Sudan University for Science and Technology

College of Graduate Studies

Assessment of Radiation Dose and Image Quality in Computed Tomography

تقييم الجرعة الإشعاعية وجودة الصورة في التصوير بالأشعة المقطعية

A thesis submitted for fulfillment of the requirements of PhD degree in Medical Physics

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الآية

قَالَ تَعَالَىٰ:

﴿ قَالَ يَٰقَوۡمِ أَرَءَيۡتُمۡ إِن كُنتُ عَلَىٰ بَيِّنَةٖ مِّن رَّبِّي وَرَزَقَنِي مِنۡهُ رِزۡقٗا حَسَنٗاۚ وَمَآ أُرِيدُ أَنۡ أُخَالِفَكُمۡ إِلَىٰ مَآ أَنۡهَىٰكُمۡ عَنۡهُۚ إِنۡ أُرِيدُ إِلَّا ٱلۡإِصۡلَٰحَ مَا ٱسۡتَطَعۡتُۚ وَمَا تَوۡفِيقِيٓ إِلَّا بِٱللَّهِۚ عَلَيۡهِ تَوَكَّلۡتُ وَإِلَيۡهِ أُنِيبُ ٨٨ ﴾

صدق الله العظيم
سورة هود (88)

Dedication

To :

The soul of my parents..

Mybeautiful family ..

My teachers ..

Myfriends ...

Acknowledgements

I would like to express my appreciation to my supervisor:
Prof. Mohammed Elfadil Mohamed
and very grateful to the family of radiology college for the encouragement and moral support.

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Abstract

This descriptive cross-sectional study aim to assess the radiation dose and image quality in CT in order to explore the potential risk of radiation. This study was conducted on three hospitals during the period from 2015-2019 in Khartoum state .CT machines were used to collect data during this study the manufacture General Electric ,Toshiba and General Electric respectively Detector Type 16,16 and 64 respectively, where the Computed Tomography Dose Index(CTDI $_{vol}$) mGy, Dose Length Product(DLP)mGy.cm, Size specific dose estimate (SSDE)mGy and effective dose (in mSV) were measured for each hospital representing the scanning of head, chest and abdomen as well the image quality that include noise, signal, signal to noise ratio and contrast in high and low intensity areas in CT scan images for the previous parts before and after image enhancement. The main results of this study showed that the maximum $CTDI_{vol}$ was to the head 60.5mGy, the highest DLP was for the abdomen 2121.45mGy.cm, to SSDE was 61.71(mGy) for head as well as the effective dose which is equal to 5.65mSV.also the study showed that the image quality in the body areas of high intensity and low intensity before and after the image enhancement is associated with linear relation as the recession coefficients could be used to estimate the values of image enhancement before processing in order to find out the image quality ,and possibility of using the method to decide the best method of image enhancement. The study conclude that, The Effective dose is higher in hospital3 than other hospitals and SSDE dose is higher in hospital2 than other hospitals. And evaluated effective dose in every Exam type is higher brain than other exams, and the following linear equation can be used to estimate the validity of enhancement before the process which as signal after enhancement (High intensity area) = (1.31×

Signal) - 64.5, noise after enhancement (Low intensity area) = (0.824× noise) +0.278 and Signal to noise ratio after enhancement (HIA) = (1.07× SNR) -3.4. Contrast after enhancement = (0.853×contrast) +16.7.

This study recommended that CT scan examination for any patient must be justified, QC program are essential to evaluate the patients dose and machine accurate performance, Future studies can be done in order to optimize the radiation dose to establish national diagnostic reference level in Sudan and Application of QC program as a routine task in all x-ray department in order to detect possible future problems.

الخلاصة

هذه دراسة وصفية تحليلية هدفت الى تقويم الجرعة الإشعاعية وجودة الصورة في الأشعة المقطعية حتى يتم اكتشاف الخطر المحتمل للإشعاع . وحسب ماهو معروف في الدراسات أن هناك موازنة بين زيادة الجرعة الإشعاعية لتحسين الصورة وتقليل الجرعة للمريض .

إجراء هذه الدراسة في ثلاثة مستشفيات في ولاية الخرطوم خلال الفترة مابين 2015- 2019 تم قياس مؤشر الجرعة للأشعة المقطعية ($CTDI_{,l}$ (mGy)، الجرعة الكلية الناتجة DLP(mGy.cm)، وتقديرالجرعة الخاصة بالحجم SSDE (mGy) الجرعة الفعالة ED (mSv) ، لكل مستشفى وتم قياسها في الجسم في منطقة الرأس والبطن والصدر بالإضافة الى جودة الصورة بما في ذلك الضوضاء والإشارة ونسبة الضوضاء الى الإشارة والتباين في مناطق الجسم ذات الكثافة العالية والمنخفضة في صورة الأشعة المقطعية قبل تحسين الصورة وبعد التحسين.

أظهرت الدراسة بعض النتائج أهمها ان قيمة مؤشر الجرعة للأشعة المقطعية ($CTDI_{vol}$)بالنسبة للرأس هي(60.5(mGy وأن أعلى قيمة للجرعة الكلية الناتجة (DLP) كانت للبطن 2121.45(mGy.cm) وكانت (mGy) 61.71بالنسبة لتقدير الجرعة الخاصة بالحجم (SSDE) للرأس أيضا والجرعة الفعالة (ED) للرأس كانت(mSV) 5.51 . أوضحت الدراسة أنقيم جودة الصورة في مناطق الجسم ذات الكثافة المنخفضة والكثافة العالية قبل تحسين الصورة وبعد التحسين ترتبط بعلاقة خطية حيث أن معاملات الإنحسار يمكن إستخدامها لتقدير قيم تحسين الصورة قبل المعالجة لإكتشاف الجودة وإمكانية إستخدام هذة الطريقة من أجل إتخاذ القرار الأفضل للتحسين.

Chapter one
Introduction

1.1 Introduction

A CT scan makes use of computer-processed combinations of many X-ray images taken from different angles to produce cross-sectional (tomographic) images (virtual "slices") of specific areas of a scanned object, allowing the user to see inside the object without cutting. Other terms include computed axial tomography (CAT scan) and computer aided tomography.

Digital geometry processing is used to generate a three-dimensional image of the inside of the object from a large series of two-dimensional radiographic images taken around a single axis of rotation(Wikipedia, 2017) Medical imaging is the most common application of X-ray CT. Its cross-sectional images are used for diagnostic and therapeutic purposes in various medical disciplines (Wikipedia, 2017) The rest of this article discusses medical-imaging X-ray CT; industrial applications of X-ray CT are discussed at industrial computed tomography scanning.

The term "computed tomography" (CT) is often used to refer to X-ray CT, because it is the most commonly known form. But, many other types of CT exist, such as positron emission tomography (PET) and single-photon emission computed tomography (SPECT). X-ray tomography is one form of radiography, along with many other forms of tomographic and non-tomographic radiography.

CT produces a volume of data that can be manipulated in order to demonstrate various bodily structures based on their ability to block the X-ray beam. Although, historically, the images generated were in the axial or transverse plane, perpendicular to the long axis of the body, modern scanners allow this volume of data to be reformatted in various planes or even as volumetric (3D)

representations of structures. Although most common in medicine, CT is also used in other fields, such as nondestructive materials testing. Another example is archaeological uses such as imaging the contents of sarcophagi. Individuals responsible for performing CT exams are called radiographers or radiologic technologists(Wikipedia, 2017)Use of CT has increased dramatically over the last two decades in many countries(Wikipedia, 2017)An estimated 72 million scans were performed in the United States in 2007. One study estimated that as many as 0.4% of current cancers in the United States are due to CTs performed in the past and that this may increase to as high as 1.5 to 2% with 2007 rates of CT use however, this estimate is disputed, as there is not a consensus about the existence of damage from low levels of radiation. Side effects from intravenous contrast used in some types of studies include kidney problems. (Wikipedia, 2017)

1-2Medical use of CT

CT has become an important tool in medical imaging to supplement X-rays and medical ultrasonography. It has more recently been used for preventive medicine or screening for disease, for example CT colonography for people with a high risk of colon cancer, or full-motion heart scans for people with high risk of heart disease. A number of institutions offer full-body scans for the general population although this practice goes against the advice and official position of many professional organizations in the field primarily due to the radiation dose applied. (Wikipedia, 2017)

Figure (1.1)Modern CT scanner

1. 3-Radiation dose units

The radiation dose reported in the gray or mGy unit is proportional to the amount of energy that the irradiated body part is expected to absorb, and the physical effect (such as DNA double strand breaks) on the cells' chemical bonds by X-ray radiation is proportional to that energy(Wikipedia, 2017).

The sievert unit is used in the report of the effective dose. The sievert unit, in the context of CT scans, does not correspond to the actual radiation dose that the scanned body part absorbs but to another radiation dose of another scenario, the whole body absorbing the other radiation dose and the other radiation dose being of a magnitude, estimated to have the same probability to induce cancer as the CT scan.(Wikipedia, 2017)Thus, as is shown in the table above, the actual radiation that is absorbed by a scanned body part is often much larger than the effective dose suggests. A specific measure, termed the computed tomography dose index (CTDI), is commonly used as an estimate of the radiation absorbed dose for tissue within the scan region, and is automatically computed by medical CT scanners.(Wikipedia,2017)

The equivalent dose is the effective dose of a case, in which the whole body would actually absorb the same radiation dose, and the Sievert unit is used in its report. In the case of non-uniform radiation, or radiation given to only part

of the body, which is common for CT examinations, using the local equivalent dose alone would overstate the biological risks to the entire organism.

1.4- CT Image Quality

Fundamentally, image quality in CT, as in all medical imaging, depends on many factors the four basic one includes: image contrast, spatial resolution, image noise, and artifacts. Depending on the diagnostic task, these factors interact to determine sensitivity (the ability to perceive low-contrast structures) and the visibility of details.

1.4-1CT Image Contrast

CT image contrast depends on subject contrast and display contrast. Because CT display contrast is arbitrary (depending only on the window level and width selected), it will not be discussed further.

As in radiography, CT subject contrast is determined by differential attenuation: that is, differences in x-ray attenuation by absorption or scattering in different types of tissue and thus resulting in differences in the intensity of the x-rays ultimately reaching the detectors. Because of the high peak kilovoage and relatively high beam filtration (beam hardness) used in CT, the x-ray/tissue interactions (except in bone) are overwhelmingly Compton-scattering events. Differential attenuation for Compton scatter arises from differences in tissue electron density (electrons/cm3), which in turn are due primarily to differences in physical density (Seibert JAetal,2005). Thus, subject soft-tissue contrast in CT comes mainly from differences in physical density. That the small differences in soft-tissue density can be visualized on CT is due to the nature of the image (a 2-dimensional image of a 2-dimensional slice), the ability to map small attenuation differences to large

differences in gray level by windowing, the near-complete elimination of scatter, and the use of a sufficient x-ray intensity.

Related to CT image contrast is the CT contrast scale. We recall that CT numbers are derived from voxel attenuation coefficients calculated during image reconstruction using the following relationship:

Whereμp and μw are linear attenuation coefficients for a given voxel and for water (μw is determined from calibration scans). Because CT number is a linear function of μp, a graph of expected CT numbers for materials with known attenuation coefficients should be linear over the clinical CT-number range (e.g., −1,000 to +1,000). For evaluation of the contrast scale of a scanner, various CT test phantoms are available that contain materials designed to provide certain CT numbers (e.g., the CT numbers for water, fat, soft tissue, bone, and air) (McColloughCHetal, 2000)

1.4-2 Image Noise

If a graphic cursor is used to display pixel CT numbers in an image of a uniform phantom (e.g., a phantom containing all water), it is seen that the CT numbers are not uniform but rather fluctuate around an average value (which should be approximately 0 for water): Some pixels are 0, some are +1, some +2, some −1, and so forth. These random fluctuations in the CT number of otherwise uniform materials appear as graininess on CT images. This graininess is the CT analog of—and is of the same nature as—radiographic quantum mottle: It is due to the use of a limited number of photons to form the image (Hounsfield GN,1976- Judy PF,2000).

In radiography, image noise is related to the numbers of x-ray photons contributing to each small area of the image (e.g., to each pixel of a direct digital radiograph). In CT, x-rays contribute to detector measurements and not

to individual pixels. CT image noise is thus associated with the number of x-rays contributing to each detector measurement. To understand how CT technique affects noise, one should imagine how each factor in the technique affects the number of detected x-rays. Examples are as follows:

X-ray tube amperage: Changing the mA value changes the beam intensity—and thus the number of x-rays—proportionally. For example, doubling the mA value will double the beam intensity and the number of x-rays detected by each measurement.

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.

Slice thickness: Changing the thickness changes the beam width entering each detector—and thus the number of detected x-rays—approximately proportionally. For example, compared with a slice thickness of 5 mm, a thickness of 10 mm approximately doubles the number of x-rays entering each detector.

Peak kilovolt age: Increasing the peak kilovolt age increases the number of x-rays penetrating the patient and reaching the detectors. Thus, increasing the kilovolt age reduces image noise but can (slightly) reduce subject contrast as well.

Although not affecting the numbers of detected x-rays, a reconstruction filter profoundly affects the appearance of noise in the image: Smooth filters blur the noise, reducing its visual impact, whereas sharp filters enhance the noise. In images of soft tissue, noise is generally more interfering than blur, and smoother filters are preferred. In images of structures with edges and small details, such as bone, blur is generally more interfering than noise, and sharper

filters are preferred. For comparison, Figure 10 shows examples of noise in scans of uniform phantoms using standard and higher-resolution (bone) filters and with standard and very low values for mAs.

Comparison of noise from scans using 270 mAs (typical clinical value) and 100 mAs. (C) Appearance of image noise is strongly affected by reconstruction filter; sharp filter such as bone also sharpens (enhances) appearance of noise.

Because CT noise appears as fluctuations in CT numbers, a measurement of image noise is a measurement of these fluctuations, and such a measurement can be made using regions of interest (ROIs) on a scan of a uniform phantom. A statistical ROI function (available on most CT scanners) allows users to place a rectangular or oval ROI on the image, within which is calculated the average and standard deviation (SD) of the CT numbers for the enclosed pixels. 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.

Because noise is the most bothersome when one is viewing low-contrast soft-tissue structures, an important test of scanner performance is how well low-contrast test objects are seen in the presence of typical noise levels. Figure 11 is an example of a low-contrast test phantom, consisting of groups of rods embedded in material producing approximately 0.6% subject contrast (i.e., a nominal CT-number difference of 6 between the rods and the background). The rod groups range in diameter from 6 to 2 mm. In this example, the 5-mm rods are visible, whereas the smaller ones are lost in the noise.

Low-contrast phantom to test CT performance in presence of typical image noise levels (from American College of Radiology accreditation phantom). Five-millimeter rods are visible, but smaller ones are obscured by noise.

Signal-to-noise ratio (abbreviated SNR or S/N) is a measure used in science and engineering that compares the level of a desired signal to the level of background noise. It is defined as the ratio of signal power to the noise power, often expressed in decibels. A ratio higher than 1:1 (greater than 0 dB) indicates more signal than noise. While SNR is commonly quoted for electrical signals, it can be applied to any form of signal (such as isotope levels in an ice core or biochemical signaling between cells).

The signal-to-noise ratio, the bandwidth, and the channel capacity of a communication channel are connected by the Shannon–Hartley theorem.

1.4-4Signal-to-noise ratio

Is sometimes used metaphorically to refer to the ratio of useful information to false or irrelevant data in a conversation or exchange. For example, in online discussion forums and other online communities, off-topic posts and spam are regarded as "noise" that interferes with the "signal" of appropriate discussion.

1.5-Problem of The study

Radiation like X-rays might lead to serious health problem to the practitioners and patient as well if certain measures were not taken seriously, which means potentially the risk exist, therefore it is essential land in fact mandatory to reduce the radiation dose in diagnostic radiology as far as possible at the same time produces a good quality image that reveal the health condition from the first exposure.

1.6-Objectives

The general objective of this study was to assess radiation dose and image quality in CT in order to explore the potential risk and image suitability.

Specific objective

- To estimate patient effective dose in three hospital
- To measure CTDvol, DLP and SSDE values for head, abdomen and chest
- To find signal, noise and signal to noise ratio in high and low intensity areas before and after image enhancement.
- To assess the image contrast before and after enhancement
- To find the significance differences between the image quality factor before and after enhancement.
- To estimate the quality of image enhancement form current quality factor.

1.7-Significance of the study

This study will make it possible to obtain an optimize image ideal for diagnosis of the disease as well as applying the lowest dose to the patient in order to protect the patient while getting the best image quality.

1.8-Overview of the study

The study consisted of five chapters; with Chapter one is an introduction which includes definition of CT scans, medical use, Radiation dose units and Image Quality parameters, as well as the problem of the study, an objective and significant of the study. Similarly Chapter two includes a comprehensive literatures review. Chapter three includes the methodology of the study which discussed the material used to collect the data and in the method followed in data acquisition. Chapter four which presented the results and Finally Chapter five which included discussion, conclusion and recommendations of the study.

