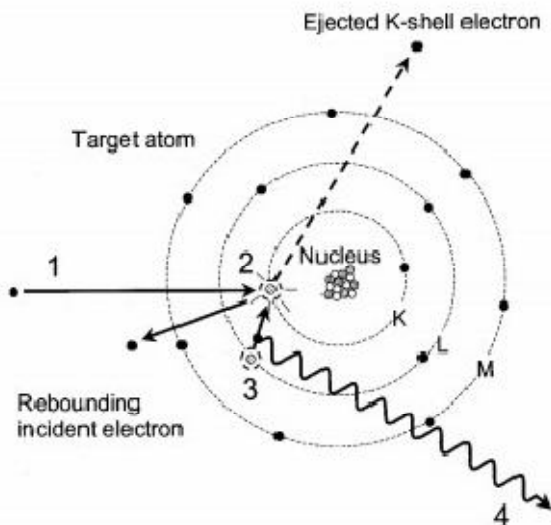
2.3.1 Bremsstrahlung Spectrum :

The conversion of electron kinetic energy into electromagnetic radiation produces x-rays. A simplified diagram of an x-ray tube . A large voltage is applied between two electrodes (the cathode and the anode) in an evacuated envelope. The cathode is negatively charged and is the source of electrons; the anode is positively charged and is the target of electrons. As electrons from the cathode travel to the anode, they are accelerated by the electrical potential difference between these electrodes and attain kinetic energy. The electric potential difference, also called the voltage. The kinetic energy gained by an electron is proportional to the potential difference between the cathode and the anode. On impact with the target, the kinetic energy of the electrons is converted to other forms of energy. The vast majority of interactions produce unwanted heat by small collision energy exchanges with electrons in the target. This intense heating limits the number of x-ray photons that can be produced in a given time without destroying the target. Occasionally (about 0.5% of the time), an electron comes within the proximity of a positively charged nucleus in the target electrode. Columbic forces attract and decelerate the electron, causing a significant loss of kinetic energy and a change in the electron's trajectory. An x-ray photon with energy equal to the kinetic energy lost by the electron is produced (conservation of energy). This radiation is termed bremsstrahlung, a German word meaning "braking radiation."The probability of an electron's directly impacting nucleus is extremely low, simply because, at the atomic scale, the atom comprises mainly empty "space" and the nuclear cross-section is very small.

Therefore, lower x-ray energies are generated in greater abundance, and the number of higher-energy x-rays decreases approximately linearly with energy up to the maximum energy of the incident electrons . A bremsstrahlung spectrum depicts the distribution of x-ray photons as a function of energy.

2.3.2 Characteristic X-Ray Spectrum :

Each electron in the target atom has a binding energy that depends on the shell in which it resides. Closest to the nucleus are two electrons in the K shell, which has the highest binding energy. The L shell, with eight electrons, has the next highest binding energy, and so forth. When the energy of an electron incident on the target exceeds the binding energy of an electron of a target atom, it is energetically possible for a collision interaction to eject the electron and ionize the atom. The unfilled shell is energetically unstable, and an outer shell electron with less binding energy will fill the vacancy. As this electron transitions to a lower energy state, the excess energy can be released as a characteristic x-ray photon with energy equal to the difference between the binding energies of the electron shells (James E. Martin,2006).

■■FIGURE 2-3 Generation of a characteristic x-ray in a target atom .
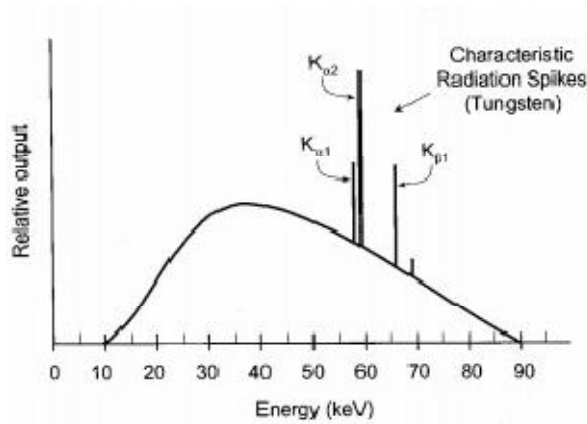
■■FIGURE 2-4 The filtered spectrum of bremsstrahlung and characteristic radiation from a tungsten target with a potential difference of 90 kV .

