

Candidate's Declaration

This is “Mahfuzur Rahman Sohag” declare that Master of Science (M.S) thesis entitled as **“CTDI Volume Measurement of CT Simulation and CT Effective Dose Calculation for Cervix and Prostate”** comprises of independent investigation carried out by me at the PET/CT Division, Institute of Nuclear Medical Physics (INMP), Atomic Energy Research Establishment (AERE), Savar, Dhaka-1349. The present research work is done under the joint supervision and guidance of **Dr. M Rafiqul Islam**, Principle Scientific Officer and Director, Institute of Nuclear Medical Physics (INMP), Atomic Energy Research Establishment (AERE), Savar, Dhaka-1349 and supervision **Professor Dr. Md. Mominul Islam**, Chairman, Department of Physics, Hajee Mohammad Danesh Science and Technology University (HSTU), Dinajpur-5200. This thesis contains no material that has been submitted previously, in whole or in part, for the award of any other academic degree or diploma. Wherever the references have been made to the work of others has been clearly indicated with the source information under reference sections. Except where otherwise indicated, this thesis is my own work."

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Abstract

The accurate measurement of Computed Tomography Dose Index (CTDI) volume and the calculation of effective dose are crucial in optimizing patient safety and ensuring the efficacy of CT simulations in radiation therapy. This study focuses on evaluating the CTDI volume during CT simulations and calculating the effective dose for patients undergoing treatment for cervical and prostate cancers. By using phantoms and patient-specific data, this study aims to quantify the radiation dose distribution and assess the potential risks associated with these diagnostic procedures. The present retrospective study was conducted on 53 cervical and prostate cancer patients (6 prostate and 47 cervical) with ages ranging from 35 to 75 years. The examinations were carried out using a Philips Ingenuity Time-of-Flight (ToF) 64-slice PET/CT system (Philips Healthcare, Cleveland, OH, USA) at the Institute of Nuclear Medical Physics, Atomic Energy Research Establishment (AERE), Savar, Dhaka. The scans were performed according to institutional protocols. Dose measurements were carried out in terms of CTDI₁₀₀. CTDI₁₀₀ values for the center and periphery of the phantom were recorded, and the average value of the periphery doses was taken. Then, CTDI_w was calculated using CTDI₁₀₀ values. Since an axial CT scan was taken on the phantom, there was no patient table movement, making CTDI_{vol} values the same as CTDI_w values. The mean $\pm$ SD CTDI_{vol} was obtained as 18.879 $\pm$ 5.914 mGy, ranging from 8.952 to 37.327 mGy. The CT effective doses were calculated using the dose conversion coefficient $k=0.015 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the pelvic region, as described in the American Association of Physicists in Medicine Report No. 96. The effective doses for the CT procedure were estimated using DLP values that represent radiation exposure over the entire scan length. The scan lengths were generated by the scanner, and the DLP values were measured as the product of CTDI_{vol} and scan length. The DLP values ranged from 411.225 to 2030.96 mGy $\cdot$ cm, with a mean value of 1094.479 $\pm$ 462.029 mGy $\cdot$ cm. The effective dose is the product of the DLP and the conversion coefficient k . The mean effective dose for the pelvic region was 16.271 $\pm$ 7.180 mSv. The doses for prostate patients were 15.125 $\pm$ 8.815 mSv, and for cervical patients, they were 16.417 $\pm$ 8.930 mSv. The present study found these values to be moderately higher compared to the effective dose values of conventional diagnostic radiology examinations. This is because the CT scans were performed on the patients as part of radiotherapy treatment planning, and the scan lengths were longer than the pelvic region, resulting in higher effective doses. However, this study suggests that the risk-benefit ratio of any radiological scan should be clarified and carefully weighed.

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CHAPTER ONE

GENERAL INTRODUCTION

1.1 Introduction

Reproductive cancers are malignancies that develop in the reproductive organs [1-2]. In men, the most prevalent type is prostate cancer [3], whereas women most commonly experience breast cancer, cervical cancer, endometrial cancer, and ovarian cancer [4]. These types of cancers make up more than 26% of all cancer diagnoses, with over 5 million new cases identified annually. Additionally, reproductive cancers result in over 1.7 million deaths each year, accounting for more than 17% of all cancer-related fatalities. Due to their significant incidence and mortality rates, reproductive cancers have been extensively studied [5-6].

Molecular imaging has the capability to observe and quantitatively assess the function of biological and cellular processes within living organisms and malignancies such as tumors and cancer cells [7-10]. This swiftly advancing field holds promise for enhancing specificity and accuracy in screening and early diagnosis, tailoring therapies to individual needs, and enabling earlier monitoring of treatment progress [11]. A key benefit molecular imaging is its ability to analyze diseased tissues without the need for invasive biopsies or surgeries. This detailed information allows for the development of more personalized treatment plans. [12-13].

Computed tomography (CT) imaging is a non-invasive diagnostic tool in molecular imaging medical science that utilizes advanced X-ray technology to produce detailed cross-sectional images of the body's internal structures [14]. By rotating an X-ray source and detectors around the patient, CT collects images from various angles, which are then processed by advanced computer algorithms to create high-resolution, three-dimensional images [15-16]. This feature enables healthcare professionals to observe fine details of bones, soft tissues, and blood vessels with exceptional clarity, aiding in precise diagnosis and planning of treatments [17]. CT scans offer rapid imaging crucial for emergencies, swiftly assessing trauma, acute abdominal issues, and other urgent conditions with detailed anatomical insight for precise diagnosis and treatment planning [18-20]. CT scanners are widely accessible in healthcare, supporting efficient patient care and timely interventions [21]. They can image various body parts and organs like the head, chest, abdomen, and others, enabling doctors to diagnose a broad spectrum of conditions effectively [22-23]. CT scans guide medical procedures like

biopsies and surgeries, offering real-time imaging for precise target localization [24-25]. They detect and monitor conditions such as cancer, heart disease, and fractures, providing detailed 3D images crucial for diagnosis and monitoring cancer [26].

However, CT scans may not always aid when X-rays are inconclusive. They might not differentiate between conditions like atelectasis and tumors due to similar densities, leading to diagnostic ambiguity and limiting the effectiveness of CT numbers for accurate interpretation [27-29]. Contrast enhancement, even with dynamic scanning, often doesn't improve imaging quality. Intrathoracic tissues enhance and lose contrast simultaneously, making it hard to distinguish structures like atelectasis, mediastinal tissues, lungs, and tumors [30-31]. Similar CT numbers and frequent tumor-atelectasis adjacency cause unclear tumor boundaries, complicating accurate visualization and delineation [29]. Despite technological advancements, CT scanners have a resolution limit of around 0.5 mm, potentially missing microscopic tumor spread [32]. Artifacts like beam hardening, motion, and metal can affect image quality. High costs can limit access, prompting efforts to improve cost-effectiveness and explore alternatives [33-34]. In some instances, intravenous contrast agents are administered to improve CT scan images [35]. Although typically safe, these agents can carry a risk of allergic reactions or renal complications, especially in patients with pre-existing renal dysfunction or allergies [36]. CT scanning entails exposure to ionizing radiation, which is a concern, especially for patients needing multiple scans or those more sensitive to radiation. Radiologists and technologists must follow radiation dose optimization protocols to guarantee patient safety. [37-39] As CT usage in clinical settings becomes more widespread, the radiation dose from these scans is increasingly contributing to the overall radiation exposure of patients and the general population [40-41]. Even within a single clinical application, patient doses can vary greatly due to the use of different imaging protocols and the inherent differences between models and manufacturers of CT scanners [42-45].

Data from various national surveys have proved that the contribution of CT to the total collective dose can be as high as 67% in the United States and 40% in Europe, although CT examinations represent only about 11% and 4%, respectively, of all radiologic examinations. Therefore, the dose levels delivered in CT examinations should be known and available to patients and their physicians [46-49]. The radiation dose patients receive from CT exams can vary greatly due to the diverse imaging protocols used and the inherent differences between various CT scanner makes and models [43-44]. Thus, it is essential for the medical imaging community to have practical tools to convert the dose quantities from commercial CT

scanners into a comprehensible and meaningful dose metric. Effective dose is widely used and understood as an indicator of radiation received by patients during radiological exams [50-53].

Effective dose is a single parameter meant to reflect the relative risk from exposure to ionizing radiation. It reflects the risk of detrimental biologic effects from a non-uniform, partial-body exposure in terms of a whole-body exposure [54-55]. It is mainly utilized to evaluate adherence to dose limits established for public and worker safety. Additionally, it can be beneficial for comparing the risks to a reference patient from various imaging methods and procedures [56-57]. Typical values of effective doses are 1-2 mSv for head CT, 5-7 mSv for chest CT, 8-12 mSv for abdomen and pelvic CT, 10-20 mSv for chest, abdomen and pelvis CT, 12-15 mSv for cardiac CT, 5-15 mSv for CT pulmonary angiography [58-60].

The computed tomography dose index (CTDI) is the most widely used quantities for measuring effective dose in CT dosimetry [61]. The computed tomography dose index (CTDI), first suggested by Jucius and Kambic and formalized by Shope et al., has been the standard measurement of radiation dose in CT since the 1980s [62-63]. Although variants of CTDI, including $CTDI_{FDA}$, $CTDI_{100}$, $CTDI_{300}$, and $CTDI_{\infty}$, exist, they are conceptually equivalent to the original definition by Shope et al., differing only in the measurement range [63-66]. Current universal dose metrics include volume CT dose index ($CTDI_{vol}$) and dose-length product (DLP). DLP accounts for scan length but not patient size. $CTDI_{vol}$ measures only scanner radiation output to a defined phantom. Neither $CTDI_{vol}$ nor DLP is suitable to estimate individual patient absorbed dose or to estimate individual patient effective dose [50, 67].

Through an in-depth exploration of CTDI volume measurement and effective dose calculation techniques, this thesis aims to contribute to the refinement of radiation dose optimization strategies in CT simulation for cervix and prostate examinations. By elucidating the dosimetric considerations and challenges inherent in these imaging procedures, this research endeavors to enhance patient safety, improve treatment accuracy, and ultimately optimize clinical outcomes in the field of oncologic radiation therapy.

1.2 Motivations of the Current Investigation

With the increasing prevalence of chronic diseases and a growing population, the demand for Computed Tomography (CT) imaging services in Bangladesh is on the rise. Radiation exposure from computed tomography (CT) scans can result in both stochastic effects such as

cancer and genetic mutations, and deterministic effects such as radiation burns, nausea, vomiting, cataracts, organ damage, radiation-induced hair loss [68-71].

Determination of radiation dose imparted to patients from CT scanning, monitoring and reduction of dose and finally assuring appropriate image quality to uplift the effective use of radiation are focused in the current study. It is important to spark attention over the long-term effect of CT because the organ doses from CT are found typically 100 times larger than those from conventional x-ray imaging [72].

1.3 Objectives and Methods of the Study

To measure computed tomography dose index (CTDI) volume that indicates the amount of radiation dose delivered by a CT scanner machine and then CT effective dose for cervix and prostate by dose-length-product (DLP) method.

The CTDI volume is first determined on a cylindrical polymethyl methacrylate (PMMA) phantom. Then this value is multiplied by the scan length of the individual cervical and prostate cancer patients during CT scanning for radiotherapy treatment planning to obtain dose-length-product (DLP) value. Then dose conversion factor for pelvic region is chosen. The effective dose of CT simulation is then calculated by multiplying DLP values with the dose conversion factor.

1.4 Importance of the Present Research Work

CTDI volume is a standardized measure of radiation dose output of a CT scanner that helps in assessing the amount of radiation exposure a patient receives during a CT scan [73]. Calculation of CTDI volume provides a uniform way to compare radiation doses between different CT scanners and also help to adjust and optimize scan protocols to minimize radiation exposure without compromising diagnostic quality [74]. Calculating CTDI volume ensures adherence to dose limits set by regulatory bodies and facilitates monitoring for compliance so that it is possible to maintain the radiation doses within safe limits, reducing the risk of radiation-induced injuries or long-term effects like cancer [75-76].

Effective dose is a measure that reflects the biological risk of radiation exposure, taking into account the type of radiation and the sensitivity of different tissues and organs [77]. Effective dose calculation provides a better understanding of the potential health risks associated with radiation exposure by considering the varying sensitivities of different organs [78-79]. It Allows comparison of radiation risks from different imaging modalities and procedures and

assists in managing and justifying the use of radiation, ensuring that the benefits outweigh the risks, especially in repeated or high-dose procedures [80].

1.5 Literature Review

E. Manssor et al. [81] estimated radiation doses during chest, abdomen and pelvis CT for 51 patients. The mean age of the patients was 48.0 ± 18.6 y, the mean patient weight was 73.8 ± 16.1 kg, the mean dose-length product was 1493.8 ± 392.1 mGy cm, Volume CT dose index ($CTDI_{vol}$) was 22.94 ± 5.64 mGy and the mean effective dose was 22.4 ± 5.9 mSv per procedure.

I Pantos et al. [82] calculated adult patient radiation doses from non-cardiac CT examination. They have used DLP method to calculate effective dose. The effective doses were 1.9 mSv (0.3–8.2 mSv) for head, 7.5 mSv (0.3–26.0 mSv) for chest, 7.9 mSv (1.4–31.2 mSv) for abdomen, and 7.6 mSv (2.5–36.5 mSv) for pelvic region.

Vicky Goh al. [83] determined radiation doses delivered during volumetric helical perfusion CT of the thorax, abdomen or pelvis over 42 consecutive examinations. Mean DLP was 1288.8 mGy. cm (range: 648 to 2456 mGy.cm), mean $CTDI_{vol}$ was 96.2 mGy (range: 32.3 to 169.4 mGy), mean effective dose was 19.6 mSv (range: 12.3 mSv to 36.7 mSv). They compared mean DLP and effective dose to be 885.2 mGy.cm (range: 504 to 1633 mGy.cm) and 13.3 mSV (range: 7.8 to 24.5 mSv) respectively for the standard staging CT thorax, abdomen and pelvis.

Richard D. Nawfel et al. [84] measured mean patient skin doses for CT urography with TLD strips and calculated from phantom data (CT dose index) were $56.3 \text{ mGy} \pm 11.5$ and $54.6 \text{ mGy} \pm 4.1$, respectively. Mean patient skin doses for conventional urography measured with TLD strips and calculated as entrance skin dose were $151 \text{ mGy} \pm 90$ and $145 \text{ mGy} \pm 76$, respectively. Correlation coefficient between measured and calculated skin doses for conventional urography examinations was 0.95. Mean effective dose estimates for CT urography and conventional urography were $14.8 \text{ mSv} \pm 9.0$ and $9.7 \text{ mSv} \pm 3.0$, respectively. Mean effective doses estimated for the pelvic-torso phantom were 15.9 mSv (CT urography) and 7.8 mSv (conventional urography).

Ezinma Helen O. et al. [85] determined $CTDI_{vol}$ for 61 patients where 39 underwent cranial CT and 22 patients underwent abdomino-pelvic CT. They utilized DLP method and find the total $CTDI_{vol}$ ranged from 135.6 to 215.5 mGy for cranial CT and 26.1 to 124.4 mGy for

abdomino-pelvic, while DLP ranged from 2631.0 to 6542.8 mGy.cm for cranial and 1147.5 to 5441.8 mGy.cm for abdomino-pelvic CT.

