Sudan University of Science & Technology

College of Graduate Studies

Assessment of Radiation Dose to Head, Chest and Abdomen of Adult Patients Underwent Computed Tomography Examination -Khartoum State - Sudan

تقييم الجرعة الاشعاعية للرأس و الصدر و البطن للمرضى البالغين الذين أجريت لهم فحوصات الأشعة المقطعية ـ ولاية الخرطوم ـ السودان

A thesis submitted for partial fulfillment for the requirement of MSc degree in Medical Physics

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الآيـة

قال تعالى:

﴿قُلْ لَا أَقُولُ لَكُمْ عِنْدِي خَزَائِنُ اللَّهِ وَلَا أَعْلَمُ الْغَيْبَ وَلَا أَقُولُ لَكُمْ إِنِّي مَلَكٌ ۖ إِنْ أَتَّبِعُ إِلَّا مَا يُوحَىٰ إِلَيَّ ۚ قُلْ هَلْ يَسْتَوِي الْأَعْمَىٰ وَالْبَصِيرُ ۚ أَفَلَا تَتَفَكَّرُونَ ﴾

سورة الأنعام ﴿٥٠﴾

Dedication

I dedicate this work:

To my two paradises, to my mom, to my mom, to my mom Halima and to my dad Hassan who supporting and helping me from my birth till now with no condition just to let me become a happier and successful person.

To my brothers and sisters.

To Dr. Hussein Ahmed Hussein who taught me.

To other doctors who taught me too.

To all my friends and colleagues.

Acknowledgement

I firstly thank Allah who gave me the power, mind and intention to go forward till I have finished my research.

Also I introduce my thanks to Dr. Hussein Ahmed. He helped me and gave my very useful information also he helped me to access patients data in his center, so great thanks to Modern medical centers and his staff especially the radiographer Abdallah.

Thanks to Al-Zytoona and Sawi hospitals for allowing me to get patient dose reports from their CT system's PACS and special thanks to the radiography technologist Abdallah for his assistance.

Abstract

This research aims to assess radiation dose received by adult patient underwent head, chest and abdomen CT examinations in three diagnostic CT centers in Khartoum-Sudan.

The Study data are from PACS of CT scanners at each center been chosen and from the literature. Dose reports for the chosen patient's examinations were firstly captured using screen-shot technique then only target parameters were recorded into excel sheets for further work. The effective dose, CTDIvol and DLP were calculated using statistical software, SPSS. Then, the effective dose been compared with maximum permissible dose values, Where CTDIvol and DLP been compared with the international DRLs.

The effective dose was found to be: 5.65, 1.08 and 1.72mSv for head, chest and abdomen respectively which mean they are within permissible range (20 mSv maximum). And the average CTDIvol are 60.5, 13.18 and 15.15mGy for head, chest and abdomen respectively. Where the average DLP was: 1332.0, 593.1 and 2121.4mGy*cm for head, chest and abdomen respectively.

The average effective dose is within the maximum permissible dose range which is 2-20mSv. And the average CTDIvol is within IDRLs, where the average DLP is within international DRLs for head and chest but higher than international DRLs for Abdomen examinations. Also were found clearly that the three CT centers have a different effective doses, SSDE, CTDIvol and DLP for the same exam type.

Future studies must use our local DRLs (after been established) and make a fair judgement on the diagnostic CT centers and its staffs.

المستخلص

يهدف هذا البحث إلى تقييم الجرعة الإشعاعية التي تلقاها المريض البالغ الذي أخضع لفحوصات الاشعة المقطعية للرأس والصدر والبطن في الخرطوم - السودان.

بيانات الدراسة مأخوذة من انظمة ارشفة الصور والاتصلات الخاصة بماسحات التصوير المقطعي المحوسب في كل مركز من المراكز التي تم اختيارها وكذلك من الدراسات السابقة. تم الحصول علي تقارير الجرعة لفحوصات المريض المختار ة باستخدام تقنية لقطة الشاشة ، ثم تم تسجيل المعاملات المستهدفة فقط في أوراق الإكسل لمزيد من العمل. تم حساب الجرعة الفعالةِ ,مؤشر الجرعة الحجمي للفحص المقطعي و حاصل ضرب الجرعة في الطول باستخدام برنامج الحزمة الاحصائية للعلوم الاجتماعية. ثم تم مقارنة الجرعة الفعالة مع أقصى قيم الجرعة المسموح بها ، في حين تم مقارنة مؤشر الجرعة الحجمي للفحص المقطعي وحاصل ضرب الجرعة في الطول مع المستويات المرجعية التشخيصية الدولية.

وُجد أن الجرعة الفعالة هي:(5.65 و 1.08 و 1.72) مللي - سيفرت للرأس والصدر والبطن على التوالي ، مما يعني أنها تقع ضمن النطاق المسموح به (الحد الأقصى 20 مللي سيفرت). ومتوسط مؤشر الجرعة الحجمي للفحص المقطعي هو:(60.5 ، 13.18 و 15.15) مللي - قري للرأس والصدر والبطن على التوالي. في حين كان متوسط حاصل ضرب الجرعة في الطول: (1332.0، 593.1و 2121.4)مللي –قري*سنتمتر للرأس والصدر والبطن على التوالي .

متوسط الجرعة الفعالة في حدود القيم المسموحة بها وهي (من 2 الي 20 مللي سيفرت). و أن متوسط مؤشر الجرعة الحجمي للفحص المقطعي ضمن المستويات المرجعية التشخيصية العالمية, في حين أن متوسط حاصل ضرب الجرعة في الحجم ضمن المستويات المرجعية التشخيصية العالمية في فحصي الرأس والصدر وغير ذلك بالنسبة لفحص البظن. كما وجد بوضوح أن: الجرعات الفعالة ،مؤشر الجرعة الحجمي للفحص المقطعي , الجرعة الفعالة المرتبطة بالحجم وحاصل ضرب الجرعة في الطول لها قيم متلفة في المراكز الاشعة المقطعية الثلاثة بالنسبة لنفس الفحص.

المحلية (بعد اكنمال إنشائها) يجب أن تستخدم الدراسات المستقبلية لدينا المستويات المرجعية التشخيصية وإصدار حكم عادل على مراكز التصوير المقطعي التشخيصي وموظفيها.

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List of abbreviations.

CT	Computed Tomography
MSCT	computed tomography
SSCT	Single slice computed tomography
ALARA	As low as reasonable achievable
LNT	Linear no threshold model
DLP	Dose length product
ED	Effective dose
$CTDI_{100}$	Computed tomography dose index 100
$CTDI_{vol}$	Volume computed tomography dose index
ACR	American collage of radiography
MSAD	Multi-slice
FDA	Food and drug association
$CTDI_{w}$	Weighted computed dose index
MRI	Magnetic resonance Imaging
AEC	Automatic exposure control
KVp	x-ray tube high tension voltage
EC	European commission

Chapter one

Introduction

Chapter one

Introduction

1.1 Introduction

Computed tomography is a well-known imaging technology used for diagnosis of various diseases. It's as an upgrade of conventional x-ray imaging in that it also uses x-ray to format the radiographic images. By the time the technology of the CT is improved, from axial scanners to helical scanners and from single-slice scanners CT (SSCT) to multi-slice scanners (MSCT) (Slovis 2002, Wiest et al. 2002, Linton and Mettler 2003). As a result many more medical examinations were introduced to be carried out by the CT(Shrimpton & Wall 1992, Brenner & Hall 2007). consequently computed tomography (CT) became the main contributor of patient medical exposure(Slovis 2002, Linton & Mettler 2003). In the mid-1990s, CT scanning accounted for about 4% of procedures and about 40% of the collective dose in diagnostic radiology. In large hospitals, CT scanning now accounts for about 15% of procedures and 75% of the diagnostic radiation dose received by patients(Wiest et al. 2002).

Several studies been conducted in this field. In a review article cried out to develop an accurate Monte Carlo program for assessing radiation dose from CT examinations and when they combined their developed program with computer models of actual patients, the program is become able to provide accurate dose estimates for specific patients(Li et al. 2011a, Li et al. 2011b). Another review article cried out to estimate cumulative radiation exposure and lifetime attributable risk (Shafer, 2001) of radiation induced cancer from CT of adult patient and they found that the 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(Sodickson et al. 2009). Another additional research have been conducted to assess and evaluate patient radian doses for adult's common CT examinants to derive local diagnose guidance levels for common CT examinants and they found that doses of patients from CT examinants is not fully optimized(Sadri et al. 2013).

Today CT is known to be the main contributor of patient dose arising from medical exposure than any other diagnostic procedure used(Durand and Mahesh 2012). Furthermore there are still wide variations in technique and dose between CT centers for similar examinations(Shrimpton et al. 2006). Sometimes the radiation dose from CT becomes unreasonably very high when a radiographer in his way to make an excellent diagnostic image(Slovis, 2002). This means ALARA principle of radiation protection is not applied which is one of three important radiation principles(Lee et al. 2008). Radiation is always considered harmful with no safety threshold (Linear no threshold – LNT model), whenever there is a radiation there is a risk. Generally, radiation detriments or effects are divided into two types according to radiation dose level, deterministic effects when the level is above specific threshold and stochastic effects when its level is very low. The estimation of radiation dose and effective dose will give a clear vision towards predicted radiation risks to the patient. Unfortunately this cannot be achieved directly from a single patient's CT scan, instead other methods should be used to do this(McNitt-Gray 2002, Li et al. 2011b). Knowledge of individual's organ absorbed dose permits the determination of the probability of inducing deterministic effects such as skin burns or epilation and any corresponding stochastic risks of carcinogenesis, and genetic effects. In addition, dose to conceptus of a pregnant patient quantifies any possible detrimental effect in the irradiated embryo or fetus(Huda & Vance, 2007). Radiation dose awareness and optimization in CT can greatly benefit from a dose-reporting system that provides dose and risk estimates specific to each patient and each CT examination(Li et al. 2011b).