2.4 X- Ray interaction with mater :

X-rays have very short wavelengths, approximately $10{-}8$ to $10{-}9$ m. The higher the energy of an x-ray, the shorter is its wavelength. Consequently, low-energy x-rays tend to interact with whole atoms, which have diameters of approximately $10{-}9$ to $10{-}10$ m; moderate energy x-rays generally interact with electrons, and high-energy x-rays generally interact with nuclei X-rays interact at these various structural levels through five mechanisms: coherent scattering, Compton scattering, photoelectric effect, pair production, and photodisintegration.

Two of these Compton scattering and photoelectric effect are of particular importance to diagnostic radiology. They are discussed in some detail here (James E. Martin,2006).

2.4.1 Compton Scattering :

In Compton scattering, the incident x-ray interacts with an outer shell electron and ejects it from the atom, there by ionizing the atom. The ejected electron is called a Compton electron. The x-ray continues in a different direction with less energy.

The energy of the Compton-scattered x-ray is equal to the difference between the energy of the incident x-ray and the energy of the ejected electron.

The energy of the ejected electron is equal to its binding energy plus the kinetic energy with which it leaves the atom, During Compton scattering, most of the energy is divided between the scattered x-ray and the Compton electron(Murat, 2010) .
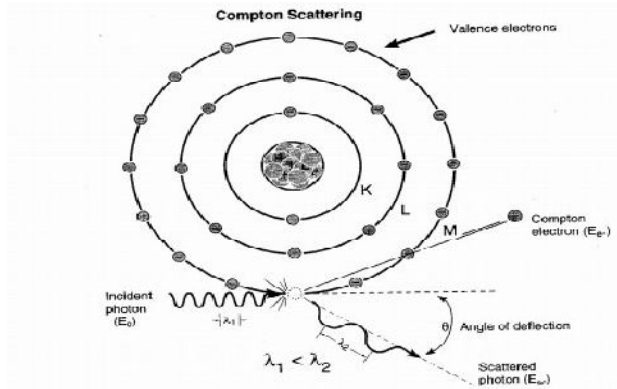
■■FIGURE 2-5 The diagram shows Compton scattering.

2.4.2 Photoelectric Effect :

X-rays in the diagnostic range also undergo ionizing interactions with inner-shell electrons. The x-ray is not scattered, but it is totally absorbed. This process is called the photoelectric effect, The electron removed from the atom, called a photoelectron, escapes with kinetic energy equal to the difference between the energy of the incident x-ray and the binding energy of the electron A photoelectric interaction cannot occur unless the incident x-ray has energy equal to or greater than the electron binding energy (Stewart, 2013).
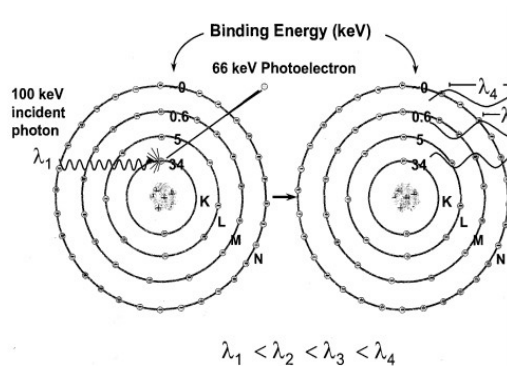
■■FIGURE 2-6 Photoelectric absorption, Left.

The diagram shows that a 100-keV photon is undergoing photoelectric absorption with an iodine atom. In this case, the *K*-shell electron is ejected with a kinetic energy equal to the difference (67 keV) between the incident photon energy (100 keV) and the *K*-shell binding energy (33 keV). Right. The vacancy created in the *K* shell results in the transition of an electron from the *L* shell to the *K* shell. The difference in their binding energies (i.e., 33 and 5 keV) results in a 28-keV Ka characteristic x-ray.

2.4.3 Beam Attenuation :

An x-ray beam consists of bundles of energy known as photons. These photons may pass through or be redirected (i.e., scattered) by a structure. A third option is that the photons may be absorbed by a given structure in varying amounts , depending on the strength (average photon energy) of the x-ray beam and the characteristics of the structure in its path. The degree to which a beam is reduced is a phenomenon referred to as attenuation ((Stewart, 2013)).

2.5 Computed Tomography (CT) :

The word tomography has as its root tomo, meaning to cut, section, or layer from the Greek tomos (a cutting). In the case of CT, a sophisticated computerized method is used to obtain data and transform them into "cuts," or cross-sectional slices of the human body.

