Sudan University of Science and Technology
College of Graduate Studies

Estimation of Radiation Risks to the Brain, Chest, Abdomen and Pelvis during CT Procedures

تقدير مخاطر الإشعاع على الدماغ ، الصدر، البطن والحوض أثناء فحوصات الأشعة المقطعية

Thesis submitted as full fulfillment for the degree of M.Sc. in Diagnostic Radiologic Technology

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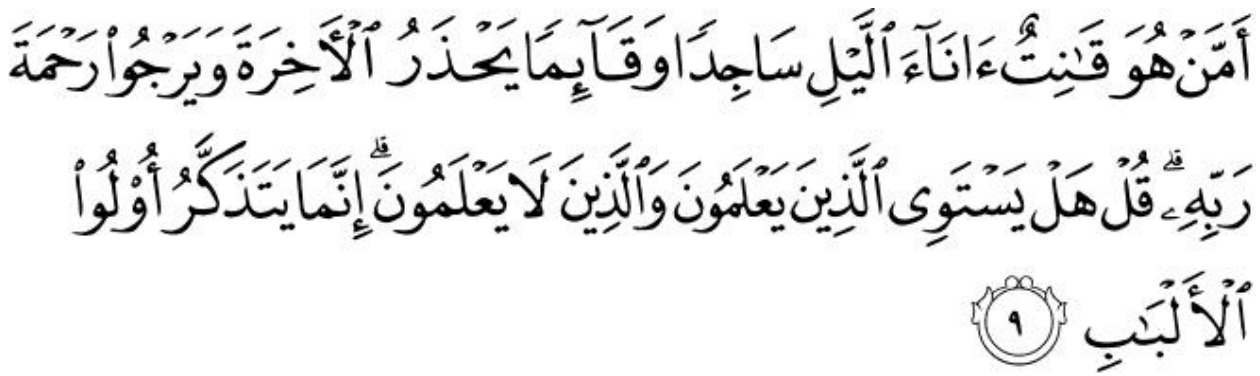
الآية

بسم الله الرحمن الرحيم



سورة الزمر الآية (9)

Dedication

To my colleagues in the college of medical radiological science. And to my friends and all people who save no effort in supporting and assisting me in the completion of this study l dedicate this simple work.

Acknowledgements

First I would like to express gratitude to my supervisor Dr. Abdelmoneim Adam Mohamed Suliman for his support, help and guidance. My deep gratitude is also due to Dr. Mohamed Elfadil, from Sudan University of science and Technology and Dr. Abdelrahman Mohamed Elnour, from Elribat and Alamal hospitals.

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Finally, I would like to gratitude my beloved family.

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List of the abbreviations

CT: computed tomography
ALARA: AS Low as Reasonably A achievable
mSv: milli Sievert
mgy: milli Gray
CTDI: computed tomography Dose Index
DLP: Dose Length Product

Abstract

The data of this study were collected form tow hospital, namely: Alamal hospital And (A)Alribatl hospital(B) in Khartoum State, chiring the period2014-2015. Three X-ray machines were used, namely: Neusoft (16 slice), Siemens Somatom (16 slice) Toshiba Aquilion (64 slice).the general objective of the study was to estimate radiation risks to the Brain ,Chest ,Abdomen and pelvis during CT Procedures. A number of (111) patients were examined. Study showed wide variation of the doses received by the different body organs in terms of DLP and CTDIvol. As for the chest examination, the DLP and CTDIvol value were lower than the value of other examination, whereas The Brain examination values were higher. The Brain examination showed the highest mA values and the chest examination showed the last mA values As for the KVP, the study showed that all the selected examination had the same value i.e (120), except the brain examination which had a value of (115).

The scan time of the brain examination was longer, where as the Chest examination scan time was shorter. Regarding the abdominal examination, the study showed a direct relationship between the DLP and CTDIvol scan time. This relationship was also a plied to the brain, chest and pelvis examination.

The study concluded that the brain examination received the highest radiation dose in terms of DLP and CTDIvol scan time. Special precautions should, there for, be considered during performing such examination.

مستخلص الدراسه

تم جمع بيانات هذة الدراسة من مستشفي الأمل ومستشفي الرباط بولاية الخرطوم خلال الفترة 2014–2015. تم أستخدام ثلاث أجهزة للأشعة المقطعيه وهي نيوسوفت (16 شريحه) وتوشيبا(64 شريحه) وسيمنس (16 شريحة)' وكان الهدف من الدراسة هو تقدير مخاطرالأشعاع على المخ والصدر والبطن والحوض اثناء التصوير الاشعاعى المقطعى.

تم فحص عدد111 مريضاً باستخدام الأجهزة المذكورة أعلاه .

أظهرت الدلراسه اختلافاً كبيراً في الجرعه الاشعاعية لاعضاء الجسم المختلفه من حيث

ففي الصدر كانت قيم DLP وCTDI أقل من قيم الفحوصات الاخرى .وفي فحص الدماغ كانت قيم DLP وCTDI أعلى من قيم الفحوصات الاخرى. وقد سجل فحص الدماغ أعلى قيمة للملى أمبير, وفي مايتعلق بالفولتية أوضحة الدراسة أن جميع الفحوصات المختارة لها نفس القسمة وهي (120)، ماعداً فحص الدماغ الذى كانت قيمة (115). كان زمن فحص الدماغ أطول من زمن الفحوصات الأخرى وزمن فحص الصدر أقل . بخصوص فحص البطن كانت هناك علاقة مباشرة بين DLP و CTDIv وزمن الفحص ، وهذا الأمر ينطبق على فحص الدماغ وفحص الصدر والحوض.

خلصت الدراسة إلى أن فحص الدماغ يتسبب في حدوث الجرعة الاشعاعية الأعلى من حيث ال DLP والـ CTDIv، ولذك ينبغى اتخاذ التحوطات اللازمة عند اجراء مثل هذا الفحص .

Chapter one

Introduction

Chapter One

1.1 Introduction

Technical progress in computed tomography (CT) has substantially increased the clinical efficacy of CT procedures and offered promising new applications in diagnostic imaging. On the other hand, data from various national surveys have confirmed, as a general pattern, the growing impact of CT as a major source of patient and population exposure. From a radiation-hygienic point of view, it is thus necessary to optimize the medical benefit of CT examinations to patients, while strictly controlling and reducing their risk from the radiation exposure. It is the purpose of this thesis to summarize relevant dosimetric concepts for dose estimation in CT, to give an overview on the specific factors determining radiation exposure to patients in MSCT, and to provide suggestions for the optimization of MSCT protocols to balance patient exposure against image quality(Christoph Hoeschen, Dieter ReGulla, Maria Zankl, Helmut Schlattl and Gunnar Brix 2009Tomography refers to the cross-sectional imaging of an object from either transmission or reflection data collected by illuminating the object from many different directions. The impact of this technique in diagnostic medicine has been revolutionary, since it has enabled doctors to view internal organs with unprecedented precision and safety to the patient. The first medical application utilized x-rays for forming images of tissues based on their x-ray attenuation coefficient. More recently, however, medical imaging has also been successfully accomplished with radioisotopes, ultrasound, and magnetic resonance; the imaged parameter being different in each case. There are numerous nonmedical imaging applications. which lend themselves to the methods of computerized tomography. Researchers have already applied this methodology to the mapping of underground resources via cross borehole imaging, some specialized cases of cross-sectional imaging for nondestructive testing, the determination of the brightness distribution over a celestial sphere, and three-dimensional imaging with electron microscopy. Fundamentally, tomographic imaging deals with reconstructing an image from its projections. In the strict sense of the word, a projection at a given angle is the integral of the image in the direction specified by that angle, as illustrated in Fig. 1.1. However, in a loose sense, projection means the information derived from the transmitted energies when an object is illuminated from a particular angle; the phrase “diffracted projection” may be used when energy sources are diffracting, as is

the case with ultrasound and microwaves. Although, from a purely mathematical standpoint, the solution to the problem of how to reconstruct a function from its projections dates back tothe paper by Radon in 1917, the current excitement in tomographic imaging originated with Hounsfield's invention of the x-ray computed tomographic scanner for which he received a Nobel Prize in 1972. He shared the prize with Allan Cormack who independently discovered some of the algorithms. His invention showed that it is possible to compute high-quality cross-sectional images with an accuracy now reaching one part in a thousand in spite of the fact that the projection data do not strictly satisfy the theoretical models underlying the efficiently implementable reconstruction algorithms. His invention also showed that it is possible to process a very large number of measurements (now approaching a million for the case of x-ray tomography) with fairly complex mathematical operations, and still get an image that is incredibly accurate.In1972 in London, Godfrey N. Hounsfield's development of Computed tomography marks the beginning of a new in diagnostic imaging .In 1974 first CT system from a medical equipment manufacture. [Siemens AG. et al.,, 2004].The developments of CT is marked by several developments in scanning Principles detectors as well as the reconstruction mathematics and computers. One major development in CT is the introduction of slip ring scanners that allow Data to be collected faster than conventional scanners.[Euclid seeram., et al., 2008]e radiation doses received by patients undergoing diagnostic radiological examinations by means of computed tomography (CT) are generally in the order of 1–24 mSv(milliSieverts) per examination for adults (UNSCEAR 2000) and 2–6.5 mSv for children (Shrimpton 2003). These effective doses can be classified as low, although they are invariably larger than those observed using conventional diagnostic radiology. The immediate question that comes to mind is whether these low doses carry any risk for the patient. Deleterious health effects induced by ionizing radiation have conventionally been separated into two different categories: deterministic effects and stochastic effects. Exposures to high acute doses in excess of one or two gray (Gy) or sievert (Sv) cause substantial levels of cell killing, which is expressed as organ and tissue damage and, soon after exposure, as deleterious clinical effects. These effects are called deterministic, and the dose–effect relationships exhibit a long threshold dose, with no observable effect, after which the effect increases in severity as the radiation dose increases. The possibility of deterministic health effects, such as radiation sickness, arising after the low doses used with computed tomography can be

dismissed. At lower doses, deleterious health effects, such as cancer or hereditary disease which may take years to be revealed, can occur as a consequence of molecular damage to the nucleus of a single cell. These effects are called stochastic effects, and the probability for their occurrence increases as the dose increases, but the severity of the effect is unrelated to the dose. The potential for stochastic health effects to occur as a result of computed tomography examinations cannot be so easily dismissed because the shape of the dose–effect relationship at low doses is not known. (D. Tack · P. A. Gevenois2007).

The radiation doses received by patients undergoing diagnostic radiological examinations by means of computed tomography (CT) are generally in the order of 1–24 mSv (milliSieverts) per examination for adults (UNSCEAR 2000) and 2–6.5 mSv for children (Shrimpton2003). These effective doses can be classified as low, although they are invariably larger than those observed using conventional diagnostic radiology. The immediate question that comes to mind is whether these low doses carry any risk for the patient. Deleterious health effects induced by ionizing radiation have conventionally been separated into two different categories: deterministic effects and stochastic effects. Exposures to high acute doses in excess of one or two gray (Gy) or Sievert (Sv) cause substantial levels of cell killing, which is expressed as organ and tissue damage and, soon after exposure, as deleterious clinical effects. These effects are called deterministic, and the dose–effect relationships exhibit a long threshold dose, with no observable effect, after which the effect increases in severity as the radiation dose increases. The possibility of deterministic health effects, such as radiation sickness, arising after the low doses used with computed tomography can be dismissed. At lower doses, deleterious health effects, such as cancer or hereditary disease which may take years to be revealed, can occur as a consequence of molecular damage to the nucleus of a single cell. These effects are called stochastic effects, and the probability for their occurrence increases as the dose increases, but the severity of the effect is unrelated to the dose. The potential for stochastic health effects to occur as a result of computed tomography examinations cannot be so easily dismissed because the shape of the dose–effect relationship at low doses is not known(D. Tack · P. A. Gevenois 2007)

1.1.1Scanning Principle :

Computed tomography is an x-ray tomographic technique in which an x-ray beam passes through a thin axial section of the patient from various directions (Fig. 1.1). Parallel collimation is used ton shape the x-ray beam to a thin fan, which defines the thickness of the scan plane. Detectors measure the intensity of the attenuated radiation as it emerges from the body. A mathematical image reconstruction (inverse Radon transformation) calculates the local attenuation at each point within the CT section. These local attenuation coefficients are translated into "CT numbers" and are finally converted into shades of gray that are displayed as an image. With conventional CT scanners the volume of interest is scanned in a sequential fashion, usually proceeding one section at a time. The first two generations of CT scanners (Table 1.2) were superseded in the late 1970s by third- and fourth-generation scanners, which are still in use today. In third -generation scan Comparison of CT scanner generations and types Image Reconstruction 3 Third Fourth Electron beam generation generation scanning Rotation Electron beam deflection Arc (30°_60°) Ring (360°) Semicircular (210°) 1-16 4 256-1000 600-4000 432/864 0.4-10 s 1-5 s ~50 ms Type Second I generation First generation Principle Translation-rotation Detectors Active detector rows Detector elements/row tube and detector array rotate synchronously around the patient. The detector array covers the full width of the fan beam in fourth-generation scanners, the detector elements cover a full circle around the scanner opening and remain stationary during the scan, while only the x-ray tube rotates around the patient. Third-generation scanners, however, offer better scatter suppression and require less detector elements, which is the reason why all multislice CT scanners use third-generation technology. Attempts to speed up the imaging process led to the development of a multi-tube CT scanner called the dynamic spatial reconstructor (the "Mayo monster" equipped with 28 tubes, able to scan up to 240 sections, each of 1 mm thickness, in one 360° rotation), electron beam CT scanning, spiral CT, and, recently, multislice CT. Of these procedures, only spiral and multislice CT have achieved large-scale clinical impact.(spiral and[Radiation protection is governed by three fundamental principles that are designed to establish a level of protection based on what is deemed acceptable (ICRP 2007a). These principles are: justification, optimization of protection, and application of dose limits. In the following, the meaningfulness of these principles in medical X-ray diagnosis–in particular, MSCT–is discussed: Justification: “Any

decision that alters the radiation exposure situation should do more good than harm" (ICRP 2007a). This means that the potential benefits of a CT examination must be balanced against the individual detriment that may be caused by radiation exposure. There must be suffient benefit for the individual patient, considering the efficiency, benefits and risks of available alternative imaging techniques that involve no exposure to ionizing radiation or result in lower patient doses. Optimization of protection: "The likelihood of incurring exposures, the number of people exposed, and the magnitude of their individual doses should all be kept as low as reasonably achievable, taking into account economic and societal factors" (ICRP 2007a).

This means that examinations have to be optimized in order to defied an acceptable balance between patient exposure and necessary diagnostic image quality. Application of Dose Limits–Diagnostic Reference Levels: "The total dose to any individual from regulated sources in planned exposure situations other than medical exposure of patients should not exceed the appropriate limits" (ICRP 2007a): this means that a clearly justified medical examination employing ionizing radiation is not limited by a specific dose value. The explicit exemption of medical exposure from the principle of dose limitation is owed both to the assumption that medical exposures are generally for the benefit of the patient and the perception that medical diagnostic procedures may lead to comparatively high doses to individual patients, e.g., when interventional procedures are considered. However, it is also recognized that the magnitude of patient exposures varies considerably among different radiological departments due to both equipment and skill of the personnel. Therefore, the ICRP recommends in its publication on 'Radiological Protection and Safety in Medicine' (ICRP 1996) the use of Diagnostic Reference Levels (DRLs) for patient examinations as a measure of adequacy of protection. The DRLs apply to an easily measurable operational dose quantity and are intended for use as a simple test for identifying situations where the levels of patient dose are unusually high. If patient doses related to a specific procedure are consistently exceeding the corresponding DRL, there should be a local review of the procedures and equipment. Measures aimed at the reduction of dose levels should be taken, if necessary. The Council of the European Union (1997) has adopted this concept in the Council Directive 97/43/EURATOM. By this means, the member states of the EU are obliged to adopt the DRLs into national legislation and regulations concerning radiation protection in medical diagnostics (for Germany: Bundesamtfür Strahlenschutz 2003).

How to Quantify Radiation Exposure to Patients Related to CT Examinations Fundamental Dose Quantities.(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer

2009)The most comprehensive way to quantify the exposure of a patient undergoing a specific investigation is to determine a dose for each organ. The absorbed dose averaged over an organ is called the organ dose. However, the complexity inherent to a large number of dose values makes it difficult to compare patient doses from different investigations or even different equipment. For such a comparison, it is desirable to have one single value–the effective dose, E. This dosimetric quantity is defied by a weighted sum of organ (or tissue) equivalent doses1 asE = TwTHT(1) where wt is the tissue-weighting factor for tissue T, Htthe equivalent dose of tissue T, and wT= □ (ICRP 1991). The sum is performed over all organs and tissues of the human body considered to be sensitive to the induction of stochastic radiation effects. Those values are chosen to represent the contributions of individual organs and tissues to overall radiation detriment from

1.1.2 Determination of Organ and Tissue Doses:

In general, organ doses cannot be measured directly; they have to be calculated by radiation transport simulations, mostly using Monte Carlo techniques and computational models of the human body. The results of these calculations are so called organ dose conversion confidents, i.e., mean organ doses normalized to a measurable dose quantity, such as the CTDI (see below). In the past, dose estimates have been based upon schematic representations of the human body where the shape of the body and its internal organs are described by relatively simple geometric bodies such as spheres, ellipsoids, elliptical cylinders and parts and combinations thereof (Cristy and Eckerman 1987; Sny Der et al. 1978). Using these "mathematical" models, various radiation protection organizations around the world have simulated X-ray examinations to determine organ dose conversion confidents (Drexler et al. 1990; Hart et al. 1994a, b; Rosenstein 1976, 1992; Stern et al. 1995; Wall 2004). During the last 2 decades, voxel models were introduced that are derived mostly from (whole-body) medical image data of real persons. Examples of voxel models are shown in Fig. 4.1. Typically, they represent realistic models of the human anatomy and offer a clear improvement compared to the mathematical models whose organs are described by relatively simple geometrical bodies. As a consequence, the dose confidents estimated for voxel models deviate system automatically from those calculated for mathematical models (Zankl et al. 2002, Schlattl et al. 2007, and

Winslow etal. 2004).(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009)

1.1.3 Measurable Dose Quantities in CT:

The dosimetric quantities typically used in CT are the "CT dose index" (CTDI) and the "dose length product" (DLP). The CTDI is defied for an axial CT scan (one rotation of the X-ray tube) by dividing the integral of the absorbed dose along the z axis by the nominal beam width. As shown in Fig. 4.2, this value is equivalent to the dose within the nominal width of the slice assuming that the absorbed dose has a rectangular profile with a constant dose inside the nominal width and zero doses outside. The CTDI is measured either free in air (CTDIair) or in a specified phantom made of PMMA. Different phantom sizes are used to reflect differences in body anatomy (M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009). This is mainly realized by different phantom diameters (16-cm diameter for head investigations, 32-cm diameter for body investigations). In practice, CTDI measurements are usually performed with a pencil ionization chamber with an active length of 100 mm, which is positioned at the center (CTDI100,c) and at the periphery (CTDI100,p) of either a standard head or body CT dosimetry phantom. On the assumption that the dose decreases linearly with the radial position from the surface to the center of the phantom, the average dose is given by the "weighted CTDI" (CTDIw) that is a weighted linear combination of the central and peripheral CTDI values: The CTDI is directly proportional to the electrical current-time product (i.e., charge, Qel, in mAs) chosen for the scan; when the CTDI is divided by the Qel value, it is called "normalized CTDI" (nCTDI). CTDIw values have to be measured for all combinations of tube potentials (U in kV) and slice collimations that can be realized at the specific type of scanner, but only for a fied Qel value. It should be noted that the CTDIw is a system specific parameter from which neither a value for a patient dose nor the dose requirements of a system can be deduced directly, without additional knowledge of specific scan parameters, such as collimation and number of rotations. According to the revised IEC standard 60601-2-44, the dose quantity displayed at the operator's console of a CT system is the "volume CTDI."C TDIVol= CTDIwp , (3) where p is the pitch, i.e., the ratio of table feed per gantry rotation and the total beam collimation h. The CTDIvol is the principal dose descriptor in CT, reflecting not only the combined effect of the scan parameters Qel,U, p, and h on the local dose level, but also of scanner specific factors, such as beam filtration, beam-shaping fiter, geometry, and over beaming (see below). The volume CTDI

(CTDIvol) describes the average local dose for the patient within the volume of investigation given in mGy. A better representation of the overall energy delivered by a given scan protocol is the dose-length product (DLP) that is the volume CTDI multiplied with the total scan length, Ltot: D LP = CTDI vol (4) According to the "European Guidelines on Quality Criteria for CT" (European Commission 1999), DRLs for CT examinations are given in terms of CTDIw or CTDIvol and DLP. DRLs valid in Germany for some of the most frequent CT examinations are listed in Table 4.2.(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009)

1.1.4 Determination of the Effective Dose:

from Device and scan Parameters A simple, but coarse estimation of effective dose can be derived from the DLP using representative conversion confidents provided by the ICRP (ICRP 2007b):

where conversion confidents (in mSv · mGy-1 · cm-1), depending on the scanned body region and patient size (respectively age). Some values of k for adult patients are presented in Table 4.3. Alternatively, and based on the above quantities, one can calculate all relevant parameters for dose estimates from the scan parameters and some system-specific components. The basic principle is summarized in Table 4.4. In the fist column the needed or calculated parameters are described; the corresponding symbols are given in column 2, while in column 3 the appropriate units are specified. (To achieve meaningful results, it is indispensable to express the quantities in their correct units. If required, suitable conversions from other units(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009)

1.1.5Advantages and limitations of CT:CT provides a rapid, non-invasive method of assessing patients. A whole body scan can be performed in a few seconds on a modern multi slice scanner with very good anatomical detail. CT is particularly suited to high X-ray contrast structures such as the bones and the lungs, and remains the cross-sectional imaging modality of choice for assessing these. It has less contrast resolution than MRI for soft tissue structures particularly for intracranial imaging, spinal imaging, and musculoskeletal imaging. CT has no major contraindications (although the use of contrast might have), providing the patient can tolerate the scan. The major disadvantage is in the significant radiation doses required for CT. An abdominal or

pelvic CT involves 3–12mSv of radiation, compared with a chest X-ray's 0.02mSv or background radiation in the UK averaging2.5mSv per year.(paul butler, 2007)

1.1.6 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 abroad range of statistical uncertainty, and there is some scientific controversy regarding the effects from very low doses and dose rates as discussed below. Under some rare circumstances of prolonged, high-dose exposure, X. rays can cause other adverse health effects, such as skin erythema (reddening), skin tissue injury, and birth defects following in-utero exposure. But at the exposure levels associated with most medical imaging procedures, including most CT procedures, these other adverse effects would not occur.[Fred 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 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 range of 1 to 10 mSv. This range is not much less than the lowest doses of 5 to 20mSv received by some of the Japaneses urvivors of the atomic bombs. The sesurvivors, 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. Mettler, Jr., 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 feed back 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].