This research is conducted to assess radiation dose received by adult patient head and abdomen in CT examination in Khartoum state(Deak et al. 2010, Li et al. 2011b).

We wonder if radiation dose is within the maximum permissible range or not.

This research is aimed to assess radiation doses received by patients. Because unnecessary additional radiation doses may lead either to deterministic effects or will increase the probability of the stochastic effects(Siegel)

1.2 Problem of the study

Radiation dose received by the patient from CT scan is much greater than any other diagnostic x-ray modality. Also there are wide variations in technique and dose between CT centers for similar examinations.

1.3 Objectives

1.3.1 General objectives

To assess radiation doses for computed tomography of head, chest and Abdomen of adult patients.

1.3.2 Specific objectives

- ❖ Calculate or/and record effective dose, CTDIvol and DLP then compare them with the DRLs.
- ❖ Calculation of effective dose using DLP and compare it with the standard dose limits.
- ❖ Compare effective dose between different exam type.

1.4 Thesis outline

The thesis is about the assessment of patients' CT dose in different diagnostic CT centers.

Chapter one is the introduction to the thesis. Where

Chapter two contains some basic background of this study and a summary of related previews studies.

Chapter three describes methods, materials, techniques used in this study.

Chapter four is dedicated for results.

Chapter five presents discussions, conclusions and recommendations.

Chapter two

Literature review

Chapter two

Literature review

2.1 Introduction

X-ray, or unknown- ray. Whereas x is expressed to the unknown thing. X-rays are a type of ionizing radiation used in medical imaging to format radiographic image for the patients chosen anatomical structure as a camera uses visible light to create a photo image. **(**Xrayrisk.com**)** The X-rays were not developed but they were discovered. (Bushong, 2013)

It is discovered by Wilhelm Conrad Roentgen in November 1895 through experimentation of cathode ray tubes. In his experiment setup he energized the cathode with gradually arising voltage and he noticed that that adjacent fluorescence screen made of barium platino-cyanide is lit up although is separated from the cathode with a heavy black cardboard. And he put another matter in between the screen and the cathode tube, and still there is a fluorescence. Even Mr. Roentgen place his hand in front of the fluoro-screen, then he observed that the shadow of his hand is on the fluoro-screen. These above facts approve the high penetration power of this new kind of electromagnetic radiation unlike any other radiation discovered before, it the x-ray. Few years later a comprehensive experiments and researches were conducted concerning with this revolutional discovery. Consequently, they have introduced to us new methods, techniques for producing x-radiation and its useful applications. (Morton, 1918)

2.2 X-ray Generation

x-ray is produced when a fast electron collides with a heavy metallic target such as tungsten; and there are two different probabilities of interaction; a high speed stops, break or slows suddenly at the target producing what is known as

Bremsstrahlung radiation which is the primary sources of x-ray produced by an x-ray tube. Here the electron either hit a target nucleus directly (not often) or their path take them closer to the nucleus. (Goaz 2011, Maher & Edyvean) The second probability is the characteristic radiation. It occurs when an electron from the cathode displaces an electron from an inner-shell of the tungsten target atom, thereby ionizing the target atom. When this displacement happens, another electron in an outer-shell of the tungsten atom is quickly into the vacancy of the inner-shell. Then a photon is emitted with an energy equivalent to the difference of the two orbital binding energies. Characteristic radiation has a higher intensity than the continuous radiation (Bremsstrahlung) thus it is preferred. But it is produced in a minor quantities by the x-ray tubes than of characteristic radiation. (Goaz, 2011)

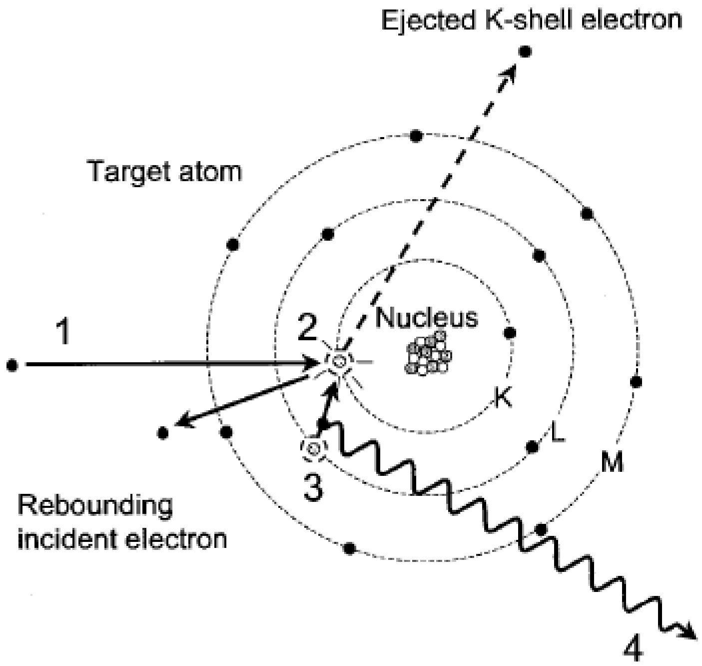

FIG. 2.1: Generation of a bremsstrahlung radiation in a target atom.

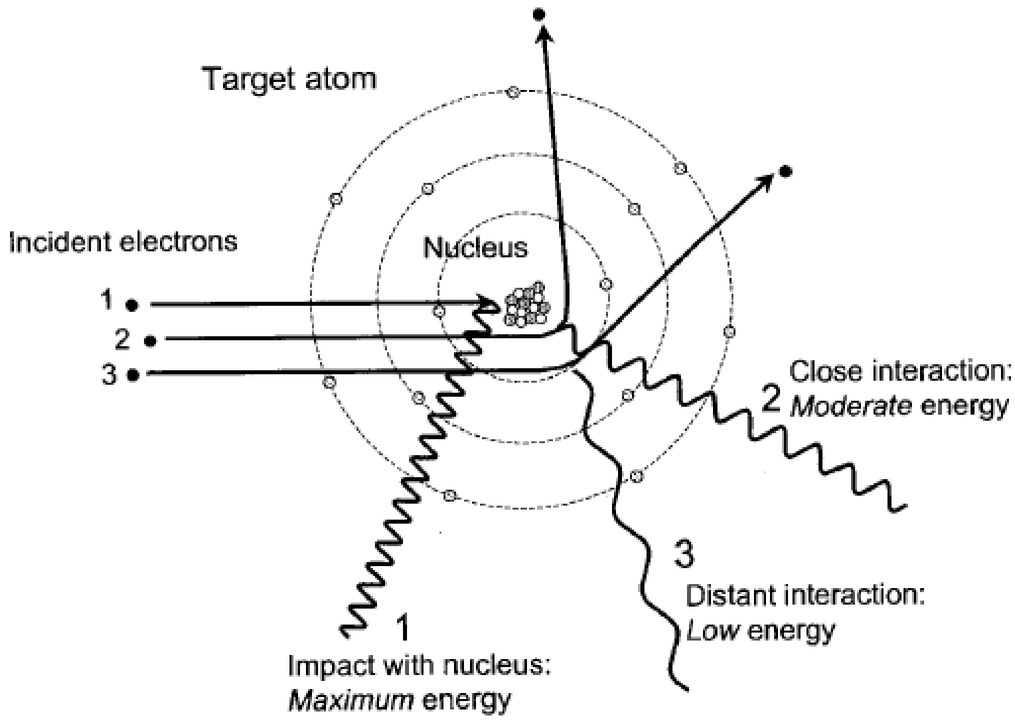

FIG. 2.2: Generation of a bremsstrahlung radiation in a target atom.

As in the discovery of the x-ray it was by using a simple configuration such as cathode tube and a power supply that radiation was not fully controlled and the amount of radiation was not known in addition it energy was not high enough either to penetrate a thick body or to form a high resolution radiographic image, A numerus researches been conducted to increase the efficiency of x-ray tubes and to theme safer to the patient.