Conventional radiographs depict a three-dimensional object as a two-dimensional image. This results in overlying tissues being superimposed on the image, a major limitation of conventional radiography. Computed tomography (CT) overcomes this problem by scanning thin sections of the body with a narrow x-ray beam that rotates around the body, producing images of each cross section. Another limitation of the conventional radiograph is its inability to distinguish between two tissues with similar densities. The unique physics of CT allow for the differentiation between tissues of similar densities.

The main advantages of CT over conventional radiography are in the elimination of superimposed structures, the ability to differentiate small differences in density of anatomic structures and abnormalities, and the superior quality of the images (Romans, 2011).

2.5.1 Principles Of Operation :

When the abdomen is imaged with conventional radiographic techniques, the image is created directly on the screen-film image receptor and is low in contrast, principally because of Compton scatter radiation. The intensity of scatter radiation is high because of the large area x-ray beam. The image is also degraded because of superimposition all of the anatomical structures in the abdomen. For better visualization of an abdominal structure, such as the kidneys, conventional tomography can be used . In nephrotomography , the renal outline is distinct because the overlying and underlying tissues are blurred. In addition, the contrast

of the in-focus structures has been enhanced. Yet the image remains rather dull and blurred. The latest advance in digital radiography is dradiographics tomosynthesis. This imaging technique uses an area x-ray beam to produce multiple digital images. The images form a three-dimensional data set from which any anatomical plane can be reconstructed. The result is even better image contrast. Conventional tomography is called axial tomography because the plane of the image is parallel to the long axis of the body; this results in sagittal and coronal images. A CT image is a transaxial or transverse image that is perpendicular to the long axis of the body .Coronal and sagittal images can be reconstructed from the transverse image set. The precise method by which a CT imaging system produces a transverse image is extremely complicated, and understanding it requires strong knowledge of physics, engineering, and computer science .The basic principles, however, canbe observed if one considers the simplest of CT imaging systems, which consists of a finely collimated   x-ray beam and a singledetector .

The x-ray source and the detector move synchronously. When   the source-detector assembly makes one sweep, or trans lation, across the patient, the internal structures of the   body attenuate the x-ray beam according to their mass density and effective atomic number . The intensity of radiation detected varies according to this attenuation pattern, and an intensity profile, or projection, is formed . At the end of this translation, the source detector assembly returns to its starting position, and the entire assembly rotates and begins a second translation . During the second translation, the detector   signal again will be proportional to the x-ray beam attenuation of anatomical structures, and a second projection will be described . If this process is repeated many times, a large number of projections are generated. These projections are not displayed visually but are stored in digital form in the computer . Computer   processing of these projections involves effective superimposition of each projection to reconstruct an image of the anatomical

structures within that slice. Superimposition of these projections does not occur as one might imagine . The detector signal during each translation has a dynamic range of 12 bits (4096 gray levels). The value for each increment is   related   to the x-ray   attenuation coefficient of the total path through the tissue. Through the use of simultaneous equations, a matrix of values is   obtained that represents the transverse cross-sectional   anatomy (Stewart, 2013) .

The individual CT slice shows only the parts of the anatomy imaged at a particular level. For example, a scan taken at the level of the sternum would show portions of lung, mediastinum, and ribs, but would not show portions of the kidneys and bladder. Computed tomography requires a firm knowledge of anatomy   in particular the under standing of the location of each organ relative to others.

Each CT slice represents a specific plane in the patient's body. The thickness of the plane is referred to as the Z axis. The Z axis determines the thickness of the slices. The operator selects the thickness of the slice from the choices available on the specific scanner. Selecting a slice thickness limits the x-ray beam so that it passes only through this volume; hence, scatter radiation and superimposition of other structures are greatly diminished. Limiting the x-ray beam in this manner is accomplished by mechanical hardware that resembles small shutters, called collimators, which adjust the opening based on the operator's selection.

In conventional film-screen radiography, the x-ray beam passes through the patient's body and exposes the photographic film. Similarly, in CT, the x-ray beam passes through the patient's body and is recorded by the detectors.

The data that form the CT slice are further sectioned into elements: width is indicated by X, while height is indicated by Y. Each one of these two-dimensional squares is a pixel (picture element). A composite of thousands of pixels creates the CT image that displays on the CT monitor. If the Z axis is taken into account, the

result is a cube, rather than a square. This cube is referred to as a voxel (volume element) ((Stewart, 2013)).