1.1.7 CT doses measurement profile: In conventional radiography, radiation dose decreases continuously from the beam's entrance in to the body to its exit, whereas in CT the dose is distributed more uniformly across the scanning plane because the patient is equally irradiated from all directions. In a head CT examination, for instance, the dose is uniform across the field of view. In larger objects such as the abdomen, the

dose is equally distributed around the periphery of the scanned object and decreases by a factor of only two near the center of the object. Hence, do comp-arisons between CT and conventional radiography in terms of skin dose are not appropriate .Furthermore , the radiation volume , scattered radiation , divergence of the radiation beam , and limits to the efficiency of beam collimation all contribute to the radiation exposure beyond the boundaries of the scan volume. In the case of the multiple scan acquisition required to image some length of a patient‘anatomy ,it becomes essential the effect of the radiation dose delivered beyond the boundaries of a single scan , the radiation dose descriptor known as the CT dose index , or CTDI, integrates the radiation dose delivered both within and beyond the scan volume . The average across the field of view to take into account variations in absorbed dose from the periphery to the center of the object results in a dose descriptor known as the weighted CTDI ,orCTDIw . CTDIw represents the average dose in the scan volume for contiguous CT scan. In the case when there is either a gap or an overlap between sequential scans ,CTDIw must be scaled accordingly , resulting in the dose descriptor volume CTDI, or CTDIvol. CTDIvol represent the average dose within ascan volume (relative to standardizedCT phantom) and is now required to be displayed on the user interface of the CT scanner .[MannudeepR.Ralra et al., 2004].The other commonly shown dose related parameters is the dose length product (DLP). The dose length product is the CTDIvol X length of the scan .Using conversion factors (for neonatal , 1.5.10 and 15 years old) , the DLP can Provide a rough estimate of the age-dependent effective dose for the protocol.[MannudeepR.Ralra et al., 2004].

1.1.8 Importance of the study: The study describes a further inadequacy of the information provided, in that the patients are not made aware that alternative, less harmful imaging techniques are available, notably, in consideration of CT.

1.1.9 The statement of the problem: The understandable excitement that current practicing radiologists experience with the increased imaging capability of modern multi detector CT is therefore not tempered with the direct experience of the harmful effects of excessive radiation exposure. (Jim Giles, 2004)Techniques that employ modern multi detector CT technology are generally performed with the intention of acquiring sufficient data to provide maximal image quality and diagnostic information, but often without enough attention paid to limiting radiation exposure.

1.1.10Objectives of the study:

To estimate Radiation Risks to the Brain, Chest, Abdomen and Pelvis during CT Procedures.

1.1.11 Organization of the study:

- Chapter One: Introduction.

- Chapter Two: Literature Review

- Chapter Three: Materials and Methods.

- Chapter Four: Results.

- Chapter Five: Discussion, Conclusion and Recommendations

- References
- Appendix

Chapter Two

Literatures Review

Theoretical Background

Previous Studies

Chapter Two

Literatures Review

Theoretical Background and Previous Studies

2.1 Theoretical Background

2.1.1 System Components of CT scanner: Three main components are—gantry assembly, computer, operating console.

2.1.1.1 CT Gantry:

The first major component of a CT system is referred to as the scan or imaging system. The imaging system primarily includes the gantry and patient table or couch. The 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 angulation motors and positioning laser lights. In older CT systems a small generator supplied power to the X-ray tube and the rotational components via cables for operation. This type of generator was mounted on the rotational component of the CT system and rotated with the X-ray tube. Some generators remain mounted inside the gantry wall. Some newer scanner designs utilize a generator that is located outside the gantry. Slip ring technology eliminated the need for cables and allows continuous rotation of the gantry components. The inclusion of slip ring technology into a CT system allows for continuous scanning without interference of cables. A CT gantry can be angled up to 30° toward a forward or backward position. Gantry angulation is determined by the manufacturer and varies among CT systems(Karthikeyan 2005). Gantry angulation allows the operator to align pertinent anatomy with the scanning plane. The opening through which a patient passes is referred to as the gantry aperture. Gantry aperture diameters generally range from 50 to 85 cm. Generally, larger gantry aperture diameters, 70 to 85 cm, are necessary for CT departments that do a large volume of biopsy procedures. The larger gantry aperture allows for easier manipulation of biopsy equipment and reduces the risk of injury when scanning the patient and the placement of the biopsy needle simultaneously. The diameter of the gantry aperture is different for the diameter of the scanning circle or scan field of view. If a CT system has a gantry aperture of 70 cm diameter it does not mean that you can acquire patient data

utilizing a 70 cm diameter. Generally, the scanning diameter in which patient or projection data is acquired is less than the size of the gantry aperture. Lasers or high intensity lights are included within or mounted on the gantry. The lasers or high intensity lights serve as anatomical positioning guides that reference the center of the axial, coronal, and sagittal planes.

2.1.1.2 X-Ray Tube, Collimation, Filtration:

X-ray is produced by an X-ray tube. The three main parts of any X-ray tube are the anode, cathode and the filament. When the filament is heated, electrons are ejected from its surface. A large voltage between the cathode and the anode force electrons to accelerat towards the anode. The electrons hitting the anode(tungsten) produce Bremstrahlung radiation at an efficiency of only 1 percent. The other 99 percent of the electrons energy is converted into heat .Most modern system use tubes with two focal spot small spot is used for high resolution examination. And large spot is used for larger anatomic coverage. Stationary anode—Used in eary scanners, oil cooled, large focal spot giving rise to higher potential radiation. Rotating anode—Air cooled, small focal spot requires large heat capacity and fast cooling rates. Mechanical stresses due to tube rotation—Up to 13 G for0.5 second rotation. CT procedures facilitate the use of large exposure factors, (high mA and kVp values) and short exposure times. The development of spiral/helical CT allows continuous scanning while the patient table or couch moves through the gantry aperture(Karthikeyan 2005). A typical spiral/helical CT scan of the abdomen may require the continuous production of X-rays for a 30 to 40 second period. The stress caused by the constant build up of heat can lead to a rapid decrease of tube life. When an X-ray tube reaches a maximum heat value it simply will not operate until it cools down to an acceptable level. CT systems produce X-radiation continuously or in short millisecond bursts or pulses at high mA and kVp values. CT X-ray tubes must possess a high heat capacity which is the amount of heat that a tube can store without operational damage to the tube. The X-ray tube must be designed to absorb high heat levels generated from the high speed rotation of the anode and the bombardment of electrons upon the anode surface. An X-ray tubes heat capacity is expressed in heat units. Modern CT systems utilize X-ray tubes that have a heat capacity of approximately 3.5 to 5 million heat unit (MHU). A CT X-ray tube must possess a high heat dissipation rate. Many CT X-ray tubes utilize a combination of oil and air cooling systems to eliminate heat and maintain continuous operational

capabilities. A CT X-ray tube anode has a large diameter with a graphite backing. The large diameter backed with graphite allows the anode to absorb and dissipate large amounts of heat. The focal spot size of an X-ray tube is determined by the size of the filament and cathode which is determined by the manufacturer. Most X-ray tubes have more than one focal spot size. The use of a small focal spot increases detail but it concentrates heat onto a smaller portion of the anode, therefore, more heat is generated. As previously described, when heat is building up faster than the tube can dissipate it the X-ray tube will not produce X-rays until it has sufficiently cooled. CT tubes utilize a bigger filament than conventional radiography X-ray tubes. The use of a bigger filament increases the size of the effective focal spot. Decreasing the anode or target angle decreases the size of the effective focal spot. Generally, the anode angle of a conventional radiography tube is between12 and 17 degrees. CT tubes employ a target angle approximately between 7 and 10 degrees. The decreased anode or target angle also helps alleviate some of the effects caused by the heel effect. CT can compensate any loss of resolution due the use of larger focal spot sizes by employing resolution enhancement algorithms such as bone or sharp algorithms, targeting techniques, and decreasing section thickness.(Karthikeyan 2005)Collimation: Important Component for Reducing Patient Dose and Improving Image Quality by Reducing Scatter Radiation In CT collimation of the X-ray beam includes tube collimators, a set of pre-patient collimators and post patient or pre-detector collimators. Some CT systems utilize this type of collimation system while other do not. The tube or source collimators are located in then X-ray tube and determine the section thickness that will be utilized for a particular CT scanning procedure. When the CT technologist selects a section thickness he or she is determining tube collimation by narrowing or widening the beam. A second set of collimators located directly below the tube collimators maintain the width of the beam as it travels toward the patient. A final set of collimators called post-patient or pre-detector collimators are located below the patient and above the detector. The primary responsibilities of this set of collimators are to insure proper beam width at the detector and reduce the number of scattered photons that may enter detector. Pre-patient collimation Depends on the focal spot size, Mounted on the tube housing, Creates more parallel beam, Reduces patient dose Pre-detector collimation, Restricts the field of view of detectors, Reduces the scatter radiation on the detector ,Aperture width helps determine the slice thickness ,The X-ray field is filtered to reduce the low energy X-rays which are not useful for imaging

but that increase the radiation dose received by the patient. This process is called collimation. The beam undergoes two-levels of collimation:(1) source collimation, and (2) detector collimation. The source collimator controls the thickness of the tomographic slice (most common thickness are 1, 2,5 or 10 mm).

2.1.1.3 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. (Karthikeyan 2005). 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).

2.1.1.4 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.

2.1.1.4.1Scintillation detectors:

Use solid materials in which the energy of X-rays is converted to light photons. Then, the emitted light is converted into an electrical current by using a photomultiplier tube or a silicon photodiode. The material which produces light when the X-ray energy is absorbed is named scintillator and the combination of a scintillator and the device converting light into a, is named scintillation detector.

Uses a scintillation crystal coupled to a photomultiplier tube to convert light to electrons. Amount of light produced is proportional to the

energy of the absorbed X-rays. Used in older generation of machines Disadvantage is that of after glow. Examples—Sodium iodide, cadmium tungstate, caesium iodide.

2.1.1.4.2Gas ionization detectors:

These are based on the ionization of a gas inside a closed chamber when the X-ray energy is absorbed into a gas. The main

disadvantage is the low efficiency of gas detectors. (Karthikeyan 2005). Favourable Detector Characters, High absorption efficiency. High conversion efficiency. High capture efficiency .High reproducibility and stability.

Ionization chamber that uses xenon or krypton gas. Ionized gas causes electrons to attach to tungsten plates creating electronic signals. Gas that is ionized is proportional to the incidentradiation.100% effective utilization of energy. Ionization chamber—xenon (under pressure) detector

2.1.1.5 Operator Console: Scan Console Technical factors, slice thickness, no of scans, angle of gantry. Initiates scan, record patient data, sets FOV.

Display Console: Used to manipulate post scan data, Post processing work—measurements, MIPS, 3Dformations. Window level and width. Computer The computer processes convert the signal from analog to digital by using a analog to digital convertor. It stores the digital signal during the scan and reconstructs the images after the scan is complete. This reconstruction can be done immediately or later. Data can be manipulated to reconstruct into various planes.

Summary of Processes: The formation of a CT image is a distinct three phase process. The reconstruction phase processes the acquired data and forms a digital image. The scanning phase produces data, but not an image. The visible and displayed analog image (shades of grey) is produced by the digital-to analog conversion phase.

2.1.2 CT generations

Part of understanding an imaging modality involves appreciation of the maturation of that modality. In addition to historical interest, the discussion of the evolution of CT scanners also allow the presentation and reinforcement of several key concepts in CT imaging. CT scanners represent a marriage of diverse technologies, including computer hardware, motor control systems, x-ray detectors, sophisticated reconstruction algorithms, and x-ray tube/generator systems. The first generation of CT scanners employed a rotate translate, pencil beam system. Only two x-ray detectors were used, and they measured the transmission of x-rays through the patient for two different slices. The acquisition of the numerous projections and the multiple rays per projection required that the single detector for each CT slice be physically moved throughout all the necessary positions. This system used parallel ray geometry. Starting at a particular angle, the x-ray tube and detector system translated linearly across the field of view (Fay), acquiring 160 parallel rays across a 24- cm Fay. When the x-ray tube/detector system completed its translation, the whole system was rotated slightly, and then another translation was used to acquire the 160 rays in the next projection. This procedure was repeated until 180 projections were acquired at I-degree intervals. A total of 180 x 160 = 28,800 rays were measured.

2.1.2.1 First-generation (rotate/translate) computed tomography((T).

The x-ray tube and a single detector (per (T slice) translate across the field of view, producing a series of parallel rays. The system then rotates slightly and translates back across the field of view, producing ray measurements at a different angle. This process is repeated at 1-degree intervals over 180 degrees, resulting in the complete (T data set. AB the system translated and measured rays from the thickest part of the head to the area adjacent to the head, a huge change in x-ray flux occurred. The early detector systems could not accommodate this large change in signal, and consequently the patient's head was pressed into a flexible membrane surrounded by a water bath. The water bath acted to bolus the x-rays so that the intensity of the x ray beam outside the patient's head was similar in intensity to that inside the head. The NaI detector also had a significant amount of "afterglow," meaning that the signal from a measurement taken at one period of time decayed slowly and carried over into the next measurement if the measurements were made temporally too close M together. One advantage of the first-generation CT scanner was that it employed pencil beam geometry-only two detectors measured the transmission of x-rays through the patient. The pencil beam allowed very

efficient scatter reduction, because scatter that was deflected away from the pencil ray was not measured by a detector. With regard to scatter rejection, the pencil beam geometry used in first-generation CT scanners was the best. The next incremental improvement to the CT scanner was the incorporation of a linear array of 30 detectors. This increased the utilization of the x-ray beam by 30 times, compared with the single detector used per slice in first-generation systems. A relatively narrow fan angle of 10 degrees was used. In principle, a reduction in scan time of about 30-fold could be expected. However, this reduction time was not realized, because more data (600 rays X 540 views = 324,000 data points) were acquired to improve image quality. The shortest scan time with a second-generation scanner was 18 seconds per slice, 15 times faster than with the first-generation system.

2.1.2.2 Open Beam Geometry:

Pencil beam geometry makes inefficient use of the x ray source, but it provides excellent x-ray scatter rejection. X-rays that are scattered away from the primary pencil beam do not strike the detector and are not measured. Fan beam geometry makes use of a linear x-ray detector and a divergent fan beam of x-rays. X-rays that are scattered in the same plane as the detector can be detected, but x-rays that are scattered out of plane miss the linear detector array and are not detected. Scattered radiation accounts for approximately 5% of the signal in typical fan beam scanners. Open beam geometry, which is used in projection radiography, results in the highest detection of scatter. Depending on the dimensions and the x-ray energy used, open beam geometries can lead to four detected scatter events for every detected primary photon (s1p=4). Incorporating an array of detectors, instead of just two, required the use of a narrow fan beam of radiation. Although a narrow fan beam provides excellent scatter rejection compared with plain film imaging, it does allow more scattered radiation to be detected than was the case with the pencil beam used in first-generation CT. The difference between pencil beam, fan beam, and open beam geometry in terms of scatter detection is illustrated in Fig. 13-6. The translational motion of first- and second-generation CT scanners was a fundamental.

Impediment to fast scanning. At the end of each translation, the motion of the x-ray tube/detector system had to be stopped, the whole system rotated, and the translational motion restarted. The success of CT as a clinical modality in its Infancy gave manufacturers reason to explore more efficient, but more costly, approaches to the scanning geometry .The number of detectors used in third-generation scanners was increased substantially (to more than 800 detectors), and the angle of the fan beam was increased so that the detector array formed an arc wide enough to allow the x-ray beam to interrogate the entire patient (Fig. 13-7). Because detectors and the associated electronics are expensive, this led to more expensive CT scanners. However, spanning the dimensions of the patient with an entire row of detectors eliminated the need for

translational motion. The multiple detectors in the detector array capture the same number of ray measurements in one instant as was required by a complete.

2.1.2.3Third-generation (rotate/rotate) computed tomography:

In this geometry, the x-ray tube and detector array are mechanically attached and rotate together inside the gantry. The detector array is long enough so that the fan angle encompasses the entire width of the patient.

translation in the earlier scanner systems. The mechanically joined x-ray tube and detector array rotate together around the patient without translation. The motion of third-generation CT is "rotate l rotate," referring to the rotation of the x-ray tube and the rotation of the detector array. By elimination of the translational motion, the scan time is reduced substantially. The early third-generation scanners could deliver scan times shorter than 5 seconds. Newer systems have scan times of one half second. The evolution from first- to second- and second- to third-generation scanners involved radical improvement with each step. Developments of the fourth- and fifth-generation scanners led not only to some improvements but also to some compromises in clinical CT images, compared with third-generation scanners. Indeed, rotate/rotate scanners are still as viable today as they were when they were introduced in 1975. The features of third- and fourth-generation CT should be compared by the reader, because each offers some benefits but also some tradeoffs. Third-generation scanners suffered from the significant problem of ring artifacts, and in the late 1970s fourth-generation scanners were designed specifically to address these artifacts. It is never possible to have a large number of detectors in perfect balance with each other, and this was especially true 25 years ago. Each detector and its associated electronics has a certain amount of drift, causing the signal levels from each detector to shift over time. The rotate/rotate geometry of third-generation scanners leads to a situation in which each detector is responsible for the data corresponding to a ring in the image (Fig. 13-8). Detectors toward the center of the detector array provide data in the reconstructed image in a ring that is small in diameter, and more peripheral detectors contribute to larger diameter rings. Third-generation CT uses a fan geometry in which the vertex of the fan is the x-ray focal spot and the rays fan out from the x-ray source to each detector on the detector array. The detectors toward the center of the array make the transmission measurement It, while the reference detector that measures 10 is positioned near the edge of the detector array. If gl is the gain of the reference detector, and Gl is the third-generation geometry in computed tomography, each individual detector gives rise to an annulus (ring) of image information. When a detector becomes mis calibrated, the tainted data can lead to ring artifacts in the reconstructed image. Gain of the other detector, then the transmission measurement is given by the following equation: In(gdo/ gzIt) = ~t The equation is true only if the gain terms cancel each other out, and

that happens when gl = gz. If there is electronic drift in one or both of the detectors, then the gain changes between detectors, so that gl =I:- g2. So, for third-generation scanners, even a slight imbalance between detectors affects the ~t values that are back projected to produce the CT image, causing the ring artifacts.

Fourth-generation CT scanners were designed to overcome the problem of ring artifacts. With fourth-generation scanners, the detectors are removed from the rotating gantry and are placed in a stationary 360-degree ring around the patient (Fig. 13-9), requiring many more detectors. Modern fourth-generation CT systems use about 4,800 individual detectors. Because the x-ray tube rotates and the detectors are stationary, fourth-generation CT is said to use a rotate/stationary geometry. During acquisition with a fourth-generation scanner, the divergent x-ray beam 3600 stationary detector array

2.1.2.4Fourth- generation (rotate/stationary) computed tomography (CT).

The x-ray tube rotates within a complete circular array of detectors, which are stationary. This design requires about six times more individual detectors than a third-generation CT scanner does. At any point during the scan, a divergent fan of x-rays is detected by a group of x-ray detectors.