There are two commonly used different methods of x-ray generation. First one is by deceleration fast electrons by firing them to a heavy metal target; by this electron will interacts with two possible way depending on distance between electron trajectory and the atom and its nucleus. If the impact factor is very close to the nucleus; is between the nucleus and the first atomic shell, electron path direction will be changed and a soft radiation is emitted called braking radiation or Bremsstrahlung (Bremsstrahlung is a German word means braking). And if the path is not very close to the atom nucleus, the incident electron will knock out the orbital electron from its shell creating a hole. This will cause the upper electron shell to drop down to fill this hole and the energy difference between the filling electron's orbit and the hole orbit will be emitted as x-ray; characteristic x-ray. (Buzug, 2008) The second method is the production of x-ray via synchrotrons. Here, the electrons beam is forced to move in a circular path with gradually increasing accelerating voltage applied to this path. Eventually the electron beam will be accelerated to the required kinetic energy. At the path exit window an angled -heavy target material is put to interact with electron beam to produce the x-ray. (Kanal)

2.3 Minimum requirements for production of x-ray

To produce x-rays at least these components must be found: source and target of electrons, an evacuated glass tube and connection of electrodes to a high voltage source. (Kanal, Bushberg & Boone 2011)

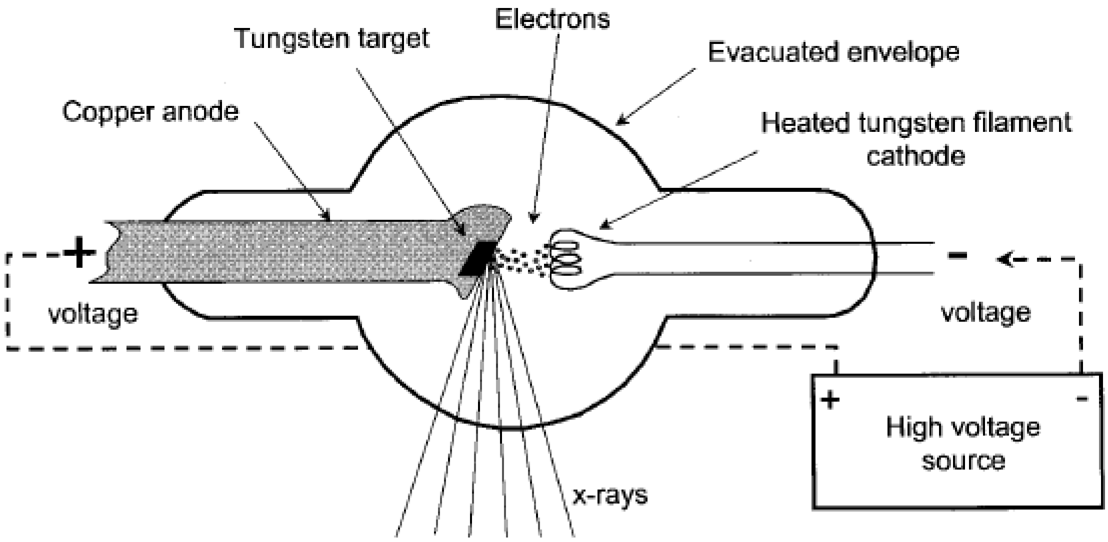

FIG. 2.3: Minimum requirements for x-ray production.

2.4 X-ray Cathode

It is source of the electrons. It consist of a helical tungsten wire called filament connected to current source. This wire is then surrounded by a metallic cup called focusing cup, this is to make electron trajectory narrower when it leave the hot cathode in their way to the anode. The mechanism of focusing is by applying electrical bias to the focusing cup. Tungsten wire is chosen because of the properties it has high thermal capacity (high melting point) to prevent anode from melting, high atomic number (Z= 74) to increase electron density and small tendency to evaporation. The current flow will heat up this filament producing thermal electrons which will be twisted through a high voltage to the target material (tungsten slap) at the anode side. Note: despite these facts in mammography they use another material other than the tungsten such as molybdenum and in order to produce soft radiation that is not harmful like the hard radiation in diagnose of the soft tissue of the breast. Filament circuit – (10V, 7A), the tube current (rate of e flow from cathode to anode) is controlled by adjusting the filament current. (Kanal)

2.5 Types of x-ray tubes

There are two types of x-ray tubes designs. The conventional x-ray tube which has static anode and the modern one which has rotating anode. The rotating anode is used to help dissipate the electric heat at the anode due to the electron collision with the target (the anode).

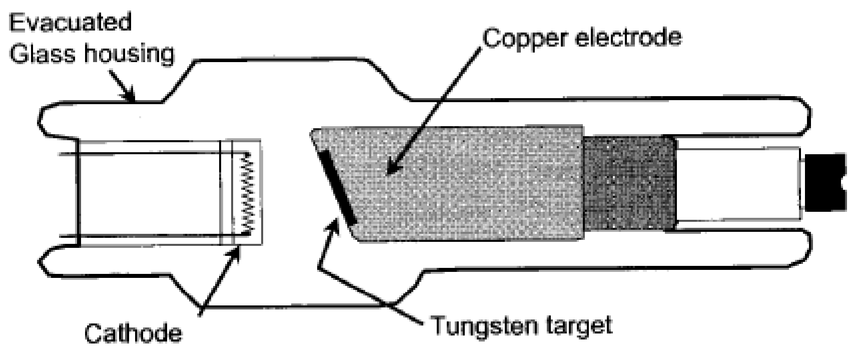

FIG. 2.4: the conventional x-ray tube with the fixed anode.

The fixed anode is inserted into a copper block. The function of this copper block is to remove the heat from the hot tungsten target.

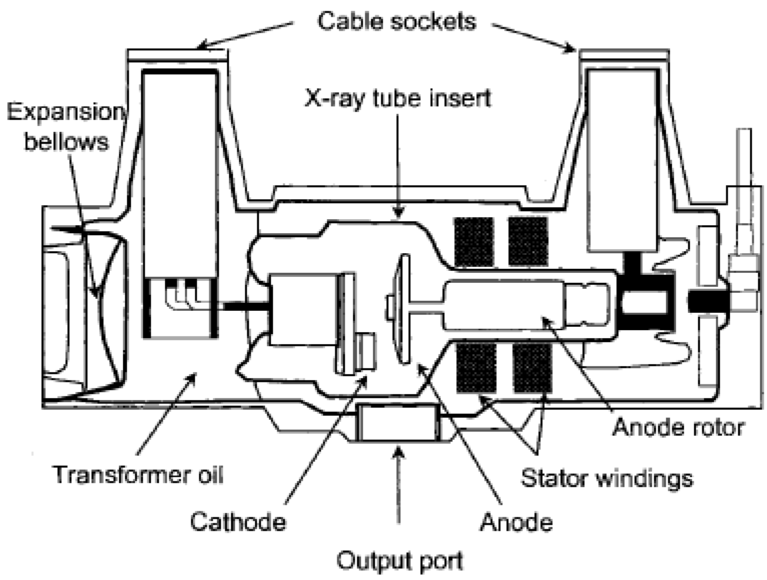

FIG. 2.5: The basic major components of a modern X-ray tube

2.6 Special Tube Designs

Much efforts had been done to increase heat capacity of X-ray tubes anode, but the power of these tubes still limited. Because there is a desire for higher power tubes with lower weight at the same time special tubes had been developed recently. One of these new x-ray designs abandons the solid-state principle of the anode.in these tubes a liquid metal jet is subjected to the fast electrons. Another one is grid-biased tube. A grid biased tube have a focusing cup that is electrically isolated from the filament and maintained at a more negative voltage. So, when the bias voltage is sufficiently large, the resulting electric field lines shut off the tube current. Turning off the grid bias allows the tube current to flow and x-rays to be produced. The grid biased tube is used tin applications such as pulsed fluoroscopy and cineangiocardiography, where rapid x-ray pulsing is required. (Bushberg & Boone, 2011)

2.7 Temperature Load

The major problem in x-ray production is the heat, almost about 99% of the electricity is converted to heat and only 1 percent is converted to x-ray, so the target material is chosen to have a high melting point or big heat capacity such as tungsten to overcome melting. Also another technique is used is the rotating anode (Bushberg & Boone, 2011) The quantum efficiency of the conversion from kinetic energy into x-ray radiation and the work efficiency using tungsten anode with acceleration voltage supply of 140 kv is roughly 0.01, Which means that 99% of the kinetic energy is transferred locally to heat at in the anode and only 1% is produced as x-ray. Hence the heat is the most common issue of x-ray tubes. To solve this, they used the rotating anode to dissipate the heat and cool the anode down. (Buzug, 2008)

2.8 Factors controlling x-ray beam

As a general though, there are different applications for x-ray in the medical sector, thus the produced x-ray should be modified to better suit the chosen procedure so as to produce a diagnostic image and to reduce x-ray detriments. It is need to adjust exposure time, exposure rate (mA), beam energy (kVp and filtration), beam shape (collimation) and target-patient distance (long or short cone). (Goaz, 2011)

2.8.1.Exposure time

Considering that the rate of generated photon is kept constant for the specific session, then, as the time period of the generated photon to reach the patient increases, radiation dose to the patient will increase. Hence an optimized time should be chosen. The amount of radiation that received by the patient is calculated from this equation; amount of radiation received equal to mA times s (mA is the tube current in mili-amperes, s is the time in seconds). (Goaz, 2011)

2.8.2.Tube current (mA)

Introduces the changed in the spectrum of photons that results from increasing tube current (mA) while maintaining constant tube voltage (kVp) and exposure time (s). There is a linear relationship between mA and the quantity of radiation, as mA is increased more thermal electrons are released from the cathode, hence more radiation produced at the anode side. Because the total quantity of received radiation in determined by summing up energies of individual same-energy-photons with constant photons fluency, thereby the quantity of radiation remains constant as long as the product of tube current and time is const. (Goaz, 2011)

2.8.3.Tube voltage (kVp)

Increasing the kVp increases the potential difference between the cathode and anode in the x-ray tube, thereby the kinetic energy of each accelerated electron when it reaching the target (the anode). The potential difference (kVp) is increased either to increase the number of generated photons, their mean energy, their maximal energy or all of these. (Goaz, 2011)

2.8.4.Filtration

X-ray beam consists of spectrum of x-ray photons of different energies, but only photons with energy high enough to penetrate the patient and low enough to be stopped by the image receptor (e.g. film), are useful for the diagnostic radiology purpose. And x-ray photons with insufficient energy to penetrate through, don't contribute in the formation of the image, they just add unnecessary radiation dose to the patient. Where photons with too high energy also are not useful although they don't add unnecessary dose to patient but they also don't contribute in image formation, rather they escape both the patient and the image receptor. A filter, a sheet of aluminum or any other material with certain thickness is positioned between the patient and the source of the x-ray. This sheet will absorbs low energy photons and allow photons of higher energy to pass through. (Goaz, 2011)

2.8.5.Collimation

A collimator is a metallic barrier (usually lead) with an aperture in the middle, used to reshape x-ray beam to fit only the desired area to be inspected or diagnosed as a result patient radiation dose will be further reduced. Also using collimator reduces the irradiated volume, image quality increases, because as the irradiated volume increases, the number of scattered photo/s increases. Consequently many of the scattered photons reach image receptor resulting in degradation of the quality of the image. (Goaz, 2011)

2.9 Interaction of x-ray with matter

As x-ray beam passes through an absorber its intensity is reduced. This reduction is due to the interactions of individual photons with the absorber atoms (the patient) that encountered. Photons interactions depend on many factors such as photon energy and target material type (density and atomic number), therefore, x-ray photon is either absorbed via photoelectric effect, scattered out of the beam or penetrate the medium without interaction. In the case of dental x-ray beam, three mechanism of scattering interactions take place; coherent scattering, Compton scattering and photoelectric scattering. The process of reduction of x-ray beam intensity or/and energy is called attenuation. (Goaz, 2011)

2.10 Mechanism of attenuation

As mentioned above attenuation of the x-ray beam is made by one or more way of the following photon interactions.