2.5.2 Multislice Helical Computed Tomography Imaging Principles:

Actually, the gantry motion in multislice helical CT is not like a slinky toy; it just appears that way.When the examination begins, the x-ray tube rotates continuously. While the x-ray tube is rotating , the couch moves the patient through the plane of the rotating x-ray beam.

The x-ray tube is energized continuously, data are collected continuously, and an image then can be reconstructed at any desired z-axis position along the patient (Stewart, 2013).

2.5.3 Hounsfield Units :

In CT, we are better able to quantify the beam attenuation capability of a given object. Measurements are expressed in Hounsfield units (HU), named after Godfrey Hounsfield, one of the pioneers in the development of CT. These units are also referred to as CT numbers, or density values.

Hounsfield arbitrarily assigned distilled water the number . He assigned the number 1000 to dense bone and −1000 to air. Objects with a beam attenuation less than that of water have an associated negative number. Conversely, substances with an attenuation greater than that of water have a proportionally positive Hounsfield value. The Hounsfield unit of naturally occurring anatomic structures fall within this range of 1000 to −1000. The Hounsfield unit value is directly related to the linear attenuation coefficient: 1 HU equals a 0.1% difference between the linear attenuation coefficient of the tissue as compared with the linear attenuation coefficient of water (Stewart, 2013).

2.5.4 Filtration :

There are two types of filtration utilized in CT. Mathematical filters such as bone or soft tissue algorithms are included into the CT reconstruction process to enhance resolution of a particular anatomical region of interest. Inherent tube filtration and filters made of aluminiumor Teflon are utilized in CT to shape the beam intensity by filtering out low energy photons that contribute to the production of scatter. Special filters called "bow-tie" filters absorb low energy photons before reaching the patient. X-ray beams are polychromatic in nature which means an X.ray beam contains photons of many different energies.

 Ideally, the X-ray beam should be monochromatic or composed of photons having the same energy. Heavy filtration of the X-ray beam results in a more uniform beam. The more uniform the beam, the more accurate the attenuation values or CT numbers are for the scanned anatomical region. Provides for a equal photon distribution across theX-ray beam. Allows equal beam hardening were the X-ray passes through the filter and object. Lessens overall patient dose by removing softer radiation. Made of aluminium, grafite can be curved, wedge or flat in shape (Karthikeyan 2005).

2.5.5 Detectors:

Detectors gather information by measuring the X-ray attenuation through objects. The most important properties of X-ray detectors used in CT are: a. Efficiency b. Response time (after glow)c. Linearity Efficiency is related to the number of X-rays reaching the detector that are detected. Response time is related to how fast the detected X-ray is converted into an electrical pulse or current.

Linearity is related to the proportionality between the output of the detector and the number of incident X-rays. The two types of detector that have been used for CT are Scintillation detectors and Gas ionization detectors.

2.5.6 Operating Console :

Computed tomography imaging systems can be equipped with two or three consoles. One console is used by the CT radiologic technologist to operate the imaging system. Another console may be available for a technologist to post process images to annotate patient data on the image (e.g., hospital identification, name, patient number, age, gender) and to provide identification for each image (e.g., number, technique, couch position).

This second monitor also allows the operator to view the resulting image before transferring it to the physician's viewing console.

A third console may be available for the physician to view the images and manipulate image contrast, size, and general visual appearance. This is in addition to several remote imaging stations.

The operating console contains meters and controls for selection of proper imaging technique factors, for proper mechanical movement of the gantry and the patient couch, and for the use of computer commands that allow image reconstruction and transfer. The physician's viewing console accepts the reconstructed image from the operator's console and displays it for viewing and diagnosis.

A typical operating console contains controls and monitors for the various technique factors . Operation is usually in excess of 120 kVp, although some recent work supports reducing patient radiation dose by using a lower kVp. The maximum mA is usually 400 mA and is modulated (varied) during imaging according to patient thickness to minimize the patient radiation dose.

The thickness of the tissue slice to be imaged also can be adjusted. Nominal thicknesses are 0.5 to 5 mm. Slice thickness is selected from the console by adjustment of the automatic collimator and by selection of various rows of the detector assembly (Stewart, 2013).

2.5.7 Computer :

Computed tomography uses a computer to process information collected from the passage of x-ray beams through an area of anatomy. The images created are cross-sectional. The computer is a unique subsystem of the CT imaging system. Depending on the image format, as many as 250,000 equations must be solved simultaneously; thus, a large computing capacity is required.