The fan beam geometry in third-generation computed tomography uses the xray tube as the apex of the fan (source fan). Fourth-generation scanners normalize the data acquired during the scan so that the apex of the fan is an individual detector (detector fan). With third-generation scanners, the detectors near the edge of the detector array measure the reference x-ray beam. With fourth-generation scanners, the reference beam is measured by the same detector used for the transmission measurement. emerging from the x-ray tube forms a fan-shaped x-ray beam. However, the data are processed for fan beam reconstruction with each detector as the vertex of a fan, the rays acquired by each detector being fanned out to different positions of the x-ray source. In the vernacular of CT, third-generation design uses a source fan, whereas fourth-generation uses a detector fan. The third-generation fan data are acquired by the detector array simultaneously, in one instant of time. The fourth-generation fan beam data are acquired by a single detector over the period of time that is required for the x-ray tube to rotate through the arc angle of the fan. This difference is illustrated in Fig. 13-10. With fourth-generation geometry, each detector acts as its own reference detector. For each detector with its own gain, g, the transmission measurement is calculated as follows: Note that the single g term in this equation is guaranteed to cancel out. Therefore, ring artifacts are eliminated in fourth-generation scanners. It should be mentioned, however, that with modern detectors and more sophisticated calibration software, third-generation CT scanners are essentially free of ring artifacts as well.

2.1.2.5 Fifth Generation: Stationary/Stationary:

A novel CT scanner has been developed specifically for cardiac tomographic imaging. This "cine-CT" scanner does not use a conventional x-ray tube; instead, a large arc of tungsten encircles the patient and He is directly opposite to the detector ring. X-rays are produced from the focal track as a high-energy electron beam strikes the tungsten. There are no moving parts to this scanner gantry. The electron beam is produced in a cone-like structure (a vacuum enclosure) behind the gantry and is electronically steered around the patient so that it strikes the annular tungsten target (Fig. 13-11). Cine-CT systems, also called electron beam scanners, are marketed primarily to cardiologists. They are capable of 50-msec scan times and can produce fast-frame-rate CT movies of the beating heart.

Third-generation and fourth-generation CT geometries solved the mechanical inertia limitations involved in acquisition of the individual projection data by eliminating the translation motion used in first- and second-generation scanners. However, the gantry had be stopped after each slice was acquired, because the detectors (in third-generation scanners) and the x-ray tube (in third- and fourth-generation machines) had to be connected by wires to the stationary scanner electronics. The ribbon cable used to connect the third-generation detectors with the electronics had to be carefully rolled out from a cable spool as the gantry rotated, and then as the gantry stopped and began to rotate in the opposite direction the ribbon cable had to be retracted. In the early 1990s, the design of third- and fourth-generation scanners evolved to incorporate slip ring technology. A slip ring is a circular contact with sliding brushes that allows the gantry to rotate continually. The use of slip-ring technology eliminated the inertial limitations at the end of each slice acquisition, and the rotating gantry was free to rotate continuously throughout the entire patient examination. This design made it possible to achieve greater rotational velocities than with systems not using a slip ring, allowing shorter scan times. Helical CT (also inaccurately called spiral CT scanners acquire data while the table is moving; as a result, the x-ray source moves in a helical pattern around the patient being scanned. Helical CT scanners use either third- or fourth-generation slip-ring designs. By avoiding the time required to translate the patient table, the total scan time required to image the patient can be much shorter (e.g., 30 seconds for the entire abdomen). Consequently, helical scanning allows the use of less contrast agent and increases patient throughput. In some instances the entire scan can be performed within a single breath-hold of the patient, avoiding inconsistent levels of inspiration. helical x-ray tube path around patient

With helical computed tomographic scanners, the x-ray tube rotates around the patient while the patient and the table are translated through the gantry. The net effect of these two motions results in the x-ray tube traveling in a helical path around the patient. The

advent of helical scanning has introduced many different considerations for data acquisition. In order to produce reconstructions of planar sections of the patient, the raw data from the helical data set are interpolated to approximate the acquisition of planar reconstruction data. The speed of the table motion relative to the rotation of the CT gantry is a very important consideration, and the pitch is the parameter that describes this relationship (discussed later).

2.1.2.6 Sixth Generation:

This generation essentially combined the principles of the third and fourth generations with the slip ring technology to grate a system that could rotate continually around the patient without being limited by electric wires .Above all, the introduction of the slip ring technology the world of CT permits much shorter acquisition times (i.e, short as 30 second to scan the entire abdomen).The main drawback of helical CT scanners lies in the nature in which the data is collected. Since the date is acquired in a helical formation , no full slices of data are available because the scanner is not producing planer. This problem can be compensated for through reconstruction process.

2.1.2.7 Seventh Generation: Multiple Detector Array:

X-ray tubes designed for CT have impressive heat storage and cooling capabilities, although the instantaneous production of x-rays is constrained by the physics governing x-ray production. An approach to overcoming x-ray tube output limitations is to make better use of the x-rays that are produced by the x-ray tube. When multiple detector arrays are used , the collimator spacing is wider and therefore more of the x-rays that are produced by the x-ray tube are used in producing image data. With conventional, single detector array scanners, opening up the collimator increases the slice thickness, which is good for improving the utilization of the x-ray beam but reduces spatial resolution in the slice thickness dimension. With the introduction of multiple detector arrays, the slice thickness is determined by the detector size and not by the collimator. This represents a major shift in CT technology.. For the same technique (kilovoltage [kV] and mAs), the number of x-rays being detected is four times that of a single detector array with 5-mm collimation. Furthermore, the data set from the 4 X 5 mm multiple detector array can be used to produce true 5-mm slices, or data from adjacent arrays can be added to produce true 10-, 15-, or even 20-mm slices, all from the same acquisition. The flexibility of CT acquisition protocols and increased efficiency resulting from multiple detector array CT scanners allows better patient imaging; however, the number of parameters involved in the CT.

2.1.3 Scanning Principle :

Computed tomography is an x-ray tomographic technique in which an x-ray beam passes through a thin axial section of the patient from various directions (Fig. 1.1).

Parallel collimation is used change the x-ray beam to a thin fan, which defines the thickness of the scan plane. Detectors measure the intensity of the attenuated radiation as it emerges from the body. A mathematical image reconstruction (inverse Radon transformation) calculates the local attenuation at each point within the CT section. These local attenuation coefficients are translated into "CT numbers" and are finally converted into shades of gray that are displayed as an image. With conventional CT scanners the volume of interest is scanned in a sequential fashion, usually proceeding one section at a time. The first two generations of CT scanners (Table 1.2) were superseded in the late 1970s by third- and fourth-generation scanners, which are still in use today. In third -generation scan Comparison of CT scanner generations and types Image Reconstruction 3 Third Fourth Electron beam generation generation scanning Rotation Electron beam deflection Arc (30°_60°) Ring (360°) Semicircular (210°) 1-16 4 256-1000 600-4000 432/864 0.4-10 s 1-5 s ~50 ms Type Second I generation First generation Principle Translation-rotation Detectors Active detector rows Detector elements/row tube and detector array rotate synchronously around the patient. The detector array covers the full width of the fan beam in fourth-generation scanners, the detector elements cover a full circle around the scanner opening and remain stationary during the scan, while only the x-ray tube rotates around the patient. Third-generation scanners, however, offer better scatter suppression and require less detector elements, which is the reason why all multislice CT scanners use third-generation technology. Attempts to speed up the imaging process led to the development of a multi-tube CT scanner called the dynamic spatial reconstructor (the "Mayo monster" equipped with 28 tubes, able to scan up to 240 sections, each of 1 mm thickness, in one 360° rotation), electron beam CT scanning, spiral CT, and, recently, multislice CT. Of these procedures, only spiral and multislice CT have achieved large-scale clinical impact.(spiral and[Radiation protection is governed by three fundamental principles that are designed to establish a level of protection based on what is deemed acceptable (ICRP 2007a). These principles are: justification, optimization of protection, and application of dose limits. In the following, the meaningfulness of these principles in medical X-ray diagnosis–in particular, MSCT–is discussed: Justification: “Any decision that alters the radiation exposure situation should do more good than harm” (ICRP 2007a). This means that the potential benefits of a CT examination must be balanced against the individual detriment that may be caused by radiation exposure. There must be sufficient benefit for the individual patient, considering the efficiency, benefits and risks of available alternative imaging techniques that involve no exposure to ionizing radiation or result in lower patient doses. Optimization of protection: “The likelihood of incurring exposures, the number of people exposed, and the magnitude of their individual doses should all be kept as low as reasonably achievable, taking into account economic and societal factors” (ICRP 2007a).

This means that examinations have to be optimized in order to define an acceptable balance between patient exposure and necessary diagnostic image quality. Application of Dose Limits–Diagnostic Reference Levels: “The total dose to any individual from regulated sources in planned exposure situations other than medical exposure of patients should not exceed the appropriate limits” (ICRP 2007a): this means that a clearly justified medical examination employing ionizing radiation is not limited by a specific dose value. The explicit exemption of medical exposure from the principle of dose limitation is owed both to the assumption that medical exposures are generally for the benefit of the patient and the perception that medical diagnostic procedures may lead to comparatively high doses to individual patients, e.g., when interventional procedures are considered. However, it is also recognized that the magnitude of patient exposures varies considerably among different radiological departments due to both equipment and skill of the personnel. Therefore, the ICRP recommends in its publication on ‘Radiological Protection and Safety in Medicine’ (ICRP 1996) the use of Diagnostic Reference Levels (DRLs) for patient examinations as a measure of adequacy of protection. The DRLs apply to an easily measurable operational dose quantity and are intended for use as a simple test for identifying situations where the levels of patient dose are unusually high. If patient doses related to a specific procedure are consistently exceeding the corresponding DRL, there should be a local review of the procedures and equipment. Measures aimed at the reduction of dose levels should be taken, if necessary. The Council of the European Union (1997) has adopted this concept in the Council Directive 97/43/EURATOM. By this means, the member states of the EU are obliged to adopt the DRLs into national legislation and regulations concerning radiation protection in medical diagnostics (Bundesamtfür Strahlenschutz 2003).

How to Quantify Radiation Exposure to Patients Related to CT Examinations Fundamental Dose Quantities.(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009)The most comprehensive way to quantify the exposure of a patient undergoing a specific investigation is to determine a dose for each organ. The absorbed dose averaged over an organ is called the organ dose. However, the complexity inherent to a large number of dose values makes it difficult to compare patient doses from different investigations or even different equipment. For such a comparison, it is desirable to have one single value–the effective dose, E. This dosimetric quantity is defied by a

weighted sum of organ (or tissue) equivalent doses1 asE = TwTHT(1) where wt is the tissue-weighting factor for tissue T, Htthe equivalent dose of tissue T, and wT= □ (ICRP 1991). The sum is performed over all organs and tissues of the human body considered to be sensitive to the induction of stochastic radiation effects. Those values are chosen to represent the contributions of individual organs and tissues to overall radiation detriment from

2.1.4 Determination of Organ and Tissue Doses:

In general, organ doses cannot be measured directly; they have to be calculated by radiation transport simulations, mostly using Monte Carlo techniques and computational models of the human body. The results of these calculations are so-called organ dose conversion confidents, i.e., mean organ doses normalized to a measurable dose quantity, such as the CTDI (see below). In the past, dose estimates have been based upon schematic representations of the human body where the shape of the body and its internal organs are described by relatively simple geometric bodies such as spheres, ellipsoids, elliptical cylinders and parts and combinations thereof (Cristy and Eckerman 1987; SnyDer et al. 1978). Using these "mathematical" models, various radiation protection organizations around the world have simulated X-ray examinations to determine organ dose conversion confidents (Drexler et al. 1990; Hart et al. 1994a, b; Rosenstein 1976, 1992; Stern et al. 1995; Wall 2004). During the last 2 decades, voxel models were introduced that are derived mostly from (whole-body) medical image data of real persons. Examples of voxel models are shown in Fig. 4.1. Typically, they represent realistic models of the human anatomy and offer a clear improvement compared to the mathematical models whose organs are described by relatively simple geometrical bodies. As a consequence, the dose confidents estimated for voxel models deviate system atically from those calculated for mathematical models (Zankl et al. 2002, Schlattl et al. 2007, and Winslow et al. 2004).(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009)

2.1.5 Measurable Dose Quantities in CT:

The dosimetric quantities typically used in CT are the "CT dose index" (CTDI) and the "dose length product" (DLP). The CTDI is defied for an axial CT scan (one rotation of the X-ray tube) by dividing the integral of the absorbed dose along the z axis by the

nominal beam width. As shown in Fig. 4.2, this value is equivalent to the dose within the nominal width of the slice assuming that the absorbed dose has a rectangular profile with a constant dose inside the nominal width and zero doses out side. The CTDI is measured either free in air (CTDIair) or in a specified phantom made of PMMA. Different phantom sizes are used to reflect differences in body anatomy (M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009). This is mainly realized by different phantom diameters (16-cm diameter for head investigations, 32-cm diameter for body investigations). In practice, CTDI measurements are usually performed with a pencil ionization chamber with an active length of 100 mm, which is positioned at the center (CTDI100,c) and at the periphery (CTDI100,p) of either a standard head or body CT dosimetry phantom. On the assumption that the dose decreases linearly with the radial position from the surface to the center of the phantom, the average dose is given by the "weighted CTDI" (CTDIw) that is a weighted linear combination of the central and peripheral CTDI values.

The CTDI is directly proportional to the electrical current-time product (i.e., charge, Qel, in mAs) chosen for the scan; when the CTDI is divided by the Qel value, it is called "normalized CTDI" (nCTDI). CTDIw values have to be measured for all combinations of tube potentials (U in kV) and slice collimations that can be realized at the specific type of scanner, but only for a field Qel value. It should be noted that the CTDIw is a system specific parameter from which neither a value for a patient dose nor the dose requirements of a system can be deduced directly, without additional knowledge of specific scan parameters, such as collimation and number of rotations. According to the revised IEC standard 60601-2-44, the dose quantity displayed at the operator's console of a CT system is the "volume CTDI."C TDIVol= CTDIwp , (3) where p is the pitch, i.e., the ratio of table feed per gantry rotation and the total beam collimation h. The CTDIvol is the principal dose descriptor in CT, reflecting not only the combined effect of the scan parameters Qel,U, p, and h on the local dose level, but also of scanner specific factors, such as beam filtration, beam-shaping filter, geometry, and over beaming (see below). The volume CTDI (CTDIvol) describes the average local dose for the patient within the volume of investigation given in mGy. A better representation of the overall energy delivered by a given scan protocol is the dose-length product (DLP) that is the volume CTDI multiplied with the total scan length, Ltot:D LP = CTDIVol ċL tot(4) According to the "European Guidelines on Quality

Criteria for CT" (European Commission 1999), DRLs for CT examinations are given in terms of CTDIw or CTDIvol and DLP. DRLs valid in Germany for some of the most frequent CT examinations are listed in Table 4.2.(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009)

2.1.6 Determination of the Effective Dose:

From Device and scan Parameters a simple, but coarse estimation of effective dose can be derived from the DLP using representative conversion confidents provided by the ICRP (ICRP 2007b):

where conversion confidents (in mSv · mGy-1 · cm-1), depending on the scanned body region and patient size (respectively age). Some values of k for adult patients are presented in Table 4.3. Alternatively, and based on the above quantities, one can calculate all relevant parameters for dose estimates from the scan parameters and some system-specific components. The basic principle is summarized in Table 4.4. In the fist column the needed or calculated parameters are described; the corresponding symbols are given in column 2, while in column 3 the appropriate units are specified. (To achieve meaningful results, it is indispensable to express the quantities in their correct units. If required, suitable conversions from other units (M. F. Reiser et al 2009)

2.1.7Advantages and limitations of CT:

CT provides a rapid, non-invasive method of assessing patients. A whole body scan can be performed in a few seconds on a modern multislice scanner with very good anatomical detail. CT is particularly suited to high X-ray contrast structures such as the bones and the lungs, and remains the cross-sectional imaging modality of choice for assessing these. It has less contrast resolution than MRI for soft tissue structures particularly for intracranial imaging, spinal imaging, and musculoskeletal imaging. CT has no major contraindications (although the use of contrast might have), providing the patient can tolerate the scan. The major disadvantage is in the significant radiation doses required for CT. An abdominal or pelvic CT involves 3–12mSv of radiation, compared with a chest X-ray's 0.02mSv or background radiation in the UK averaging2.5mSv per year.(PAUL BUTLER, ADAM W. M. MITCHELL and HAROLD ELLIS 2007) Applied Radiological Anatomy for Medical Students.

2.1.8 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 abroad range of statistical uncertainty, and there is some scientific controversy regarding the effects from very low doses and dose rates as discussed below. Under some rare circumstances of prolonged, high-dose exposure, X. rays can cause other adverse health effects, such as skin erythema (reddening), skin tissue injury, and birth defects following in-utero exposure. But at the exposure levels associated with most medical imaging procedures, including most CT procedures, these other adverse effects would not occur.[Fred 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]

2.1.9Image Reconstruction:

The computer receives a signal in analog form and converts it to a binary digit by using a analog to digital convertor. The digital signal is stored and the image is reconstructed after the scan is over. Each picture is displayed on a matrix, each square in a matrix is called a pixel, its assigned a number based on the amount of energy reaching the detector. This number is called as Hounsfield unit.

The reconstructed anatomy of an object is in the digital format composed of a large number of Tiny elongated blocks. Representing a volume of tissue called voxel Voxel—3D tissue element that has a width, height and depth. Depth of a voxel is a important parameter which depends on the slice thickness and each unit is assigned a shade of grey. Pixel is the 2D projection of a voxel on the computer screen and it has only height and width.

2.1.10 CT Numbers:

CT Numbers and Hounsfield Units The digital value ascribed to each pixel is called the Hounsfield units or HU, which lies on a scale were water has a value of 0 and air has a value of –1000.Bone has a value in order of +1000. HU values reflect

Pixel values for some biological tissues are: Tissue CT number range in HU: Air –1000Lungs –900 to –300Fat –120 to –80Water 0Muscle 10 to 30Soft tissue 10 to 30Cortical bone 50 to 100Trabecular bone 500 to 1000

Algorithms for Image Reconstruction: The fundamental problem in CT is to calculate the linear attenuation coefficient of the pixels using a large number of X-ray transmission measurements and to use the results to build up an image of the object by means of computer processing algorithm an algorithm is a mathematical method for solving a problem. Various methods are used for reforming the image.1. Back projection method2.Iterative method3. Analytical method.

Back Projection Method: Simplest method also known as linear or summation method. Involves obtaining profiles of an object and then combining them. Does not produce sharp images. Analytical Reconstruction Algorithm. Commonly used now because of their speed.

• It is a filtered back projection method –no stair artifacts. Data is reconstructed using a fourier transform.

Windowing and Grey Scale: Technique of windowing is a electronic manipulation

of the data to enable the shades of grey to be used to represent a limited range of HU values so that different structures can be imaged. CT scans are displayed as a monochrome image on a TV screen. The value of the pixel at a specific point in the image is converted to a grey level. However, the range of pixel values is

approximately–1000 (air) to +800 (dense bone) and the eye can only distinguish 32 grey levels at best. The majority of the soft tissues range from -100 to +100 so a system know as windowing has been developed to allow Radiologists to dynamically view images. Decreasing the window width increases the contrast in the image so is good for looking at differences in soft tissues. Increasing the window width allows structures with a large pixel range (i.e. bones and lungs) to be viewed. Decreasing the window level allows the lungs and other airways to be viewed. Increasing the window level allows the denser bones to be viewed.

Windowing allows you to dynamically alter the image. Film hard copies are taken at specific (user defined) window settings so are just a representative copy of the original image. Good diagnostic practices is to have access to the images on a diagnostic console to allow windowing to be performed as required.

Image Quality: Spatial resolution. Contrast resolution. Noise and spatial uniformity. Linearity. Image artifacts. Image noise and artifacts are the two biggest enemies of CT image quality. CT parameters can be manipulated to either decrease or eliminate the adverse effects of these image. Spatial Resolution

It is the CT system's ability to differentiate small objects that are adjacent to one another. The CT scanner's resolving power relies on how well small objects that are close together but have very different attenuation values or CT numbers are imaged (Karthikeyan 2005). There are parameters that a CT technologist can manipulate to increase the spatial resolution when scanning high frequency regions. Utilizing a bone, sharp, high frequency or high pass algorithm during reconstruction can improve the spatial resolution. Other factors that influence spatial resolution include pixel size, which is influenced by the chosen, scanned field of view and matrix size, width of the detector, spacing between detectors, number of projections or views obtained and focal spot size.

Field of View (FOV): Diameter of FOV has a proved effect on the image quality. FOV should be adjusted to the size of the anatomic areas to be examined. Ideally pixels should be smaller than the minimum ,distance resoluble by the scanner. Contrast Resolution: It is the ability of a CT scanner to differentiate small

Attenuation differences on the CT image. Contrast resolution is also known as Low Contrast Resolution and Tissue Resolution. Contrast resolution is limited by noise, as noise in an image increases, contrast resolution decreases thereby, inhibiting the ability of the CT scanner to image slight differences in tissue density. A soft tissue, standard or smooth algorithm is used during the reconstruction process to enhance soft tissue and contrast resolution.

