Sudan University of Science and Technology

College of Graduate Studies and Scientific Research

Optimization of Radiation Dose in Multislices CT scan Examinations

A thesis submitted in partial fulfillment for the requirements

Of Master degree in Diagnostic radiology Technology

By:

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Dedication

To my parents,

To anyone who taught

me a letter,

To my friends,

And to all my family.

Acknowledgements

First and foremost, I would like to express my deepest gratitude to Dr. Abdelmoneim Adam Mohamed Sulieman for his support and guidance. Without his help this work could not have been accomplished.

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ABSTRACT

Computed Tomography (CT) is a diagnostic imaging modality giving higher patient dose in comparison with other radiological procedures, so the calculation of patient dose in CT exams is very important. CT improved the diagnosis many of the diseases. The increasing use of CT in the Sudan in recent years is what has to think in the attempt to reduce the exposure of the patient and that the risks known to the X-ray. This study aimed to measure the radiation dose and estimating the risks resulting from exposure to X-rays during the imaging by CT scan.

A total of 130 patients were examined in two hospitals using two spiral CT scans 64 slices (Alamal National and Alzaytouna specialist hospitals in the period (March 2011-June 2011). The average age of the samples was 45±18 years. The mean effective dose for Al-amal National hospital was 17.4±12.7 mSv, 22.9±14.3 mSv and 2.4±0.9 mSv for the chest, abdomen and brain examinations, respectively. The mean effective doses for Al-Zaytouna specialist hospital were 26.3±7.8 mSv, 47.6±33.0 mSv, and 3.7±1.5 mSv for the chest, abdomen and brain, respectively.

The dose of this study is relatively higher compared to previous studies locally and internationally. This can be attributed to lack of training in CT dose optimisation and CT modality. The study showed the urgent need to review and evaluation of dose and also the need for continuous training of workers in this field and establishing the diagnostic reference level in the Sudan.

الملخص

الأشعة المقطعية هي تقنية تصوير طبي ،عند إستخدامها تعطي جرعة إشعاع عالية للمريض بالمقارنة مع الفحوصات الإشعاعية الأخرى، لذلك حساب الجرعات الإشعاعية للمرضى مهم جدا. كما أن الأشعة المقطعية زادت من المقدرة على تشخيص كثير من الأمراض. زيادة إستخدام الأشعة المقطعية في السودان في السنوات الأخيرة حفز الباحثين لخفض الجرعة الإشعاعية والخطر الإشعاعي المصاحب.

هدفت هذه الدراسة إلى قياس الإشعاع وتقدير الخطر الإشعاعي الناجم عن فحوصات الأشعة المقطعية للأجهزة ذات الأربعة وستون شريحة.

تم فحص 130 مريضا بكل من مستشفى الأمل الوطني و مستشفى الزيتونة التخصصي في الفترة من مارس وحتى يونيو 2011.

بلغ متوسط العمر للمرضى (18±45) سنة. وبلغ متوسط الجرعة الفعالة (ملي سيفرت) في مستشفى الأمل الوطني (0.9±2.4 and 14.3±22.9, 12.7±17.4) لكل من فحوصات الصدر والبطن والرأس على الترتيب، كما بلغ متوسط الجرعة الفعالة(ملي سيفرت) في مستشفى الزيتونة التخصصي (7.8±26.3 1.5±3.7, 33.0±47.6) لكل من فحوصات الصدر والبطن والرأس على الترتيب.

أظهرت هذه الدراسة أن المرضى يتعرضون لجرعة من الإشعاع أعلى من الدراسات السابقة محليا وعالميا. يعزى ذلك إلى قلة التدريب في مجال أمثلة الجرعة الإشعاعية للمرضى وكذلك لنوع الأجهزة. كما بينت الدراسة الحوجة الماسة لتدريب العاملين ووضع مستوى مرجعي للجرعة الإشعاعية بالسودان.

Chapter one

Introduction

Chapter One

Introduction

1.1 Historical background:

Computed Tomography (CT) technology and its clinical applications have shown enormous resilience against alternative diagnostic methods and at the moment is stronger than ever. Enabled by technology that provides high power x-ray tubes, magnificent computing power, multi channel detectors to give sub millimeter slices with wider scan coverage, faster rotation times to complete one rotation in one third of a second, all have moved CT to dynamic applications in cardiology and 3-dimensional imaging of vascular and musculoskeletal anatomy [ICRP 2006].

A mathematical description of the method for reconstruction of a 2-D image from projections was given by Johann Radon in 1917. Imaging instrumentation and computers involved to enabling researchers to apply the theory to the reconstruction of cross-section images for projections acquired from physical objects. In 1970s X-ray computed tomography (CT) was introduced for clinical use and immediately followed by extensive technological refinements. In 1970 A.M Cormack and Godfrey Hounsfield were jointly awarded the Nobel Prize for CT Invention [Edyean 1998, ICRP 2006]. Since then, CT has rapidly involved in terms of both technical performance and clinical use. Although initial experiences rapidly predicted widespread implementation of technique, it could hardly CT would become one of the most important of all X-ray procedures worldwide. Spiral CT and in particular the latest generation of the scanners with muti-slice capability in subsecond time frames have allowed improvement in speed of acquisition and image quality. This has resulted in highly reliable information about every part of the body, without motion artifacts from peristalsis and breathing. This consequence

has been further unexpected growth of the modality. Thus, completely new indications for CT are being reported, as well as completely new methods for performing and reading the studies. Twenty years ago, a standard CT examination of the thorax took several minutes to conduct, while total body similar information can be accumulated within a single breath hold period. This makes it more comfortable for patients examination, since the investigation is fast, well tolerated, accessible and the last not least, regarded as highly reliable in its outcome [Edyean 1998, Itoh et al. 2000], (Fig.1.1).

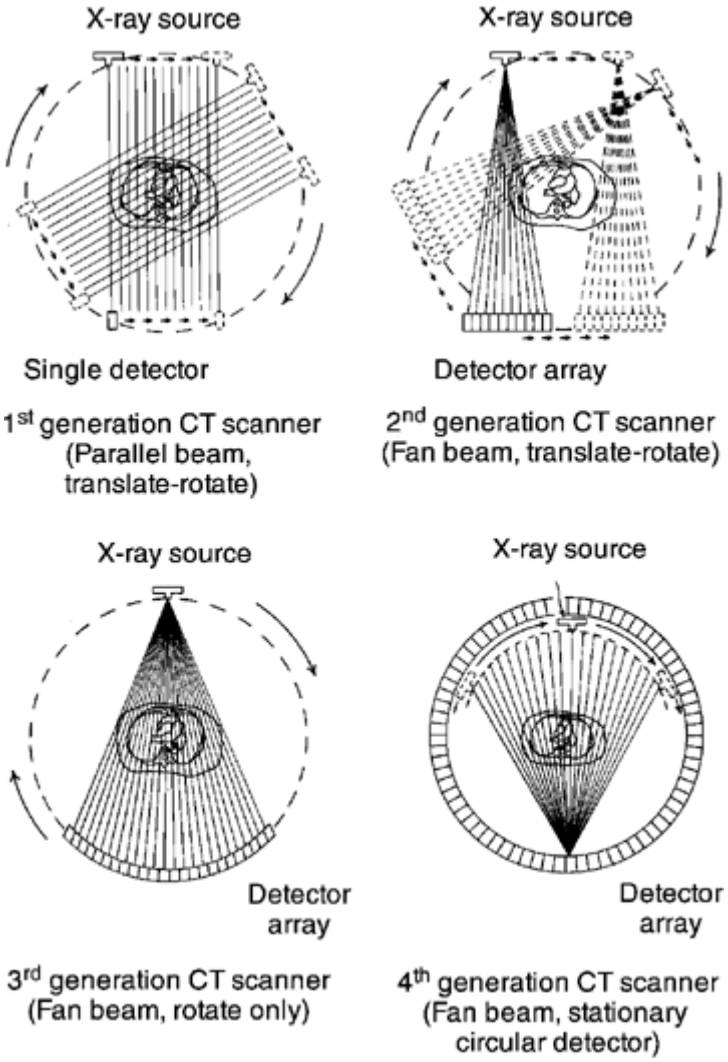


Figure 1.1 Generations of CT and tube motions [D.Tack 2007]

A number of terminologies are in use for this technology, namely multi-detector row computed tomography (MDCT), multi-detector CT (MDCT), multi-detector array helical CT, multi-channel CT and multi-slice CT (MSCT). The number of simultaneous but independent measurements along the patient long axis is often referred to as the number of "slices", and this value is commonly used to represent the technical capabilities of a system (e.g. 64-slice MDCT) [ICRP 2006].

In 2000, ICRP published a report on "Managing Patient Dose in Computed Tomography" [ICRP 2006]. At that time there was an urgent need to focus the attention of radiologists, physicians, medical physicists and other personnel involved in CT on the relatively higher effective doses to individual patients, increasing frequency of CT examinations, changes in clinical applications and the increasing contribution of CT to the collective dose. Further, the technology in use dominantly utilized a single row of detectors (SDCT), permitting scanning of only a single slice at a time in either a discrete (sequential acquisition) or continuous fashion (spiral acquisition). Multiple-detector rows along the z-axis (longitudinal axis of the patient, i.e. head to toe) permit simultaneous scanning of more than one slice. MDCT was in its infancy at the time of the 2000 report [ICRP 2006] and thus there was brief mention in the report of its impact on radiation dose. The concrete data and experience was insufficient to make any judgment. In the following years there has been a phenomenal increase in use of MDCT and technology has been advancing very rapidly to move from 4 slices to 8, 16, 32, 40 and 64-slice. Furthermore, dual source MDCT has been recently made available and 256-slice MDCT is expected to be released soon. The improved speed of MDCT scanning has also meant new applications (cardiac CT, whole body scanning) as well as improved patient throughput and workflow. In the last two decades, use of CT scanning has increased by more than 800% globally. In the United States, over the period of

1991 to 2002, a 19% growth per year in CT procedures has been documented. Also in the United States during this period, CT scanning for vascular indications has shown a 235% growth, followed by a 145% growth in cardiac applications. An increase has also been demonstrated in abdominal (25%), pelvic (27%), thoracic (26%) and head & neck (7%) applications. With 64-slice MDCT a further substantial increase is expected in cardiac applications. A 10% annual growth in the global CT market was reported in the year 2002 and this trend seems to continue [ICRP 2006].

1.2 Radiation risk:

The individual risk from radiation associated with a CT scan is quite small compared to the benefits that accurate diagnosis and treatment can provide. Still, unnecessary radiation exposure during medical procedures should be avoided. Unnecessary radiation may be delivered when CT scanner parameters are not appropriately adjusted for the patient size [Anne et al. 2001]. In conventional X-ray procedures, medical personnel can tell if the patient has been overexposed because of the film is overexposed, producing a dark image [ICRP 2006]. However, with CT there is no obvious evidence that the patient has been overexposed because the quality of the image may not be compromised. Several recent articles [Kalender et al. 1999, Rehani M, Berry M 2000, Rehani M 2000] stress that it is important to use the lowest radiation dose necessary to provide an image from which an accurate diagnosis can be made, and that significant dose reductions can be achieved without compromising clinical efficacy.

The United Nation Scientific Committee on the Effects of Atomic Radiation (UNSCEAR, 2000) has highlighted that the worldwide there about 93 million CT examinations performed annually at a rate of about 57 examinations per 1000 persons. UNSCEAR also estimated that CT

constitutes about 5% of all X-ray examinations worldwide while accounting for about 34% of the resultant collective dose. In the countries that were identified as having the highest levels of healthcare, the corresponding figures were 6% and 41% respectively.

1.3 Radiation dose optimization:

The main tools generally to achieve this aim are justification of practices optimization of protection. Optimization is even more important than in other practices using ionizing radiation [Edyean 1998]. Optimization means keeping the dose "As low as reasonably achievable, economic and social factors being taken into account', for diagnostic medical exposure this is interpreted as being a dose as low as possible, which is consistent with the required image quality [European Commission 1999], and necessary for obtaining the desired diagnostic information. It is inevitable that some complex cases will require a larger number of CT sections and multiple phases, but the disparity occurring between apparently similar applications is of serious concern. It is now widely accepted that un-optimized CT examination protocols are a significant contributor of unnecessary radiation dose. There appears to be much scope for dose optimization through use of appropriate protocols.

1.4 Statement of the Problem:

The motivations for this study are the relatively high radiation dose to the patient in CT examinations and the increasing frequency and variety of examinations. This study intends to provide technologists, radiologists and clinical staff with the means to successfully manage patient doses. Absorbed dose in tissues from CT are among the highest observed from diagnostic radiology (i.e. 10-100 mGy) [Edyean 1998, Hart D et al. 1994, Hounsfield GN 1973]. These

doses can often approach or exceed levels known to increase the probability of cancer due to the following factors:

The doses tissues from CT can often approach or exceed levels known to increase the probability of cancer, technologists are responsible for managing the dose in collaboration radiologists and medical physicists, CT examinations are increasing in frequency, newer CT techniques have often increased doses when compared with standard CT, referring physicians and radiologists should make sure that the examination is indicated and many practical possibilities currently exist to manage dose. The most important is reduction in mA.

1.5 Objectives of the study:

The general objective of the study is to reduce patient radiation dose during CT examinations without affecting the image quality.

The study intended specifically to:

Measure the radiation dose for patients during CT examinations, increase the benefits of CT examination compare with the risk of radiation, and estimate the effective dose for the patients undergoing CT examination.

1.6 Thesis outlines:

This thesis is concerned with the assessment of radiation dose for adult and pediatric patients during CT examinations.

Accordingly, it is divided into the following chapters:

Chapter one is the introduction to this thesis. This chapter presents the historical background and radiation risks, in addition to study problem, objectives and scope of the work. It also provides an outlines of the thesis.

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

Chapter three describes the materials and methods that used to measure dose for CT machines and explains in details the methods for calculation and optimization.

Chapter four presents the results of this study.

Finally *Chapter five* presents the discussion, conclusion and recommendations of this thesis and presents the suggestions for future work.

Chapter Two

Theoretical Background

Chapter Two

Theoretical Background

2.1 CT scan:

British engineer Godfrey Hounsfield of EMI laboratories in England invented CT in 1972. CT combined X-ray images with computer. A computer could put information from X-ray together to create across-sectional image. The CT scanner for clinical use was first installed in 1975. The original systems were dedicated to head scanning but whole body scanners with larger patient opening became available in 1976. The first CT scanner developed by Hounsfield in his laboratory at EMI took several hours to acquire the raw data for single scan (slice) and took days to reconstruct a single image from this raw data [Hounsfield GN 1973]. The latest multi-slice CT systems can collect up to 4 slices of data in about 350 ms reconstructs a 512*512 matrix image from million of data points in less than a second [Geleinjs J, et al 1994].

2.1.1 Principles of CT:

This part describes the principles and evolution of multi-slice CT (MSCT), including conceptual differences associated with slice definition, cone beam effects, helical pitch, and helical scan technique. MSCT radiation dosimetry is described, and dose tissues associated with MSCT-and with CT in general- as well as techniques for reducing patient radiation dose are discussed. Factors associated with the large volume of data associated with MSCT examinations are presented.

2.1.2 Principles of MSCT:

Soon after their introduction in the late 1980s, slip ring scanners and helical (spiral) CT were rapidly adopted and soon became the indisputable standard of care for body CT. However, a significant problem became evident: helical CT was very hard on X-ray tubes. For example, an abdomen-pelvis helical CT covering 60 cm (600mm) of anatomy with a 5 mm slice thickness, a pitch of 1.0 (thus requiring 120 rotations), and typical technique factors (120 kVp, 250 mAs, 1-s rotation time) deposits a total of 3.6x1,000,000 J of heat in the X-ray tube anode. Before slip ring CT, individual slice obtained with an equivalent technique (120 kVp, 250 mAs, 1-s scan time) would deposit only 30,000 J, much of which could be dissipated during the relatively lengthy (several seconds) interscan delay [Anne et al. 2001].