At the heart of the computer used in CT are the microprocessor and the primary memory. These determine the time between the end of imaging and the appearance of an image -the reconstruction time. The efficiency of an examination is influenced greatly by reconstruction time, especially when a large number of image slices are involved (Stewart, 2013).

2.5.8 Gantry :

The gantry includes the x-ray tube, the detector array, the high-voltage generator, the patient support couch, and the mechanical support for each. These subsystems receive electronic commands from the operating console and transmit data to the computer for image production and post processing tasks (Stewart, 2013).

2.5.9 Computed Tomography Quality Contorol :

Computed tomography imaging systems are subject to all the misalignment, miscalibration, and malfunction difficulties of conventional x-ray imaging systems. They have the additional complexities of the multimotional gantry, the interactive console, and the associated computers.

Each of these subsystems increases the risk of drift and instability, which could result in degradation of image quality. Consequently, a dedicated quality control (QC) program is essential for each CT imaging system. Such a program includes daily, weekly, monthly, and annual monitoring in addition to an ongoing preventive maintenance program (Stewart, 2013).

2.6 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. Such estimates of cancer and genetically heritable risk from X-ray exposure have broad 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 A. Mettler, Jr., 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 (abbreviated mSv; 1 mSv =1 mGy in the case of x rays.) 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. Nevertheless, 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 A. Mettler, Jr., 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 that 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 arenas. The effective doses from diagnostic CT procedures are typically estimated to be in the angle of 1 to 10 mSv.

This range is not much less than the lowest doses of 5 to 20mSv 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 encountered in CT, have demonstrated a small but increased radiation-related excess relative risk for cancer mortality.[Fred A et al., 2008].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.[Brenner DJ 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 maybe 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. 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 R. Bruesewitz,2006] .

2.7 General Thoracic Scanning Methods :

Most thoracic protocols are performed while the patient lies in a supine position on the scan table with the arms elevated above the head. In a few instances, primarily high resolution CT protocols of the lungs, additional scans are obtained with the patient in the prone position. Using the shortest scan time possible helps to reduce artifacts created by respiratory motion .

Whenever possible, scans of the chest should be acquired within a single breath-hold, as this will prevent misregistration that may be caused by uneven patient breathing between scans. The thorax has the highest intrinsic natural contrast of any body part. The pulmonary vessels and ribs have significantly different attenuation values compared with the adjacent aerated lung. In most adults, the mediastinal vessels and lymph nodes are surrounded by enough fat to be easily

identified . Because of this intrinsic natural contrast, intravenous (IV) iodinated contrast administration is not necessary for all thoracic indications. For example, scans done for the screening, detection, or exclusion of pulmonary nodules or primary lung diseases such as emphysema or fibrosis are typically done without IV contrast a dministration. The use of IV contrast material is typically requested by the radiologist to differentiate vascular from nonvascular structures, particularly lymph nodes, to evaluate cardiovascular structures by seeing the inside of these structures, and to further characterize lesions by observing their pattern of enhancement. The demarcation of the esophagus and the gastroesophageal junction can be improved by giving the patient an oral contrast agent, most often a barium suspension, shortly before beginning the scan, but is not necessary for most thoracic CT examinations.

2.8 CT Dose Descriptors :

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

• First, the dose distribution inside the patient is completely different from that for a conventional radiogram, where the dose decreases continuously from the entrance of the X-ray beam to its exit, with a 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 distributed in the scanning plane. A dose comparison of CT with conventional projection radiography in terms of skin dose therefore does not make any sense.

• Second, the scanning 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–unlike with conventional projection radiography–is further complicated by the circumstances in which the volume to be imaged is not irradiated simultaneously. This often leads to confusion about what the dose from a complete series of, for example, 15 slices might be compared with the dose from a single slice.

As a consequence, dedicated dose quantities that account for these peculiarities are needed: the 'computed tomography dose index (CTDI)', which is a measure of the local dose, and the 'dose–length product (DLP)', representing the integral radiation exposure associated with a CT examination. Fortunately, a bridge exists that enables comparison of CT with radiation exposure from 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 (BUSHBERG, SEIBERT,Sacramento, JR, 2012).