2.10.1. Rayleigh or Thomson scattering

Rayleigh or Thomson scattering is an elastic scattering event that occurs when the diameter of the scattering nucleus is smaller than the wave length of the incident radiation. In this type of scattering, incident and scattered x-ray wave are the same, meaning that there is no energy transfer process is accompanied. (Buzug, 2008)

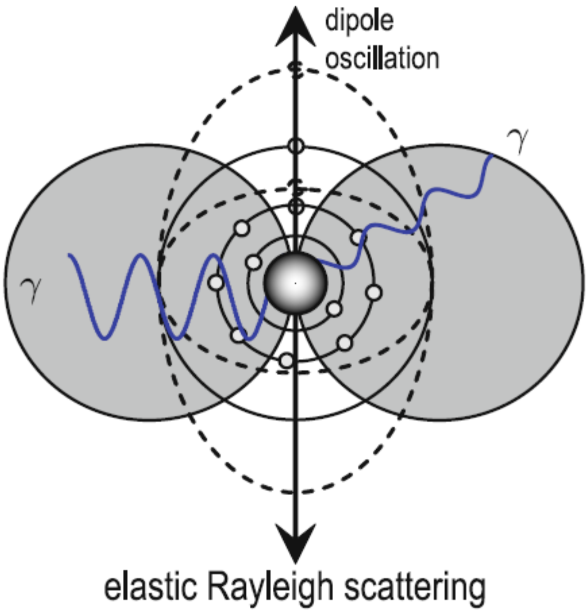

FIG. 2.6: Principle of photon-matter interaction. For the Rayleigh process, the dipole antenna characteristic is illustrated.

2.10.2. Photoelectric absorption

The entire energy of an X-ray photon, hv, can be absorbed by an atom if hv is higher than the binding energies of the atomic electrons of this atom. However, the entire energy of the incident electron is transferred to the atomic electron. This energy is divided into two parts, the major part is to release the atomic electron and the remaining is converted to kinetic energy to the electron thereby it leaves the atom as a photo-electron. The vacancy left by the electron that was kicked out is filled by electron from outer shells or, in the case of solids, by electrons from the band. For the fact that electron energy of higher shells have higher binding energies than that of the lower shells. The filling electron that will come from a higher energy level emits the excessive energy as a photon with energy equal to the energies difference between the higher level and the lower level. (Buzug, 2008)

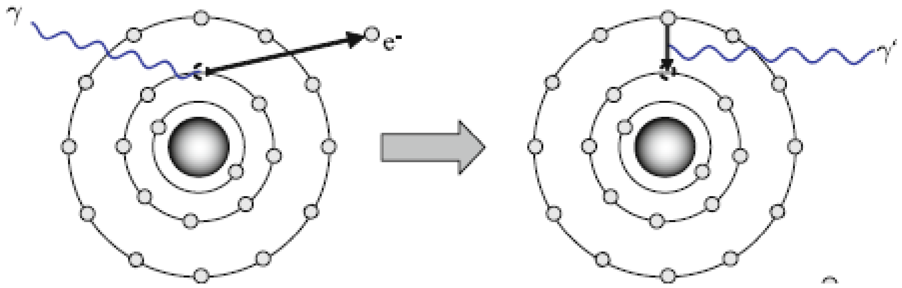

FIG.2.7: Photoelectric absorption

2.10.3. Compton scattering

This when an x-ray photon with energy hv collides with a quasi-free electron. Unlike to photoelectric absorption, x-ray photon loses only a part of its energy during the Compton collision. Thus the scattered photon is of lower energy and is continues traveling through the matter. The complementary part of this energy is carried by the kinetic energy of the recoiled electron that was kicked off the atom. This produced Compton electron is also called a secondary electron. (Buzug, 2008)

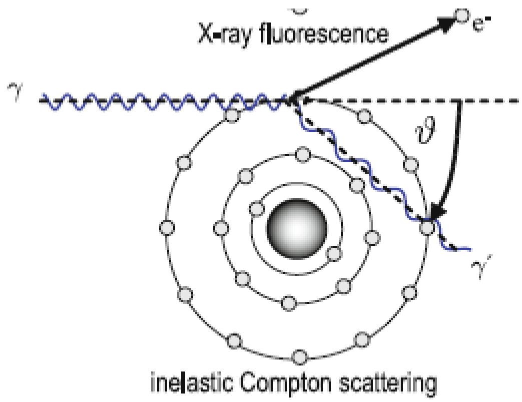

FIG. 2.8: Inelastic Compton scattering.

2.10.4. Pair production

Pair production occurs only for x-ray photons of energies greater than 1.022 MeV. The mechanism of pair production interaction is as this, the incident photon of high energy came closer to the nucleus, the strong Coulomb force totally converts this photon into two (pair) opposite sign electrons, a negative electron (e^-) and a positive electron (called positron, or the anti-electron, e^+). Positron has the same physical properties of the electron except their signs are different. Positron is short lived particle, in a few seconds it collides with a free electron and annihilates with it producing a pair of gamma photons. (Buzug, 2008)

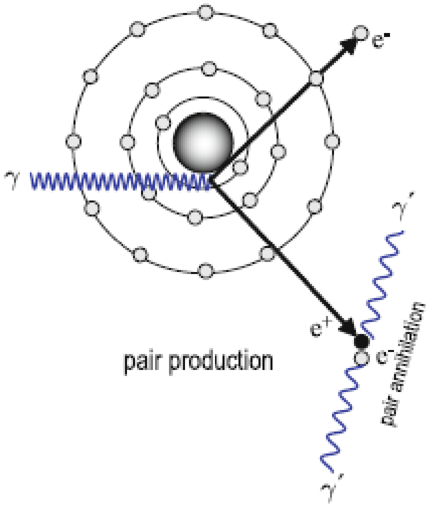

FIG. 2.9: A Successive pair production

2.11 Dosimetry

Dosimetry is the process of determination of radiation exposure quantity or dose to any field or body. Where the term dose is used to describe the amount of energy absorbed per unit mass at the site of interest. In other words, exposure is a measure of radiation based on its ability to produce ionization in air under standard conditions of temperature and pressure.

2.12 General dosimetry measures and its units

As the radiation measurement is an important task to do in order to get most of radiation advantages and avoid its detriments as much as possible, radiation studies are co process. Meaning that, a unified international measurement units is need, it is the international system of measurement (SI).

2.12.1 Exposure

It is the measure of radiation quantity, the capacity of radiation to ionize air. Traditional unit of exposure is Roentgen (R), where 1 R is amount of x-radiation or gamma that produce 2.08 ×109 ion pairs in 1 cc of air (STP). . The quantity must always be defined with respect to the specific material in which the interactions are taking place (eg, air kerma; kinetic energy released per unit mass, water kerma). From radiation exposure, one can calculate the skin entrance dose, which is important for deterministic effects such as tissue fibrosis and cataract. (Lee et al., 2008) There is no specific SI unit equivalent to the R, but in term of other SI units it is equal to coulombs per kilogram (C/kg). Note, the roentgen applies only for x-rays and gamma rays. Recently, roentgen had been replaced by air kerma (an acronym for kinetic energy (KE) released in matter). Kerma measures the KE transferred from photons to electrons and is expressed in units of dose (Gray).

2.12.2 Absorbed dose

Absorbed dose is a measure of energy absorbed by any type of ionizing radiation per unit mass of any type of matter. The SI unit for absorbed dose is Gray (Donnelly et al. 2001), 1Gy= 1 J/kg. (Siegel) Traditional unit of absorbed dose is Rad (radiation absorbed dose), whereas 1 rad is equivalent to 100 erg/g of absorbed dose. And 1 Gy = 100 rads. Dose not take into account where the radiation dose is absorbed or the relative radiosensitivity of the tissue being irradiated (Siegel)

2.12.3Equivalent dose

The equivalent dose is used to compare the biological effects of different types of radiation on a tissue or organ. The SI unit for equivalent dose Sievert, Sv. For diagnostic x-ray examination, 1 Sv equal 1 Gy.

