Sudan University of Sciences and Technology

College of Graduate Studies

Optimization of Radiation Dose and Image Quality in Computed Tomography Imaging

أمثلة الجرعة الاشعاعية وجودة الصورة فى التصوير بالاشعة المقطعية

A Thesis submitted in Fulfillment of the Academic Requirement for the degree of Doctor of Philosophy (Ph.D.) in Medical Physics

By

Nissren Mohamed Ahmed Tamam
B.Sc. and M.Sc. Medical Physics

Supervisor:

Dr. Abdelmoniem Adam Mohamed
Associate Professor

Co-supervisor

Dr.Abdelrahman Mohamed Elnour

Assistant Professor

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قال تعالى :

(إِنَّ الَّذِينَ آمَنُوا وَعَمِلُوا الصَّالِحَاتِ إِنَّا لا نُضِيعُ أَجْرَ مَنْ أَحْسَنَ عَمَلا)

[الكهف :30]

Dedication

To My beloved family

AKNOWLEDGMENTS

First and foremost after God, I would like to point out It is a genuine pleasure to express my deep sense of thanks and gratitude to my supervisor Dr. Abdelmoniem Adam. His dedication and keen interest above all his overwhelming attitude to help his students had been solely and mainly responsible for completing my work.

I would like also to owe a deep sense of gratitude and thanks to my co supervisor Dr. Abdelrahman Mohamed El Nour, Dean of Radiological Sciences Alribat University, Department of diagnostic in Alamal Diagnostic Center, for his constructive input and support to the very end.

This dissertation would not be possible without the help of great many people in so many ways. I owe my gratitude thanks to all, participants in this research undertaking for their time and willingness to be part of this study.

Abstract

Improvements in the benefit of CT have been so dramatic that a tendency exists to the overuse. CT is a diagnostic imaging modality giving higher patient dose in comparison with other radiological procedures. While the benefits of CT exceed the harmful effects of radiation exposure in patients, increasing radiation doses to the population have raised.

The main objective of this work was to find an optimization approach to minimize the radiation dose to adult patients undergoing CT examinations, while maintaining the diagnostic image quality.

This study was done on four different CT scanners (2, 4, 16 and 64), in Khartoum state, during the period 2013-2016. One way to achieve optimization is to reduce tube rotation time, which has been shown to be effective in reducing absorbed dose to patients undergoing CT examinations.

A total of 404 CT patients' examinations 240 before and164 after optimization were included in the study. The results from this study indicate that radiation dose DLP was reduced significantly by (14.3%-59.7%) mGy.cm in Brain Protocol and by (1.1%-28.2%) mGy.cm in Chest Protocol and by (16.2%-55.4%) mGy.cm in Abdomen protocol for the four scanners. Image noise generally increases, subjective image quality was affected by an increased noise level in the images but was judged to be acceptable in all groups.Using this protocol, effective dose was reduced by (22.9%-47.0%) mSv in Brain and (2.6%- 25.3%) mSv in Chest and (15.6%-49.2%) mSv in Abdomen which in turns reduced the cancer probability. This study showed that optimizing the dose and image quality for the four CT scanners is dependent on choosing the appropriate parameter for the exam protocol. Finally, concerted efforts and research should be directed to define

diagnostic image quality, and research efforts must focus on patient- and technology- based methods to achieve a diagnostic- quality CT image at an optimum radiation dose. A team approach is essential in CT protocol review. And there is still considerable room for optimization and continuous developments of new technologies aim to optimize image quality and radiation absorbed dose to the patient.

المستخلص

ساهم التطور التقنى الملحوظ لجهاز الاشعة المقطعية فى زيادة الاستخدام . يعتبر جهاز الاشعة المقطعية الاعلى جرعة اشعاعية للمرضى مقارنة مع الاجهزة التشخيصية الاخرى.بالرغم من ان فوائد الاشعة المقطعية الكبيرة تتجاوز الاثار الضارة الناتجة من الاشعة المؤينة تظل الجرعة الممتصة للمرضى فى ازدياد.

تهدف هذه الدراسة للحصول على طريقة مثلى لخفض الجرعة الاشعاعية للمرضى الذين بخضعون للفحوصات عن طريقة الاشعة المقطعية مع الحفاظ على جودة الصورة التشخيصية .

اجريت هذه الدراسة فى الفترة بين2013الى 2016 وكانت لاربعة اجهزة اشعة مقطعية مختلفة فى 4 مراكز فى ولاية الخرطوم. والتي تتراوح موصفاتها بين الاقل (2 شريحة الى 64 شريحة) و اختير اكثر ثلاث فحوصات روتينية وهى الدماغ والصدر والبطن.الطريقة التى استخدمت لتحقيق الهدف هى تقليل زمن دوران انبوب الاشعة المقطعية والتى اظهرت فعاليتها فى خفض الجرعة الاشعاعية للمرضى.

شملت الدراسة مجموعة404 فحوصات مرضى يتراوح متوسط وزن يتراوح بين (55 الى 85) كيلوغرام وذوي تشابهه في الشكوى في الفحص الواحد, 240 عينة قبل و164عينة بعد امثلة الجرعة الاشعاعية.عندقياس الجرعة الاشعاعية بمفهوم DLP و $CTDI_{VOL}$ كوحدات اساسية لقياس الجرعة, تبين الاتى:بالنسبة للدماغ انخفضت الجرعة الاشعاعية بفعالية بمقدار يتراوح ما بين (13.3%-33.6%) ملى قرى و (14.7%-59.7%)ملى قرى.سم للاجهزة الاربعة.

وبالنسبة للصدر كانت النسبة بين(6.3%-34.1%) ملى قرى و (1.1%-28.2%)ملى قرى.سم للاجهزة الاربعة. وكان الانخفاض فى الجرعة بالنسبة لفحص البطن يتراوح بين(10.6%-30.1%) ملى قرى و(16.2%-55.2%) ملى قرى .سم. والتقليل فى الجرعة الممتصة ادى الى زيادة التشويش فى الصورة ولكنه لميكن بمقدار يمنع تشخيص الصورة.

باستخدام هذا البروتوكول، تم تخفيض الجرعة الفعالة من قبل الامثلة من (22.9 ٪ -47.0٪) ملي سيفرت في الدماغ و (2.6٪ - 25.3٪) ملي سيفرت في الصدر و (15.6٪ -49.2٪) ملي سيفرت في البطن، مما ادى الى تراجع في احتمالية الإصابة بالسرطان. وقد خلصت هذه الدراسة أن الاستفادة المثلى من جرعة وجودة الصورة لاجهزة الاشعة السينية المقطعية الاربعة تعتمد على اختيار للبروتوكول المناسب .

وأخيرا، ينبغي أن توجه البحوث لتحسين جودة الصورة التشخيصية مع امثلة الجرعة الممتصة للمريض وذلك باستخدام الاساليب التكنولوجية. والعمل الجماعي طريقة اساسية لوضع برتوكلات مختلفة لامثلة الجرعة الاشعاعية والحفاظ علي جودة الصورة.

List of Tables

List of Figures

List of Abbreviations

CT	Computed Tomography
3D	Three dimension
mSv	milli Sevirt
ICRP	International Committee of Radiation protection
UNSCEAR	United Nation Scientific Committee on the Effects of Atomic Radiation.
MSCT	Multi Slice Computed Tomography
DRL	Diagnostic reference level
Kvp	Kilo volt peak
UK	United kingdom
CT_{no}	CT number
ED	Effected dose
DLP	Dose Length Product
CTDI	Computed Tomography Dose Index
$CTDI_{vol}$	Computed Tomography Dose Index volume
mGy	milli Gray
$CTDI_{w}$	Computed Tomography Dose Index weight.
SD	Standard Deviation
NDRLs	National Diagnostic Reference Levels.
HU	Hounsfield Unit
SIU	System International Unit.
AEC	Automatic Exposure Control
RT	Rotation Time
SSTC	Single Slice Computed Tomography

Table of contents

Chapter One

Introduction

1.1: Introduction

Medical imaging provides tremendous and undeniable benefits for patients in modern health care. During recent years, substantial developments have been made in imaging techniques with progress continuing today (Söderberg, 2012).

The invention of computed tomography is considered to be the greatest innovation in the field of radiology since the discovery of X-rays. (Goldman, 2007).

Today, CT is one of the most important methods of radiological diagnosis. Since its inception in the 1970s, its use has increased rapidly. It is estimated that more than 62 million CT scans per year are currently obtained in the United States; including at least 4 million for children.

Recent data from the Organization for Economic Co-operation and Development(OECD,2011) and the IMV Medical Information Division (IMV,2012) show that 13.4 million more CT examinations were performed in 2011 compared to 2009 in the United States. The worldwide average annual per-capita effective dose from medical procedures has approximately doubled in the past 10-15 years. A study (IMV, 2012) has also found an uneven distribution of medical radiation exposure, which is greater in highly developed countries. For example, the 2006 United States data showed that medical imaging contributed to approximately half (3.0 mSv) of the total radiation dose (5.6 mSv). The greatest contributor to medical radiation exposure is CT. In the United States, the number of CT scans is increasing by approximately 10% per year (Fig 1.1).

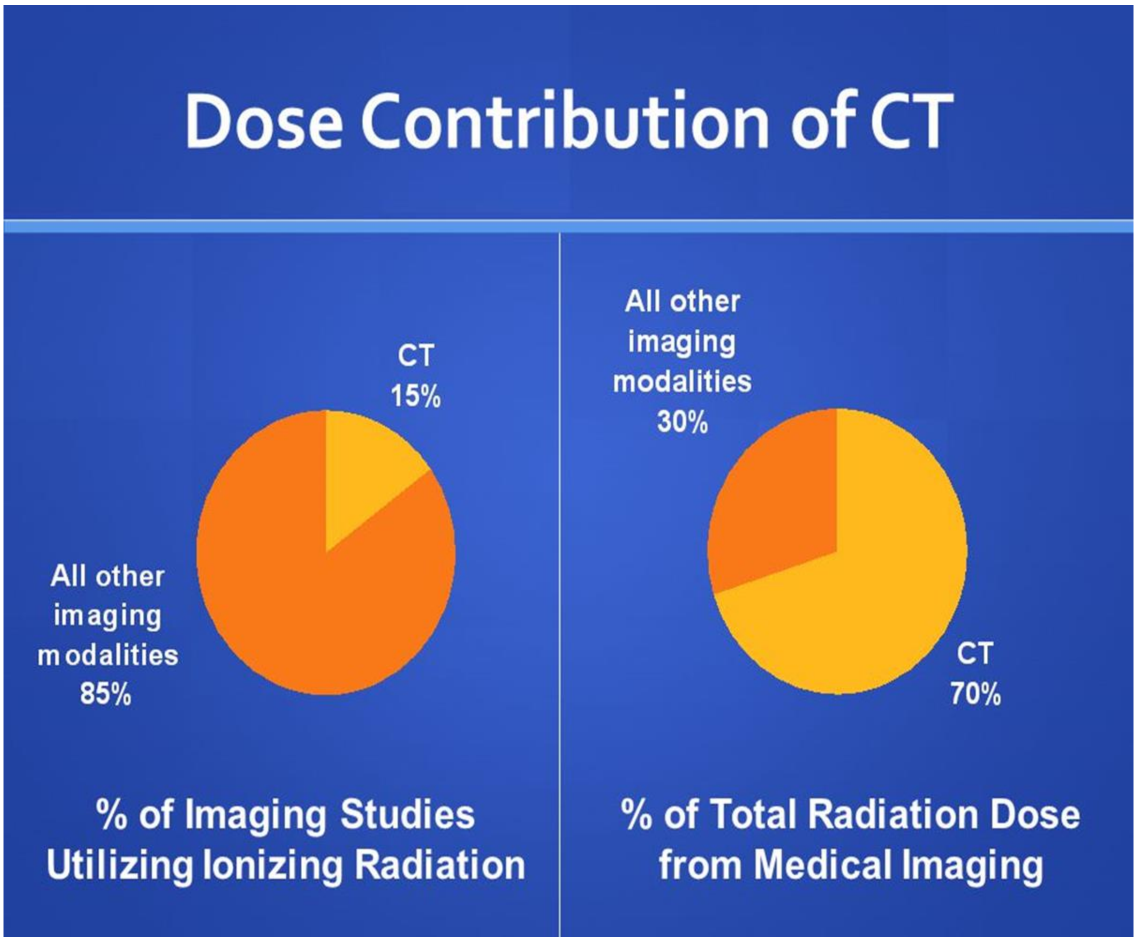


Fig 1.1: Shows the radiation dose received by a patient undergoing a CT examination, and national surveys generally show that this imaging technique, is the dominant contributor to medical radiation exposure (Mettler et.al, 2008).

Overall CT scans are the single most important contributor to the estimated worldwide collective effective dose from diagnostic imaging of approximately 4 million person-Sv/years (UNSCEAR, 2010).

Some reasons for this increased use of CT scans might be the use of CT in the follow-up of cancer patients, the use of CT at the emergency rooms to get an overview of the injuries, and an increasing use of CT on symptom free patients requesting examinations themselves (NRPA 2010; Bakke 2011). This sharp increase has been driven largely by advances in CT technology that make it extremely user-friendly for both the patient and the physician and much faster and wider scan coverage(Sutton,2008).However, as with any tool, the greatest benefit is derived from a combination of sufficient technical understanding and appropriate application (Fig 1.2).

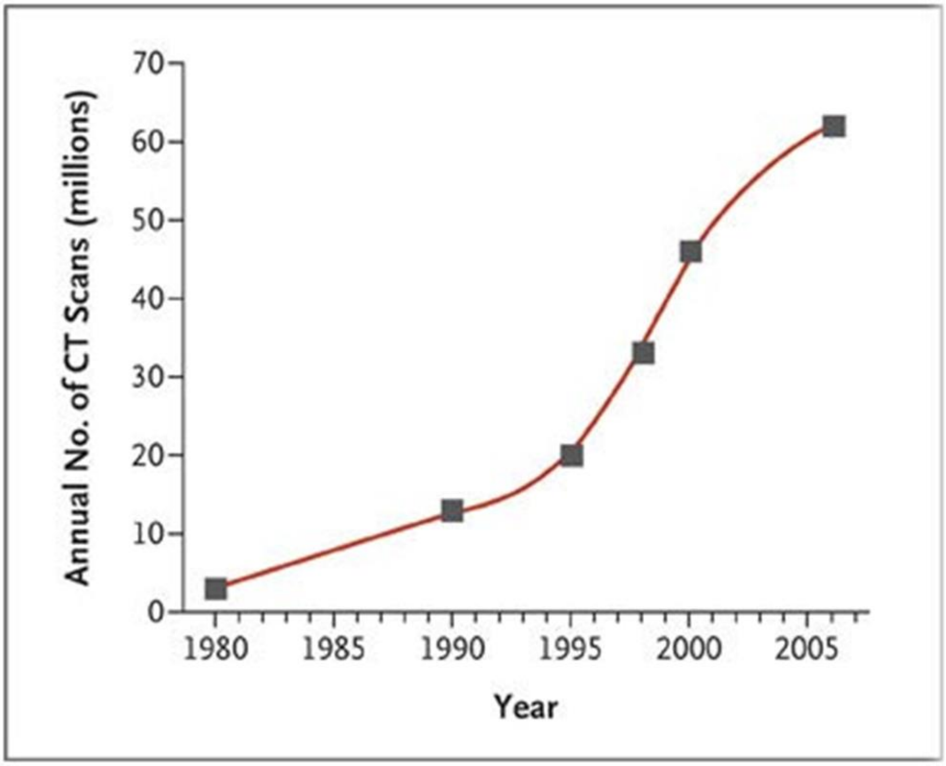


Fig 1.2: shows the sharp increase in the CT dose during time.

1.2: CT in Sudan

The first CT machines installed in Sudan in 1990 was single slice which from GE company. At last 20years was increased more than 30 machines of computerized tomography and in different specification tools and software applications, so this are increased the clinical used and replaced some radiological investigations. And lead to increased radiation dose to the patients so produced the needs justification optimization and how reduce the dose.

1.3: Radiation risks associated with CT scans

The main problem of the computed tomography its high radiation dose to the patient compared to other imaging modalities using x-rays. It delivers more than two thirds of the total radiation dose from all sources of radiological imaging using ionizing radiation. Because ionizing radiation is used in CT scans, and with the increased use of CT, the very small but finite cancer risk associated with CT scans has attracted greater attention in both medical physics and clinical societies (Kubo et al, 2009). There is now direct credible epidemiological evidence for a small risk of radiation-associated cancer at doses comparable to a few CT scans, or from other high dose radiological procedures(Hall and Brenner,2008) . Indeed, as early as 2002, the International Commission on Radiological Protection (ICRP) commented that: ‘‘The absorbed dose to tissue from CT can often approach or exceed the levels known to increase the probability of cancer’. (ICRP , 2002).

The association of ionizing radiation and cancer risk is assumed to be continuous and graded over the entire range of exposure, and approximately 29,000 future cancers have been related to computed tomography (CT) examinations performed in the United States in 2007. (De Mauri et al, JASN 2011). Radiation exposure should always operate under the‘‘As Low As Reasonably Achievable’’ (ALARA) principle and opportunities do exist in the CT field for collective dose reduction, both by reducing the numbers of CT scans and by reducing the doses per scan. Taking all these into consideration, as well as the continuous need to balance between the net benefits and the risks of using such a modality, various international organizations have published guidelines so as to standardize CT examinations and optimize radiation dose (Radiology Rounds 2003.Rehani et-al, Elnour, 2015).