Colin J. Martin et al. [86] carried out large-scale computed tomography (CT) patient dose surveys from the USA, UK and Germany undertaken during the period 1999–2001. They reported values for the three surveys were similar with a mean $CTDI_{vol}$ of 59 ± 6 mGy and a mean DLP of 760 ± 90 mGy cm. For the body, the values for $CTDI_{vol}$ in the USA (21 ± 5 mGy) were nearly double those in the European countries (12 ± 2 mGy). Average head $CTDI_{vol}$ values are comparable with the diagnostic reference levels (DRLs) recommended by the European Commission and the UK Health Protection Agency, but lower than the value of 75 mGy, which is currently used in the USA. DRLs for abdominal imaging are nearly a factor of 2 lower in the UK (14 mGy) than in other parts of the European Community and the USA (25 mGy).

C. Anam et al. [87] finds the differences between the calculated $CTDI_{vol}$ and the $CTDI_{vol}$ reported by the CT scanner were $4.4 \pm 1.2\%$ and $6.0 \pm 2.0\%$ for abdominal and thoracic examinations, respectively. The average tube current was found to be linearly correlated with D_w with R^2 values of 0.707 and 0.696 for abdominal and thoracic examinations, respectively.

K FUJII et al. [88] investigated and compared patient doses in pediatric and adult CT examinations. In chest CT examinations, organ doses for organs within the scanning area were 2–21 mGy for children and 7–26 mGy for adults. Thyroid doses for children were frequently the highest with a maximum of 21 mGy. In abdominal CT examinations, organ doses for organs within the scanning area were 3–16 mGy for children and 10–34 mGy for adults. Effective doses evaluated for children and adults were found to be proportional to the effective mAs of CT scanners, where linear coefficients were specific to the types of CT examinations and to the manufacturers of CT scanners.

Jian Xu et al. [89] compared the metrics of volume computed tomography index ($CTDI_{vol}$) and size specific dose estimate (SSDE), and quantity the differences in head CT examinations of adult patients. A total of 4239 SSDE(s) values were obtained from all patients. $CTDI_{vol}$ value was higher than $SSDE_{aver}$ value in 136 patients and was lower than $SSDE_{aver}$ value in 20 patients, and both estimated dose values were equal in only 1 patient. The WED increased by about 5.25% compared with the phantom diameter of 16 cm, resulting in an increase in $CTDI_{vol}$ value by about 2.99% compared to $SSDE_{aver}$ value. The average $CTDI_{vol}$ and $SSDE_{aver}$ values were 55.10 ± 3.06 mGy and 53.50 ± 2.29 mGy, respectively, the difference was

statistically significant ($t=15.54$, $P=0.000$). The distribution range of $CTDI_{vol}$ values was wider and covered that of $SSDE_{aver}$ values.

Hussin et al. [90] measured the effective dose from PET-CT scanner over 93 pediatric patients with ages ranging from 2 to 15 years. 68 patients of which undergone the scan with the use of the Automatic Exposure Control (AEC) function and 25 patients gone without it. The effective doses from PET and CT examinations were calculated based on ICRP Publications 106 and 102. They reported the average effective doses for PET and CT examinations with and without AEC as 6.03mSv and 4.96 mSv, and 6.67 mSv and 6.77 mSv, respectively. The total effective doses of PET-CT with and without AEC were 10.99 and 13.44 mSv respectively.

Issahaku Shirazu et al. [91] estimated CTDI and DLP values for patients' dose optimization procedures. Technical parameters were obtained for three groups of randomly selected patients undergoing abdominal CT examinations of 1320 patients of age 20-80 years. They showed that the mean CTDI volume and DLP values were 6.33mGy and 936.25 mGy-cm.

Kim et al. [92] measured and compared radiation doses from three imaging modalities- X-ray, CT, and PET/CT for pediatric patients with neuroblastoma by estimating the effective dose for each system. For 63 patients with a mean age of 3.243.5 years, the mean radiation dose per exam for this group was 0.04-0.19 mSv for X-ray, 1.09 ± 1.11 mSv for conventional CT, and 8.3547.45 mSv for PET/CT.

Khamwan et al. [93] calculated the effective dose for 35 oncology patients undergoing "F-FDG PET/CT scan in Thailand. The age range of the patients was from 28 to 60 years. The patient absorbed dose and effective dose calculations were done using the Medical Internal Radiation Dosimetry method and Monte Carlo simulation. The average effective dose from PET was 4.4040.61 and from CT was 14.45 ± 2.82 mSv, with a total patient dose of 18.85 mSv.

Hamid Osman et al. [94] measured the CT dose of the abdomen and pelvic examinations for adult patients, as well as establishes the provincial diagnostic reference level (DRL) for hospitals in Taif city, Saudi Arabia. A total of 200 patients who underwent pelvis and abdomen examination. The patients' average age and BMI were 37.3 years and 25.5 kg/m², respectively. The mean DLP and $CTDI_{vol}$ for the pelvis were found to be 379.6 mGy-cm and 10.8 mGy, respectively, whereas the corresponding mean values for the abdomen were 369.5 mGy-cm and 11.5 mGy, respectively.

1.6 Thesis Outline

Chapter One: Importance and objective, motivation, research background related to this work is illustrated.

Chapter Two: Radiation exposure, different dosimetric quantities and their units, various health effects caused by ionizing radiation and their mechanisms and estimation of damage effects is discussed.

Chapter Three: Theory, working principle of computed tomography and its instrumentation that is used in the measurement of the effective dose is delineated.

Chapter Four: The role of computed tomography in radiotherapy treatment planning is portrayed.

Chapter Five: In this chapter the methodology of the CTDI volume measurements, and procedures of effective dose calculation is shown.

Chapter Six: Results of the calculation of CTDI volume and effective dose and brief discussion on the obtained results is given.

Chapter Seven: This chapter includes the conclusion and the outcome of the study.

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CHAPTER TWO

BIOLOGICAL EFFECTS OF IONIZING RADIATION

2.1 Introduction

The biological effects of ionizing radiation are complex and can vary depending on factors such as the type of radiation, dose received, and the tissue or organ exposed [1-2]. Ionizing radiation has the ability to penetrate living tissue and cause damage by directly interacting with atoms and molecules, leading to the ionization of atoms and the generation of free radicals. This can result in DNA damage, which may lead to mutations, cell death, or carcinogenesis. Acute exposure to high doses of ionizing radiation can cause immediate effects such as radiation sickness, while chronic exposure to lower doses may increase the risk of developing cancer over time [3-4].

The severity of effects depends on the radiation dose, with high doses causing immediate issues like skin burns and nausea. Lower doses may not have immediate effects but can increase the risk of cancer later in life. These long-term effects are more likely in rapidly dividing cells, which is why radiation therapy can target and kill cancer cells [5]. Additionally, ionizing radiation can affect reproductive cells, potentially causing hereditary effects in future generations. Protective measures such as shielding and dose monitoring are essential in minimizing the biological effects of ionizing radiation exposure [6].

2.2 Ionizing Radiation

Ionizing radiation is a form of energy with enough punch to rip electrons from atoms and molecules. This type of radiation can be classified into two main categories: directly ionizing radiation and indirectly ionizing radiation. Directly ionizing radiation interacts with matter directly, dispersing its energy through numerous small electrostatic interactions along its path. An example of directly ionizing radiation is the α -particle, which directly ionizes matter upon interaction. Indirectly ionizing radiation initially transfers its energy to charged particles within the traversed matter [7]. These charged particles subsequently convey the energy to the surrounding matter. Electromagnetic radiations, such as X-rays and gamma rays, as well as neutrons, which are uncharged particles, initially transfer their energy to electrons and protons within the matter. These electrons and protons then transmit the energy to the surrounding matter [8].

As ionizing radiation propagates through matter, it disrupts the delicate balance within cells, directly damaging DNA or creating harmful free radicals. This cellular mayhem can hinder cell division and function, potentially leading to cell death. The severity of effects depends on the amount of radiation absorbed (the dose). High doses, like those experienced in a nuclear accident, can cause immediate cell death and radiation sickness. Lower doses, however, may not have immediate consequences but can increase the risk of cancer later in life. This is why controlling radiation exposure is paramount in medical applications like X-rays and CT scans, where the benefits of diagnosis outweigh the minimal risks associated with low-dose radiation [9].

2.3 Radiation Exposure

Radiation exposure refers to the absorption of ionizing radiation by living organisms or materials [10]. We encounter radiation constantly, both from natural and human-made sources. Natural background radiation includes cosmic rays from space, radon gas from the earth's crust, and ultraviolet (UV) rays from the sun. Human activities like X-ray machines, nuclear power plants, and even smoke detectors contribute additional radiation exposure [11]. The key factor determining health effects is the type and amount of radiation absorbed (dose). High doses of ionizing radiation, like those in a nuclear accident, can cause immediate cell death and radiation sickness. Lower doses, however, may not have immediate consequences but can increase the risk of cancer later in life. This is why managing radiation exposure is crucial [12].

Medical professionals carefully balance the diagnostic benefits of X-rays and CT scans with the minimal risks associated with low-dose radiation. In daily life, limiting exposure to sources like UV rays through sun protection helps reduce risks. Understanding radiation exposure and its potential effects empowers us to make informed choices about our health and environment. The conventional unit of exposure is the Roentgen (R), and the SI unit is Coulomb per Kilogram in the air (C/Kg):

$$1 \text{ R} = 2.58 \times 10^{-4} \text{ C/Kg}$$

$$1 \text{ C/Kg} = 3877 \text{ R.}$$

2.3.1 Types of Radiation Exposure

Depending upon different exposure pathways, radiation exposure can be divided into two categories [13]: internal and external.

Internal exposure occurs when radioactive materials are inhaled, ingested, or absorbed through the skin. This can happen through contaminated dust, food, or water, or even certain medical procedures [13]. Radioactive isotopes then lodge inside the body, continuously emitting radiation and damaging surrounding cells [14].

External exposure involves being bombarded by radiation from outside the body. X-rays, gamma rays, and neutrons are common culprits here. These high-energy particles or waves can penetrate tissues to varying degrees, depending on their energy and the body's composition. The severity of effects depends on the type, duration, and intensity of the exposure. High doses from external sources can cause immediate harm like skin burns or radiation sickness, while lower doses may increase the risk of cancer over time [15].

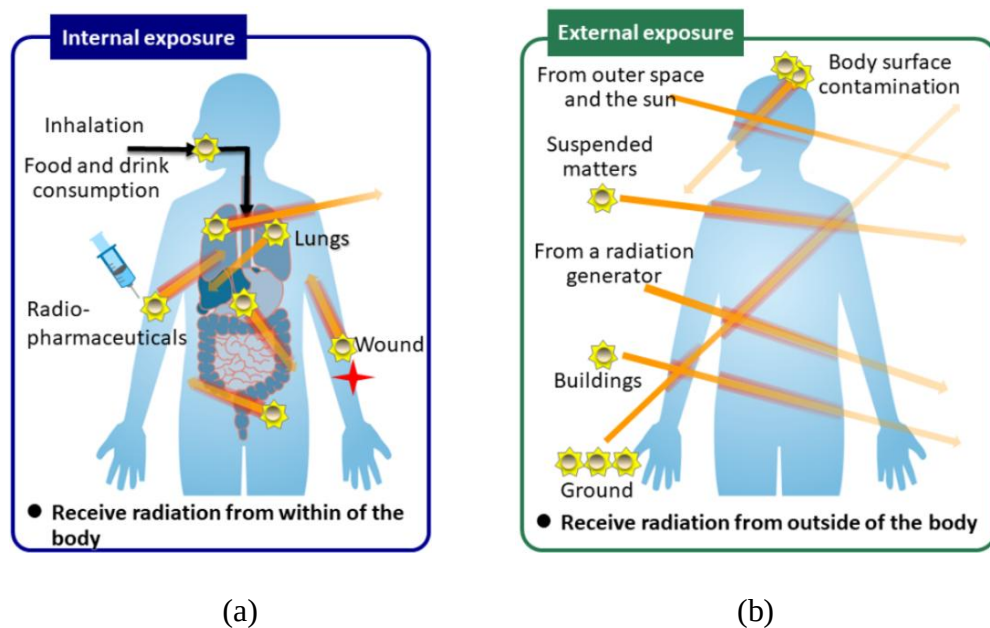


Figure-2.1: (a) Internal Radiation Exposure (b) External Radiation Exposure [17]

Both internal and external exposure highlights the importance of radiation safety measures. Medical professionals take precautions to minimize radiation during procedures, and regulations govern safe handling of radioactive materials to protect workers and the public [16].

2.4 Basic Dosimetric Quantities

2.4.1 Absorbed Dose

Absorbed dose is the energy deposited by ionizing radiation per unit mass of a material, usually measured in an organ or tissue. It is key in radiation protection, quantifying the impacts of radiation on living organisms and materials. Ionizing radiation, such as X-rays, gamma rays, and alpha and beta particles, imparts energy through ionization and excitation, potentially causing cellular and molecular damage [18-19].

The unit of absorbed dose is the Gray (Gy), which is defined as one joule of energy absorbed per kilogram of matter. Another unit of absorbed dose, the rad, which is acronym for radiation absorbed dose, is still used in some countries, where 1 Gy of absorbed dose is equivalent to 100 rad [20]. Absorbed dose is a key factor in the assessment of radiation exposure and its impacts on health. It is crucial in medical applications like radiation therapy, radiation protection etc. where controlled doses of ionizing radiation are used to treat abnormal cells, such as, tumors or cancer cells. However, excessive absorbed doses can result in acute radiation sickness and long-term health issues, including cancer [19].

2.4.2 Equivalent Dose

Precise delivery of radiation doses to bodily tissues is crucial in both diagnostic and therapeutic procedures. However, it's essential to recognize that different types of radiation can have varying effects on tissues, even when the dose quantity remains constant. For instance, α -rays pose a greater risk to tissues due to their heightened ionization capacity compared to γ -rays [21-22]. Consequently, to account for these discrepancies in the biological impacts of different radiation types, the International Committee for Radiological Units (ICRU) and the International Commission on Radiological Protection (ICRP) introduced the concept of a Quality Factor. This factor, denoted as 'Q', serves as a dimensionless measure or numerical representation of a particular radiation's effectiveness when interacting with the human body. Subsequently, the term "Quality Factor" was replaced with a new term, Radiation Weighting Factor ' W_R ' [23].

Table-2.1: Radiation Types with their Radiation Weighting Factors [24]

Radiation Type and Energy Range	Radiation Weighting Factor (W_R)
X and γ rays, all energies	1
Electrons, positrons and muons, all energies	1
Neutrons:	
< 10 keV	5
10 keV to 100 keV	10
> 100 keV to 2 MeV	20
> 2 MeV to 20 MeV	10
> 20 MeV	5
Protons, (other than recoil protons) and energy > 2 MeV	2-5
α particles, fission fragments, heavy nuclei	20

ICRU introduced a new parameter known as Equivalent Dose as a consequence of introducing the radiation weighting factor [23]. This quantity is determined by multiplying the absorbed dose by the radiation weighting factor. Mathematically, Equivalent Dose 'H' is expressed as follows.

$$H = \sum D.W_R \quad (2.1)$$

Equivalent dose is measured in the units of Sievert ' S_v '.