2.8.1 Computed Tomography Dose Index (CTDI) :

The CTDI is the primary dose measurement concept in CT, Where

$$CTD = 1 \ (1/N)^{\infty}\!\int_{\infty} D \ (z)dz$$

D(z)= the radiation dose profile along the z-axis Where: N is the number of tomographic section imaged in a single axial. This is equal to the number of data channels used in a particular scan T = the width of the tomographic section along the z-axis imaged by one data channel. In multiple-detector-row (multiline) CT scanner, several detector elements may be grouped together to form one data channel. In single-detector-row (single-slice) CT, the z-axis collimation(T) is the nominal scan width. CTDI represents the average absorbed dose, along the z-axis forma series of contiguous irradiations. It is measured form one axial CT scan (one rotation of the x-ray tube), and is calculated by dividing the integrated absorbed dose by the nominal total beam collimation. The CTDI is always measured in the axial scan mode for a single rotation of the x-ray source, and theoretically

estimates the average dose within the central region of scan volume consisting of multiple, contiguous CT scans [Multiple Scan Average Dose (MSAD)] for the case where the scan length is sufficient for the central dose to approach its asymptotic upper limit. The MSAD represents the average dose over a small interval (-1/2, 1/2) about center of the scan length (z=0) for scan interval 1, but requires multiple exposure for its direct measurement. The CTDI offered a more convenient yet nominally equivalent method of estimating this value, and required only a single-scan acquisition, which in the early days of CT ,saved a consider- able amount of time (BUSHBERG, SEIBERT,Sacramento, JR, 2012).

2.8.2 Dose–Length Product (DLP) **:**

(DLP; unit: mGy.cm)'.

DLP takes both the 'intensity' (represented by the CTDIvol) and the extension (represented by the scan length L) of an irradiation into account :

$DLP = CTDI_{vol} \cdot L$

So the DLP increases with the number of slices (correctly: with the length of the irradiated body section), while the dose (i.e., CTDIvol) remains the same regardless of the number of slices or length. 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 (BUSHBERG, SEIBERT, Sacramento, JR, 2012).

2.8.3 Effective Dose :

CTDI and DLP are CT-specific dose descriptors that do not allow for comparisons with radiation exposures from other sources, e.g., 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 irradiation of the body are converted into an equivalent uniform dose to the entire body.

Effective dose E [unit: millisievert (mSv)] according to ICRP 60 (ICRP 1991) is defined as the weighted average of organ dose values HT for a number of specified organs:

$E = \sum w_i . H_{T,I}$

How much a particular organ contributes to calculation of effective dose depends on its relative sensitivity to radiation-induced effects, as represented by the tissue-weighting factor w_i attributed to the organ:

• 0.20 for gonads .

• 0.12 for each of lungs, colon, red bone marrow and stomach wall .

• 0.05 for each of breast, urinary bladder, liver, thyroid and esophagus .

• 0.01 for each of skeleton and skin .

• 0.05 for the 'remainder' .

The 'remainder' consists of a group of additional organs and tissues with a lower sensitivity to radiation-induced effects for which the average dose must be used: small intestine, brain, spleen, muscle tissue, adrenals, kidneys, pancreas, thymus and uterus. The sum of all tissue-weighting factors w_i is equal to 1.

Effective dose, can be assessed in various ways using conversion factors.

For coarse estimates, it is sufficient to multiply the DLP with mean conversion factors, depending on which one of three body regions has been scanned and whether that scan was made in head or body scanning mode:

$E \approx DLP . f_{mean}$

For adults of standard size, the following generic mean conversion factors f_{mean} apply:

1. 0.025 mSv/mGy.cm for the head region .

2. 0.060 mSv/mGy.cm for the neck region, scanned in head mode .

3. 0.100 mSv/mGy.cm for the neck region, scanned in body mode .

4. 0.175 mSv/mGy.cm for the trunk region .

In order to apply Equation , the DLP or at least the CTDIvol and the (gross) scan length L, from which the DLP can be calculated , must be available. If the scanner is not equipped with a dose display, or if a more detailed assessment of effective dose is desired (e.g., to be more specific for the scanned region of the body, to distinguish between males and females, to assess pediatric doses, or to take differences between scanners into account), dedicated CT dose calculation software should be used. These programs make use of more detailed conversion factors and also allow for calculation of organ doses. Currently, five different programs are in general use. They are available either commercially or as freeware and differ significantly in specifications, performance, and price (BUSHBERG, SEIBERT,Sacramento, JR, 2012).