1.4: Optimization

The gradually increasing awareness of radiation exposure mainly from CT examinations has forced manufacturers to develop techniques to reduce radiation doses. The implementation of these methods, as well as recommendations from authorities, requires close collaboration between medical physicists, manufacturers, radiologists, nuclear medicine physicians, technologists, and referring physicians in order to be effective. The challenge is to establish sufficient image quality for specific diagnostic task with the lowest effective dose to the patient (Lidinus, 2011). As CT utilization increases, the concern about radiation hazards from CT also increases (Goo, 2012).

Yet, an intrinsic problem of reducing radiation dose in CT examinations is magnification of noise and thereby loss of signal. However, modern CT technology includes advanced techniques for image reconstruction and dose reduction. During the last 30 years, manufacturers have developed new reconstruction techniques and several post processing tools to improve image quality. In spite of still being power- and potentially time-consuming, iterative reconstruction methods appear to improve image quality and thereby give potential for dose reduction .(Sæther et al, 2012) . The image quality and radiation dose is affected by the detector system, output from the X-ray tube, and the image reconstruction techniques, among other factors. So when talking about optimization we must think of balancing between Dose and Image Quality.

1.5: Objectives:

1.5.1:General objective:

The overall objective of this work was to investigate the potential of dose reduction and the possibility to maintain adequate image quality in CT.

The specific objectives were to:

- Measure patient dose during CT investigation using four different CT modalities.
- Optimize the radiation dose versus image quality for patients during CT examinations (Chest, Abdomen, and Brain, Pelvis, and CT angiography.
- Evaluate the role of continuous education in patient dose reduction.
- Study the effect of CT modality in dose optimization in clinical practice.

- Design a flowchart for referral criteria based on image quality and CT modality.
- Estimate the radiation risks.

1.6: Thesis outlines

This thesis is concerned with the optimization of CT dose with regards to diagnostic requirements on Image Quality for adult patients during Chest, Brain and Abdomen CT exams. Accordingly, it is divided into the following chapters:

Chapter one: is the introduction to this thesis. This chapter discusses the objectives and scope of work and introduces necessary background. It also provides an outline of the thesis.

Chapter two: contains the background material for the thesis. Specifically it discusses the dose for all absorbed dose measurements and calculations. This chapter also includes a summary of previous work performed in this field.

Chapter three: describes the materials and a method used to measure dose for CT machines and explains in details the methods used for dose calculation and optimization .Also it shows how to measure the noise for evaluation of the Image Quality.

Chapter four: reveals and demonstrates the results of this study. Additionally, the measurements obtained will be analyzed in order to determine if there exist significant trends between different scanners, patient-sizes and types of examination.

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

1.7: Thesis outcome

The following publications and conference registration are limited to those whicare based on work undertaken during the period of registration.

1.7.1: Publications:

1. **N. Tammam** , A. M. Elnour , H. Omer, A. Suleiman. *Rotation Time and Dose Reduction in Chest CT scans*. Scholars Journal of Applied Medical Sciences (SJAMS). 4(3) 1039-1041 (2016).
2. A. Sulieman, **N. Tammam**, K. Alzimami, A. M. Elnour, E. Babikir and A.Alfuraih . *Dose reduction in chest CT examination*. Radiation Protection Dosimetry Journal, 165(1-4):185-9 (2015).

1.7.2: Conference Presentations:

1. Khalid Alzimami, **Nissren Tammam**, Abdelrahman M. Elnour, Abdelmoneim Sulieman. Optimization of Radiation Dose in CT Chest Examination. *EPRBioDose 2013 International Conference / 24 –28 March 2013. Leiden, TheNetherlands.*

2. Abdelrahman M. Elnour, Abdelmoneim Sulieman Khalid Alzimami, **Nissren Tamam**, Optimization of Radiation Dose in CT Chest Examination. *RPM 2014,*

2nd International conference on radiation protection in medicine, 30.05-02.06, 2014.

Chapter Two

Theoretical Back ground

2.1: Types of Ionizing Radiation

Radiation is classified as either ionizing or non-ionizing according to its ability to affect matter at an atomic level, specifically whether or not it has sufficient energy to break chemical bonds and separate electrons from atoms.Ionizing radiation is furthermore divided into two separate groups, directly- and indirectly ionizing, based on the nature of the ionizing particle. Charged particles (electrons, protons, alpha particles) are included in directly ionizing radiation, as they carry sufficient energy to ionize or excite atoms and molecules. Uncharged particles (neutrons, photons), however, initiate direct ionizing radiation, but are not in themselves directly ionizing (IAEA, 2006) (SIS, 2011).

The thesis will focus on the subclass of electromagnetic radiation named X-rays, which are defined by extremely short wavelengths of 10^{-8} to 10^{-12} meters and having resulting energies in the range of 120 electron volts [eV] to 1.20 MeV as per Eq. 2-1(Encyclopedia Britannica, 2011).

$$Ep=h\upsilon \dots\dots\dots\dots\dots\dots\dots\dots\dots\dots\dots\dots \quad (2\text{-}1)$$

Where Ep is the energy of the photon, h is Planck's constant and υ is the frequency of the photon (Young, 2004).

X-rays carry the image information when acquiring a CT-image, in the form of absorption of photons.

2.2: X-ray Imaging and the X-ray Tube

The production of an X-ray image is the result of the successful detection of the incident photons, which initially hit the patient and subsequently pass through the patient without being absorbed. Conventional X-ray imaging goes back to Wilhelm Roentgen creating the first X-ray image in history on December 22, 1895. The basic principle behind X-ray imaging has not changed significantly since then. X-ray tubes used in modern scanners are still based on the same principle as the models used by Roentgen in the late 19th century, as shown on Figure 2.1 below.

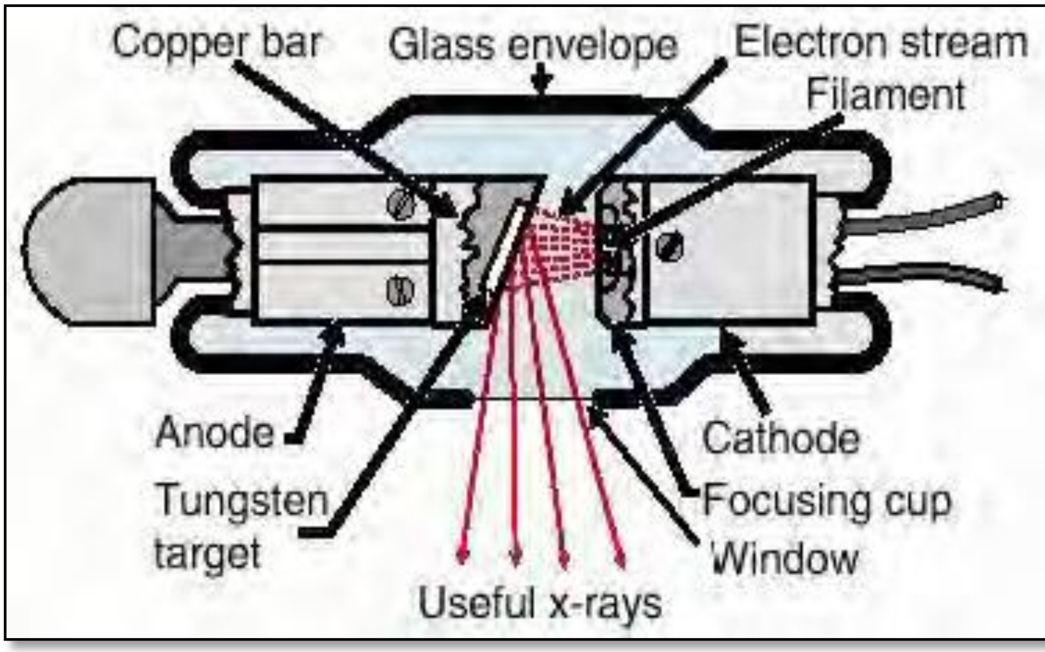


Fig 2.1: Showed the X-ray tube (Mikkel Oberg, 2011)

A glass envelope constitutes the exterior shell of the tube, with a vacuum inside. In this vacuum, a cathode emits a steady stream of electrons whose paths are controlled by use of a focusing cup. An anode is positioned directly opposite the cathode, with a metal target fastened to the anode. Normally copper or tungsten is used, either alone or in combination. A high voltage exists across the anode and cathode, usually in the magnitude of 30-100 kV, and as a result, the electrons wander towards the anode. Because of the vacuum, the electrodes do not interact with anything before reaching the metal target, with which they collide. Electrons are charged particles. They are directly ionizing radiation and will bring the atoms of the metal target to an excited state. This will result in the emission of X-rays.

These X-rays exit the tube through a window in the glass envelope, and are called characteristic X-rays, as their energy is characteristic for the type of metal target.(Mikkel Oberg , 2011).

2.3: History of CT and Evolution of Spiral Scanners

The term tomography stems from the Greek word "tomos" meaning "section". Scientists and mathematicians have described, "Body section radiography" in many different ways since the 1920's. It wasn't until the 1960's after much research, that the world's first CT scanner emerged. The inventor was Godfrey Newbold Hounsfield. He and Alan Cormack, a medical physicist, together developed and placed the first brain scanner into operation in 1971 for a company called EMI Ltd. In 1979, they were awarded the Nobel Prize in medicine and physiology.Initially data acquisition in CT scanning was very slow. The first experimental brain scan in 1967 took 9 days Fig 2.2. By 1971 they had reduced the scan time to 20 minutes.

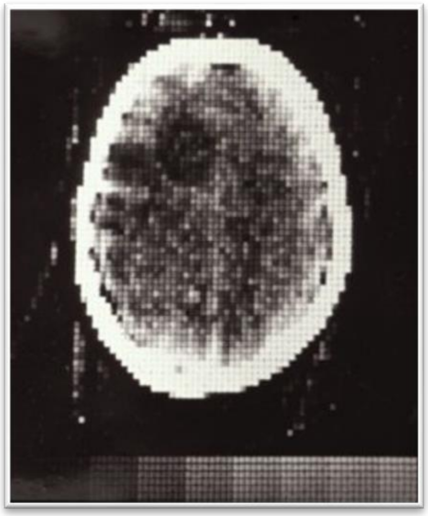


Fig 2.2: Showed the first clinical scan: Atkinson Morley's Hospital, October 1971

(Impactscan.org)

The basic designs of these CT scanners of the early days are illustrated in figures2.3.and 2.4.

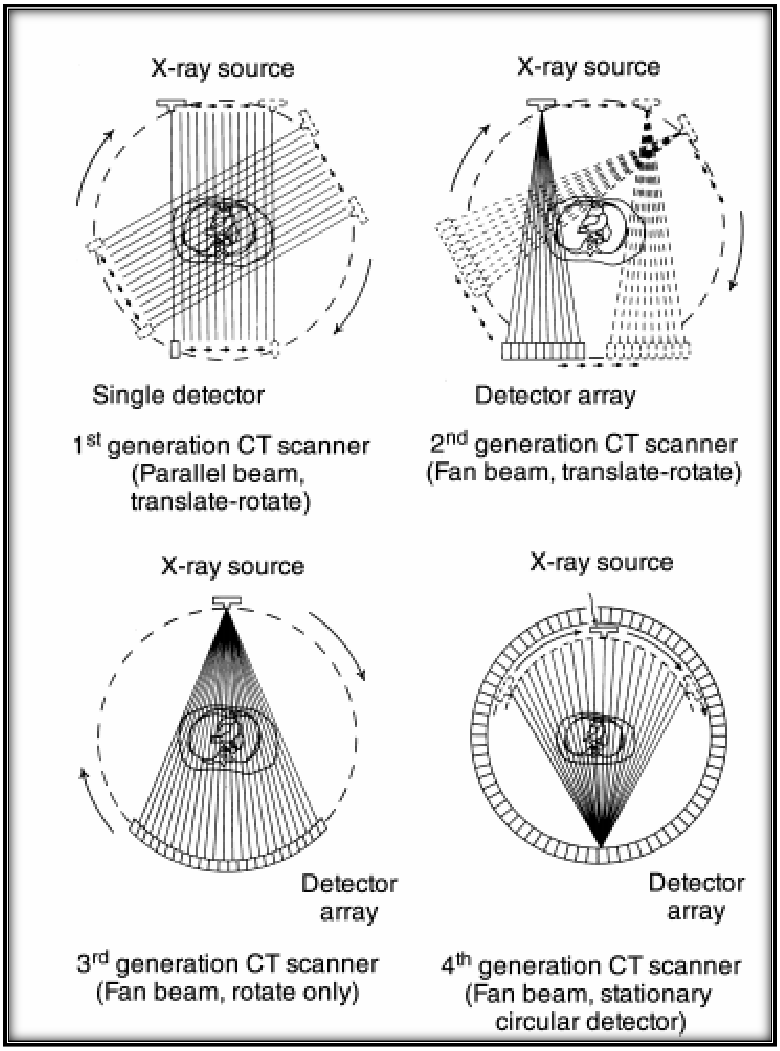


Fig 2.3: Illustrated the development of CT technology and the four generations (Robb, 1982)

2.4: Principles of Helical CT Scanners

In 1989, the helical (spiral) concept was considered one of the most significant developments in CT scanningthat finally allowed true 3D image acquisition within a single breath hold. This development meant continuous rotation of the x-ray tube without reversal between images.(Fig 2.5) (Kalender,et al, 1990). Three technological developments were required: slip-ring gantry designs, very high power x-ray tubes, and interpolation algorithms to handle the non-coplanar projection data (Beck, 1996).

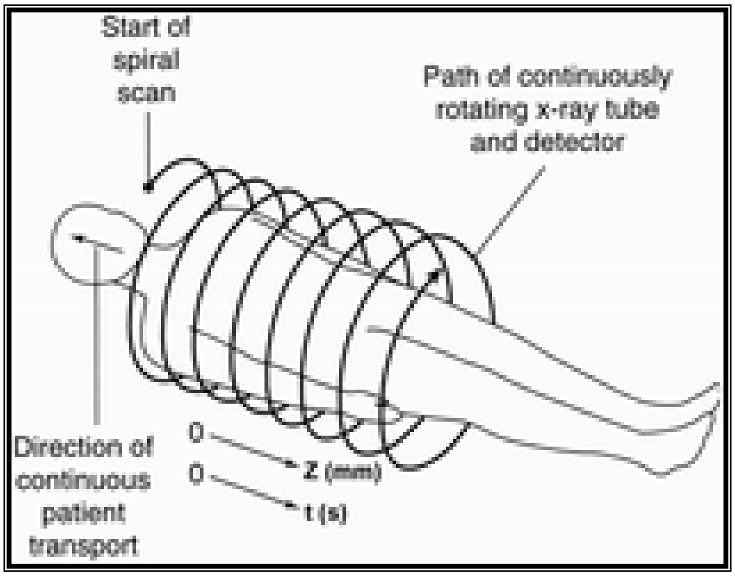


Fig 2.4: Principles of helical CT. As the patient is transported through the gantry, the x-ray tube traces a spiral or helical path around the patient, acquiring data as it rotates. t = time in seconds, from (Mahesh, 2002).

2.5: Slip-Ring Technology

The new continuous motion was given the name "slip-ring" technology.Slip rings are electromechanical devices consisting of circular electrical conductive rings and brushes that transmit electrical energy across a moving interface.The slip-ring design consists of sets of parallel conductive rings concentric to the gantry axis thatconnect to the tube, detectors, and control circuits by sliding contactors (Fig 2.6) and it reduced brain scan times to as low as 0.8 seconds. As technology continues to develop with multi-slice systems, times are getting even shorter (0.4 sec.).(Seerman, 1994),(Bushong, SC. 2001).

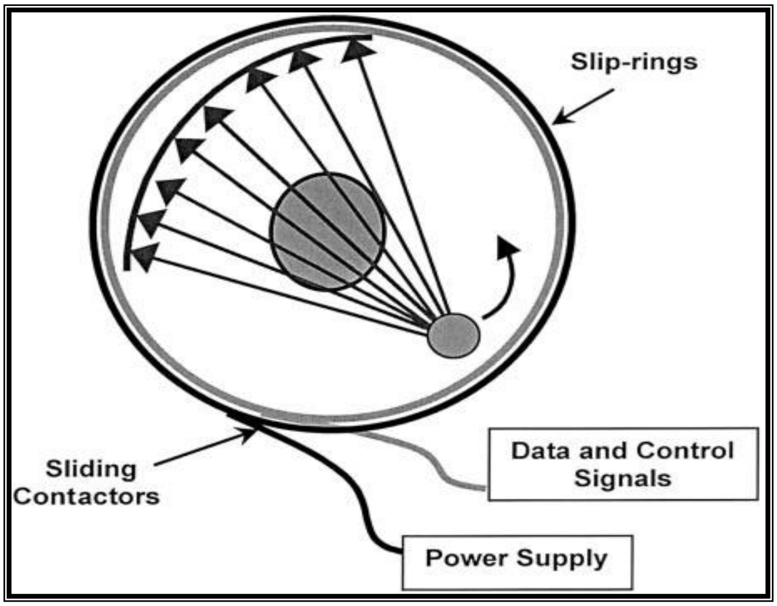


Fig 2.5: Diagram of the slip-ring configuration. Sliding contactors permit continuous rotation of the x-ray tube and detectors while maintaining electrical contact with stationary components (reference).