2.4.3 Effective Dose

Another crucial consideration in the utilization of radiation for diagnostic and therapeutic purposes in the human body is the response of tissues to radiation exposure. Despite administering equivalent doses to different body parts, the sensitivity of various tissues varies. Certain tissues exhibit greater sensitivity to radiation, whereas others are comparatively less sensitive [25]. Consequently, the damage inflicted on different tissues differs, even when the equivalent dose remains constant. To address this variability in tissue response to radiation, the International Commission on Radiological Units and Measurements (ICRU) introduced a new parameter known as the Tissue Weighting Factor (W_T) [26].

Table-2.2: Name of Tissues or Organs with Corresponding Weighting Factors [27]

Tissue or Organ	Tissue Weighting Factor (W_T)
Bladder	0.04
Bone marrow (red)	0.12
Bone surface	0.01
Brain	0.01
Breast	0.12
Colon	0.12
Esophagus	0.04
Gonads	0.08
Liver	0.04
Lung	0.12
Salivary glands	0.01
Skin	0.01
Stomach	0.12
Thyroid	0.04
Subtotal	0.88
<i>Remainder</i>	0.12
Total	1.00

This factor accounts for the varying sensitivities of different tissues to radiation. Incorporating W_T into radiation exposure considerations yields a new quantity termed Effective Dose (E). The effective dose is calculated by multiplying the equivalent dose by the tissue weighting factor of the irradiated tissue, thereby providing a more comprehensive assessment of radiation-induced effects on the human body [28]. Mathematically,

$$E = \sum H \cdot W_T \quad (2.2)$$

Since from equation (2.1),

$$H = \sum D \cdot W_R$$

Then equation (2.2) becomes,

$$E = \sum D \cdot W_R \cdot W_T \quad (2.3)$$

Effective dose is measured in the units of Sievert ' S_v '.

2.5 Units of Different Dosimetric Quantities

2.5.1 Gray

The gray (Gy) is the SI unit of absorbed dose and is defined as the absorption of one joule of energy, in the form of ionizing radiation, per kilogram of matter. That is, 1 gray = 1 J/kg. One gray is a large unit and is usually used with either milligray (mGy) or microgray (μ Gy) [29].

2.5.2 RAD

The RAD is an acronym for “radiation absorbed dose” and was introduced before the introduction of the SI units. The rad is defined as the dose represented by 100 ergs of energy being absorbed by one gram of matter. The erg is the unit of energy in the CGS system, where one erg = 1.0×10^{-7} joules. When converted into SI units, one rad = 0.01 Gy = 0.01 J/Kg or alternatively one gray = 100 rads [30].

2.5.3 Roentgen

The conventional unit for exposure is Roentgen (R). One roentgen is defined as the amount of x or gamma radiation required to produce ions carrying a charge of 1 electrostatic unit (esu) per cubic centimeter (2.58×10^{-4} Coulomb/Kg) of dry air under standard conditions [31].

Thus, 1 R = 2.58×10^{-4} Coulomb/Kg.

2.5.4 Sievert

The Sievert is the SI unit for measuring ionizing radiation effective dose, which accounts for relative sensitivities of different tissues and organs exposed to radiation. The radiation quantities measured by the sievert are called equivalent dose and effective dose. The larger the value expressed in sieverts, the larger the effects of radiation to which the human body is exposed. The dose equivalent in sieverts is equal to the product of the absorbed dose in grays and the quality factor. The sievert is inconveniently large for various applications, and so the millisievert (mSv) is frequently used. One millisievert corresponds to 10 ergs of energy of gamma radiation transferred to one gram of living tissue. Thus, 1 mSv = 1 mGy [32].

2.5.5 Roentgen Equivalent Man

The rem is a unit of measurement that takes into account different biological responses to different kinds of radiation. The radiation quantity measured by the rem is the equivalent

dose. The rem has the same relationship with Sievert as rad has with Gray. 1 Sv equals 100 rem, and 1 rem equals 10 mSv [33].

2.6 Health Effects of Ionizing Radiation

Health effects of ionizing radiation in human body occur either in the irradiated individuals themselves (somatic effects) or in their descendants (hereditary or genetic effects). Somatic effects are divided into deterministic effects (also known as tissue reactions) and stochastic effects, where hereditary and genetic effects are all of stochastic origin only. The damage caused by ionizing radiation to any organs or tissues on human body depends upon the dose of radiation received, type of radiation and the radiosensitivity of different tissues and organs [34]. The deterministic and stochastic health effects are discussed below.

2.6.1 Deterministic Health Effects

Deterministic health effects of radiation, also known as non-stochastic effects, are characterized by a clear dose-response relationship. These effects occur only after a certain threshold dose is crossed, meaning there's a minimum amount of radiation needed for them to appear. Below this threshold, our bodies are resilient enough to handle the exposure without any noticeable consequences [35].

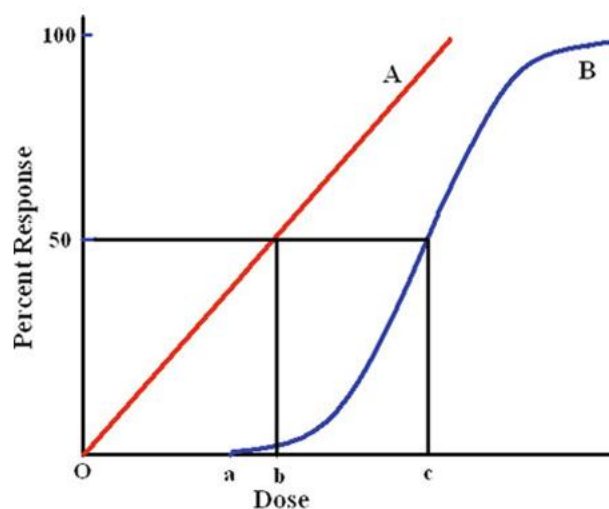


Figure-2.2: Curve A and B illustrates Deterministic and Stochastic effects respectively [36]

As long as radiation levels stay below a certain point, there's no impact. However, once the threshold is surpassed, the effects become increasingly severe with higher doses. These effects primarily stem from the direct damage radiation inflicts on rapidly dividing cells. Since these cells are constantly replicating their DNA, they're more vulnerable to the

disruptive effects of radiation. The severity of the effect hinges on the radiation dose absorbed by the tissue. The higher the dose, the more pronounced the damage. Radiation sickness, for instance, with symptoms like nausea and vomiting, worsens progressively with increasing radiation exposure [37]. Due to the clear dose threshold, deterministic effects are predictable. Scientists and medical professionals can establish safe radiation exposure limits to minimize or completely prevent them. This predictability allows for controlled use of radiation in medical treatments like X-rays and CT scans, where the benefits outweigh the minimal risks associated with low-dose exposure.

Common examples of deterministic effects include skin reddening (erythema) from excessive radiation exposure, hair loss, cataracts in the eyes, and suppression of bone marrow activity leading to a weakened immune system. At extremely high doses, acute radiation syndrome can develop, causing rapid cell death and multiple organ failure [38].

2.6.2 Stochastic Health Effects

Stochastic health effects of radiation refer to those effects that occur by chance and whose probability increases with the dose of radiation but do not exhibit a threshold below which they do not occur. Unlike deterministic effects, which have a clear dose-response relationship and a threshold below which they do not manifest, stochastic effects do not have a minimum dose threshold. Instead, they are characterized by a probability of occurrence that increases with the dose, meaning that even low doses of radiation exposure carry a risk of causing stochastic effects. These effects include the induction of cancer and hereditary genetic mutations [39].

Cancer is the most widely studied stochastic effect of radiation exposure, with extensive research linking ionizing radiation to an increased risk of various types of cancer, including leukemia, thyroid cancer, and solid tumors. While the probability of developing cancer as a result of radiation exposure increases with dose, the exact risk for an individual depends on factors such as age at exposure, duration of exposure, and genetic susceptibility [40].

Hereditary effects refer to mutations passed on to offspring as a result of radiation exposure to reproductive cells. Stochastic effects underscore the importance of radiation protection measures and regulatory guidelines aimed at minimizing radiation exposure to reduce the risk of long-term health consequences [41].

2.7 Effect of Ionization Radiation on Human Cells

Ionizing radiation of high energy, such as α -particles, γ -rays, X-rays, protons, and heavy ions, induces diverse damages to living cells, potentially resulting in the loss of genetic material, genetic mutations, the exacerbation of genomic instability, and cellular apoptosis. Among the array of cellular lesions caused by high-energy ionizing radiation, those that significantly disrupt or halt cellular functions, particularly those impacting the human genome, are regarded as the most prevalent and severe [42].

These encompass single-strand and double-strand breaks, DNA-protein cross-links, the release of bases, and other alterations in chemical structure, as well as the creation of multiply damaged regions within DNA. Such harm can occur either through direct interaction of radiation with individual components of DNA or indirectly through the interaction of reactive species generated from surrounding molecules [43].

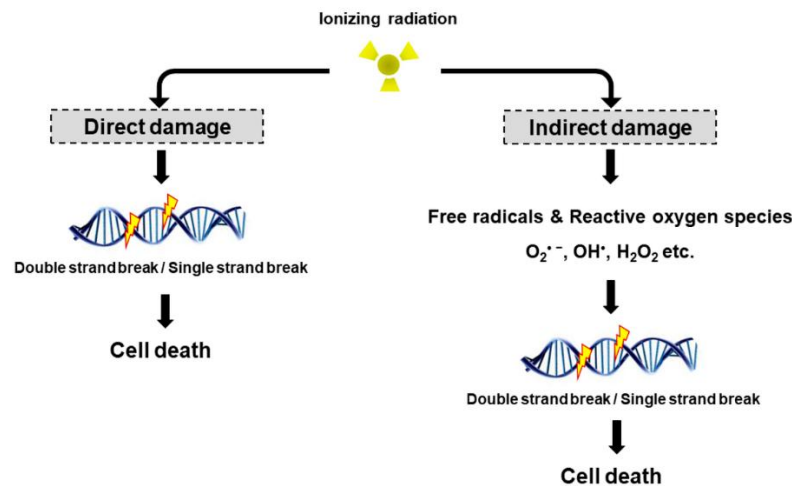


Figure-2.3: Direct and Indirect effect of cell damage caused by radiation [44]

This latter process, commonly termed the indirect effect, involves radiation interacting with the local molecular environment surrounding the DNA molecule, including water, salts, proteins, and oxygen molecules. The indirect effect primarily stems from the reaction of radiolysis products of water (such as hydroxyl radicals, solvated electrons, and hydrogen atoms) with DNA. It is generally assumed that damage to the human genome caused by high-energy radiation is approximately one-third direct and two-thirds indirect [1].

In both direct and indirect effects, the energy transferred to the biological medium primarily occurs through ionization, creating numerous secondary entities along the path of radiation, such as ions, radicals, and secondary electrons.

2.8 Mechanisms of Radiation Effects

2.8.1 Direct Effect

The direct effect of ionization on human cells occurs when ionizing radiation interacts directly with cellular components, such as DNA, proteins, and membranes, leading to molecular damage and biological consequences [45-46]. This interaction occurs when radiation deposits energy within the cell, causing ionization and the formation of reactive chemical species, including free radicals and ions [47-48]. These reactive species can directly damage DNA by causing breaks in the DNA strands, altering bases, or inducing cross-links between DNA molecules, disrupting the cell's ability to replicate and repair its genetic material accurately [49-50].

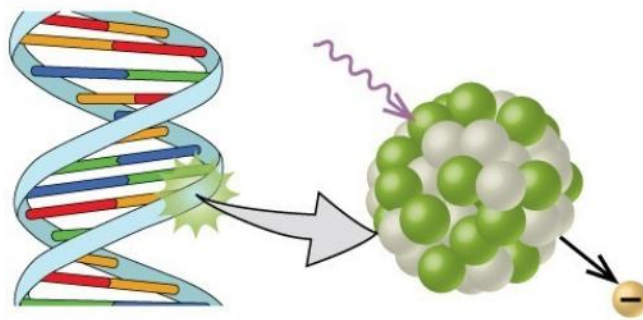


Figure-2.4: Direct effect of radiation [51]

The direct ionization effect can lead to damage to other cellular structures and molecules, such as proteins and lipids, affecting cell function and viability [52]. The extent of damage depends on various factors, including the type and energy of the radiation, the dose received, and the sensitivity of the cell type [53]. Understanding the direct effects of ionization on human cells is crucial for assessing the risks associated with radiation exposure, developing effective radiation therapy strategies, and implementing radiation protection measures in medical, occupational, and environmental contexts.

2.8.2 Indirect Effect

The indirect effects of ionization on human cells refer to the biological consequences that arise not from direct interaction between ionizing radiation and cellular components, but rather from the generation of secondary reactive species within the cellular environment [54]. When ionizing radiation passes through biological tissue, it can ionize water molecules,

leading to the formation of highly reactive free radicals such as hydroxyl radicals ($\bullet\text{OH}$). A free radical is a fragment of a compound or an element that contains an unpaired electron. These free radicals are capable of causing oxidative damage to cellular components, including DNA, proteins, and lipids [55].

When radiation interacts with water, two types of free radicals are formed:

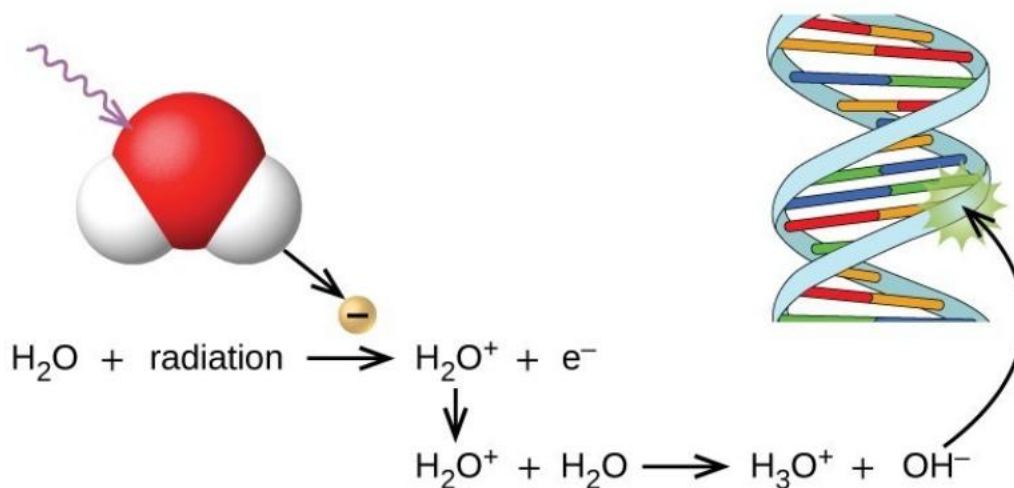
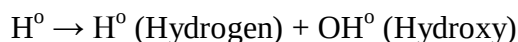
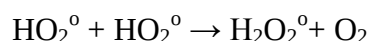


Fig-2.5: Indirect Effect of radiation [56]

The presence of an excess of oxygen during irradiation of cells allows the formation of additional free radicals:



In some cases, free OH radicals combine before they combine with free H, leading to the formation of hydrogen peroxide:



It is worth noting that antioxidants block hydroxyperoxy free radical combination into highly unstable hydrogen peroxide (H_2O_2). H_2O_2 has a short lifetime of the order of microsecond and can affect molecules or cells that did not suffer radiation damage directly. It can also cause damage to the structure of DNA, proteins and enzymes, creating subsequent biological effects of radiation such as cancers or genetic mutations that may arise later [57].