Chapter Two
Background and Literature Review

2.1 CT Machine

2.1.1The CT scanner components

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

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

The Gantry which is a central opening gantry is a moveable frame that contains the x-ray tube including collimators and filters, detectors, data acquisition system (DAS), rotational components including slip ring systems and all associated electronics such as gantry angulations motors and positioning laser lights. A CT gantry can be angled up to 30 degrees toward a forward or backward position.

The Table is where the patients is positioned (lie down), and it moves through the gantry. The patient's table and the gantry constitute CT scanner itself.

The power Distribution unit supplies power to the gantry, the patient's table and the computers of the Computing System, which is localized in a separate room as will be explained next.

The computing System(or operator's console) is installed in separate room, making it possible for the operator (technician)to control the acquisition process, introducing patient data and selecting several acquisition parameters such as the kVp , mA values the protocol is going to use (Nunes, 2010). Also there is another operator's console for editing and post-processing is also

necessary, so it possible to analyze and review previous exam data, without interfering with the current examinations taking place.

Figure (2.1) CT scanner

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

2.2 CT Generations

2.2.1 First-Generation CT Scanners

The EMI Mark I scanner, the first commercial scanner invented by Hounsfield,was introduced in 1973 (Hounsfield GN., 1973). This scanner acquired datawith an x-ray beam collimated to a narrow “pencil” beam directed to a singledetector on the other side of the patient; the detector and the beam were alignedin a scanning frame. A single projection was acquired by moving the tube anddetector in a straight-line motion (translation) on opposite sides of the patient(Fig 2.2). To acquire the next projection, the frame rotated 1°, then translated inthe other direction. This process of translation and rotation was repeated until180 projections were obtained. The

earliest versions required about 4.5 minutesfor a single scan and thus were restricted to regions where patient motion could be controlled (the head). Since procedures consisted of a series of scans,procedure time was reduced somewhat by using two detectors so that two parallel sections were acquired in one scan. Although the contrast resolution ofinternal structures was unprecedented, images had poor spatial resolution (onthe order of 3 mm for a field of view of 25 cm and 80 × 80 matrix) and verypoor z-axis resolution (13-mm section thickness).

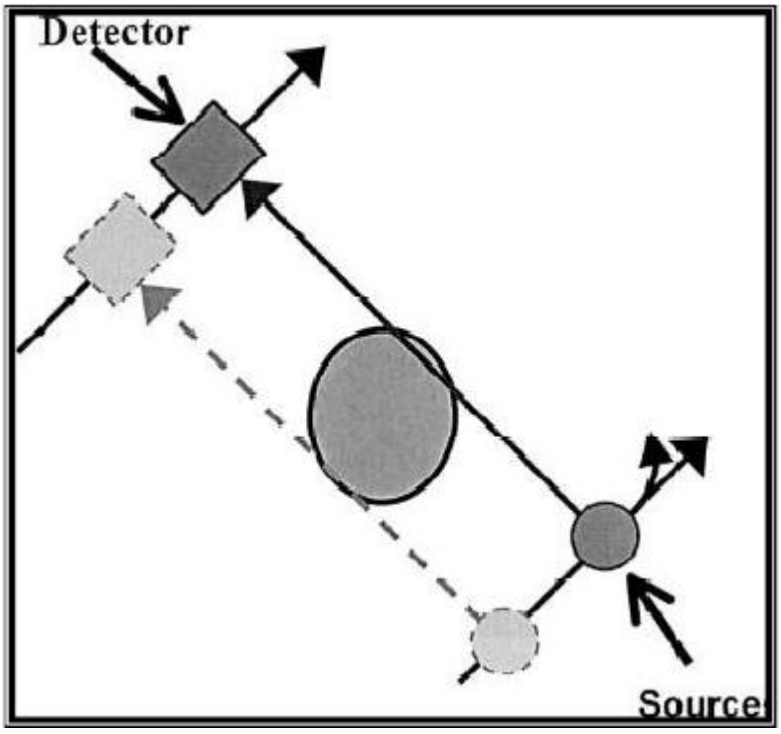
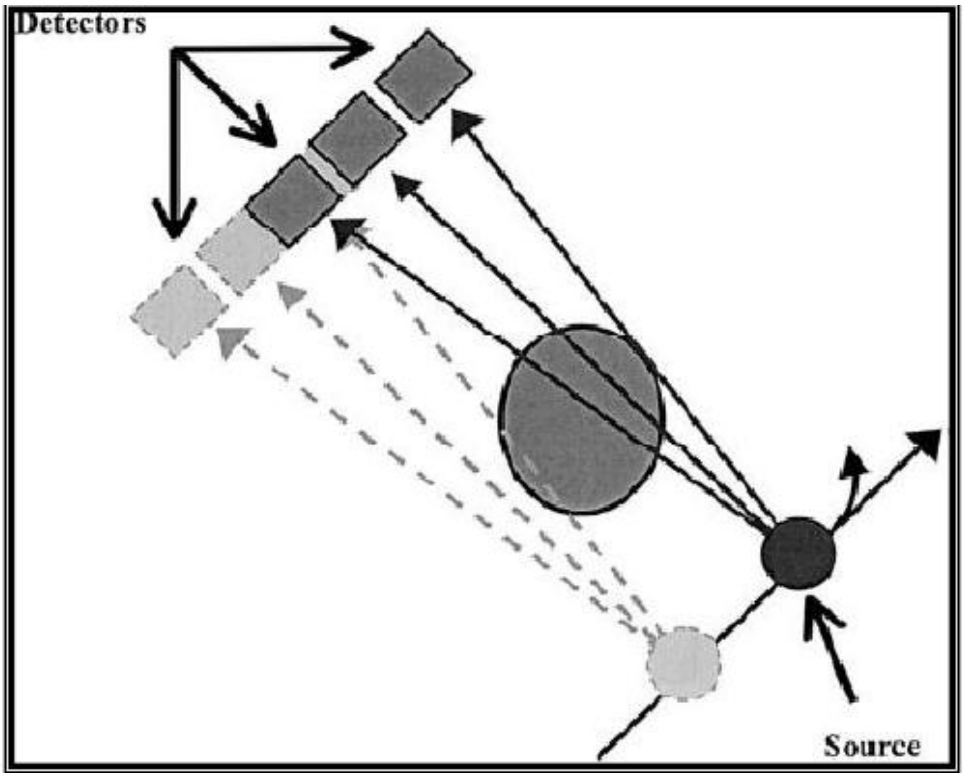


Fig (2.2): Diagram of the first-generation CT scanner, which used aparallel x-ray beam with translate-rotate motion to acquire data. From (Mahesh,2002).

2.2.2 Second-Generation CT Scanners

The main impetus for improvement was in reducing scan time ultimately to the point that regions in the trunk could be imaged. By adding detectors angularly displaced, several projections could be obtained in a single

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



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

2.2.3 Third-Generation CT Scanners

Designers realized that if a pure rotational scanning motion could be used, thenit would be possible to use higher-power, rotating anode x-ray tubes and thusimproves scan speeds in thicker body parts. One of the first designs to do so wasthe so-called third generation or rotate-rotate geometry. In these scanners, the x-raytube is collimated to a wide, fan-shaped x-ray beam and directed toward anarc-shaped row of detectors. During scanning, the tube and detector array rotatearound the patient (Fig 2.4), and different projections are obtained during rotation by pulsing the x-ray source or by sampling the detectors at a very high rate. The number of detectors varied from 300 in early versions to over 700 in modern scanners. Since the slam-bang translational motion was replaced withsmooth rotational motion, higher-output rotating

anode x-ray tubes could beused, greatly reducing scan times. One aspect of this geometry is that rays in asingle projection are divergent rather than parallel to each other, as in earlierdesigns. Beam divergence required some modification of reconstructionalgorithms, and sampling considerations required scanning an additional arc ofone fan angle beyond 180°, although most scanners rotate 360° for each scan.

Nearly all current helical scanners are based on modifications of rotate-rotatedesigns. Typical scan times are on the order of a few seconds or less, and recentversions are capable of sub second scan times.

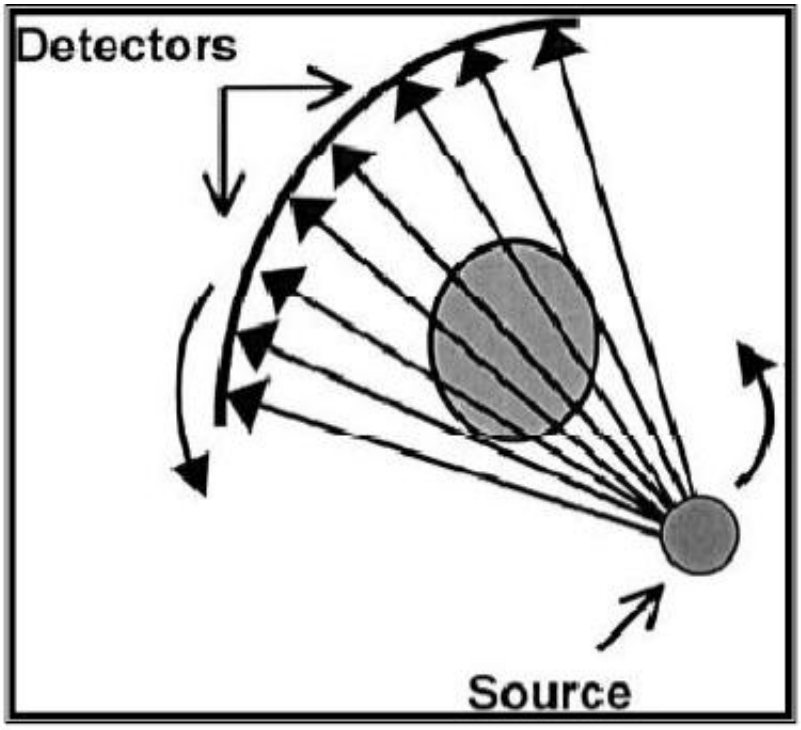


Fig (2.4): Diagram of the third-generation CT scanner, which acquires data by rotating both the x-ray source with wide fan beam geometry and the detectors around the patient. Hence, the geometry is called rotate-rotate motion.

2.2.4: Fourth-Generation CT Scanners:

This design evolved nearly simultaneously with third-generation scannersand also eliminated translate-rotate motion. In this case, only the source rotateswithin a stationary ring of detectors (Fig 2.5). The x-ray tube is positioned torotate about the patient within the space between the patient and the detectorring. One clever version, which is no longer produced, moved the x-ray tube outof the detector ring and tilted the ring out of the x-ray beam in a wobbling(nutation) motion as the tube rotated. This design permitted a smaller detectorring with fewer detectors for a similar level of performance. Early fourth generationscanners had some 600 detectors and later versions had up to 4,800.Within the same period, scan times of fourth-generation designs werecomparable with those of third-generation scanners. One limitation of fourth generation;designs is less efficient use of detectors, since less than one-fourthare used at any point during scanning. These scanners are also more susceptibleto scatter artifacts than third-generation types, since they cannot use anti scattercollimators. CT scanners of this design are no longer commercially availableexcept for special-purpose applications.(Mahesh, 2002)Until around 1990, CT technology had evolved to deliver scan plane resolutionsof 1–2 lp/mm, but z-axis resolution remained poor and interscan delay wasproblematic due to the stop-start action necessary for table translation and forcable unwinding, which resulted in longer examination times. The z-axisresolution was limited by the choice of section thickness, which ranged from 1to 10 mm. For thicker sections, the partial volume averaging between different issues led to partial volume artifacts. These artifacts were reduced to some extent by scanning thinner sections. In addition, even though it was possible toobtain 3D images by stacking thin sections, inaccuracy dominated due toinvoluntary motion from scan to scan. A

typical 3D reconstruction of this era isshown in Figure 2.7 the step like contours could be minimized by overlappingof CT sections at the expense of a significant increase in radiation to the patient.Also, the conventional method of section-by-section acquisition producedMisregistration of lesions between sections due to involuntary motion ofanatomy in subsequent breath holds between scans. It was soon realized that ifmultiple sections could be acquired in a single breath hold, a considerableimprovement in the ability to image structures in regions susceptible tophysiologic motion could result. However, this required some technologicaladvances, which led to the development of helical CT scanners. (Mahesh, 2002)

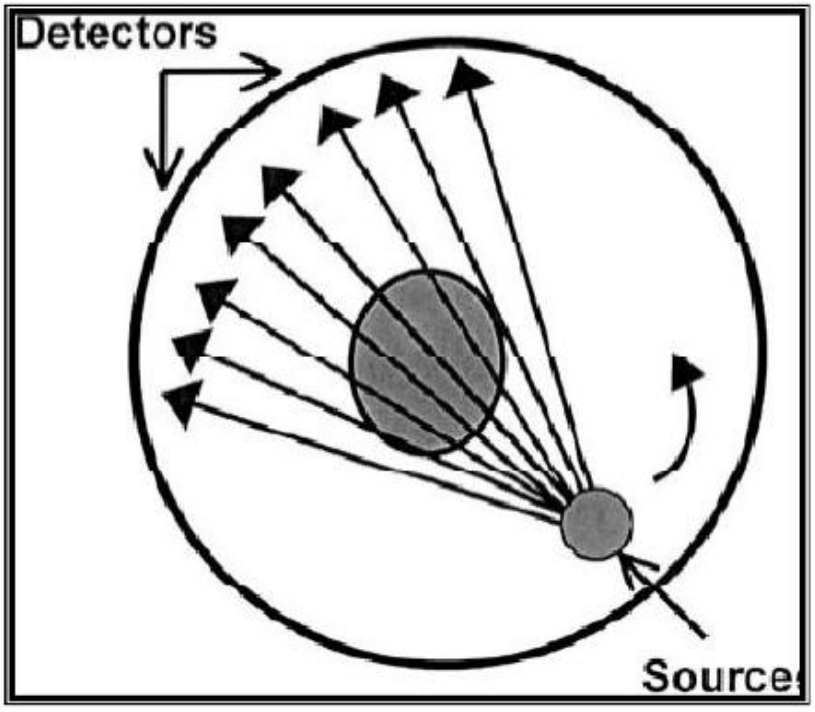


Fig (2.5): Diagram of the fourth-generation CT scanner, which uses a Stationary ring of detectors positioned around the patient. Only the x-ray source Rotates with wide fan beam geometry, while the detectors are stationary. Hence,the geometry is called rotate-stationary motion. From (Mahesh, 2002)

2.3 Principles of Helical CT Scanners

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

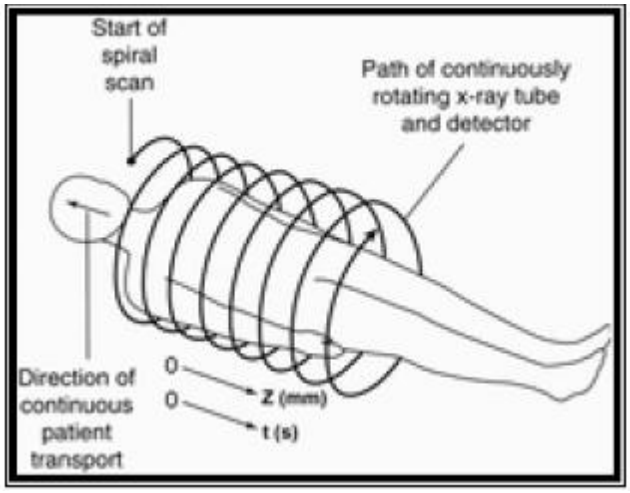


Fig (2.6): Principles of helical CT. As the patient is transported throughthe gantry, the x-ray tube traces a spiral or helical path around the patient,acquiring data as it rotates. t = time in seconds.(Mahesh, 2002).

2.4 Slip-Ring Technology

Slip rings are electromechanical devices consisting of circular electricalconductive rings and brushes that transmit electrical energy across a movinginterface. All power and control signals from the stationary parts of the scannersystem are communicated to the rotating frame through the slip

ring. The slip ringdesign consists of sets of parallel conductive rings concentric to the gantry axis that connect to the tube, detectors, and control circuits by sliding contactors(Fig 2.7).