2.6: Multiple-Row Detector Helical CT

Multi-detector (or multi-slice) CT was introduced to maximize the effectiveuse of available x-ray beam. The x-ray beam is widened in the z-direction (slice thickness) and multiple rows of detectors were employed for dataacquisition for more than one slice at a time (Goldman,2008).Multi-detector CT

(MDCT) scanner differs from the single-slice CT scanner mainly in terms ofdesign of detector assembly (Figure2.7). The post-processing of the 'volume'data from these scanners was almost isotropic i.e. reformatted images in anyplane other than the original plane exhibit spatial resolution (in the z-direction)that is equal to that of the original images. MSCT thus allows ‘volume’ imaging of thepatient which can be later post-processed into the desired number of slices indifferent planes depending upon the clinical indication.For a single slice axial scanner, the detector unit will have over 700 elements arranged along an arc to intersect the exit beam of the tomographic plane. This is known as 3rd generation scan geometry and is the basic design for modern CT scanners. In multidetector CT scanners, the detector typically has additional adjoining arcs, or rows, of detector elements.

Such multi row detectors may have up to 128 rows, allowing a total acquisition width of 32–40 mm (measured at the isocenter). This type of acquisition can produce slicethicknesses varying from 0.5 mm to 10 mm. With such a detector, the acquisitiontime is reduced and the occurrence of motion artefacts is considerably reduced.

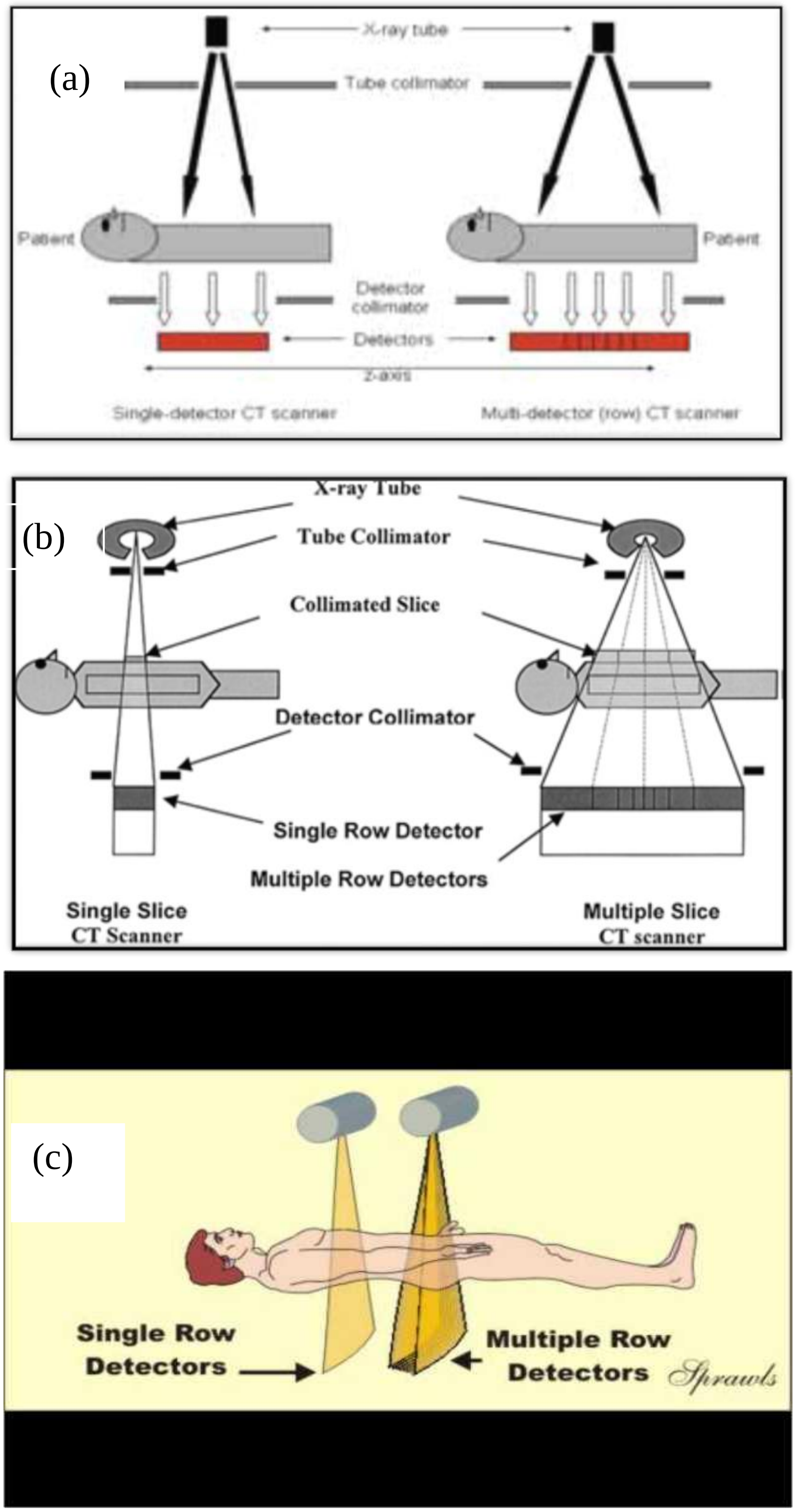


Fig 2.6: Diagrams (a,b,c) show the difference between single-row detector and multiple-row detector CT designs. The multiple-row detector array shown is asymmetrical and represents that of one particular manufacture (sprawls)

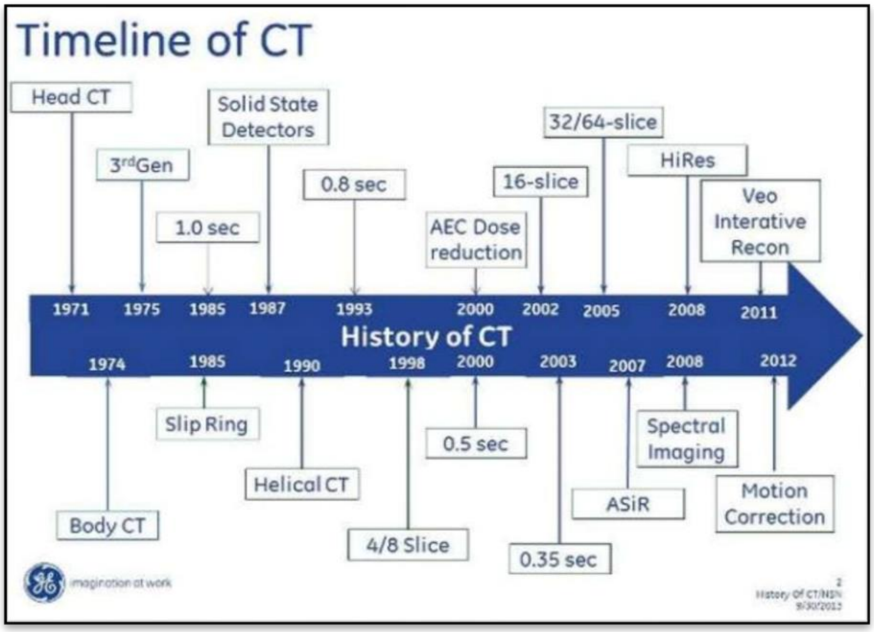


Fig 2.7: Showed the time line and the development in CT (GEhealthcare.com).

2.7: CT Basics

CT is a fundamental method for acquires and reconstructing an image of a thin cross section of an object. It differs from conventional projection in two significant ways: CT forms a cross sectional image, eliminating the superimposition of the structures that occurs in plane film imaging because of compression of three-dimensional body structures onto the two-dimensional recording system and, second the sensitivity of CT is subtle differences in x-ray attenuation is at least a factor of 10 greater than normally achieved by film screen recording system (American Institute of physics, 1994).

CT is the science that creates two-dimensional cross sectional images from three-dimensional body structures. CT utilizes a mathematical technique called reconstruction to accomplish this task. It is important for any individual studying the CT science to recognize that CT is a mathematical process. In a basic science, a CT image is the result of "breaking apart" a three-dimensional structure and mathematically putting it back together again and displaying it as a two-dimensional image on a television screen. The primary goal of any CT system is to accurately reproduce the internal structures of the body as two-dimensional cross-sectional images. This goal is accomplished by computed tomography's superior ability to overcome superimposition of structures and demonstrates slight differences in tissue contrast. It is important to realize that collecting many projections of an object and heavy filtration of the x-ray beam play important roles in CT image formation. Each component of a CT system plays a major role in the accurate formation of each CT image it produces (Reddinger, 1997).

2.8: Basic Principles of CT

Fundamentally, a CT scanner makes many measurements of attenuation through the plane of a finite-thickness cross section of the body. The system uses these data to reconstruct a digital image of the cross section, with each pixel in the image representing a measurement of the mean attenuation of a box like element (a voxel) that extends through the thickness of the section (Fig2. 8).(Mahesh, 2002).

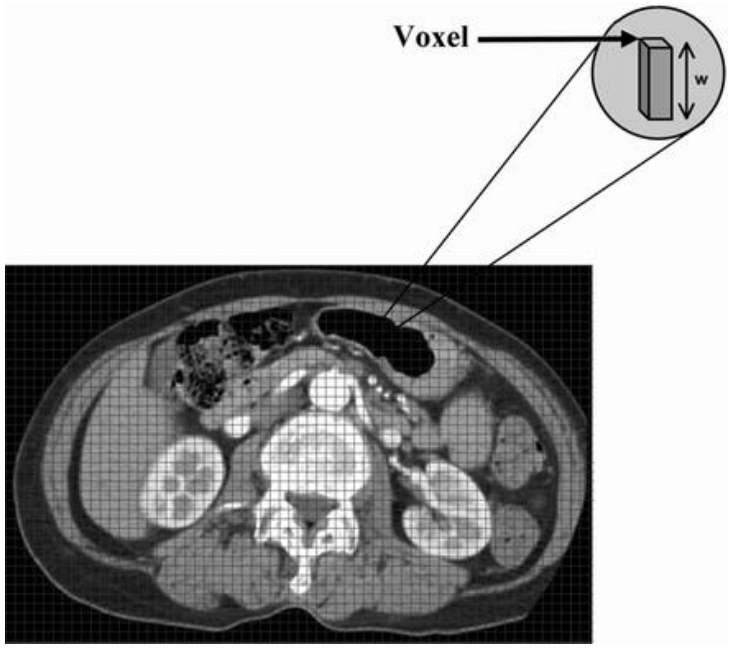


Fig 2.8: Sample CT image. A CT image is composed of pixels (picture elements). Each pixel on the image represents the average x-ray attenuation in a small volume (voxel) that extends through the tissue section. (In this example, the pixel size is exaggerated. In addition, in a real CT image, all tissues within a single pixel would be the same shade of gray), from (Mahesh, 2002).

An attenuation measurement quantifies the fraction of radiation removed in passing through a given amount of a specific material of thickness Δx (Fig 2.9, A). Attenuation is expressed as follows: where I_t is the x-ray intensity measured with the material in the x-ray beam path,I_0 is the x-ray intensity measured without the material in the x-ray beam path, and μ is the linear attenuation coefficient of the specific material.

To illustrate CT principles, any material can be considered as a stack of voxels along the beam path (Fig 2.9, B). Each attenuation measurement is called a ray sum because attenuation along a specific straight-line path through the patient from the tube focal spot to a detector is the sum of the individual attenuations of all materials along the path. If it is assumed that the ray path through the tissue is broken up into incremental voxel thicknesses Δx, the transmitted intensity is given by the following formula:

$$I_t = I_o e^{-\mu \Delta x} \quad \text{.......................... (2-2)}$$

This formula is expressed as the natural logarithm:

$$ln\left(\frac{I_0}{I_t}\right) = \sum_{i=1}^{k} \mu_i \Delta x \quad \text{.......................... (2-3)}$$

The image reconstruction process derives the average attenuation coefficient (μ) values for each voxel in the cross section by using many rays from many different rotational angles around the cross section. The specific attenuation of a voxel (μ) increases with the density and the atomic numbers of tissues averaged through the volume of the voxel and declines with increasing x-ray energy.

Mathematically, the attenuation value (μ) for each voxel could be determined algebraically with a very large number of simultaneous equations by using all ray sums that intersect the voxel. However, a much more elegant and simpler method called filtered back-projection was used in the early CT scanners and remains in use today (Naples S., 1995). Rays are collected in sets called projections, which are made across the patient in a particular direction in the section plane. There may be from 500 to 1,000 or more rays in a single projection. To reconstruct the image from the ray measurements, each voxel must be viewed from multiple different directions. A complete data set requires many projections at rotational intervals of 1° or less around the cross section. Back-projection effectively reverses the attenuation process by adding the attenuation value of each ray in each projection back through the reconstruction matrix. Because this process generates a blurred image, the data from each projection are mathematically altered (filtered) prior to back-projection, eliminating the intrinsic blurring effect. There are a number of advanced reconstruction techniques that are currently used in the CT image reconstruction process; however, these are beyond the scope of this article (Naples S., 1995).

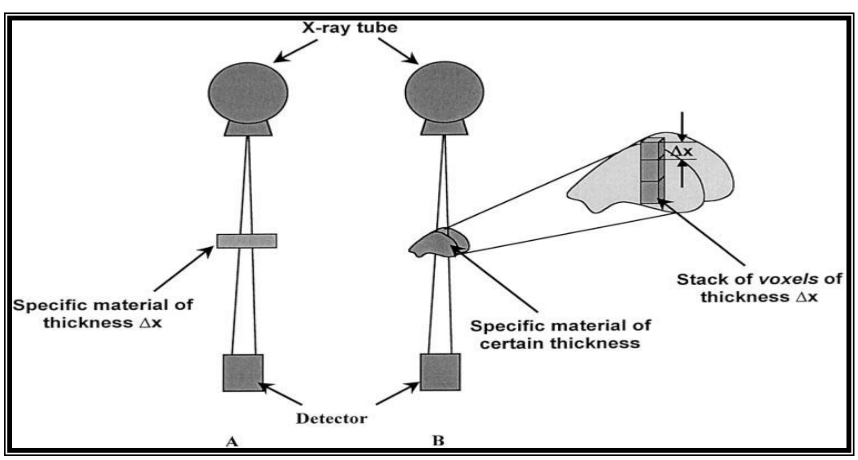


Fig 2.9: Principles of CT. Diagram shows the x-ray attenuation through a specific material of finite thickness (Δx) (Eq 2.1) (A) and through a material considered as a stack of voxels with each voxel of finite thickness (Δx)(Eq 2.1)(B), from (Mahesh,2002).

2.9: CT gantry: Tube, collimator, filters and detector

The large X ray tube located within the gantry (Fig. 2.11) operates typically at high voltage and high tube currentvalues for long periods of time, which requires the rapid dissipation of heat toavoid tube failure. The tube cooling system is designed to deal with this.However, it is essential that the ambient temperature around the scanner or heatexchanger be controlled by effective air conditioning to allow optimal operation.

The X ray beam, after leaving the tube, passes through filter material toremove low energy photons. Typically, specially shaped filters are then applied tocompensate for attenuation differences in a patient's head or body. It is essentialto use the correct filter for the correct body part. The slice width collimator, positioned at the filter exit, determines the width of the X ray beam. In modernscanners, multiple slices (currently up to 640) are acquired simultaneously. These scanners are known as multidetector, multislice or multirow CT scanners. Width of the beams for these acquisitions is the product of the individual slicewidth and the number of slices acquired simultaneously.

The X ray detector element is typically an ionization chamber using highpressure xenon or a scintillation detector. Early scanners used scintillationdetectors such as sodium iodide (NaI) or cadmium tungstate (CdWO4);

later highpressure xenon generally replaced these early materials and in later years scintillator doped ceramics have been used, such as gadolinium oxysulphide (Gd_2O_2S) or yttrium gadolinium oxide (YGdO). Important specifications for suchdetector elements, and factors in their development, include a high dynamicrange, high quantum absorption efficiency and a fast temporal response with lowafterglow.