2.9 Radiation Damage to DNA

When cells are exposed to radiation, the main culprit behind cell death is damage to their DNA. This radiation exposure can cause a variety of injuries to the DNA itself, including breaks in one or both strands, damage to the individual building blocks (bases), and even the creation of links between DNA and proteins or between proteins themselves. These various forms of DNA damage disrupt the cell's ability to function properly, and in severe cases, can lead to cell death [58-59].

Different types of DNA damages are briefly outlined below:

2.9.1 Single Strand Breaks

Radiation-induced single-strand breaks in DNA occur when the backbone of one strand of the DNA molecule is severed. This type of damage can result from exposure to ionizing radiation, such as X-rays or gamma rays, which have sufficient energy to disrupt the chemical bonds holding the DNA molecule together, usually a break of the sugar-phosphate chains [60]. Single-strand breaks can lead to genetic mutations, interfere with DNA replication and transcription processes, and contribute to cell death. Repair mechanisms within cells can often mend single-strand breaks, but if left unrepaired or improperly repaired, they can pose risks to genomic stability and cellular function. Normally 90% of this damage repairs within 1 hour at 37° C [61].

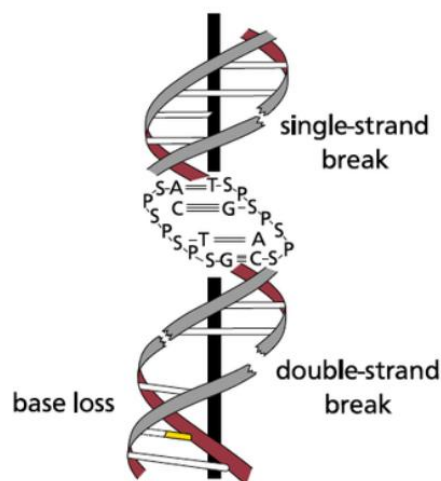


Figure-2.6: Different types of DNA damage [44]

2.9.2 Double Strand Breaks

This form of damage affects both strands of the DNA helix, causing breaks that occur opposite each other or within a short distance of a few base pairs. Repairing double strand breaks presents a greater challenge due to their complexity [62]. Nevertheless, the DNA molecule is inherently resilient due to its packaging with supportive proteins that maintain its structure and prevent disintegration, even in instances of breaks occurring on both strands of the helix [63]. Complex organisms like humans have evolved various mechanisms to address these double strand breaks [64]. However, it is evident that repairing such breaks is more arduous and often leads to cell death and observable chromosomal damage [65].

2.9.3 Base Damage

Base damage to DNA occurs when individual nucleotide bases within the DNA molecule are chemically altered or physically damaged [66]. This damage can arise from various sources such as exposure to ionizing radiation, ultraviolet light, certain chemicals, or metabolic processes within the cell. Common forms of base damage include the formation of thymine dimers caused by UV radiation, oxidation of guanine bases, and alkylation of bases by chemicals [67]. Base damage can interfere with the accurate replication and transcription of DNA, leading to mutations if not repaired promptly. Cells have sophisticated repair mechanisms to detect and correct base damage, helping to maintain genomic integrity and prevent the accumulation of mutations that could contribute to diseases like cancer. Understanding base damage and the repair mechanisms involved is crucial in elucidating the molecular basis of disease and developing therapeutic strategies to mitigate its effects [68].

2.9.4 Clustered Damage

In clustered damage, a variety of scenarios can occur, including one where two neighboring bases, T and T, on the same strand undergo chemical alterations [69]. This represents just one among numerous potential outcomes. What remains consistent across these possibilities is that two or more damaged sites are situated close to each other [70-71].

2.10 Concepts of Cell Death

Exposure to radiation doses in the range of several sieverts can result in cellular death. This "cell death" in radiation terms refers to the loss of reproductive potential, not necessarily immediate physical destruction. Several processes can lead to this loss, including programmed cell death (apoptosis), uncontrolled cell death (necrosis), abnormal cell division

(mitotic catastrophe), and a permanent state of non-division (induced senescence). While all except the last process eventually result in the physical disappearance of the cell, this might take a considerable amount of time [71].

Apoptosis, also known as programmed cell death, can occur either as a natural process or as a response to cellular damage. Specifically, apoptosis is observed in certain types of cells

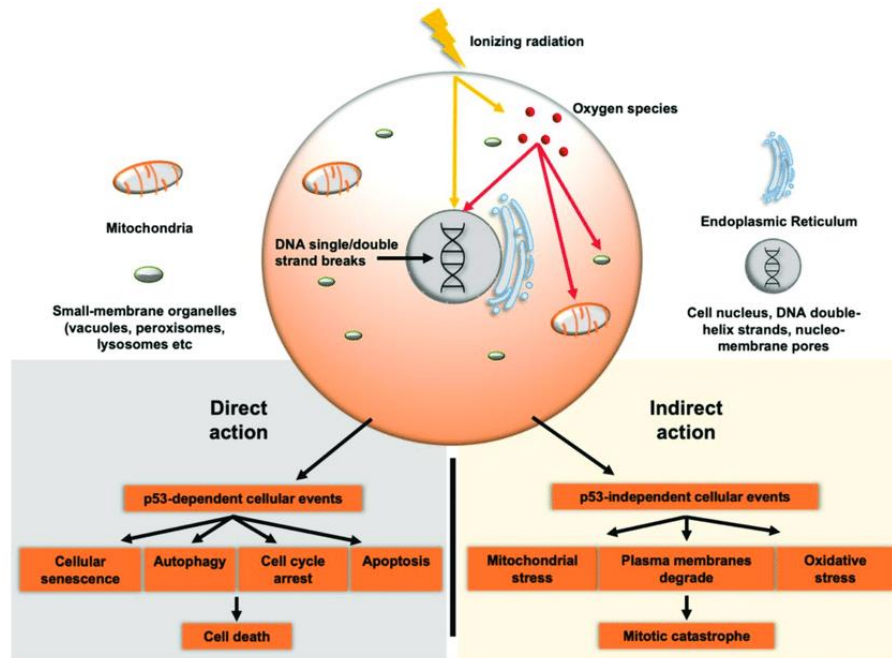


Figure-2.7: Cellular damage caused by ionizing radiation [72]

following exposure to low doses of radiation, such as lymphocytes, serous salivary gland cells, and specific cells located in the stem cell regions within the testis and intestinal crypts [73].

Necrosis represents a type of cellular demise marked by the cessation of cellular membrane functions. Typically, cellular necrosis is linked with exposure to elevated doses of radiation [74].

Reproductive cell death occurs due to mitotic catastrophe, a phenomenon where cells attempt to divide despite inadequate repair of DNA damage. This can happen within the initial few cell divisions following irradiation and becomes more frequent as the radiation dose increases [75].

Ionizing radiation can also induce senescence, a state where cells remain metabolically active but lose their capacity to undergo division.

2.11 Consequences of Exposure to Ionizing Radiation in Organs and Tissues

The unit "radiation equivalent man" (rem) measures the radiation dose received by an individual, indicating the amount absorbed by the body. Here is an overview of the effects of various radiation doses on the human body following acute, whole-body exposure [76-79]:

- 0-5 rem received in a short period of time is safe. We don't expect observable health effects.
- 5-10 rem received in a short period or over a long period of time is safe. We don't expect observable health effects. At this level, the effect is either non-existent or too small to observe.
- 10-50 rem received in a short period or over a long period of time - We don't expect observable health effects although above 10 rem the chances of getting cancer are slightly increased. A short term decrease of blood cells is also seen for doses of about 50 rem received in a matter of minutes.
- 500-100 rem received in a short period will likely cause some observable health effects and over a long period will increase the chances of getting cancer. Above 50 rem, some changes in blood cells can be seen, which quickly recover.
- 100-200 rem received in a short period will cause nausea and fatigue. When received over a long period, increases the chances of getting cancer.
- 200-300 rem received in a short period will cause nausea and vomiting within 24-48 hours. Medical attention should be sought.
- 300-500 rem received in a short period will cause nausea, vomiting and diarrhea within hours. Loss of hair and appetite occurs within a week. Medical attention should be sought for survival, half the people exposed to radiation at this level will die if they do not receive medical attention.
- 500-1200 rem received in a short period will likely lead to death within a few days.
- Above 10,000 rem received in a short period of time will lead to death within a few hours.

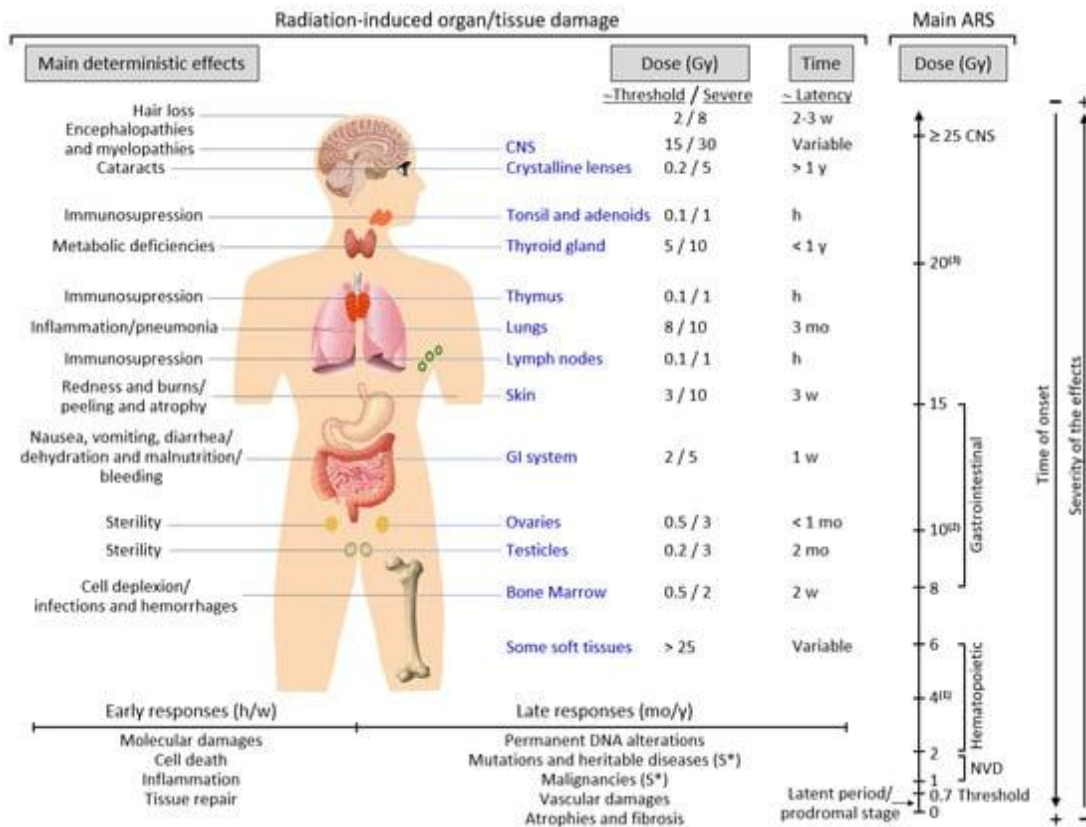


Figure-2.8: Consequences of exposure to ionizing radiation in organs and tissues are time and dose-dependent [80].

The previously mentioned health effects refer to radiation exposure affecting the entire body. If radiation is confined to a small body area, different effects might occur, but illness or death is not anticipated unless specifically stated [81-82]:

- 40 rem or more locally to the eyes can cause cataracts.
- 100-500 rem or more can cause hair loss for a section of the body that has hair.
- 200 rem or more locally to the skin can cause skin reddening similar to a sunburn.
- 1000 rem or more can cause a breakdown of the intestinal lining, leading to internal bleeding, which can lead to illness and death when the dose is to the abdomen.
- Above 1500 rem or more locally can cause skin reddening or blistering.

Table-2.3: Health Effects due to different ranges of radiation dose with time [83-84]

Exposure (rem)	Health Effect	Time to Onset (Without Treatment)
5-10	Changes in blood chemistry	Within Hours
50	Nausea	
55	Fatigue	
70	Vomiting	
75	Hair loss	2-3 Weeks
90	Diarrhea	
100	Hemorrhage	
400	Possible death	Within 2 months
1000	Destruction of the intestinal lining	1-2 weeks
	Internal bleeding	
	Death	
2000	Damage to central nervous system	Within minutes
	Loss of consciousness	
	Death	Hours to days

Rem is a CGS unit used for radiation dose. The SI unit of dose equivalent is Sievert (Sv), where $1 \text{ Sv} = 100 \text{ rem}$.

2.12 Radiation Dose Limits

The International Commission on Radiological Protection (ICRP) recommends dose limits to prevent individuals from being exposed to excessive ionizing radiation. These limits are essential for radiation protection, and exceeding them is illegal in many countries [85]. The limits are categorized into two groups: occupationally-exposed workers, and non-radiation workers and general public [86].

2.12.1 Radiation Dose Limits for Occupationally-Exposed Workers

The effective dose limit for radiation workers, as prescribed by ICRP, is 20 mSv a year, averaged over defined periods of 5 years with no single year $>50 \text{ mSv}$, for whole body exposure. The equivalent dose limit to the lens of the eye is 20 mSv a year, averaged over

defined periods of 5 years with no single year >50 mSv; to the skin and to the hands and feet are 500 mSv respectively [87].

2.12.2 Radiation Dose Limits for non-radiation workers and general public

According to ICRP, the effective dose limit for non-radiation workers and general public is 1 mSv a year (higher values are permitted if the average over 5 years is not above 1 mSv a year). The equivalent dose limit to the lens of the eye is 15 mSv a year and to the skin is 50 mSv a year respectively [87].

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CHAPTER THREE

THEORY OF COMPUTED TOMOGRAPHY IMAGING

3.1 Introduction

Computed Tomography (CT) has revolutionized medical imaging since its inception, offering unparalleled insights into the internal structures of the human body with exceptional clarity and detail [1]. As one of the most widely used imaging modalities in clinical practice, CT provides clinicians with invaluable diagnostic information for a myriad of medical conditions, ranging from trauma and oncology to cardiovascular and neurological disorders [2]. In recent years, there has been a growing emphasis on optimizing CT imaging protocols to ensure both diagnostic efficacy and patient safety, particularly concerning radiation dose management [3].

3.2 Computed Tomography (CT)

Computed Tomography (CT), also known as Computed Axial Tomography (CAT) scanning, is a medical imaging technique that utilizes X-ray technology to create detailed cross-sectional images of the body [4]. The word tomography has as its root “tomo”, meaning to cut, section, or layer from the Greek “tomos” (a cutting) and “graphia”, meaning to write or writing or to describe. Therefore, the term "tomography" literally translates to "slice writing" or "sectional description" [5]. This aptly reflects the core principle of CT scans, which is to generate detailed images by piecing together information from multiple thin slices of the body [6]. Invented in the 1970s, CT scans are essential in medical diagnostics. Unlike traditional X-rays, which create flat images, CT scans use computer processing to generate detailed 3D reconstructions, significantly improving medical diagnosis and treatment planning [7].