Image Noise: Noise is considered to be the number one limiting factor of CT image quality. Noise is the portion of a signal that contains no information. Noise is characterized by a grainy appearance of the image. The major types of noise include quantum noise, electronic noise and computational noise. Quantum noise is a result of too few photons reaching a detector after being attenuated by the body. Any factor that limits the number of attenuated photons at the detector will increase image noise. Anatomical structure size, reduction of slice thickness without increasing technical factors, decreasing pixel size and scatter radiation are all factors that contribute to image noise. Electronic noise is noise contained within the image that can be caused by vibrations of any of the physical components, especially the rotational components or power fluctuations. Computational noise is primarily caused by all the statistical fluctuations that occur from the reconstruction mathematics that are essential to produce a CT image. The following factors influence image noise: Voxel size, slice thickness, matrix, FOV. Filter algorithm, mAs. Image Artifacts: "An artifact is any distortion or error in the image that is unrelated to the subject being studied."Artifacts can appear as geometrical inconsistencies, blurring, streaks or inaccurate CT numbers. Streak artifacts are the most common distortions or errors that affect the quality of CT images. Motion, metallic objects, out-of-field, edge gradient effects, high-low frequency interfaces, equipment malfunctions and sampling errors are all causes of streak artifacts. Equipment malfunctions such as tube arching, electrical malfunctions and detector malfunctions produce streak artifacts on a CT image. Source of Artifacts, Data formation ,Patient motion. Polychromatic effects. Equipment misalignment. Faulty X-ray source. **(**Karthikeyan 2005).

CT Image Parameters—Definitions and Ranges Noise All CT images contain noise or pixel to pixel variations in the pixel value in the image of an object of uniform linear attenuation coefficient. It is measured as the standard deviation (σ) of the pixel values within an area of the image. The units are HU and it typically ranges from 3 to 30HU

.Spatial This is quoted either as the full width at half resolution in maximum (FWHM) of the point spread function the image plane (PSF) measured in mm or as the spatial frequency of the 50% modulation transfer function (MTF) measured in cycles/cm. (Note percentages other than the 50% can be quoted. A simple method of converting is: FWHM of PSF in mm = 50% of MTF in cycles/cm The typical range of the of the PSF is from0.75 to 2.0 mm. Spatial resolution This is the slice width. The image is the "average" in the long axis across the slice width so in general the spatial resolution is up to an order of magnitude poorer in this direction. This also gives rise to the partial volume effect (see later).Dose CT doses are measured in mGy and range upwards from about 50 mGy. CT has one of the highest dose consequences to the patient. Data acquisition , Slice geometry .Profile sampling. Angular sampling. Data measurement ,Detector imbalance. Scatter collimation , Data processing Algorithm effects. Because the patient moves continuously through the Gantry for a 360° rotation, the reconstructed image will be blurred with only the same filtered backprojection algorithm as conventional CT. That's why we should interpolate our image data before the filtered back-projection is used. This process leads toa higher noise level and artifacts such as stair-step artifact.

Patient Motion Artifacts ,Motion can be voluntary or involuntary. No matte which kind of motion we are dealing with, the most efficient way to reduce motion artifact is to reduce our scanning time. Methods to reduce patient motion artifacts include patient immobilization, ECG gated CT, and some correction algorithms. Metal Artifacts (Metallic materials such as prothetic devices, dental fillings, surgical clips, and electrodes produce streak artifacts on the image. Several methods have been provided to remove the artifacts coming from metal.

Beam Hardening Artifacts Beam hardening is a phenomenon results from theincrease of mean energy of the X-ray beam when it the exposure to radiation of patients undergoing computed tomography (CT) examinations is determined by two factors: equipment-related factors, i.e., design of the scanner with respect to dose efficiency, and application-related factors, i.e., the way in which the radiologist or the radiographer makes use of the scanner. In this chapter, the features and parameters influencing patient dose are outlined. First, however, a brief introduction on the dose

descriptors applicable to CT is given. 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 (Tack.Gevenois2007).

2.1.5Computed Tomography Dose Index:

The 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 along a line that is parallel to the axis of rotation for the scanner (= z-axis) and is recorded for a single rotation of the X-ray source.

2.1.11 Application-Related Factors:

Although the scanner design is of some importance, surveys on CT practice have regularly shown that the way that the scanner is used has the largest impact on the doses applied in a CT examination. The application-related factors on which patient exposure depends can be grouped into:• Scan parameters, i.e., those factors that directly determine the local dose level (CTDIvol) and that are often pre-installed or recommended by the manufacturer (e.g., in application guides) • Examination parameter, i.e., those factors that–in combination with CTDI vol–determine the integral exposure (i.e., DLP) and depend on the preferences of the user• Reconstruction and viewing parameters, which implicitly influence the dose settings First, however, the principal inter-dependence between dose settings and image quality shall be outlined(D. Tack · P. A. Gevenois (2007)

2.1.12 Scan Parameters

2.1.12.1 Tube Current–Time Product (Q):

As in conventional radiology, a linear relationship exists between the tube current–time product and dose; i.e., all dose quantities will change by the same amount as the applied mAs. The mAs product Q for a single sequential scan is obtained by multiplying the tube current I and exposure time t; in spiral scanning mode, Q is the product of the tube current I and rotation time trot .This should not be mixed up with the total mAs product of the scan which is the product of tube current I and (total) scan time T. The consequences on image quality resulting from variations in the tube current–time product are relatively simple to understand. The only aspect of image quality so affected is image noise, which is–as indicated in Equation 4.18–inversely proportional to the square root of dose (i.e., mAs). The tube current–time product is often used as a surrogate for the patient dose (i.e., CTDI). However, this is highly misleading, as the normalized CTDI values and thus the dose that results for the same mAs setting can vary by up to a factor of six between different scanners. So it makes absolutely no sense to communicate dose information or recommendations on the basis of mAs. Instead, only CTDI vol (and DLP) should be used for this purpose. With the advent of multi-slice scanners, additional confusion arose due to the introduction of a different, pitch-corrected mAs notation (‘effective mAs’ or ‘mAs per slice’, Eq. 4.15) by Elscint, Philips and Siemens. As most multi-slice scanners make use of a multi-

point spiral interpolation scheme as outlined in section 4.2.6, effective mAs is the most appropriate notation for MSCT. Nevertheless, General Electric and Toshiba still prefer the traditional electrical mAs notation, which further makes it difficult to compare mAs settings among different scanners. This particularly holds for cardiac CT where very low pitch settings are used.(D. Tack · P. A. Gevenois (2007)

The settings for the tube current–time product should be adapted to the characteristics of the scanner, the size of the patient (see section 4.3.2.5), and the dose requirements of each type of examination.

Examinations with high inherent contrast, such as for chest or skeleton, that are characterized by viewing with wide window settings, can regularly be conducted at significantly reduced mAs settings. (D. Tack · P. A. Gevenois (2007)

2.1.12.2 Tube Potential (U):

When the tube potential is increased, both the tube output and the penetrating power of the beam are improved, while image contrast is adversely affected. In conventional projection radiography, increased tube potentials are applied in order to ensure short exposure times for obese patients, to equalize large differences in object transmission (e.g., during chest examinations) or to reduce patient dose. In the latter case, automatic exposure control devices guarantees that the improved penetrating power of the beam is exclusively for the benefit of the patient. In CT, increased tube voltages are used preferentially for improvements in tube loading and image quality. Contrary to the case for mAs, the consequences of variations in kV cannot easily be assessed. The relationship between dose and tube potential U is not linear, but rather of an exponential nature which varies according to the specific circumstances. The intensity of the radiation beam at the detector array, for example, varies with U to the power of 3.5.

If the tube potential is increased, e.g., from 120 kV to 140 kV, the electrical signal obtained from the detectors therefore changes by a factor 1.7 .The decrease in primary contrast which normally results from this action is largely overcompensated by the associated decrease in noise, i.e., the higher the tube potential, the better the CNR (except for the application of iodine as contrast agent). The only reason why this analysis generally holds true is the absence of any kind of automatic exposure control

devices in the majority of scanners which might prevent unnecessary increases in the detector signal. This clearly demonstrates that dose is not reduced by applying higher kilovolt settings, but merely increased as long as mAs settings are not changed: weighted CTDI and effective dose increase with U to the power of 2.5 (Fig. 4.28), which means that both are increased by approximately 50% if kilovolt settings are changed from 120 kV to 140 kV Therefore the question of whether and when it might be reasonable to deviate from the 120-kV setting usually applied is justified. As can be seen from Figure 4.29, this depends on the attenuation characteristics of the detail that is diagnostically relevant. The figures are given in terms of contrast-to-noise ratio squared (CNR2) at constant patient dose; this notation allows direct conversion of the percentage differences into dose differences. For soft tissue contrast (e.g., differences in tissue density), higher tube potentials perform slightly better than lower ones, but the differences are quite small. The opposite holds true for bone contrast (i.e., bone versus tissue). For iodine contrast, however, there is a strong dependence on tube potential that is much in favor of lower kilovolt settings. Thus, 80 kV instead of 120 kV would allow reduction of the patient dose by almost a factor of two without sacrificing image quality . (D. Tack ·P. A. Gevenois2007)
Tube potentials other than 120 kV should be considered only in the case of:• Obese patients in whom mAs cannot be further increased: use higher kilovolt settings• Slim patients and pediatric CT, where mAs cannot be further reduced: use lower kilovolt settings• CT angiography with iodine: use lower kilovolt settings Variations in tube potential should not be considered for pure dose reduction purposes except in the case of CT angiography. Due to the complexity involved, adaptation of mAs settings should not be left to AEC systems, as these do not account for changes in contrast. Dose settings in CT angiography should not be higher than in unenhanced scans of the same body section and should be lowered if performed at reduced kilovolt settings.

(D. Tack · P. A. Gevenois 2007)

2.1.12.3 Slice Collimation and Slice Thickness :

With single-slice CT (SSCT), the slice collimation used for data acquisition and the reconstructed slice thickness used for viewing purposes were identical (except for slice profile broadening in spiral scans with increased pitch, as discussed in section 4.2.6). So there was no need to distinguish between them. With MSCT, the slice collimation (e.g., 0.75 mm) and the reconstructed slice thickness (e.g., 5 mm) are usually different.

Frequently, the selection of the reconstructed slice thickness is made with respect to multi-planar reformatting (MPR) purposes (e.g., 1 mm), thus creating a socalled ‘secondary raw data set’, i.e., a stack of thin slices from which MPR slabs with larger thickness (e.g., 5 mm) can be made for viewing purposes. The ability to acquire longer body sections with thin slices in order to achieve an almost isotropic spatial resolution is the most important achievement of multi-slice technology. As reduced slice thickness is associated with increased image noise, this may have a significant impact on patient dose as expressed by the Brooks’ formula (Eq. 4.18). Therefore, it is worthwhile to treat this matter in a somewhat more detailed fashion. A narrow slice collimation is a precondition for a narrow slice thickness, but its impact on patient dose is restricted to aspects of over beaming and over ranging only. As these show, opposed dependence on beam width, as outlined in sections 4.2.3 and 4.2.8, the question arises as to the optimized beam-width settings. As demonstrated in Figure 4.30 for a typical MSCT scanner, dose performance is almost equal with beam-width settings greater than 10 mm (a), except at short scan ranges (spine, pediatrics) where a beam width of between 10 mm and 20 mm is more. (D. Tack · P. A. Gevenois (2007)appropriate (b). Beam-width settings below 10 mm should be avoided due to increased over beaming effects unless there are other important aspects to justify overriding this recommendation. The decisive determinant with respect to image noise and its implications for patient dose, however, is the slice thickness h rec used for viewing purposes. The relationship among slice thickness, noise and dose expressed in the Brooks’ formula attempts to correct any reduction in slice thickness by a corresponding increase in dose to ensure a constant image noise, and some AEC systems exactly do so. However, any variation in slice thickness also affects image contrast due to a modification in partial volume effect, which is not taken into account by the Brooks’ formula. image noise and image contrast of small details will react in a different fashion on reduction of the slice thickness: while image quality in terms of noise is impaired proportionally to the square root of the change in slice thickness only, the contrast is improved in proportion to the slice thickness. As a result, there is gain in image quality in terms of CNR without any increase in dose whenever partial volume effect is of importance. This is clearly demonstrated by the clinical example given in, where the visibility of a liver lesion (approximately 3 mm in size) diminishes continually with increasing slice thickness, despite reduced image noise. In addition, a detailed analysis of the results of the German survey on CT practice in 1999

(Galanskiet al. 2001) has revealed that slice thickness has only minor or no influence on clinical dose settings. for liver examinations with slice thicknesses of between 3 mm and 10 mm that were used in practice. Therefore, it is essential to understand that the selection of a narrow slice collimation is only a means to an end: to enable MPR images without or with reduced step artifacts, and, if necessary, to overcome partial volume effects..(D. Tack · P. A. Gevenois (2007)

The slice collimation should be selected as small as compatible with aspects of over beaming/over ranging, total scan time and tube power. Viewing should preferentially be made with thicker slabs (e.g., 3–8 mm), thereby reducing image noise and other artifacts. Thinner slabs should only be used if partial volume effect is of importance. This should preferentially be done in conjunction with workstations that allow one to change the slab thickness in real-time. Except for very narrow slices, there should be no need for any increase in dose settings on reduction of slice thickness.(D. Tack · P. A. Gevenois (2007)

2.1.12.4 Pitch (p)

With SSCT scanners, scanning at increased pitch settings primarily serves to increase the speed of data acquisition. As a side effect, however, patient dose is reduced accordingly, at the expense of impaired slice profile width, i.e., z-resolution. As already outlined in section 4.2.6, MSCT scanners make use of a spiral interpolation scheme that is different from SSCT. Thus, the slice profile width remains unaffected from changes in pitch settings. Instead, image noise changes with pitch (Fig. 4.34a) unless the tube current is adapted accordingly. Scanners that make use of the effective mAs (mA sper slice) concept not only keep slice profile width, but also image noise constant when pitch changes (Fig. 4.34a). To achieve this goal, the electrical mAs product supplied to the X-ray tube automatically changes linearly with pitch (Fig. 4.34b). As a consequence, patient dose (CTDIvol) is no longer reduced at increased pitch settings in contrast with SSCT scanners; neither will dose increase at reduced pitch settings. MSCT scanners without automatic adaptation of mAs will still save dose at increased pitch setting, but this will happen at impaired image quality (more noise) as long as mAs is not adapted manually. Frequently, image quality in terms of artifacts depends on pitch settings. In general, spiral artifacts are reduced at

lower pitch settings. For similar reasons, some scanners allow the setting of a limited number of ‘preferred’ pitches only. Reduced pitch settings can also be applied to enhance the effective tube power, however, at the expense of reduced scanning speed..(D. Tack · P. A. Gevenois (2007)Pitch settings with MSCT scanners should be made exclusively with respect to scan speed, spiral artifacts and tube power. Dose considerations no longer play a role if scanners that employ effective mA sare used or if (electrical) mAs is adapted to pitch to achieve constant image noise.(D. Tack · P. A. Gevenois (2007)

2.1.12.5 Object Diameter (d) or Patient Weight (m):

Patient size, although not a parameter to be selected at the scanner’s console, represents an important influencing parameter that needs to be considered in this context. Considerable reductions in mAs settings are appropriate whenever slim patients, and particularly children, are examined. In order to avoid unnecessary over-exposure, the mAs must be intentionally adapted by the operator unless AEC-like devices are available. Due to the decreased attenuation for the smaller object, image quality will not be impaired if mAs is selected appropriately. This means that the image quality will be at least as good as for patients of normal size, although the dose has been reduced. The diameter is typical for a standard patient to whom the standard protocol settings refer to? From theoretical considerations (half-value thickness for CT beam qualities), mAs should be altered by a factor of two for each change in patient diameter of 4 cm tissue-equivalent thickness. However, dedicated studies (Wilting et al. 2001) have shown that this algorithm does not work well in practice: although objective (i.e., measured) noise was almost constant for patient diameters of between 24 cm and 36 cm, it was found that the subjective (i.e., perceived) image quality continually decreased with the patient diameter and vice versa. This is most likely due to the circumstance that adipose patients have more fatty tissue around their organs. Thus, the inherent contrast is better, and more noise can be tolerated. The opposite holds true with slim patients. Consequently, a more gentle adaptation of mAs with patient diameter (factor of two in mAs per 8-cm change in patient diameter) will better comply with clinical needs. Among the AEC systems .currently in use, those from Philips and Siemens already make use of this modified algorithm that ensures a constant ‘adequate’ image quality, while those implemented by General Electric and Toshiba simply attempt to ensure a constant noise level. As already outlined in

sections 4.3.2.2 for tube potential and 4.3.2.3 for slice thickness, strategies for automatic dose control that do not account for image contrast will fall short with respect to clinical needs. Similar considerations apply to the longitudinal dose modulation functionality: in examinations comprising several consecutive body sections with differing attenuation properties (e.g., in tumor staging of chest, abdomen and pelvis in a single spiral acquisition), mAs adjustment is often made in a way that ensures constant image noise, thus producing the highest settings in the pelvis region. However, inherent contrast in the pelvis region is much better than in the upper abdomen; consequently, reduced mAs settings would be more appropriate, as recommended in ICRP publication 88 (ICRP 2001). Although not specified explicitly, standard protocol settings implemented by the manufacturers are usually tailored to satisfy the vast majority of clinical situations except for obese patients in whom higher mAs or kilovolt settings must be applied. So, there is good reason to refer these standard settings to patients of about 80–85 kg body weight, which is also the average weight of European males. This corresponds to a lateral diameter of 33 cm, according to a detailed analysis of patient data from a large children's hospital in Germany (Schneider 2003;.). The following formula can be used to convert from lateral patient diameter (in cm) to patient weight m (in kg) and vice versa: In current literature, numerous differing recommendations can be found on how to reduce mAs settings with patient weight or diameter. In Fig. 4.35b, three examples are shown, which are representative of weak (Donellyet al. 2001), moderate (Rogalla 2004) and strong (Huda et al. 2000) adaptations of mAs to patient weight. As indicated by the dashed lines, mAs adaptation by a factor of two per 8-cm change in patient diameter is almost perfectly met by Rogalla'sr ecommendation, which follows a very simple relationship: Relative mAs ∝ □body weight + 5 kg (4.20) A similar relationship has been proposed by another research group (Honnefet al. 2004). This formula can be used to create a set of standard protocols for different weight classes (e.g., 0–5 kg, 6–10 kg, 11–20 kg, 21–40 kg, 41–60 kg, 61–80 kg, etc.), which can easily be applied in daily practice..(D. Tack · P. A. Gevenois (2007).

mAs settings should be adapted to patient size in a more gentle way (factor of two per 8-cm change in diameter) than predicted by theoretical considerations that only account for image noise. In addition, body regions with better inherent contrast should be scanned at reduced mAs settings. Preferentially, AEC systems that measure rather

than estimate patient absorption should be used, provided that their algorithm makes use of this more gentle mAs adjustment. Failing this, manual adjustment using a set of patient-weight-adapted protocols based on Rogalla's formula (4.20) should be applied instead. For head examinations, mAs adaptation should not be made with respect to patient weight, but to patient age.(D. Tack · P. A. Gevenois (2007)

2.1.13 Examination Parameters:

2.1.13.1 Scan Length (L):As already pointed out in section 4.1, the local dose, i.e., CTDI, is almost independent of the length of the scanned body section. The same does not hold, however, for the integral dose quantities, i.e., DLP and effective dose. Both increase in proportion to the length of the body section. Therefore, limiting the scan length according to the clinical needs is essential.(D. Tack · P. A. Gevenois (2007) On most scanners, the scan length, L, is usually not indicated explicitly. Instead, the positions of the first and the last slice are stated only; the same holds for the information that is documented on the images or in the DICOM data file..(D. Tack · P. A. Gevenois (2007)

For each patient, the scan length should be selected individually, based on the scan projection radiograph that is generally made prior to scanning for the purposes of localization, and should be kept as short as necessary. Moreover, a reduction in the scan range should be considered in multi-phase examinations and follow-up studies. Whenever feasible, critical organs, such as the eye lenses or the male gonads, should be excluded from the scan range.