These sliding contactors allow the scan frame to rotate continuouslywith no need to stop between rotations to rewind system cables (Brunnett, et al.1994). This engineering advancement resulted initially from a desire to reduceinterscan delay and improve throughput. However, reduced interscan delayincreased the thermal demands on the x-ray tube; hence, tubes with much higherthermal capacities were required to withstand continuous operation overmultiple rotations. (Mahesh, 2002)

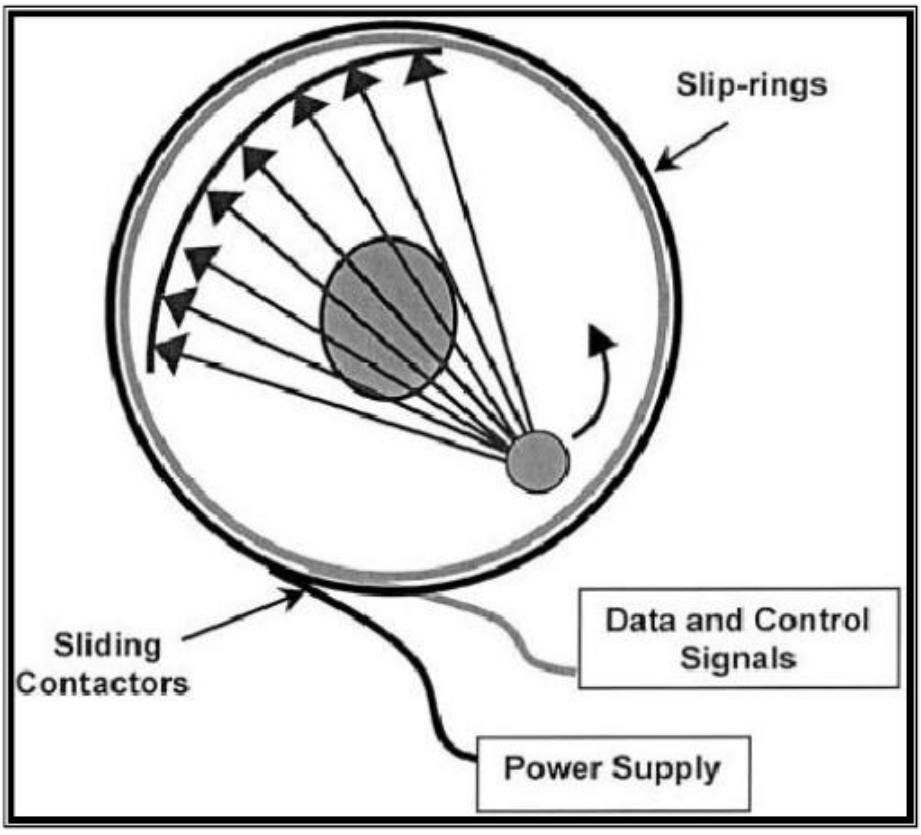


Fig (2.7): Diagram of the slip-ring configuration. Sliding contactors permit continuous rotation of the x-ray tube and detectors while maintaining electrical contact with stationary components.

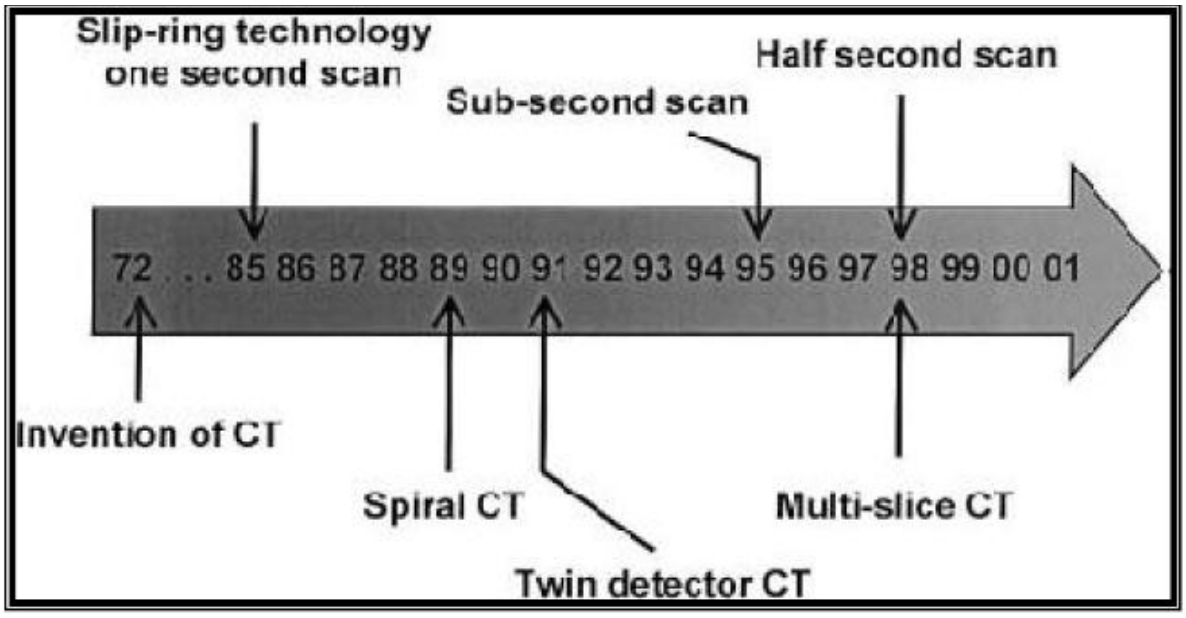


Fig (2.8): Time line of the key technological developments in CT. From (Mahesh,2002).

2.5 Capabilities of Single-Row Detector Helical CT

With the advent of helical CT, considerable progress was made on the roadtoward 3D radiography. An example of a 3D reconstruction from single-rowdetector helical scanning is shown in Fig (2.9).Complete organs could bescanned in about 30–40 seconds; artifacts due to patient motion and tissuemisregistration due to involuntary motion were virtually eliminated. It becamepossible to generate sections in any arbitrary plane through the scanned volume.Significant improvements in z-axis resolution were achieved due to improvedsampling, since sections could be reconstructed at fine intervals less than thesection width along the z axis. Near-isotropic resolution could be obtained withthe thinnest (□1 mm) section widths at a pitch of 1, but this could be done onlyover relatively short lengths due to tube and breath-hold limitations (Kalender1995), (Levy, 1995). Higher-power tubes capable of longer continuousoperation coupled with faster rotation speeds could scan

greater lengths withhigher resolution. The practical limit on such brute force approaches, however,became the length of time a sick patient could reliably suspend breathing. Thisturns out to be no more than 30 seconds. Even though the z-axis resolution forhelical CT images far exceeds that of conventional CT images, the type of interpolation algorithm and the pitch still affect the overall image quality.

The section sensitivity profiles of helical CT images are different compared withthose of conventional CT images, which are influenced by the type of interpolation algorithm and the selected pitch.

2.6 Multiple-Row Detector Helical CT

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

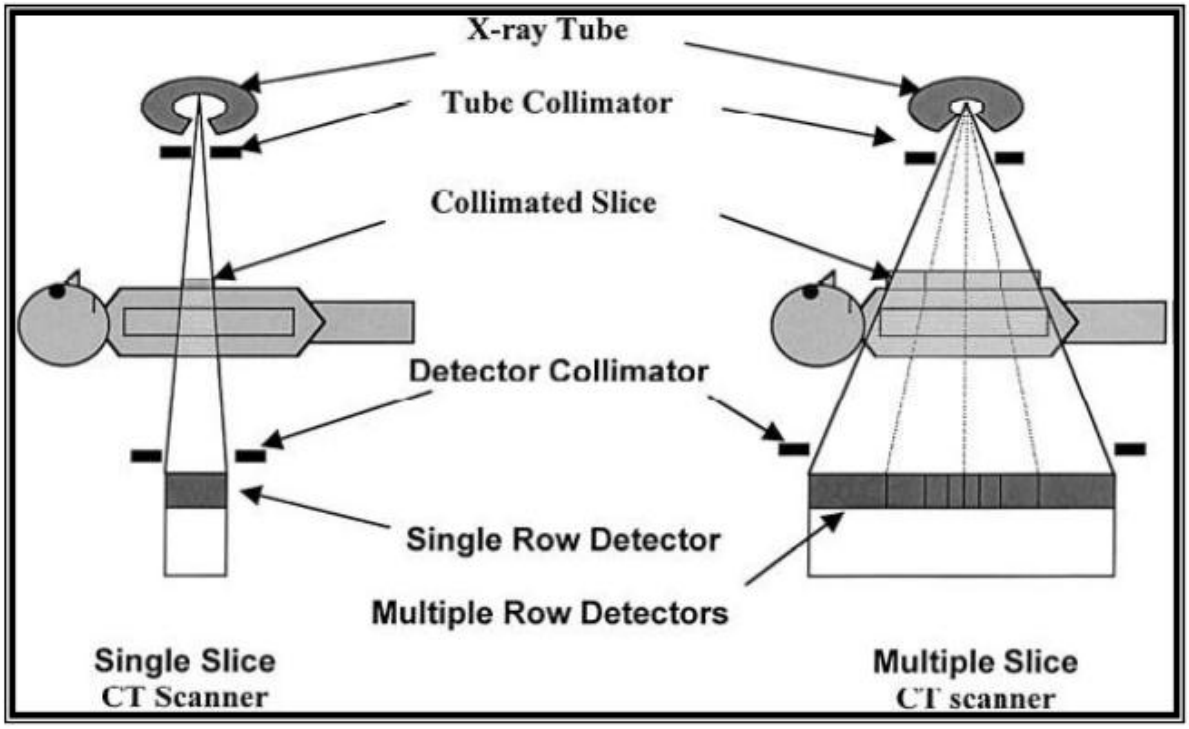


Fig (2.9): Diagram shows the difference between single-row detector andmultiple-row detector CT designs.

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

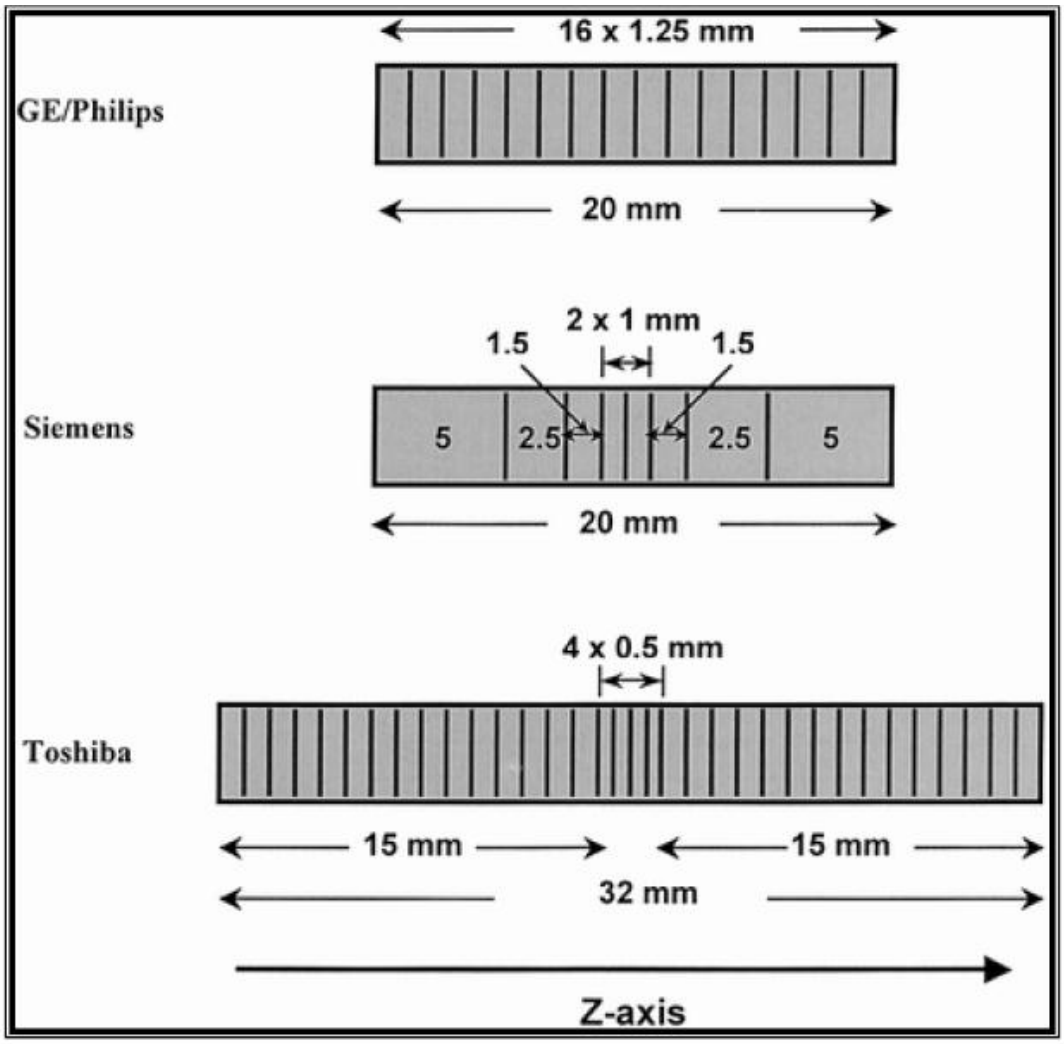


Fig (2-10): Various detector array designs used in multiple-row detector CT Scanners

In single-row detector helical CT designs, scan volume can be increased with an increased pitch at the expense of poorer z-axis resolution, whereas z-axisresolution can be preserved in multiple-row detector designs. For example, ifa 10-mm collimation were divided into four 2.5-mm detectors, the same scan length could be obtained in the same time but with a z-axis resolutionimproved from 10 mm to 2.5 mm. In another example, a multiple-rowdetector scanner with four 5-mm detectors and a beam width of 20

mmreduces the scan time by a factor of 4–15 seconds for the same z-axisresolution (Mahesh, 2002). By increasing the number of CT scanner detectorrows, data acquisition capability dramatically increases while greatlyimproving the efficiency of x-ray tubes. Further developments in scannerrotational speeds and tube outputs have made isotropic resolution a practicalpossibility with even better improvements on the horizon. Current multiple rowdetector scanners can scan large 40-cm volume lengths in less than 30seconds with near-isotropic resolution and image quality that could not be envisioned at the time of Hounsfield's invention.MDCT systems are CT scanners with a detector array consisting of morethan a single row of detectors. The "multi-detector-row" nature of MDCTscanners refers to the use of multiple detector arrays (rows) in thelongitudinal direction (that is, along the length of the patient lying on the patient table) MDCT scanners utilize third generation CT geometry inwhich the arc of detectors and the x-ray tube rotate together. All MDCTscanners use a slip-ring gantry, allowing helical acquisition at rotation

speeds as fast as 0.33 second for a full rotation of 360 degrees of the X-raytube around the patient. A scanner with two rows of detectors (Mahesh,2002) had already been on the market since 1992 and MDCT scanners withfour detector rows were introduced in 1998 by several manufacturers. The primary advantage of these scanners is the ability to scan more than one slice simultaneously and hence more efficiently use the radiation delivered fromthe X-ray tube (Fig.2.11). The time required to scan a certain volume couldthus be reduced considerably.The number of slices, or data channels, acquired per axial rotation continuesto increase, with 64-detector systems now common (Flohr et al., 2005a ;Flohr et al., 2005b). It is likely that in the coming years even larger arrays ofdetectors having

longitudinal coverage per rotation > 4 cm will be commercially available. Preliminary results from a 256-detector scanner(12.8 cm longitudinal coverage at the center of rotation) have already beenpublished (Mori et al., 2004). Further, an MDCT system with two x-raysources is now commercially available, signaling continued evolution of CTtechnology and applications (Flohr et al., 2006).

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

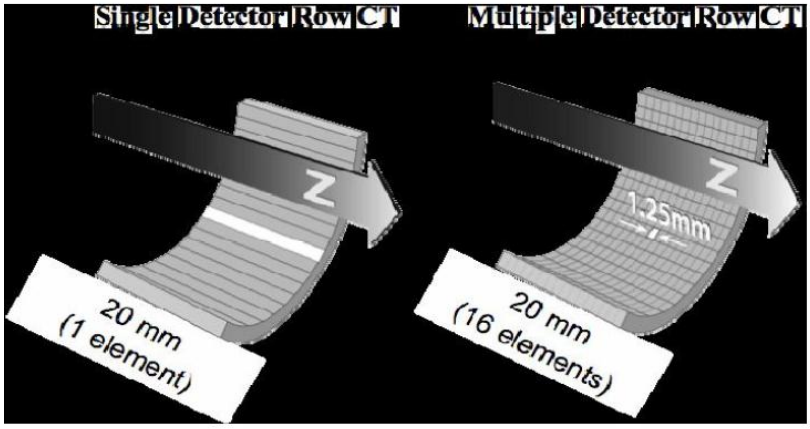


Fig (2.11): single CT detector versus Multi slice CT detector. From (ICRP 32/219,2006).