CT imaging works on the principle of X-ray attenuation [8]. A CT scanner is a doughnut-shaped machine with an X-ray source and detectors rotating within its center. The patient lie on a motorized table that moves you slowly through the scanner's opening [9]. During the scan, the X-ray source emits a narrow beam that passes through your body. The detectors on the opposite side measure the amount of radiation absorbed or transmitted by various tissues. Since different tissues have varying densities, they absorb X-rays to different degrees [10].



Figure-3.1: PET/CT machine at INMP, AERE, Savar, Dhaka

The varying attenuation of X-rays by different body tissues creates differences in the intensity detected by sensors. Dense tissues like bones attenuate more X-rays, appearing whiter (high attenuation), while less dense tissues like air attenuate fewer X-rays, appearing darker (low attenuation). This contrast allows for clear visualization of anatomical structures. This data is processed by a powerful computer [11]. The computer uses algorithms to process data from all angles, reconstructing detailed cross-sectional body slices. These slices can be viewed individually or stacked for a high-resolution 3D image, helping doctors see internal structures clearly and understand anatomy and potential issues [12].

Early CT scanners used parallel-beams to form the images [13]. These were then improved by using wider, fan-shaped beams with detectors that constantly recorded information while rotating around the body [14]. The newest CT technology uses helical or spiral scanning, which is faster and more comfortable for patients. It continuously acquires multiple scans to create detailed 3D images, enabling quicker and more efficient measurement and assessment of body structures' size (volume) by doctors [15].

CT scans can be done in two ways: a single scan with a limited number of slices or a multi-scan approach. Multi-scan CT scanners use dual-row detectors to capture information from multiple slices simultaneously, significantly increasing the number of slices per scan. Today, most CT scanners are either 64-slice or 128-slice, employing helical or cone-shaped beam systems [16-17].

3.2.1 Basic Principle of Computed Tomography

The process of acquiring CT images involves determining the characteristics of X-ray transmission through a subject from multiple angles. Generally, a detector arc, also called a detector row, with 800-900 detector elements (dels), is used to get a profile from each perspective. A variety of views can be obtained by revolving the detector row and X-ray tube around the patient. Faster data collection is possible by using tens or even hundreds of detector rows placed along the rotating axis. The CT image is rebuilt using the acquired transmission profiles, which are made up of a matrix of picture elements (pixels) [18].

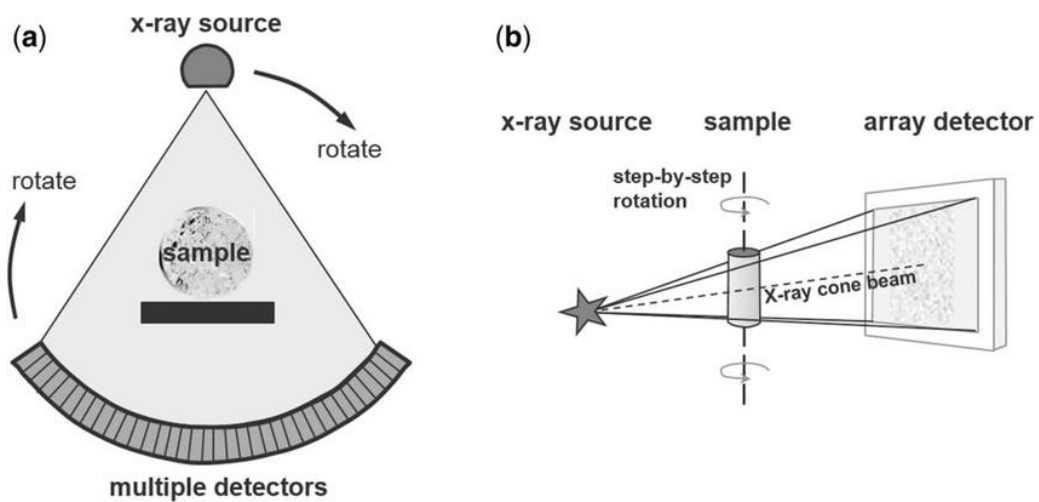


Figure-3.2: Imaging technique of a CT scanner [19]

Different shades of gray in a CT scan represent anatomical structures, formed through fundamental radiation principles. X-ray beams consist of photons, which structures can either guide, scatter, or absorb in varying proportions based on beam strength and structure characteristics. Attenuation describes the reduction in beam intensity. The interaction of photons depends on object thickness, density, and atomic composition, with density indicating the concentration of matter within a volume [20]. Elements characterized by high density, indicated by their high atomic number, possess abundant circulating electrons and hefty nuclei, thereby offering increased potential for photon interaction compared to less dense elements.

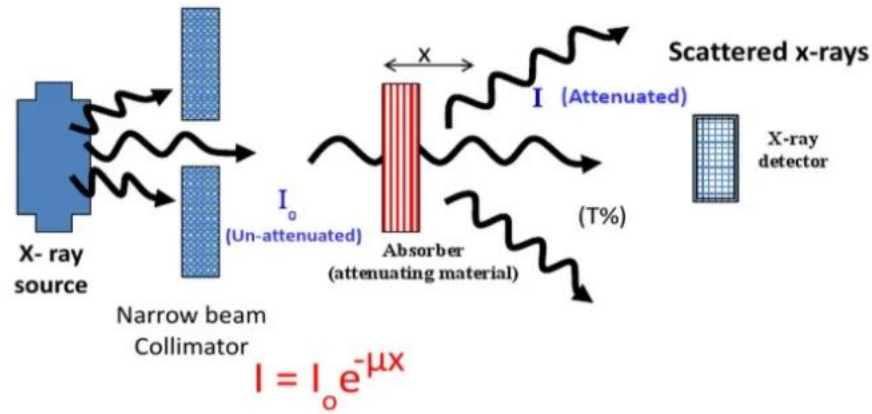


Figure-3.3: Attenuation of X-rays in Computed Tomography

The values that are assigned to the pixels in a ct image are connected with the attenuation of the relevant tissue, or, more specifically, to their linear attenuation coefficient μ (m^{-1}). The linear attenuation coefficient relies on the composition of the medium, the density of the material and the photon energy, as seen in Beer's law [21]:

$$I(x) = I_0 e^{-\mu x} \quad (1)$$

where x is the material's thickness, I_0 is the incident X-ray beam, and $I(x)$ is the attenuated X-ray beam's intensity plus the linear attenuation coefficient of μ . Beer's law describes only the attenuation of the original beam, not the intensity of dispersed radiation. It needs integration over all photon energies in the X-ray spectrum for polyenergetic beams. CT reconstruction algorithms often assume a single average photon energy, causing errors and beam hardening artifacts [22]. Various tissues are contacted by the X-ray beam as it passes through the patient, and these tissues have varying linear attenuation coefficients [23].

Table-3.1: Linear absorption coefficients of various body tissues [23]

Types of Tissue	Linear Absorption Coefficient, μ (cm^{-1})
Air	0.0004
Fat	0.185
Water	0.206
Blood	0.208
Gray Matter	0.212
White Matter	0.213
Bone	0.528

If the pathway through the patient runs from 0 to d, then the intensity of the attenuated X ray beam, transmitted a distance d, can be stated as:

$$I(d) = I_0 \exp \left(- \int_0^d \mu(x) dx \right) \quad (2)$$

Since a ct image is constituted of a matrix of pixels, the scanned patient can also be seen as being made up of a matrix of distinct linear attenuation coefficient volume elements (voxels). Figure 1 shows a simplified (4×4) matrix that represents the measurement of transmission along one line. For such a discretization, the equation for the attenuation can be expressed as:

$$I(d) = I_0 \exp \left(- \sum_{i=1}^4 \mu_i \Delta x \right) \quad (3)$$

As can be seen from the foregoing, the fundamental information required for CT is the measurable intensities of the attenuated and unattenuated X-ray beams, or $I(d)$ and I_0 , respectively. The matrix of linear attenuation coefficients, which forms the foundation of the CT image, can then be obtained using image reconstruction techniques. Utilizing kilovoltage photons enables accurate differentiation between tissues with varying atomic numbers. Even though adipose and muscle tissue have similar atomic numbers, CT is able to distinguish between them because it uses a large number of beams. The fact that the kilovoltage photons are solely impacted by the intervening tissue and that the spatial resolution - the precision with which each voxel's physical location can be ascertained, is exceptionally high are also noteworthy characteristics that facilitate precise planning [24].

3.2.2 CT numbers and Hounsfield Units

In CT scans, each tiny image element (called a pixel) has a CT number. This number reflects how much X-rays are absorbed by the tissue at that specific point. Different tissues absorb X-rays differently, and the strength of the X-ray beam (measured in kilovolts or kV) can also affect this absorption. To account for these variations, CT numbers are calculated relative to water. By comparing the X-ray absorption of a tissue to water (which is used as a reference point), CT numbers become consistent regardless of the X-ray beam strength. These CT numbers are then expressed in units called Hounsfield units (HU). CT number or Hounsfield unit can be calculated as follows [25-26]:

$$HU = [(\mu_{\text{tissue}} - \mu_{\text{water}}) / (\mu_{\text{water}} - \mu_{\text{air}})] \times 1000 \quad (4)$$

Since the linear absorption coefficient of air is very small to be negligible, hence,

$$HU = [(\mu_{\text{tissue}} - \mu_{\text{water}}) / \mu_{\text{water}}] \times 1000 \quad (5)$$

It can be seen that $HU_{\text{water}} = 0$ ($\mu_{\text{tissue}} = \mu_{\text{water}}$), $HU_{\text{air}} = -1000$ ($\mu_{\text{tissue}} = 0$) and $HU = 1$ is associated with 0.1% of the linear attenuation coefficient of water. Table-3.2 shows typical HU values for different body tissues.

Table-3.2: Typical HU Values and Ranges for Different Tissues and Materials [25-26]

Substance	Hounsfield Units (HU)
Air	-1000
Lung	-750 (-950 to -600)
Fat	-90 (-100 to -80)
Water	0
Brain, White matter	+25 (+20 to +30)
Brain, Grey matter	+35 (+30 to +40)
Muscle	+25 (+10 to +40)
Kidneys	+30 (+20 to +40)
Blood	+55 (+50 to +60)
Liver	+60 (+50 to +70)
Compact bone	+1000 (+300 to +2500)

Images in CT scans require a minimum data storage capacity (bit depth) of 12 for each picture element (pixel). This allows for a scale called the Hounsfield Unit (HU) to range from -1024 HU to 3071 HU, which covers most tissues important for medical diagnosis. For situations where very dense materials are present, a higher bit depth of 14 can be used. This expands the HU scale to a maximum of 15,359 HU, providing better accuracy when imaging these high-density materials [29].

3.2.3 Gray Scale

In CT scans, images use a gray scale representing Hounsfield Unit (HU) values for each pixel, indicating X-ray absorption by tissues. Denser tissues absorb more X-rays, showing higher HU values as brighter shades of gray. In contrast, cone-beam CT (CBCT) initially displays X-ray absorption as gray scale, but some systems convert it to HU-like values. Both CT and CBCT visualize tissue density using gray scale, with HU being the standard in CT due to its direct link to X-ray absorption [28-29].

3.3 Conventional X-ray Imaging vs CT Imaging

Both conventional X-ray and CT imaging use ionizing radiation for medical imaging. CT provides superior diagnostic capabilities, especially for complex conditions needing detailed anatomical information. However, factors like radiation exposure, cost, and availability

influence the choice between these modalities based on clinical scenarios and patient needs [30].

Conventional X-rays produce flat, 2D images, which can be limited in distinguishing overlapping structures, especially in dense areas like the chest or abdomen. In contrast, CT scans use multiple X-ray images from different angles, processed by computers to create detailed 3D reconstructions. This clarity helps doctors visualize internal structures more accurately, reducing misinterpretations from overlapping tissues [31].

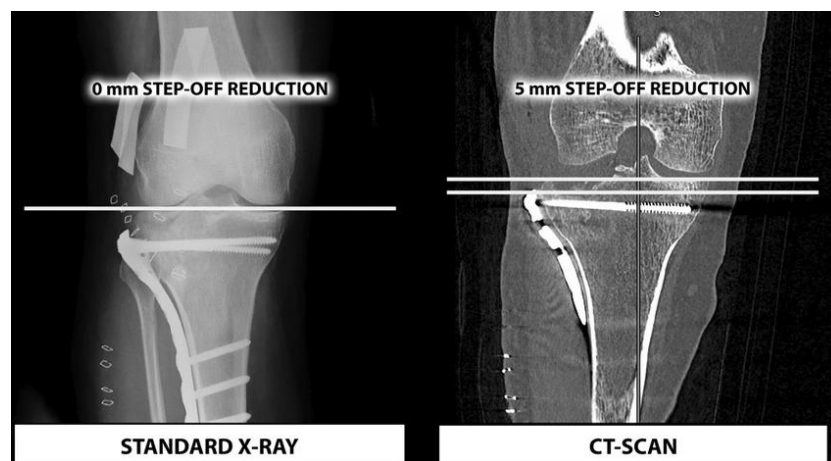


Figure-3.4: A radiological comparison between standard X-ray and CT scan [32]

X-rays provide basic images of bones and some organs, highlighting fractures, bone issues, and calcifications. However, they struggle with soft tissues, blood vessels, and subtle injuries. In contrast, CT scans offer higher resolution, distinguishing between soft tissues, blood vessels, and organs. They are ideal for detecting tumors, bleeding, and soft tissue abnormalities. X-rays use one radiation burst with low risk, while CT scans use multiple beams, resulting in higher radiation, despite modern safety measures being in place, making it crucial for young or frequently scanned patients to consider [33].

X-ray procedures are quick and simple, usually taking just a few minutes and are less expensive than CT scans. In contrast, CT scans involve more complex machinery and longer scan times (5 to 30 minutes), making them generally more costly. X-ray imaging is typical in emergency departments, orthopedics, dentistry, and general medicine for routine screenings. In contrast, CT imaging is widely used in radiology, oncology, neurology, cardiology, and surgery for detailed evaluations, disease staging, treatment planning, and monitoring therapeutic responses [34].

The decision between X-ray and CT scan hinges on the specific medical need. X-rays are favored initially for their speed, affordability, and lower radiation dose. However, for detailed 3D views or enhanced soft tissue visualization, CT scans are often recommended [35].

3.4 Modes of CT scans

CT (Computed Tomography) scanning can be accomplished with multiple different scan modes which have different characteristics. The different CT scan modes offer a toolbox for doctors to choose the most appropriate technique for the specific situation. This allows for optimized image quality, reduced radiation exposure when possible, and improved diagnostic capabilities based on the clinical need. Some of the modes are discussed below [36-37].

3.4.1 Axial CT scan

In axial CT scans, the X-ray source rotates 360 degrees around the patient, who remains stationary on a table. Variations include shorter arcs (around 180 degrees plus a fan angle) for quicker scans, or extended arcs (up to 720 degrees) for enhanced visibility of subtle details, utilizing higher mAs for stronger X-ray beams.