2.12.4 Effective dose

Takes into account where the radiation dose is being absorbed and radiosensitivity of the tissue been irradiated. It can be calculated by summing up the product of absorbed doses of individual organs and their tissues weighted factors. However, because accurate measurements cannot be achieved for all patient organ doses and the risk coefficients specific to age, gender, and organ being irradiated, the estimated dose is calculated for an idealized 70-kg, 30-year-old anteropospatient**.** (Lee et al. 2008) The effective dose allows estimate of stochastic risks (cancer induction). Also effective dose is used for comparison between radiological procedures and between different radiological centers. Its unit is Sievert (Sv). It is given by the following equation;

$$E = \sum_T (w_T \times D_{T,R})$$

Where, w_T is the tissue weighting factor for tissue T and D_T is the absorbed dose of tissue T (Siegel)

Table (2.1): Tissue Weighting Factors from: ICRP Publication 103 (ICRP PUBLICATION 10**3)**

Tissue	$\mathbf{w_T}$	$\sum \mathbf{w}$
Bone marrow, (red). Colon. Lung. Stomach. Breast. Remainder tissues*	0.12	0.72
Gonads	0.08	0.08
Bladder, Oesophagus, liver, Thyroid	0.04	0.16
Bone surface, Brain, Salivary glands, Skin	0.01	0.04
	Total	1.00

* Reminder tissues: Adrenals, Extrathoracic (ET) region, Gall bladder, Heart, Kidneys, Lymphatic nodes, Muscle, Oral mucosa, Pancreas, Prostate(man), Small intestine, Spleen, Thymus, Uterus/cervix(women).

2.13 Computed tomography (CT)

In the past where there is no CT there was a major problem with the radiographic images that was the superimposition. Superimposition is the situation where images of the over-laying tissues of a chosen anatomic region overlap altogether, accordingly, successful diagnosis is lost. This problem is no longer encountered after invent of CT. In CT the final attenuation-based image is produced by imaging of very thin consecutive slices of the patient in addition it provides the opportunity to localize in three dimension. As a result, computerized tomography does not suffer from interference of structures of the patient inside the slice being imaged as conventional tomography does. Furthermore CT images have a superior contrast resolution compared to the planar radiography, but have inferior spatial resolution. (Sandborg, 1995)

FIG. 2.10: A photograph of CT machine (courtesy of AL-Zytoona hospital, Khartoum-Sudan)

2.14 Principles of operation

Typically there two necessary steps to derive a CT image. Firstly, physical measurements of attenuation of X rays traversing the patient in different directions and secondly mathematical calculations of the linear attenuation coefficients, m, all over the slices. The process is carried as follows. There is a stationary examination table where the patient lies on. Then an x-ray tube starts rotating around the patient in the table in a circular orbit in a plane perpendicular to the length-axis of the patient. A fan-shaped beam of variable thickness [1 - 10 mm], with width enough to cover both sides of the patient is used. Another part called image receptor is used. It is an array of several hundreds of small separated receptors. Readings from these receptors are been fed to a computer to produce the radiographic image from this raw data. Actually the computer does a numerous calculations before produces the final image. (Sandborg, 1995)

2.15 CT Specific Dose Measurements

2.15.1 CT Dose Index

The computed tomography dose index were firstly introduced by Shope et al in 1981 as a metric of quantifying radiation output from a CT examination which consists of multiple contiguous CT scans (multiple adjacent transverse rotations of the x-ray tube along the patient longitudinal axis) which cannot be assessed by the old dosimetry procedures. The CTDI method sought to create an index to reflect the average dose to a cylindrical phantom in the central region of a series of scans. DLP is one of the $CTDI_{vol}$derivatives DLP is the product of CTDI and the irradiated scan length. (Mccollough et al. 2011). This allows direct comparison of the radiation dose at different scanning parameter settings, even between scanners made by different manufacturers. However, the CT dose index does not indicate the precise dose for any individual patient, but is rather an index of the dose as measured and calculated in a polymethylmethacrylate phantom. Although the CT dose index is a valuable tool for protocol comparison, it does not take into account patient-associated parameters such as size, shape, and inhomogeneous composition. The dose index can be used in conjunction with patient size to estimate the absorbed dose (wiki.org). For any given scanning technique, patient dose depends on the size and attenuation of the patient (i.e., the greater the patient attenuation, the smaller the patient dose). Therefore, the displayed CT dose index is smaller than the actual dose delivered to young children and infants. The CT dose index is now commonly measured from one axial rotation of the scanner with use of a 100-mm pencil ionization chamber. An average (weighted) CT dose index is calculated by adding one-third of the central value and two-thirds of the peripheral values together. For scanning with a pitch that does not equal 1, weighted CT dose index has to be corrected by pitch factor (dose index divided by pitch) and then termed as volume CT dose index.

2.10.2 Dose-Length Product (DLP)

DLP, as its name implies, it is an indicator of the integrated radiation dose of an entire CT examination. The DLP incorporates number of scans and scan width (DLP = volume CT dose index × total scan length (in centimeters)). Unit is the mGy.cm. For conventional (non-spiral) scanning, scan length is the sum of all section collimations— for example, 25 mm (25 ×1 mm) for high-resolution CT. However, spiral and multi-detector CT oversample data at the beginning and at end of scan range. This because data are needed for raw data interpolation of the first and last sections. There are differences between manufacturers, but approximately one half-rotation at the beginning and another half rotation at the end have to be added to calculate the radiation exposure to the patient. Thus, the scan length, as provided by the scanner, should be increased by at least one table feed. (Lee et al. 2008) DLP is used to estimate effective dose (Siegel)

2.15.3 $CTDI_{100}$

It represents the accumulated multiple scan dose at the center of a 100-mm scan and underestimates the accumulated dose for longer scan lengths. It is thus smaller than the equilibrium dose or the MSAD. The CTDI100, like the CTDIFDA, requires integration of the radiation

2.15.4 $CTDI_{vol}$

$CTDI_{vol}$ is the metric used by the ACR for CT practice accreditation (Siegel). It is displayed on the scanner console before the initiation of a scan. It is a standardized measure of the radiation output of a CT system, measured in a cylindrical acrylic phantom. It enables users to gauge the amount of emitted output between different scan protocols or scanners. (Mccollough et. al 2011) The reference phantoms used to do this are two reference phantoms, 16 cm and 32 cm, depending on scan protocol. The 16 cm reference phantom represents all heads and for some pediatric bodies. (Boone, 2012)

2.15.5 $CTDI_w$

Stands for weighted computed tomography dose index. It is measured by placing a 100-mm-long ionization chamber in an acrylic phantom (reported by the manufacturer). It is given by the following equation;

$CTDI_w$ = 1/3 CTDI (center) + 2/3CTDI (periphery=surface) (Siegel)

2.16 Automatic Exposure Control

AEC is analogous to acquisition timing in general radiography. The user determines image quality requirements (as regards noise or the contrast-to-noise ratio), and the CT system determines the right tube current–time product. In practice, it is relatively straightforward for the system to deliver the desired image quality once that has been defined. However, it can be quite difficult to achieve agreement on the image quality requirement for various CT examination types and patient age groups. In the way to define the required image quality, user needs to remember that pretty pictures are not needed for all diagnostic tasks, instead a choice can be made between low noise and a low dose, depending on the diagnostic task. The CT system then adjusts the tube current during the gantry rotation, during movement along the z-axis, or during movement in all three dimensions, according to the patient's body habitus and the user's image quality requirements. Thus, differentiation between modulation of tube current to achieve a defined image quality, and the prescription of the desired image quality by the user together these tasks are referred to as AEC. (Mccollough et.al 2006) There are a variety of dose modulation systems(AEC), but their main purpose still the same, is to reduce the radiation dose received by the patient while sustaining diagnostic image quality by adjusting radiation dose according to the patient's attenuation (e.g. patient habitus, anatomical structure, angle of projection, …etc.). AEC systems have a number of potential advantages, including better control of patient radiation dose, avoidance of photon starvation artifacts, reduced load on the x-ray tube, and the maintenance of image quality in spite of different attenuation values on CT scans. With these benefits of AEC systems in mind, users should learn how to use and apply this systems properly. However, concerns about routine use of AEC still remain. Although AEC systems generally reduce radiation dose, image noise inevitably increases, particularly in the region adjacent to contrast material– and prosthesis-related artifacts (Mccollough et al. 2006).

2.17 CT generations

The arrangement of the x ray tube and the receptors have changed during years, e.g. in the first invent of the CT the receptors array (in fact it wasn't an array, it was only a single x-ray detector cell) and the x ray tube were fixed relative to each another and translating in a straight line one of them is over the patient and the another is under the patient. This movement is step-shot-record mode and because was one detector, just single scan is made in each traversal scan past the patient till the scan process completes 180 steps over 24 cm field of view (Hagi). In the second generation array of detectors are used and the x-ray tube emits the radiation over a large angle but the source and the array of detectors are translated as in the first generation. Multiple projections obtained during each traversal past the patient, hence scan time be reduced compared with the first generation. These different technical solutions being made to get better results and to solve some problems observed in computed tomography technique. These different technical solution being named ‘generations’. (Lee et al. 2008) Bellow, third and fourth generation were taken as an example of CT generations.