A limitation imposed by tube heating was that the thin slices (less than 3 mm) desired for acceptable-quality reformatting into off-axis (coronal, sagittal or oblique) or 3-dimentional reconstructions were impractical unless the scanned region was very limited or the scan technique was severely constrained. It was not uncommon for scanners to limit a helical technique with thin slices to 100 mAs (tube current in milli-amperes X-ray scan time in seconds) or less per rotation, yielding low quality, noisy images. A straightforward solution to this heat issue, of course, is to develop X-ray tubes with a higher heat capacity; such tubes have been developed. Another approach is to more effectively use the available X-ray beam: if the X-ray beam is widened in the z-direction (slice thickness) and if multiple rows of detectors are used, then data can be collected for more than one slice at time. This approach would reduce the total number of rotations and therefore the total usage of the X-ray tube needed to cover the desired anatomy. This is the basic idea of MSCT (other use the term multi-row CT and multi-detectors row CT (MDCT)). Although both third and forth generations scanners were in common use as

single slice scanners, all multi-slice scanners are based on a third generation platform. Therefore, in the following discussion, the third generation scanner geometry (tube and detector bank linked and rotating together) is assumed [Anne et al. 2001].

2.1.2.1 MSCT detectors:

The primary difference between single slice CT (SSCT) and MSCT hardware is in the design of the detector arrays, as illustrated in figure 2.1.

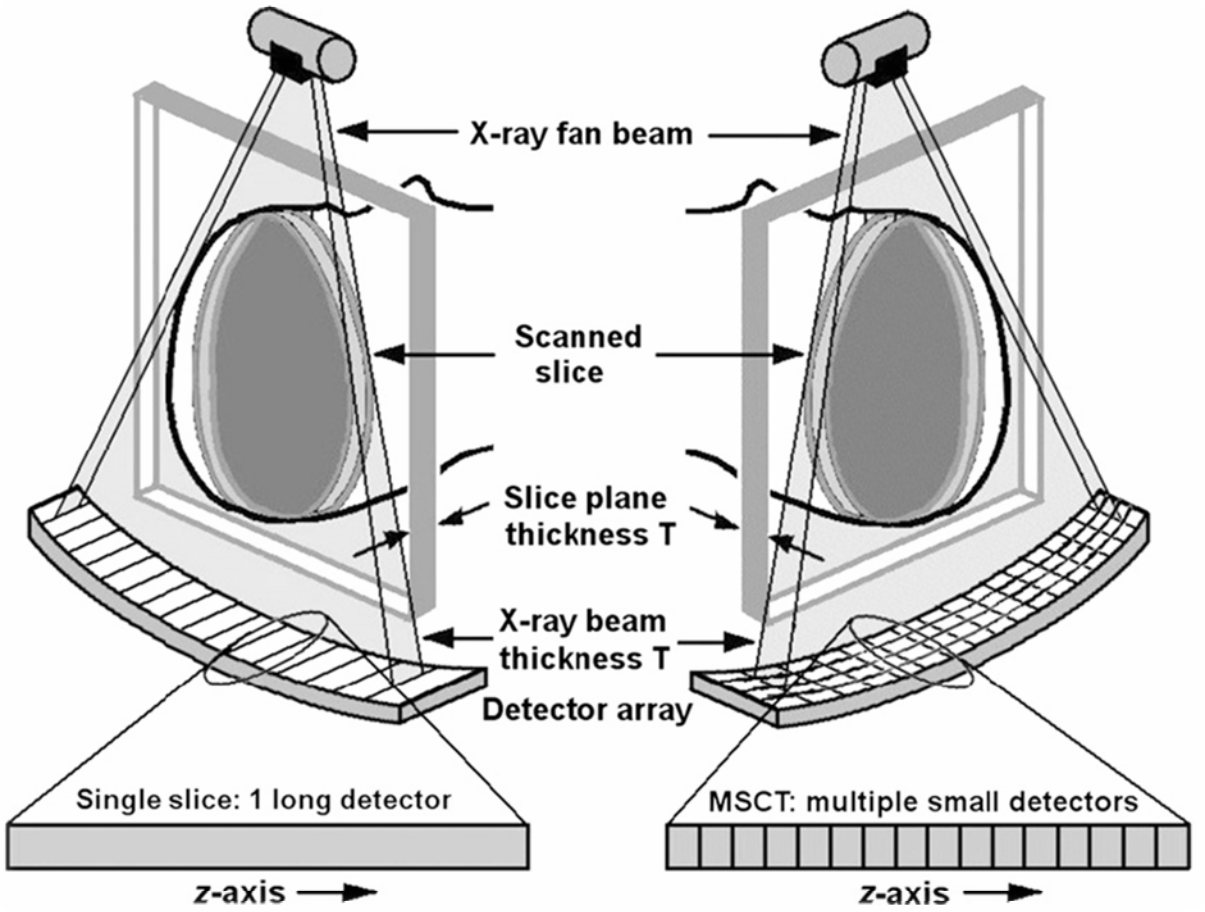


Fig.2.1 (Left) SSCT arrays containing single, long elements along z-axis. (Right) MSCT arrays with several rows of small detector elements [D.Tack 2007].

SSCT detector arrays are one dimensional (Fig.2.1); that is, they consist of a large number (typically 750 or more) of detector elements in a single row across the irradiated slice to intercept the X-ray fan beam. In the slice thickness direction (z- direction), the detectors are monolithic, that is, single elements long enough (typically about 20 mm) to intercept the entire X-ray beam width, including part of the penumbra (here, the term "X-ray beam width" always refers to the size of the X-ray beam along z-axis- that is, in the slice thickness direction). In MSCT, each of the individual, monolithic SSCT detector elements in the z- direction is divided into several smaller detector elements, forming a 2-dimensional array [Fig.2.1]. Rather than a single row of detectors encompassing the fan beam, there are now multiple, parallel rows detectors [D.Tack 2007].

2.1.2.2 MSCT Era:

The first scanner with more than one row of detectors and a widened z-axis X-ray beam was introduced by Elscint in 1992 (CT Twin). This scanner had two rows of detectors, allowing data for 2 slices to be acquired simultaneously, and was developed primarily to help the address the X-ray tube heating problem. As a curious historical note, according to the description given earlier in this article, the first generation EMI Mark 1. With 2 adjacent detectors and widened X-ray beam, this scanner collected data for 2 slices at the same time and thereby reduced the lengthy examination time associated with the 5 to 6 min scan time [Edyean 1998]. The first scanners of "the modern MSCT era" were introduced in late 1998 and are described in the following discussions [Edyean 1998].

2.1.2.3 MSCT data acquisition:

A detector design used in one of the first modern MSCT scanners [Fig.2.1] consisted of 16 rows of detector elements, each 1.25 mm long in the z- direction, for a total z-axis length of 20 mm. Each of the 16 detector rows could, in principle, simultaneously collect data for 16 slices, each 1.25 mm thick; however, this approach would require handling an enormous amount of data very quickly, because a typical scanner may acquire 1000 views per rotation. If there are 800 detectors per row and 16 rows, then almost 13 million measurements must be made during a single rotation with duration of as short as 0.5 s. Because of the initial limitations in acquiring and handling much large amount of data, the first versions of modern MSCT scanners limited simultaneous data acquisition to 4 slices. Four detector "rows" corresponding to the 4 simultaneously collected slices fed data into 4 parallel data "channels", so that these 4- slices scanners were said to possess 4 data channels. These 4- slices scanners, however, were quite flexible with regard to how detector rows could be configured; groups of detector elements in the z- direction could be electronically linked to function as single, longer detector, thus providing much flexibility in the slice thickness of the 4 acquired slice. Examples of detector configurations used with the 4 channels are illustrated in (Fig. 2) for 2 versions of 4-slice MSCT detectors: one based on the detector design described earlier (16 rows of 1.25 mm elements) and the other based on an "adaptive array" consisting of detector elements of different sizes (other detector design were used by other manufacturers) [ICRP 2006, Itoh et al. 2000, Galanski et al. 1999]. Possible detector configurations for detector design encompassing 16 rows of 1.25 mm elements for the acquisition of slices are illustrated in figure 2.2 A and figure 2.2 B. In (Fig. 2.2 A, 4 elements in a group are linked to act as a single 5mm detector (4 x1.25). The result is four 5mm detectors covering a total z-axis length of 20 mm. When a 20 mm wide X-ray beam is used,

4 slices with a thickness of 5 mm are acquired. The acquired 5 mm slices can also be combined into 10 mm slices, if desired. In figure 2 B, 4 pairs of detectors elements are linked to function as four 2.5 mm detectors (2xs1.25). When a 10 mm wide X-ray beam is used, four 2.5 mm slices can be acquired simultaneously. Again, the resulting 2.5 mm slices can be combined to form 5 mm slices (5 mm axial slices generally preferred for interpretation purpose). A third possibility is to use a 5 mm wide X-ray beam to irradiate only the 4 innermost individual detector elements for the acquisition of four 2.5 mm slices. Yet another possibility is to link coverage and tube heating limitations. Sub millimeter scanning had to await the introduction of 16 slices scanners [D.Tack 2007].

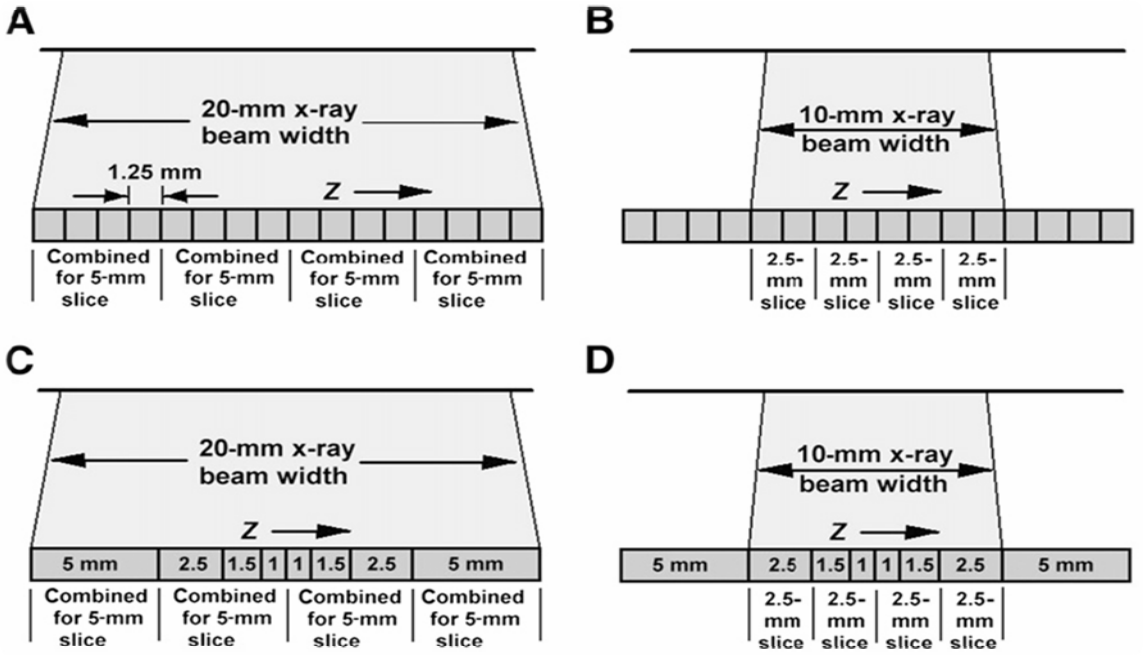


Figure: 2.2 Flexible uses of detectors in 4 slice MSCT scanners. (A) Groups of fours 1.25 mm wide elements are linked to act as 5 mm detectors. (B) Inner 8 elements are linked in pairs to act as 2.5 mm detectors. (C) Inner, adaptive array are linked to act as 5 mm detectors (1 1 1.5 1 2.5) and, together with outer, 5 mm elements, yield four 5 mm slices. (D) The 4 innermost elements are linked in pairs to form 2.5 mm detectors (1 1 1.5), which along with two 2.5 mm detectors, collect data for four 2.5 mm slices [D.Tack 2007].

2.1.3 16- Channels (16-slices) scanners and more:

The installation of MSCT scanners providing 16 data channels for 16 simultaneously acquired slices began in 2002. In addition to simultaneously acquiring up to 16 slices, the detector arrays associated with 16-slice scanners were redesigned to allow thinner slices to be obtained as well. Detector arrays for various 16-slice scanners models are illustrated in figure 3. Note that in all of the models, the innermost 16 detectors elements along the z-axis are half the size of innermost elements, allowing the simultaneous acquisition of 16 thin slices (from 0.5 mm thickness to 0.75 mm thickness, depending on the model). When the inner detectors were used to acquire sub millimeter slices, the total acquired z-axis length and therefore the total width of the X-ray beam ranged from 8 mm for the Toshiba version to 12 mm for the Philips and Siemens versions. Alternatively, the inner 16 elements could be linked in pairs for the acquisition of 16 thicker slices [D.Tack 2007].

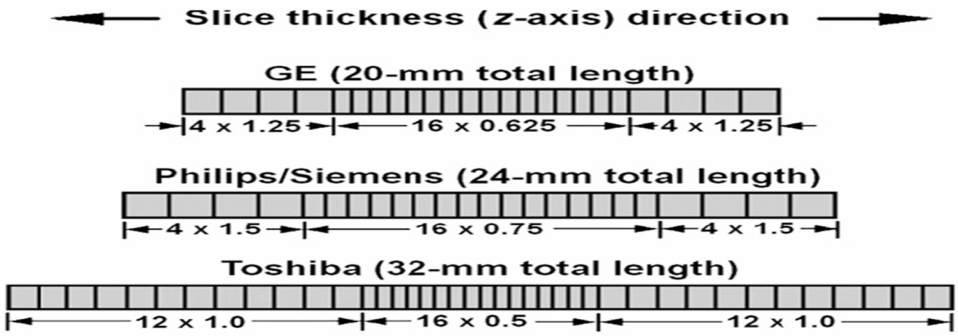


Figure: 2.3 Diagrams of various 16-slices detector designs (in z-direction). Innermost elements can be used to collect 16 thin slices or linked in pairs to collect 16 thicker slices [D.Tack 2007].

During 2003 and 2004, MSCT manufacturers introduced models with both fewer than and more than 16 channels. 6-slices and 8- slices models were introduced by manufacturers are cost effective alternatives. At the same time, 32-64- slices scanners were announced, and installations by most manufacturers began [D.Tack 2007]. Detector array designs used by several manufacturers are illustrated in figure 2.4.

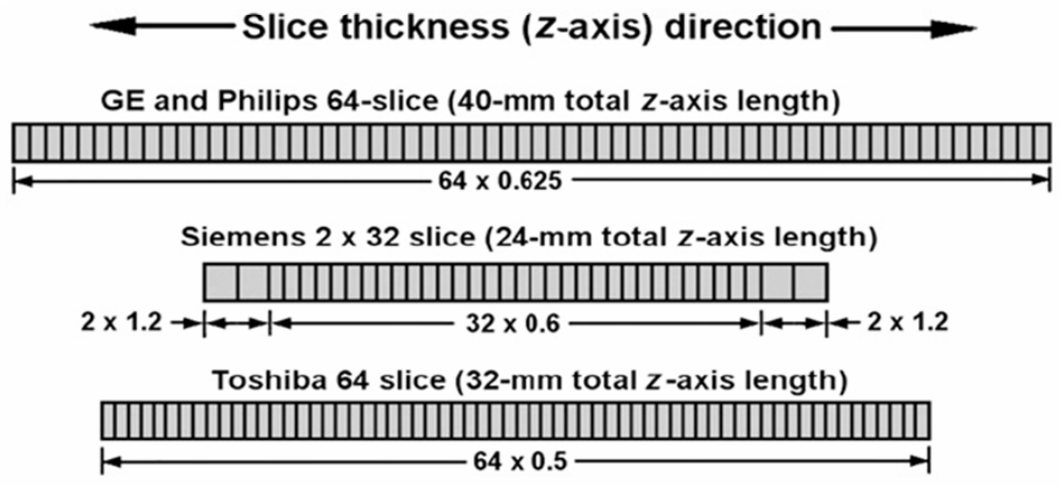


Figure: 2.4. Diagram of various 64-slice detector designs (z-direction). Most designs lengthen the arrays and provide all sub millimeter elements. Siemens scanner uses 32 elements and dynamic-focus X-ray tube to yield measurements per detector [D.Tack 2007].