2.7 CT imaging protocol

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

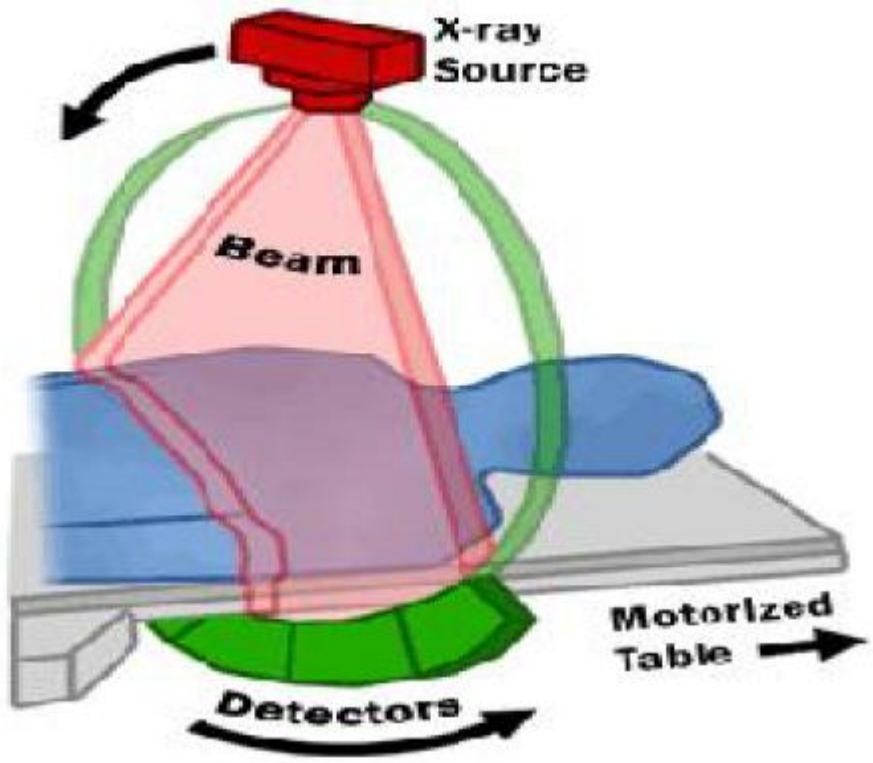


Figure (2.12) Simple overview of a third generation CT-imaging system

2.7.1 Parameters

In order to properly calculate and compare doses, it is imperative to have astandardized nomenclature to ensure that all data is comparative (Kalra, M. Ket al 2006). Without this, it will be difficult to reproduce measurements, and todevelop consistent protocols. When performing a CT examination, a number ofparameters are defined by the operator. The thesis will cover the

parametersdeemed important for correct, uniform dosimetery: tube current, tube voltage,

rotation time, total scan length, slice thickness and pitch. Automatic exposurecontrol (AEC) and iterative reconstruction will be briefly covered, as theirimpact on dose and image quality is more of a qualitative influence than a quantitative one.

2.7.1.1 Tube current

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

2.7.1.2 Tube Voltage

The tube voltage [kV] determines the voltage across the anode and cathode ofthe X-ray tube, and therefore the acceleration of the electrodes across theinterior vacuum. This determines the kinetic energy of the electrodes when theyreach the anode, and therefore the number of interactions they can initiatebefore being absorbed. As a consequence, an increase in tube voltage willincrease the dose, all other factors kept constant; however, the increase is notdirectly proportional as was the case with current. Voltage determines theenergy of the electrons, and therefore the energy distribution of the incident X-rays.

It is rarely adjusted from the customary value of 120 kV. Certainexaminations use a different voltage, but seldom outside the range of 80 to 140 kV (Kalra, M. K ,et al 2006).

2.7.1.3 Rotation Time

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

2.7.1.4 Total Scan Length

It is apparent that the total scan length [cm] influence the absorbed dose, as anincrease 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 thediagnostically relevant part of the patient; otherwise, an unnecessary increase indose will be seen (ICRP, 2000). This is relatively easy with SSCT; however, thesituation is more complicated for MSCT. At the initiation of the scan, the X-raytube will be activated the moment the first row of detectors reach the diagnosticarea. The X-ray beam will irradiate the entire detector-array, but only the firstrow of detectors will be acquiring image data. The remaining detector rows willnot acquire data, but the area will still be irradiated. This is called over scan, anda small degree of over scan is required for correct reconstruction. As the tablemoves, more rows of detectors are entering the diagnostic area, contributing tothe image. At the reverse end of the patient, the same scenario occurs, and anoteworthy part of the dose is absorbed in the patient outside the diagnostic area

(M. K., Maher et al 2004).

2.7.1.5 Slice Thickness

In SSCT, with only a single row of detectors, the slice thickness [cm] isdetermined by simple collimation. The maximum slice thickness is limited bythe width of the individual detector element (typically 10 mm (M. K.,

Maher,etal 2004)), and by collimating the beam, this thickness can be decreased. In otherwords, the width of the beam is equal to slice thickness. In MSCT, the width of each individual detector element in the longitudinal direction determines theminimum slice thickness, and by merging multiple adjacent detector elementsduring detection, one can increase the slice thickness.

This has a significantimpact on image quality, as thin slices have better spatial resolution comparedto thick slices, but lower SNR. To address the decrease in SNR, it is necessaryto increase for instance the tube current, resulting in a significant increase indose to the patient (Kalender,et al,2005).

2.7.1.6 Pitch

With the prevalence of helical MSCT, it is necessary to incorporate theincremental movement of the table, in relation to the irradiated area. This isdefined as pitch, being the increment of the table per rotation, divided by thewidth of the beam. In Figure 2.13 below, a 4-slice MSCT is rotated twicearound 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 ismoving during exposure.

pitch = table feed per rotation/collimation

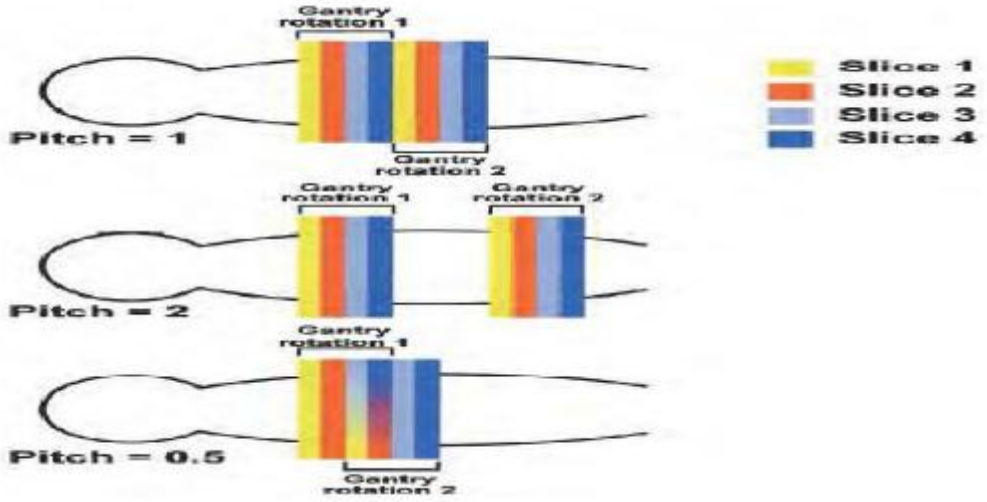


Figure (2.13) The effect of pitch on irradiated area, with a overlap for pitch < 1

2.7.1.7 Automatic Exposure Control

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

2.7.2 Protocols

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

2.8 CT Dose and unit

2.8.1 Radiation dose units

The specific units of measurement for radiation dose commonly referred to as effective dose (mSv). Other radiation dose measurement units include; Rad ,Rem, Rontgen and Sievert. Because different tissues and organs have varying in sensitivity to radiation exposure, the actual effective dose to different parts of the body for X-ray procedure varies. The term effective dose is used when referring to the dose averaged over the entire body. The effective dose accounts for the relative sensitivities of different tissues exposed. More importantly, it allows for qualification of risk and comparison to more familiar sources of exposure that range from natural background radiation to radiographic medical procedure. As with other medical procedures, X-rays are safe when used with care. Radiologists and X-ray technologists have been trained to use the minimum amount of radiation that is necessary to obtain the needed results. The decision to have an X-ray examination is a medical one, based on the like lihood of benefit from the examination and the potential risk from radiation (ICRP1990, ICRP 1991).

2.8.2 Effective dose

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

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

Effective Dose (Gy) = Absorbed Dose (Gy) x WT (2.1)

If more than one area has been exposed, then the total body effective dose is justthe sum of the effective doses for each exposed area. It is a simple as that.

Now let's see why effective dose is such a useful quantity. There is often a need to compare the amount of radiation received by patients for different types of X-ray procedures, for example, a chest radiograph and a CT scan. The effective dose is the most appropriate quantity for doing this. Also, by using effective dose it is possible to put the radiation received from diagnostic procedures into perspective with other exposures, especially natural background radiation (PerrySprawls.org, Online).

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

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

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

2.9.1 Computed tomography dose index (CTDI)

The ‘Computed Tomography Dose Index (CTDI)’ is the fundamental CT dose descriptor. By making use of this quantity, the first two peculiarities of CT scanning are taken into account: The CTDI (unit: Milligray (mGy)) is derived from the dose distribution a long a line which is parallel to the axis of rotation for the scanner (=z axis) and which is recorded for a single rotation of X-ray source. (Fig.2.14 the) illustrates meaning of the term: CTDI is the equivalent

ofthe dose value inside the irradiated slice (beam) that would result if theabsorbed radiation dose profile were entirely concentrated to a rectangular ofwidth equal to the nominal beam width with N being the number of independent(i.e. non-overlapping) slices that are acquired simultaneously. Accordingly, alldose contributions from outside the nominal beam width, i.e. the areas under thetails of the dose profile, are added to the area inside the slice (Nagel 2007).

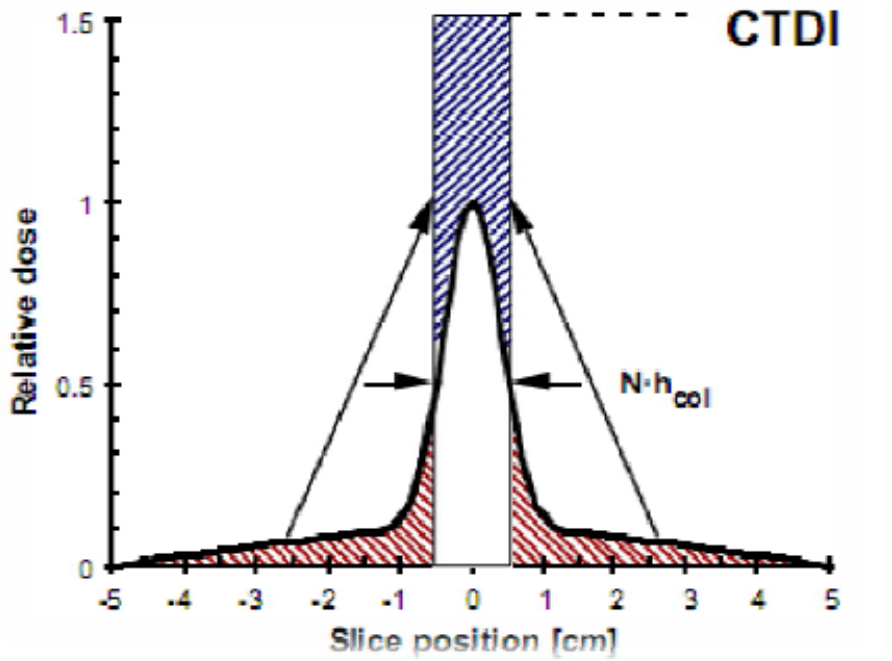
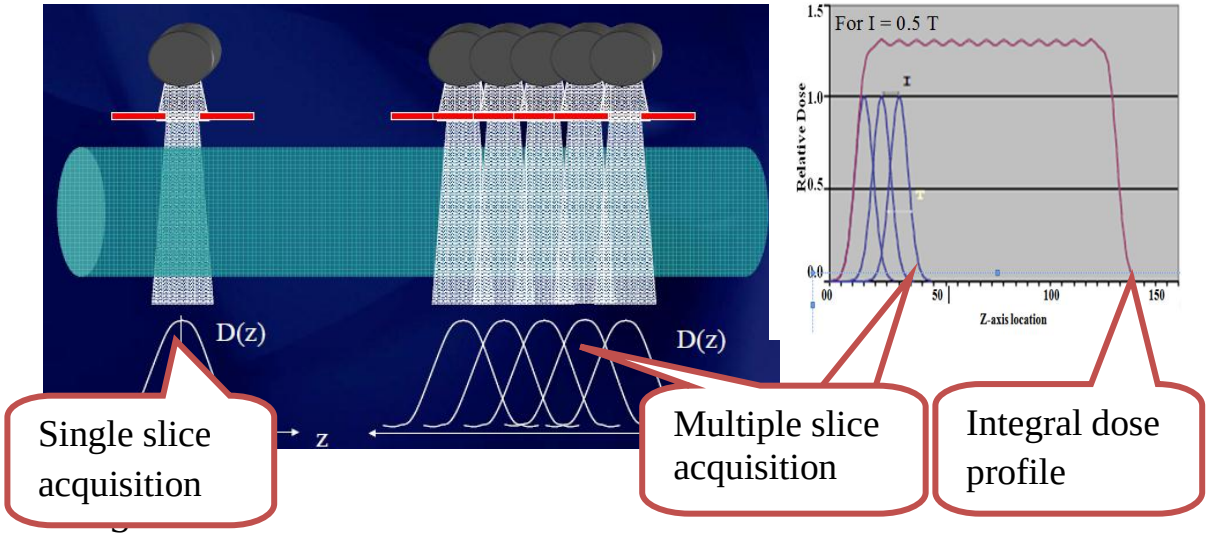


Figure (2.14) Illustrates meaning of the term: CTDI

The local tissue dose from a single slice is not the same as the dose in the very same tissue when additional adjacent slices are made, because each additional slice scatters radiation into adjacent slices.

D(z) = dose profile along z-axis from:

Even if slices are non-overlapping (and ignoring beam penumbra) scatter tails of multiple contiguous scans overlap and contribute to an increased integral dose Profile, which is a Function of:



- an Profile Width (T)
- Number of scans (N)
- Spacing (I) between slices

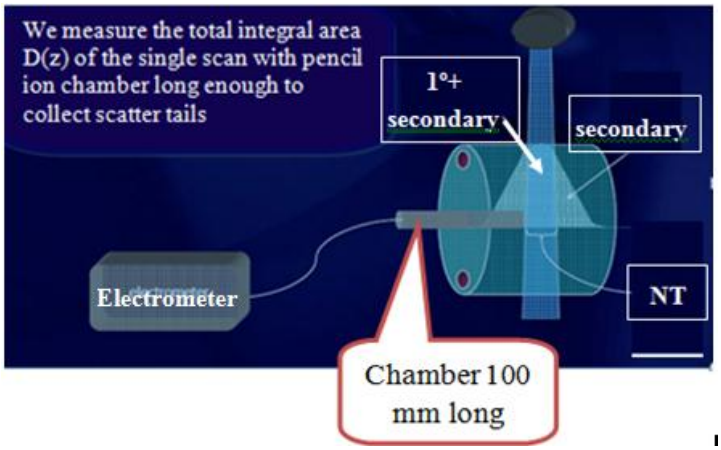
- Computed Tomography Dose Index (CTDI) – defined:

  Total area of D(z) *under width T* of central scan of multiple scan profile = total area of single scan dose profile (including scatter tails).

CTDI is calculated from measured D(z)

Calculating CTDI:

- Single axial scan (in phantom to emulate patient scatter) of nominal *beam* thickness T is given by:



2.10 Image quality

2.10.1 Contras

Contrast is a measure of the relative signal difference between two locations in an image, especially between the image of an object and the background (Cunningham IA ,2000). The contrast of an imaging system is described by the characteristic response curve of the system. This curve has a typical Sshape for a screen-film system and an X-ray film can easily become over- or underexposed because there for most systems is not a simple linear relationship. In digital systems the characteristic curve is generally linear and thus the risk of over- or underexposure can be avoided; however, if the user is not properly trained, there is an obvious risk that the patient exposure can be unnecessarily high since a digital detector does not set the limit as film does with respect to film blackening.

2.10.2Resolution

Image degradation due to spatial resolution properties can be characterized effectively by observing the response of a system to known, simple objects and such information can be used to predict the response to more complex objects. The point spread function (PSF), line spread function (LSF) and edge spread function (ESF)are the response of a system to point, line and step-edge objects, respectively. The modulation transfer function(MTF) of a system is defined as its response to a sinusoidal input; it specifies the relative amplitude of the output signal as a function of the spatial frequency of the sinusoid. In general, the response will decrease as the frequency increases. The MTF curves can be derived simply by computing the two-dimensional Fourier transform of the PSF or, by the one-dimensional transform of the line spread function(Metz CE etal, 1979)A discussion of the measurement of MTF in radiographic systems is given by ICRU Report 54 (1996)

2.10 .3Noise

The introduction of noise into an imaging process means that the system is no longer deterministic and that its performance must be analyzed using statistical methods.