2.8.4 Dose Display :

Newer scanners must be equipped with a dose display. At present, only the display of $CTDI_{vol}$ is mandatory (IEC 2001). However, many scanners already also show the DLP, either just per scan series or both per scan series and per exam. An example with display of CTDIvol and DLP per scan series .

2.8.5 Dose Reduction and Optimization in Computed Tomography of the Chest :

The number of CT examinations performed has increased dramatically, as have the average scanned volume per patient and the number of acquisitions perexamination . The subsequent increase in collective radiation dose has been of concern to radiologists, medical physicists and governmental regulatory authorities and it has been suggested that the radiation dose used for CT was excessive (Rogers 2001).

The radiation dose received by patients undergoing diagnostic radiological examinations by CT is generally in the order of 1–24 mSv per examination for

adults (UNSCEAR 2000) and 2–6.5 mSv for children (Shrimpton et al. 2003). These effective doses can be classified as low even though they are invariably greater than those from conventional radiography. Typically, a chest radiographic examination with two views delivers a dose ranging from 0.08 to 0.30 mSv. In contrast, a standard-dose MDCT delivers 8 mSv, i.e. a 100-fold risk of death by cancer. In other words, one death by cancer is expected every 250,000 chest X-rays and every 2,500 MDCT examinations. Most importantly, more than one-half of the collective radiation dose delivered for diagnostic imaging procedures is due to CT examinations (Golding and Shrimpton 2002). Consequently, particular attention has to be paid to dose optimization and dose reduction, and radiologists and medical physicists should be aware of their responsibility in achieving the appropriate balance between the image quality necessary for diagnostic purposes and the amount of radiation dose delivered to patients (Golding and Shrimpton , 2002). In the rapidly evolving field of MDCT, the quest for the highest image quality supposed to lead to the greatest diagnostic efficacy has obscured possible issues regarding the radiation dose. In this chapter we review the interactions between image quality, diagnostic performances and radiation dose. We specifically focus on clinical advances in dose reduction in chest CT. Although CT is an imaging technique that uses relatively high radiation doses, it should be noted that it has replaced other techniques – such as pulmonary angiography and bronchography – that delivered even higher doses. Nevertheless, a further step in reducing the radiation dose is needed as CT has become the main source of the radiation delivered by medical procedures.

2.9 Routine Chest CT :

The concept of reducing the radiation dose in chest CT was first introduced by (Naidich et al.1990), who reduced the tube current on incremental 10-mm-collimation CT, and demonstrated that with low tube current settings (i.e. 20 mAs),

the image quality is sufficient for assessing the lung parenchyma. While these images are sufficient for assessing lung parenchyma, the increased noise results in marked degradation of the quality of images photographed with mediastinal window settings. Because of this, these authors recommended that such lowdose techniques should be most suitable for children and for screening. As such, these recommendations have been implemented and further studied in lung cancer screening programs (Henschke et al. 1999; Itoh et al. 2000; Swensen et al. 2002). Similar dose reduction strategies have been applied to thin-section CT, in which no significant difference in lung parenchyma structures was detectable between low doses (i.e. 40 mAs) and high doses (i.e. 400 mAs) (Lee et al. 1994; Zwirewich et al. 1991). Although observed differences were not statistically significant, changes in ground glass opacity were difficult to assess at low-dose CT because of the increased noise. Therefore, it was recommended that 200 mAs should be used for initial thin-section CT and lower doses (i.e. 40–100 mAs) for follow-up examinations. An example of a tree in-bud patter demonstrated at 10 mGy (CTDIvol) and 1 mGy is shown in Figure 10.1. The relationship between radiation exposure and image quality at mediastinal and pulmonary window settings has been evaluated on conventional 10-mmcollimation CT images on a single model of CT scanner with mAs settings ranging from 20 to 400 mAs (Mayo et al. 1987). Although this study showed a consistent increase in image quality with radiation dose, no difference in detection of mediastinal and lung abnormalities could be detected. These findings were confirmed on MDCT by Dinkel et al. (2003), who showed that a 90% reduction in dose compared with standard-dose techniques was not associated with impaired detection of suspicious lesions of malignant lymphoma and extra pulmonary tumours.

In order to investigate the effect of dose reduction without scanning patients several times at several dose levels, it is now possible to use computed simulation

of dose reduction by adding random noise to the image obtained at standard dose. In a validation trial, it has been shown that experienced chest radiologists were unable to distinguish CT images obtained with simulated reduced doses from those obtained with really reduced doses (Mayo et al. 1997).