The approach used by manufacturers for 64-slice detector arrays designs to lengthen the arrays in the z-direction and provide all sub millimeter elements: 64x0.625 mm (total z-axis length of 40 mm) for the Philips and GE Healthcare models and 64x0.5 mm (total z-axis length of 32 mm) for the Toshiba model. The design approach of Siemens was quite different. The detector arrays of Siemens 32-slice scanner (containing 32 elements, each element 0.6 mm long for a total z-axis length of 19.2 mm) was combined with a “dynamic-focus” X-ray tube for the simultaneous acquisition of 64-slices. This X-ray tube could electronically and very quickly shift the focal spot

location on the X-ray tube target so as to emit radiation from a slightly different position along the z-axis. Each of the 32 detector elements then collected 2 measurements (samples), separated along the z-axis by approximately 0.3 mm. The net result was a total of 64 measurements (32 detectors, 2 measurements per detector) along a 19.2 mm total z-axis field of view (this process is referred to in Siemens literature as "Z-sharp" technology) [D.Tack 2007].In the preceding examples, in addition to simultaneous acquisition of more slices, MSCT X-ray beam width can be considerably wider than those for SSCT. Sixteen-slice MSCT beam widths are up to 32 mm; 64 –slice beams can be up to 40 mm wide; and even wider beams are used in systems currently under development or in clinical evaluation. A possible consequence is that more scatter may reach the detectors, compromising low contrast detection. Generally, however, an anti- scatter septa traditionally used with 3rd generation CT scanners can be made sufficiently deep to remain effective with MSCT. An example of a section of a 16-slice detector with the associated scatter removal septa is shown in (Fig.2.5) [D.Tack 2007].

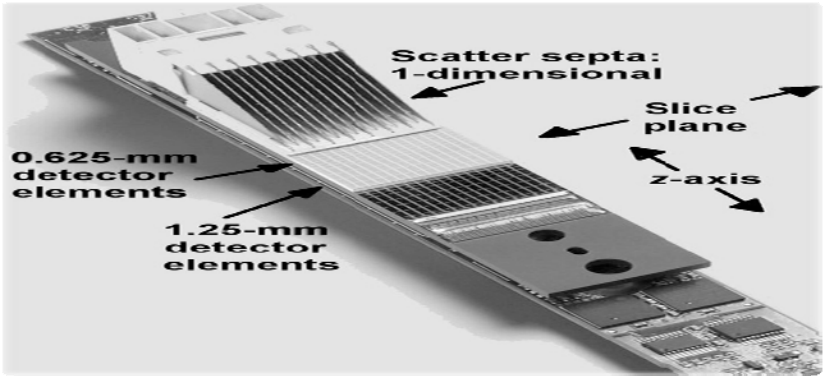


Figure: 2.5. Section of a 16-slice detector with scatter removal septa. Septa are sufficiently deep to eliminate nearly all scatter. Note smaller elements (0.625 mm, in this example) in center of array and larger (1.25 mm) outer elements. Also note dead spaces (lighter lines) between elements [D.Tack 2007].

2.1.4 MSCT concepts: Differences between MSCT and SSCT

Before the further development of MSCT technology is described, certain concepts that are associated with MSCT and that may differ fundamentally from those associated with SSCT are addressed. One of these concepts is the relationship between slice thickness and X-ray beam width. Another involves the notion of cone beam effects [D.Tack 2007, IEC 1999].

2.2 Optimization of MDCT technique:

Once the clinical induction of CT is well established, the appropriate CT technique is the required on order to optimize the image quality with the lowest possible radiation dose. What can be easy modified and adapted by the operator performing the examination, as general rule, it should be noted that the use of standardized and fixed acquisition parameters lead to necessary over exposure of patient.

2.2.1 CT parameters:

2.2.1.1 Tube potential (kVp):

The relationship between the dose and the tube potential is not straight and linear one tube potential is usually modified only through the (kVp) setting. These kVp values differ from one manufacturer to another, as well as from one CT scanner to another, and vary from 80 kVp to 140 kVp, as the effect of increasing tube potential has a huge influence on radiation dose.

A general rule for selecting kVp could be the following [ICRU 1976]:

- To avoid 140 kVp except for CT of the chest, abdomen, and pelvis in extremely obese patient (Body mass index (BMI) is greater than 35 kg/m2) and for CT of the Lumbo-sacral (L/S) in obese patient (BMI more than 30 kg/m2).

- Prefer 100 - 110 kVp for CT chest, abdomen, and pelvis in this patient (i.e. with BMI less than 22 kg/m2) and in 10 -15 years old for children.

- Prefer 80 - 90 kVp for CT angiography and in children younger than 10 years old.

2.2.1.2 Tube current – Time product (mAs):

As in conventional radiography, a straight linear exists between the mAs and the dose. The setting for mAs should be adapted to the characteristics of the scanner unit, the patient's size, and the dose requirements for each type of examination.

Appropriate use of mAs also depends on the patient's size, which is an important parameter to consider in dose optimization. In order to avoid unnecessary over exposure, mAs should be internationally adapted by the operator unless automatic exposure control (AEC) devices, or similar, are available [ICRU 1976].

General rule; mAs setting may be halved when the patient's trunk diameter – typically 30 cm decreases by 4 cm without loss of image quality. If CT scanner is not equipped with (AEC) following this rule.

2.3 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 a 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 likelihood of benefit from the examination and the potential risk from radiation [ICRP 1990, ICRP 1991].

2.3.1 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 (W_T) values. For a specific organ or body area the effective dose is:

$$\text{Effective Dose (Gy)} = \text{Absorbed Dose (Gy)} \times W_T$$

If more than one area has been exposed, then the total body effective dose is just the 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 [Perry Sprawls.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 mrad 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.

Table: 2.1 Tissue Weighting Factors [UNSCEAR 2008]:

Weighting factors for different organs			
Organs	Tissue weighting factors		
	ICRP30(I36) 1979	ICRP60(I3) 1991	ICRP103(I6) 2008
Gonads	0.25	0.20	0.08
Red Bone Marrow	0.12	0.12	0.12
Colon	-	0.12	0.12
Lung	0.12	0.12	0.12
Stomach	-	0.12	0.12
Breasts	0.15	0.05	0.12
Bladder	-	0.05	0.04
Liver	-	0.05	0.04
Oesophagus	-	0.05	0.04
Thyroid	0.03	0.05	0.04
Skin	-	0.01	0.01
Bone surface	0.03	0.01	0.01
Salivary glands	-	-	0.01
Brain	-	-	0.01
Remainder of body	0.30	0.05	0.12

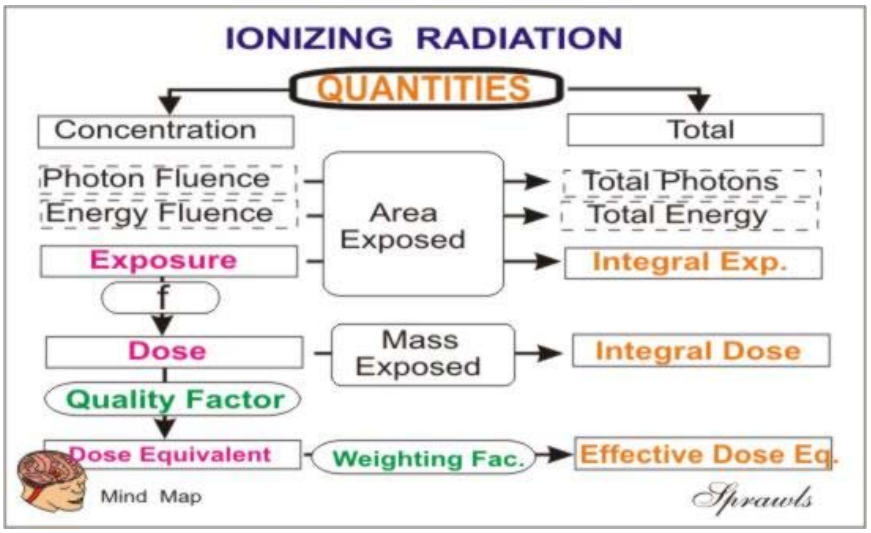


Figure: 2.6. Radiation quantities and units [ICRP 1990, ICRP 1991]

2.4 CT dose measurements:

Although CT presents only a small percentage of radiology examinations, it results in a significant portion of the effective radiation dose from medical procedures; (I) with the increasing use of CT for screening procedures, (II) and advances in scanner technology, they tend for increasing numbers of procedures performed with this modality may increase. Although CT is clearly providing many clinical benefits, the motivation to understand radiation dose in general as well as the specific concepts related to CT grows with prevalence of this modality [ImPACT 2007, Jones et al. 1993].

2.4.1 CT parameters that influence the radiation dose:

The radiation exposure to the patients undergoing CT examinations is determined by two factors: equipment-related factors, .e. the design of the scanner with respect to dose efficiency, and applications-related factors, i.e. the way in which the radiologist and X-ray technologist makes

use of the scanner [Nagel 2007]. In this chapter the features and parameters influencing patient dose are outlined. First, however, a brief introduction on the dose descriptors applicable to CT is given [Nagel 2007].

2.4.2 CT dose descriptors:

The dose qualities used in this projection radiography are not applicable to CT for three reasons [ImPACT 2007, Jones et al. 1993]:

First, the dose distribution inside the patient is completely different from that of a conventional radiography where the dose decreases continuously from entrance of the X-ray beam to its exit, with the ratio of between 100 and 1000 to 1. In the case of CT, as a consequence of the scanning procedure that equally irradiates the patient from all directions; the dose is almost equally distribution in the scanning plane. A dose comparison of CT with conventional projection radiography in term of skin dose therefore does not make any sense.

Second, the scan procedure using narrow beams along the longitudinal z-axis of the patient implies that a significant portion of the radiation energy is deposited outside the nominal beam width. This is mainly due to penumbra effects and scattered radiation produced inside the beam.

Third, the situation with CT is further complicated by the circumstances in which-unlike in conventional projection radiography-the volume to be imaged is not irradiated simultaneously. This often leads to confusion about what dose from a complete series of e.g. 15 slices might be compared with the dose from a single slice [ImPACT 2007, Jones et al. 1993].

As a consequence, dedicated dose quantities that account for these peculiarities are needed. The ‘Computed Tomography Dose Index (CTDI)’, which is a measure of the local dose, and the

Dose Length Product (DLP), representing the integral radiation exposure associated with a CT examination. Fortunately, a bridge exists that enables to compare CT with radiation exposure from the other modalities and sources; this can be achieved by the effective dose (E). So there are three dose descriptors in all, which everyone dealing with CT should be familiar with [Nagel 2007].

2.4.2 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.2.7) illustrates the meaning of the term: CTDI is the equivalent of the dose value inside the irradiated slice (beam), that would result if the absorbed radiation dose profile were entirely concentrated to a rectangular of width equal to the nominal beam width with N being the number of independent (i.e. non-overlapping) slices that are acquired simultaneously. Accordingly, all dose contributions from outside the nominal beam width, i.e. the areas under the tails of the dose profile, are added to the area inside the slice [Nagel 2007].

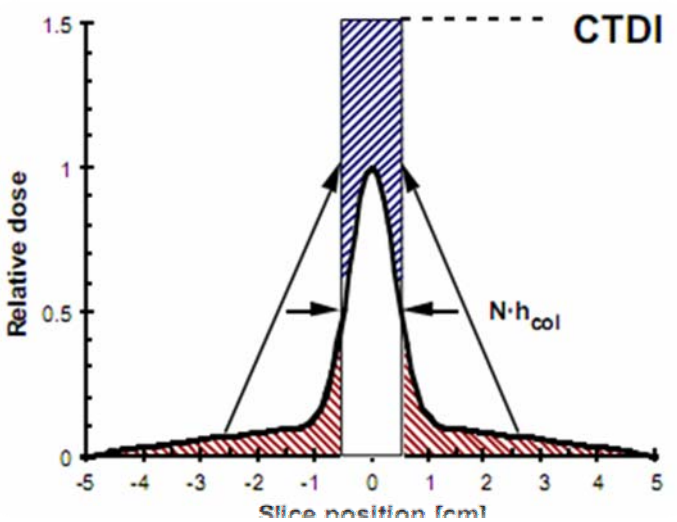


Figure: 2.7: Illustration of term 'Computed Tomography Dose Index (CTDI)': is the equivalent of the dose value inside the irradiated slice (beam) that would result if the absorbed radiation dose profile were entirely concentrated to a rectangular of width equal to the nominal beam width N.hcol, with N being the number of independent (i.e. non-overlapping) slices that are acquired simultaneously [Nagel 2007].

The corresponding mathematical definition of CTDI therefore describes the summation of all dose contributions along the z-axis:

$$CTDI = \frac{1}{N \cdot h_{col}} \cdot \int_{-\infty}^{+\infty} D(z) \cdot dz$$

Equation (2.1)

Where D(z) is the value of the dose at a given location, z, and N.hcol is the nominal value of the total collimation (beam width) that is used for data acquisition. CTDI is therefore equal to the

area of the dose profile (the 'dose-profile integral') divided by the nominal beam width. In practice, the dose profile is accumulated in a range of -50 mm to +50 mm relative to the centre of the beam, i.e. over a distance of 100mm.

The relevancy of CTDI becomes obvious from the total dose profile of a scan series with e.g. n=15 subsequent rotations (Fig.2.8). The average level of the total dose profile, which is called 'Multiple Scans Average Dose (MSAD)'- [Shope 1981], is higher than the peak value of each single dose profile. This increase results from the tails of the single dose profiles. Obviously MSAD and CTDI are exactly equal of the table feed (TF) is equal to the nominal beam width N.hcol, i.e. if the pitch factor

$$p = \frac{TF}{N \cdot h_{col}}$$

Equation (2.2)

is equal to 1. In general (i.e. if the pitch factor is not equal to 1, Fig.2.9), the relationship between CTDI and MSAD is given by:

$$MSAD = {}^{1}\!/_{p} \cdot CTDI$$

Equation (2.3)

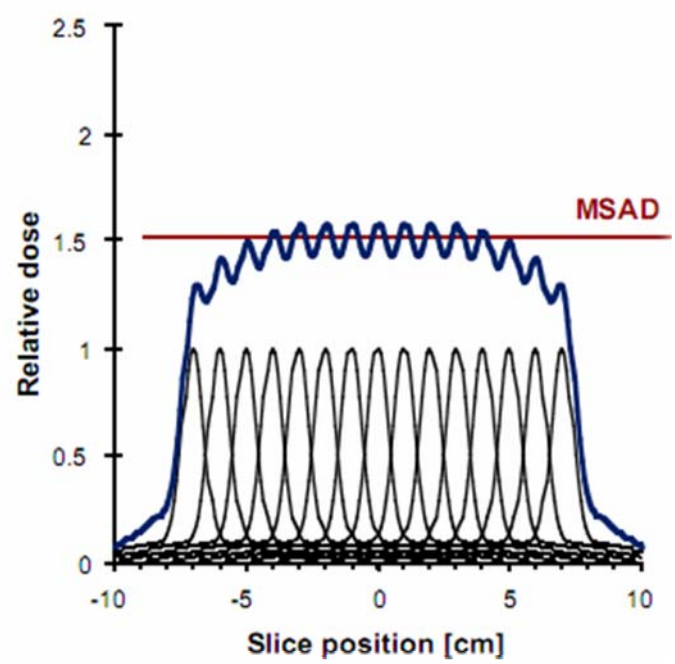


Figure: 2.8: The average level of the total dose profile, which is called 'Multiple Scans Average Dose (MSAD)'- [Shope 1981], is higher than the peak value of each single dose profile. This increase results from the tails of the single dose profiles [Nagel 2007].

Each pair of CTDI (central and peripheral) can be combined into a single are named weighted CTDI (CTDIw):

$$CTDI_w = \frac{1}{3}CTDI_{100}^{central} + \frac{2}{3}CTDI_{100}^{peripheral}. \quad \text{Equation (2.4)}$$

If pitch-related effects on radiation exposure are taken into account at level of local dose (i.e. CTDI) already, a quantity named volume CTDI (CTDIvol)' is defined [IEC 2001]:

$$CTDIvol = CTDIw/P \quad \text{Equation (2.5)}$$

So CTDIvol is the pitch-corrected CTDIw. Apart from the integration length, which is limited to 100 mm, CTDIvol is practically identical to MSAD based on CTDIw (i.e.MSADw). Since averaging includes both the cross section and the scan length, CTDIvol therefore represents the average dose for a given scan volume. CTDIvol is used as the dose quantity that is displayed at the operator's console of newer scanners [Nagel 2007].