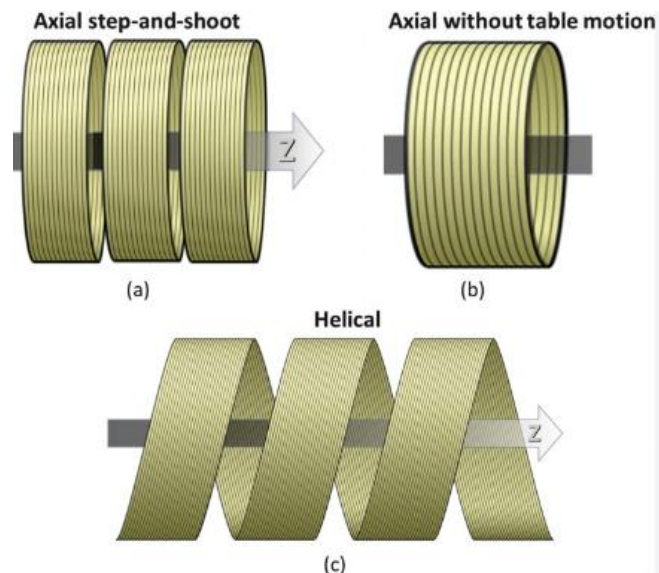


Figure-3.5: Axial and Helical CT [38]

To capture a larger body area, a complete CT scan involves acquiring multiple axial slices sequentially. This is done by taking individual images (or "shots") and incrementally moving the table (a "step") before each shot. Known as "step-and-shoot" acquisition, this method ensures all required slices are obtained. The table movement between shots matches the slice thickness, enabling seamless stacking of images to form a 3D view of the scanned area [39].

3.4.2 Helical CT scan

In 1989, the medical field saw a breakthrough with the introduction of helical CT scans, which combined a rotating X-ray source with a moving patient table. This innovation significantly improved CT scan performance. Helical CT scans reduced scan time and provided more consistent and detailed 3D information [40].

Helical scanning revolutionized CT by enabling doctors to capture a large area in a single breath hold, essential for high-quality CT angiography. In helical scans, "pitch" describes the table's movement relative to the X-ray beam width during one rotation. Unlike traditional CT with fixed slice thickness, helical scans allow varying pitch factors, offering more flexibility in slice thickness and table movement for faster, more detailed examinations. Traditional CT typically captures images with a slice thickness of 5-10 millimeters and a rotation time of 1-2 seconds [41].

3.4.3 MDCT scan

A decade after helical CT revolutionized the field, Multi-Detector Row CT (MDCT) scanners emerged as the next major leap in CT technology. These scanners boasted a significant upgrade: up to 64 rows of detectors working simultaneously. This innovation allowed for capturing a much larger number of slices at once. Additionally, the rotation time of the scanner drastically reduced to a mere 0.3-0.4 seconds. This remarkable speed meant scanning nearly an entire adult body in a single breath hold, all while achieving slice thicknesses well below 1 millimeter. Notably, MDCT scans are primarily performed in helical mode for optimal efficiency. However, there are exceptions, such as high-resolution lung scans and specific cardiac CT techniques used for assessing coronary calcium or visualizing coronary arteries [42-43].

3.5 Components of a CT Scanner

CT scanners have a complex structure with various components essential for image generation. Different manufacturers often modify these components' designs. To understand a CT system's functionality, it's important to know each component's role and their design variations. Generally, CT scanners include a scanning gantry, x-ray generator, computer system, operator's console, and physician's viewing console. Traditional hard-copy film has mostly been replaced by workstation viewing and electronic archiving, though many systems still use a laser printer for transferring CT images to film. CT operations can be divided into

three parts: data acquisition, image reconstruction, and image display. This chapter focuses on the initial phase. Data acquisition happens as x-rays pass through a patient, hit a detector, and are recorded. The main components involved in this phase are the gantry and the patient table. [44].

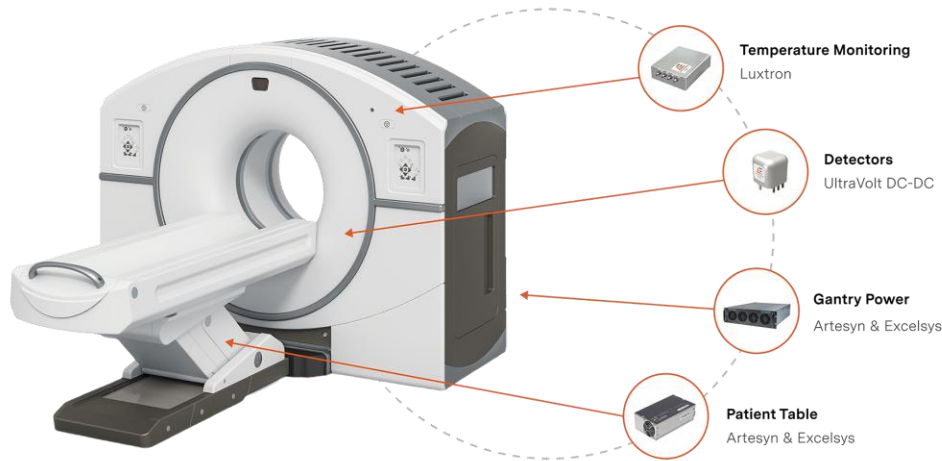


Figure-3.6: Components of a CT machine [45]

3.5.1 Gantry

The CT scanner has a donut-shaped opening called the gantry. This ring-like structure holds the machinery for generating and picking up X-rays. These components are mounted on a frame that spins inside the gantry. The size of the gantry and the opening (aperture) can vary, with typical openings ranging from 70 to 90 centimeters. For different patients and exams, the gantry can be tilted up or down to provide the best positioning. The extent of tilt varies across different systems, typically ranging from approximately $\pm 15^\circ$ to $\pm 30^\circ$. Within the gantry, there are mechanisms for controlling the tilt, alignment lights, and table movement, allowing technologists to ensure precise positioning of the patient. Additionally, a laser light system assists in accurately positioning the patient within the scanner before the scan begins. The gantry of a CT scanner plays a crucial role in facilitating accurate imaging by housing essential components, providing precise positioning, and enabling seamless communication and control throughout the scanning process [46].

3.5.2 CT Generators

Modern CT scanners use compact, high-frequency generators that fit within the machine's gantry. Older, bulkier three-phase generators required separate housings and cables, making

them less desirable and obsolete. These generators provide elevated voltage levels to the x-ray tube, with power capacity measured in kilowatts (kW). This capacity influences the range of exposure techniques, such as kilovolts (kV) and milliamperes (mA). CT generators typically operate at 120–140 kV, which enhances the x-ray beam's penetration and reduces patient radiation exposure. Higher kV settings also allow for lower mA settings, reducing the x-ray tube's heat load and extending its lifespan [47].

3.5.3 X-ray Source

X-ray tubes in CT scanners rotate around the patient within a gantry, emitting narrow beams of X-rays from multiple angles essential for detailed CT images. High voltage from the CT scanner generator controls the beam's strength, allowing for deeper penetration into the body with lower heat buildup, thus extending the tube's lifespan. Typically, tungsten, boasting an atomic number of 74, serves as the preferred material for the anode target due to its capacity to yield a more potent x-ray beam. This phenomenon arises from the fact that the intensity of x-ray emission is approximately correlated with the atomic number of the target substance [48]. CT tubes frequently incorporate multiple focal spot sizes, commonly 0.5 and 1.0 mm. Analogous to conventional x-ray tubes, smaller focal spots in CT tubes yield crisper images, thereby enhancing spatial resolution. However, their limitation lies in their inability to withstand as much heat, as they concentrate heat onto a smaller portion of the anode, resulting in reduced tolerance for thermal stress [49].

3.5.3.1 Production of X-rays in CT scanner

X-rays are a type of electromagnetic radiation characterized by wavelengths shorter than approximately 10 nanometers. A decrease in wavelength signifies an increase in energy, as outlined in Equation 4. The energy of individual X-ray photons, denoted as E , is directly proportional to their frequency, f , and can be expressed as follows:

$$E = hf = \frac{hc}{\lambda} \quad (4)$$

where h is Plank's constant ($h = 6.63 \times 10^{-34}$ Js), c is a speed of light ($c = 3 \times 10^8 \text{ms}^{-1}$), and λ is a wavelength of the X-ray.

Therefore, X-ray photons with longer wavelengths have lower energies than the photons with shorter wavelengths. The X-ray energy is usually expressed in eV ($1\text{eV} = 1.602 \times 10^{-19}$ J).

X-rays are generated by a process where a swift stream of electrons is decelerated by a metallic object (typically the anode), leading to the emission of X-ray photons. In a typical X-ray setup, as illustrated in Figure 3.7, there is a vacuum tube housing a heated cathode (usually made of tungsten) and an anode [50].

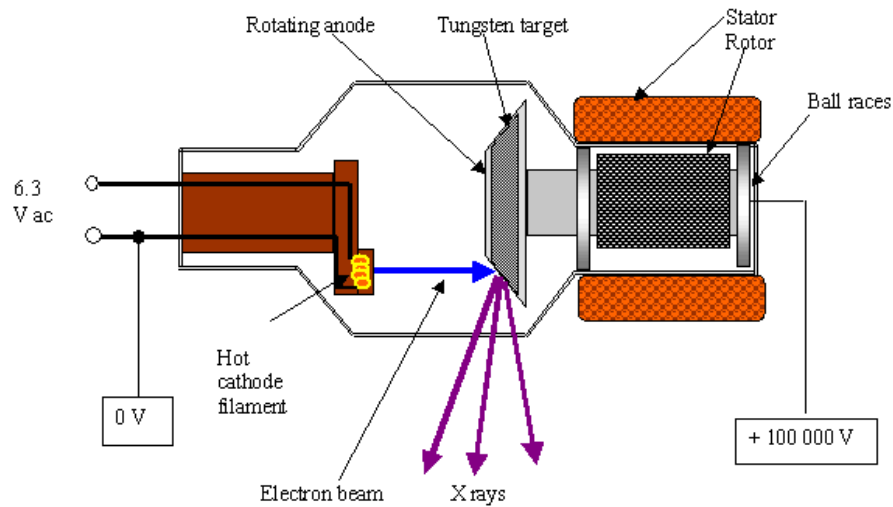


Figure-3.7: Production of X-rays in a CT scanner [51]

An electric potential is applied between these two components. Electrons are emitted from the cathode's surface and accelerated towards the anode. Upon reaching the anode, these electrons collide with its atoms, transferring their energy to the anode. This collision process happens within a very shallow layer of the anode. As electrons slow down, they eventually halt, allowing them to pass through the anode and into the electronic circuitry. These electrons interact either with the orbital electrons or the nuclei of the metal atoms. This interaction converts their kinetic energy into both thermal and electromagnetic energy, manifesting as X-rays [52].

Due to the conversion of over 98% of energy into heat, it is necessary for the anode to undergo water cooling. The emitted X-rays exhibit two distinct components: Firstly, Bremsstrahlung radiation, characterized by a continuous spectrum, and secondly, radiation resulting from the surplus energy of accelerated electrons colliding with atoms, leading to the expulsion of electrons from the inner electronic orbits of the target atom. This excess energy is released as X-ray photons with consistent peak wavelengths, commonly known as characteristic $K\alpha$ radiation or peak X-ray energy. Since these spectral peaks vary depending on the material, this type of radiation is termed Characteristic radiation [53].

3.5.4 Patient Table

During a CT scan, the patient lies on a table that moves through a large donut-shaped machine called a gantry. This movement, known as incrementation, is measured in millimeters per second as the table continuously moves during the scan. The table's sideways movement determines the scannable range, indicating how much of the patient's body can be scanned without repositioning. The CT scanner shows the table's position relative to the gantry. When the patient is positioned, a specific anatomical reference point is aligned with the scan point, and the technologist manually calibrates the table to zero, marking the reference level. Precise table referencing ensures consistency across examinations [54].

3.5.5 Control Console

The control console of a CT machine is a vital component that enables operators to control scanning parameters, manage patient positioning, oversee image reconstruction, ensure safety, and interact with the CT system effectively, ultimately contributing to the production of high-quality diagnostic images for patient diagnosis and treatment. The control console is designed for user-friendliness, allowing the technologist to precisely control factors like X-ray dose, scan region, and image quality. It also provides vital information about machine status and ensures patient safety throughout the examination [55].

3.5.6 CT Collimators

Collimators limit the x-ray beam to a specific area, reducing scatter radiation, which improves image clarity and lowers patient radiation exposure. By adjusting the x-ray beam width, collimators control slice thickness. Positioned near the x-ray source, the source collimator, or "pre-patient collimation," narrows the x-ray beam before it reaches the patient. This not only affects patient exposure but also influences dose distribution within the slice. The collimator's aperture size can be adjusted for different slice thicknesses. In MDCT systems, the detector configuration also affects slice thickness, typically ranging from 0.5 to 10 mm [56-57].

3.5.7 CT Filters

Filters in CT scanners serve several key functions. They remove low-energy photons from the x-ray beam, reducing beam hardening effects and ensuring more accurate image reconstruction. By filtering out these low-energy photons, filters lower the radiation dose to the patient and improve the efficiency of higher-energy photons. This results in clearer, more

detailed images with enhanced contrast resolution and reduced noise. Additionally, filters help minimize imaging artifacts caused by beam hardening and scatter radiation, leading to more accurate and reliable diagnostic information [58].

3.6 Detectors used in CT

As the x-ray beam moves through the patient, it gets weakened to a certain extent. To generate an x-ray image, we need to gather data on how much each part of the body weakened the beam. In traditional radiography, we relied on a film-screen setup to capture this weakening. In contrast, in CT scans, we employ detectors for this purpose. The term "detector" can refer to a single component or a specific type of detector utilized in a CT system. Meanwhile, the term "detector array" encompasses all detectors integrated into a CT system. Typically arranged in an arc or ring, the detector array consists of individual detector elements, each measuring the intensity of x-ray radiation passing through from the source to that particular element. Additionally, the array includes reference detectors to aid in data calibration and minimize artifacts [59].

The total effectiveness of a detector is influenced by various components. These include: the ability of the detector material to halt particles; the efficacy of scintillators (for solid-state variations); the effectiveness in collecting charges (for xenon variations); geometric effectiveness, which measures the ratio of the space taken up by the detector collimator plates to the detector's surface area; and the ability to reject scattered particles [60].

3.6.1 Xenon Gas Ionization Detector

CT scanners use special detectors filled with xenon gas under pressure. This gas absorbs a large portion, between 60% and 87%, of the X-ray photons that hit it. Xenon gas is chosen for its stability under pressure. In contrast to solid-state detectors, xenon gas-based detectors are notably cheaper to manufacture, relatively simpler to adjust, and exhibit strong stability [61].

A xenon detector setup comprises three tungsten plates. When a photon penetrates the channel, it triggers ionization of the xenon gas. These resulting ions experience acceleration and amplification due to the electric field generated between the plates. The accumulated charge then generates an electric current, which is subsequently treated as primary data. However, one drawback of using xenon gas is its necessity to be maintained under pressure within an aluminum enclosure. This enclosure partially filters the x-ray beam. The primary

factors that impede detector efficiency are the loss of x-ray photons through the casing window and the spatial occupation of the plates [62].