Random noise is the fluctuations of the signal over an image, as result of a uniform exposure, and can be characterized by the standard deviation of the signal variations over the image of an uniform object. To get a more complete description of the spatial correlation of the noise its Wiener spectrum has to be used which measures the noise power as a function of spatial frequency (Geiger ML, etal1984).While the difference in signal amplitude of an object relative to the background (contrast) and the noise properties of the image are important characteristics of the image quality, it is the ratio between them

that is most significant indicator of image quality (Dobbins JT .2000). The signal-to noise ratio (SNR) is defined by the following equation:

$$SNR = \Delta S/\sigma \qquad (1)$$

Where ΔS is difference in signal of an object and of the background, and σ is standard deviation of the signal distribution in a background area . The SNR of a radiographic system describes the ability of the system to reproduce low-contrast objects. The quantity SNR is especially useful in digital radiography systems, where the image noise can be lowered by increasing the number of photons in the image and thereby also increasing the patient dose , and the contrast can be varied arbitrarily by changing the slope of the characteristic curve of the system. A combination of the primary physical characteristics (contrast, spatial resolution and noise) into a common quantity will be a measure of the overall imaging capability of a radiographic system. Such a quantity can be used to compare the total image quality of two radiographic systems. For instance, a system with high spatial resolution but with very high in here noise can have lower overall image quality than the system with moderate spatial resolution and low noise. The combination of the primary physical characteristic scan be achieved analytically or indirectly. The detective quantum efficiency (DQE) and the image quality

index (IQI) are performed by combining primary physical characteristics into one equation. The contrast detail resolution test combines the primary physical

Characteristic indirectly, by a study of the visibility of small low-contrast objects in the final image. The detective quantum efficiency is a parameter that describes the overall capability of the system to use the information of the incoming photon fluency distribution for the formation of the image. It is defined as the square of the quotient of the signal-to-noise ratio of the signal

out of the imaging system over the signal-to-noise ratio impinging on the imaging system: $DQE = \{SNRout / SNRin\}$ (2)

Determination of the DQE is made by combining the MTF, the Wiener spectrum and the response of the system, but there is at the moment no widely accepted standard method, although it is a valuable quantity for comparing the performance of various imaging systems especially digital systems. Another method for describing the overall performance of an imaging system is the image quality index(IQI) (Desponds L, 1991) . The IQI of an image system is the diameter of the smallest high-contrast sphere that is detectable with the system. The IQI is determined by combining the MTF, the Wiener spectrum and the contrast of the imaging system and is mostly used in mammography(Verdun FR etal ,1996).A tool for quick determination of the performance of an imaging system is the use of a contrast-detail phantom (Thijsen MAO,1989)The phantom consists of a PMMA slab with cylindrical holes of different diameter and depth or for mammography, gold slices of different diameter and thickness included in a PMMA slab. For a given diameter the observer marks the detail that is on the edge of detection as the smallest visible contrast in the image .Due to the subjective nature of human observers, a detection algorithm may be used by a computer program(Jansen JTM etal, 2000)

2.11Previous studies

Breikietal. (2008) Evaluated Radiation Dose and Image Quality for Patients Undergoing Computed Tomography (CT) Examinations , to study the CT practice in some CT units in different hospitals in Egypt, in order to investigate the radiation doses imparted to patients during CT examinations and image quality. Method used to CTDI measurements free-in-air and in quality control (QC) phantom using ion chamber, and hence DLP

calculations. Monte Carlo technique utilizing CTDI measurements to calculate organ and effective doses . Direct approach using TLD and Rando phantom to measure organ dose . Computed Tomography Dose Index (CTDI)w was calculated for each scan from an average of three measurements in the head phantom and another three measurements in the body phantom. DLP values were estimated for each type of examination. Mean values of CTDIw had a range of 36.0-69.0 mGy for head and11.0-30.0 mGy for chest, abdomen and pelvis examinations. Organ dose and hence effective dose, calculated using Monte Carlo simulation technique. The effects of selecting tube KV and mAs on both spatial resolution and low contrast detestability were examined for two groups of KV values (90and 120), the mAs values were degraded from 100 to 300 mAs in 100 mAs interval in first case, and from 50 to 300mAs, in 50mAs interval in second case.

Loubele etal. (2008) investigated Image quality vs. radiation dose of four cone beam computed tomography scanners, To evaluate image quality by examining segmentation accuracy and assess radiation dose for cone beam CT (CBCT) scanners .Methods: A skull phantom, scanned by a laser scanner, and a contrast phantom were used to evaluate segmentation accuracy. The contrast phantom consisted of a polymethylmethacrylate (PMMA) cylinder with cylindrical inserts of air, bone and PMMA. The phantoms were scanned on the (1) Accuitomo 3D, (2) MercuRay, (3) NewTom 3G, (4) i-CAT and (5) Sensation 16. The structures were segmented with an optimal threshold. Thicknesses of the bone of the mandible and the diameter of the cylinders in the contrast phantom were measured across lines at corresponding places in the CT image vs. a ground truth. The accuracy was in the 95th percentile of the difference between corresponding measurements. The correlation between accuracy in skull and contrast phantom was calculated. The radiation dose

was assessed by DPI100,c (dose profile integral 100,c) at the central hole of a CT dose index (CTDI) phantom. Results: The results for the DPI100,c were 107 mGy mm for (1), 1569 mGy mm for (2),446 mGy mm for (3), 249 mGy mm for (4) and 1090 mGy.mm for (5). The segmentations in the contrast phantom were sub millimeter accurate in all scanners. The segmentation accuracy of the mandible was 2.9 mm for (1), 4.2 mm for (2), 3.4 mm for (3), 1.0 mm for (4) and 1.2 mm for (5). The correlation between measurements in the contrast and skull phantom was below 0.37 mm..

Jumaa Yousif Tamboul, (2014), Measured of adult and pediatric Patient doses during head CT scan The objectives of this study were to measure doses from CT examinations of adult and pediatric patients and to compare the doses with international standards as provided in DRLs. Method used A total of59 patients (pediatric and adults) were examined at the Department of Radiology, Al -Ribat University Hospital-Khartoum. The mean age was 40.80 years for adults while the mean weight was 70.04 kg and the mean age for pediatric was 5.10 years while the mean weight was 20kg.Result : DLP for adults were 1000.25 mGy.cm, 733.33 for pediatrics. The mean effective dose for adults patient was 0.48 mSv in rang (0.49-0.44) mSv, while for pediatric patients was 0.31mSv in rang between (0.49- 0.11) mSv. The DRL was 1120 mGy.cm, a value which is higher than the European Guidelines on Quality Criteria for Computed Tomography. The study has shown a great need for referring criteria, continuous training of staff in radiation dose optimization concepts. Further studies are required inJörgHausleiter et al(2006) Estimated Radiation Dose From Cardiac Multi slice Computed Tomography in Daily Practice the aim of the present retrospective study was (1) to estimate the effective dose from cardiac CT angiography in daily practice, (2) to compare the dose estimates from 16- and 64-slice cardiac CT angiography, and (3) to evaluate

the impact of 2 different strategies on dose savings and image quality with both scanner types. Methods : Radiation dose was estimated for 1035 patients undergoing coronary CTA. Scanning algorithm switch and without an ECG-dependent dose modulation and with a reduced tube voltage were investigated on dose estimates and image quality. Result: radiation dose estimates were 6.4_1.9 and 11.0_4.1 mSv for 16- and 64-slice CTA, respectively (P_0.01). The reduction in radiation dose estimates ranged between 37% and 40% and between 53% and 64% with the use of ECG-dependent dose modulation and with the combined use of the dose modulation and a reduced tube voltage, respectively. The reduction in dose estimates was not associated with a reduction in diagnostic image quality as assessed by the signal-to-noise ratio and by the frequency of coronary segments with diagnostic image quality.

Christophe etal (2004) Assessed Patient, Physician, and Radiologist Awareness of Radiation Dose and Possible Risks this study aimed to determine the awareness level concerning radiation dose and possible risks associated with CT scans among patients, emergency department (ED) physicians, and radiologists. Method s : Adult patients seen in the ED of a U.S. academic medical center during a 2-week period with mild to moderate abdominopelvic or flank pain and who underwent CT were surveyed after acquisition of the CT scan. Patients were asked whether or not they were informed about the risks, benefits, and radiation dose of the CT scan and if they believed that the scan increased their lifetime cancer risk. Patients were also asked to estimate the radiation dose for the CT scan compared with that for one chest radiograph. ED physicians who requested CT scans and radiologists who reviewed the CT scans were surveyed with similar questions and an additional question regarding the number of years in practice. The X2 test of independence was used to compare the three respondent groups

regarding perceived increased cancer risk from one abdominopelvic CT scan RESULTS: Seven percent (five of 76) of patients reported that they were told about risks and benefits of their CT scan, while 22% (10 of 45) of ED physicians reported that they had provided such information. Forty-seven percent (18 of 38) of radiologists believed that there was increased cancer risk, whereas only9%(four of 45) of ED physicians and3%(two of 76) of patients believed that there was increased risk (X22= 41.45, P<.001). All patients and most ED physicians and radiologists were unable to accurately estimate the dose for one CT scan compared with that for one chest radiograph

Frederiksen et al,(1995) assessed Effective dose and risk from computed tomography of the maxillofacial complex, This study was therefore undertaken to establish Å from this radiographic technique.(maxillofacial complex in CT.) Methods :Dosimetry was performed using thermoluminscent dosimeters (TLD). For each scan sequence 64 TLDs were placed in 27 selected sites in the upper portion of a tissue-equivalent human phantom to record the equivalent dose in radiosensitive organs/tissues. Results The greatest equivalent dose recorded was that delivered to the salivary glands during scans of the mandible(20.4 mSv) and the least was that received by the bone marrow during scans of the maxilla (0.11 mSv).With the salivary glands assigned a weighting factor of one half the remainder, the effective doses were 0.1mSv and 0.76 mSv for the maxilla and mandible respectively. From these data, it was calculated that scanning themaxilla corresponded to a probability of stochastic effects of 8 x 10- 6 and for the mandible of 56 ÷ 10"6.The total probability was therefore 63 x 10"6.

Cohnen et al. (2006) investigated Radiation Exposure of Patients in Comprehensive Computed Tomography of the Head in Acute Stroke ,This

study To assess patient radiation exposure in comprehensive stroke imaging using multidetector row CT (MDCT) combining standard CT of the head, cerebral perfusion (CTP), and CT angiography (CTA) studies. Methods: Examination protocols for CT and CTA of cerebral and cervical vessels, as well as CTP were simulated using a Somatom Sensation Cardiac 64. Effective doses were derived from measurements with the use of lithium-fluoride thermoluminscent dosimeters (LiF-TLD) at several organ sites using an Alderson-Rando phantom .RESULTS: LiF-TLD measurements resulted in effective doses of 1.7 mSv for CT, 1.9 mSv for CTA of intracranial vessels, and 2.8 mSv for CTA of cervical vessels, respectively. Depending on examination parameters, effective doses varied between 1.1 and 5.0 mSv for cerebral CTP. For CTP, local doses in the area of the primary beam ranged between 114 and 444 mGy.

Chapter three
Material and Method

3.1 Material

Three CT machines were used to collect data during this study. These machines were installed in three private radiological departments. All quality control tests.

Performed for the machine prior any data collection, the tests were carried out by experts from Sudan Atomic Energy Commission (SAEC).

Table (3.1)Showed CT machines were used to collect data during this study

Hospital	manufacture	Model	Detected Type
Hospital 1	General electric	Optima	16
Hospital 2	Toshiba	Aquilion	16
Hospital 3	General electric	Optima	64
Hospital4	General electric	Optima	16

3-2 design of the study

The data of this study collected prospectively, it is a descriptive study intends to evaluate the received dose and the quality of the image obtained with the current exposure setting.

3.3. Population of the study

The data of this study collected from adult patient from three types of CT scan, which include head, chest and abdomen in three hospital in Khartoum state during the period from 2017 to 2018.

3-4 Sample size and type

A total of111 Patients60 adult were examined for the abdomen 29 of them were female and the others were males and 34 adult were examined for the brain15of them were female and the others were males and17 adult were examined for the chest8 of them were female and the others were males there aged ranged 25-75years

3.5 Method of data collection

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

Effective Dose (E) in CT given by the following

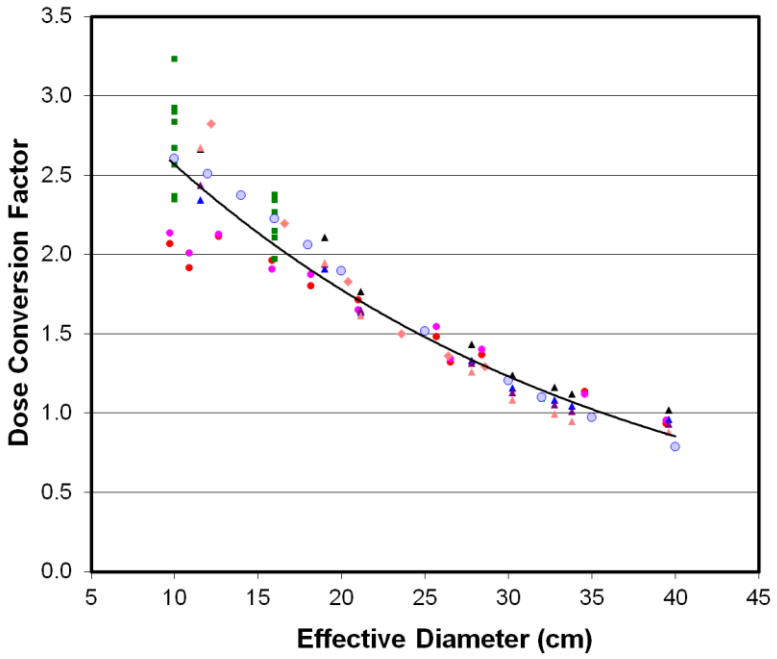
*E=DLP *k*

Where k is the tissue weighting factor based on region of body scanned

Representative adult values for k are:Head/Neck =.0031, Head = .0021, Chest = 014 and Abdomen = .015

$SSDE = Dose_{patient} \approx CTDI \times f$

f is a dose conversion factor that depends on patient size (effective diameter).



Chapter Four
Results

Table(4.1)Showed The overall mean of CTDvol, DLP and SSDE values for the three Hospitals

Exam	CTDIvol(mGy)	DLP(mGy)	SSDE(mGy)
Head	60.5	1332.025	61.71
ABD	15.1525	2121.45	16.0025
Chest	13.18	593.09	13.18

Table(4.2) The Mean values of CTDvol, DLP and SSDE values for Hospital 1

Exam	CTDIvol(mGy)	DLP(mGy)	SSDE(mGy)
ABD	13.925	1665.405	14.2035
Chest	13.18	593.09	13.4436

Table(4.3) The Mean values of CTDvol, DLP and SSDE values for Hospital 2

Exam	CTDIvol(mGy)	DLP(mGy)	SSDE(mGy)
Head	60.5	1211.05	61.71
ABD	31.3	1475.05	31.926

Table(4.4) The Mean values of CTDvol, DLP and SSDE values for Hospital 3

Exam	CTDIvol(mGy)	DLP(mGy)	SSDE(mGy)
Head	54.1	8115	55.182
ABD	13.225	5125.5	13.4895
Chest	9.300	2906	7.5888

Table(4.5) Showed Mean effective dose (in mSv) in the three hospitals

Hospitals	Mean	St. Deviation
Hospital1	1.37	12.12
Hospital	5.40	6.34
Hospital	7.66	47.08

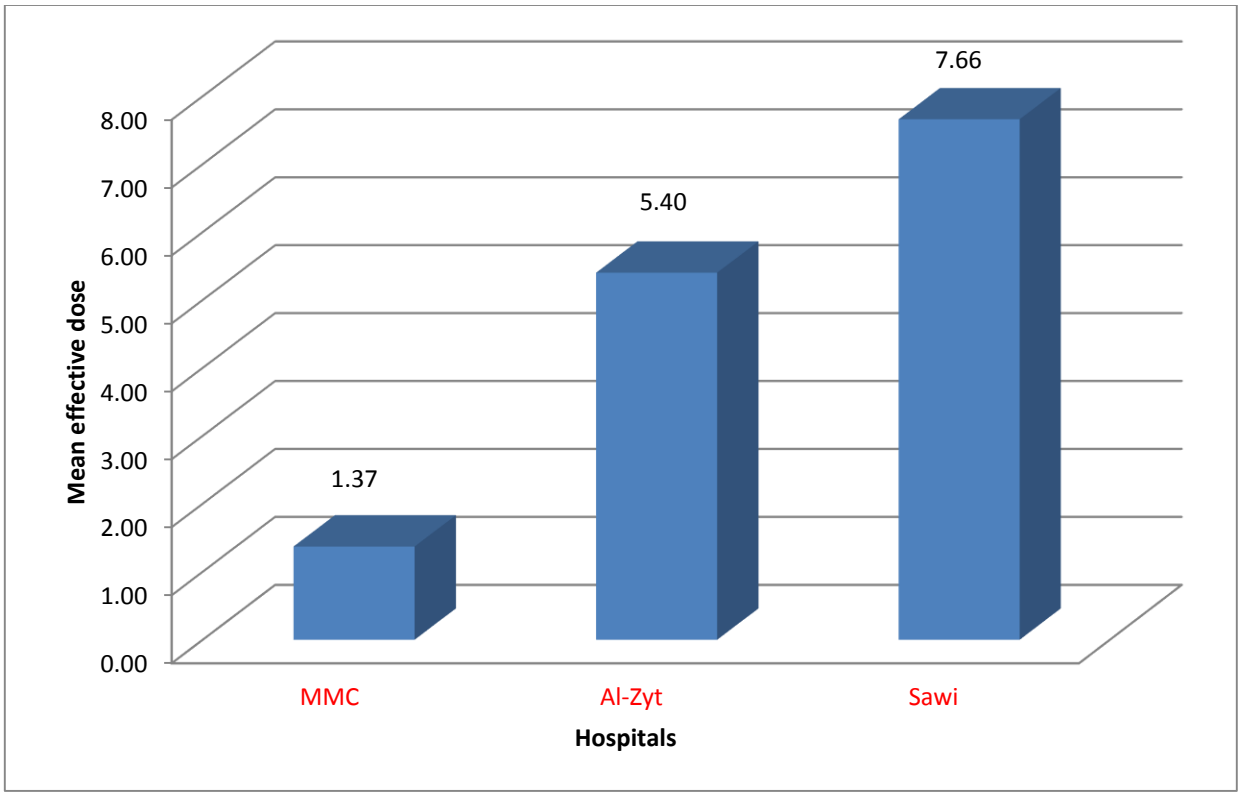


Figure (4.1): Bar graph show the mean effective dose of the three hospitals

Table (4.6): Showed mean SSDE (in mGy) in the three hospitals

Hospital	Mean	St. Deviation
Hospital1	1.88	18.31
Hospital2	4.85	21.40
Hospital3	1.84	54.15