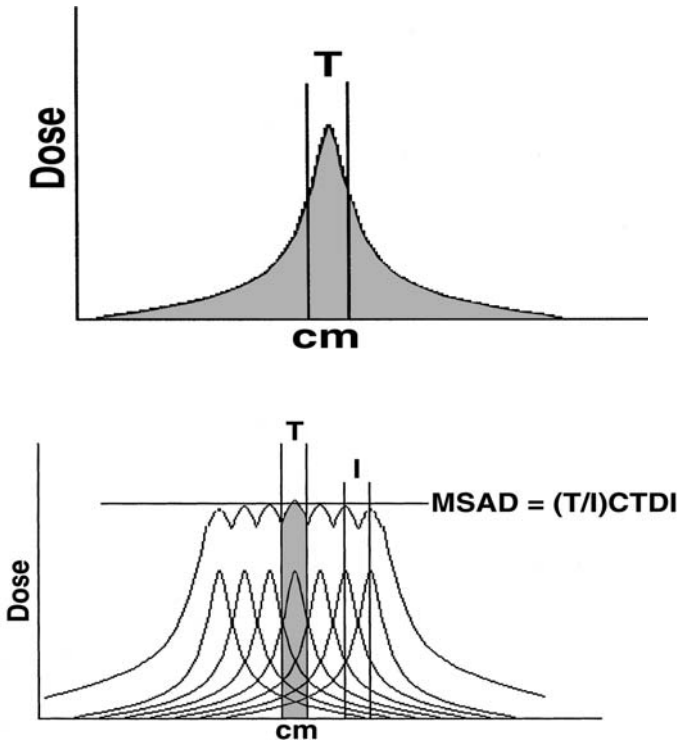


Figure: 2.9: (1) Schematic illustrates the profile of radiation dose delivered during a single CT scan. The CTDI equals the shaded area under the curve divided by the section thickness (*T*). (**2)** Schematic illustrates the profile of radiation dose delivered during multiple CT scans. *T* represents section thickness, and *I* represent the interval between sections. The MSAD includes the contributions of neighboring sections to the dose of the section of interest [D.Tack 2007].

2.4.4 Dose length product (DLP) unit (mGy):

DLP = CTDIw**.** L (mGy-cm). DLP takes both the 'intensity') represented by CTDIvol) and the extension (represented by scan length L) of an irradiation into account (Fig.2.7).

$$DLP = CTDI_{vol} \cdot L$$

Equation **(**2.6)

So DLP increases with number of slices (correctly: with length of irradiated body section), while the dose (i.e. CTDIvol) remains the same regardless of the number of slices or length, respectively. In fig. 2.7, the area of the total dose profile of the scan series represents the DLP. DLP is the equivalent of the dose-area product (DAP) in projection radiography, a quantity that also combines both aspects (intensity and extension) of patient exposure. In sequential scanning, the scan length is determined by the beam width N.hcol and number of the table feed (TF):

$$L = n*TF + N.hcol$$ Equation **(**2.7)

While in spiral scanning the scan length only depends on the number (n) of rotations and the table feed (TF):

$$L = n \cdot TF = {}^{T}\!/_{t_{rot}} \cdot p \cdot N \cdot h_{col}$$

Equation **(**2.8)

Where T is the total scan time, trot is the rotation time, and p is the pitch factor. While in sequential scanning the scan length L is equal to the range from the begin of the first slice till the end of the last, the (gross) scan length for spiral scanning not only comprises the (net) length of the imaged body section but also includes the additional rotations at the begin and the end of the

scan (‘over-ranging’) that are required for data interpolation [European Commission 1999]. If an examination consists of several sequential scan series or spiral scans, the dose-length product of the complete examination (DLP exam) is the sum of the dose-length products of each single series or spiral scan:

$$DLP_{exam} = \sum_{i} DLP_i$$

Equation (2.9)

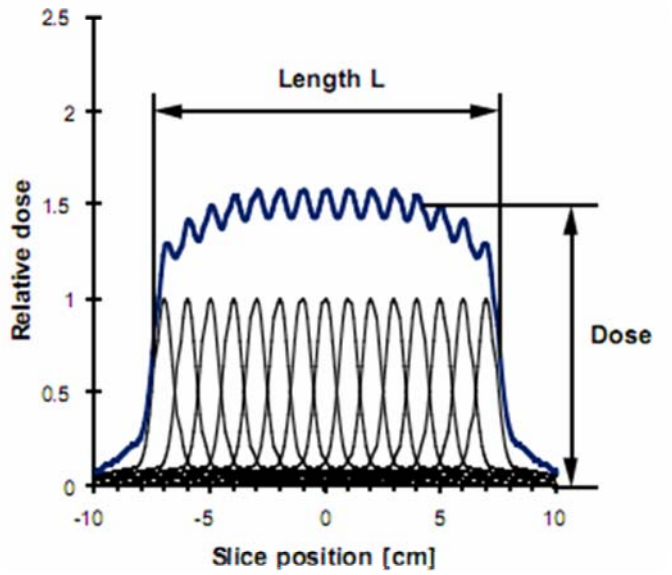


Figure 2.10: Dose length product (DLP) in CT (Total dose profile of a scan series with n=15 sub-sequent rotations. The dose-length product (DLP) is the product of the height (dose, i.e. CTDIvol) and the width (scan length L) of the total dose profile and is equal to the area under the curve [Nagel 2007].

2.4.5 Effective Dose:

CTDI and DLP are CT specific dose descriptors that do not allow for comparisons with radiation exposure from other sources, projection radiography, nuclear medicine or natural background radiation. The only common denominator to achieve this goal is the (Effective Dose). With effective dose, the organ doses from a partial radiation of the body are converted into an equivalent uniform dose to the entire body. An effective Dose E unit (millisevert, mSv) according to ICRP 60 [ImPACT 2007] is defined as the weighted average of organ dose values HT for a number of specific organs:

$$E = \sum i \; Wi^* \; HTi \qquad \text{Equation (2.10)}$$

2.5 Equipment – related factors:

2.5.1 Beam filtration:

In conventional projection radiography, beam filtration is a well-known means to reduce those portions of the radiation spectrum with no or little contribution to image formation. In the early years of the CT history, beam filtration was comparatively large in order to compensate for beam hardening artifacts [Nagel 1989]. The present generation of scanners typically employs a beam filtration for the X-ray tube assembly of between 1 mm and 3 mm Al and an additional filtration (flat filter) of 0.1 mm Cu, given a total beam filtration of between 5 to 6 mm Al.

Newer surveys on CT practice [ICRP 1991] revealed that scanners of comparable age but with largely differing beam filtration are operated at almost similar dose levels. Similar results in terms of dose efficiency have been found in comparative tests on scanners with differing beam filtration conducted by Imaging Performance Assessments of CT [IMPACT 2007]. Beam

filtration plays only a minor role in CT.

Contrary to projection radiography, this operates at comparatively lower tube potentials; beam filtration plays only a minor role in CT where higher tube potentials are applied. A return to increased beam filtration - as sometimes recommended or practiced - is less advantageous than expected and should only be made if sufficient X-ray tube loading capacity is available or if other important aspects exist (e.g. improved performance of reconstruction filters).

Apart from this, there are a number of older and also newer scanners which operate with an added filtration of approximately 0.2 mm Cu, resulting in a total beam filtration of between 8 and 9 mm Al, and sometimes even more (currently up to 12 mm Al quality-equivalent filtration). Likewise, there are also scanners that employ less filtration. Consequently, the normalized dose values for these scanners (nCTDI in terms of mGy/mAs) differ significantly. Very often these lower or higher values are misunderstood as being an indicator that the equipment is more or less dose-efficient compared with other scanners. This might not necessarily be the case in reality [ImPACT 2007].

2.5.2 Beam shaper:

Most scanners are equipped with a dedicated filter device named ‘beam shaper’ or ‘bow tie filter’, which modifies the spatial distribution of radiation emitted within the fan beam. The purpose of this kind of filter (which is characterized by increasing thickness towards its outer edges) is to adapt the beam intensity to match the reduced attenuation of objects in outer portions of the fan beam.

Beam shapers preferentially affect the dose in outer portions of object, thereby reducing the peripheral CTDIp values. But as the dose at the centre is mainly caused by scattered radiation

from the periphery of the object, the central CTDIc value is also somewhat reduced. The ratio of dose at the periphery to the dose at the centre therefore decreases, making the dose distribution inside an object more homogeneous and so improving the uniformity of noise in the image. In addition, different types of beam shapers can selected on some scanners depending on the nature and diameter of the object (e.g. for head and body scanning mode) [Tack 2007].

2.5.3 Beam collimation:

The beam collimation for defining the thickness of the slice to be imaged is made in the first instance close to the X-ray source (Primary collimation). The shape of the dose profile is determined by the aperture of the collimator, its distance from the focal spot, and the size and shape (i.e. the intensity distribution) of the focal spot. Due to the narrow width of collimation, penumbral effects occur. These effects become more and more pronounced as collimation is further narrowed.

The primary collimator and secondary are which is close to the detector (Post patient collimation). In addition, there is a secondary collimation close to the detector ('post-patient collimation') that primarily serves to remove scattered radiation. The primary serves to remove scattered radiation. In some SSCT and dual scanners, this secondary collimation is further narrowed in order to improve the shape of slice profile. On some single-slice and dual-slice scanners this secondary collimation is further narrowed in order to improve the shape of the slice profile ('restrictive post-patient collimation', see Fig. 2.12a, b). For multi-slice scanners with more than two detector rows, the primary collimation must necessarily be made wider than N times the selected slice collimation in order to avoid (or at least to reduce) penumbral effects in the outer portions of the detector array (Fig. 2.12c). In both cases, the dose profile is wider than the slice profile or the nominal beam width, and the patient is exposed to a larger extent

('overbeaming'), as becomes obvious from normalized CTDI values that increase with reduced beam width. Overbeaming can be expressed by a single parameter, the 'overbeaming parameter' dz that is equal to the combined width of the portion of the dose profile that is not used for detection (Fig. 2.12c). Overbeaming itself, i.e. the percentage increase in CTDI due to the unused portion of the dose profile, is then given by:

$$\Delta CTDI_{rel} = \frac{dz}{N \cdot h_{col}} \cdot 100$$

Equation (2.11)

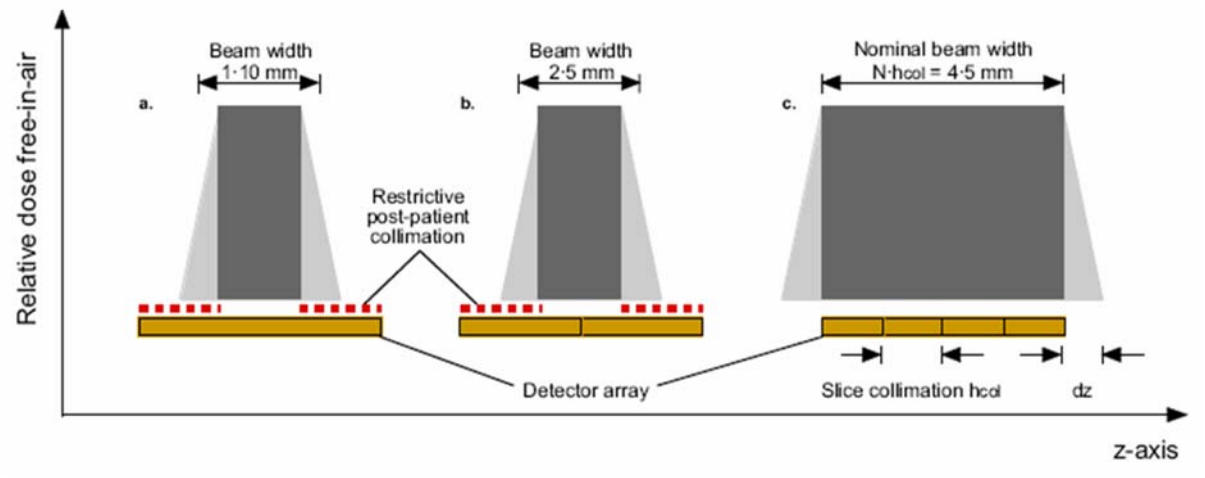


Figure: 2.11: Dose profile in free air umbra (dark grey) and penumbra (light grey) portions for single slice scanner (a), dual slice scanner (b), and quad slice scanner (c)(Dose profiles free-in-air with umbra (dark grey) and penumbra (light grey) portions for a single-slice scanner (a.), a dual-slice scanner (b.), and a quad-slice scanner (c.). With single- and dual-slice scanners, the width of the active detector rows is sufficient to capture the entire dose profile, penumbra included (except for some scanners which employ restrictive post-patient collimation). For scanners with four and more slices acquired simultaneously, penumbra is excluded from detection in order to serve all detector channels equally well. The combined width of the penumbra triangles at both sides is characterized by the overbeaming parameter dz.) [Vander et al. 1998].

In practice, overbeaming is not real issue for single and dual scanners, as the limited coverage restricts the use of narrow beam width to a few examinations with a short scan range (e.g. inner ear). With multi-slice scanners, however, overbeaming effects have to be taken seriously, as MSCT technology aims to provide improved resolution along the z-axis, which requires reduced slice collimation. Overbeaming, i.e. the increase in CTDI that results from beam width setting that is typical for each type of scanner is shown in (Fig. 3.9) for a number of scanners from different manufacturers. As indicated by the trend line, overbeaming is most pronounced with quad-slice scanners and is diminished with an increasing beam width N·hcol provided by scanners with more slices [Vander et al. 1998].

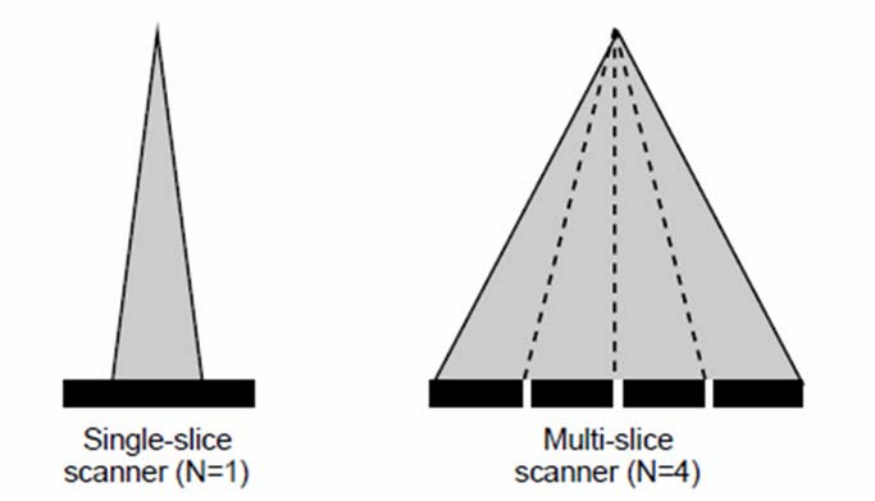


Figure: 2.12: MSCT scanner, with simultaneous scanning of four slices, compared with a conventional single-slice scanner. Due to the additional septa between the detector rows, the geometric efficiency of MSCT detector arrays is comparatively lower by 10 to 20% [Nagel 2007].

2.5.4 Detector array:

In general, solid-state detectors are more dose-efficient than gas detectors, but require additional means to suppress scattered radiation (anti-scatter-grids) that inevitably cause a certain loss of primary radiation, too [Vlassenbroek 2004]. Each detector design had its specific advantages and drawbacks: separating strips and decreasing sensitivity. All 16-slice scanners introduced in 2001, now made use of the same hybrid design, with 16 smaller central detectors, accompanied by number of larger detectors at both sides. Apart from the number of detector rows (between 24 and 40) and array width (between 20 and 32 mm), there were differences in the size of the detectors (between 0.5 and 1.5 mm), and each manufacturer claimed his solution to be the best one. As in real life, there are a number of conflicting needs (spatial resolution, dose efficiency, coverage) that must be met, especially with respect to cardiac imaging where scan times below 20 s (one breath hold) are mandatory. Consequently, designs, which put emphasis to a single one of these criteria, only were definitely not the best compromise. Due to the increased number of septa (from 0.6 per mm (4-slice) to 1.1 per mm (16-slice) on average), the geometric efficiency of 16-slice detector arrays is somewhat lower. In the latest generation of 64-slice scanners, matrix arrangements that allow for simultaneous acquisition of 64 sub-millimetre slices are employed by the majority of manufacturers (Fig. 3.13). By electronically combining several adjacent rows, thicker slices can be acquired, too, but at a reduced number of slices (e.g. 32·1.25 mm, 16·2.5 mm etc.). Once again, the number of septa was increased (to 1.6 per mm on average), resulting in an additional loss in geometric efficiency. The hybrid detector design exclusively used by Siemens for its Sensation 64 scanner is particular insofar as the number of simultaneous slices claimed by the manufacturer (64) is much larger than the number of rows (32·0.6 mm or 24·1.2 mm). The claim is based on a special acquisition mode that employs two

alternating focal spot positions to simultaneously produce 64 data sets per rotation with 50% overlap in order to achieve a somewhat improved spatial resolution in z-direction. With respect to all other important features (collimation, coverage, overbeaming effects etc.), however, this model behaves as a 32-slice scanner in submillimetre mode and a 24-slice scanner in all other modes at maximum. In addition, the thickness of the smallest slice that can be reconstructed (relevant for partial volume effects) is at least equal to the smallest slice collimation, i.e. 0.6 mm not lowers [ImPACT 2005].