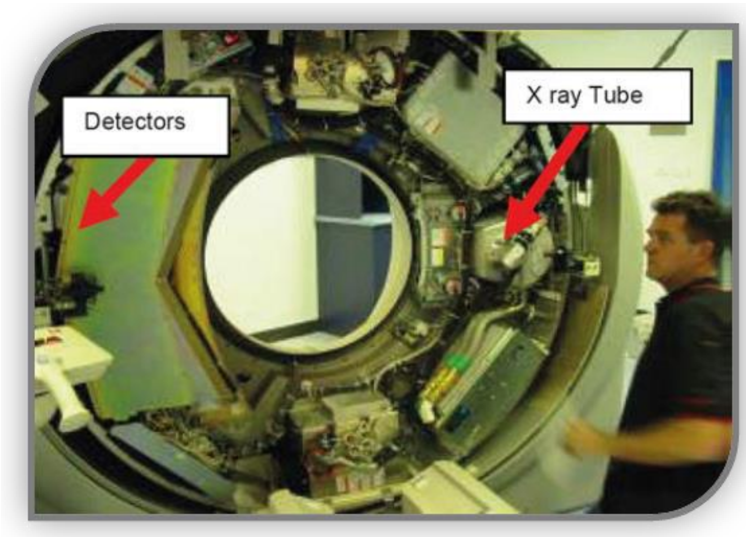


Fig 2.10: Showed the CT gantry from inside (slideshare.com)

After the introduction of multi-slice CT (MSCT) in 1997 (Hu, 1999), the number of Slices acquired per rotation have rapidly increased from 4 up to 8, 16, 32, 40, 64, 128, and 320 (Hsieh, 2009). The primary advantage of MSCT is improved temporal (<250 ms) and spatial resolution (<0.5 mm) and shorter scan times (Flohr and Ohnesorge, 2007).CT has undergone a tremendous technical development since the invention of the first CT equipment 1970s.

To assess radiation exposure tohumans and correlate it with the risk of exposure, mean absorbed dose in an organ or tissue is used (ICRP, 2007b). Based on the dose quantities prescribed by the ICRU and ICRP, the International Atomic Energy Agency (IAEA) has established an international code of practice for dosimetry in diagnostic radiology (IAEA, 2007).

2.10: The principles of CT dosimetry

When the X-rays penetrate the object, parts of its energy is absorbed by the object. The amount of energy imparted per unit mass at a point is expressed in terms of absorbed dose as defined by the International Commission on Radiation Units and Measurement (ICRU, 1998). The absorbed dose is the fundamental dosimetric quantity, and its unit is joule per kilogram, denoted as gray (Gy). For CT, estimates of absorbed doses to organs and tissues and effective doses arebased on two quantities: CTDI and dose-length product (DLP) (AAPM, 2008).

The CTDI concept was originally introduced for single slice axial scanning (Shope et al., 1981). CTDI represents the average absorbed dose along the z-axis (table feeddirection) from a series of contiguous irradiations. The most commonly used index is $CTDI_{100}$, which refers to absorbed dose in air or in cylindrical polymethylmethacrylate phantoms (15 cm in length) representing head (16 cm in diameter) and body (32 cm in diameter). The International Electrotechnical

Commission (IEC, 2009) has defined $CTDI_{100}$ as the absorbed dose integrated over a length of 100 mm for a single axial scan using a pencil ionisation chamber with an active length of 100 mm, divided by the collimated beam width (if n·T<100 mm) or 100 mm (if n·T ≥100 mm):

$$CTDI_{100} = \int_{-50\,mm}^{+50\,mm} \frac{D(z)}{\min\{n.T, 100\,mm\}} dz \text{ ....(2-4)}$$

where n is the number of slices per rotation, T is the nominal slice thickness, and D(z) is the absorbed dose profile along the z-axis.

To account for spatial variation ofthe absorbed dose in the scan plane (x, y), a weighted dose index ($CTDI_w$) was introduced (Leitz et al., 1995):

$$CTDI_w = \frac{1}{3} CTDI_{100_c} + \frac{2}{3} CTDI_{100_p} \text{ ....(2-5)}$$

To take axial scan spacing into account, CTDI by volume ($CTDI_{vol}$) was introduced (Bongartz et al., 2004):

$$CTDI_{vol} = \frac{CTDI_w}{pitch} \text{ ....(2-6)}$$

where pitch is defined as the ratio of the table transportation per rotation to thecollimated beam width (Silverman, 2001). $CTDI_{vol}$ is expressed in mGy and isdisplayed on the CT consoles. The $CTDI_{vol}$ is a measure of the radiation output of a CT scanner and represents an estimation of the average absorbed dose within the irradiated volume of an object of similar attenuation to the CTDI phantom. $CTDI_{vol}$

needs to be adjusted for patient size because it does not represent the average absorbeddose for objects of substantially different size or shape (AAPM, 2011).To better represent the overall energy delivered for an entire CT exam, DLPexpressed in mGy·cm was introduced (Bongartz et al., 2004):

$$DLP = CTDI_{vol} \times L \text{ ....(2-7)}$$

where L is the scan length. DLP is a measure of the total energy deposited in the phantom or patient. Quantity effective dose is the sum of weighted equivalent doses in the principaltissues and organs of the body (ICRP, 1991; 2007b). The different tissues and organs have been assigned a tissue weighting factor that reflect the radiosensitivity. The equivalent dose expresses the biological impact of a given type of radiation.Consequently, effective dose reflects the stochastic risk, such as cancer induction, andthe unit is sievert (Sv) (ICRP, 1991). Broad estimates of the effective dose can beobtained by multiplying DLP by a conversion factor (k) appropriate to differentanatomical regions (Bongatz et al., 2004; Huda et al., 2008; Shrimpton, 2004).

$$ED = DLP \times k \text{ .... (2-8)}$$

Where ED is Effective Dose, *k* is weighting factor

It is nearly impossible to measure the dose to individual organs directly during CT examinations, but it is possible to estimate the expected dose by use of phantoms as reference. In contrast to conventional X-ray imaging, in CT imaging the X-ray tube is rotated around the patient during exposure.It is necessary to know the relation between parameters, the dose for the individual scanner and its associated protocols.

2.11: Clinical Scanning Factors Affecting CT Radiation Dose

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

2.11.1: Tube current

The tube current [mA] influences the number of photons exiting the X-ray tube, as it determines the number of electrons leaving the cathode. The tube current is directly proportional to radiation dose, and as such is a prime parameter in adjusting the dose. Instead of tube current is sometimes used the tube-current-time-product [mAs], which is the tube current multiplied with the scan time. Taking into account the patient's body size makes it possible to achieve significant reductions in dose by reducing mAs.

2.11.2: Tube Voltage

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

2.11.3: Rotation Time

Patient dose is in principle proportional to rotation time when all other CT scan parameters remain constant.

The rotation time of the gantry [s] has decreased greatly over the last few decades, with modern scanners having a rotation time in the area of 0.4 seconds (Philips, 2011). The main consequence of the decreased rotation time is an increase in the noise and a reduction in absorbed dose. To avoid the noise, it is customary to increase the tube current accordingly (Kalra, M. K., 2004). CT, slice thickness, slice spacing, and helical pitch may affect dose as well. In single-slice CT with well-designed collimators, dose (as indicated by CTDI) is relatively independent of slice thickness for contiguous slices. Of course, the total length of the area scanned, as well as slice spacing, will determine how much total energy is deposited in the patient. For the same techniques, doses for helical scans with a pitch of 1.0 are equivalent to axial scans with contiguous slices. Pitches greater or less than 1 again affect CTDI values proportionally. Total Scan Length

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

2.11.4: Slice Thickness

In SSCT, with only a single row of detectors, the slice thickness [cm] is determined by simple collimation. The maximum slice thickness is limited by the width of the individual detector element (typically 10 mm (Kalra, M. K., 2004)), and by collimating the beam, this thickness can be decreased. In other words, the width of the beam is equal to slice thickness. In MSCT, the width of each individual detector element in the longitudinal direction determines the minimum slice thickness, and by merging multiple adjacent detector elements during detection, one can increase the slice thickness. This has a significant impact on image quality, as thin slices have better spatial resolution compared to thick slices, but lower SNR. To address the decrease in SNR, it is necessary to increase for instance the tube current, resulting in a significant increase in dose to the patient (Kalender, W.A., 2005). As an example, changing the slice thickness from 10 mm to 1 mm will increase the noise by a factor of 3.2, other factors held constant (McNitt-Gray, M.F., 2002).

2.11.5: Pitch

With the prevalence of helical MSCT, it is necessary to incorporate the incremental movement of the table, in relation to the irradiated area. This is defined as pitch, being the increment of the table per rotation, divided by the width of the beam. In Figure 2.11 below, a 4-slice MSCT is rotated twice around the patient, resulting in the acquisition of eight slices in pairs of two (indicated by color). The slices are in reality at an incline, as the patient is moving during exposure.

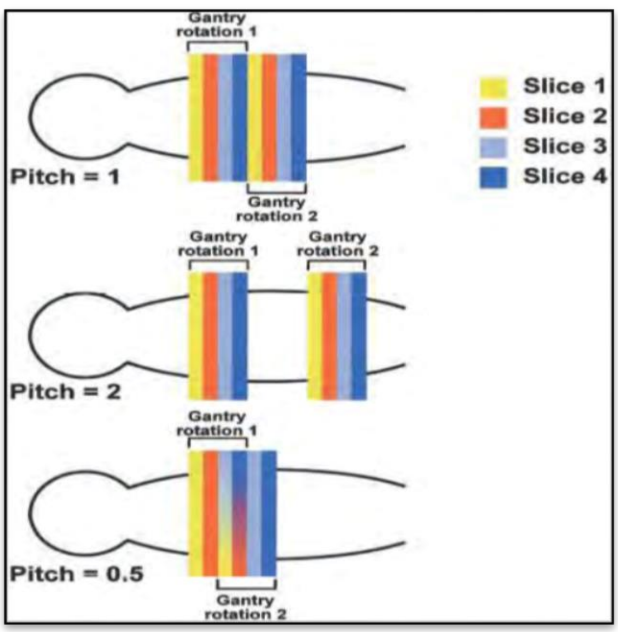


Fig 2.11: The effect of pitch on irradiated area, with a overlap for pitch < 1 (Cattin, P., 2010)

With pitch of 1, the last slice of the first rotation will be directly adjacent to the first slice of the second rotation, i.e. a distance of zero between them. With increasing pitch, this distance will increase. With pitch of 2 it is equal to the beam width, as the table has moved twice the beam width during a single rotation. This result in less irradiation of the patient, but the lack of full 360 degrees image date for all slices lowers the SNR. With pitch lower than 1, the slices will be overlapping, resulting in an increased dose to the patient as some areas are exposed multiple times. The SNR, however, improves as a result of overlapping image data.

Table 2.1: Adjustable Scan Parameters and Their Effect on Radiation Dose

Parameter	Effect on Radiation Dose
X-ray beam energy	Higher energy increases radiation dose (at matched tube current)
Tube current	Higher tube current increases radiation dose
Gantry rotation	Faster gantry rotation decreases radiation dose
Section thickness	Thinner collimation is linked with increased dose
Pitch	Higher pitch decreases radiation dose (at matched tube current)
Distance of x-ray tube to CT isocenter	Optimal patient placement decreases radiation dose
Scan length	Lengthening the scan range increases radiation dose

2.12: CT Image Quality

Fundamentally, image quality in CT, as in all medicalimaging, depends on 4 basic factors: image contrast, spatialresolution, image noise, and artifacts. Depending on the diagnostictask, these factors interact to determine sensitivity (the ability to perceive low-contrast structures) and thevisibility of details

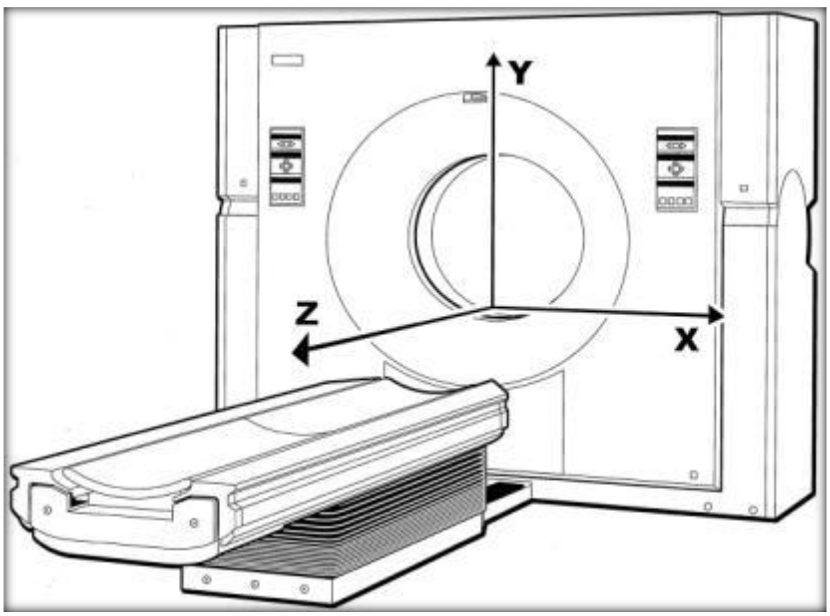


Fig 2.12: Coordinate system used for CT imaging (Morinet al 2003)

The objectives in CT development have changed from increasing the number of slices to focusing on improvements in X-ray tube performance, detector efficiency, and data processing (Fleischmann and Boas, 2011). The technology has provided further improvements in scan speed and temporal resolution.

2.13: AEC in CT

A clinical CT examination often covers different anatomic regions with variableattenuation. Because the selected tube current normally is based on the region withthe highest attenuation, such as the shoulder and pelvis, or the region that requiresthe highest image quality, the tube current is usually set to a high level when an AEC system is not used. Standard protocols are usually established to generate good quality images for average patient sizes. Thus, if an AEC system is not used, smaller patients will be exposed to unnecessarily high doses of radiation and images of larger patients may be of lower quality. AEC systems were developed to enable tube current modulation according to a patient's size, shape, and attenuation, and to improve the consistency of image quality among patients.

2.14: The Necessity of Optimization

Optimization is the process of maintaining diagnostic quality while minimizing the ionizing radiation dose required to capture an image (Mansson LG et al, 2005). The increasing exposure to radiation from CT has been of concern for some years and is now receiving more attention from health professionals, authorities, manufacturers, and patient groups. The number of publications on radiation exposure in CT, and management thereof, has since seen a yearly increase. Manufacturers whose main focus had been on reducing scan time started to put radiation exposure reduction on their agenda. In recent years, improved management and optimization of radiation exposure in CT has been high on the agenda for all CT manufacturers.

Previously published studies (Jaffe TA 2009, Kulama E. 2004, Muhogora WE, 2009, McRobbie DW. 2001, Shrimpton PC, 2003, Wall BF. 2004) have focused on the technical parameters that may be modified in establishing an efficient radiation dose for image capture. The tests were selected to verify if the scanner is technically adequate, if preprogrammed patient protocols are up-to-date, and if exposure values displayed at the console are sufficiently correct. In addition, they will ensure that the participating medical physics expert (MPE) gets a full understanding of the system to enable him/her to guide optimization processes and allow automated patient dosimetry. It is expected that major optimization studies will be triggered by annual testing based on the new documents.

CT examinations are among the highest-dose procedures encountered routinely in medical imaging. The qualitative criteria for acceptability in RP 162 address some functional and operational issues, and the quantitative criteria, in the form of suspension levels, focus primarily around hardware aspects of the CT scanner, though consideration is also given to software, operator aspects and selection of scan protocols. Some of the specific aspects and challenges in modern CT systems, in particular multi-slice and wide beams are also addressed.

However, nobody really knows exactly what this low-dose radiation does to the patient. Now, researchers from Stanford University have shown that cellular damage is detectable in patients after CT scanning. The team state that the new study has shown that even exposure to small amounts of radiation from computed tomography scanning is associated with cellular damage. They go on to add that whether or not this causes cancer or any other negative effect to the patient is still unclear, however, these results should encourage physicians toward adhering to dose reduction strategies. The study is published in the Journal of the American College of Cardiology: Cardiovascular Imaging.

In last few years widespread concerns were raised regarding the radiation dose from CT imaging and its impact over population as claimed by Brenner DJ et al 2007. This issue has initiated a sort of ‘dose war’ in the CT technology and manufacturers now claim to generate the desired image quality at lower radiation dosage for a particular imaging study. This is indeed a positive step and the use of radiation for diagnostic imaging should be essentially based on rationalism.

CT image quality, as in most imaging, is described in terms of contrast, spatial resolution, image noise, and artifacts. A strength of CT is its ability to visualize structures of low contrast in a subject, a task that is limited primarily by noise and is therefore closely associated with radiation dose: The higher the dose contributing to the image, the less apparent is image noise and the easier it is to perceive low-contrast structure Heggie JC, Kay JK, Lee WK. Importance in optimization of multi-slice computed tomography scan protocols. They discussed the type of optimization of multi-slice scan protocols that may be undertaken to keep patient doses to acceptable levels without compromising image quality.

Beeres M et al. (2014) found that faster CT gantry rotation reduces scan time and motion artifacts. However, accelerating rotation time increases image noise and streak artifacts. Applications of reduced- dose CT for specific clinical indications other than detection of pulmonary nodules. Reduced-dose CT was reported to be useful in follow-up chest CT of oncology patients.

As the number of computed tomography (CT) procedures performed worldwide continues to increase, there is growing concern about patient protection issues. Currently, no system is in place to track a patient's lifetime cumulative dose from medical sources, and questions have arisen regarding the possible threat to public health from the widespread use of CT.

In this part, the author reviewed the published literature to determine whether patients are receiving a higher absorbed dose of radiation and explored several proposed models to optimize the radiation dose delivered to patients and track cumulative lifetime dose.