This may be difficult for MSCT scanners that allow for large beam-width settings due to increased over ranging effects..(D. Tack · P. A. Gevenois (2007)

2.1.13.2Number of Scan Series (nSer):

In CT terminology, a scan series is usually referred to as a series of consecutive sequential scans or one complete spiral scan. With the limited tube power available for many SSCT scanners, CT examinations of long body sections (e.g., tumor staging of the entire trunk) had to be separated into several consecutive subsections. If the same protocol settings are applied to each series, the local dose will always be the same, while the integral dose is the sum of the DLP or effective dose values of each series. So it would not make a difference whether the body section is scanned as a whole or in

several shorter subsections, except for over ranging effects that will increase proportionally to the number of subjections. However, mAs settings can be adapted to the particular needs of each subsection, e.g., lower settings for the chest, higher settings for the upper abdomen and reduced settings for the pelvis, as indicated in section 4.3.2.5. If the same body section (or parts of it) is scanned more than once, this is usually denoted as ‘multiphasic’. However, this not only applies to examinations with administration of contrast agents, but also to examinations where the same body section is scanned with different orientation (such as in facial bone exams) or with different slice collimation settings (e.g., chest standard plus high resolution). Although more than one scan is made at the same position, the length of each single scan of a multiphasic exam does not necessarily have to be the same. While it is meaningful to sum up the integral doses (DLP, effective dose) of each phase, this is not true for the local doses (i.e., CTDIvol). Nevertheless, multi-phasic exams result in an increase in integral radiation exposure that is roughly proportional to the number of phases. The number of scan series (phases) should be kept as low as necessary. This holds true particularly for liver examinations, where studies with up to six different phases are sometimes recommended in literature..(D. Tack · P. A. Gevenois (2007)

2.1.13.3 Number of Rotations in Dynamic CT Studies (n) :

In dynamic CT studies, e.g., in CT fluoroscopy or in perfusion studies, a multiple number of scans is made at the same position. Therefore, it is meaningful to sum up the local doses, also. For this particular situation, the main issue is the avoidance of deterministic radiation effects. Local doses can be quite high if the scans are made with the standard dose settings used for that body region. Integral doses are normally comparable to the values encountered in standard examinations of the same region. However, with the advent of wider detector arrays, which may become even larger in future, integral dose will also be significantly increased. The doses applied in dynamic CT studies depend on two factors: the dose, i.e., the CTDIw, per rotation, and the number of rotations. As perfusion studies are regularly made with administration of contrast agents, the benefits of reduced kilovolt settings as described in section 4.3.2.2 should be used to reduce the dose settings. The number of rotations can be kept low by limiting the total length of the study, by reducing the image acquisition rate or by

intermitting the procedure (in CT fluoroscopy) whenever possible..(D. Tack · P. A. Gevenois (2007)

Dynamic CT studies should be made with the lowest dose settings, the most narrow beam width, the shortest length and the smallest image rate that is compatible with the clinical needs of the examination.(D. Tack · P. A. Gevenois (2007)

2.1.14 Quality Assurance Methods:

Quality control programs are designed to ensure that the CT system is producing the best possible image quality using the minimal radiation dose to the patient. An effective program provides a method for the systematic monitoring of the system's performance allowing the identification of specific problems or malfunctions. Responsibility for performing and documenting quality control tests is often shared between CT technologists and medical physicists. Technologists typically perform and record routine quality control tests; testing done by physicists is typically annual or semiannual. A medical physicist is required to obtain necessary dosimetric data. Quality assurance programs should adhere to three basic rules: 1) the tests that make up the program must be performed on a regular basis, 2) the results from all quality control tests must be documented in a consistent format, and 3) the quality control test should indicate whether the tested parameter is within specified guidelines (Lois E. Romans, RT, 2011).

2.2 Previous Studies:

Alhadi Abakar (2011) studied : Evaluation of Radiation Risks in Computed Tomography in Sudan .The purpose of his study was to evaluate the patient radiation dose in three different modalities of CT scanners (64 slices, 16 slices and 2 slices for routine CT investigations. A total of 108 patients were examined in this study in three different hospitals in Khartoum state. The data was collected from the following examinations: brain, chest, abdomen and pelvis. Dual slice scanner delivered the lowest radiation dose while 16 and 64 slice scanners delivered the highest radiation dose. The dose values of this study were comparable to the international reference levels CT examination. CT dose optimization protocol is not implemented in all hospitals.

Asim Abedelmajed , (2009) studied "Measurement of Radiation Dose During Extracorporeal Using Shock Wave Lithotripsy" The mean entrance surface dose is 0.3675 mGy and 0.3432 mGy for N.C and KADC respectively , these values are

considered an acceptable values if compared with 4 Gy as a lethal dose for adult man weighing about 70 kg , also these values regard as insignificant amount for producing ionization event in each cell , because of 1 mGy is the threshold of inducing ionization event per cell , although the measured dose were infinitesimal but ESWL cannot consider a safety treatment procedures in addition to the exposures from per- and post-procedure

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

Abdelrahman Mohamed (2009) studied: "Optimization of Radiation Dose in Abdomen Using Computerized Tomography "Optimization could be achieved through optimal study , body region of interest being scanned , and patient size.

Abdalazeem Ahmed Khalifa,(2009) studied :"Evaluation of Entrance and Exit Dose in Tangential Fields in Breast Cancer "This study showed there is a linear relationship between the separation , skin surface dose (entrance ,exit dose) and the lung dose . The skin dose and lung dose measured by (TPS) . The maximum dose received 5000 Gy in 25 fraction . The study showed that the dose received to the skin in entrance and exit points and to the lung in cases of SAD technique is higher than in cases of SSD technique. The skin dose increases proportionally to the separation in case of SAD technique. Therefore the SSD is much better in case of breast cancer and in opposing fields generally.

Lubna Osman Elbehery,(2008) studied: "Evaluation of Skin Dose in Nasopharyngeal Center Irradiation " The study investigated experimentally 30 nasopharyngeal center patients who had external radiation therapy in the period from August 2007 to may

2008 , a random sample was used . This study showed there is a linear relationship (direct proportionality) between skin surface dose separation and the area of the filed . The skin dose measure by TLDs and TPS was 51% and 53.2% respectively from the maximum dose .

KawtherYousif Shambal,(2011) studied: " Assessment of Radiation Dose Received by Personnel and Patient in Fluoroscopic Urology" The dose at the exposure center varies according to exposure factors , but the noted high dose wes xx mSv. The exposure at right side was higher than left side with 0.1 mSv /exam .The patient right eye would receive 0.62 mSv /exam and the left eye would receive 0.66 mSv /exam , while the legs would receive 0.64 mSvexam in mean.

Omer Osman Omer ,(2012) studied: " Effective Dose Estimation During Pediatric Chest X.ray Radiography" The chest radiography are the most commonly performed pediatric radiological exam .Because of this it is imperative to understand the effects of dose rates on the pediatric patients . Pediatric patients in general are more sensitive to the effects of ionizing radiation due to nature of their rapidly dividing cells and with radiological exams on the rise it is important to understand that the effects of ionizing radiation are cumulative. In order to help reduce the effects of ionizing radiation within the pediatric population we need to utilize manual techniques .

Einas Mohamed Ahmed ,(2012) studied: "Calculation of Dose Received by Sensitive Organs in Breast Cancer Radiotherapy Using Day's Method" Day's method was used to calculate the % dose received by the critical organs where they considered as point outside the irradiated volume, therefore they received dose from the scattered dose inside the patient as a function of field size and transmitted rays.

Nazar Alhassan Mohamed,(2012) studied: "Measurement of Radiation Dose in Intravenous Urography" This study indicates that the impactions of digital radiography in departments made a considerable dose reduction possible . Reduction of number of radiographs taken per exam is also factor of dose reduction. The result the radiation dose is lowest than the reported in previous studies done in Sudan. The radiation risk from IVU procedure based on the results of very low compared to the previous studies.

Daniele Marin et al, (2011) in their study titled "Body CT: Technical Advances for Improving Safety" reported that although CT is a powerful tool that has transformed

the practice of medicine; the benefits are accompanied by important hazards. Radiologists must understand these hazards and the strategies available to minimize them as well as the hazards associated with contrast medium delivery in abdominal CT.

Ali S. Raja, et al, (2010),studied: "Negative Appendectomy Rate in the Era of CT: An 18-year Perspective" concluded there was a significant reduction in both the NAR and the number of appendectomies in patients who presented to the emergency department during an 18-year period, which was associated with a significant increase in the use of preoperative abdominal CT.

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

Materials and Methods

Chapter Three
Materials and Methods

3.1: Materials

In this study the data were collected from the following radiological departments: (A) Alamal And (B) Alribat Hospital in Khartoum State. Data of the technical parameters used in CT procedures were taken during 2014-2015 ,.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.

3.1.1 CT equipment Used:

Table 3.1 demonstrates CT machines used in this study

Slice No.	Manufacture	Detector type
16 slice	Neusoft	16 rows
16 slice	Siemens Somatom	16 rows
64 slice	ToshibaAquilion	64 rows

3.1 CT machines:

Fig(3.1) Shows the CT machine Siemens-Sensation-16-Slice

Fig(3.1) Shows the CT machine Toshiba_Aquilion_64-Slice

3.1.2: Patient: A total of 111 patients (male and female) examined in this study: as illustrated in Table 3.2

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

Number of slice	Abdomen	Chest	Brain	total
16 slice Siemens Sensation	12	11	13	36
16 slice Neusoft	12	12	13	37
64 slice	12	11	12	35
Total	36	34	38	111

3.2:Methods:

3.2.1 Technique used:

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

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

For abdomen scan by "16 and 64 slice" 5mm slice thickness .Axial cuts all the abdomen.

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

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

3.2.1.1: Patient preparation, protocol and technique:

3.2.1.1.1 Patient preparation:

All metallic objects should be removed. Use sedation or anesthesia (no motion during scan). Empty stomach if anesthesia or contrast media indicated.

3.2.1.1.2 Patient position:

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

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

Head rest on head holder (CT brain).

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

Chapter Four

Results

Chapter Four

Results

In this study, a total of 111 patients were examined in two Hospitals of Khartoum state by three CT machines.

Table 4-1 the mean and standard deviation of the radiation parameters for the different organs

Organs	DLP	CTDIV	mA	kv	scan time
Abdomen	653.9± 315.7	14.2± 7.8	175.5±59.4	120±0	0.64±0.1
Brain	1427.4±625.2	69.5± 18.2	311.8± 112.4	115±11.3	1.35±0.9
Chest	430. 6 ±254.7	11.4±7.1	129.5±80.4	120±0	0.63± 0.2
Pelvis	821.3± 273.1	21.8± 6.7	192.6± 74.9	120±0	0.76± 0.1

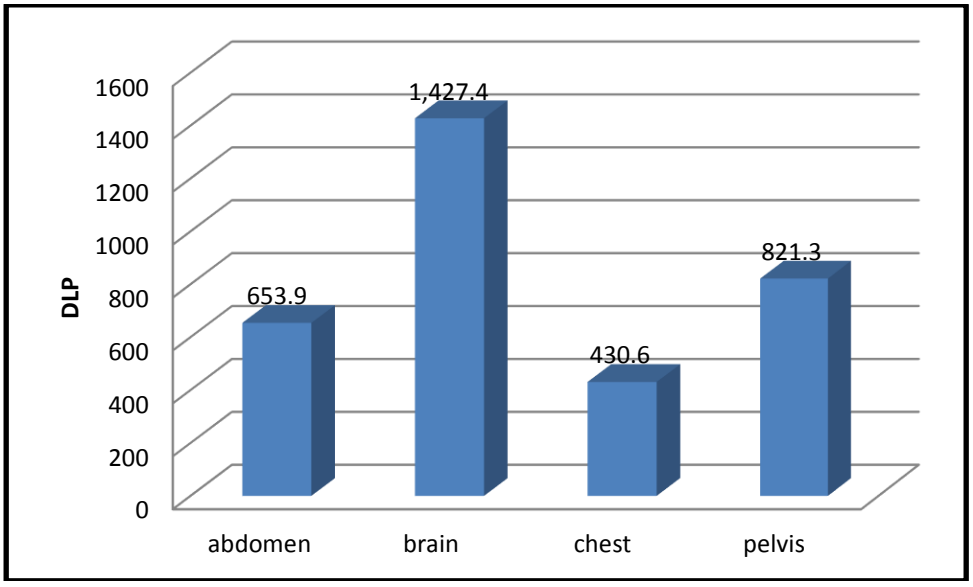

Figure 4-1 a bar graphs shows the average distribution of DLP body structures

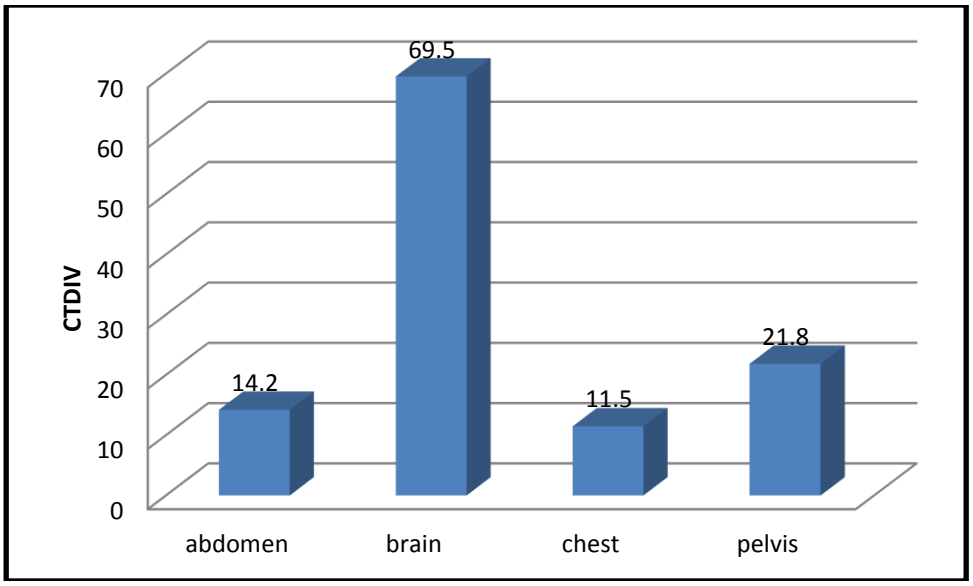

Figure 4-2 a bar graphs shows the average distribution of CTDIv body structures

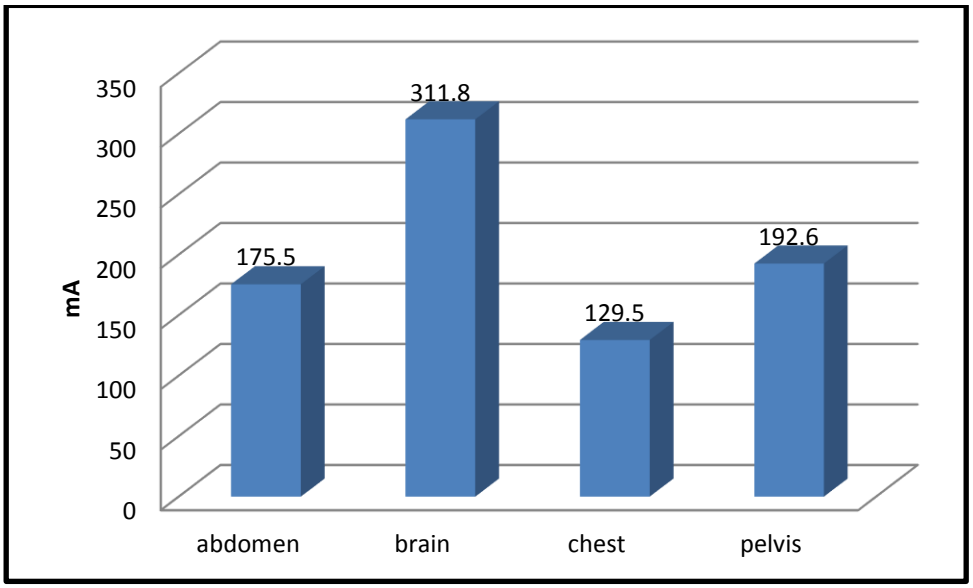

Figure 4-3 a bar graphs shows the average distribution of mA body structures.

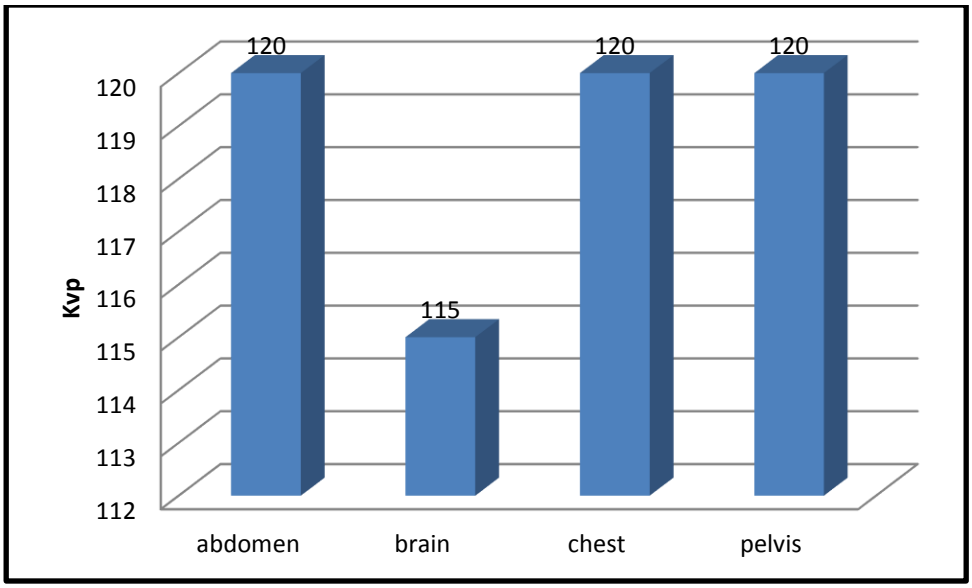

Figure 4-4 a bar graphs shows the average distribution of Kvp body structure

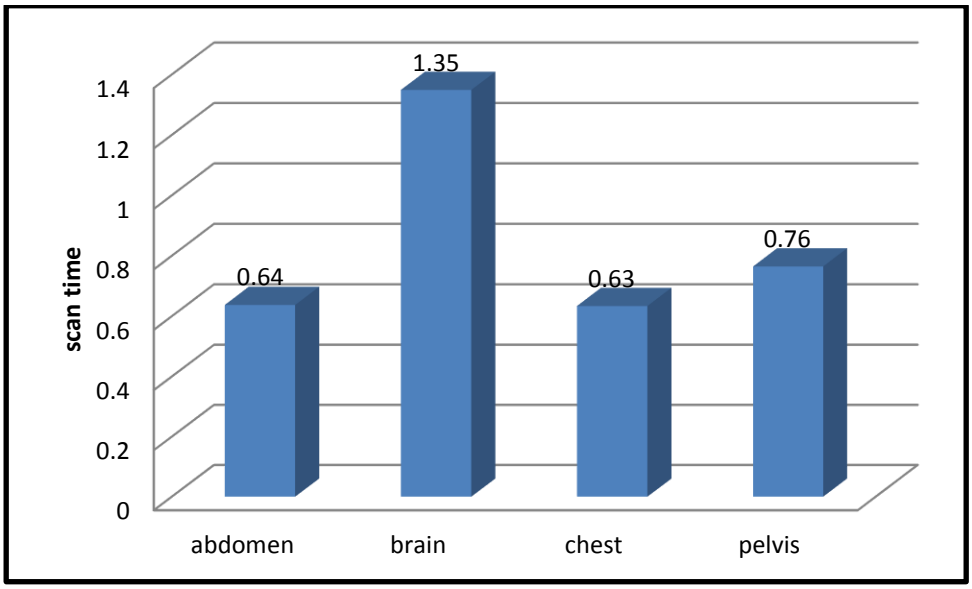

Figure 4-5 a bar graphs shows the average distribution of scan time body structures

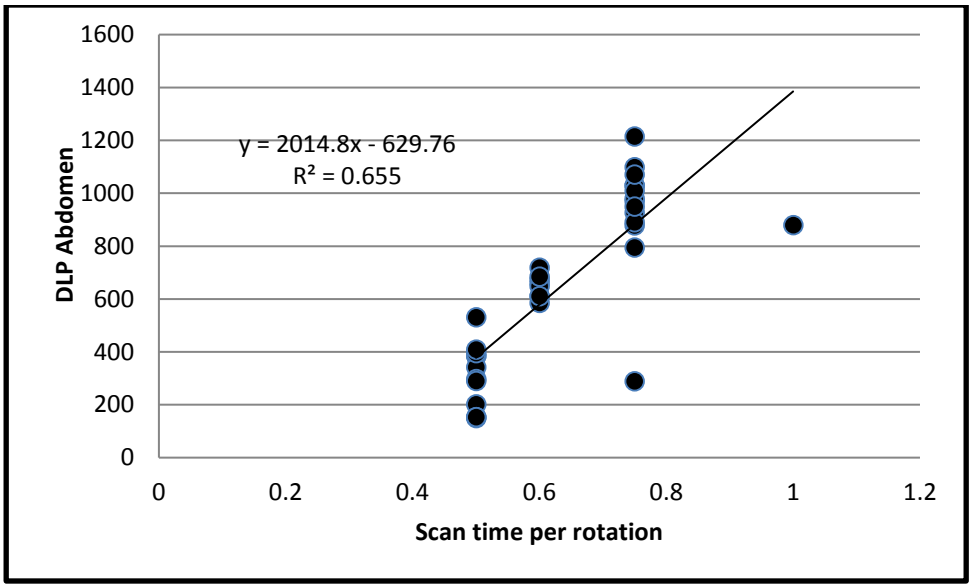

Figure 4-6 scatter plot show a direct linear relationship between the DLP of abdomen per scan time