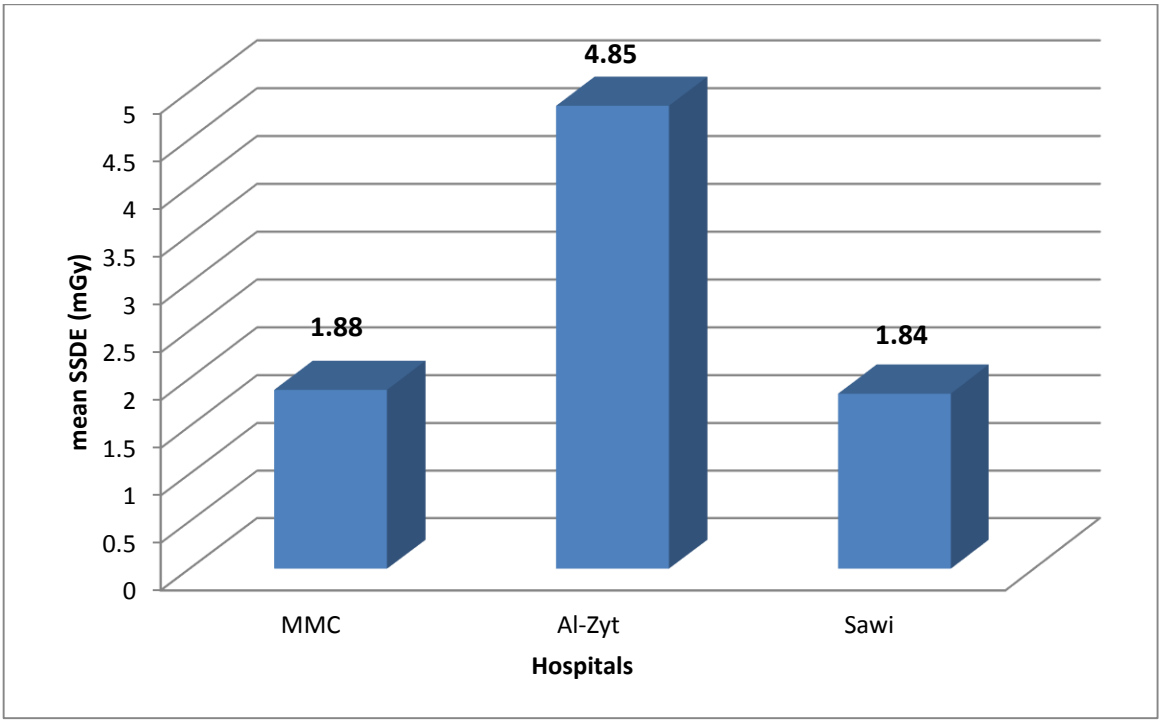


Figure (4.2): Bar graph show mean SSDE in the three hospitals

Table (4.7): Showed Mean effective dose (in mGy) for every exam type

Exam type (our Code)	Mean	St. Deviation
ABD(1)	1.72	14.42
Head(3)	5.65	15.22
Chest(2)	1.08	12.13

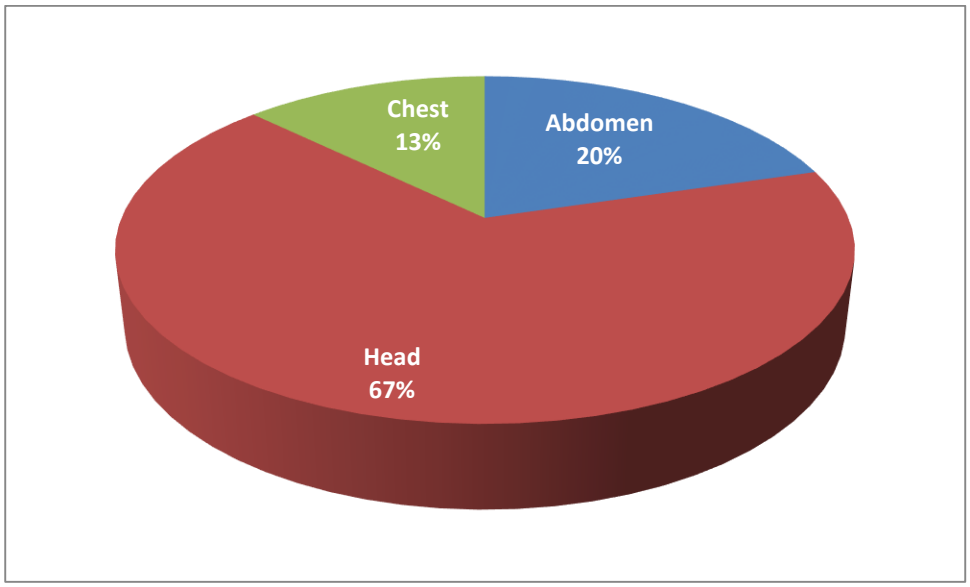


Figure (4.3) Pie graph shows the relative percentage of the mean effective dose (in mGy) for every exam type

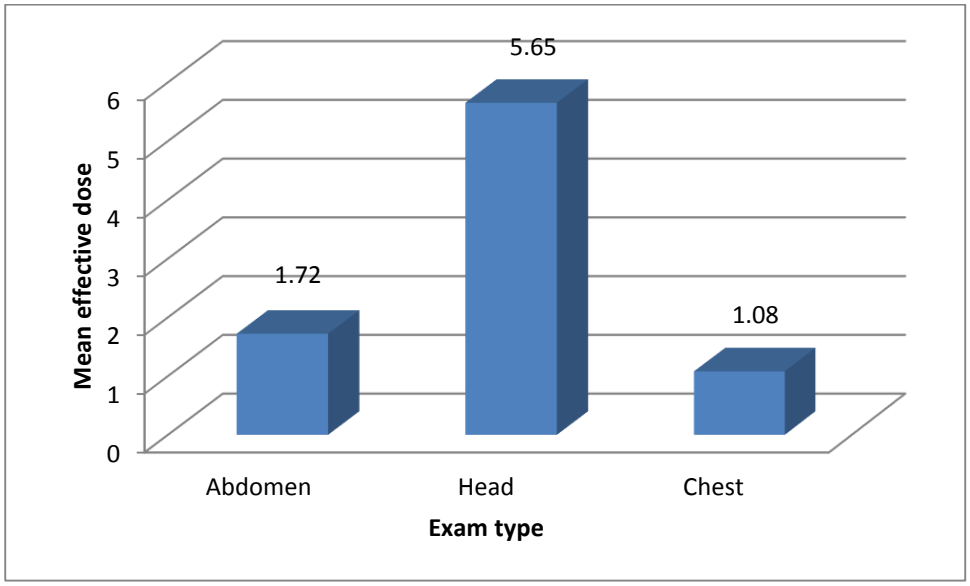


Figure (4.4) Bar graph shows mean effective dose (in mGy) for every exam type

Table (4.8): Showed Mean SSDE (in mGy) in the three hospitals

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

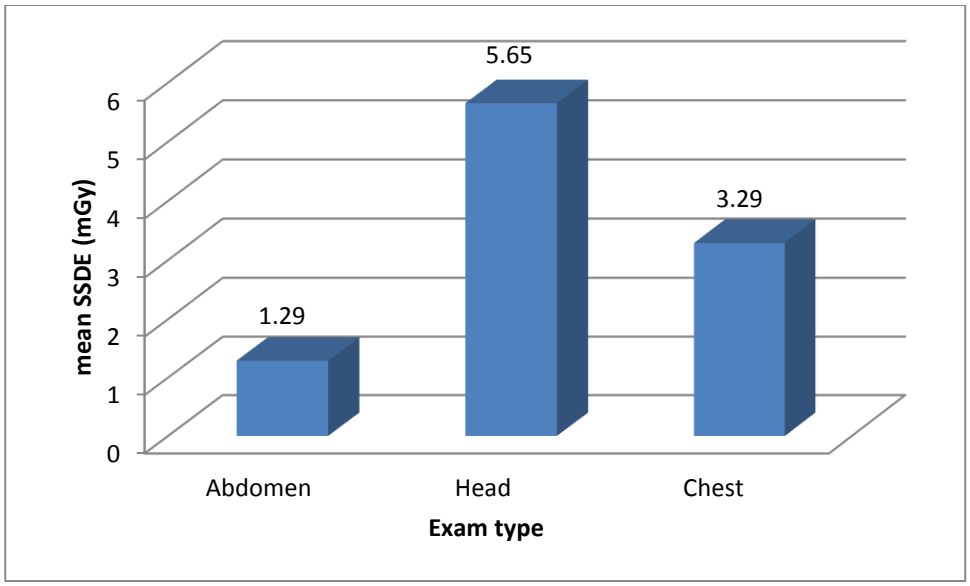


Figure (4.5) Bar graph shows mean SSDE (in mGy) in the three hospitals

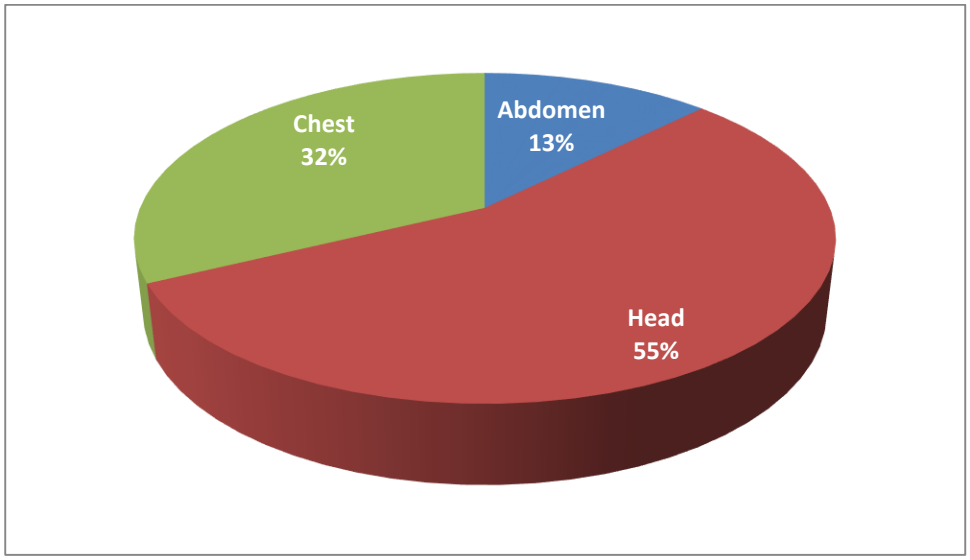
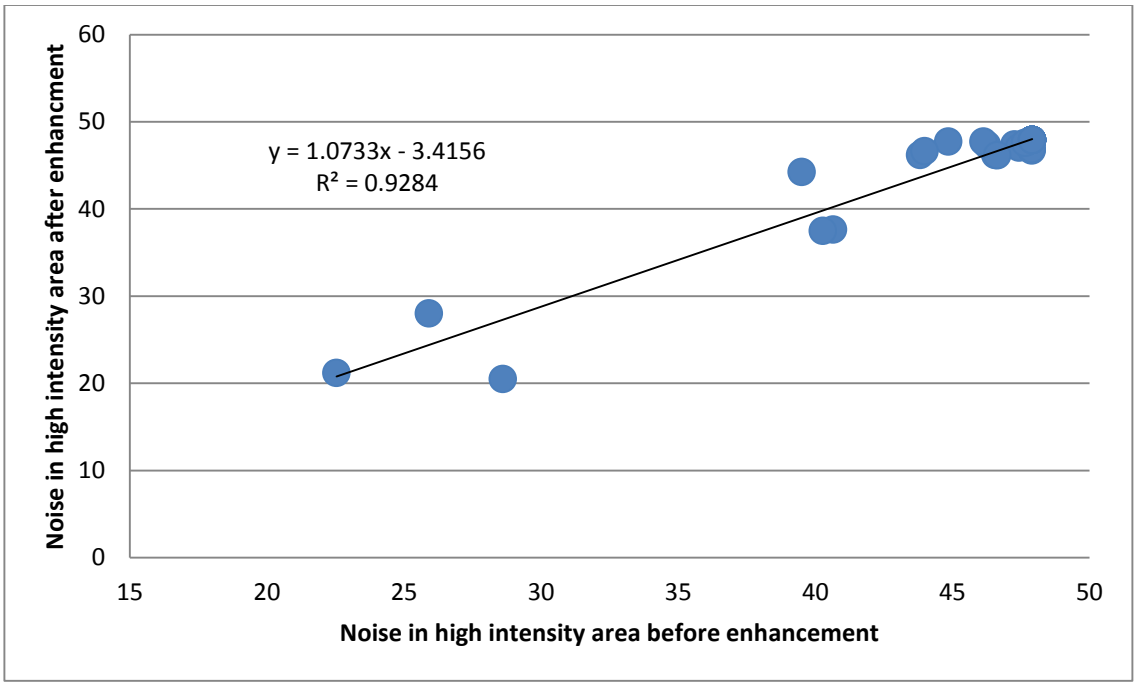
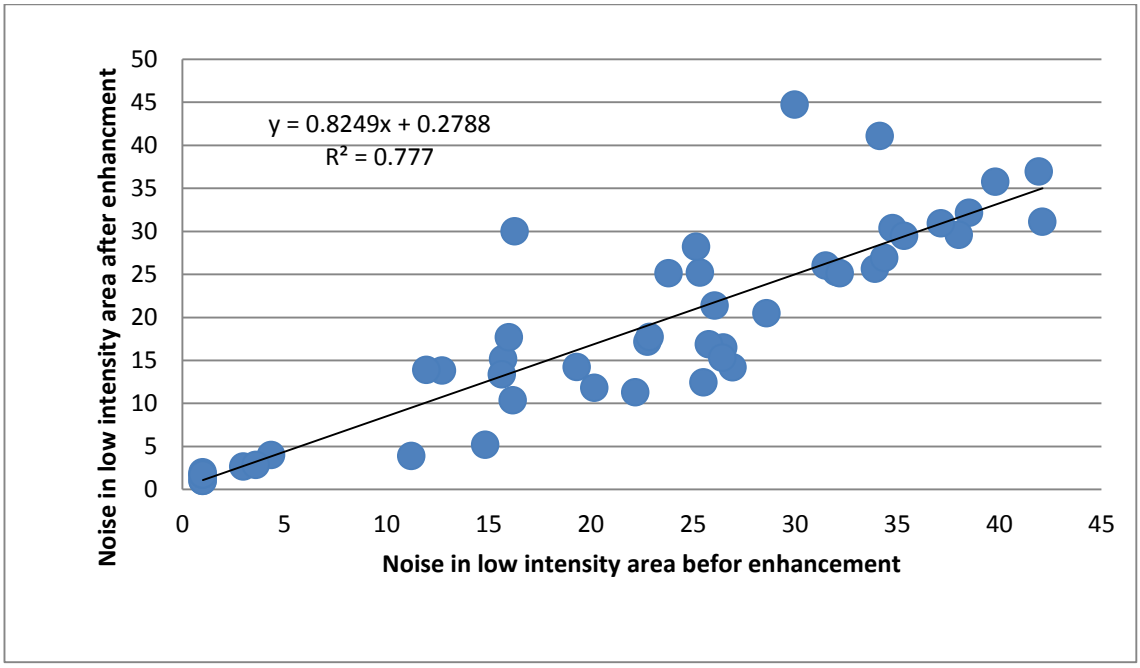