A recent study by Aldrich and Williams quantified changes in numbers of radiology exams in order to examine the correlation to the radiation dose received by the patient. In addition to a 4-fold increase in CT exams, they also found that the average annual effective dose per patient almost doubled during the study period, from 3.3 mSv in 1991 to 6.0 mSv in 2002. CT is the largest contributor to patient dose in radiology. This could be because more CT scanners are in use and their performance has been enhanced, along with increasing indications for CT exams.

CT is not the only modality that has experienced more use and has the potential to deliver higher patient radiation doses. It drew attention to the fact that optimizing technique and standardizing practice could benefit the field of radiology and protect patients from overexposure to ionizing radiation. Although not pivotal to the discussion of correlating increased use of CT to an increased patient radiation dose (Sodickson,2001) study calls attention to the fact that dose to the patient can be reduced by careful attention to technique and optimization.

Yoshizumi and Nelson (Vano et al., 2002) pointed out the need to balance optimization of image quality against radiation dose in developing clinical protocols. Their study described fundamental concepts of radiation dose in detail, including the CT dose index and other technical factors such as pitch effect, dose profile in the penumbra and signal-to-noise ratio. Yoshizumi and Nelson concluded that multi-detector CT (MDCT) radiation dosimetry issues have not been addressed adequately and have lagged behind advances in the actual technology.

Other researchers also are questioning the effect of newer imaging technologies on patient radiation dose. Berland and Smith (Makayama et al.,2001) proposed that the absorbed dose could be up to 40% higher using MDCT compared with older generation scanners. Golding and Shrimpton suggested that "evidence indicates a strong trend of increasing population dose owing to rising use of CT and to increased dose per examination." A significant body of literature focuses on discovering a causal link between increased use of the CT scanner and an increase in radiation absorbed dose to the patient population.

Numerous studies have suggested that, although CT is not the most commonly performed radiologic examination, it is the largest source of radiation dose. Nagel et al found that, although CT represents only about 4% of all radiologic examinations, it is responsible for up to 35% of the collective radiation dose to the population from radiologic examinations. In a related National Cancer Institute report, data suggested that the use of CT in adults and children has increased approximately 7 fold in the past 10 years. In large U.S. hospitals, CT represents 10% of diagnostic procedures and accounts for approximately 65% of the effective radiation dose for all medical examinations.

Aldrich et al (ICRP, 2000) conducted a study to compare the dose length product and effective radiation dose to patients from CT examinations. They compared data from 1070 CT exams and concluded that considerable variation existed in the dose length product and patient radiation dose for a specific exam. This study called attention to the need to optimize the effective dose to the patient and to conduct more research to determine which additional efforts are needed to minimize patient exposure. Optimizing technical factors for exams can help reduce the patient radiation dose, thereby reducing risks.

A pivotal study by (Lee et al., 2000) assessed awareness levels among patients, emergency department physicians and radiologists concerning radiation dose and the risks involved with CT scans. Lee and colleagues concluded that patients were not given information about the risks, benefits and radiation dose for a CT scan. Regardless of their experience levels, few of the participants in the study (including the emergency department physicians and the radiologists) were able to provide accurate estimates of CT radiation doses. This study underscores the prevalent lack of attention to the issue of lifetime cumulative radiation dose. This must become a central issue so that risk can be studied and monitored. One disadvantage to communicating the risk of a cumulative radiation dose would be the natural instinct of some patients to defer or cancel the exam. Professionals should highlight the benefits of the examination when discussing risks with the patient. Physicians improve their understanding of radiation risks from medical imaging exams.

Amy K et al.,2003 evaluate the image noise, low-contrast resolution, image quality, and spatial resolution of adaptive statistical iterative reconstruction in low-dose body CT.

Adaptive statistical iterative reconstruction was used to scan the American College of Radiology phantom at the American College of Radiology reference value and at one-half that value (12.5 mGy). Test objects in low- and high-contrast and uniformity modules were evaluated. Low-dose CT with adaptive statistical iterative reconstruction was then tested on 12 patients (seven men, five women; average age, 67.5 years) who had previously undergone control-dose CT. Two radiologists blinded to scanning technique evaluated images of the same patients obtained with control-dose CT and low-dose CT with and without adaptive statistical iterative reconstruction. Image noise, low-contrast resolution, image quality, and spatial resolution were graded on a scale of 1 (best) to 4 (worst). Quantitative noise measurements were made on clinical images.

In the phantom, low- and high-contrast and uniformity assessments showed no significant difference between control-dose imaging and low-dose CT with adaptive statistical iterative reconstruction. In patients, low-dose CT with adaptive statistical iterative reconstruction was associated with CT dose index reductions of 32–65% compared with control imaging and had the least noise both quantitatively and qualitatively ($p < 0.05$). Low-dose CT with adaptive statistical iterative reconstruction and control-dose CT had identical results for low-contrast resolution and nearly identical results for overall image quality (grade 2.1–2.2). Spatial resolution was better with control-dose CT ($p = 0.004$). These preliminary results support body CT dose index reductions of 32–65% when adaptive statistical iterative reconstruction is used. Studies with larger statistical samples are needed to confirm these findings.

Sodickson A et al 2001, estimate cumulative radiation exposure and lifetime attributable risk (LAR) of radiation-induced cancer from computed tomographic (CT) scanning of adult The cohort comprised 31- 462 patients who underwent diagnostic CT in 2007 and had undergone 190 712 CT examinations over the prior 22 years. Each patient's cumulative CT radiation exposure was estimated by summing typical CT effective doses, and the Biological Effects of Ionizing Radiation (BEIR) VII methodology was used to estimate LAR on the basis of sex and age at each exposure. Thirty-three percent of patients underwent five or more lifetime CT examinations, and 5% underwent between 22 and 132 examinations. Fifteen percent received estimated cumulative effective doses of more than 100 mSv, and 4% received between 250 and 1375 mSv. Associated LAR had mean and maximum values of 0.3% and 12% for cancer incidence and 0.2% and 6.8% for cancer mortality, respectively. CT exposures were estimated to produce 0.7% of total expected baseline cancer incidence and 1% of total cancer mortality. Seven percent of the cohort had estimated LAR greater than 1%, of which 40% had either no malignancy history or a cancer history without evidence of residual disease.

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. Smith A, et al, 2003) quantified retrospectively the effect of systematic use of tube current modulation for neuroradiology computed tomographic (CT) protocols on patient dose and image quality.

The authors evaluated the effect of dose modulation on four types of neuroradiology CT studies: brain CT performed without contrast material (unenhanced CT) in adult patients, unenhanced brain CT in pediatric patients, adult cervical spine CT, and adult cervical and intracranial CT angiography. For each type of CT study, three series of 100 consecutive studies were reviewed: 100 studies performed without dose modulation, 100 studies performed with z-axis dose modulation, and 100 studies performed with x-y-z–axis dose modulation. For each examination, the weighted volume CT dose index ($CTDI_{vol}$) and dose-length product (DLP) were recorded and noise was measured. Each study was also reviewed for image quality. Continuous variables ($CTDI_{vol}$, DLP, noise) were compared by using *t* tests, and categorical variables (image quality) were compared by using Wilcoxon rank-sum tests.

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

Use of dose-modulation techniques for neuroradiology CT examinations affords significant dose reduction while image quality is maintained.

Finally, a unique study conducted in Sudan regarding patient dose in CT (Gala, 2007). The study assessed the radiation doses for patients undergoing control ct examinations in four centers in Khartoum state for various CT examinations of head, neck, abdomen, pelvis and chest; $CTDI_{vol}$, DLP and Effective Dose were calculated using CT-expo software. The mean $CTDI_w$, $CTDI_{vol}$, DLP and effective do dose were found to be 32.6mGy, 26.5 mGy, 454mGy.cm and 3.3mSv respectively (Honef et al 2004).

Yamada et al., noted no difference in terms of detectability of abnormal findings between standard- dose (140 kVp, 96 mAs) and low-dose (140 kVp, 45mAs) CT images. Chiu et al., found almost perfect concordance in image interpretation between standard-dose 120 kVp, 240 mAs with contrast dose were found to be 32.6 mGy, 26.5 mGy, 454 mGy and 3.3 mSv respectively.

Enhancement and reduced-dose (140 kVp, 43 mAs without contrast enhancement) CT Dinkel et al investigated CT in follow-up studies of lymphomas and extra pulmonary primary tumors using a low-dose protocol (15 mAs at 120 kVp) and a standard-dose protocol (150 mAs at 120 kVp). Although disease conspicuity decreased in the lung apex and mediastinum, detectability of lesions was not affected in the reduced-dose CT images.

Chapter Three

Materials and Methods

This chapter presents the work on collecting the data and the subsequent sorting into relevant and manageable categories, which is a requirement for proper statistical analysis. Having sorted and uniformed the data, the chapter concludes with a statistical analysis programs of relevant phenomena and tendencies within the data. This will be used as basis for the revised guideline in the subsequent chapter, in accordance with the defined goal. All data used in this thesis is from January 2014-October 2015.

3.1: Computer Tomography Scanners

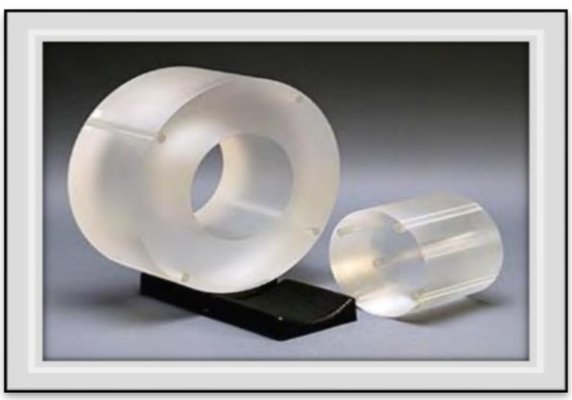
Table 3.1 showed the different type of the CT-Scanners used

CT scanner(slices)	**Modality**
2(Dual)-Slice	Somtom Siemens
4-Slice	Siemens
16-Slice	Siemens
64-slice	Toshiba Aquillon

Data from thousands projections around the patients is collected for each rotation to create CT images. This type of data is denoted as raw data i.e. data that has not been processed yet. All raw data used in this thesis originated from four different CT scanners modalities.

Dual- slice Siemens somtom this scanner with rotation time 1.5- 1.0 Sec. And , 4-slice scanner with rotation time 1.0, 0.75, 0.5 Sec. The 16- slice scanner with minimum rotation time 0.5 Sec. And 64- Toshiba – Aquillon scanner with minimum rotation time 0.4 Sec.

All quality control test were performed and done to the machine prior any data collection. These tests were carried out by experts from Sudan Atomic Energy Commission (SAEC) (Fig3.1a, b). Below are some pictures of callibration for dual and 16- slice CT scanners.



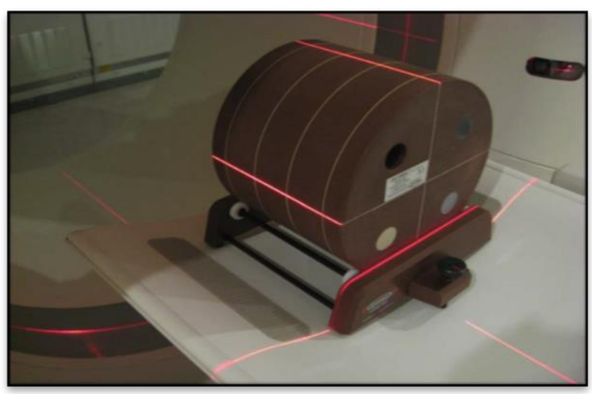


Fig 3.1a: Standard two-part Plexiglas phantom, with a large body-phantom and a smaller head-phantom

Fig 3.1b: Quality control for a CT scanner

3.3: Collection and Sorting of Data

A total of 384 patients were divided into two groups one before optimization and the other after optimization .These data were collected from four big diagnostic centers for the common exams (Brain, Chest and Abdomen). The choice of these examinations was pragmatic in that they were commonly performed at the majority of centers, and thus it was likely that representative numbers examinations would be achieved during the required data periods. For each examination, the following parameters were entered: kV, mAs (tube current), Gender, Height, Weight $CTDI_{vol}$ and DLP. These values were very valuable for statistical purposes, as they might possibly allow for analysis of scanner dependency. Gender, height and weight were normalized to m/f+ (male/female). We tried in purpose to collect data from standard patients (65-85 kg). The patient's

Body Mass Index (BMI) was calculated. BMI is defined as the weight in kilograms divided by the square of the height in meters, and has the unit of [kg/m^2]. The collection of patient exposure parameters was done using patient dose data sheet, as added in the appendix.

3.4: Absorbed Dose measurements

$CTDI_{vol}$ and DLP were obtained from the CT scanner directly. Then the data was sorted and every scanner with its three exams was tabulated together in addition of the absorbed dose and the reduction. So in this study the rotation time was decrease (tube speeding) according to the capability of every scanner Table 3.1.

Table 3.2: Showed the rotation time for every CT scanner according to number of slice before and after optimization.

CT scanner(slice)	Rotation time before optimization (sec)			Rotation time after optimization(sec)		
2 slice	1.5	1.0	1.0	1.0	0.7	0.7
4-slice	0.7	1.0	0.7	0.5	0.7	0.5
16-slice	0.7	0.7	0.7	0.5	0.5	0.5
64-slice	0.7	0.6	0.6	0.5	0.5	0.5

3.5: Image Quality (Noise) measurements

The SD indicates the magnitude of random fluctuations in the CT number and thus is related to noise: The larger the SD, the higher the image noise.

3.6: CT dose optimization strategies steps

Changes in Protocols:

• Scan (rotation) time: Changing the scan time changes the duration of each measurement—and thus the number of detected x-rays—proportionally. Because amperage and scan time similarly affect noise and patient dose, they are usually considered together as mA · s, or mAs.

3.7: Evaluation of Image Quality

There are several ways to evaluate image quality in medical imaging systems. In this study we measure Image Noise .Which is expressed as a standard deviation of the measured density valuesof the CT numbers for the enclosed pixels (in Hounsfield units, HU) within a selected ROI in an image.

3.8: Analysis of the Data

All the collected data was analysis using SPSS, continuous variables (age, body mass index (BMI), $CTDI_{vol}$, DLP, effective dose (E), image noise (SD) were presented in means ± standard deviation, median (range), reduction and increase (%).Categorical variables (number of studies and gender) were present in number. Comparison between before and after implementation groups was done using student t-test.

Chapter four

Results and Analysis

The results of optimize the Radiation Dose for the four different CT scanners after selecting the suitable imaging technique in computed tomography CT are given in Tables 4.1 to 4.14.The data was analyzed using Statistical Package for the Social Sciences (SPSS) version. 16.0 Chicago, Illinois, USA, SPSS Inc.). Descriptive statistics, bivariate statistics (t-test, ANOVA). The following statistical methods were used: Mean, Std. Deviation, Maximum, Minimum, Range, Test (One Way ANOVA): to determine the level of significance of the differences in the variables (Age, kVp, mAs, DLP, $CTDI_{vol}$ and E dose) .CT exams protocols are used to obtain the diagnostic image quality required, while minimizing radiation dose to the patient and ensuring the proper utilization of the scanner features and capabilities.

4.1: Radiation dose ($CTDI_{vol}$, DLP and E dose) and Image Quality (Noise)

4.1.1: Brain protocol

A total of 128 patients were subjected to the brain protocol. 82 before and 46 after optimization undergo the Brain routine exams. The patients were assigned randomly by the physician in different CT scanner 2-slice, 4-slice, 16-slice and 64-slice data pertaining to demographic; exposure parameters, dose information and image quality parameters were collected. Details of patient demographics were provided in Tables 4.1, 4.2 and 4.3.Rotation time was reduced from 1.5 sec in 2-Slice scanner to 0.5 sec in 64-Slice scanner as shown in table 4.4.Radiation exposure parameters were presented in Table 4.4 for tube voltage (kVp) and tube current time product (mAs), before and after optimization respectively. Patient dose information in terms of $CTDI_{vol}$ and DLP (mGy.cm) before and after optimization was presented in Table 4.7 in that order and the reduction in percentage. In addition, t-test was done to see the significance difference effect between the two groups. CT_{no} and its difference and image noise (SD) and its increase were represented in tables 4.9 and 4.10 in order. It was apparent from table 4.1 that irrespective of the CT scanner modality, no significant differences ($p>0.05$) were found between gender of patients exposed to radiation before optimization and those patients exposed to radiation after optimization. In Table 4.4, 2-slice machine has a high kV 130 and slice thickness 8mm compare with the other CT scanners.

The 4-CT machine operated with mAs values (120-360mAs). The induced reduction in rotation time led to substantial reduction imaging from 28.57% to 50%. Table 4.7 showed the reduction in $CTDI_{vol}$ after optimization was highly significant $p>0.01$ in four machines and ranged from 12.5% in 2-slice to 33.6% in 64 –slice CT scanners. Total DLP was reduced from 14.3% to 59.7%.

4.1.2: Chest protocol

With respect to chest protocol a total of 148 patients undergo routine Chest exams. Rotation time was changed from 1.5 sec in 2-slice scanner to 0.5sec in the four different types of scanners in order .Radiation exposure parameters were presented in Table 4.4 for tube voltage (kVp) and tube current time product (mAs), before and after optimization respectively. Patient dose in terms of DLP (mGy.cm) and $CTDI_{vol}$ were presented in Table 4.7 before and after optimization in that order and the reduction in percentage. In addition to CT_{no} and its difference and image noise (SD) and its increase were represented in tables4.9 and 4.10 in order. Test of significance using *t*-test.