2.17.1 Third generation

In this CT scanner generation, the x-ray tube and the receptor array are located on opposite sides of the patient and both rotate around the patient during data acquisition. In this generation the receptor array consists of about 700 pressurized xenon detectors. (Lee et al. 2008)

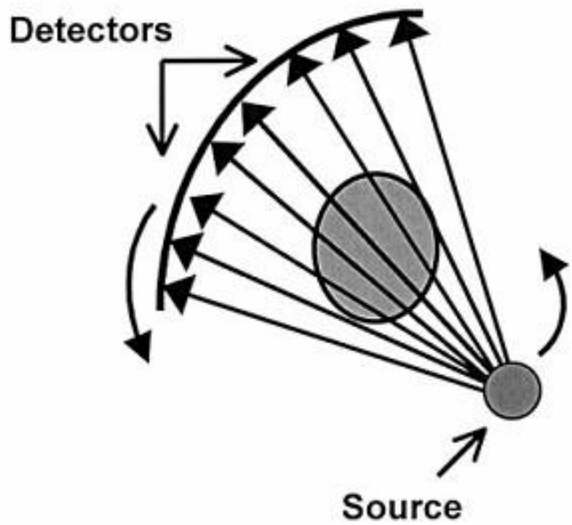

FIG. 2.11: Third Generation CT. (Hagi RAD309)

2.17.2 Fourth generation

In fourth CT generation only the x-ray tube rotates around the patient while receptors array remains stationary. Here, receptors are made from solid-state material and can be as many as 4000. Both, the third and fourth CT generation uses fan –beams and makes about 1000 projections. (Lee et al. 2008)

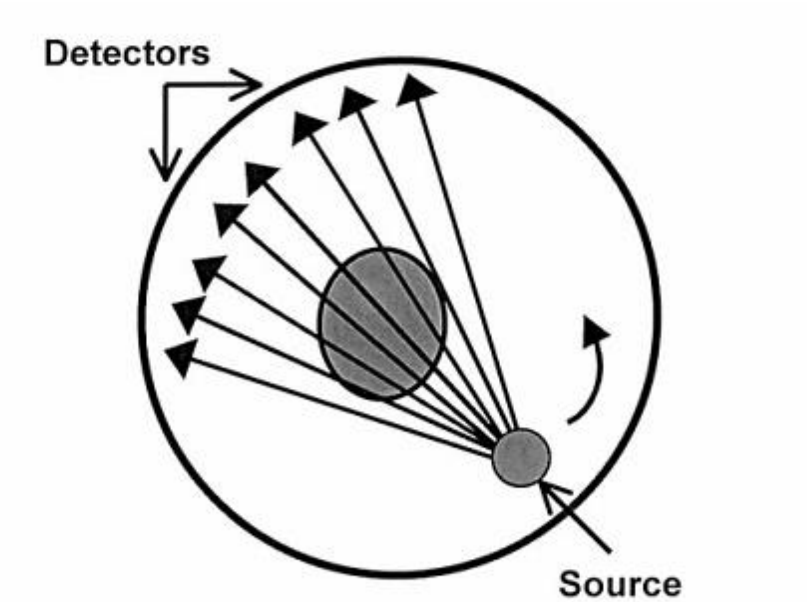

FIG. 2.12: Fourth Generation CT. (Hagi RAD309)

2.18 Some CT parameters

2.18.1 Slice thickness

Slice thickness and slice increment are central concepts that surround CT/MRI imaging. Slice thickness refers to the (often axial) resolution of the scan (2 mm in the illustration). Slice increment refers to the movement of the table for scanning the next slice (ranging from 1 mm to 4 mm in the illustration). Slice thickness is an important factor for the determination of image resolution of CT scanners. Generally there are three choices of choosing values of both slice thickness and slice increment as in the figure bellow, each choice of has its advantages and disadvantages, so optimization process must be applied. (Materialist.com)

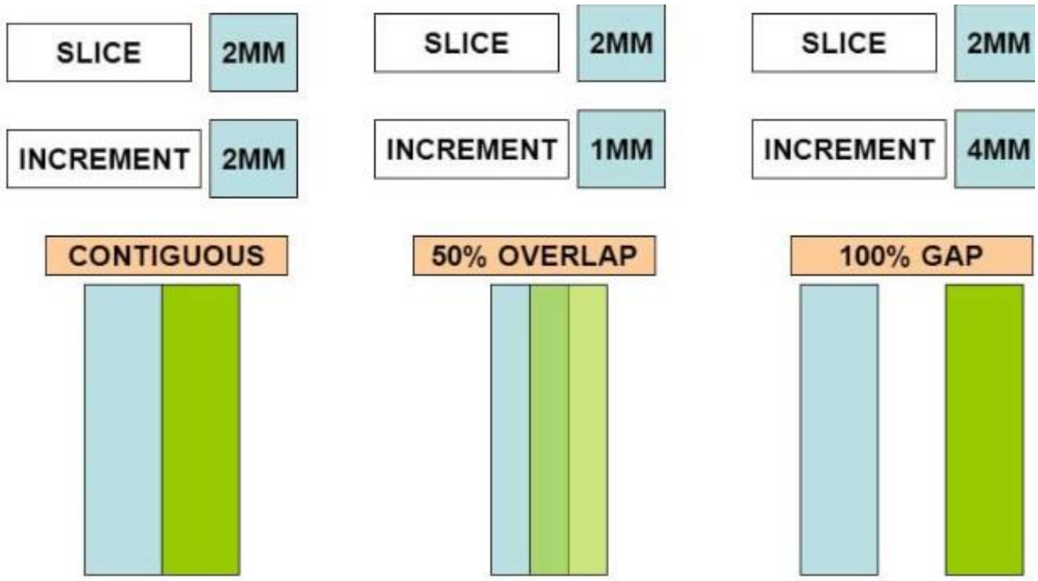

FIG. 2.13: The effect of choosing values for slice thickness and the slice increment.

2.18.2 Tube Current

The increase in either tube current (mA) or tube current duration (s) results in an improved image quality and decreased image noise but increased patient dose. The relationship between tube current or product of tube current and scan time (mAs) is known to be linear, increase of one results in of increase in the second almost the same percentage. Hence tube current must be controlled either manually or by using the new technique which is known as automate exposure control AEC. This AEC system is essentially a computer software and is named according to the specific vendor been used, for instant it is called CARE Dose 4D on Siemens, Dose –Right on Philips, Auto mA/Smart mA on GE and SUREExposure 3D on Toshiba. (Raman et al. 2013)

2.18.3 Tube voltage (KVp)

Radiation dose to the patient also changes with tube voltage. As tube voltage increases, patient dose increases and vice versa but fortunately image quality degradation is minimal compared to amount of change in the tube voltage thus this advantage can be used to decrease patient dose in some studies. Low –KVp protocols (100, 80 and even 70 KV) can be used for thin non-obese patients undergoing angiography, venous phase studies, cardiac or coronary studies, etc. According to the general rule of the thumb, the radiation dose changes with the square of Kvp, consequently a reduction in KVp from 120 to 100 KV reduces radiation dose by 33%, while a reduction of Kvp to 80 KV can reduce dose by 65%. (Raman et al. 2013)

2.18.4 Scan range

Scan range is the area boundaries over which the irradiation will be made by the CT machine, and it essentially covers at least the region or organ of interest. Reducing of scan range is needed as much as possible, because this reduces the area of irradiation, so the radiation dose be reduced. (Raman et. al. 2013)

2.18.5 Pitch

Pitch in multidetector and helical era is defined as table travel per rotation divided by beam collimation. Pitch <1 suggests overlap between adjacent acquisitions, pitch <<1 implies gaps between adjacent acquisitions and pitch of 1 suggests that acquisitions are contiguous, with neither overlap nor gaps. It is clear that a smaller pitch with increased overlap of anatomy and increased sampling at each location results in an increased radiation dose. Alternatively, a large pitch implies gaps in the anatomy and hence lowers radiation dose received by the patient. (Raman et al. 2013)

2.9 Previous studies

The arising number of CT examinations made the CT the major contributor of patient medical exposure across all other radiographic diagnostic modalities. It is impractical to directly measure the radiation dose absorbed by individual patients even when the radiation emitted by a machine is precisely known. Instead, radiation exposure may be quantified using various methods. Effective dose is used to quantify the radiation exposure associated with each CT examination, as this is one of the most frequently reported measurements. Further, effective dose allows comparison across the different types of CT studies and between CT and other imaging tests, facilitating comparison of CT to the most common radiology studies that patients undergo. The effective dose accounts for the amount of radiation, the exposed organs, and each organ's sensitivity to developing cancer from radiation exposure(Smith-Bindman et al. 2009**)**.

Early attempts of estimating CT doses were by using dose measurement versus depth summed over all x-ray tube angles and positions. As the CT been developed, other methods of radiation dose assessment been discovered (Goldman, 2007).

In a review article L. Sadri1 et.al assessed and evaluated patient doses for adult's common CT examinations. They measured volume computed tomography dose index and dose length product of four common CT examinations including head, head sinus, chest, abdomen and pelvis for 8 different CT scanners using standard head and body phantoms. The CTDIw for head base; head cerebrum, head sinus, chest and Abdomen were 71.8, 29.47, 35.8, 9.8, and 12.9 mGy, respectively. And the DLP for head, head sinus, chest and Abdomen were 400, 371, 225 and 482 mGy.cm. It is clear that the patient dose in terms of DLP values for head sinus are higher compared with the other studies while CTDIw values for head base and sinus were higher than EC measurements. Therefore, radiation dose of patient from CT examination in not fully optimized(Sadri et al. 2013).

Current methods for assessing and reporting radiation dose from CT examinations give inaccurate estimations because they don't use the patient body size which certainly will affect the radiation dose be received. A review article was conducted to develop a new technique. They estimated patient-specific radiation dose and also the stochastic effects from CT by combining a validated Monte Carlo program with patient-specific models that are derived from the patients' clinical CT data and supplemented by transformed models of reference adults and compared it with patient-generic CT dose quantities in current clinical use: the volume-weighted CT dose index ($CTDI_{vol}$) and the effective dose derived from the dose-length product (DLP). They found that for the two paediatric-aged patient that were chosen in their study, $CTDI_{vol}$ underestimated dose to large organs in the scan coverage by 30%-48%. And the effective dose derived from DLP using published conversion coefficients differed from that calculated using patient-specific organ dose values by -63% to 28% when the tissue weighting factors of ICRP 103 were used(Colang et al. 2007).