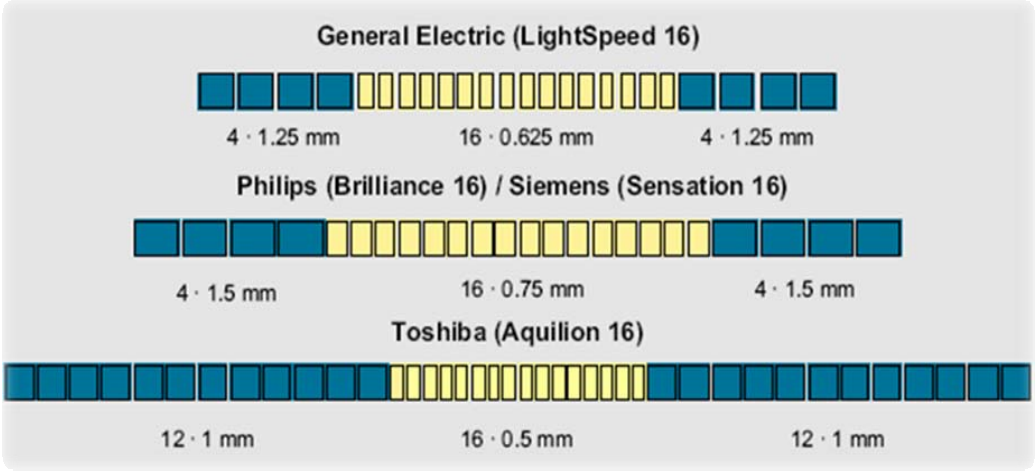


Figure: 2.13: Detectors array in different manufacturers (Detector arrangement of 16-slice scanners, all of them employing a hybrid design, but with differences in the number of rows, detector size, and array width) [ImPACT 2005]**.**

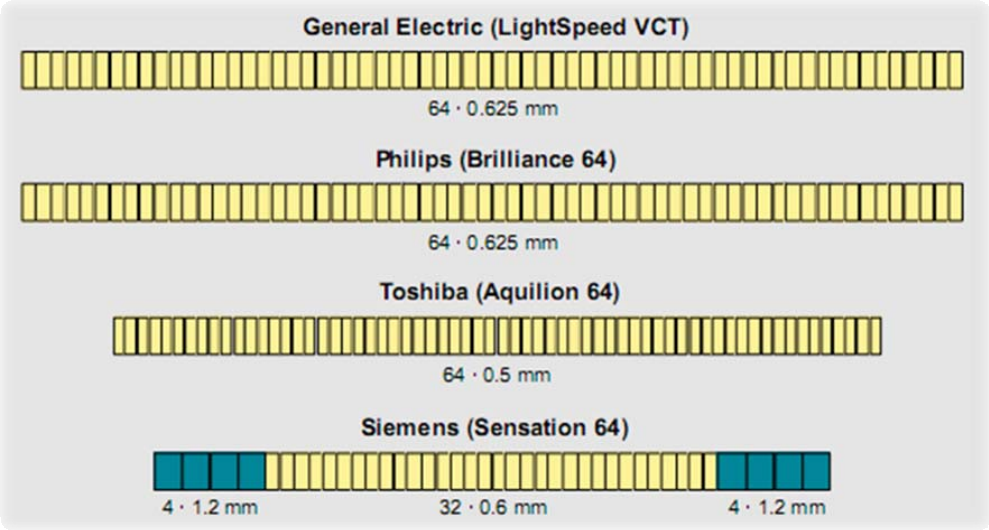


Figure 2.14: Detector arrangement of 64-slice scanners, most of them employing a matrix design with 64 rows of uniform size. The Siemens design refers to a 32-slice scanner that makes use of a particular acquisition mode (alternating focal spot) with 64 overlapping (i.e. non-independent) slices [ImPACT 2005]**.**

2.5.5 Data acquisition system (DAS):

The data acquisition system (DAS) serves to collect the detector signals, to convert them to digital information and to transfer the data to the image reconstruction system. The number of DAS channels, not the number of detector rows, is the decisive parameter that limits the number N of independent slices that can be acquired simultaneously. With the advent of 16-slice scanners at latest, the spatial requirements of an increased number of detector rows and exorbitantly increased data rate no longer use of traditional circuit board [ImPACT 2005]**.**

2.5.6 Spatial interpolation:

Data acquisition in spiral scanning mode requires an additional interpolation step to obtain axial slice.

2.5.7 Adaptive filtration:

Adaptive filtration (AF) is a dedicated data processing technique for projections that are subject to strong attenuation. Without AF, images e.g. from the pelvis region often exhibit inhomogeneous noise patterns due to 'photon starvation'. AF used either to improve image quality or to lower the dose settings [ImPACT 2005]**.**

2.5.8 Overranging:

'Overranging' is an increase in dose-length product (DLP) due to the additional rotations at the beginning and at the end of a spiral scan required for data interpolation to reconstruct the first and the last slice of the imaged body region. With single-slice scanners, theory requires that $\Delta n = 1$ additional rotation is usually made in total [Kalender 1998]. For multi-slice scanners, the situation is much obvious. Overranging effects can be expressed both in terms of the additional number Δn of rotations and increase ΔL in the scan length. ΔL primarily depends on two factors: the beam width N.hcol and the pitch factor p. This can be fairly well described by a linear relationship [Nagel 2005]. The implications of overranging effects for the radiation exposure to the patient, i.e. the dose-length product (DLP), not only depend on ΔL, but also on the length L net of the imaged body region. The percentage increase in DLP is given by:

$$\Delta \text{ DLP net} = (\Delta L / L \text{ net}) \times 100 \quad \text{Equation (2.12)}$$

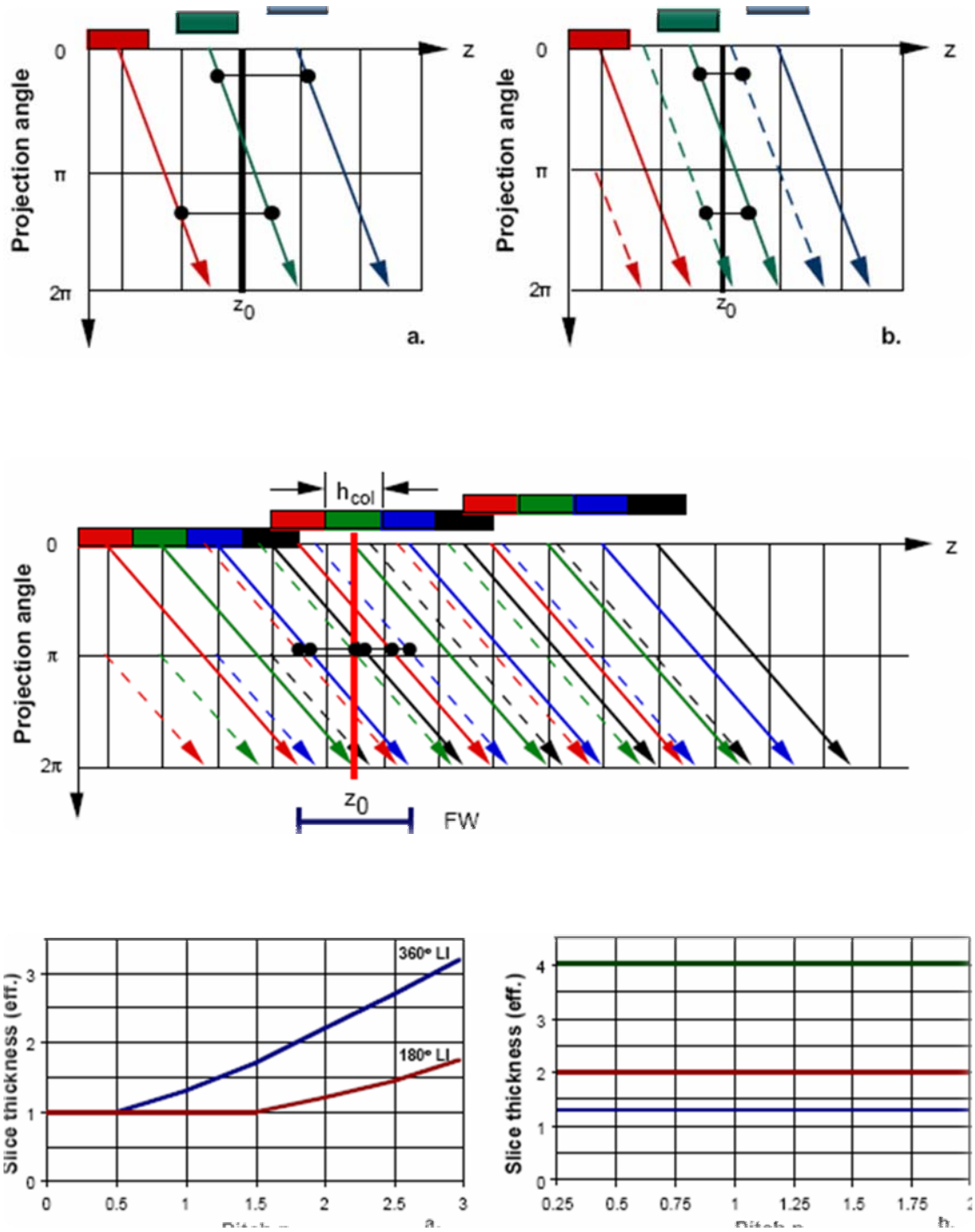
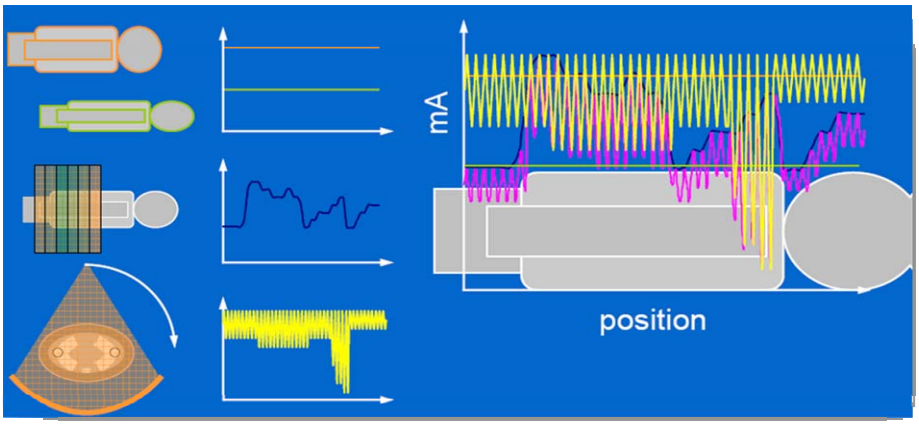


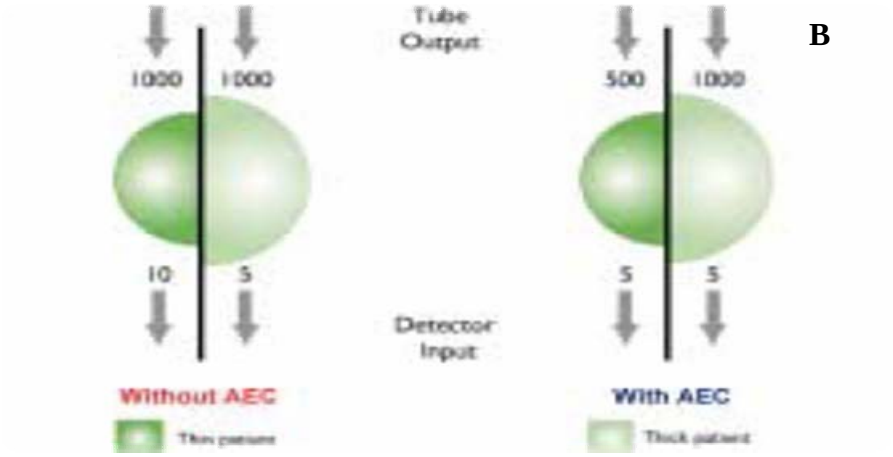
Figure 2.15: Relationship between pitch and slice thickness [Vander 1998].

2.5.9 Devices for automatic dose control (ADC):

Newer scanners are equipped with means that automatically adapt the mAs settings to the individual size and shape of the patient. Automatic dose control systems offer up to four different functionalities that can be use either alone or in combination. Automatic exposure control (AEC Fig. 2.13) that accounts for the average attenuation of the patient's body region that is to be scanned. Information on the patient's attenuation properties are derived from the scan projection radiogram (SPR) usually recorded prior to the scan for planning purposes [Kalender 1998].



A



B

Figure: 2.16: A & B Automatic exposure control **(**AEC) [Nagel 2007].

2.5.9.1 Longitudinal dose modulation (LDM):

Longitudinal dose modulation (LDM, Fig. 2.13), which is a refinement of AEC by adapting the mAs settings locally, i.e. slice-by-slice or rotation by rotation. Those parts of the scan range with reduced attenuation will be less exposed.

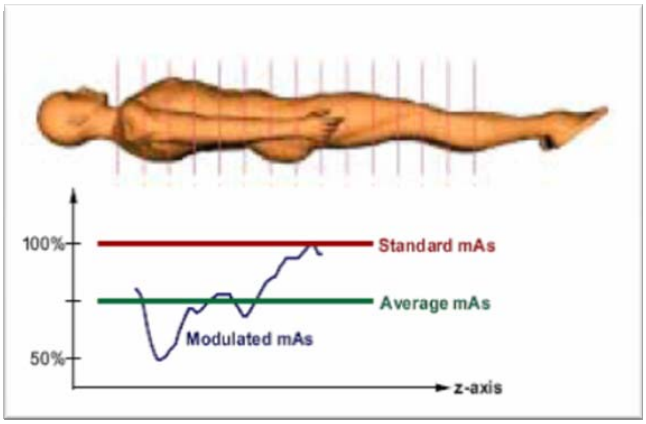


Figure 2.17: Longitudinal dose modulation (LDM) [Vander 1998].

2.5.9.2 Angular dose modulation (ADM):

Angular dose modulation (ADM, Fig. 2.14), another refinement of AEC, which adapts the tube current to the varying attenuation at different projection angles. Those projections with reduced attenuation will be less exposed. Information on the patient's attenuation properties are derived from two scan projection radiogram (SPR) or real-time from the preceding rotation.

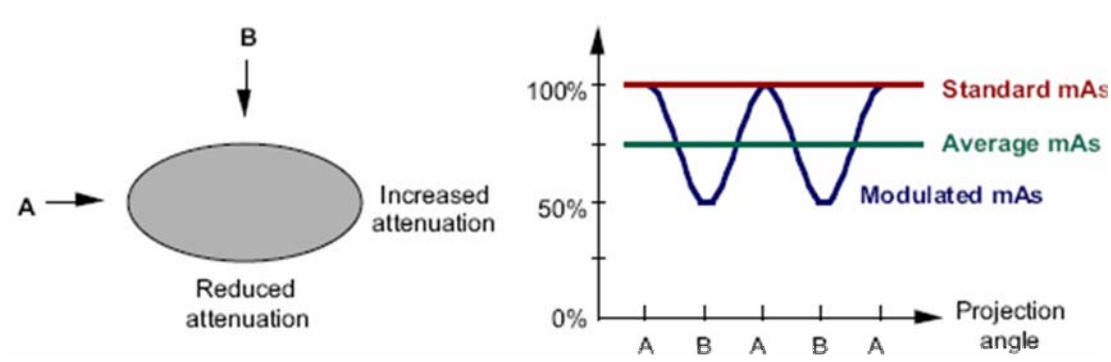


Figure 2.18: Angular dose modulation (ADM) [Nagel 2007].