4.1.3: Abdomen protocol

A total of 128 patients undergo the Abdomen routine exams, 80 and 48 before and after optimization. Rotation time was changed from 1.5 to 0.5 Sec in the four scanners mentioned before .Radiation exposure parameters were presented in Table 4.2 for tube voltage (kVp) and tube current time product (mAs), before and after optimization respectively. Patient dose in terms of DLP (mGy.cm) and $CTDI_{vol}$ were presented in Table 4.7 before and after optimization in that order and the reduction in percentage. In addition to CT_{no} and its difference and image noise (SD) and its increase were represented in tables4.9 and 4.10 in order.

Table 4.1: Patients demographics information for the Brain's protocol. For the four scanners before and after optimization.

	Before optimization	After optimization	t-test	P-value
2-Slice scanner				
N	22	12		
Age(mean)	46.9±21.1 (18-85)	42.7±18.2 (20-70)	0.7	p>0.05 p(0.50)
Gender(number)	9(F),11(M)	5(F),7(M)		
BMI(mean)	(18-25)	(18-25)		
4-Slice scanner				
N	20	12		
Mean Age	53.9±20.9 (19-90)	55.2±19.8 (25-80)	0.2	p>0.05 p(0.8)
Gender(number)	7(F),13(M)	6(F),6(M)		
BMI(mean)	(18-25)	(18-25)		
16-Slice scanner				
N	20	11		
Age(mean)	48.8±18.3 (20-75)	39.7±18.7 (19-74)	1.2	p>0.05 (0.2)
Gender(number)	6(F) ,14(M)	2(F) ,9(M)		
BMI(mean)	(18-25)	(18-25)		
64-Slice scanner				
N	20	11		
Age(mean)	55.2±20.9 (23-85)	44.8±19.5 (20-75)	0.3	p>0.05(0.9)
Gender(number)	6(F) ,14(M)	4(F) ,7(M)		
BMI(mean)	(18-25)	(18-25)		

Table 4.2: Patients demographics information for the Chest's protocol. For the four scanners before and after optimization.

	Before optimization	After optimization	t-test	P-value
2-Slice scanner				
N	20	12		
Age(mean)	48.5±13.9 (29-78)	48.4±15.6 (23-70)	0.4	p>0.05(0.90)
Gender(number)	9(F),11(M)	6(F),8(M)		
BMI(mean)	(18-25)	(18-25)		
4-Slice scanner				
N	20	12		
Age(mean)	51.6±15.0 (20-75)	48±13.4 (27-67)	0.5	p>0.05(0.6)
Gender(number)	13(F),7(M)	5(F),7(M)		
BMI(mean)	(18-25)	(18-25)		
16-Slice scanner				
N	20	12		
Age(mean)	55.4±17.4 (26-80)	55.5±16.3 (28-81)	0.9	p>0.05(0.3)
Gender(number)	8(F),12(M)	5(F),7(M)		
BMI(mean)	(18-25)	(18-25)		
64-Slice scanner				
N	20	12		
Age(mean)	53.1±18.4 (26-80)	62.0±13.6 (34-85)	1.5	p>0.05(0.1)
Gender(number)	4(F),16(M)	5(F), 7(M)		
BMI(mean)	(18-25)	(18-25)		

4.3: Patients demographics information for the Abdomen's protocol. For the four scanners before and after optimization.

	Before optimization	After optimization	t-test	P-value
2-Slice scanner				
N	20	14		
Age(mean)	46.5±11.7 (35-80)	39.6±16.9 (20-74)	1.6	p>0.05(0.1)
Gender(number)	7(F),13(M)	4(F),9(M)		
BMI(mean)	(18-25)	(18-25)		
4-Slice scanner				
N	20	12		
Age(mean)	56.2±13.7 (14-70)	56.5±13.3 (30-70)	0.4	p>0.05(0.6)
Gender(number)	15(F),(M)5	6(F),(M)6		
BMI(mean)	(18-25)	(18-25)		
16-Slice scanner				
N	20	12		
Age(mean)	49.9±16.1 (20-80)	43.4±8.3 (35-62)	1.7	p>0.05(0.1)
Gender(number)	7(F),13(M)	3(F),9(M)		
BMI(mean)	(18-25)	(18-25)		
64-Slice scanner				
N	20	10		
Age(mean)	53.0±16.5	48.3±17.7	1.2	p>0.05(0.2)
Gender(number)	7(F),13(M)	4(F),6(M)		
BMI(mean)	(18-25)	(18-25)		

Table 4.4: Dose acquisition parameters for Brain Protocol from the four types of scanners

CT scanner type		Kv (volt)	mAs (amp)	Slice thickness (mm)	Pitch	Rotation time (sec)	Rotation % reduction
2-slice	Before	130	120	8	1.0	1.5	33.33
	After	130	120	8	1.0	1.0	
4-slice	Before	120	200	5	3.0	0.7	28.57
	After	120	200	5	3.0	0.5	
16-slice	Before	120	360	5	0.84	1.0	50.00
	After	120	360	5	0.84	0.5	
64-slice	Before	120	250	5	0.656	0.7	28.57
	After	120	250	5	0.656	0.5	

Table 4.5: Dose acquisition parameters for Chest Protocol from the four types of scanners:

CT scanner type		Kv volt	mAs amp	Slice thickness mm	Pitch	Rotation time (sec)	Rotation % reduction
2-slice	Before	120	120	5	1.0	1.0	30.00
	After	120	120	5	1.0	0.7	
4-slice	Before	120	200	5	4.5	1.0	30.00
	After	120	200	5	4.5	0.7	
16-slice	Before	120	100	5	0.84	0.7	28.57
	After	120	100	5	0.84	0.5	
64-slice	Before	120	100	5	0.83	0.6	16.67
	After	120	100	5	0.83	0.5	

Table 4.6: Dose acquisition parameters for Abdomen Protocol from the four types of scanners

CT scanner type		Kv volt	mAs amp	Slice thickness mm	Pitch	Rotation time (sec)	Rotation % reduction
2-slice	Before	110	120	5	2.0	1.0	30.00
	After	110	120	5	2.0	0.7	
4-slice	Before	120	200	5	4.5	0.7	28.57
	After	120	200	5	4.5	0.5	
16-slice	Before	120	160	5	0.84	0.7	28.57
	After	120	160	5	0.84	0.5	
64-slice	Before	120	150	5	1.48	0.6	16.67
	After	120	150	5	1.48	0.5	

Table 4.7: CT dose information for Brain Protocol in four CT scanners

	Before optimization mean (range)	After optimization mean (range)	Mean dose reduction (%)	p-value
2-slice CT				
$CTDI_{vol}$(mGy)	27.36±0.0 (27.4–27.4)	24.13±0.0 (24.1–24.1)	12.5%	p<0.01
Total DLP (mGycm)	167.55±56.0 (134.0–257.0)	130.0±0.0 (130.0–130.0)	25.2%	p<0.01
Effective dose (mSv)	3.5	2.7	22.9%	p<0.01
4-slice CT				
$CTDI_{vol}$(mGy)	52.61±3.3 (40.9-54.5)	45.4±6.6 (40.5-54.0)	14.7%	p<0.01
Total DLP (mGycm)	651.8±133.2 (463.7–967.3)	490.6±83.4 (358.1-673.1)	14.3%	p<0.01
Effective dose (mSv)	1.61	1.40	13.0%	p<0.01
16-slice CT				
$CTDI_{vol}$(mGy)	68.1± 0.0 (68.1–68.1)	59.6±3.6 (54.9–69.1)	13.3%	p<0.01
Total DLP (mGycm)	1195.2±328.2 (913.0-1559.0)	876.2±65.9 (746.0-968.0)	30.8%	p<0.01
Effective dose (mSv)	2.5	1.8	27.9%	p<0.01
64-slice CT				
$CTDI_{vol}$(mGy)	75.7±12.8 (21.9–80.8)	53.9±9.1 (48.0–80.8)	33.6%	p<0.01
Total DLP (mGycm)	1622.0±234.0 (1360-2069.4)	875.9±312.4 (113.0-1463.0)	59.7%	p<0.01
Effective dose (mSv)	3.4	1.8	47.0%	p<0.01

*$CTDI_{vol}$, volume CT dose index; DLP, dose length product.

Table 4.8: CT dose information for Chest protocol in four CT scanners.

	Before optimization mean (range)	After optimization mean (range)	Mean dose reduction (%)	*P* value
2-slice CT				
$CTDI_{vol}$ (mGy)	7.13±0.9 (6.4-8.8)	4.62±.01 (3.1–6.5)	21.3%	<0.01
Total DLP (mGycm)	173.9±69.6 (112.0–268.0)	138.50±20.0 (111.0–162.0)	22.7%	<0.01
Effective dose (mSv)	2.4	1.9	20.8%	<0.01
4-slice CT				
$CTDI_{vol}$ (mGy)	20.6±6.3 (11.7–15.6)	14.60±1.7 (11.0–15.0)	34.1%	<0.01
Total DLP (mGycm)	651.8±133.2 (463.1–1063.1)	490.6±83.4 (358.1-673.1)	28.2%	<0.01
Effective dose (mSv)	9.1	6.8	25.3%	<0.01
16-slice CT				
$CTDI_{vol}$ (mGy)	4.67± 1.3 (2.6–7.3)	4.40±1.4 (1.9–6.8)	6.3%	<0.01
Total DLP (mGycm)	180.6±57.7 (74.0- 337.0)	152.6±90.4 (49.0–335.0)	16.8%	<0.01
Effective dose (mSv)	2.5	2.1	15.9%	<0.01
64-slice CT				
$CTDI_{vol}$ (mGy)	14.60±0.5 (12.70–15.2)	16.70±1.8 (10.6–15.2)	6.4%	<0.01
Total DLP (mGycm)	549.1± 135.7 (1360.0–1663.2)	546.6±66.6 (111.3-1463.2)	1.1%	<0.01
Effective dose (mSv)	7.8	7.6	2.6%	<0.01

*$CTDI_{vol}$, volume CT dose index; DLP, dose length product

Table 4.9: CT dose information for Abdomen protocol in four CT scanners.

	Before optimization mean (range)	After optimization mean (range)	Mean dose reduction (%)	*P* value
2-slice CT				
$CTDI_{vol}$ (mGy)	5.54±1.1 (2.8–6.2)	4.25±1.1 (2.2–6.2)	26.4%	<0.01
Total DLP (mGy●cm)	222.20±57.5 (105.0–312.0)	173.92±54.3 (82.0–276.0)	24.4%	<0.01
Effective dose (mSv)	3.3	2.6	21.2%	<0.01
4-slice CT				
$CTDI_{vol}$ (mGy)	59.1±15.3 (41.2-82.8)	45.3±4.2 (41.2-56.7)	26.4%	<0.01
Total DLP (mGy●cm)	2906.6±674.8 (1500.2-3709.1)	1475.7±378.8 (903.1-2119.1)	55.4%	<0.01
Effective dose (mSv)	43.5	22.1	49.2%	<0.01
16-slice CT				
$CTDI_{vol}$ (mGy)	7.10± 1.8 (4.4-11.3)	5.20±0.8 (4.2–6.8)	30.1%	<0.01
Total DLP (mGy●cm)	339.9±100.7 (146.0-534.0)	241.0±46.2 (184.0-316.0)	34.1%	<0.01
Effective dose (mSv)	5.1	3.6	29.4%	<0.01
64-slice CT				
$CTDI_{vol}$ (mGy)	21.9±0.0 (21.9–21.9)	19.70±1.8 (18.30-21.90)	10.6%	<0.01
Total DLP (mGy●cm)	938.5±200.0 (629-1297.0)	797.8±135.4 (623.7-973.50)	16.2%	<0.01
Effective dose (mSv)	14.1	11.9	15.6%	<0.01

*$CTDI_{vol}$, volume CT dose index; DLP, dose length product

Table 4.10: Represents the CT_{no} (White matter) before and after Optimization for Brain protocol

CT scanner type	Before	After	Difference
2-slice scanner			
CT_{no}. (HU)	28.2±2.8 (21.7-33.4)	27.8±2.3 (23.5-31.8)	0.4
4-slice scanner			
CT_{no}. (HU)	28.4±3.0 (23.8-33.8)	30.3±2.2 (26.6-33.4)	1.9
CT_{no}. (HU)	30.3±3.0 (24.3-36.3)	28.5±4.0 (19.1-34.0)	1.8
64-slice scanner			
CT_{no}. (HU)	27.7±3.2 (21.6-32.0)	28.9±2.7 (24.2-31.2)	1.2

Table 4.11: Represent the CT_{no} (Lung) before and after Optimization for Chest protocol

CT scanner type	Before	After	Difference
2- slice scanner			
CT_{no}. (HU)	-792.9±49.0 -(870.5-701.1)	-785.6±27.4 -(817.3-750.7)	7.3
4-slice scanner			
CT_{no}. (HU)	-853.1±53.3 -(947.4-742.9)	-851.8±61.4 -(916.2-705.6)	1.3
16-slice scanner			
CT_{no}. (HU)	774.2±102.0 -(988.3-541.1)	-770.4±67.8 -(882.0-650.7)	3.4
64-slice scanner			
CTno. (HU)	-837.3±58.0 -(946.0-702.5)	-835.6±35.6 -(907.0-782.7)	1.7

Table 4.12: Represent the CT_{no} (liver) before and after Optimization for Abdomen protocol.

CT scanner type	Before	After	Difference
2-Slice CT scanner			
CT_{no}. (HU)	51.3±5.7 (40.6-61.0)	47.2±5.1 (40.0-57.0)	4.1
4-Slice CT scanner			
CT_{no}. (HU)	64.2±7.6 (51.9-77.1)	61.3±8.3 (42.1-77.3)	2.9
16-Slice CT scanner			
CT_{no}. (HU)	52.5±7.3 (42.0-70.6)	55.93±8.4 (42.6-69.2)	4.4
64-Slice scanner			
CT_{no}. (HU)	52.8±8.0 (40.8-64.2)	50.3±6.9 (41.3-61.1)	2.4

Table 4.13: Image Quality (Noise) in four scanners type before and after optimization for Brain Protocol

CT scanner type	Before	After	Increase %	*P* value	interpretation
2-slice scanner					
Image noise (HU) (SD)	3.0±0.5 (2.1-4.1)	3.8±1.4 (1.5-6.1)	26.67	p<0.5 (0.3)	The difference is not statistically significant
4-slice scanner					
Image noise (HU) (SD)	2.5±0.9 (2.1-5.6)	2.9±0.8 (1.6-4.5)	15.9	P<0.5 (0.1)	The difference is not statistically significant
16-slice scanner					
Image noise (HU) (SD)	2.67±1.4 (0.8-5.9)	2.70±0.5 (2.1-3.4)	1.1	P<0.5 (0.01)	The difference is not statistically significant
64-slice scanner					
Image noise (HU) (SD)	3.1±1.0 (1.9-6.9)	3.9±1.5 (1.5-6.3)	25.8	P<0.5 (0.04)	The difference is not statistically significant

HU, Hounsfield unit; SD , standard deviation.

Table 4.14: Image Quality (Noise) in four scanners type before and after optimization for Chest Protocol

CT scanner type	Before	After	increase	*P* value	interpretation
2-slice scanner					
Image noise (HU) (SD)	26.9±9.6 (12.8-50.3)	29.7±13.5 (14.3-52.7)	10.4	P<0.5 (0.6)	The difference is not statistically significant
4-slice scanner					
Image noise (HU) (SD)	18.3±4.5 (8.6-28.5)	18.5±2.5 13.8-22.3)	1.1	P<0.5 (0.7)	The difference is not statistically significant
16-slice scanner					
Image noise (HU) (SD)	39.0±23.6 (17.9-87.3)	43.9±22.8 (8.8-75.8)	12.6	P<0.5 (0.9)	The difference is not statistically significant
64-slice scanner					
Image noise (HU) (SD)	24.5±12.1 (11.3-53.6)	24.9±10.9 (10.1-53.0)	1.6	P<0.5 (0.2)	The difference is not statistically significant

Table 4.15: Image Quality (Noise) in four scanners type before and after optimization for Abdomen Protocol

CT scanner type	Before	After	increase	*P* value	interpretation
2-slice scanner					
Image noise (HU) (SD)	13.4±3.3 (80-.0-23.0)	14.8±3.4 (9.6-21.1)	10.4	P<0.5 (0.4)	The difference is not statistically significant
4-slice scanner					
Image noise (HU) (SD)	7.8±1.7 (4.4-9.7)	8.2±2.6 (3.9-13.0)	5.13	P<0.5 (0.7)	The difference is not statistically significant
16-slice scanner					
Image noise (HU) (SD)	11.49±3.1 (5.3-16.3)	11.64±4.0 (5.4-18.6)	1.3	P<0.5 (0.5)	The difference is not statistically significant
64-slice scanner					
Image noise (HU) (SD)	7.16±1.3 (4.4-9.6)	7.53±1.2 (6.0-10.3)	5.17	P<0.5 (0.1)	The difference is not statistically significant

Chapter Five

Discussion and Conclusion

5.1: Discussion

Recent advances in MDCT technology increased the number of CT examinations by approximately 100% during the last decade of the 20th century. (Mettler et al, 1993), despite a significant reduction of CT doses in recent years, mainly due to improved technology, CT is still a predominant source of medical radiation absorbed dose to the general population. (Söderberg et al, 2012), essential to the appropriate use of CT is the appropriate selection of the imaging technique, based on the patient's age, anatomy (e.g., size), and the imaging task (e.g., tracking an interventional instrument, surveillance imaging of lung nodules, or diagnosing a suspicious soft tissue lesion in the abdomen) optimization.