Although CTDI and DLP are given in the scanner output screen but neither of them provides the absorbed dose by any specific patient. They are very precise but not necessary accurate of patient dose(Durand and Mahesh, 2012). Because they are an estimate to a homogeneous cylindrical phantom which is not the patient(Bauhs et al. 2008). Dose estimation in computed tomography is challenging due to the vast variety of CT scanners and imaging protocols in use. In this article the authors attempted to evaluate the reliability and accurateness of the theoretical formalism implemented in CT-EXPO for dose assessment. In order to achieve this, they performed phantom dose measurement for three body regions (head, chest and pelvis) of an anthropomorphic Alderson phantom on a variety of SSCT and MSCT scanners (10 scanners; four 1-slice, four 4-slice and two 16-slice spiral CT scanners) for a representative scan protocols. Firstly they measured the scanners-specific normalized weighted CT dose index (nCTDIw) values and compared it with corresponding standards values used for dose calculations. Secondly they calculated effective doses for three CT scans (head, chest and pelvis), they measured it for the all 10 CT scanner been used from the readings of thermoluminescent dosimeters distributed inside an anthropomorphic Alderson phantom and compared it with the corresponding dose values computed with CT-EXPO. They found the differences between standard and individually measured nCTDIw values were less than 16%. And the statistical analysis yielded a highly significant correlation ($P<0.001$) between calculated and measured effective doses. And the systematic and random uncertainty of the dose values calculated using standard nCTDIw values was about -9 and +11%, respectively. Eventually, the phantom measurements and model calculations been carried out for a variety of CT scanners and representative scan protocols validate the reliability of the dosimetric formalism considered—at least for patients with a standard body size and a tube voltage of 120 kV been selected for the majority of CT scans performed in their study(Brix et al. 2004).

The dose of CT scan acquisitions has been substantially reduced, therefore the main contributor to the CT dose today is seems to be the localizer radiograph as they think. To approve this; they firstly, measured the dose distributions in anthropomorphic phantoms for three different body regions (head, thorax and Abdomen) and three positions of the x-ray tube (AP, PA and lateral views), then they compared the measured values to simulated data using Monte Carlo techniques for validation purposes. Then secondly, they calculated organ and effective dose values for various investigated localizer radiograph scenarios were and compared it with published dose values for standard CT and low-dose CT examinations. They have come to conclude that the localizer radiographs substantially contribute to the total dose of the whole CT examination, particularly in the case of dedicated low-dose scan used, e.g., for young patients or screening purposes(Schmidt et al. 2013). Today, the most accurate method of assessing CT dose and the effective dose is a Monte Carlo based program such as CT-EXPO(Boone, Linton & Mettler 2003).

Chapter three

Material and method

Chapter three

Material and method

3.0 Preface

A retrospective dose reports of adult patients who undertaken brain, chest and Abdomen examinations were collected from three diagnostic CT centers. In the first center, patients were 40 abdomen exams (20 female, 20 male) and 15 chest studies (8 females and 7 males). In the second center were 28 brain studies (16 females and 14 males). In the last center, 12 abdomen studies (10 females and 7 male). In fact there were some differences between these three centers in the information presented on their dose reports, this may be due to the scanners type, protocol used, or as preference of the specific center to show whatever they want. Information that are related to the study is recorded consisting DLP, $CTDI_{vol}$, KVp, Pitch and age. Then by using excel sheets, some parameters were been calculated using the existing variables. After that, SPSS v.16 was used to examine the statistical relationships between the most important parameters to our investigation such as the effective dose then record them for further explains.

3.1 Material

Because this study is retrospective, all study data are from PACS of CT scanners at each center been chosen and from the literature. Three CT machines in three different centers were chosen. Type, manufacturer and version of each CT scanner been used in this study is as in the following table:

Table (3.1): CT machines been used and its specifications.

Center	Manufacturer	No, of slices	Version
1	General Electric	16	Opima 350
2	Toshiba	16	Aguilion
3	General Electric	16	Optima

Also the statistical analyzing softwares (SPSS, version 20 and Microsoft excel program, version 2010) were been used.

3.2 Method

3.2.1 Technique used (data collection)

Dose reports of the chosen studies were firstly captured using screen-shot technique. Then target parameters from the dose reports were recorded into excel sheets for further work. Effective dose and SSDE were calculated for each study and recorded. The CTDIvol, CTDIw, DLP were compared between the different centers and with the maximum permissible value for the effective dose and the DRLs for the rest.

3.2.2 Analysis of data

All the data in this study were calculated and summarized using Microsoft excel program (version 2010) and SPSS software (version 20).

Chapter four

Results

Chapter four

Results

Table (4.1) show Total number of patients examination been chosen for this study.

Exam type	No. of patient studies
Head	38
Chest	17
ABD	52

Table (4.2) show DLP (mGy.cm) in Center A

Exam type	Minimum	Maximum
Chest	129.6	1314.3
ABD	267.2	3505.2

Table (4.3) show CTDIvol (mGy) in Center A

Exam type	Minimum	Maximum
Chest	1.9	402.0
ABD	4.0	45.2

Table (4.4) show Effective dose (mSv) in Center A

Exam type	Minimum	Maximum
Chest	4.0	52.6
ABD	1.8	18.4

Table (4.5) show in DLP (mGy.cm) in Center B

Exam type	Minimum	Maximum
Head	842.0	1650.0
ABD	235.4	1584.7

Table (4.6) show CTDIvol (mGy) in Center B

Exam type	Minimum	Maximum
Head	60.5	61.9
ABD	5.2	31.3

Table (4.7) show Effective (mSv) in Center B

Exam type	Minimum	Maximum
Head	1.8	3.6
ABD	3.5	23.8

Table (4.8) show DLP (mGy.cm) in Center C

Exam type	Minimum	Maximum
Head	2232.0	8115.0
ABD	2145.0	6201.0

Table (4.9) show CTDIvol (mGy) in Center C

Exam type	Minimum	Maximum
Head	12.4	54.1
ABD	5.7	15.9

Table (4.10) show Effective dose (mSv) in Center C

Exam type	Minimum	Maximum
Head	39.1	170.4
ABD	32.2	93.0

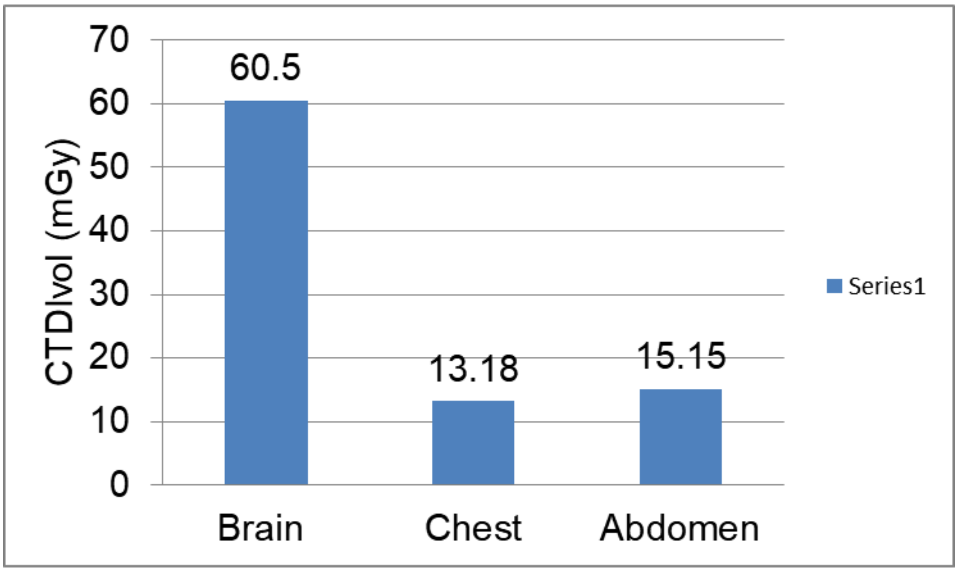

FIG. 4.1: Shows $CTDI_{vol}$ (mGy) vs exam type at 75^{th} percentile of the three centers.

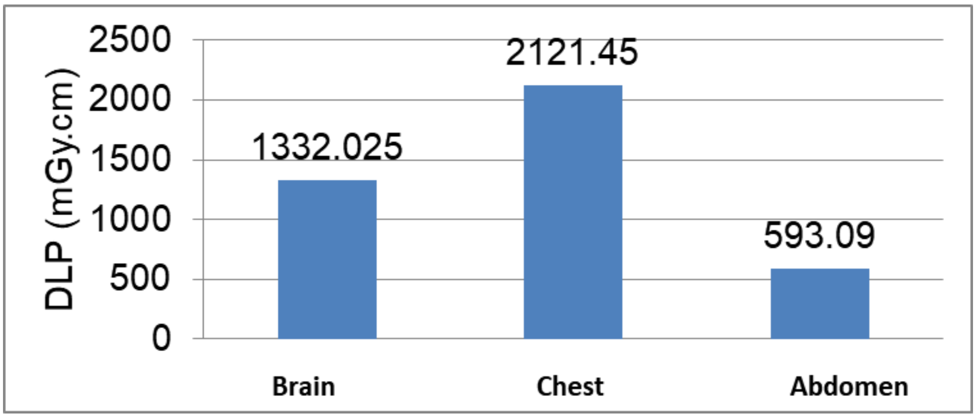

FIG. 4.2: Shows DLP (mGy.cm) vs exam type at 75^{th} percentile of the three centers.