2.5.9.3 Temporal dose modulation (TDM):

Temporal dose modulation (TDM, Fig. 2.15), that reduces the tube current in cardiac CT (or other ECG-gated CT examinations) during those phases of the cardiac cycle that are not suited for image reconstruction due to excessive object motion. All major CT manufacturers now offer some or all of these functionalities with their latest scanners. A comprehensive report on the current status of automatic dose control system has been published by Impact [Brooks RA, Dichiro G 1976].

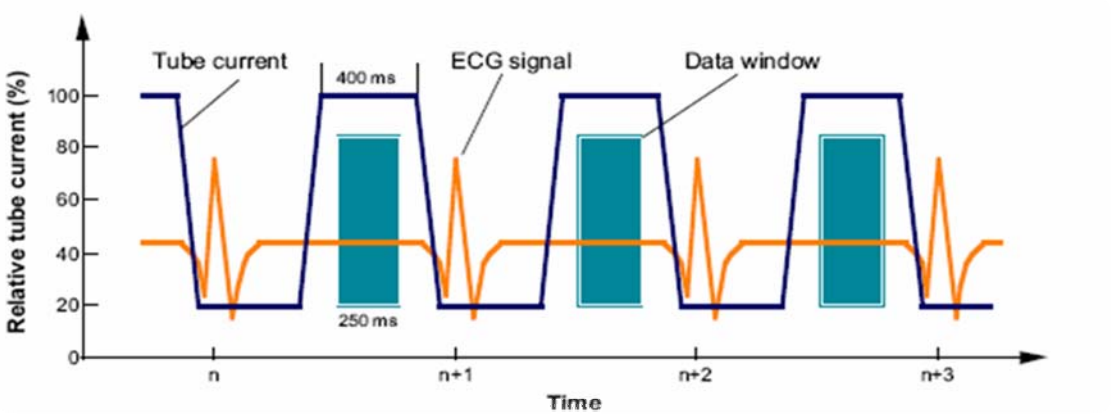


Figure 2.19: Temporal dose modulation (TDM)[D. Tack 2007]

2.5.10. Dose display:

Newer scanners must be equipped with a dose display. At present, only the display of CTDIvol is mandatory [ICRP 2007]. However, many scanners also show DLP, too, either per scan series or both DLP per scan series and DLP per exam. The dose display can be used for purposes of dose optimization. Finally, CTDIvol can be used as a fair estimate for the dose to organs that are entirely located in the scan range. The interpretation of the dose values displayed at the scanner's console needs special attention in the following situations:

Many dose recommendations are given interns of weighted CTDI (CTDIw); in order to allow comparison, the pitch correction involved in CTDIvol must be reverted by multiplying CTDIvol must be reverted by multiplying CTDIvol with the pitch factor.

Up to now, the dose values for examinations carried out on body scanning mode have always based on body-CTDI regardless of patient size. In pediatric CT examinations, the displayed figures should be multiplied by 2 for children and by 3 for infants in order to give a realistic estimate of patient dose.

2.6 Application-related factors:

Although the scanner design is of some importance, surveys on CT practice have regularly shown that the way how the scanner is used has the largest impact on the doses applied in a CT examination. The application-related factors on which patient exposure depends are subdivided in:

Scan parameters: i.e. those factors that directly determine the local dose level (CTDIvol) and that are often pre-installed or recommended by the manufacturer (e.g. in application guides),

examination parameters: i.e. those factors that - in combination with CTDIvol - determine the integral exposure (i.e. DLP) and depend on the experiences of the user, and reconstruction and viewing parameters, which implicitly influence the dose settings.

2.6.1 Brook's formula:

As in conventional projection radiography, aspects of dose and image quality are linked. For CT, Brook and DiChiro [Wilting 2001] have formulated the correlation between these two opposed quantities:

$$D \propto \frac{B}{\sigma^2 \cdot a^2 \cdot b \cdot h} \quad with \quad B = \exp^{-\mu \cdot d}$$

Equation (2.13)

Where:

D = patient dose

B = attenuation factor of the object

μ = mean attenuation coefficient of the object

d = diameter of the object

σ = standard deviation of CT numbers (noise)

a = sample increment

b = sample width

h = slice thickness

This fundamental equation - commonly known as the 'Brooks' formula' - describes what happens with respect to patient dose if one of the parameters is changed while image noise remains constant:

Dose must be doubled if slice thickness is cut by half; dose must be doubled if object diameter increases by 4cm; an eightfold increase in dose is required if spatial resolution is doubled (by cutting sample width and sample increment by half).

In this context, the term ‘dose’ is applicable to each of the dose quantities that are appropriate for CT. Dose and noise are inversely related to each other in such a way that a fourfold increase in dose is required if noise is to be cut by half. It should be noted, however, that the Brooks’ formula is incomplete in that image quality is only considered in terms of quantum noise and spatial resolution. Other important influences, such as contrast, electronic noise or artefacts, are not taken into account and will therefore modify optimization strategies under particular circumstances [Nagel 2007].

2.6.2 Scan parameters:

2.6.2.1 Tube current - time product (Q):

As in conventional radiology, a linear relationship exists between the tube current-time product and dose; i.e. all dose quantities will change by the same amount as the applied mAs. The mAs product Q for a single sequential scan is obtained by multiplying the tube current 1 and exposure time (t); in spiral scan mode, Q is the product of tube current 1 and rotation time (trot). This should not be mixed up with the total mAs product of the scan which is the product of tube current I and (total) scan time T. The consequences on image quality resulting from variations in the tube current-time product are relatively simple to understand. The only aspect of image quality so affected is image noise, which is - as indicated in equation (2.13) - inversely proportional to the square root of dose (i.e. mAs).

The tube current-time product is often used as a surrogate for patient dose (i.e. CTDI). With the advent of multi-slice scanners, additional confusion arose due to the introduction of a different, pitch – corrected mAs notation ('effective mAs', or mAs per slice'). The settings for the tube current-time product should be adapted to the characteristics of the scanner, the size of the patient and the dose requirements of each type of examination. Examinations with high inherent contrast, such as for chest or skeleton, that are characterized by viewing with wide window settings, can regularly be conducted at significantly reduced mAs settings.

2.6.2.2 Tube potential (kVp):

When the tube potential is increased, both the tube output and the penetrating power of the beam are improved, while the image contrast is adversely affected**.** In conventional projection radiology, increased tube potential are applies in order to ensure short exposure times for obese patients, to equalize large differences in object transmission (e.g. during chest examinations) or to reduce patient dose.

In CT, increased tube voltages are used preferentially for improvement in tube loading and image quality. Contrary to the case for mAs, the consequences of variations in kV cannot easily be assessed. The relationship between dose and tube potential U is not linear, but rather of exponential nature which varies according to the specific circumstances. The intensity of the radiation beam at the detector array, for example, varies with U to the power of 3.5. The decrease in primary contrast which normally results from this action is largely over-compensated by the associated decrease in noise, i.e. the higher the tube potential, the better the contrast-to-noise ratio CNR (except for the application of iodine as contrast agent). The only reason why this analysis generally holds true is the absence of any kind of AEC in the majority of scanners which

might prevent unnecessary increases in the detector signal. This clearly demonstrates that dose is not reduced by applying higher kV settings, but merely increased as long as mAs settings are not changed. This means that both are increased by approximately 50% if kV settings are changed from 120 to 140 kV. Decreasing tube current by 50% will essentially decrease radiation dose by 50%. So 80 kV instead of 120 kV would allow to reduce of the patient dose by almost a factor of two without scarifying image quality. Therefore the question is justified whether and when it might be reasonable to deviate from the 120 kV setting usually applied, this depends on the attenuation characteristics of the detail that is diagnostically relevant. The figures are given in terms of contrast-to-noise ratio squared (CNR2) at constant patient dose; this notation allows to directly convert the percentage differences into dose differences. For soft tissue contrast (e.g. differences in tissue density), higher tube potentials perform slightly better than lower ones, but the differences are quite small. The opposite holds true for bone contrast (i.e. bone vs. tissue). For iodine contrast, however, there is a strong dependence on tube potential that is much in favour of lower kV settings. So 80 instead of 120 kV would allow reducing the patient dose by almost a factor of two without scarifying image quality.

Tube potentials other than120 kV should be considered only in case of:

Obese patients in where mAs cannot be further increased: use high kV settings, slim patients and pediatric CT where mAs cannot be further reduced: use lower kV settings, and CT angiography with iodine; use lower kV settings.

Variations in tube potential should not be considered for pure dose reduction purposes except in the case for CT angiography. Due to the complexity involved, adaptation of mAs settings should not be left to automatic exposure control systems, as these do not account for changes in contrast.

Dose settings in CT angiography should not be higher than in unenhanced scans of the same body section and should be lowered if performed at reduced kV settings.

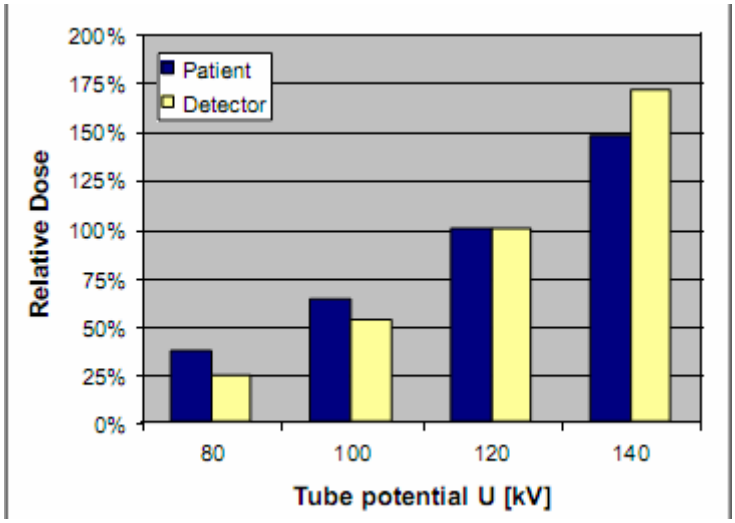


Figure 2.20: Voltage dependence of patient dose (CTDIw) and detector signal [Nagel 2007].

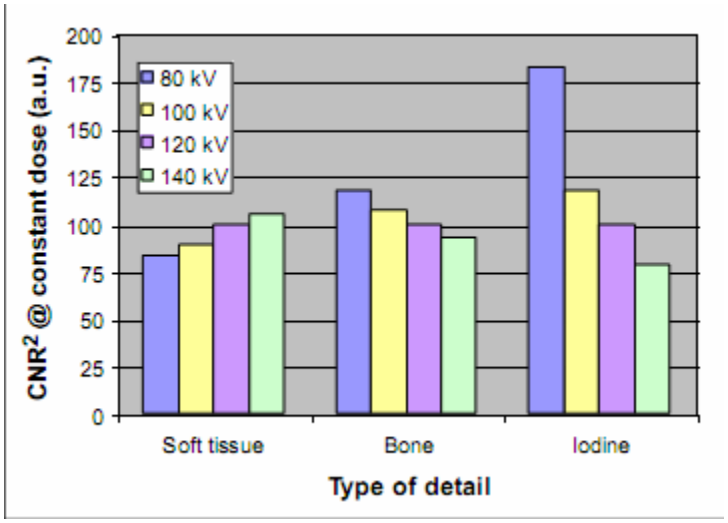


Figure 2.21: Voltage dependence of contrast-to noise ratio squared (CNR2) at constant patient dose (CTDIw) for different types of detail. While CNR2 is almost constant for imaging of soft tissue and bone, imaging performance is significantly improved for iodine at lower voltages [Nagel 2007].

2.6.2.3 Slice collimation (hcol) and slice thickness (hrec):

With single-slice CT, the slice collimation hcol used for data acquisition and the reconstructed slice thickness hrec used for viewing purposes were identical. So there was no need to distinguish between them. With multi-slice CT, the slice collimation hcol (e.g. 0.75 mm) and the reconstructed slice thickness hrec (e.g. 5 mm) are usually different. Frequently, the selection of hrec is made with aspect to multiplanar reformatting (MPR) purposes (e.g. 1 mm).

As reduced slice thickness is associated with increased image noise, this may have a significant impact on patient dose as express by the Brook's formula. The slice collimation should be selected as small as compatible with the aspects of overbeaming/over ranging, total scan time and tube power. Viewing should preferentially be made with thicker slabs (e.g. 3 to 8 mm), thereby reducing image noise and other artifacts. Thinner slab should only be used if partial volume effect is of importance. This should preferentially be done in conjunction with workstations that allow changing the slab thickness. Except for very narrow slice, there should be no need for any increase in dose setting on reduction of slice thickness.

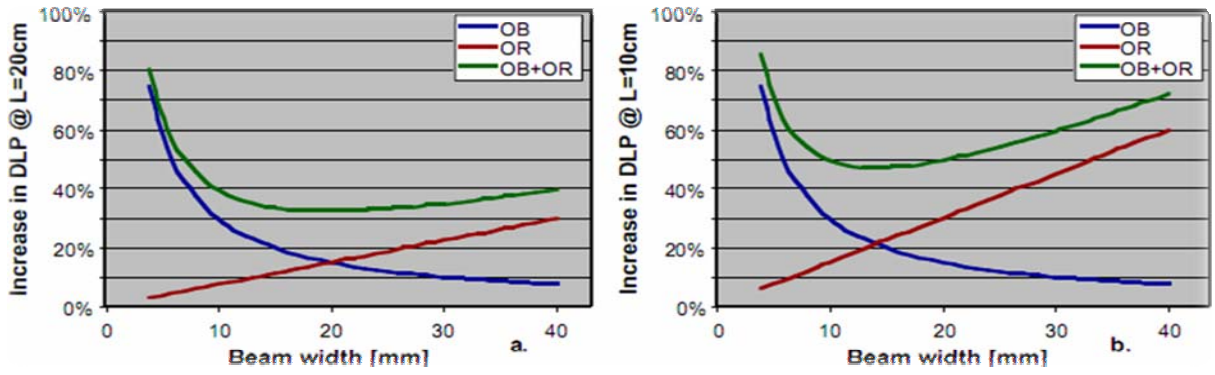


Figure 2.22: Increased dose-length product due to overbeaming (OB) and overranging (OR) effects for a typical MSCT scanner. For average to long scan ranges (L = 20 cm and more, a.), all beam width settings above 10 mm perform almost equally well. For short scan ranges (L = 10 cm as in paediatric and spine exams, b.), beam width settings between 10 and 20 mm should be preferred [Brooks RA, Dichiro G 1976].

2.6.2.4 Pitch:

Defined as tube distance traveled in one 360º rotation/total collimated width of X-ray tube beam. Is inversely proportional to patient dose (Longer pitch lower radiation dose). The relationship between pitch and radiation dose is linear, specifically increasing the pitch from 1.0 to 1.5 mm will reduce the dose by 33%.

With SSCT scanners, scanning at increased pitch settings primarily serves to increase the speed of data acquisition. As a side effect, patient dose is reduced accordingly, at the expense of impaired slice profile width, i.e. z-resolution, however. As already outlined in section 3.2.6, MSCT scanners make use of a spiral interpolation scheme that is different from SSCT. Thus the slice profile width remains unaffected from changes in pitch settings. Instead, image noise changes with pitch (Fig. 3.34a) unless the tube current is adapted accordingly. Scanners that make use of the effective mAs (mAs per slice) concept not only keep slice profile width, but also image noise constant when pitch changes (Fig. 3.34a). To achieve this goal, the electrical mAs product supplied to the x-ray tube automatically changes linearly with pitch (Fig. 3.34b). As a consequence, patient dose (CTDIvol) is no longer reduced at increased pitch settings in contrast to SSCT scanners. On the other hand, dose will also not increase at reduced pitch settings. MSCT scanners without automatic adaptation of mAs will still save dose at increased pitch setting, but this will happen at impaired image quality (more noise) as long as mAs is not adapted manually. Frequently, image quality in terms of artefacts depends on pitch settings. In general, spiral artefacts are reduced at lower pitch settings. For similar reasons, some scanners allow the setting of a limited number of ‘preferred’ pitches only. Reduced pitch settings can also be applied to enhance the effective tube power, however, at the expense of reduced scanning speed. Pitch settings with MSCT scanners should be made exclusively with respect to scan

speed, spiral artefacts and tube power. Dose considerations no longer play a role if scanners that employ effective mAs are used or if (electrical) mAs are adapted to pitch to achieve constant image noise.