3.6.2 Solid-State Crystal Detector

Solid-state detectors, alternatively known as scintillation detectors, utilize a crystalline material that emits light upon being struck by an x-ray photon. Coupled with a photodiode, this crystal converts the light energy into electrical energy. These detectors are integrated into a circuit board for operation. Various materials, such as cadmium tungstate, bismuth germinate, cesium iodide, and ceramic rare earth compounds like gadolinium or yttrium, have been employed to fabricate solid-state crystal detectors. Due to their high atomic numbers and density compared to gases, solid-state detectors exhibit superior absorption coefficients, capturing nearly all incident photons without losses through a front window, as seen in xenon systems. This heightened absorption efficiency stands as the primary advantage of solid-state detectors. Although some solid-state detectors may exhibit a brief afterglow, modern CT detectors have largely mitigated this phenomenon. Nonetheless, solid-state detectors are more susceptible to temperature and moisture fluctuations compared to gas-based counterparts [63-64].

The arrangement, form, and dimensions of the sensors impact the quantity of scatter radiation reaching the image. A detector design that is deep and narrow will intercept less scatter compared to one that is short and wide. Spacing bars are used to separate the detectors, allowing them to be arranged in either an arc or a circle. The measurement of detector spacing extends from the center of one detector to the center of the adjacent detector, taking into account the spacing bar. Ideally, detectors should be positioned as closely together as possible to ensure that all X-rays are converted into data. The aperture size of the detector determines its opening. A smaller aperture is crucial for achieving high spatial resolution and rejecting scatter. Optimal utilization and a compact aperture are preferred attributes [65].

3.7 CT Image Reconstruction

3.7.1 Pixel

The smallest individual unit of a physical display or image is known as a pixel (also referred to as a pel or picture element). While pixel diameters are commonly measured in micrometers (microns), there is no standard size for pixels. Although pixels are not a unit of measurement, the resolution or sharpness of photographs is often expressed in terms of pixels. The term

"pel," short for "picture element," has been used historically as a synonym for "pixel," but it has not become popular outside of specialized fields and is not utilized in medical imaging [66-67].

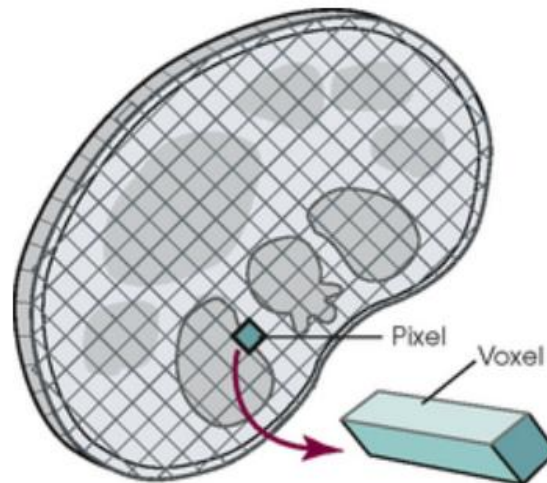


Figure-3.8: Matrix of a CT image: Pixel and Voxel [68]

3.7.2 Voxel

A voxel, or volume element, is the three-dimensional counterpart to a pixel. Both pixel size and slice thickness are connected to the voxel. In a CT scan, each voxel represents a small volume element due to the limited thickness of the section. The pixels forming a CT image are arranged in a square matrix, typically ranging from 256 x 256 to 1024 x 1024. Given that a CT slice has a fixed thickness, each pixel represents a tiny voxel. The voxel size is influenced by the matrix size, the field of view (FOV), and the section thickness. In most CT scans, the section thickness along the z-axis is 10 to 20 times greater than the voxel size in the x-y plane. To achieve more uniformly shaped voxels, the section thickness must be significantly reduced. Multi-slice CT is the only method to obtain nearly isotropic (cube-shaped) voxels for larger body sections [69-71].

3.7.3 Image Reconstruction

The objective of CT image reconstruction is to determine the level of attenuation of the narrow X-ray beam in each voxel of the reconstruction matrix [72]. To create an image, a CT scan calculates the attenuation coefficients of each voxel by using the sum of attenuation coefficients obtained from X-ray intensity measurements taken from various irradiation directions [73]. If a slice of heterogeneous tissue is divided into volume elements, or voxels,

each with its own attenuation coefficient, and exposed to radiation, the sum of the voxel attenuation coefficients for each X-ray beam direction can be determined from the experimentally measured beam intensities for a specific voxel width [74].

Tomograms are digital images produced by assembling all the photographs of the patient's individual body slices. These tomograms are used to diagnose the patient undergoing the scan. Tomography software is employed to categorize and magnify a region of interest (ROI), measure distances on the tomogram, calculate the radiation attenuation coefficients of specific areas, and display the attenuation coefficients of any slice. The images are documented and stored for future use. With the advancement of modern spiral CT or multi-slice scanners and the introduction of picture archiving and communication systems (PACS), printing CT images is becoming less common, as they are now more frequently viewed directly on CRT or flat-screen monitors and archived digitally [75-76].

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CHAPTER FOUR

CT SIMULATION IN RADIOTHERAPY TREATMENT PLANNING

4.1 Introduction

Computed Tomography (CT) simulation has revolutionized radiotherapy by providing precise anatomical details essential for planning and administering treatments [1-2]. This imaging method enables accurate pinpointing of tumors and critical structures in the body, facilitating personalized radiation therapy to optimize tumor control while minimizing harm to healthy tissues [3-4]. As a cornerstone in contemporary radiotherapy, CT simulation integrates advanced technologies like intensity-modulated radiotherapy (IMRT) and image-guided radiotherapy (IGRT), thereby improving treatment accuracy and patient outcomes [5].

4.2 CT Simulation

The swift advancement of computer technology, which allows for the rapid reconstruction of three-dimensional images, has revealed the significant value of the vast amount of data collected by CT scanners and its application in radiotherapy planning. CT simulation involves obtaining a CT scan for planning radiation treatment, representing the initial and essential phase in the workflow of radiation therapy. The images generated during CT simulation are utilized to delineate the anatomy of tumors and surrounding organs, achieved through manual methods or increasingly, through the adoption of commercial semi-automated or fully automated techniques [6-10].



Figure-4.1: CT simulation procedure [45]

For radiation treatment to be as effective as possible, doctors need to clearly identify the exact size and shape of the tumor and surrounding healthy organs in the CT scans. This is

achieved by considering how well the tumor and organs show up in the scans, understanding the body's anatomy from the scans, and considering how the cancer might have spread based on its biology. Fundamentally, this relies on the assumption that the CT scan accurately reflects the patient's anatomy [11-16].

Two primary technologies are competing in the realm of CT simulation. One is a diagnostic CT (DCT) scanner that is transformed into a pseudo-simulator using specialized software and laser marking, referred to as CT-Sim [17-18]. This system offers advanced CT images and high-speed acquisition capabilities, but its effectiveness is limited by a small gantry aperture and challenges in producing a simulator X-ray beam with divergence across all planes. The alternative approach involves converting a conventional treatment simulator into a CT scanner (Sim-CT), providing highly accurate simulations. However, the image quality of this Sim-CT is inferior to that of a cutting-edge dedicated CT (DCT) system, and it is restricted in the number of acquisitions it can perform [19-20]. Recent advancements in CT simulation have reignited the discussion on whether these techniques provide additional benefits to radiotherapy [21].

Multi-slice helical scanning with high-capacity CT tubes has significantly decreased the time needed to acquire a 100-slice CT dataset to just a few seconds. Wide bore scanners have eliminated most limitations related to patient size and technique. Enhanced computing power and speed enable real-time reconstruction of slice images at the scanner and real-time manipulation of images in virtual simulation software. Additionally, the DICOM protocol allows for rapid transfer of image data between systems [22]. Most CT simulation systems are set up as single virtual simulator workstations connected to a CT scanner. Issues occur when additional virtual simulation stations are added to manage a higher workload, leading to multiple copies of patient data and unsynchronized files. Future systems should integrate patient images, structures, and treatment files into a database that allows multi-user access while ensuring full data protection, such as locking files while in use to prevent simultaneous access [4].

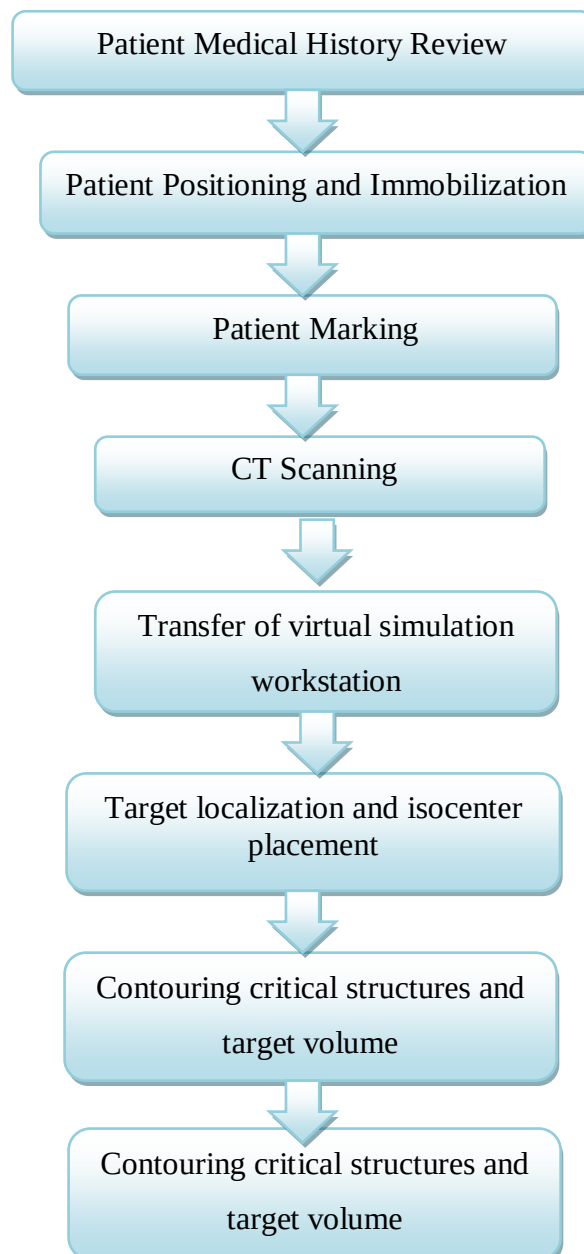
4.2.1 Steps of CT Simulation Process

The CT simulation process includes the following steps [23-28]:

- Patient Consultation and Medical History Review
- Patient Positioning and Immobilization
- Patient Marking

- CT Scanning
- Transfer of virtual simulation workstation
- Localization of targets and placement of isocenter
- Marking of patient and immobilization devices based on isocenter coordinates
- Contouring of critical structures and target volume
- Beam placement design, design of treatment portals
- Transfer of data to treatment planning system (TPS) for dose calculation
- Prepare documentation for treatment
- Perform necessary verifications and treatment plan checks

4.2.2 Flowchart of CT simulation Process



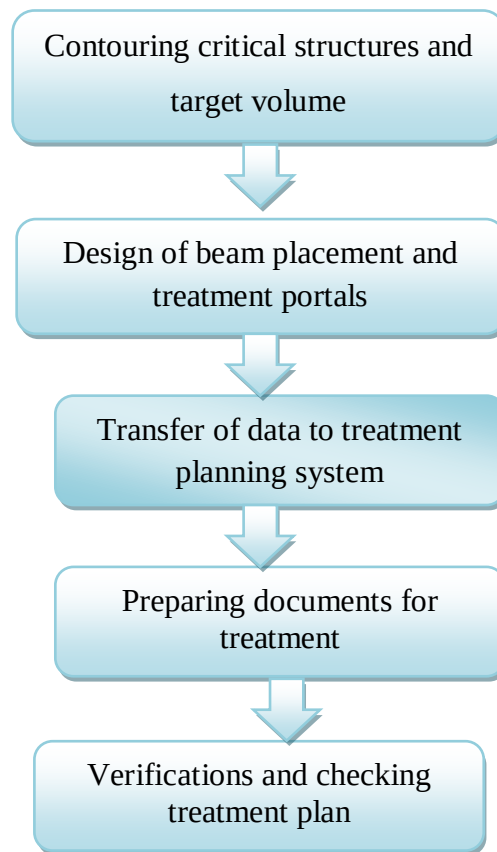


Figure-4.2: Flowchart of CT Simulation Process [23-29]

4.3 Specifications of a CT Simulator

The basic essentials for a CT simulator include a CT scanner equipped with a flat table, positioning lasers, immobilization devices and virtual simulation software [4,22,29]:

4.3.1 CT Scanner

A CT scanner with a flat couch is a crucial component in CT simulation for radiotherapy. This specialized equipment ensures patients can be positioned precisely and consistently during imaging, aligning accurately with the treatment plan coordinates. The flat couch offers a stable and comfortable surface for patients during CT scans, facilitating the capture of detailed cross-sectional images of their anatomy. These high-resolution images are essential for radiation oncologists and medical physicists to accurately outline tumor volumes and critical structures, crucial for optimizing treatment plans. Integrating a CT scanner with a flat couch in CT simulation allows healthcare providers to thoroughly examine and interpret the patient's internal anatomy, aiding in the development of personalized radiation therapy approaches that enhance tumor targeting while minimizing radiation exposure to healthy tissues [22,29].

4.3.2 Positioning Laser

A setup with three lasers is necessary to ensure precise patient positioning and alignment. The lateral lasers can be mounted on walls or frames, either fixed or adjustable in a vertical plane. The sagittal laser should have lateral movement capability to compensate for any limitations in lateral movement on the CT couch. These lasers are controlled by a computer to accurately establish the treatment plan's is center based on shifts from reference marks [22]. In modern CT scanners, positioning lasers are mounted within the scanner.

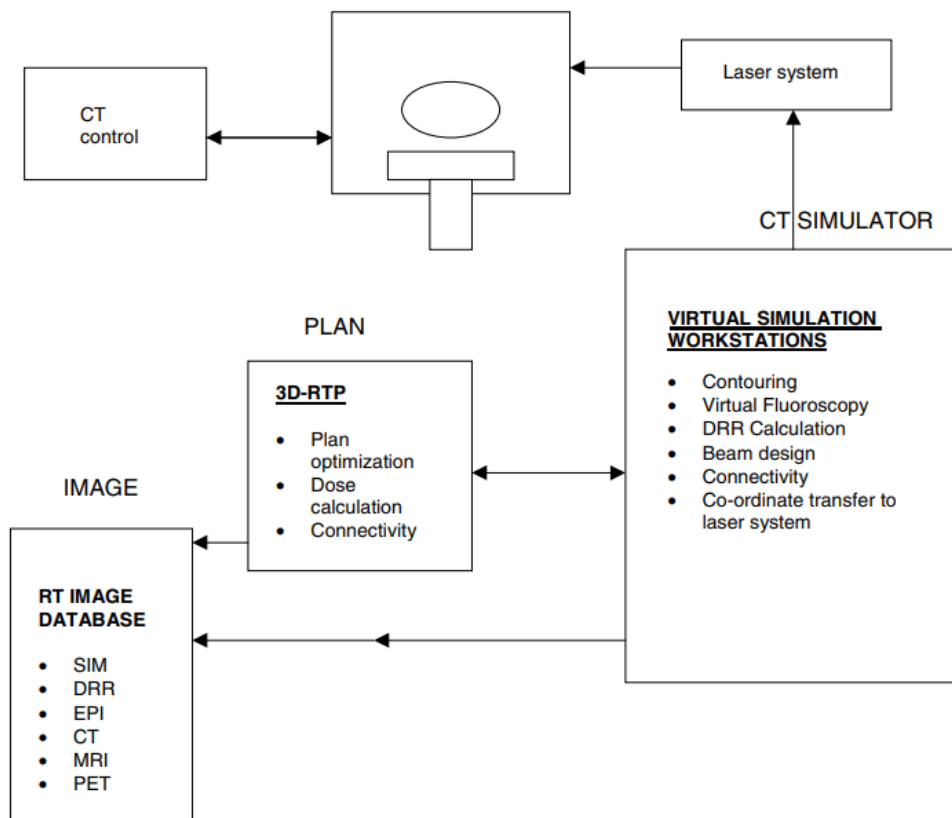


Figure-4.3: Schematic diagram of a CT Simulator [4].