Although all the major CT manufacturers offer significant tools to reduce radiation dose, many centers do not take advantage of the dose reduction capabilities of their scanners because of a lack of familiarity and understanding as to how these tools work. Today, several approaches are used to minimize radiation absorbed dose and improve image quality in medical X-ray. One way to succeed is speed up the gantry rotation time. So when performing CT, two important criteria need to be considered: radiation dose needs to be minimized while maintaining adequate image quality. Unfortunately, these are competing forces and as radiation dose decreases, image quality suffers. There is still considerable room for optimization and continuous developments of new technologies aim to optimize image quality and radiation absorbed dose to the patient. These technologies will continue to require close collaboration between medical physicists, manufacturers, radiologists, nuclear medicine physicians, technologists, consequently, although there are several studies in literature about dose reduction. Recently, it has been demonstrated that low and medium dose reduction is possible by CT protocols and developed softwares, and the recent expectation is to provide advanced dose reduction. It is important to determine the necessary parameters and to know its contribution to dose reduction in order to be able to develop techniques and software programs for CT dose reduction. We suggest that it is possible to provide an advanced level of radiation dose reduction by optimizing CT protocols.

Determining the contributing parameters that affect dose will allow the development of new protocols and reveal the importance of previously unused parameters. This information will enable the development of new software. Thus, it will be possible to reduce the damage to humans by radiation.

It is clear that computed tomography (CT) overwhelmingly benefits patients when used for appropriate settings. Concerns have been raised regarding the potential risk of cancer induction from CT due to the exponentially increased use of CT in medicine. Keeping radiation dose as low as reasonably achievable, consistent with the diagnostic task, remains the most important strategy for decreasing this potential risk (Yu, et al 2009). Multiple strategies in CT dose reduction have been developed and well described by many authors. Managing CT protocols and scan parameter adjustment are simple methods.

Therefore, reduction of dose remains to be a challenge in CT scans. Nevertheless it is associated with increase in noise. Image Noise is one of the primary factors in CT Image Quality Noise (specifically, quantum noise) is generally characterized by graininess, or a salt and pepper pattern on the image.

In this study, the scan exposure parameters for 4-Slice scanner were 120kVp and 200mAs and standard pitch of 1.0 and rotation time 1.0Sec, keeping these parameters constant only changing in the time per rotation to 0.7Sec the reduction in dose was evaluated (Table 4.1) . For 16-Slice scanner the exposure parameters were 120kVp and 100mAs and a pitch of 0.87and rotation time 0.7Sec. After changing the rotation time to 0.5Sec with the same scan exposure parameters reduction to dose was also examined. The increase in noise determined by increase in Hounsfield units was tested (Tables 4.1- 4.6).

The CTDI, which is defined to represent an approximation to the average absorbed dose to a particular location in a standard acrylic phantom from multiple CT slices. $CTDI_w$ is a weighted average of the CTDI's at the center and periphery of the phantom. $CTDI_{vol}$ is similar to $CTDI_w$ but also includes the effect of pitch on the radiation dose. It is the $CTDI_{vol}$ that is displayed on the CT console and dose report. $CTDI_{vol}$ is a useful indicator of the radiation output for a specific exam protocol, because it takes into account protocol-specific information such as pitch.

In this study the mean $CTDI_{vol}$ for the 4-Slice was 20.60mGy before rotation time reduction and 14.6mGy afterwards. And the reduction obtained was 34.1%. For the 16 –Scanner the mean $CTDI_{vol}$ before and after rotation time reduction were 4.67mGy and 4.40mGy respectively (Table 4.5).

The dose length product (DLP) is an indicator of the integrated radiation dose of an entire CT examination, or radiation deposit in patients. The DLP for the 4 –Slice scanner was reduced by 28.2% after rotation time reduction. For the 16-Scanner the reduction was 16.8%. The reduction in dose was slightly less than that obtained by reducing the mAs and pitch (10) but was still significant while reproducing acceptable images.

Noise is one of the primary factors in CT Image Quality Noise (specifically, quantum noise) is generally characterized by graininess, or a salt and pepper pattern on the image. Noise is inversely related to the number of X-rays. In our study there was no significant change in the noise for the chest exam for 4-Slice scanner which increased from 18.2HU to 18.5HU. However there was a significant change in the 16-Slice scanner the mean noise increased from 39.0 to 43.9HU. The image was still within in the acceptance range. The challenge is in finding a balance between dose and noise that allows the images to be of diagnostic quality while utilizing the lowest dose possible.

5.1.1: Brain protocol

The results of our study demonstrate that use of rotation time modulation as a radiation reduction tool for the CT brain examination resulted in significant reduction in radiation dose to adults ($P < .001$). Indeed, for 2-slice scanner the $CTDI_{vol}$ and DLP, respectively, were reduced by 12.5% and 25.2% by reducing rotation time from 1.5Sec to 1.0Sec. In 4-slice scanner a significant reduction ($P < .001$) was 14.7% and 14.3% for $CTDI_{vol}$ and DLP, when rotation time decreased from 0.7Sec to 0.5Sec.

Speeding tube rotation time from 1.0 Sec to 0.5 Sec radiation doses was reduced by 13.3% for $CTDI_{vol}$ and 30.8% for DLP in 16- slice CT scanner, which is also a significant reduction. When we transitioned to a 64- slice CT scanner, a significant reduction of 33.6% and 59.7% for $CTDI_{vol}$ and DLP respectively. The rotation time was changed from 0.7Sec to 0.5 Sec. And these was agree with one of the study that concerned with dose reduction techniques in Brain; In 2008 (Alice B. Smith, et al) For unenhanced CT of adult brains, the $CTDI_{vol}$ and DLP, respectively, were reduced by 60.9% and 60.3%. Significant dose reductions ($P < .001$) were also observed for adult cervical and intracranial CT angiography performed. Image quality and noise were unaffected by the use of either dose modulation technique ($P > .05$).

5.1.2: Chest Protocol

Reducing rotation time method as a CT radiation dose reduction tool has led to significant dose reduction in the three chosen body regions. Effective doses were reduced by 2.6-25.3% in the four CT scanners compared with the pre-optimized period. Technical parameter changes led to 6.4–34.1% reduction of $CTDI_{vol}$ per acquisition. The reduction is largest in 4-slice CT and smallest in 16-slice and 64-slice CT, which could be explained by a larger reduction in rotation time, when compared with other modalities. DLP was reduced by 22.7%, 28.2%, 16.8% and 1.1% in 2-slice, 4-slice, 16-slice and 64-slice CT scanners in order. This small reduction in chest protocol regarding 64-slice CT, probably from the appropriate scan length performed. CT image quality (noise) was compared for the two groups. However, accelerating rotation time increased image noise. There was no significant difference in noise ($p > 0.05$). An excellent review of dose reduction was given by Kubo *et al*., 2008. They summarized the available data on reducing radiation dose exposure in routine chest protocol.

Kubo et al (2008) compared and evaluated the image quality of CT examinations with reduced tube current time product. The image quality criteria was identified of structures based on (Naidich et al.1990), on noise and uniformity (Mayo et al 1995and Prasad et al.2002); their work derived the smallest acceptance of mAs for each organ. Their approach was an acceptance alternative approach in image quality which is core of current study based on dose reduction on rotation time. According to (Beeres et al 2014) investigated the influence of faster gantry rotation time on image quality in Chest CT .They found that faster CT gantry rotation times (0.28s/rot and 0.33s/rot) , reduced scan time and motion artifacts, but at the same time resulted in image noise which decreased image quality.

5.1.3: Abdomen Protocol

Choosing minimizing rotation time method for radiation dose reduction these methods have led to significant dose reduction in abdomen region in all four scanners. Effective doses were reduced by % compared with the pre-implementation period. Technical parameter changes led to % reduction of $CTDI_{vol}$ per acquisition. The reduction is largest in CT and smallest in CT, the difference in CT doses between the two groups was mainly due to the very high nature of the initial scan technique.

The post-implementation doses were under diagnostic reference levels as recommended by the European Commission, American College of Radiology, Australian Radiation Protection and Nuclear Safety Agency, and United Kingdom Department of Health (Table 5.1).

Table 5.1: Comparison with adult CT diagnostic reference levels of Australia, the United Kingdom, Europe and United States

Examination	Median values		Australia (2014)	United Kingdom (2005)†	Europe (2014)‡	United States (2014)§
	Before 2,4,16,64	**After 2,4,16,64**				
Head						
$CTDI_{vol}$	27.3,52.6,68.1,75.7	24.13,45.4,59.6,53.6	60	65–100	60	75
DLP	167.5,651.8,1195.2,1622.0	130.0,490.6,876.2,875.9	1,000	760–930	1,000	–
Chest						
$CTDI_{vol}$	**7.1,2.6,4.6,14.6**	4.6,14.6,4.4,14.7	15	13	10	21
DLP	173.9,651.8,180.6,549.1	138.5,490.6,152.6,546.6	450	580	400	–
Abdomen						
$CTDI_{vol}$	5.54,59.1,7.1,21.9	4.25,45.3,5.20,19.7	–	14	–	–
DLP	222.22906.6,339.9,938.5	173.9,1475.7,241.0,797.8	-	470	400–740	–
Pelvis						
$CTDI_{vol}$	-	-	–	–	–	–
DLP	-	-	–	–	550	–
Abdominopelvic						
$CTDI_{vol}$	-	-	15	14	25	25
DLP	-	-	700	560	800	

†Specified for multi-detector CT. ‡Most common values for each anatomical region. §Guideline of the American College of Radiology. $CTDI_{vol}$, volume CT dose index; DLP, dose length product.

They found that images with acceptable quality and reliable detection ability could be obtained using smaller doses of radiation, compared to protocols commonly used by operators. Effective doses reduced by without a negative impact on image quality, these substantial dose reductions were remarkable.

5.2: Conclusion

This study showed that optimizing the dose for the four CT scanners is dependent on multiple factors. One of the major factors effecting radiation dose is speed up the tube rotation (minimizing the rotation time). An import consideration in our study is that there is a strong linear relationship of rotation time with radiation dose, when the other scanning parameters, including kilovolt peak, tube current, pitch, and section thickness are kept constant .The reduced rotation time led to a linear decreased in radiation dose. CT radiation dose optimization and reduction is a complex process that seems to stay motionless since years. Several parameters can result in dose reduction. Reducing the rotation time was found to significantly reduce the dose. The rotation time reduction is limited by the amount of increase in noise. A significant reduction in $CTDI_{vol}$ and DLP were observed with the 4-slice scanner, with a considerably low increase in noise. A lower reduction in dose and larger increase in noise was observed with the 16-slice scanner. The results are encouraging for further efforts in reducing the CT dose without compromising the clinical findings with an aim to fulfill the optimization requirements. A significant dose reduction of CT performed in emergency patients was achievable via strategies that target simple adjustments (e.g., reduction of scan coverage, number of acquisition), technical optimization (e.g., mAs and kVp reduction, ATCM) and use of indication-specific protocoling while maintaining an acceptable level of diagnostic image quality. Reduction of tube rotation time reduced the patient radiation doses up to 30% from its original value without compromising the diagnostic findings.

5.3: Recommendations and Future Studies

- Training and continuous education should be provided to concern people.
- Master the use of every dose reduction tool available (Think optimal - do the optimal study with the optimal scanning technique with the optimal protection in place for your patients.
- A comprehensive low-dose program should be established in order to achieve measurable dose reduction goals.
- Make use of information resources Consult: user manuals ,manufacturers, websites, scientific literature and colleagues

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Appendix A

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DOSE REDUCTION IN CHEST CT EXAMINATION

A. Sulieman[1,2,*], N. Tammam[2], K. Alzimami[3], A. M. Elnour[2], E. Babikir[3] and A. Alfuraih[1]
[1]Radiology and Medical Imaging Department, College of Applied Medical Sciences, Prince Sattam bin Abdulaziz University, Alkharj, Kingdom of Saudi Arabia
[2]College of Medical Radiologic Science, Sudan University of Science and Technology, Khartoum, Sudan
[3]Radiological Sciences Department, College of Applied Sciences, King Saud University, Riyadh, Saudi Arabia

*Corresponding author: abdelmoneim_a@yahoo.com

Computed tomography (CT) examinations involve relatively high doses to patients. The objectives of this study were to optimise the radiation dose for patient during CT chest scan and to estimate the lifetime cancer risk. A total of 50 patients were studied: control group (A) (38 patients) and optimisation group (B) (12 patients). The optimisation protocol was based on CT pitch increment and lowering tube current. The mean volume CT dose index ($CTDI_{vol}$) was 21.17 mGy and dose length product (DLP) was 839.0 mGy cm for Group A, and $CTDI_{vol}$ was 8.3 mGy and DLP was 339.7 for Group B. The overall cancer risk was estimated to be 8.0 and 3.0 cancer incidence per million for Groups A and B, respectively. The patient dose optimisation during CT chest was investigated. Lowering tube current and pitch increment achieved a radiation dose reduction of up to 60 % without compromising the diagnostic findings.

INTRODUCTION

Man-made sources of radiation account for ~14 % of the annual radiation dose from all sources of radiation[1, 2]. The average level of radiation exposure due to the medical applications in developed countries is equivalent to 50 % of the global average level of natural exposure, although obviously there will be marked variations in the doses received by individuals worldwide depending on health-care level[1]. Medical exposure is the largest source of man-made exposure to ionising radiation that accounts for nearly 96 % of all man-made radiation exposure to human and continues to grow substantially[1, 2]. CT scanning is recognised as a high radiation dose modality and estimated to be 17 % of the radiological procedure and responsible for 70 % medical radiation exposure[3, 4]. Advances in CT technology have made possible new CT applications and have expanded the role of CT into new types of clinical diagnoses[4, 5]. The doses can often approach or exceed levels known with certainty to increase the probability of cancer, and some deterministic effects were reported in some angiography/perfusion brain studies[6, 7]. It has been estimated that 1 individual in 1000 develops cancer from exposure to a radiation dose of 10 mSv[8], and 2 % of current cancers in the United States are due to CTs performed in the past[9]. CT scan of the chest is widely used to evaluate different clinical conditions. The effective dose in chest CT is in the order of 8 mSv (around 400 times more than chest radiograph dose), and in some CT examinations like that of pelvic region, it may be around 20 mSv[6]. During CT chest procedure, breast dose in female patients may be as much as 30–50 mGy, even though breasts are not the target of imaging procedure[6]. In previous literature, CT dose reduction achieved by tube current modulation has been reported to be up to 26–50 %[10], and a dose reduction of up to 40–50 % could be achieved by means of iterative reconstruction algorithms without degrading image quality and reconstruction speed. In addition to that radiosensitive organs shielding reduced the radiation exposure to radiation-sensitive organs, such as the breast, thyroid and eye lens by 20–50 %[11]. Shields are, however, associated with greater image noise and streak artefacts[11]. Furthermore, a reduction of patient dose by 10–50 % was documented when automatic exposure control (AEC) is used without loss of image quality[6].

In recent years, concerns have been raised about the radiation exposures to patients during CT procedures and some studies have been published in patient radiation protection[12–17]; yet, still few studies have been performed in dose optimisation during CT chest procedures[2, 13–17]. These studies have shown that there is a wide range of dose values and acquisition protocols. In addition to that the data available on patient doses in CT procedures are generally outdated because of the continuous development of CT X-ray generators and technologic innovation that have taken place over the past decade from single-slice CT in 1998[18] to 320-slice CT in 2009 and 640 slices in 2013. The objective of this study was to evaluate and optimise the radiation dose to patients undergoing CT chest exam with 64-slice CT scanner.

MATERIALS AND METHODS

CT machine

The study was performed with 64-slice CT scanner Toshiba Aquilion (Toshiba Medical Systems, Otawara,

Appendix A

Japan). It consists of 64 × 0.5 mm detector rows and a maximum gantry rotation speed of 0.4 s. The system has quantum de-noising software resulting in 15 % less image noise than that produced by the 16-slice system. The CT machine was manufactured in 2008 and installed in 2011. All quality control tests were performed to the machine prior to the data collection. These tests were carried out by experts from the Sudan Atomic Energy Commission (SAEC). All the parameters were within the acceptable range.

Patient data

A total of 50 patients were divided into two groups: the first group (A) as control group (38 patients), and the other as optimisation group (B) (12 patients). Procedures in Group A were performed with the department's local protocol. Ethics and research committee approved the study, and informed consent was obtained from all patients prior to the procedure. All patients suffered from chest problems that required referring them to the CT department. Data were collected to study the effects of patient-related parameters [age, sex, weight, height and body mass index (BMI)] and diagnostic purpose of examination on radiation dose. The exposure-related parameters were taken into consideration: gantry tilt, potential in kilovoltage (kVp), tube current (mA), rotation time, slice thickness, number of slices, and start and end points of scans, but special consideration was paid to the effect of pitch table increment on patient dose. The collection of patient exposure parameters was done using survey forms prepared for collection of patient exposure-related parameters.