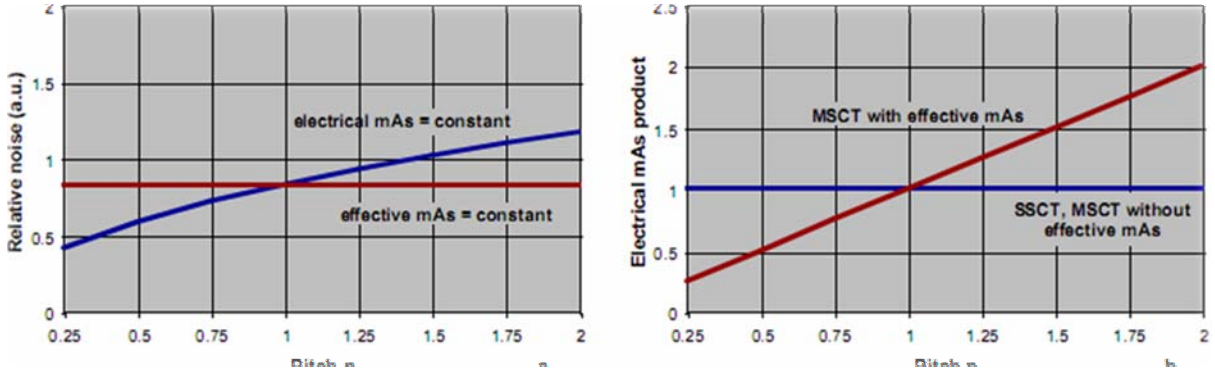


Figure.2.23: For MSCT systems that employ multi-point spiral data interpolation (z-filtering), image noise changes with pitch unless effective mAs is held constant (a.). This implies that the electrical mAs product supplied to the tube changes with pitch (b.). Contrary to SSCT, changes in pitch settings therefore no longer have any influence on patient dose in terms of CTDIvol [Brooks RA, Dichiro G 1976].

2.6.2.5 Gantry rotation time:

Decreasing gantry rotation time decreases radiation dose in linear fashion. The faster gantry rotation, the lower the dose increasing the cycle speed of rotation from 1.0 to 0.5 seconds per 360º rotation reduced the dose essentially by 50%.

2.6.2.6 Object diameter (d) or patient (m):

Patient size, although not a parameter to be selected at the scanner's console, represents an important influencing parameter that needs to be considered in this study. Considerable reductions in mAs settings are appropriate whenever slim patients, and particularly children, are

examined. In order to avoid unnecessary over-exposure, the mAs must be intentionally adapted by the operator unless AEC-like devices are available. Due to decreased attenuation for the smaller object, image quality will not be impaired if mAs is selected appropriately. This means that the image quality will be at least as good as for patients of normal size, although the dose has been reduced [Honnef et al. 2004, Van E et al.1998].

$$Relative\ mAs \propto body\ weight + 5\ kg$$

Equation (2.14)

mAs settings should be adapted to patient size in more gentle way (factor 2 per 8 cm change in diameter) than predicted by theoretical considerations that only account for image noise. In addition, body regions with better inherent contrast should be scanned at reduced mAs settings.

Preferentially, AEC systems that rather measure than estimate patient absorption should be used, provided that their algorithm makes use of this more gentle mAs adjustment.

If not, Manual adjustment using a set of patient-weight adapted protocols that are based on Rogalla's formula should better be applied instead. For head examinations, mAs adaptation should not be made with respect to patient weight, but to patient age. The following formula can be used to convert from lateral patient diameter d lat (in cm) to patient weight m (in kg) and vice versa:

$$d_{lat} = 6.5 + 3 \cdot \sqrt{m}$$

Equation (2.15)

2.6.3 Examination parameters:

2.6.3.1 Scan length (L):

The local dose, i.e. CTDI is almost independent of the length of the scanned body section. The same does not hold, however, for the integral dose quantities, i.e. DLP and effective dose, both are increased in proportional to length of the scanned body section. Both increase in proportion to the length of the body section. Therefore, limiting the scan length according to the clinical needs is essential. On most scanners, the scan length L is usually not indicated explicitly. Instead, the positions of the first and the last slice are stated only; the same holds for the information that is documented on the images or in the DICOM data file. For each patient, the scan length should be selected individually, based on the scan projection radiograph that is generally made prior to scanning for the purposes of localization, as should be kept as short as necessary. Moreover, a reduction in the scan range should be considered in multi-phase examinations and follow-up studies. Whenever feasible, critical organs like the eye lenses or the male gonads should be excluded from the scan range. This may be difficult for MSCT scanners that allow for large beam width settings due to increased overranging effects.

2.6.3.2 Number of scan series (nser):

In CT terminology, a scan series is usually referred to as a series of consecutive sequential scans or one complete spiral scan. The number of scan series (phases) should be kept as low as necessary. This holds particularly for liver examinations where the studies with up to six different phases are sometimes recommended in literature.

2.6.3.3 Number of rotations in dynamic CT studies (n):

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

Dynamic CT studies should be made with the lowest dose settings, the most narrow beam width, the shortest length and the smallest image rate that is compatible with clinical needs of the examination.

2.7 Patient dose from CT:

As the number of computed tomography (CT) procedures performed worldwide countries to increase, there is growing concern about patient protection issues. Currently, no system is in place to track a patient life time cumulative dose from medical sources, and questions have arisen regarding the possible threat to public health from the widespread use of CT. In this part, the authors reviewed the published literature to determine whether patients are receiving a higher

absorbed dose of radiation and explored several proposed models to optimize the radiation dose delivered to patients and track the cumulative lifetime dose. A recent study by Aldrich and Williams [Aldrich and Williams 2007] quantified changes in numbers of radiology examinations in order to examine the correlation to radiation dose received by the patient. In addition to a 4-folds increase in CT examinations, they also found the average annual effective dose per patient almost doubled during the study period; from 3.3 mSv in 1991 to 6.0 mSv in 2002. CT is the largest contributor to patient dose in radiology. This could be because more CT scanners are in use and their performance has been enhanced, along with increasing indications for CT examinations. CT is not the only modality that has experienced more use and has the potential to deliver higher patient radiation doses. It drew attention to the fact that optimizing technique and standardizing practice could benefit the field of radiology and protect patient from overexposure to ionizing radiation. Although not pivotal to discussion of correlating increased use of CT to an increased patient radiation dose [Shrimpton PC, et al. 1991], study calls attention to the fact that to the patient dose can be reduced careful attention to technique and optimization.

Yoshizumi and Nelson [Shrimpton PC, et al. 1991] pointed out the need to balance optimization of image quality against radiation dose in developing clinical protocols. Their study described fundamental concepts of radiation dose in detail, including the CT dose index and other technical factors such as pitch effect, dose profile in the penumbra and signal to noise ratio. Yoshizumi and Nelson concluded that multi-detector CT (MDCT) radiation dosimetry issues have not been addressed adequately and have lagged behind advances in the actual technology.

Other researchers also are questioning about the effects of newer imaging technologies on the patient radiation dose. Berland and Smith [Task Group 2000] proposed that the absorbed dose could be up to 40% higher using MDCT compared with older CT generation scanners. Golding

and Shrimpton [Lee 2001] suggested that “evidence indicates a strong trend of increasing population dose owing to rising use of CT and to increase dose per examination”. A significant body of literature focuses on discovering a causal link between increased use of CT scanner and an increase radiation absorbed dose to the patient population.

Numerous studies have suggested that, although CT is not the most commonly performed radiologic examination, it is the largest source of radiation dose. [Nagel et al. 1989] found that, although CT represents only about 4% of all radiologic examinations, it is responsible for up to 35% of collective radiation dose to the population from radiologic examinations. In related National Cancer Institute report, data suggested that the use of CT in adults and children has increased approximately 7 folds in the past 10 years. In large U.S hospitals, CT represents 10% of diagnostic procedures and accounts for approximately65% of the for all medical effective radiation dose examinations.

[Aldrich et al. 2007] conducted a study to compare the dose length product (DLP) and effective radiation dose to the patients from CT examinations. They compared data from 1070 CT examinations and concluded that considerable variation existed in the dose length product and patients radiation dose for specific examination. This study called attention to the need to optimize the effective dose to the patient and conduct more research to determine which additional efforts are needed to minimize patient exposure. Optimizing technical factors for examinations can help reduce patient radiation dose, thereby reducing risks. A pivotal study by [Lee 2001] assessed awareness levels among patients, emergency department physicians and the radiologists concerning radiation dose and the risks involved with CT scans. Lee and colleagues concluded that patients were not given information about the risks, benefits and radiation dose from a CT scan. Regardless of their experience levels, few of the participants in the study

(including the emergency department physicians and the radiologists) were able to provide accurate estimates of CT radiation doses. This study underscores the prevalent lack of attention to the issue lifetime cumulative radiation doses. This must be become a central issue so that risk can be studied and monitored. One disadvantage to communicating instinct of cumulative radiation dose would be the natural instinct of some patients to defer or cancel the examination. Professionals should highlight the benefits of the examination when discussing risks with the patient. Physicians improve their understanding of radiation risks from medical imaging examinations.

[Amy K et al. 2009] evaluate the image noise, low-contrast resolution, image quality, and spatial resolution of adaptive statistical iterative reconstruction in low-dose body CT. Adaptive statistical iterative reconstruction was used to scan phantom at the American College of Radiology reference value and at one-half that value (1.25 mGy). Test objects in low and high contrast and uniformity modules were evaluated. Low-dose CT with adaptive statistical iterative reconstruction was then tested on 12 patients (7 men & 5 women; average age is 67.5 years) who had previously undergone routine-dose CT. Two radiologists blinded to scanning technique evaluated images of the same patients obtained with routine-dose CT and low-dose CT with and without adaptive statistical iterative reconstruction. Image noise, low-contrast resolution, image quality, and spatial resolution were graded on scale of 1 (best) to 4 (worst). Quantitative noise measurements were made on clinical images.

In the phantom, low- and high-contrast and uniformity assessments showed no significant difference between routine-dose imaging and low-dose CT and with adaptive statistical iterative reconstruction. In patients, low-dose CT and with adaptive statistical iterative reconstruction was associated with CTDI reductions of 32– 65% compared with routine imaging and had the least

noise both quantitatively and qualitatively ($P < 0.05$). Low-dose CT with adaptive statistical iterative reconstruction had identical results for low-contrast resolution and nearly identical results for overall image quality (grade 2.1 – 2.2). Spatial resolution was better with routine-dose CT ($P = 0.004$). These preliminary results support body CTDI reductions of 32– 65% when adaptive statistical iterative reconstruction is used. Studies with larger statistical samples are needed to confirm these findings.

[Aaron, et al. 2009] estimate cumulative radiation exposure and lifetime attributable risk (LAR) of radiation induce cancer from CT scanning of adult. The cohort comprised 31462 patients who underwent diagnostic CT in 2007 and had undergone 190712 CT examinations over the prior 22 years. Each patient's cumulative CT radiation exposure was estimated by summing typical CT effective dose, and the Biological Effect of Ionizing Radiation (BEIR) VII methodology was used to estimate LAR on the basis of sex and age at each exposure. 33% of patients underwent five or more lifetime CT examinations, and 5% underwent between 22 and 132 examinations. 15% received estimate cumulative effective doses of more than 100 mSv, and 4% received between 250 and 1375 mSv. Associated LAR had mean and maximum values of 0.3% and 12% for cancer incidence and 0.2% and 6.8 for cancer mortality, respectively. 7% of the cohort had estimated to produce LAR greater than 1% of which 40% had either no malignancy history or a cancer without evidence of residual disease. Cumulative 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.

[Alice B, et al. 2009] quantified retrospectively the effect of systematic use of tube current modulation for neuroradiology CT protocols on patient dose and image quality. The authors evaluated effect of dose modulation on four types of neuroradiologic CT studies: brain CT

performed without contrast, material (unenhanced CT) in adult patients, unenhanced brain CT in pediatric patients, adult cervical spine CT, and adult cervical and intracranial CT angiography. For each type of CT study, three of 100 consecutive studies were reviewed: 100 studies performed without dose modulation, 100 studies performed with z-axis dose modulation, and 100 studies performed with x-y-z-axis dose modulation. For each examination, the weighted volume CT dose index (CTDIvol) and dose length product (DLP) were recorded and noise was measured. Each study was also reviewed for image quality. Continuous variables (CTDIvol, DLP, noise) were compared by using t test and categorical variables (image quality) were compared by using Wilcoxon rank-sum test. For unenhanced CT of adult brains, the CTDIvol and DLP, respectively, were reduced by 60.9% and 60.3%, respectively, by using z-axis dose modulation and by 50.4% and 22.4% by using x-y-z-axis dose modulation. Significant dose reductions ($P < 0.001$) were also observed for pediatric unenhanced brain CT, cervical spine CT, and adult cervical and intracranial CT angiography performed with each dose modulation technique. Image quality and noise were unaffected by use of either dose modulation technique ($P < 0.05$). Use of dose modulation techniques for neuroradiology CT examinations affords significant dose reduction while image quality is maintained.

Finally, a unique study conducted in Sudan regarding patient dose in CT [M A Aziz 2007]. The study assessed the radiation doses for patients undergoing routine CT examinations in four centers in Khartoum state for various CT examinations of head, neck, abdomen, pelvis and chest. CTDIvol, DLP and effective dose were calculated using CT-exposure software. The mean CTDIw, CTDIvol DLP and effective dose were found to be 32.6 mGy, 26.5 mGy, 454 mGy and 3.3 mSv respectively.

Chapter Three

Materials and Methods

Chapter Three

Materials and Methods

3.1 Materials and Methods:

The data of this study were collected from Alzaytouna specialist and Alamal National hospitals. Data of the technical parameters used in CT procedures was taken during April 2011 –June 2011.

3.2 CT machines:

Multislice CT scanners (MSCT) 64 slice (Toshiba Sensation aquilion 64) were installed in 2010. All quality control tests were carried out for the machines by experts from Sudan Atomic Energy Commission (SAEC) prior to any data collection. All data were within acceptable ranges.

3.3 Patient data:

A total of 130 patients with different CT examinations were referred to Alzaytouna specialist and Alamal National hospitals in the period of study. All the patients were performed using departments' protocols with Multislice CT (MSCT) 64 slice (Toshiba Sensation aquilion 64). Data were collected to the study the effects of patient-related parameters (e.g. age, sex, diagnostic of examination, and use of contrast media) on the patient dose. Data were collected to investigate the effect of exposure-related parameters (gantry tilt, kilo voltage (kV), tube current (mA), exposure time, slice thickness, table increment, number of slices, and start and end positions of scans) on patient dose. The collection of the patient exposure

parameters was done using patient dose survey forms prepared for collection of patient exposure- related parameters.

3.4 CT dose measurements:

The patient dose estimation from CT examinations using the Monte Carlo technique requires measurements of CTDI and conversion coefficient data packages (1-1). In theory, the CTDI, which is a measure of the dose from single slice irradiation, is defined as the integral along a line parallel to the axis of rotation (z) of the dose profile, D(z), divided by the nominal slice thickness, t (1, 1-5,41). In this study, CTDI was obtained from a measurements of dose, D (z), a long the z-axis made in air using a special pencil shaped ionization chamber (Diados, type 11003, PTW-Freiburg) connected to an electrometer (Diados, type 11003, PTW-Freiburg). The calibration of the ion chamber is traceable to the standards of the German National Laboratory and was calibrated according to the international Electrical Commission standards [IEC 1999]. The overall accuracy of ionization chamber measurements was estimated to be ±5% Measurements of CTDI in air (CTDI100, air) were made as recommended by the EUR 16262N [European Commission 1999] based on each combination of typical scanning parameters obtained from the machine [Hart 1996]. The required organ dose for this study was estimated using normalized CTDI values published the IMPACT group [Hart 1996]. For the sake of simplicity, the CTDI100, air will henceforth be abbreviated as CTDIair.

3.5 Radiation dose optimization steps were included:

Optimize CT settings. Based on patient weight or diameter and anatomic region of interest, evaluate whether the CT operating conditions are optimally balanced between image quality and radiation exposure. To reduce dose while maintaining diagnostic image quality; reduce tube

current, increase table increment, (axial scanning) or pitch (helical scanning).