Figure (4.6) Pie graph shows the relative percentage of mean SSDE (in mGy) in the three hospitals

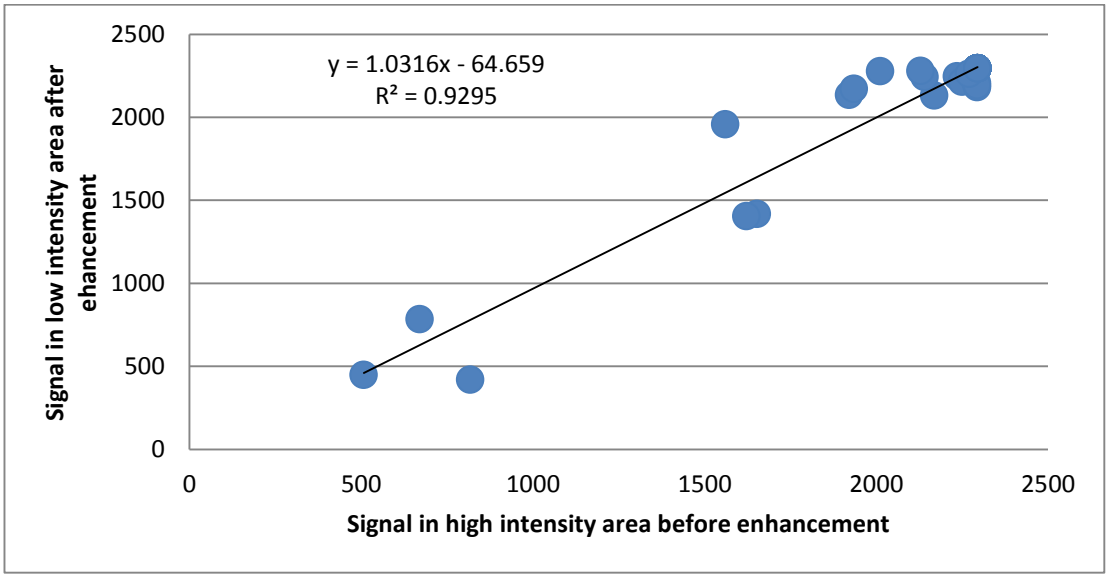


(A)

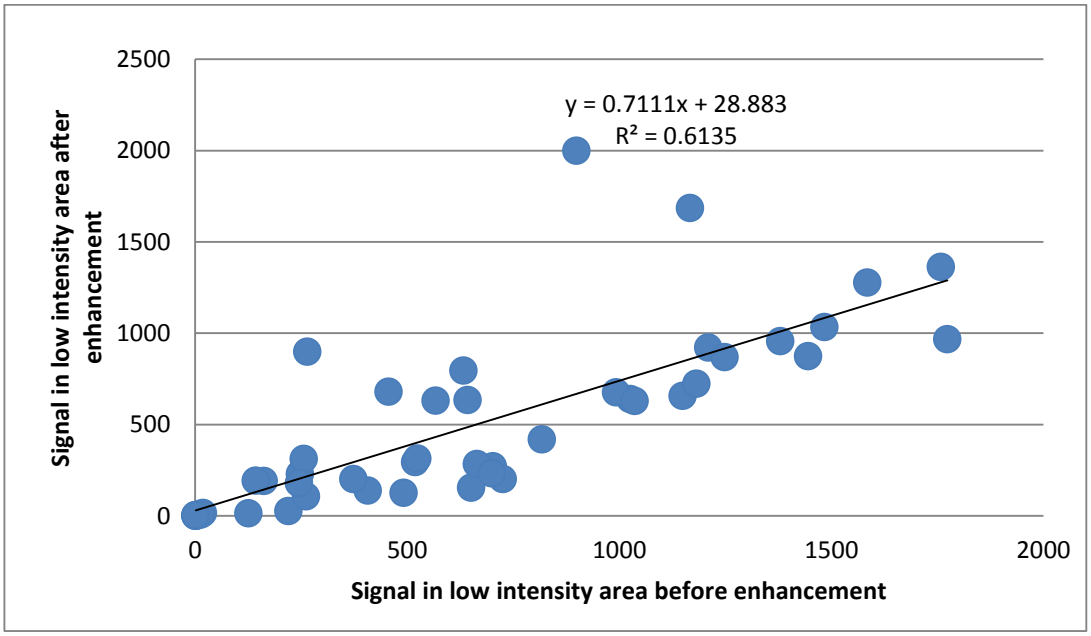


(B)

Figure (4.7): Scatter plot show a direct linear relationship between noise before the enhancement and after enhancement (A) in the high intensity area and (B) low intensity area

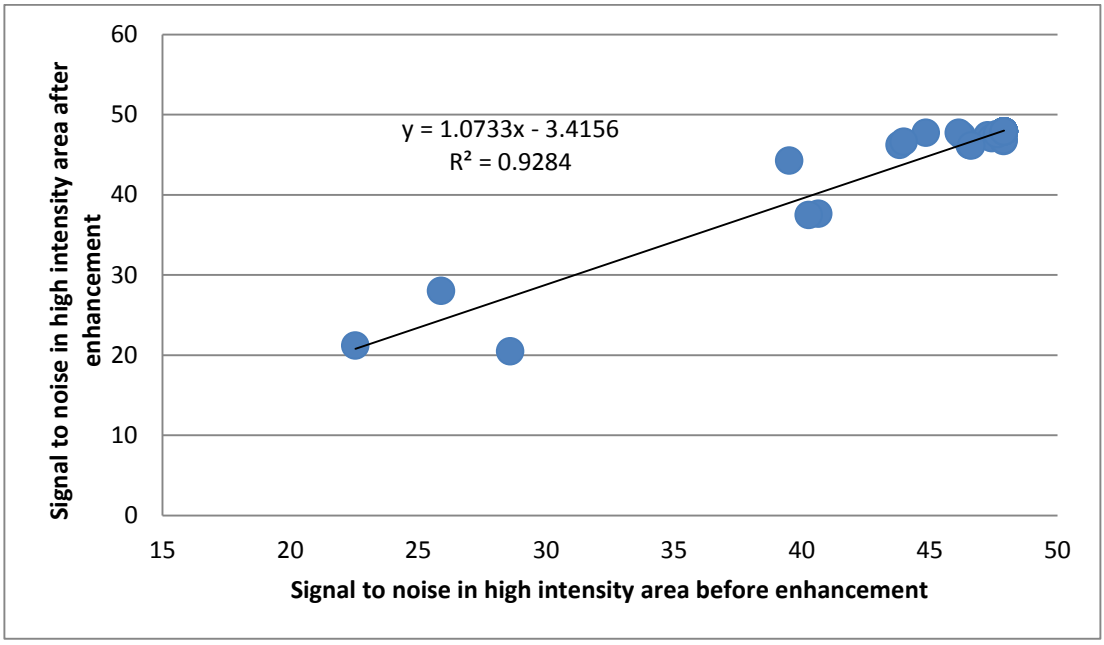


(A)

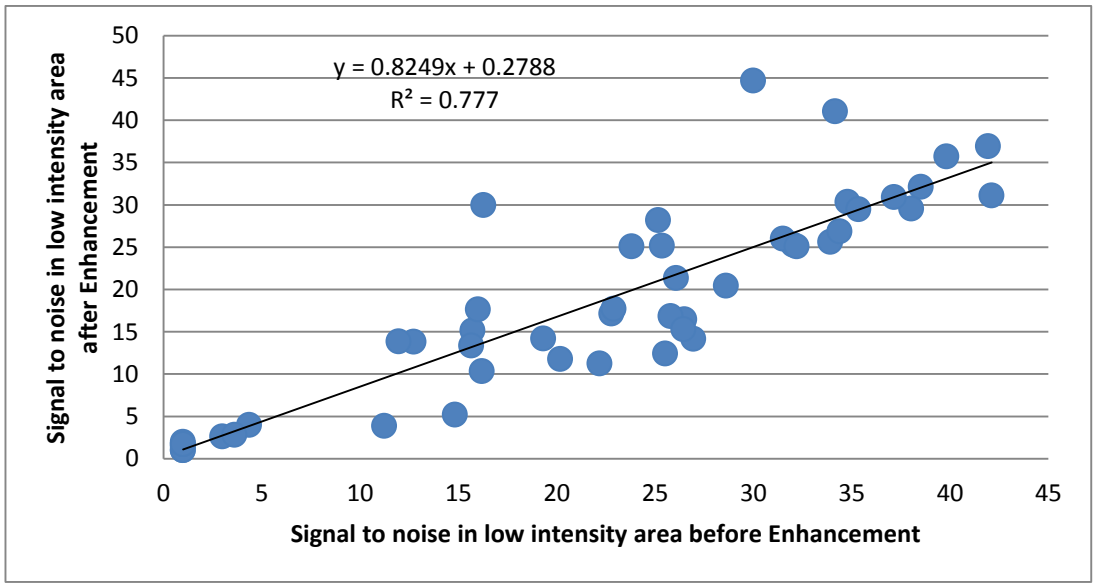


(B)

Figure (4.8): Scatter plot show a direct linear relationship between signal before the enhancement and after enhancement (A) in the high intensity area and (B) low intensity area



(A)



(B)

Figure (4.9): Scatter plot show a direct linear relationship between signal to noise before the enhancement and after enhancement (A) in the high intensity area and (B) low intensity area

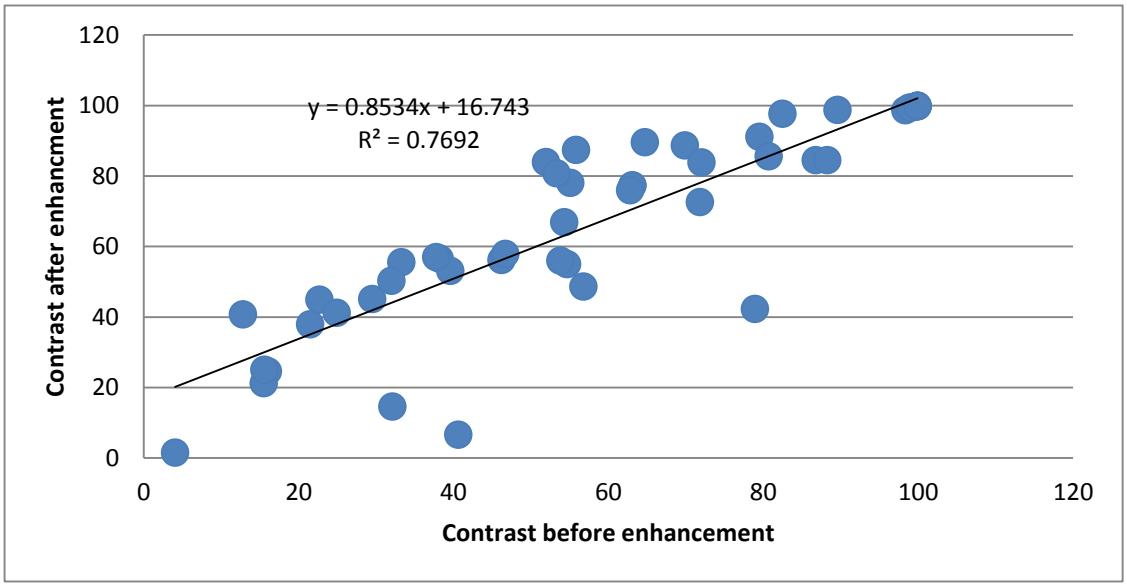


Figure (4.10): Scatter plot show a direct linear relationship between contrast before the enhancement and after enhancement

Table(4-9) Significance paired t-test between the image quality factor before and after enhancement

Before and after enhancement	*t*	Sig. (2-tailed) (*p* value)
Signal in high intensity area	-.085	.932
Noise in high intensity area	.276	.784
Signal to noise in high intensity area	.276	.784
Signal in low intensity area	-3.382	**0.001**
Noise in low intensity area	-11.286	**<0.001**
Signal to noise in low intensity area	-11.286	**<0.001**
Contrast	-4.029	**<0.001**

Table(4.10) The mean, standard deviation, minim and maximum values for the image quality factors

Image quality factors		mean	STD	MIN	MAX
Before enhancement	signal high intensity area	2118.46	416.41	508	2295
	noise high intensity area	45.69	5.58	22.54	47.91
	signal to noise high intensity area	45.69	5.58	22.54	47.91
	signal low intensity area	638.09	518.93	1	1774
	noise in low intensity area	22.01	12.44	1	42.12
	signal to noise low intensity area	22.01	12.44	1	42.12
	Contrast	59	29	4	100%
After enhancement	signal high intensity area	2119.91	445.45	419	2295
	noise high intensity area	45.62	6.21	20.47	47.91
	signal to noise high intensity area	45.62	6.21	20.47	47.91
	signal low intensity area	480.53	471.99	1	1999
	noise in low intensity area	18.42	11.64	1	44.71
	signal to noise low intensity area	17.44	10.85	1	41.06
	contrast	67	28	1%	100%

Chapter Five
Discussion, Conclusion and Recommendation

5-1 Discussion

The results obtained by this study which demonstrates the effective dose in hospital (1, 2 and 3) was1.37±12.12mSv, 5.40±6.34mSv and 7.66±47.08mSv respectively, also the Size specific dose estimate (SSDE) was1.88±18.31mGy, 4.85±21.40mGy and 1.84±54.15mGy. As well as CTDvol, DLP and SSDE values in every exam Type firstly head was 60.5mGy, 1332.025mGy and61.71mGy respectivelyand for abdomenwas15.1525mGy, 2121.45mGy and16.0025mGy respectively, finally Chest was 13.18mGy, 593.09mGy and13.18mGy.

CTDvol of every Exam Type in Three hospitals shown in Figs(4.2,4.3and4.4) agree with National DRLs Of CT Examination(Uk) as shown below and agree with Ali (2011).

The DLP for the three hospitals as shown in Figs (4.2,4.3 and4.4) which concern the chest Exam in hospital 1 agree with National DRLs Of CT Examination (Uk) and other Exams(head and abdomen) were not agree with National DRLs of CT Examination (Uk) because the obtained values were significantly greater than the international one but it agree with Ali (2011) since his study done in the same environment.

For the Effective Dose values in every Exam Type in the three hospitals ABDHead and Chest was 1.72mSv , 5.65mSv and 1.08mSv respectively, this results agree with Ali (2011), Nour (2016) and EuropeanCommission (1999)

For image quality which quantified using noise, signal, signal to noise ratio (in low and high intensity) and contrast before and after enhancement; the results show Mean ±SD, for signal high intensity before 2118.46±416.41, for

noise is 45.69±5.58, and for signal to noise 45.69±5.58. Mean ±SD, for signal inlow intensity area before 638.09±518.93, for noise is 22.01±12.44, and for signal to noise is 22.01±12.44. Mean ±SD, for contrast before 59±29. And Mean ±SD, for signal high intensity after 2119.91±445.45, for noise is 45.62±5.58, and for signal to noise is 45.62±5.58. Mean ±SD, for signal inlow intensity area after 480.53±471.99, for noise is 18.42±11.64, and for signal to noise is 17.44±10.85. Mean ±SD, for contrast after 67±28.

There is a direct linear relation between the signal in high intensity area before enhancement and after enhancement where the signal after enhancement increase by 1.07 versus one unit of signal before the enhancement as well the graph shown in fig.(4.8) revel strong correlation between the signal before and after, which mean there is a homogenous increase in the signal.

There is a direct linear relation between the signal in low intensity area before enhancement and after enhancement where the signal after enhancement increase by0.711 versus one unit of signal before the enhancement as well the graph shown in fig.(4.8) revel strong correlation between the signal before and after, which mean there is a homogenous increase in the signal.

There is a direct linear relation between the noise in high intensity area before enhancement and after enhancement where the noise after enhancement increaseby1.031 versus one unit of noise before the enhancement as well the graph shown in fig.(4.7) revel strong correlation between the noise before and after, which mean there is a homogenous increase in the noise.

There is a direct linear relation between the noise in low intensity area before enhancement and after enhancement where the noise after enhancement increase by o.824versus one unit of noise before the enhancement as well the graph shown in fig.(4.7) revel strong correlation between the noise before and after, which mean there is a homogenous increase in the noise.

There is a direct linear relation between the signal to noise in high intensity area before enhancement and after enhancement where the signal after enhancement increase by 0.928versus one unit of signal before the enhancement as shown in fig.(4.9) which reveal strong correlation between the signal before and after, which mean there is a homogenous increase in the signalto noise ratio.

There is a direct linear relation between the signal to noise in low intensity area before enhancement and after enhancement where the signal to noise after enhancement increase by 0.777versus one unit of signal before the enhancement as well the graph shown in fig.(4.9) revel strong correlation between the signal to noise before and after, which mean there is a homogenous increase in the signal to noise ratio.

There is a direct linear relation between the contrast before enhancement and after enhancement where the contrast after enhancement increase by 0.853 versus one unit of contrast before the enhancement as well the graph shown in fig.(4.10) revel strong correlation between the contrast before and after, which mean there is a homogenous increase in the contrast.