Table (4.11) show $CTDI_{vol}$ (mGy) between the three centers at 75^{th} percentile.

Exam	Hospital A	Hospital B	Hospital C
Brain	-	60.5	54.1
Chest	13.2	-	
Abdomen	13.9	31.3	13.2

Table (4.12) show DLP (mGy.cm) between the three centers at 75th percentile.

Exam	Hospital A	Hospital B	Hospital C
Brain	-	1211.0	8115.0
	593.1	-	5125.5
abdomen	1665.4	1475.0	

Table (4.13) show Comparison of mean effective dose (mSv) in this study with other studies'.

Body region	This study	EC	UK	Orggi et al.	Osei & Darko	Clarke et al.	Tsai et al.	Aldrich et al.
Head	5.65	2.0	1.5	1.8	1.8	1.3	1.6	2.8
Chest	1.08	8.8	5.8	7.9	7.9	5.6	8.4	9.3
Abdomen	1.72	9.0	5.3	7.9	–	5.8	7.4	10.1

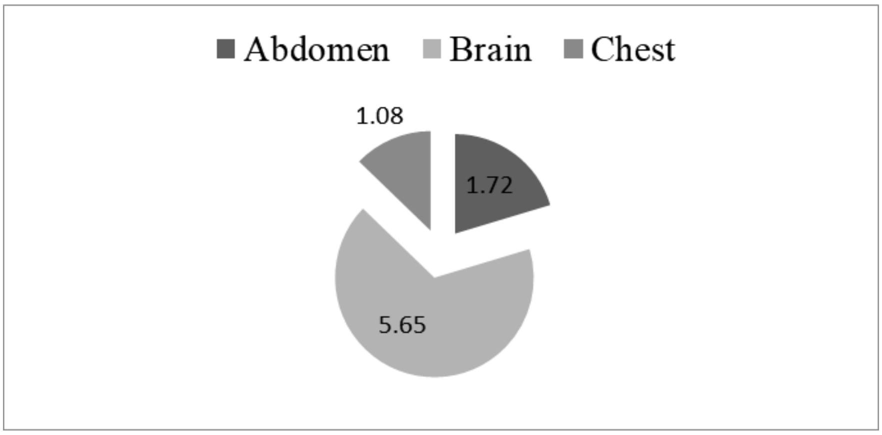

FIG. 4.3: Shows contribution of each exam type on total mean effective (in mGy) received by a patient.

Table (4.14) show Mean SSDE dose (in mGy) for every exam Type.

Exam type	Mean	St. Deviation
Abdomen	1.29	8.06
Brain	5.65	15.22
Chest	3.29	97.26

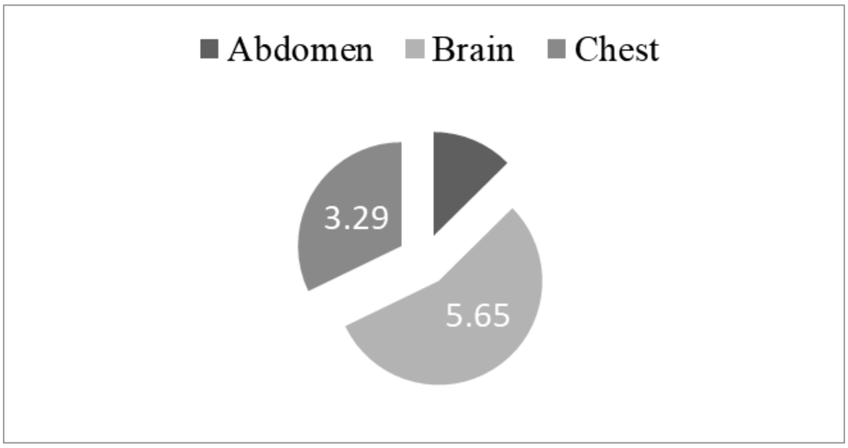

FIG. 4.4: Mean SSDE (mGy) vs. Exam Type.

Table (4.15) show Comparison of DLP (mGy.cm) in this study and other studies at the 75th percentile.

Exam	This study	UK	Switzerland	Malta	Canada	USA
Brain	1332.0	-	1000	736	1302	962
ABD	2121.4	610	-	492	874	443
Chest	593.1	745	650	539.4	521	781

Table (4.16) show Comparison of $CTDI_{vol}$ (mGy) in this study and other studies at the 75^{th} percentile.

Exam	This study	UK	Switzerland	Malta	Canada	USA
Brain	60.50	80	65	41	82	56
Chest	13.18	12	-	13.1	14	12
Abdomen	15.15	15	15	12.1	18	16

Chapter five

Discussion, Conclusion and recommendation

Chapter five

Discussion, Conclusion and recommendation

5.1 Discussion

Now a days there is an increased use of computed tomography in many diagnostic studies and it is preferred over other methodologies because of its unique advantages. Consequently it became the biggest contributor of patient's medical exposure. Therefore meaningful radiation measures must be done to know if the process of medical referral and practice of imaging are optimized or not.

The input parameters were 120 kv, 124=400mA and pitch in the range 0.7-2mm.

The average $CTDI_{vol}$ for head, chest and abdomen were, and they were: 60.5, 13.18 and 15.15mGy respectively (Table 4.1). Average DLP for head, chest and abdomen and they were: 1332.0, 2121.4 and 593.1mGy.cm respectively (Table 4.2). And finally, the average effective dose for head, chest and abdomen were calculated, and they were: 5.56, 1.08 and 1.72mSv respectively (Table 4.5).

The effective dose for the chest and abdomen are very lower compared to the other studies chosen for this purpose. The permissible dose range for the diagnostic CT is 2-20 mSv (Mettler, 2008). This means, the measured effective dose is within the range. In contrast, the effective dose to the head is very high compared with the other studies, but still within the permissible dose range. This increase in the effective dose of the head may be due to the protocol used.

Head $CTDI_{vol}$ is 13.9% lower than of UK's, 3.6% lower than Switzerland, 19% higher than Malta, 15% lower than Canada and 3.8% higher than USA. As it seems our measured head $CTDI_{vol}$ is higher than some studies (countries) by about 15%, there were other studies have a head $CTDI_{vol}$ higher than ours by about 19%, meaning that the net difference between lower one and the upper is about 4%, in other words, about 95% confidence that our measured head

$CTDI_{vol}$ is within the range. Keeping in mind that each country has its own DRLs which may not fit with any other country.

Chest $CTDI_{vol}$ is 4.7% higher than UK's, 3.0% higher than Malta, 3.0%lower than Canada and 4.7% higher than USA. Abdomen $CTDI_{vol}$ is 0.5% higher than both UK and Switzerland, 11.1% higher than Malta, 8.6% lower than Canada and 2.7% lower than USA. These differences in $CTDI_{vol}$ of both chest and Abdomen may be acceptable in light of the local DRLs (if they were stated) as discussed in the above paragraph.

Head DLP is 14.2% higher than Malta, 28.8% higher than Switzerland, 1.3% higher than Canada and 16.1% higher than USA. Abdomen DLP is 55.3% higher than UK, 62.3% higher than Malta, 41.6% higher than Canada and 65.4$ higher than USA. Chest DLP is 11.3% lower than UK, 4.6% lower than Switzerland, 4.7% higher than Malta, 6.5% higher than Canada and 13.7% lower than USA. The differences between our measured DLP head, chest and other countries been chosen are acceptable. But the DLP of Abdomen is far higher than most of the other countries chosen in this study. It is clear that this value need to be investigated in a urgent manner to know what makes it such higher like this so as to safe our patients from over dosing.

There are a remarkable differences in $CTDI_{vol}$ and DLP values between the three centers. For instance, $CTDI_{vol}$ of abdomen in center A (13.9mGy) is 38.5% higher than of center B (31.3mGy) and 2.6% lower than center C. And the DLP for the abdomen in center A (593.1mGy.cm) is 79.2% lower than of center C which is 5125.5mGy.cm. Really it is big differences between the centers in our measured $CTDI_{vol}$ and DLP. Although differences between centers are expected to found even if they are in same country with the same DRLs. These differences may refer to the protocol used, technique, or patient (age and weight) etc and they are quite small. But such difference of 79.2% difference as between abdomen DLP of center A (593.1mGy.cm) and C (5125.5mGy.cm) is very big and need to be investigated urgently.

5.2 Conclusion

Today computed tomography is the most contributor in radiation dose received by the patient from the medical modalities, so, optimization is hardly needed to get rid of the unnecessary radiation exposure from CT exams.

In this study, radiation dose experienced by adult patient in common CT examinations were assessed. Measured effective dose of this study were compared with other studies from other countries and found that it is within the maximum permissible dose range. Also measured $CTDI_{vol}$ and DLPs values were compared with the international DRLs and found that the radiation dose is within the maximum permissible dose range in some exams and not for others. Also, were found that some centers need a urgent investigation because of their abnormal high DLPs.

Although it is better to compare with the local (Sudan) DRLs than by international DRLs, but our local DRLs were not established yet till the writing of this research discussion. Other limitation is that the samples taken were not random enough, hence, some parameters such as patient weight, length may affect the measured values somehow if they were taken in consideration.

Our first intend was to use a Mote Carlo-base software (certainly CT- Expo program) in the assessment of patient dose from CT because of its high accuracy, but it was quite difficult and time consuming to us possess this program.

Future researchers should overcome these limitations to make a better evaluation study.

5.3 Recommendations

Future studies must use our local DRLs and make a fair judgement on diagnostic CT centers of our country.

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