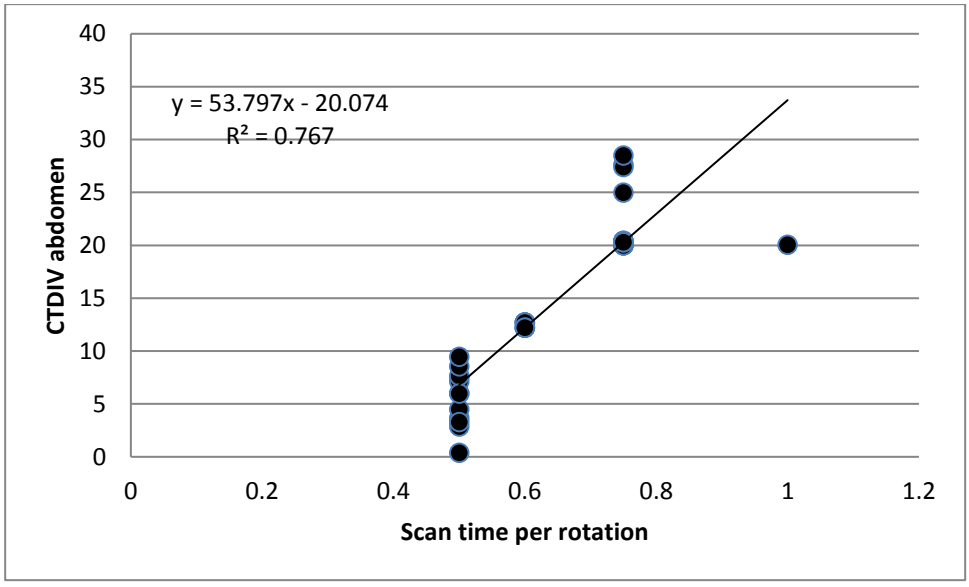

Figure 4-7 scatter plot show a direct linear relationship between the CTDIv of abdomen per scan time

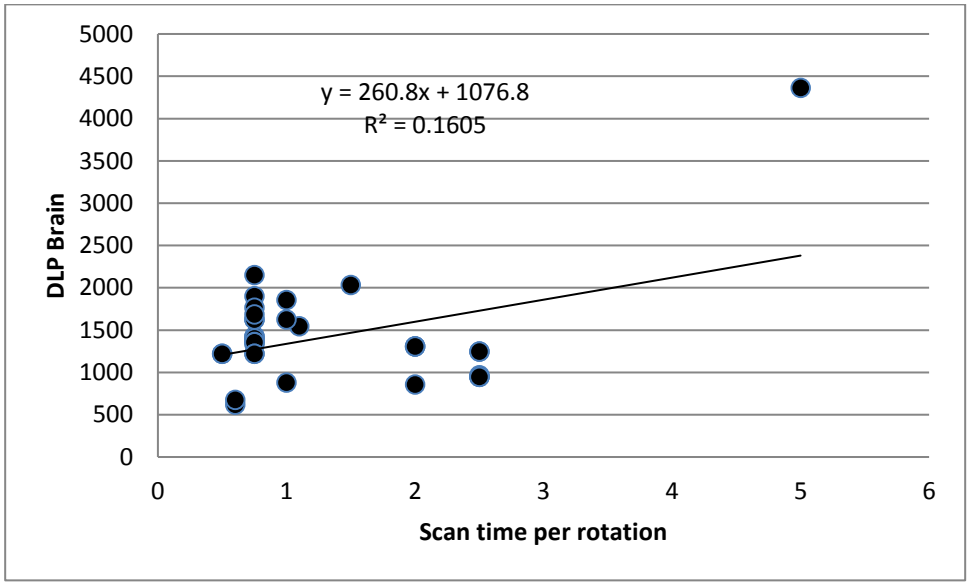

Figure 4-8 scatter plot show a direct linear relationship between the DLP of Brain per scan time

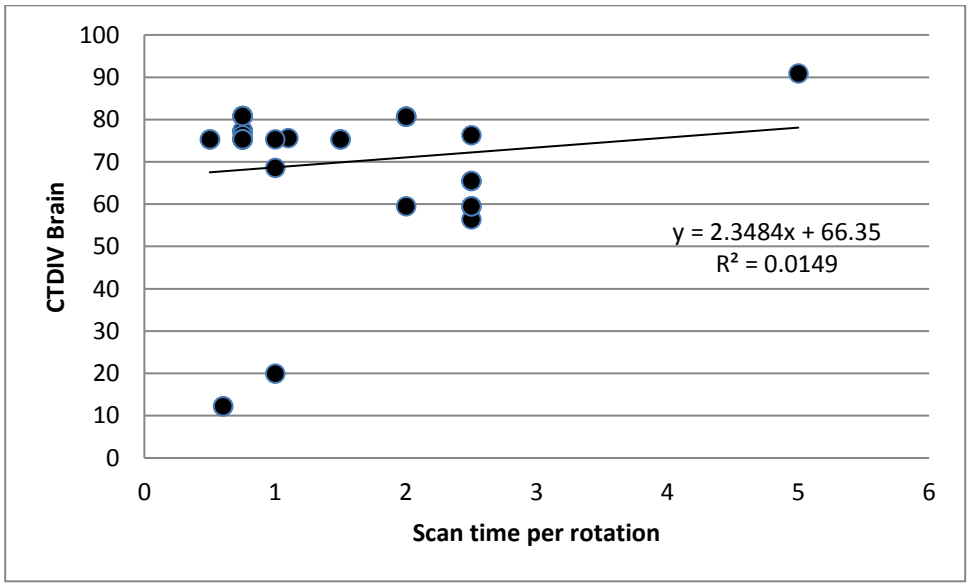

Figure 4-9 scatter plot show a direct linear relationship between the CTDIv of Brain per scan time

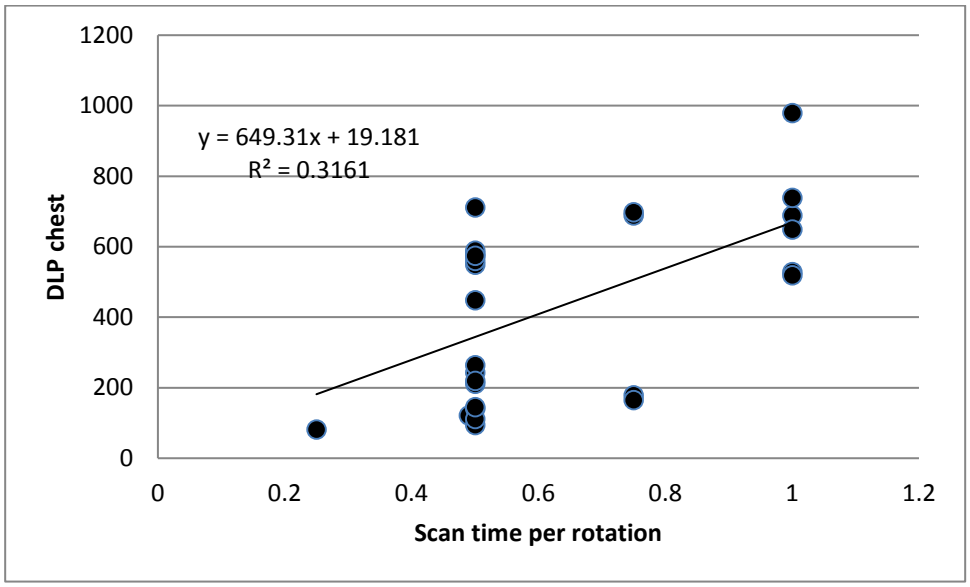

Figure 4-10 scatter plot show a direct linear relationship between the DLP of chest per scan time

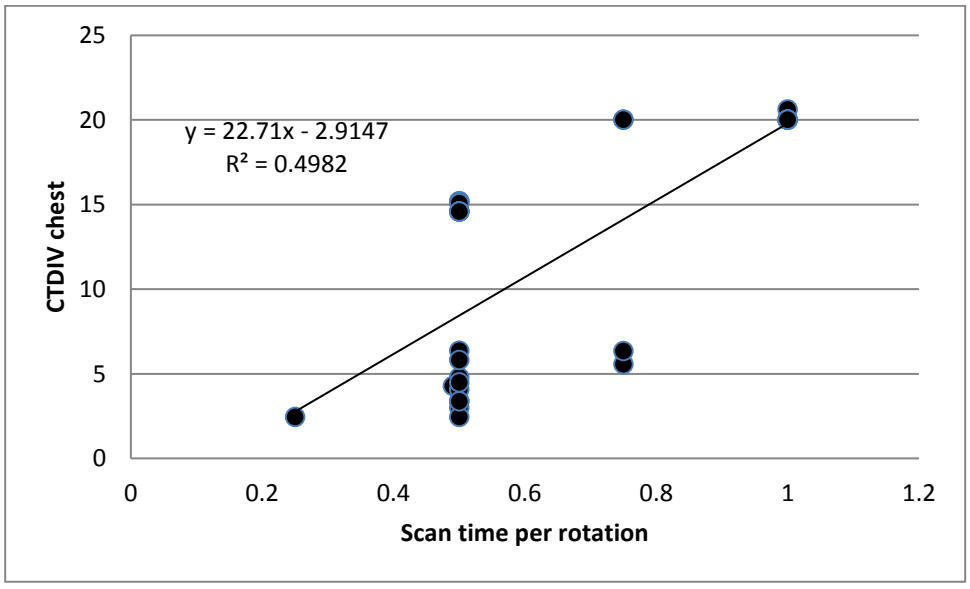

Figure 4-11 scatter plot show a direct linear relationship between the CTDIv of Chest per scan time

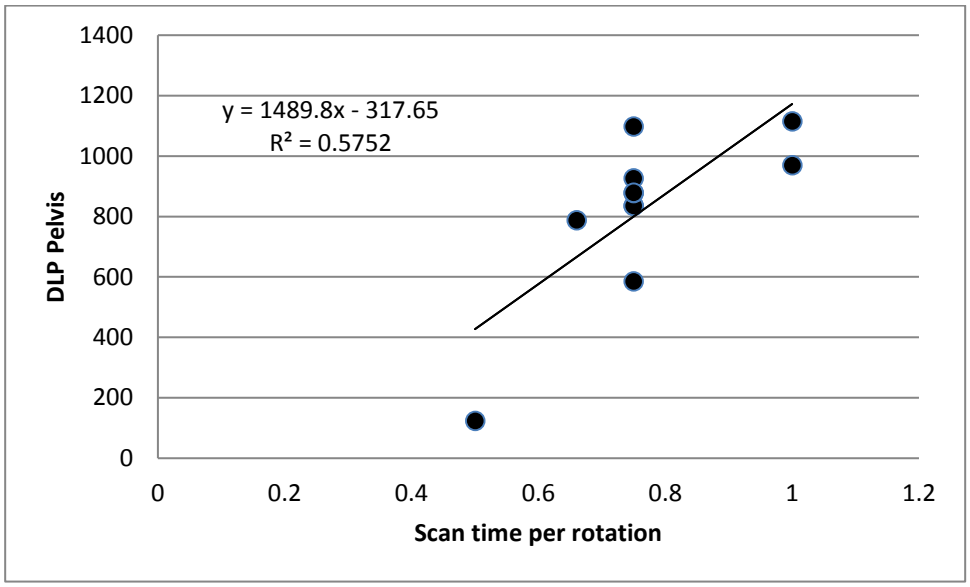

Figure 4-12 scatter plot show a direct linear relationship between the DLP of the pelvis per scan time

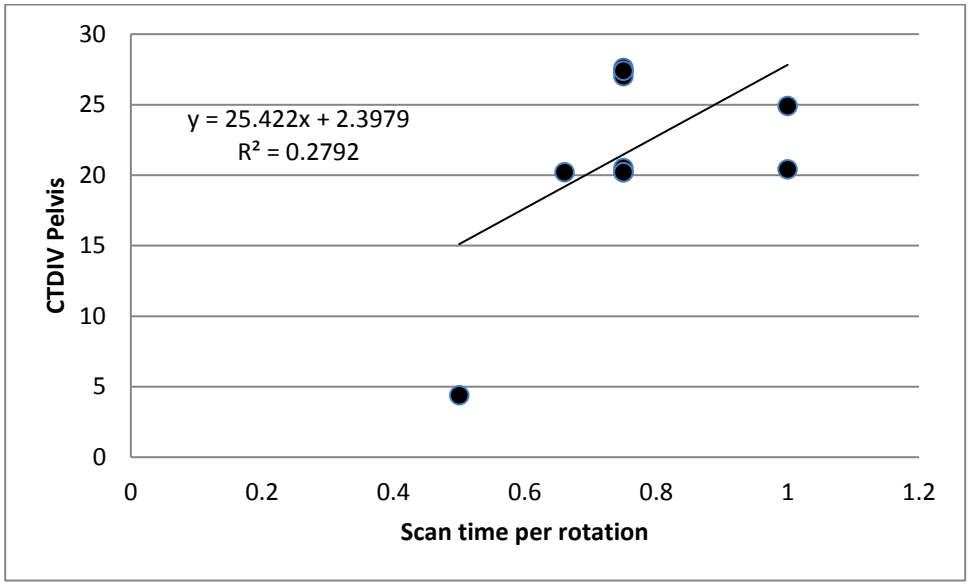

Figure 4-13 scatter plot show a direct linear relationship between the CTDIv of the pelvis per scan time

Abdomen

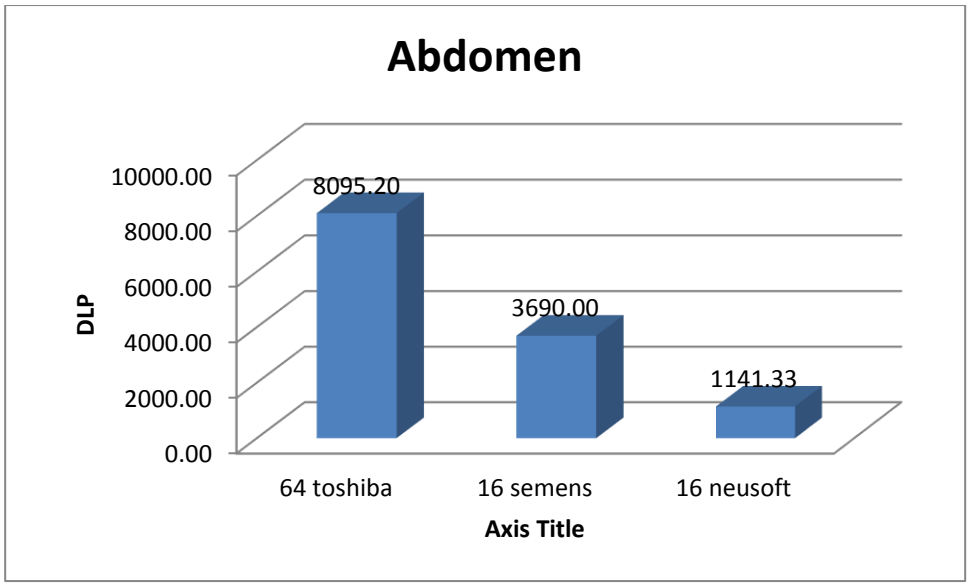

Figure 4-14 a bar graphs shows the relationship between the DLP and three types of CT machines for the abdomen.

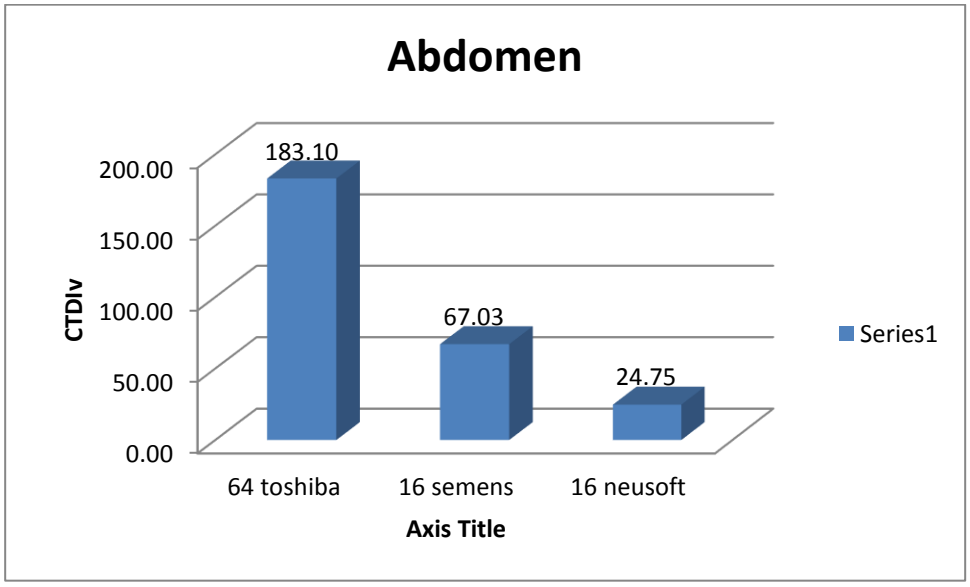

Figure 4-15 a bar graphs shows the relationship between the CTDIv and three types of CT machines for the abdomen.

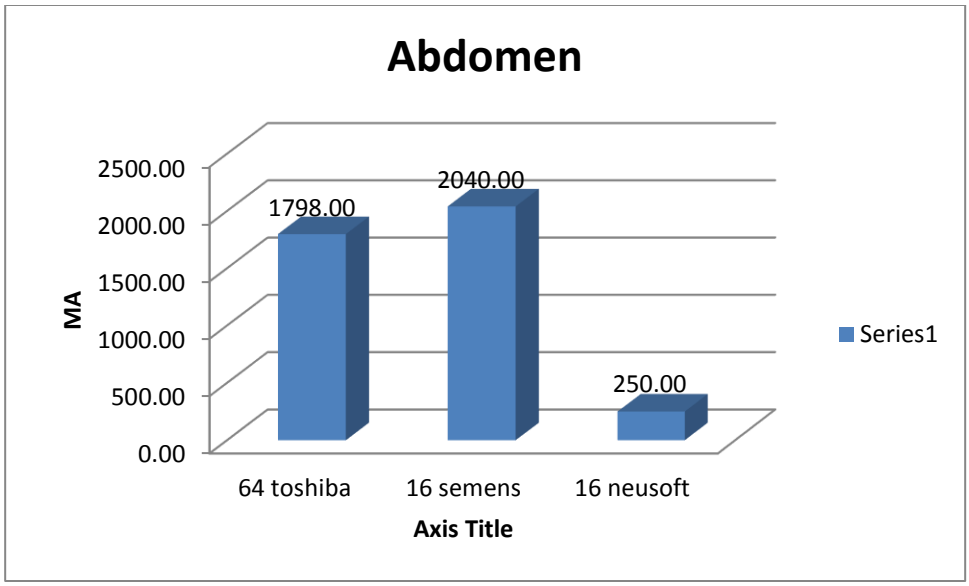

Figure 4-16 a bar graphs shows the relationship between the MA and three types of CT machines for the abdomen.

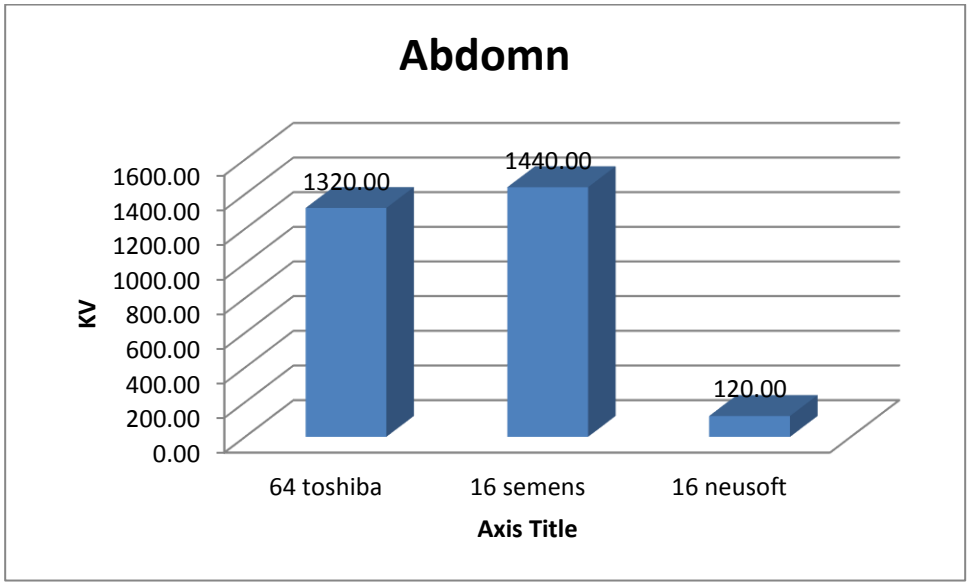

Figure 4-17 a bar graphs shows the relationship between the K and three types of CT machines for the abdomen.