This technique of simulated reduced doses allows investigators to determine the impact of dose reduction on the diagnostic performances without exposing patients to additional radiations and/or several injections of iodinated contrast material .

2.10 Previous Studies :

Mona Taha Idris (2012) studied: Estimation of Radiation Hazards of Computed Tomography Dose in Khartoum State. The purpose of her study was to measure and estimates the patient radiation dose in three different detectors of CT scanners (64 slices ,16 slices and dual slice) for routine CT investigations. A total of 108 patients were examined in this study. The study concluded that: Dual slice scanner delivered the lowest radiation dose while 16 and 64 slice scanners delivered the highest radiation dose. CT dose optimization protocol is not implemented in all departments.

Madan M. Rehani(2010), studied: "Radiation protection in newer imaging technologies" concluded that the Computed tomography (CT) happens to be a common element in most of these technologies. Radiation protection is high on the agenda of manufacturers and researchers and that is becoming a driving force for users and international organizations. The media and thus the public have their own share in increasing the momentum. The slice war seems to be shifting to dose war. Manufacturers are now chasing the target of sub-mSv CT. The era of two digit mSv effective dose for a CT procedure is far from losing ground, although cardiac CT within 5 mSv seems possible.

Aaron Sodickson, et al, (2009) studied: "Recurrent CT, Cumulative Radiation Exposure, and Associated Radiation-induced Cancer Risks from CT of Adults"

concluded to cumulative CT radiation exposure added incrementally to baseline cancer risk in the cohort. While most patients accrue low radiation-induced cancer risks, a subgroup is potentially at higher risk due to recurrent CT imaging.

Rebecca Smith-Bindman, et al, (2009) studied: "Radiation Dose Associated With Common Computed Tomography Examinations and the Associated Lifetime Attributable Risk of Cancer" 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

In his study titled "The Control Of Radiation Exposure From CT Scans" 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 .

CHAPTER THREE
MATERIAL AND METHODES

3.1 MATERIAL :

In this study the data of CT-scan has been collected from Modern Medical center , used to assess doses for 30 adult patients(male and female) underwent chest, CT examinations. The local ethics committees of all participating institutions approved the study protocol. The collected information in regard to:

3.1.1 Patient data :

o Age and gender

3.1.2 CT equipment-specific information :

Devise UID : 1.2.840.113619.6.278 .

Devise name : CT99 .

Device model name : Optima CT520 series .

3.1.3 CT scan parameters :

o kV, mA, rotation time and scan time (spiral mode),

o Scan length (start and end of scan region),

o Number of slices, slice thickness, pitch.

3.2 Method :

3.2.1 Dosimetric calculations :

Monte Carlo software was used to calculate common CT dose descriptors: (i) CT dose index (CTDIw) and volume dose index (CTDIvol) provides an indication of the average absorbed dose in the scanned region, (ii) CT dose –length product (DLP) the integrated absorbed dose along a line parallel to the axis of rotation for

the complete CT examination, and (iii)effective dose (E): a method for comparing patient doses from different diagnostic procedures (Effective dose) .

3.2.1 Monte Carlo Dose Computation :

Most modern x-ray dosimetry is based upon Monte Carlo calculations, with reference to the stochastic nature of gambling in the principality of Monaco. Monte Carlo. Monte Carlo procedures use computer simulation to study the dose deposition during an x-ray exposure. Computer programs can be used to produce random numbers, similar conceptually to flipping a coin, except that the output of the random number generator is usually a continuous variable between 0 and 1. Using these values, any probability distribution can be produced. Using the known physics of x-ray interaction in tissues, the known probabilities of scattering events (both Rayleigh and Compton scattering) as a function of angle and x-ray energy, a geometrical model of the simulated patient, and the random number generator, Monte Carlo calculations compute the three-dimensional dose distribution for a given exposure geometry . The power of modern computers allows Monte Carlo simulations to be performed using many simulated photons, which reduces statistical uncertainty in the computation. Typically, 106 to 108 photons are simulated and their individual trajectories and interactions are tracked by the Monte Carlo program, photon by photon and interaction by interaction. Even though millions of photons (i.e., 106) can be simulated in Monte Carlo programs, the number of simulated photons is still much lower than that which occurs in a physical radiographic acquisition (BUSHBERG, SEIBERT,Sacramento, JR, 2012).