Organ dose calculation

Organ doses were estimated using normalised CT dose index (CTDI) values published by the ImPACT group[19]. For the sake of simplicity, the $CTDI_{100,air}$ will henceforth be abbreviated as $CTDI_{air}$. In this study, volume CTDI ($CTDI_{vol}$, mGy) and dose length product (DLP, mGy cm) were indicated by the scanner software, and by using these parameters and applying conversion factors for chest, effective dose (mSv) was calculated. The organ dose conversion factor f (organ, z) was obtained from the National Radiological Protection Board (NRPB) datasets (NRPB-SR250) based on the Monte Carlo simulations[18].

Estimation of effective dose

Patient doses were determined by using the $CTDI_{vol}$ expressed in mGy and the DLP in mGy cm as provided on the scanner console. The CTDOSE software supplied by the ImPACT group (ImPACT CT Patient Dosimetry Calculator, version 0.99×; ImPACT, London, UK)[19] was used, and typical scanning parameters such as kVp, mA, exposure time, pitch, slice thickness, gender, and start and end positions of each scan were used as input data to the CTDOSE spreadsheet in organ dose estimations[19].

Table 1. Image acquisition parameters for both groups during CT chest procedures.

Parameter	Control	Optimisation
kVp	130 (120–140)	130 (120–140)
mAs	175 (100–250)	132 (100–164)
Detector configuration	0.5 × 64	0.5 × 64
Rotation time	0.5	0.5
Noise[a]	28.4	48.2
SD index	8.5	19.2
Slice thickness, mm	5.0	5.0
Pitch		
PF	0.84	1.48
HP	0.53	0.95
SDOFOV, L	400.0	400.0
Reconstruction mode	Helical 3D	Helical 3D

[a]Before processing.

CT dose optimisation strategies steps

CT dose optimisation was performed for patients during CT chest. Routine image acquisition was performed using AEC settings. Using AEC dose can increase or decrease depending on reference image quality setting at the time of installation. Toshiba Aquilion 64 slice has the capability to automatically alter the tube current (mA) on the basis of each individual patient's size and shape. However, in this study, the dose reduction strategy was based on reduction in tube current (mA) and increase in pitch while maintaining diagnostic image quality based on patient characteristics. Image acquisition for both groups is illustrated in Table 1. Three consultant radiologists evaluated all the medical images.

Cancer risk estimation

The risk (RT) of developing cancer in a particular organ (T) following CT chest after irradiation was estimated by multiplying the mean organ equivalent (H_T) dose with the risk coefficients (f_T) obtained from the ICRP publication[20]. The overall lifetime mortality risk (R) per procedure resulting from cancer probability was determined by multiplying the effective dose (E) by the risk factor (f). The risk of genetic effects in future generations was obtained by multiplying the mean dose to the ovaries by the risk factor[20].

RESULTS

Patient demographic data and scan parameters are presented in Table 2. Patient demographic data were

Appendix A

Table 2. Demographic data of patient and scan parameters for both groups: mean and the range in the parenthesis.

Patient group	N	Age (y)	Weight (kg)	BMI (kg m^{-2})
A	38	50.21 (15–77)	71.6 (40.0–84.0)	26.2 (19–32.1)
B	12	54.42 (29–75)	72.33 (65.0–80.0)	25.8 (22.8–28.3)

Table 3. Dose parameters.

Patients	DLP (mGy cm)	CTDI$_{vol}$ (mGy)	Effective dose (mSv)
Control group	832.7 (209–1860)	21.2 (8.20–120.0)	14.2 (3.6–31.6)
Optimised group	339.6 (209–374)	8.3 (8.6–8.2)	5.8 (3.5–6.3)
Reduction, %	52.0	60.8	59. 2

Table 4. Organ equivalent dose (mSv) and risk estimation.

Organ	Patient group	Organ equivalent dose (mSv)	Risk factor × Sv^{-1} × 10^{-4}	Cancer probability 10^{-6}
Breast	A	13.4	116	155.4
	B	5.2		60.3
Thyroid	A	1.6	20	3.2
	B	4.1		8.2
Uterus	A	0.05	6.3	0.03
	B	0.11		0.07

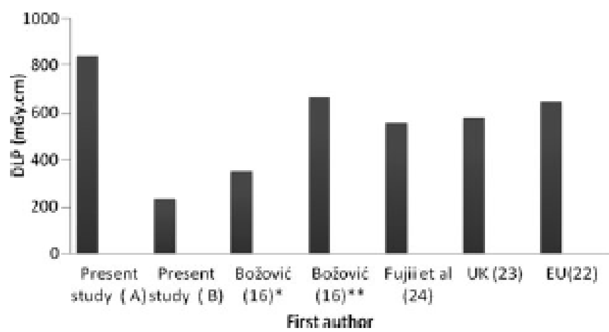

Figure 1. Comparison of DLP (mGy cm) for CT chest procedures with previous studies. *,**Different CT modality.

comparable. Although many patients were elderly, 55 % of them were below 40 years old. The tube voltage was constant for both groups, while the mAs for control group is higher by 25 % compared with optimised group (Table 1). Table 3 presents the patient's dose values in terms of CTDI$_{vol}$, DLP and effective dose. Dose reduction of 60 % was achieved by using optimisation technique. Table 4 shows the effective dose values used to estimate the cancer risks associated with the organ dose to adjacent organs. It is also reveals that the probability of radiation-induced cancer for different organs was in a magnitude of 10^{-6}. The breast has the highest dose due to its position inside the radiation field (Table 4).

DISCUSSION

The radiation dose depends on patients' parameters (weight) and scan parameters. No significant difference was noticed in terms of weight, height and BMI between the two patient groups. Hence, the comparison between the two groups will be more reliable. According to the result in Table 2, the mean CTDI$_{vol}$ was 21.2 mGy and DLP was 839.7 mGy cm for Group A and CTDI$_{vol}$ was 8.3 mGy and DLP was 239.6 mGy cm for Group B. The main reason for higher doses is different pitch value in this study. Other factors such as insufficient education of operators and practitioners in the newly emerging technology and patient-related factors were also reported in literature[2]. A reduction of radiation dose of up to 60 % of the total scan dose and effective dose was achieved (Table 2). All CT images were acceptable and easy to diagnose. After optimisation, the effective dose was 5.7 mSv per procedure showing a reduction of 59 %, while there was increase in noise, but within the acceptable range. Image quality was judged subjectively by three consultant radiologist. Average scan lengths, calculated dividing DLP and CTDI, were 39.2 cm and 40.1 for Groups A and B, respectively. Although scan length depends on patient height, the results indicate that the scan range is not optimally determined compared with a study published by Bozovic *et al.*[17]. The mean DLP per CT chest procedure for Group A was higher than previously reported studies[21–24], while DLP after optimisation was lower than those values (Figure 1).

Appendix A

CT chest involves direct irradiation of the breast, and the thyroid and uterus lie adjacent to the field (Table 3), which necessitates estimating the organ dose received by scattered radiation. Breast has the highest organ dose with the highest cancer probability compared with thyroid and uterus. Therefore, CT procedure of chest in young girls and young females needs to be carefully justified in view of high breast dose and probability of cancer incidence.

The overall cancer risk was estimated to be 8.0 and 3.0 cancer incidence per million for Groups A and B, respectively. The cancer risk of developing cancer following a CT scan is significantly reduced by radiation dose optimisation. Consequently, referring doctors must justify the decision to perform each CT scan weighing the undoubted benefits of CT scans against the potential risks. The study protocol of dose reduction that allows patient dose reduction without the loss of diagnostic accuracy was designed for optimisation of image quality to meet clinical requirements. With this protocol, dose reduction of up to 60 % was achieved. The disadvantage of this technique is that it needs a good level of clinical experience.

CONCLUSION

The patient dose optimisation during CT chest was investigated. By lowering tube current and pitch increment, a radiation dose reduction of up to 60 % was achieved without compromising the diagnostic findings. Optimisation requires continuous efforts and close cooperation between radiologists, radiographers and regularity authorities. Optimising protocols must be applied with care to ensure that they are tailored to clinical need and patient size.

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Appendix B

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Original Research Article

Rotation Time and Dose Reduction in Chest CT scans

N. Tammam[1], A. M. Elnour[2], H. Omer[3], A. Sulieman[3]
[1]College of Medical Radiologic Science, Sudan University of Science and Technology, Khartoum, Sudan
[2]Faculty of Radiology and Nuclear Medicine, the National Ribat University, Khartoum, Sudan
[3]College of Medicine, University of Dammam, Dammam, Kingdom of Saudi Arabia
[4]Radiology and Medical Imaging Department, College of Applied Medical Sciences, Prince Sattam bin Abdulaziz University, Alkharj, Kingdom of Saudi Arabia

***Corresponding author**
A. Sulieman
Email: abdelmoneim_a@yahoo.com

Abstract: Computed tomography is associated with exposing patients with high radiation doses. This study was conducted to evaluate the effect of reducing the rotation time to reduce the dose without compromising the image quality. Two types of CT scanners were involved in this study: a 4-slice and a 16-slice scanner. A significant reduction in CTD Ivol and DLP were observed with the 4-slice scanner, with a considerably low increase in noise.
Keywords: Computed tomography, CTD Ivol

INTRODUCTION

Computed Tomography (CT) was introduced into clinical practice in 1972 and revolutionized x-ray imaging by providing high quality images, which reproduced transverse cross sections of the body. The simultaneous introduction in 1998 of computed tomography with multislice acquisition and half-second rotation times allowed major advances in CT imaging.

Multislice CT (MSCT) with sub-second rotation times allows for the scanning of long ranges (advantageous in, for example, peripheral multislice CTA), for shorter scan times (advantageous in, for example, pediatric CT and trauma), and for a reduction in movement artifacts (as, for example, in ECG gated cardiac CT). With the reconstructed thin axial sections provided by MSCT, a near-isotropic 3-dimensional volume with sub-millimeter sized voxel can be constructed, that is well-suited for review on advanced 3D workstations. This is particularly true for 16 (or more) slice scanners [1].

16-slice CT allows applications of three-dimensional (3D) images in clinical fields such as diagnosis, surgical simulation, planning of radiation therapy and monitoring of interventional therapy [2]. Because of its geometry and usage, CT is a unique modality and therefore has its own set of specific parameters for radiation dose [3]. CT-scanners are becoming more and more popular imaging modality amongst medical practitioners as their tools for diagnostic practices. Continuing advances in CT technology coincide with increasing utilization of CT as diagnostic tools. However, CT is associated with relatively high radiation doses; with a corresponding increased risk of carcinogenesis [4]. The high radiation dose from CT procedures has increase the concern regarding the associated radiogenic risk. Unlike conventional radiography, CT exposes patients to higher radiation doses than do conventional diagnostic x-rays. For example, a chest CT scan (8 mSv) typically delivers more than 400 times the radiation dose of a routine chest X rays (0.02 mSv) [5, 6]. It had been estimated that CT radiation doses generate 0.7% of total expected cancer prevalence and 1% of total cancer death [7].

Image acquisition factors affect patient doses include tube voltage, tube current; scan length and imaging technique (helical or sequential). However, the wide variation in patient doses can be minimized if proper exposure factors were selected, and patients will exposed to radiation to justifiable radiation doses consistent with the diagnostic purposes [8].

Chest CT are commonly used to detect various disorders such as abnormalities and disorders of the lung found on conventional chest x-rays or ultrasound, help diagnose the causes of clinical signs or symptoms of disease of the chest, such as cough, shortness of breath, chest pain, or fever. It is also used to detect and evaluate the extent of tumors that arise in the chest, or

Appendix B

tumors that have spread there from other parts of the body. And to evaluate the progress of disease and effects of therapy [9].

MATERIALS AND METHODS

Two detectors were used in this study: a 4-slice and a 16-slice CT scanner by Siemens. A total of 60 patients undergoing chest CT at Alneilain diagnostic center were included in this study. Routine image acquisition was performed using AEC settings. Using AEC dose can increase or decrease depending on reference image quality setting at the time of installation. The patients were divided into two groups: one before rotation time reduction and the other afterwards. The following parameters were entered: kV, (tube current), gender, height, weight. Patient doses were determined by using the CTD Ivol expressed in mGy and the DLP in mGy.cm as provided on the scanner console. These values were very valuable for statistical purposes, as they might possibly allow for analysis of scanner dependency. The collection of patient exposure parameters was done using patient dose data sheet. The rotation time was altered from 1 second to 0.7 seconds for the 4-slice detector and from 0.7 seconds to 0.5 seconds for the 16-slice detector. The effects on the dose and image quality were assessed. Three consultant radiologists evaluated all the medical images. Patients with gross pathology or studies that required special scanning parameters for any reason were excluded.

RESULTS

Table 1 below shows the comparison of CTD Ivol before and after reduction of rotation time. Table 2 below shows the comparison of DLP before and after reduction of rotation time. Table 3 below shows the comparison of noise expressed in SD before and after reduction of rotation time.

Table 1: CTDIvol comparisons

Detector	Before time reduction	After time reduction	Reduction %
4-slice	20.60±6.3 (11.7-15.6)	14.60±1.7 (11-15)	34.1%
16-slice	4.67±1.3 (2.6-7.3)	4.40±1.4 (1.9-6.8)	5.9%

Table 2: DLP comparisons

Detector	Before time reduction	After time reduction	Reduction %
4-slice	651.8±133.2 (463.7-967.3)	490.6±83.4 (358.1-673.1)	28.2%
16-slice	180.6±57.7 (74.0-337.0)	152.6±90.4 (49.0-335.0)	16.8%

Table 3: Noise SD

Detector	Before time reduction	After time reduction	Increase
4-slice	18.3±4.5 (8.6-28.5)	18.5±2.5 13.8-22.3)	0.2
16-slice	39.0±23.6 (17.9-87.3)	43.9±22.8 (8.8-75.8)	4.3

DISCUSSIONS

Reduction of dose remains to be a challenge in CT scans. Nevertheless it is associated with increase in noise. Image Noise is one of the primary factors in CT Image Quality Noise (specifically, quantum noise) is generally characterized by graininess, or a salt and pepper pattern on the image.

In this study, the scan exposure parameters for 4-Slice scanner were 120 kVp and 200 m As and standard pitch of 1.0 and rotation time 1.0 sec. Keeping these parameters constant only changing in the Time per rotation to 0.7 sec the reduction in dose was evaluated.

For 16-Slice scanner the exposure parameters were 120 kVp and 100mAs and a pitch of 0.87and rotation time 0.7 sec. After changing the rotation time to 0.5 sec with the same scan exposure parameters reduction to dose was also examined. The increase in noise determined by increase in Hounsfield units was tested.

The CTDI is defined to represent an approximation to the average absorbed dose to a

Appendix B

particular location in a standard acrylic phantom from multiple CT slices. CTDIw is a weighted average of the CTDI's at the center and periphery of the phantom. CTDIvol is similar to CTDIw but also includes the effect of pitch on the radiation dose. It is the CTDIvol that is displayed on the CT console and dose report. $CTDI_{vol}$ is a useful indicator of the radiation output for a specific exam protocol, because it takes into account protocol-specific information such as pitch. In this study the mean $CTDI_{vol}$ for the 4-Slice was 20.60 mGy before rotation time reduction and 14.6 mGy afterwards. And the reduction obtained was 34.1%.For the 16 –Scanner the mean $CTDI_{vol}$ before and after rotation time reduction were 4.67mGy and 4.40mGy respectively.

The dose length product (DLP) is an indicator of the integrated radiation dose of an entire CT examination, or radiation deposit in patients. The DLP for the 4 –Slice scanner was reduced by 28.2% after rotation time reduction. For the 16-Scanner the reduction was 16.8.

The reduction in dose was slightly less than that obtained by reducing the m As and pitch (10) but was still significant while reproducing acceptable images Noise is one of the primary factors in CT Image Quality Noise (specifically, quantum noise) is generally characterized by graininess, or a salt and pepper pattern on the image. Noise is inversely related to the number of X-rays. In our study there was no significant change in the noise for the chest exam for 4-Slice scanner which increased from 18.2HU to 18.5HU. However there was a significant change in the 16-Slice scanner the mean noise increased from 39.0 to 43.9 HU. The image was still within in the acceptance range. The challenge is in finding a balance between dose and noise that allows the images to be of diagnostic quality while utilizing the lowest dose possible.

CONCLUSION

CT radiation dose optimization and reduction is a complex process that seems to stay motionless since years. Several parameters can result in dose reduction. Reducing the rotation time was found to significantly reduce the dose. The rotation time reduction is limited by the amount of increase in noise. A significant reduction in CTDIvol and DLP were observed with the 4-slice scanner, with a considerably low increase in noise. A lower reduction in dose and larger increase in noise was observed with the 16-slice scanner. The results are encouraging for further efforts in reducing the CT dose without compromising the clinical findings with an aim to fulfill the optimization requirements

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Appendix C

P ID	sex	age	kv	mAs	Slice thickness	$CTDI_{vol}$	DLP	CT_{NO}	SD	Protocol