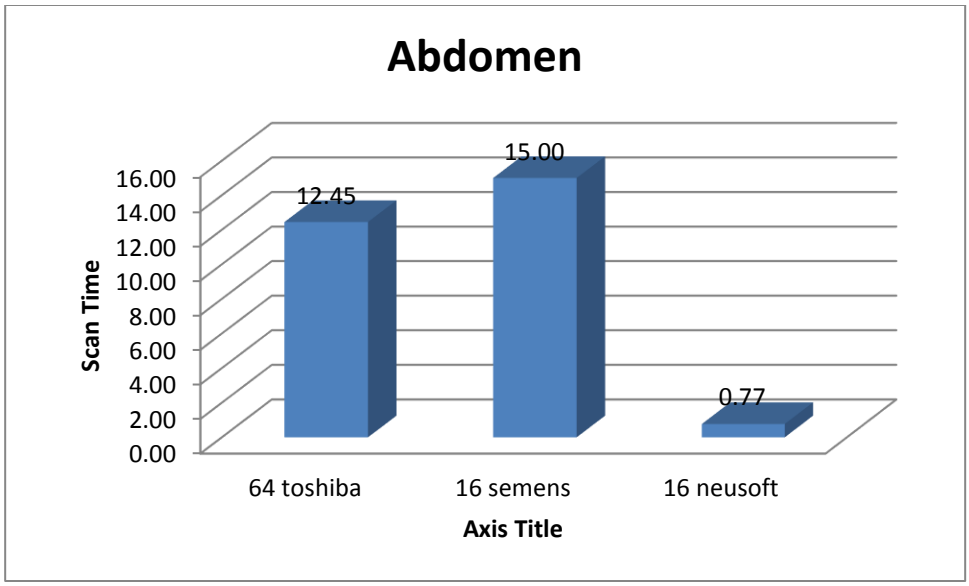

Figure 4-18 a bar graphs shows the relationship between the scan time and three types of CT machines for the abdomen.

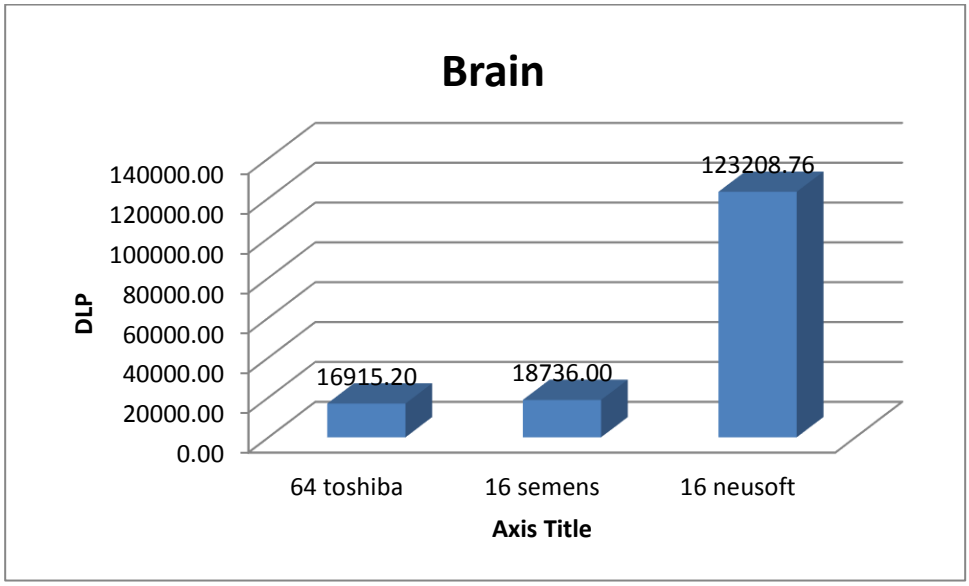

Figure 4-19 a bar graphs shows the relationship between the DLP and three types of CT machines for the brain.

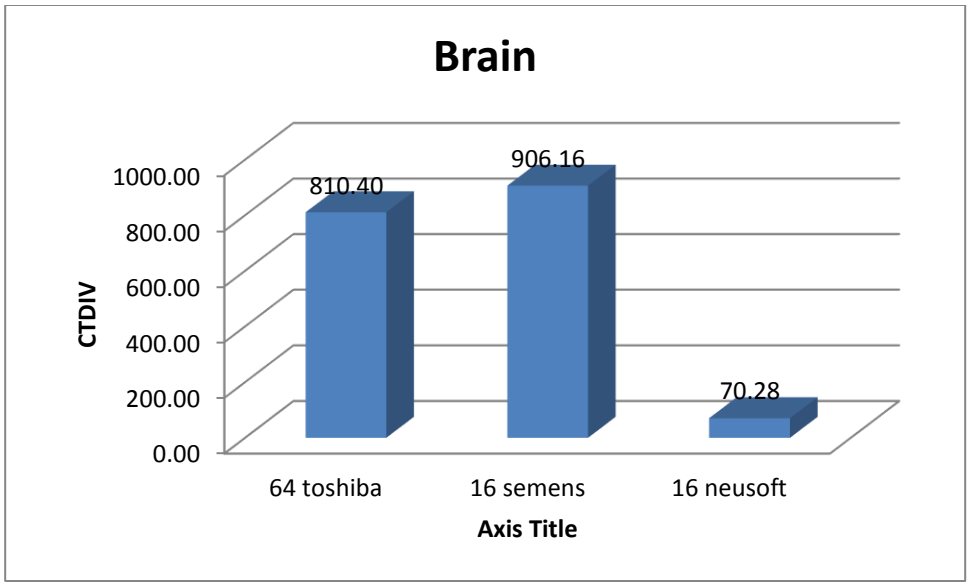

Figure 4-20 a bar graphs shows the relationship between the CTDIv and three types of CT machines for the brain.

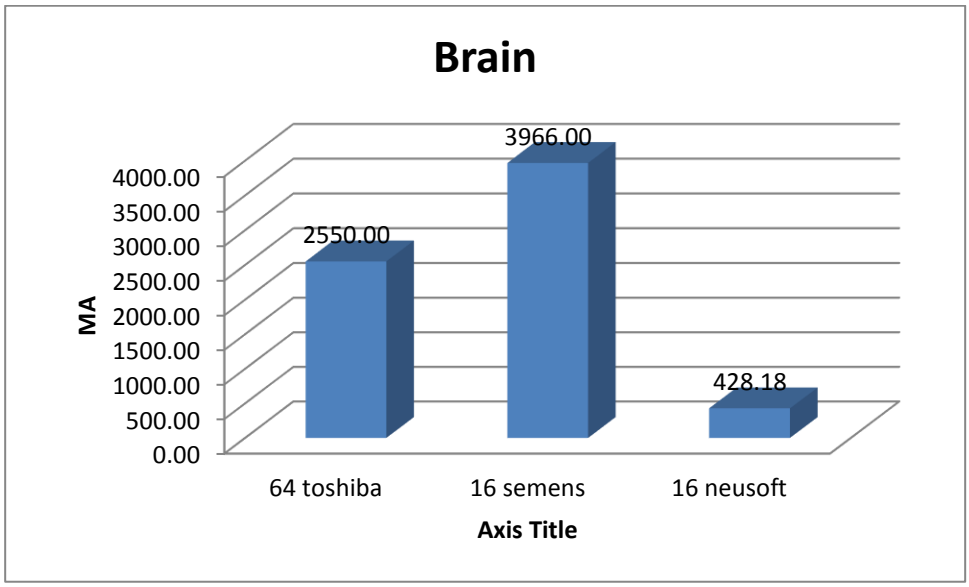

Figure 4-21 a bar graphs shows the relationship between the MA and three types of CT machines for the brain.

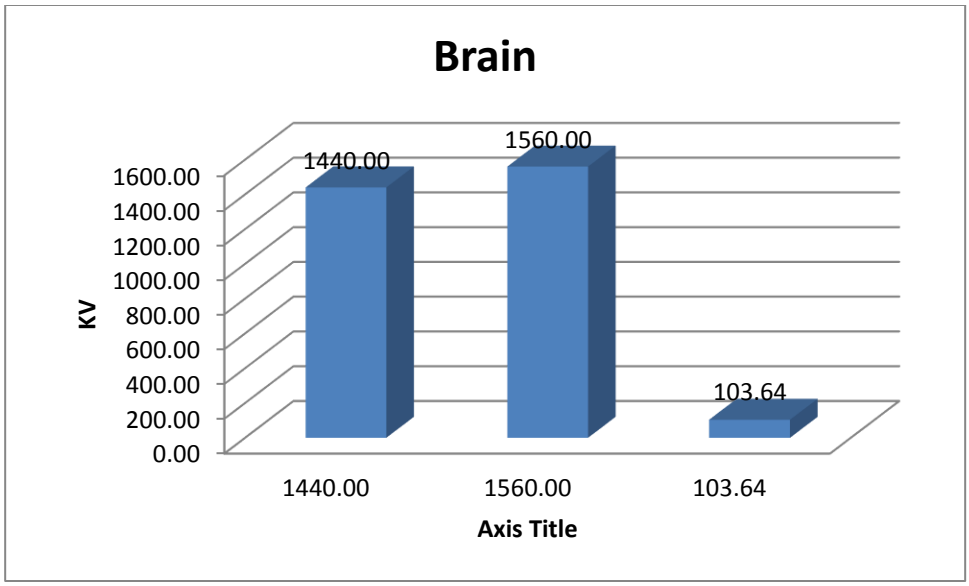

Figure 4-22 a bar graphs shows the relationship between the KV and three types of CT machines for the brain.

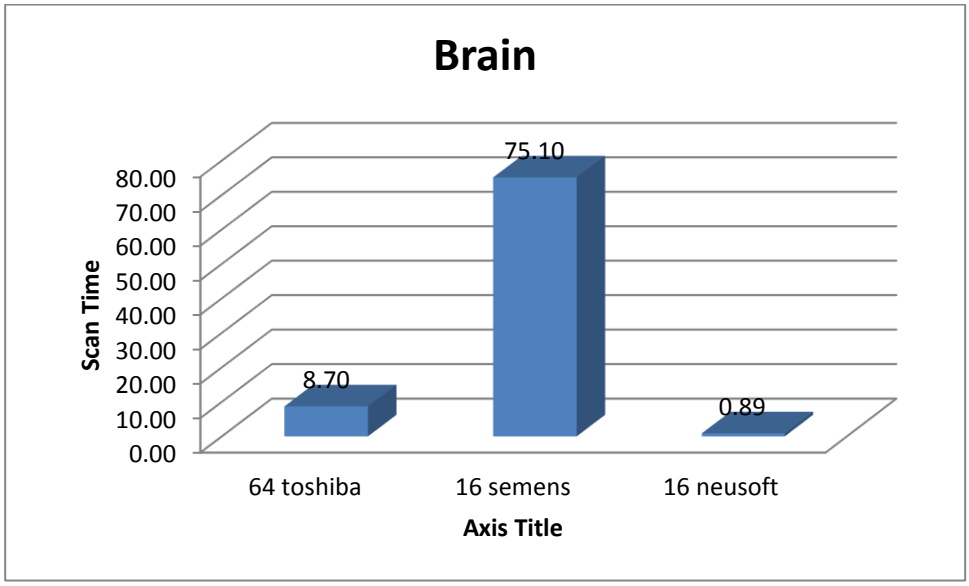

Figure 4-23 a bar graphs shows the relationship between the scan time and three types of CT machines for the brain.

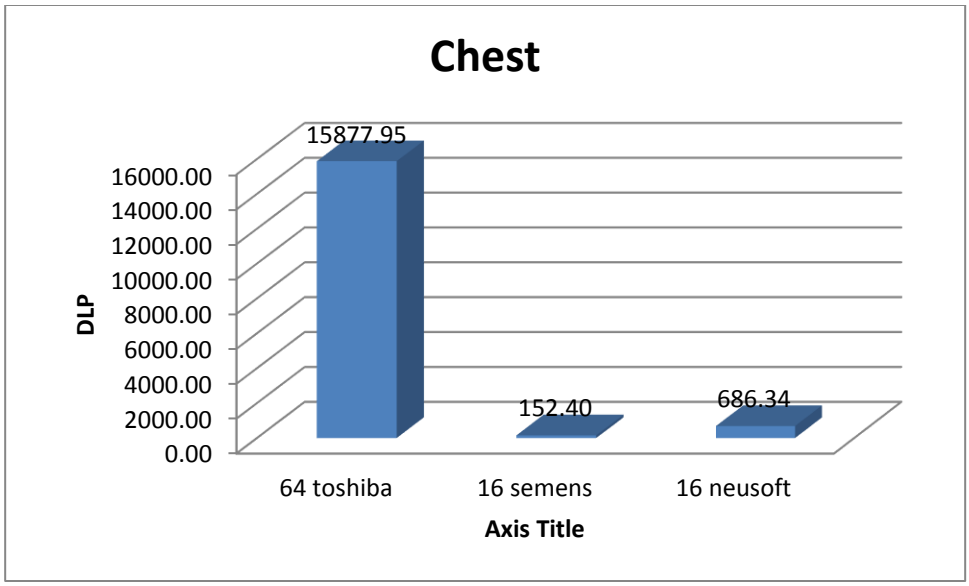

Figure 4-24 a bar graphs shows the relationship between the DLP and three types of CT machines for the chest.

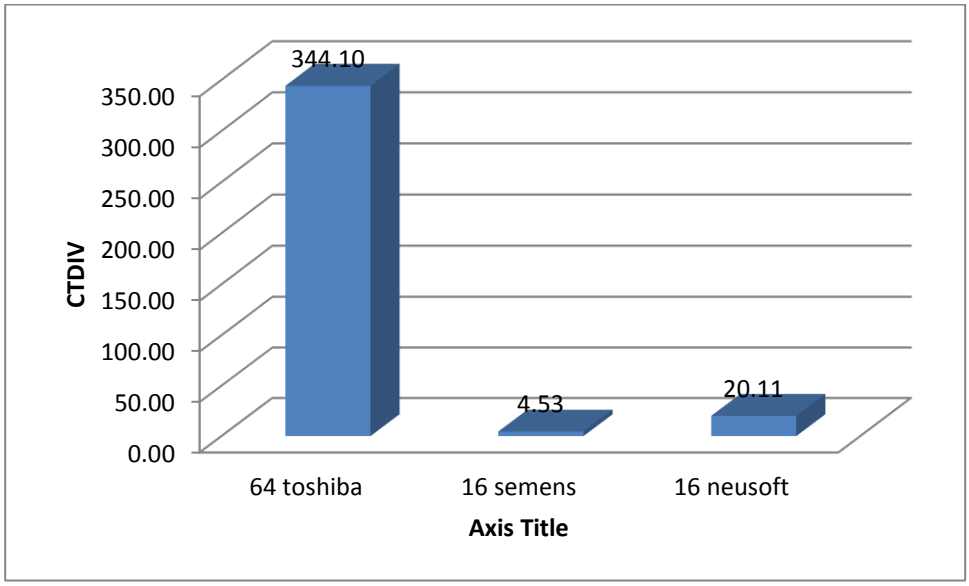

Figure 4-25 a bar graphs shows the relationship between the CTDIv and three types of CT machines for the chest.

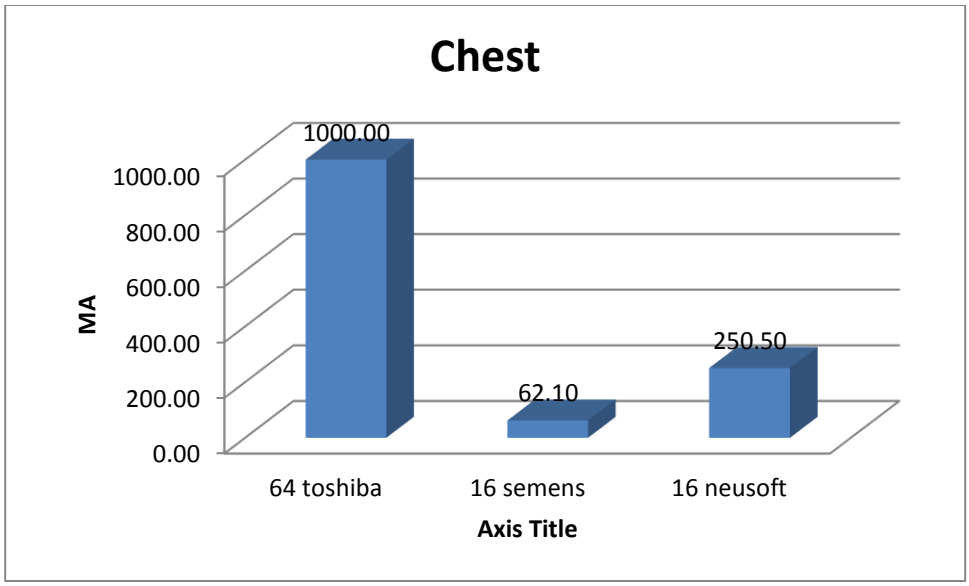

Figure 4-26 a bar graphs shows the relationship between the MA and three types of CT machines for the chest.

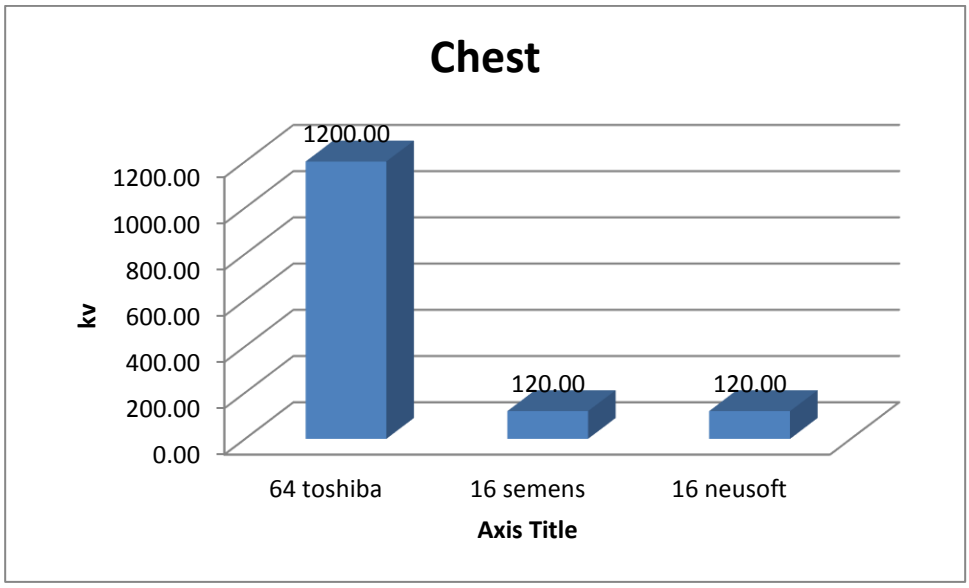

Figure 4-27 a bar graphs shows the relationship between the KV and three types of CT machines for the chest.

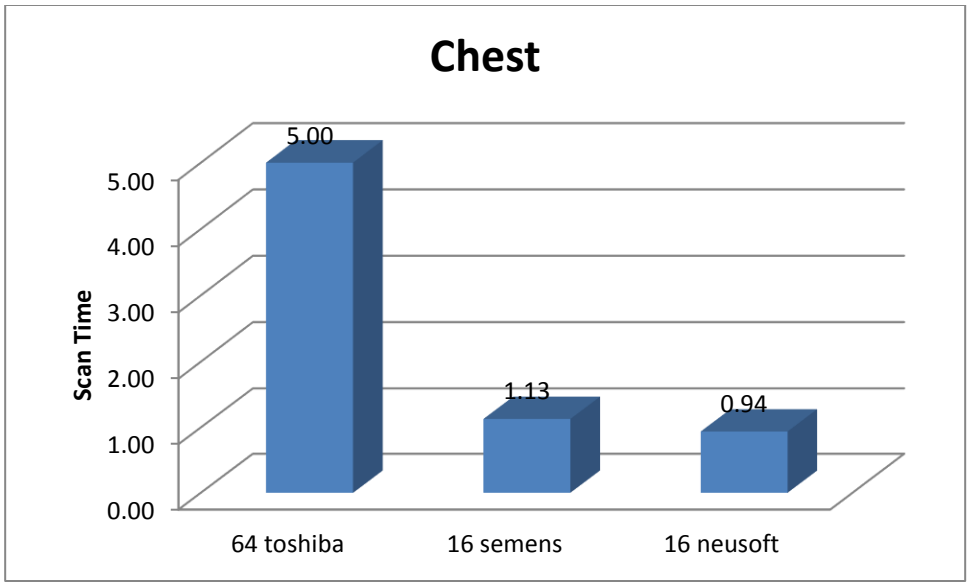

Figure 4-28 a bar graphs shows the relationship between the scan time and three types of CT machines for the chest.