Reduce the number of multiple scans with contrast material. Often, CT scans are done before, during, and after injection of IV contrast material. When medically appropriate, multiple exposures may reduced by eliminating pre-contrast images (i.e. unenhanced images).

Eliminate in appropriate referrals for CT. In some cases, conventional radiography, sonography, or magnetic resonance imaging (MRI) can be just as effective as CT, and with lower radiation exposure.

3.6 Cancer risk estimation:

The risk (RT) of developing cancer in a particular organ (T) following CT exam after irradiation was estimated by multiplying the mean organ equivalent (H_T) dose with the risk coefficient (f_T) obtained from ICRP [ICRP 2006, ICRP 2007]. The overall lifetime mortality risk per procedure resulting from cancer/heritable was determined by multiplying the effective dose (E) by the risk factor (f). The risk of genetic effects in future generations was obtained by multiplying the mean dose to the gonads by the risk factor [ICRP 2006, ICRP 2007].

$$R = E_{.f} = \sum R_T \quad \text{Equation (3.1)}$$

Table 3.1: Radiation risk for adults and workers:

Exposed population	Cancer		Heritable effects		Total	
	ICRP 103(2)	ICRP 60 (19)	ICRP 103	ICRP 60	ICRP 103	ICRP 60
Whole population *	5.5	6.0	0.2	1.3	6.0	7.3
Adults workers**	4.1	4.8	0.1	0.8	4.0	5.6
Children	NA	13	0.08	0.1	NA	NA

*Age between 0-90 years old ** Adults workers aged 18-64

It is important to note that alternative methods and conversion coefficient exist to calculate the effective dose. This estimate only and can differ from other estimates by as much as a factor of 2. This estimate is not the dose for any given individual, but rather, for a standardized anthropomorphic phantom, representative of the "whole body equivalent" radiation detriment (risk) associated with the "partial body" CT examination. These values can be used to optimize protocols, and as a broad indication of the relative risk of the CT examination compared to background radiation or examinations from other modalities.

Chapter Four

Results

Chapter Four

Results

CT examinations for adults and pediatrics have contributed greatly to diagnose different diseases. However, the radiation exposure to the patient is significantly higher compared with other radiological examinations. Dose monitoring during CT scan procedures and re-evaluation of equipment and techniques used, if necessary, are mandatory to keep the patients radiation risk as low as reasonable achievable. As previously mentioned the aims of this study are to measure, estimate the risk for patients and optimize the radiation dose to the patients during the CT examinations.

Table 4.1: Adult patients' data (Sex, number per hospital)

Hospital	Brain		Chest		Abdomen		Others		Total
	M	F	M	F	M	F	M	F	
Alzaytouna	14	11	9	11	18	10	10	4	87
Al-Amal	9	2	4	5	6	5	-	-	31
Total	23	13	13	16	24	15	10	4	118

Table 4.2: Pediatric patients' data (age, No per hospital):

Hospital	Brain		Chest		Abdomen		Others		Total
	M	F	M	F	M	F	M	F	
Alzaytouna	4	2	-	-	-	-	2	-	8
Al-Amal	-	1	1	1	1	-	-	-	4
Total	4	3	1	1	1	-	2	-	12

Table 4.3: Doses of adults; Mean±sd and range (min-max) in the parenthesis:

Hospital	kVp	mAs	Total mAs	Slice thickness	No of Slices	Total time
Alzaytouna	120±0 (120-120)	242.8±144.2 (87-743)	6922±4340.2 (1713-19910)	5.5±1.7 (3-12)	81.8±60.7 (23-277)	26.8±16.9 (7.4-70.9)
Al-Amal	119.4±3.6 (100-120)	162.9±47.8 (100-225)	NA	6.5±3.1 (0.5-14)	1141.6± (24-3706)	NA

Table 4.4: Doses of adults from Alzaytouna hospital; Mean±sd, and range (min-max) in the parenthesis:

	CTDI vol (mGy)	DLP (mGy.cm)	Effective dose (mSv)
Chest	191±260.4 (26.6-887.2)	1878.7±553.2 (1085-3228.3)	26.3±7.8 (15.2-45.2)
Abdomen	199.4±239.3 (12.2-826.9)	3174.1±2200.9 (672.6-7619.8)	47.6±33.0 (10.1-114.3)
Brain	114.6±158.1 (50.2-864.8)	1773.1±723.4 (1003.7-4262)	3.7±1.5 (2.1-8.9)
Others	231.4±259.9 (12.7-887.2)	2370.5±1089.4 (699.5-3588.9)	Not applicable

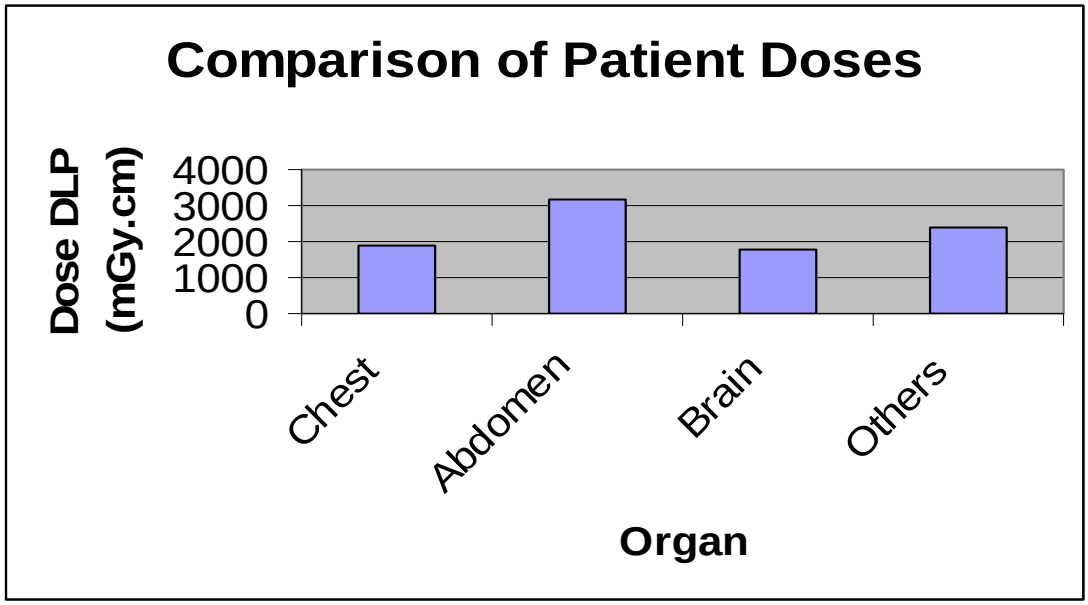


Figure 4.1: illustrates the comparison of patients' doses.

Table 4.5: Doses of pediatrics from Alzaytouna hospital; Mean±sd, and range (min-max) in the parenthesis:

	CTDI vol (mGy)	DLP (mGy.cm)	Effective dose (mSv)
Chest	N.A	N.A	N.A
Abdomen	N.A	N.A	N.A
Brain	70.9±19.2 (31.8-80.8)	1458.1±399.5 (671.7-1793.2)	3.1±0.8 (1.4-3.8)
Others	116±54.6 (77.4-154.6)	3457.25±162 (3342.7-3571.8)	Not applicable

Table 4.6: Doses of adults from Al-Amal hospital; Mean±sd, and range (min-max) in the parenthesis:

Examination	CTDIvol (mGy)	DLP (mGy.cm)	Effective dose (mSv)
Chest	26.7±15.5 (14.6-58.4)	1246.3±906.8 (537.7-2873.1)	17.4±12.7 (7.5-40.2)
Abdomen	62.1±85.2 (12.2-246)	1528.7±952.3 (604.2-3136.5)	22.9±14.3 (9.1-47.0)
Brain	73.1±22.1 (31.8-96.4)	1128.6±441.7 (464.8-1669.9)	2.4±0.9 (0.9-3.5)

Chapter Five

Discussion

Chapter Five

Discussion

5.1 Discussion:

A total of 130 patients were examined over 4 months. Table 4.1 and Table 4.2 represent the anatomical regions of all patients. Anatomical regions are important in radiation dosimetry because in CT imaging the applied protocol affects drastically radiation dose (region of interest ROI) and use of contrast media. There were large variations in the radiation dose to the patients. In general, these variations of doses can be attributed to the differences in tube voltages (kVp), number of scan, tube current time product (mAs) and repeated scans. There may be justifiable reasons for some variability in practice; of which the most important one is the difference in anatomical regions. This difference is greater if operators and practitioners are insufficiently educated in newly emerging technology. Further, increasing demand in CT examination in Sudan may induce operators to overcome the ideal referring criteria because the vast majority of CT services are concentrated in Khartoum state. It is perceived that this is more likely to occur with relatively inexperienced workers and it is also possible that some examinations are carried out more intensively than needed as a mean of radiation risk limitations. These factors indicate strongly against measures to provide effective radiation protection. It is necessary to establish the minimum exposure threshold that will deliver adequate image quality in each application, preferably expressed in terms of clinical effectiveness.

Table: 5.1: Comparison of patient dose during CT with previous studies**:**

Author	No. of pts	Exam	Machine model	Pitch	kVp	mAs	Slice th	Dose mSv		
								CTDIvol	DLP	Effective dose
Present study	31	Chest, Abd, Brain& s	Toshiba Sensation aquilion 64	1.5	120	242.8	5.5	178.3	2344.4	20.05
A.M Nour	83	Abd	Siemens Somatom emotion	0.75-1	80-120	42-243	24	18.87 mGy	865.3 mGy.cm	13.5
Entisar Omer	51	CTU	Siemens Somatom emotion duo		110-130	37-111		25.1-10.95	85-425	1.29-6.37
European Commission1999		Routine Abd, Pelvis & Liver							780	11.7
I.I.Suliman	445	Head, chest, abd& pelvis	Toshiba Somatom sensation 16	NA	120	41±17	5.8±1	65	507.3	11.3

Patients were referred to the CT scan department for different examinations (e.g. brain, chest, abdomen and others). The patients were scanned with routine protocols which were desired by the manufacturer adopted by the operator.

The patients were scanned with kVp range from (100-120), and scanned with mAs range (87-743) as shown in the Table 4.3, and with the pitch (0.75) or less than (1), and feed/rotation about 18 mm, and slice thickness collimation was (16x1.5), and slice thickness was 5 mm, and rotation time was (0.5 s) and region of scan included thee ROI. These parameters produced these radiation values represented in Table 4.4 which shows the values of DLP average (2275.3 mGy.cm), CTDIvol average (168.3 mGy) and effective dose average (20.05 mSv). The range of DLP was (464.8-7619.8 mGy.cm) and the large variable in DLP due to repeated of scans and patients.

In CT examination, patients are exposed to high radiation dose. Therefore, the used of ordinary dose values (CTDI, or DLP) will provide less information regarding the radiation risks. Effective dose is the unit of choice in this situation (partial exposure) and furthermore, comparisons between different procedures are possible with different imaging modalities. In this study, the mean effective dose for Al-amal hospital was 17.4±12.7 mSv, 22.9±14.3 mSv and 2.4±0.9 mSv for the chest, abdomen and brain, respectively. The mean effective dose for Al-Zaytouna hospital were 26.3±7.8 mSv, 47.6±33.0 mSv, 3.7±1.5 mSv for the chest, abdomen and brain, respectively

DRLs can be used to verify the practices for typical examinations for group of standardized patients in order to ensure that the dose should not be exceeded in normal practice without

adequate justification [ICRP 1991]. Nevertheless, the available data is still not enough to establish national reference levels, but this could be a baseline for further studies concerning dose optimization.

To the best of our knowledge, no values have been proposed to data for DLP during CT procedures. CT examinations of the whole body are justified by the ability to detect alternative/or additional diagnoses. However, since the body contains sensitive organs, the radiation dose delivered to the patients becomes a particular concern, especially in young patients and in those with chronic diseases who undergo repeated CT studies.

Strategies to reduce the radiation dose delivered by CT have been developed and clinical investigations have shown that in several body organs disorders the diagnostic performance of CT is not decreased by the dose reduction. Reduction of the dose was first investigated in conditions characterized by intrinsic high contrast between structures, such as urethral stones, and later on in conditions characterized by intrinsic high contrast between structures. By referred to the table (5.2) of the previous studies these lowering probably reflected the increasing concern in reducing the dose as observed recently as well as technological advances in CT technology (i.e. the introduction of solid state detectors). The indication of each examination is very important to consider in order selecting the required image quality and subsequently the lowest acceptable radiation dose. As many examinations are actually performed with unnecessarily elevated radiation doses. With MDCT scanners, the ability to rapidly scan large volumes tempts the operators to increase this volume along the z-axis, and/ or to use multiple-phase CT instead of single-pass CT. Therefore, z-coverage should be adapted to the clinical indication and to the possible alternative diagnoses.

Radiation dose from head CT scans may vary considerably as a result of inherent differences in equipment and because of variations in exposure technique and scanning protocol. Previous studies where systemic changes in scanning parameters were analyzed with respect to resulting image quality have reported dose reductions of up to 40% in CT scans of the head without loss of relevant information or diagnostic image quality [Alice B 2008, Cohnen, et al. 2000].

Unjustified screening the entire i.e. Abdomen, chest, head, etc because of a "you never know" policy should thus be banished. Such policy is unacceptable in young patients who are at a low risk of having an incidental associated disease. Similarity, repeated acquisition should not be performed in circumstances where they do not specifically yield additional information.

Automatic Exposure Control (AEC) devices that are nowadays available in modern equipment module the tube current as function of the table position along the z-axis and of the image quality requested by the radiologist. Such increase it in obese and overweight patients, tending to maintain the image quality constant. Therefore, radiologists using these devices should think on terms of image quality and not of the tube current. Mulkens et al. 2005 showed that systems based on both angular and z-axis modulation reduce the mean tube current by 20% – 68% when applied to the standard MDCT protocols at constant tube current. With such systems, these authors also showed a good correlation between the mean effective tube current and patient's body mass index (BMI), with an adaptation in obese and overweight patients leading to reference tube current level being exceeded. These devices, which are only a partial response to the issue of the radiation dose. Survey studies have shown that the collective doses have increased as MDCT has replaced SDCT. However, the radiation dose has been optimized over the last decade, mainly through AEC devices and reasonable use of tube current and tube voltage preset. This was achieved thanks to the technological improvements and willpower of several study

groups to investigate the effect of dose reduction in terms of image quality and diagnostic performance. Nevertheless, as both the number of examinations and number of clinical indications for CT increase, a major effort should be made in order to optimize the radiation dose. In addition, as survey studies have shown that great variations in doses among institutions remain, a supplementary effort should be made in order to recommend standardized acquisition protocols. One of the several problems limited the study: first one is CT requested form, which was not included ideal clinical information, and some more time not justified and all cases referred for e.g. CT abdomen pelvis, brain and upper chest CT, chest abdomen CT, etc.

The clinical needs to ask, before referring a patient for MDCT, “do not really need this investigation? Will it change what I do”? If the answer to these questions is positive, the next critical question is to ask whether the information that is needed could be obtained without the use of ionizing radiation. In many CT examinations there are some applications such as ultrasound and MRI provide acceptable alternatives to MDCT, and MRI is also an effective competitor elsewhere in the body. Even where these two techniques may not be as sensitive as MDCT, there may be a case for employing them first, especially in young patients, on the basis that if they yield the required information then exposure of the patient to radiation may not be required.

5.2 Conclusions:

The assessment of radiation dose to the patients undergoing CT examinations investigated. In this study, large variations of radiation dose to various organs were observed. Different data in request form were responsible for these variations. The main contributor for high dose was the use a protocol for various CT examinations (abdomen, chest, brain, pelvis, etc) than that used in some of these countries.

5.3 Recommendations:

Clear justification of examination is highly recommended, avoid repetition of examination (CT examinations should not be repeated without clinical justification) and use of tube current modulation. If clinical situation and pathology of the patient permit, increase the pitch of examination. Requests of CT scanning must be generated only by qualified medical practitioners and justified by both the referring doctor and the radiologist. Design a proper requesting form for CT examination and rule out the responsibility of technologists, physicians and radiologists for reducing of radiation dose and image quality.

5.4 Suggestions for future studies:

Future studies should be done to optimize the radiation dose to all others CT examinations in order to establish national diagnostic protocol levels in Sudan.

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- **Appendixes:**

Data collection sheet

No	Examination	kVp	mAs	Total mAs	Total Rot. Time	CTDIvol	DLP	Slice thickness	No of slice