5-2 Conclusion

Computed tomography (CT) is an imaging technique which produces a digital topographic image from diagnostic x-ray. It always considered a "high dose" technique, there is growing realization that image quality in CT often exceeds the level needed for confident diagnosis and that patient doses are higher than necessary.

In this study, a total of 111 adult patients underwent the abdominal, head and chest CT scanning exams in Three hospitals were evaluated using effective Dose AND SSDE. The result of this study revealed that the mean effective dose in hospital (1,2 and 3).The Effective dose is higher in hospital3 Than other hospitals and SSDE dose is higher in hospital2 Than other hospitals. And evaluated effective dose in every Exam type is higher brain than other exams. In respect to received dose the CT abdominal examination dose was within the international diagnostic reference level values.

Inconclusion the image quality which include noise, signal, signal to noise and contrast in low and high intensity areas can be tested for the applicability and suitability of enhancement before starting the process using the following linear regression equation to estimate the validity as follows:

Signal after enhancement (HIA) = (1.31× Signal) - 64.5
Signal after enhancement (LIA)=(0.711× Signal)+28.8
Noise after enhancement (HIA)=(1.07× noise)-3.41
Noise after enhancement (LIA)=(0.824× noise)+0.278
SNR after enhancement(HIA)= (1.07× SNR) -3.4
SNR after enhancement(LIA)=(0.824× SNR) +0.278
Contrast after enhancement = (0.853×contrast)+16.7

Where HIA: High Intensity Area, LIA: Low Intensity Area and SNR:Signal To Noise Ratio

5-3 Recommendation

- CT scan examination for any patient must be justified
- QC program are essential to evaluate the patients dose and machine accurate
  performance.
- Future studies can be done in order to optimize the radiation dose to establish national diagnostic reference level in Sudan.
- Application of QC program as a routine task in all x-ray department in order to detect possible future problem

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Appendix (A)
Data

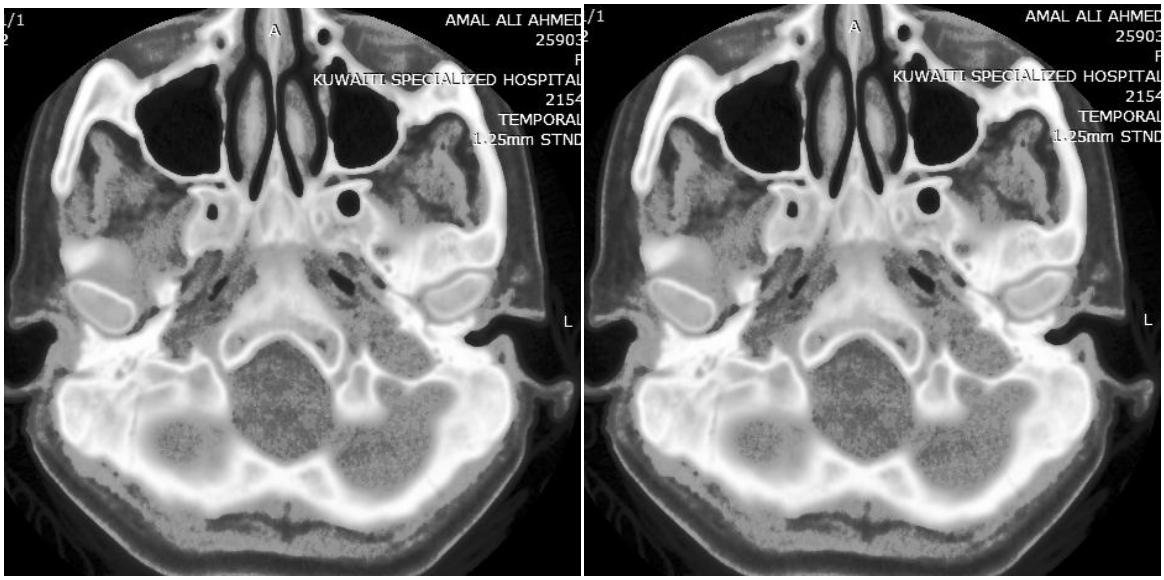
Data sheet for CT dose

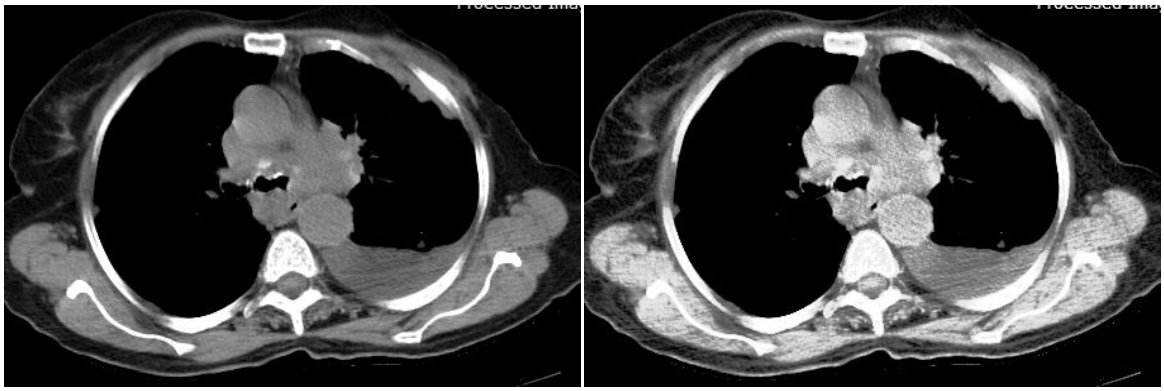
Exam	Gander	Age	Kv	$CTDI_{vol}$	$CTDI_{w}$	DLP	Pitch	ED	SSDE

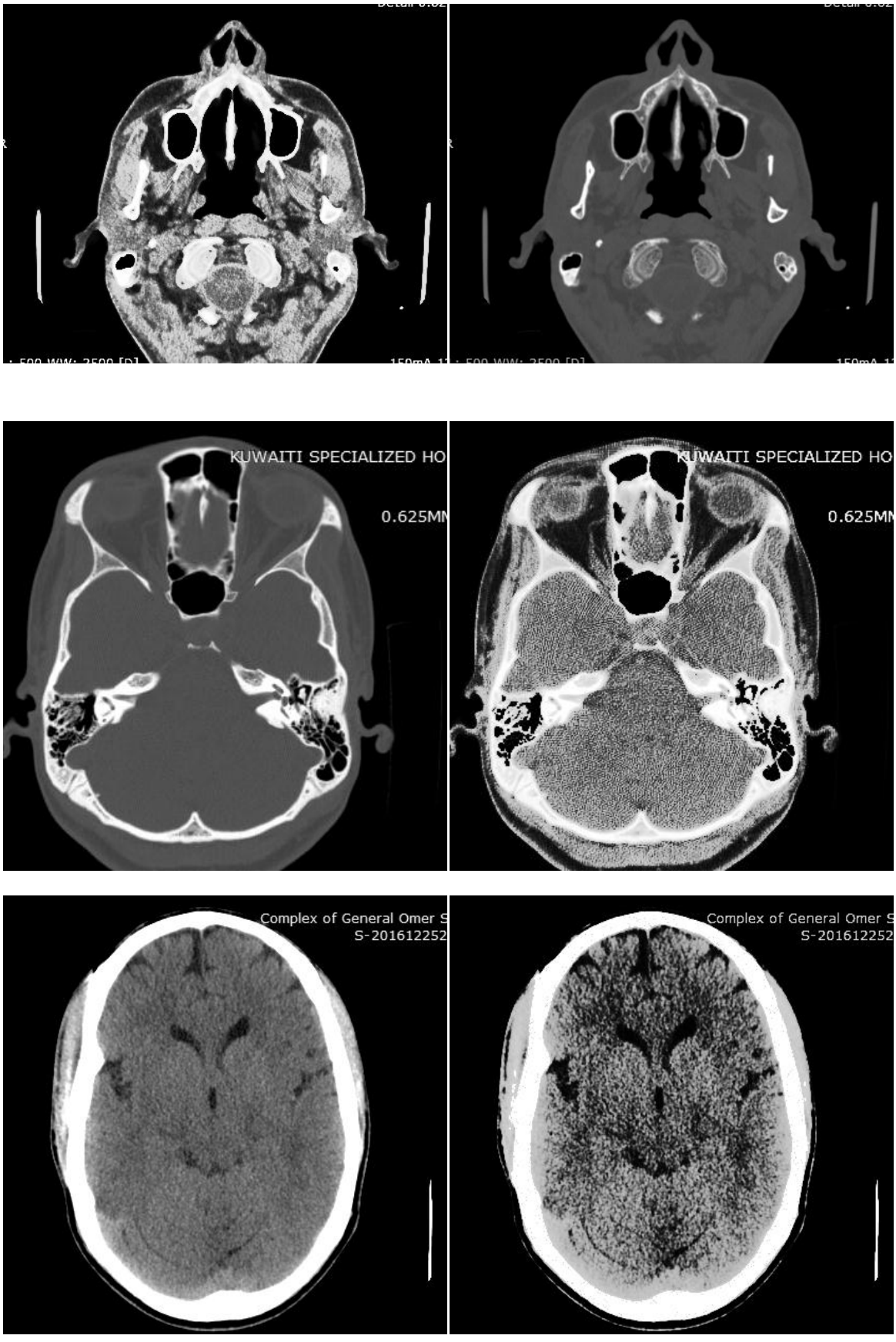
Data sheet for CT Image Quality

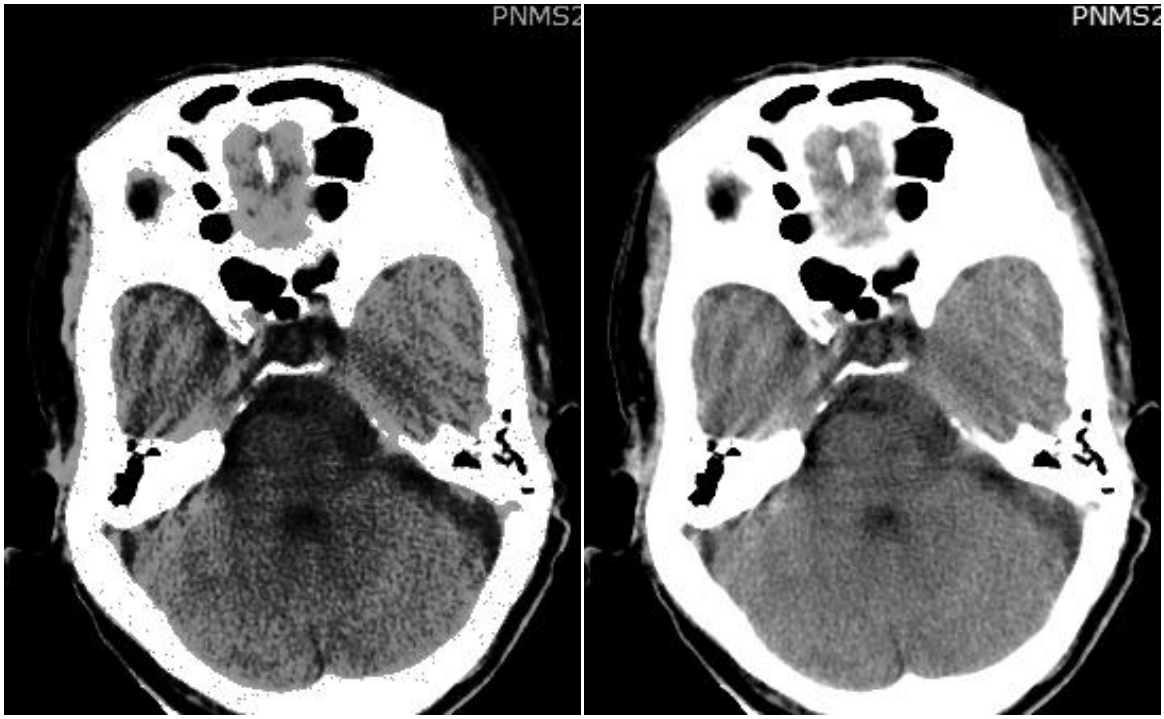
sigw	noisew	snw	sigb	noiseb	snb	contbefor	sigwe	noisewe	snw	sigbe	noisebe	snbe	contafter

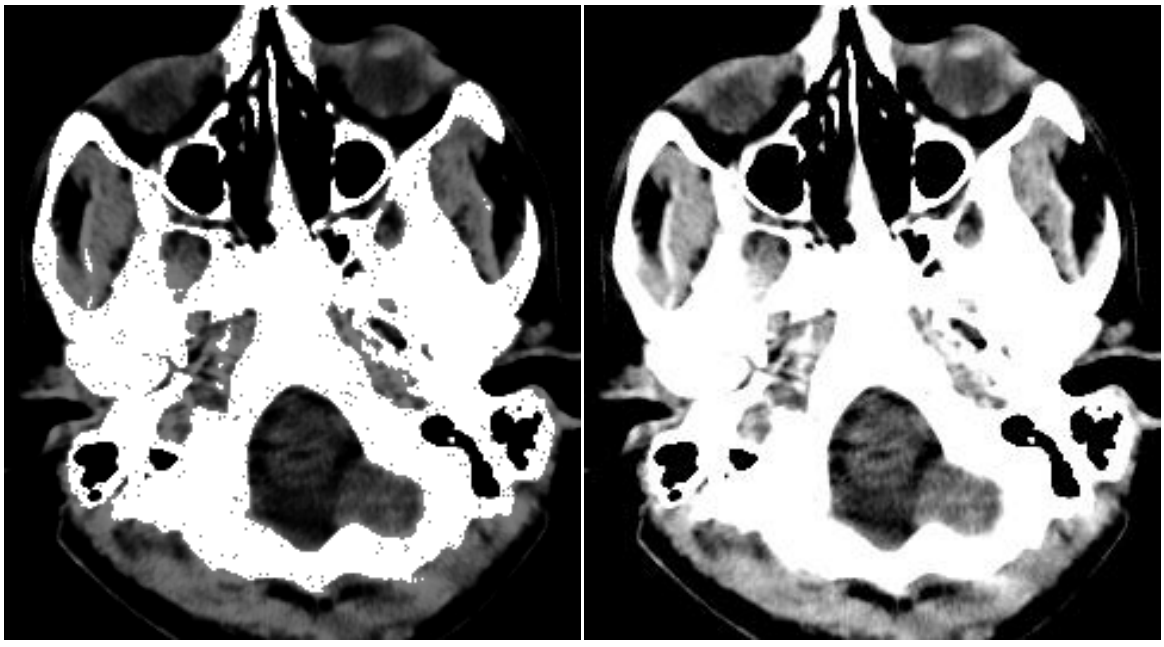
Appendix (B)
Medical images

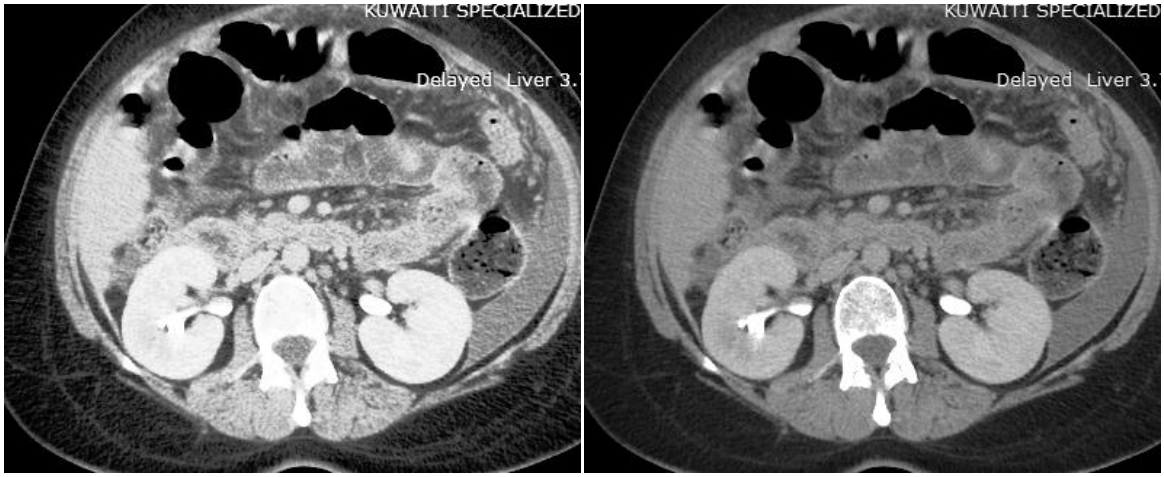





KUWAITI SPECIALIZED HO
0.625MM
KUWAITI SPECIALIZED HO
0.625MM
Complex of General Omer S
S-201612252
Complex of General Omer S
S-201612252


PNMS2
PNMS2




KUWAITI SPECIALIZED
Delayed Liver 3.
KUWAITI SPECIALIZED
Delayed Liver 3.