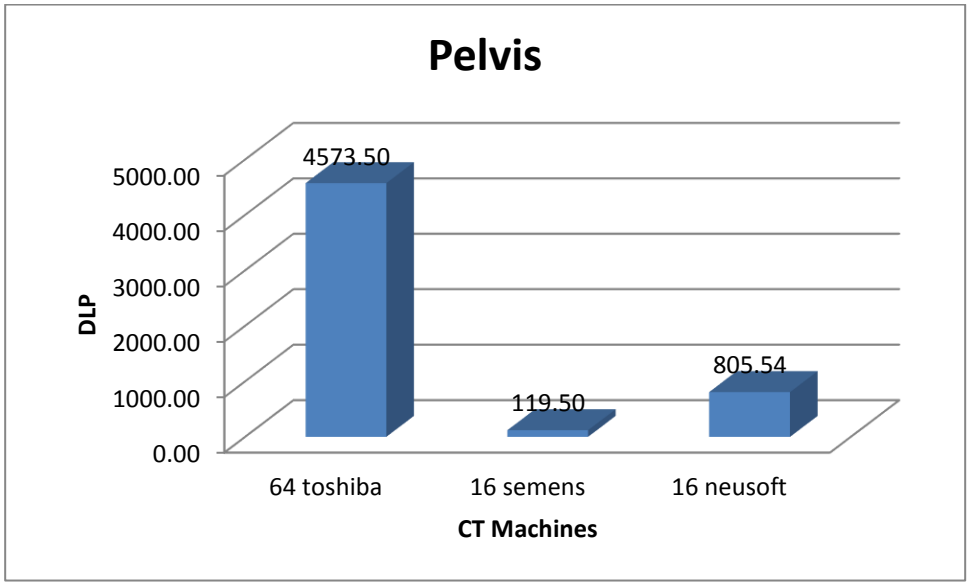

Figure 4-29 a bar graphs shows the relationship between the DLP and three types of CT machines for the Pelvis.

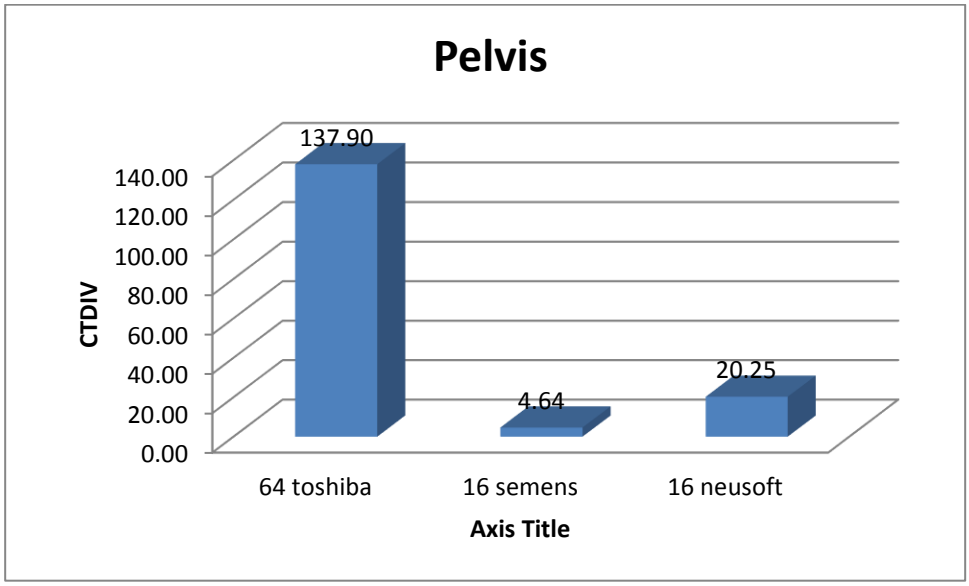

Figure 4-30 a bar graphs shows the relationship between the CTDIv and three types of CT machines for the Pelvis.

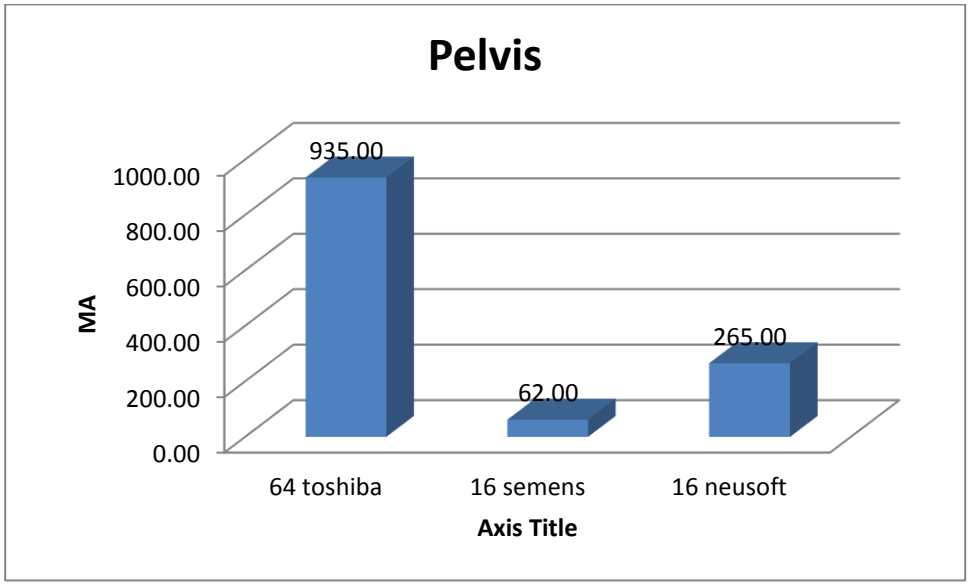

Figure 4-31 a bar graphs shows the relationship between the MA and three types of CT machines for the Pelvis.

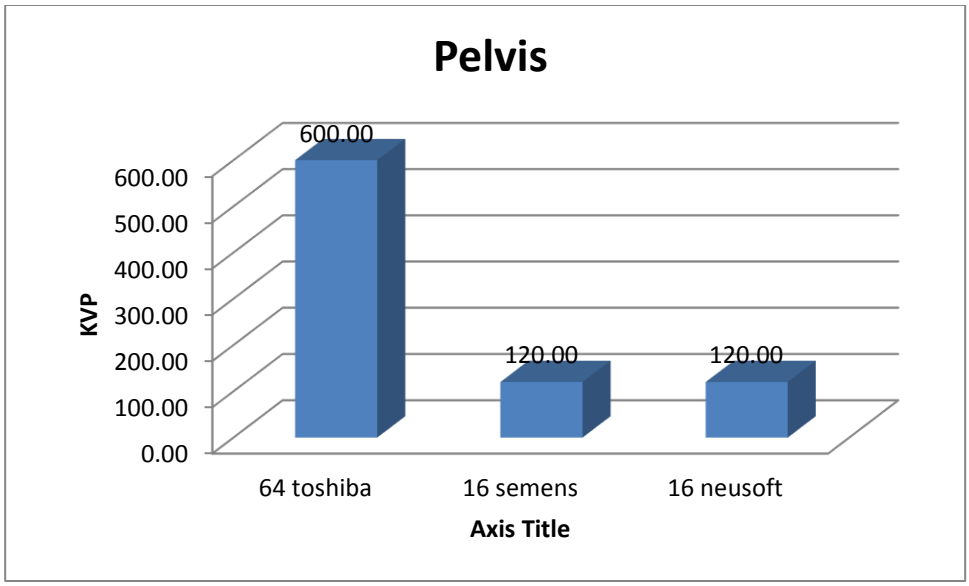

Figure 4-32 a bar graphs shows the relationship between the KV and three types of CT machines for the Pelvis.

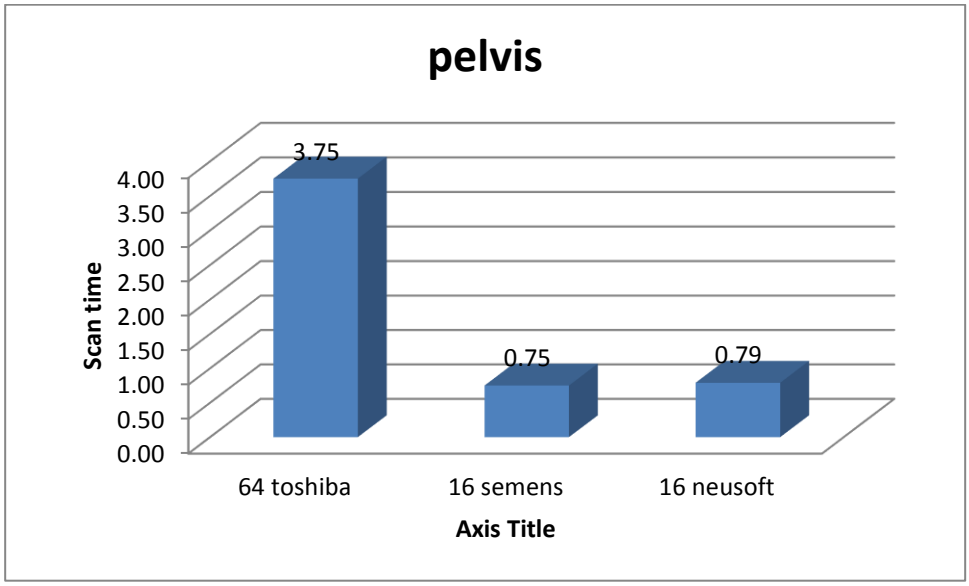

Figure 4-33 a bar graphs shows the relationship between the scan time and three types of CT machines for the Pelvis.

Chapter Five

Discussion, Conclusion and Recommendations

Chapter Five

Discussion, Conclusion and Recommendations

5.1 Discussion:

CT scanning has been recognized as high radiation dose modality ,when compared to other diagnostic X.ray techniques , since its launch into clinical practice more than 30 years ago over that time , as scanner technology has developed and its use has became more widespread , concerns over patient radiation dose from CT have grown, the introduction of multi-slice scanners has focused further attention on this issue , and it is generally believed that it will lead to higher patient doses. Measure and estimates of the impact of irradiating by CT machine during the CT examined is important in order to quantify the dose for further dose optimization. Radiation doses from CT vary widely, and they could be reduced significantly if strategies for minimizing exposure were more widely followed. in the current work we measured and estimated radiation exposure from different CT scans in two hospitals. In this study, a total of 111 patients suffering from brain, chest ,pelvis and abdominal disorders were subject to CT scans.

With respect to the Relationship between the body structures and each variable by the three types of CT machines , Patients CT dose in this study were measured and estimated in different CT machines. Figure(3.1) shows 16 (Siemens Somatom) system and 16-slice (Neusoft)system and sixty four slice (Toshiba).The results of this study showed wide variation in patients dose among different detectors number in terms of DLP and $CTDI_{vol}$, for chest examination the DLP and $CTDI_{vol}$ shown in Figure(4.1) was less than the other studies in terms of DLP , $CTDI_{vol}$. In Brain examination the DLP and $CTDI_{vol}$ were higher than other examinations shown in Figure(4.1). The Brain examination gave a highest mA value while chest examination gave the lowest one as shown in Figure (4-3).

Regarding Kvp all the selected examinations have the same value which was 120 except the Brain examination value which was 115 as shown in Figure (4.4). The scan time of the Brain examination was higher while the chest examination was lower as shown in Figure (4.5).

In the abdominal examination there was direct relationship between DLP, CTDIv and scan time as shown in the scatter plot Figure (4.6), (4.7).

i.e. DLP, CTDIv α scan time.

In Brain examination, like the abdominal examination, there was direct relationship between DLP, CTDIv and scan time as shown in the scatter plot Figure (4.8), (4.9).

i.e. DLP, CTDIv α scan time.

Chest examination as shown in Figure (4.10) and (4.11), had direct relationship between DLP, CTDIv and scan time.

i.e. DLP, CTDIv α scan time.

Pelvic examination in Figure (4.12) and (4.13) demonstrates direct relationship between DLP, CTDIv and scan time.

i.e. DLP, CTDIv α scan time.

Concerning the Relationship between each body structures and each variable by the three types of CT machines.
The study regarding the Abdomen DLP found that: 16-slice (Neusoft) had the lowest DLP, the 64 slice (Toshiba) has highest value while the 16 (Siemens Somatom) recorded the middle value . the MA ,KVp and scan time found that: 16-slice (Neusoft) has the lowest MA ,KVp and scan time , the16 (Siemens Somatom) has highest value while 64 slice(Toshiba) the recorded the middle value . For The Brain. the DLP in 16-slice (Neusoft) has the highest 0DLP, the16 (Siemens Somatom) has middle value while 64 slice (Toshiba) recorded the lowest value . in the study regarding the CTDIv ,MA ,KVp and scan time it was found : 16-slice (Neusoft) has the lowest CTDIv ,MA ,KVp and scan time , the16 (Siemens Somatom) had highest value while 64 slice(Toshiba) the record the middle value .,this mean that the Brain is different from the other body structures in recording the highest values for different variables ,while the chest recorded the lowest ones.

As for the DLP,CTDIv and MA of the chest ,The reacher found that: 16-slice (Neusoft) had the middle DLP,CTDIv and MA the16 (Siemens Somatom) had lowest value while 64 slice(Toshiba) the recorded the highest value
As for Kvp it found that: 16-slice (Neusoft) and the16 (Siemens Somatom) have equal value while 64 slice(Toshiba) the recorded the highest value
On the other hand the Scan time we found that: 16-slice (Neusoft) had the lowest , the16 (Siemens Somatom) had middle value while 64 slice(Toshiba) recorded the highest value.

Concerning the DLP,CTDIv, MA ,Kvp and Scan time for the Pelvis. we found that: 16-slice (Neusoft) has the middle DLP,CTDIv, MA ,Kvp and Scan time , the16 (Siemens Somatom) has lowest value while 64 slice(Toshiba) the record the highest value .

Concerning the DLP of the Brain .16 scanner (Neusoft) delivered the lowest radiation dose while 16 and 64 slice scanners delivered the highest radiation, the current study agree with (Mona 2012)and(Alhadi 2011).

5.2 Conclusions:

The current study concluded that:

In the abdominal examination there was direct relationship between DLP, CTDIv and scan time i.e. DLP, CTDIv α scan time.

In Brain examination, like the abdominal examination ,there was direct relationship between DLP, CTDIv and scan time i.e. DLP, CTDIv α scan time.

Chest examination, had direct relationship between DLP, CTDIv and scan time. i.e. DLP, CTDIv α scan time.

Pelvic examination demonstrates direct relationship between DLP, CTDIv and scan time.i.e. DLP, CTDIv α scan time.

Brain examination takes the highest radiation dose in term of DLP and CTDIv so especial care should be taken during performing such examination.

5.3 Recommendations:

*Regular and continuous training for the technologists who works in CT departments.

* Further studies about this topic should be encouraged to improve the performance and minimize the radiation hazards.

*CT machines should be available at each radiology department.

References

References:

Alhadi Abakar (2011) , Evaluation of Radiation Risks in Computed Tomography in Sudan

Avinash C.Kak Computerized To mographic Imaging School of Electrical Engineering P undue University Malcolm Slaney SchfumbergerPalo Alto Research

Brenner D, Elliston C, Hall E, Berdon W. Estimated risks of radiation-induced fatal cancer from pediatric CT. AJR Am J Roentgenol 2001.

Brenner DJ"2004" Radiation risks potentially associated with low-dose CT screening of adult smokers. For lung cancer Radiology.

Brenner DJ, Hall EJ. Computed tomography: an increasing source of radiation exposure. N Engl J Med 2007.

Brenner DJ. Radiation risks potentially associated with low-dose CT screening of adult smokers. For lung cancer.Radiology. 2004.

Cohnen M, Poll LJ, Puettmann C, Ewen K, Saleh A, Modder U. Effective doses in standard protocols for multi-slice CT scanning. EurRadiol 2003.

D Karthikeyan ,DeepaChegu ,(2005)Step by Step CT ScanFirst Edition : 2005JAYPEE BROTHERS MEDICAL PUBLISHERS (P) LTDMEDICAL PUBLISHERS (P) LTDNew Delhi

D. Tack · P. A. Gevenois (2007) Radiation Dose from Adult and Pediatric Multidetector Computed Tomography¤Springer-Verlag Berlin Heidelberg ,Printed in Germany.

Euclid seeram , RT (R) (2008)Edition , Burnaby , bccanda , (1-235) .

Euclid seeram , RT (R) , B.Sc. MSc British Columbia institute of technology , computed tomography (Alstudy Guide and review), 6th . Edition , Burnaby , bccanda , (1-235) 2008.

Fred A. Mettler, et al2009" "Effective Doses in Radiology and Diagnostic nuclear Medicine: A Catalog , " Radiology Vol. 248, No. 1, pp. 254-263, July 2008.

Fred A. Mettler, Jr., et al., "Effective Doses in Radiology and Diagnostic Nuclear Medicine: A Catalog," Radiology Vol. 248, No. 1, pp. 254-263, July 2008.

Herman, G. T., Fundamentals of computerized tomography: Image reconstruction from projection, 2nd edition, Springer, 2009.

Huda W. Radiation doses and risks in chest computed tomography examinations. Proc Am ThoracSoc 2007.

ICRP (International Commission on Radiological Protection) (1991) 1990 recommendations of the ICRP.Publication 60.Pergamon Press, Oxford ICRP (International Commission on Radiological Protection) (2001) Managing patient dose in computed tomography. Publication 88, Annals of the ICRP Vol 31, No2.Pergamon Press, Oxford

JessenkA , Panzer W. Shrimpton PC.et. al . EUR 16262:mEuropeanGuidkine on Quality Criteia Computed Tomography 2000 .

Jim Giles. Study warns of 'avoidable' risks of CT scans. Nature. 2004.

KosucuP.Ahmtoglu A, Koramasi,et al 2001. Low dose MDCT and visual.

Lois E. Romans, RT, (R)(CT) 2011Computed Tomography for Technologists: A Comprehensive Text

M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer (2009)Multislice CT 3rd Revised Edition Springer-Verlag Berlin Heidelberg

Mannudeep R.Ralra , MD , DNB et al 2004 ,strategies for CT radiation dose optimization , Radiology.

MannudeepR.Ralra , MD , DNB et ,strategies for CT radiation dose Optimization , Radiology 2004 .

Mathias_Prokop,_Michael_Galanski,_Aart_J._Van_Der(BookZZ.org) Mathias Prokop Michael Galanski Co-Editors Aartj. van der Molen Cornelia M. Schaefer-ProkopGeneral Radiation Protection Principles and Their Applicability to Medical X-Ray Diagnosis.(M. F. Reiser · C. R. Becker · K. Nikolaou G. Glazer 2009)

McHugh K. CT radiation doses.Arch Dis Child. 2003.

Mettler FA Jr, Huda W, Yoshizumi TT, Mahesh M. Effective doses in radiology and diagnostic nuclear medicine: a catalog. Radiology 2008.

Michael R. Bruesewitz, et. al., CT Dose Reduction and Dose Management Tools: Overview of Available Options, USA, 2006.

Michael R. Bruesewitz, et. al.,2006 CT Dose Reduction and Dose Management Tools.

Mona Idris (2012) Estimation of Radiation Hazards of Computed Tomography Dose in Khartoum State

Mona Taha Idris (2012): Estimation of Radiation Hazards of Computed Tomography Dose in Khartoum State

Podgorsak, E. B., ed. (2005). Radiation Oncology Physics: A Handbook for Teachers and Students. Vienna: International Atomic Energy Agency. November 2012.

(PAUL BUTLER, ADAM W. M. MITCHELL and HAROLD ELLIS 2007) Applied Radiological Anatomy for Medical Students.

Shapiro, J. Radiation Protection. 4th ed., Harvard University Press, 2002.

Shrimpton P. Assessment of patient dose in CT. Chilton, England: National Radiological Protection Board, 2004.

Siemens AG , Medical solution Hekestr , 127 , D-91052 Erlargen Germany . www. Siemensmedical.com .Siemens AG , Medical solution Computed Tomography siemensstr . 1.0-9/301 for Chheim Germany 2004 .

Siemens AG , Medical solution Hekestr , 127 , D-91052 Erlargen Germany . www. Siemensmedical.com .Siemens AG , Medical solution Computed Tomography siemensstr . 1.0-9/301 for Chheim Germany 2004 .

Smith-Bindman R, Lipson J, Marcus R, et al. (December 2009). "Radiation dose associated with common computed tomography examinations and the associated lifetime attributable risk of cancer". 2009

The 2007 Recommendations of the International Commission on Radiological Protection. Annals of the ICRP.ICRP publication, May 2012.

Turner, J.F. Atoms, Radiation, and Radiation Protection 3rd ed. Wiley, 2007.

Wiest PW, Locken JA, Heintz PH, Mettler FA Jr. CT scanning: a major source of radiation exposure. Semin Ultrasound CT MR 2002.

Appendix

WORKSHEETS

Patient doses measurements

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

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

Indication..

Previous examination1...findings...

Previous examination2...findings...

Interested organ..

Imaging Protocol,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,,1................................2.............................

3...

1. Phase*	Value	Dosimetry	
2. Tube Potential (kVp)		Displayed CTDIw	
3. Milliampere (mA) per rotation		DLP	
4. Time for one rotation (s) Or mAs per rotation		$CTDI_{Vol}$	
5. Slice thickness (mm)		7. Table feed (T mm)	
6. Slice Beam collimation (total detector number (N) and beam collimation (h))		8. Pitch	

PRC = Pre contrast, PTC = Post contrast, Other

Final diagnosis .....Pathology.......................................Normal.....................

Referring physician.......................qualification......................................

Comments...