4.3.3 Immobilization Devices

In CT simulation, immobilization devices are essential for ensuring treatment precision and patient comfort. These devices reduce patient movement during scanning, resulting in clearer CT images. This accuracy is critical for radiation oncologists to accurately delineate the tumor and surrounding healthy tissues on CT scans. Common immobilization devices include headrests, masks (thermoplastic shells for head and neck), vacuum bags for body molding, and bite blocks for stabilizing the jaw [29-30].

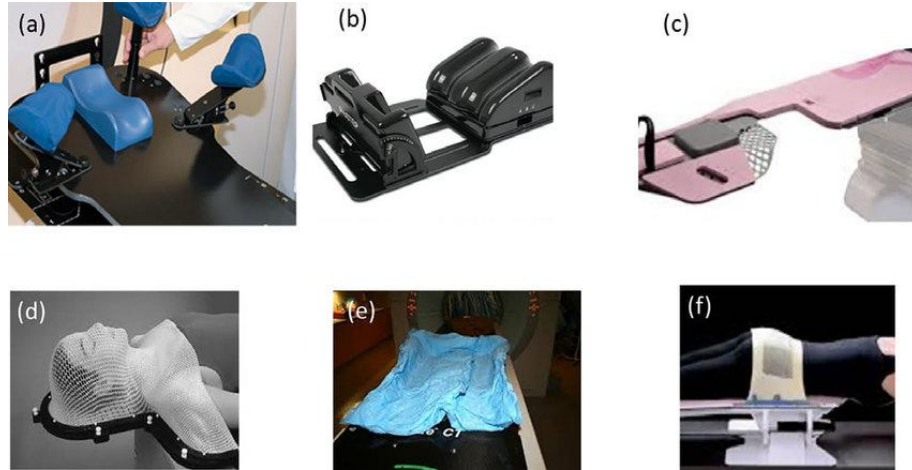


Figure-4.4: Various types of immobilization devices. Upper panels are reusable and customizable to patient (a) breast and thorax board, (b) Leg support used in prostate, (c) prone breast board. Lower panels are patient specific fixation devices (d) aquaplast mold for head and neck, (e) Vaclock fixation used for trunk and (f) solid aquaplast device for pelvic immobilization. [26]



Figure-4.5: Thermoplastic mask used for CT simulation at INMP, Savar Dhaka

The selection of the device depends on the treatment area targeted; for instance, headrests and masks stabilize head and neck regions, while vacuum bags conform to body contours for lung or abdominal tumors. Immobilization devices also enhance patient comfort during scanning by providing support and minimizing the need for repositioning, which is particularly beneficial for patients experiencing discomfort or difficulty remaining still due to pain, anxiety, or weakness [30].

4.3.4 Virtual Simulation Software

Virtual simulation software serves as a robust digital tool in CT simulation for radiotherapy. It assumes a central role following the acquisition of CT scans, importing high-resolution 3D images to transform static data into an interactive virtual environment. Here, oncologists and medical physicists utilize the software to meticulously delineate the target area (tumor) and critical organs at risk (OARs) within the patient's anatomy with precision derived from CT images. Beyond visualization, the software empowers clinicians to virtually model the placement of radiation beams and explore various treatment plans. This capability allows for optimizing beam angles, dose distribution, and overall treatment strategies. Moreover, virtual simulation software can digitally replicate immobilization devices used during treatment, further enhancing the accuracy of treatment plan development. By facilitating virtual rehearsals of the treatment process, this software plays a crucial role in refining treatment plans and ultimately improving the precision and effectiveness of radiation therapy [4,22,29,31].

4.4 CT Simulation Vs Conventional Simulation

The enhanced differentiation of soft tissues, coupled with the detailed axial anatomical data from CT scans, enables highly accurate localization of target volumes and critical structures. During the CT simulation phase, precise identification and outlining of these structures are directly performed on the CT dataset. This capability, combined with the creation of Digitally Reconstructed Radiographs (DRRs) and Beam's Eye Views (BEVs), where organs and targets are projected onto simulated X-ray images, enables users to delineate treatment fields based on the location of the target volume and critical structures [32-34].



(a)



(b)

Figure-4.6: (a) CT Simulator at INMP (b) Conventional Simulator

In contrast, conventional simulation relies on identifying the tumor's location relative to visible landmarks on radiographs from a diagnostic-quality simulator. These radiographs offer limited contrast for soft tissues, restricting the ability to define treatment boundaries based on either bony structure visible on the images or anatomical features enhanced by contrast agents like barium [36]. Another significant benefit of the CT simulation process compared to conventional methods is that once the scanning is complete, the patient does not need to remain on the simulator couch. They only need to stay for the minimum time required to acquire the CT data set. This allows radiotherapy staff the flexibility to take their time in planning the treatment and experiment with different beam configurations without the patient needing to wait on the couch [4,34].

A CT simulator enables users to create digitally reconstructed radiographs (DRRs) and beam's eye views (BEVs) for beam configurations that were previously impractical to simulate using traditional methods. For instance, planning vertex fields was not feasible with conventional simulators because the film plane is located within the patient's anatomy. There is ongoing discussion regarding the necessity of a conventional simulator in radiotherapy clinics where a CT simulator is available. Apart from logistical and economic considerations associated with CT scanning every patient, there are specific scenarios where CT simulation may not be essential, such as in cases involving cord compression, bone metastases, and brain metastases. Furthermore, performing a fluoroscopic simulation of patients after CT simulation is beneficial for verifying the position of the treatment isocenter, ensuring accurate field limits, and marking the patient for treatment [4,22,34].

4.5 Advantages and Disadvantages of CT Simulation

CT simulation offers several advantages, particularly in the fields of radiotherapy and medical imaging. Some key advantages include:

- a) The availability of a three-dimensional dataset enhances the visualization of tumors and lymph node involvement, leading to a decrease in side effects [22].
- b) CT simulation generates detailed cross-sectional images of the body, allowing for precise identification of tumors and vital structures. This precision improves the effectiveness of radiotherapy by ensuring the radiation dose is accurately targeted to the treatment area, minimizing exposure to healthy tissues [35].

- c) By enabling more accurate radiation targeting, CT simulation can enhance patient outcomes, resulting in higher tumor control rates and reduced side effects. This can lead to an improved quality of life for patients receiving treatment [36].
- d) The detailed, high-resolution images produced by CT simulation improve the visualization of internal structures such as bones, soft tissues, and blood vessels. This enhanced clarity supports more accurate diagnosis and treatment planning [37].
- e) CT simulation optimizes the treatment planning process, enhancing efficiency. It minimizes the necessity for multiple imaging sessions, enabling quicker planning and treatment delivery, which improves overall workflow in clinical environments [38].
- f) CT simulation can be combined with cutting-edge technologies like intensity-modulated radiotherapy (IMRT) and image-guided radiotherapy (IGRT), significantly improving the accuracy and efficiency of treatments [39].
- g) Radiation oncologist not required during the scanning process [22].

Despite numerous advantages of CT simulation, it comes with certain disadvantages. Here are some of the disadvantages:

- a) CT simulation uses ionizing radiation, which may elevate the likelihood of radiation-induced cancers over time, particularly with frequent scans or in younger patients [40].
- b) The costs of acquiring and maintaining CT simulation equipment can be significant, presenting financial hurdles for healthcare facilities, especially those operating with limited resources [41].
- c) Certain patients may feel claustrophobic or uncomfortable during CT scans, which could potentially compromise image quality if they involuntarily move during the procedure [42].
- d) Contrast agents utilized during CT scans have the potential to induce allergic reactions in certain patients, varying in severity from mild to severe, and in some cases, they can result in kidney impairment [43].
- e) CT scans may be excessively used, potentially resulting in unnecessary radiation exposure and increased healthcare costs without yielding proportional clinical benefits [44].

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CHAPTER FIVE

MATERIALS AND METHODS

5.1 Introduction

The ICRP introduced the effective dose to establish radiation protection limits. It is a derived, mathematical construct rather than a measurable physical quantity. The calculation formula for effective dose to a reference model includes factors for various radiation types, organ and tissue sensitivities, population groups, and multiple biological endpoints [1].

In this chapter we will discuss about phantom used to determine CTDI volume ($CTDI_{vol}$), procedures and steps for calculation of CTDI volume and the dose-length-product (DLP) method to calculate effective dose.

5.2 Materials

5.2.1 Phantom

Dose distribution in radiation treatment is rarely measured directly in patients. Instead, measurements are typically conducted in phantoms, materials mimicking human tissue, to replicate conditions for the radiation beam. These measurements form the basis for dose calculations used to predict distribution in patients [2]. Water phantoms, chosen for their similarity to human tissue in radiation absorption and scattering, are commonly used despite practical challenges like waterproofing detectors [3].

Due to the challenges of placing radiation detectors in water, solid dry phantoms have been created as alternatives. The ideal tissue or water equivalent material should possess equivalent effective atomic number, number of electrons per gram, and mass density [4]. However, as the Compton effect prevails as the primary interaction mode for megavoltage photon beams in clinical applications, achieving water equivalence for such beams primarily requires matching electron density (number of electrons per cubic centimeter) to that of water.



Figure-5.1: Cylindrical PMMA Phantom used at INMP, AERE, Savar, Dhaka

The electron density (ρ_e) of a material may be calculated from its mass density (ρ_m) and its atomic composition according to the formula [5]:

$$\rho_e = \rho_m \cdot N_A \cdot (Z/A) \quad (5.1)$$

where,

$$Z/A = \sum a_i \cdot (Z_i/A_i) \quad (5.2)$$

N_A is the Avogadro's number and a_i is the fraction by weight of the i th element of atomic number Z_i , and atomic weight A_i . Electron densities of various human tissues and body fluids have been calculated according to Equation (5.1).

Lucite and polystyrene are commonly used in dosimetry due to their stable atomic composition and electron density, making them suitable for photon and electron dosimetry, particularly in high-energy applications [6]. Anthropomorphic phantoms like the Alderson Rando Phantom, designed to mimic human tissues such as muscle, bone, lung, and air spaces, are also widely used for clinical dosimetry assessments [7].

White et al. [8] has devised detailed procedures for creating tissue substitutes. This technique involves incorporating particulate fillers into epoxy resins to produce a blend with radiation characteristics closely resembling those of specific tissues. The key radiation characteristics

considered include the mass attenuation coefficient, mass energy absorption coefficient, electron mass stopping, and angular scattering power ratios.

Constantinou et al. [9] developed a solid water substitute using epoxy resin, building upon prior methods. This solid material serves as a dosimetric calibration phantom suitable for photon and electron beams within the energy range utilized in radiation therapy.

5.3 Methodology

5.3.1 Study Population

A total of 53 oncology patients (6 male and 47 female) were included in the present study. The age of male subjects was ranging from 54-75 year and for female subjects ranged from 35-65 year. The male patients had height range from 1.52 m to 1.67 m had weight range from 41 Kg to 93 Kg. Female patients had height range from 1.37 m to 1.65 m, had weight range from 35 Kg to 80 Kg. The patient's demographics are shown in Table 5.1.

Table-5.1: Anthropometric and demographic characteristics of the patients

Categories	Age Range (years)	Height Range (m)	Weight Range (Kg)
Male, n=6	54-75	1.52-1.67	41-93
Female, n=47	35-65	1.37-1.65	35-80
Total, n=53	35-75	1.37-1.67	35-93

5.3.2 Collection of Exam Data

The data for anthropometric and demographic characteristics of the patients such as age, gender, height and weight have been noted. The amount of radionuclide injected was collected from the medical records of each patient. The CT parameters: tube voltage (volt), tube current (mA), scan length (cm), were obtained from the display screen of the CT images of individual patients. Volume CTDI (mGy) were calculated and multiplied with the scan length to obtain Dose-Length Product.

5.4 Calculation of Patient Radiation Dose

5.4.1 Computed Tomography Dose Index (CTDI)

The CT dose index (CTDI) is a ubiquitous dose quantity used in CT. The CTDI represents an approximation of the average absorbed dose in a standard geometry (PMMA cylinder with a diameter of 32 cm for body scans or 16 cm for head scans and a length of approximately 14 cm) in a central rotation, including scatter contributions from adjacent rotations [10-11].

The CTDI is determined by integrating the dose profile from a single axial scan along the z-direction, $D(z)$, and normalizing that integral with the nominal total collimation width (nT) of the scan. Measured in milligray (mGy), CTDI offers a basis for estimating the radiation absorbed by a patient undergoing a CT scan [10,12-13].

5.4.2 CTDI₁₀₀

In 1999, the International Electrotechnical Commission (IEC) published the European Standard “Particular requirements for the safety of X-ray equipment for computed tomography” where the integration length of the CTDI was fixed to ± 50 mm (CTDI₁₀₀) and it also stated that $D(z)$ should be measured in air kerma (K_{air}) [10,12,14]. The CTDI₁₀₀ was defined for an integration length of 100 mm as,

$$\text{CTDI}_{100} = \frac{1}{nT} \int_{-50\text{mm}}^{+50\text{mm}} D(z) dz \quad (5.3)$$

Where, n is the number of slices in a single axial scan, T is the width of one slice (mm) and $D(z)$ is the air kerma at position z from a single axial rotation at $z = 0$. The CTDI₁₀₀ measurement must be performed with a stationary patient table and a 100-mm-long pencil ionization chamber [10,12-13,15].

5.4.3 CTDI_w

As the CTDI varies across the field of view (FOV), an average quantity must be estimated by weighting the different contributions of the periphery and center values of the FOV [10-16]:

$$\text{CTDI}_w = \frac{1}{3} \text{CTDI}_{100}(\text{Centre}) + \frac{2}{3} \text{CTDI}_{100}(\text{Periphery}) \quad (5.4)$$

Where $CTDI_{100}(\text{Centre})$ is measured at the central hole in the phantom and $CTDI_{100}(\text{Periphery})$ is calculated as an average of four measurements made at the peripheral positions [10-16].

5.4.4 $CTDI_{vol}$

The Volume Computed Tomography Dose Index ($CTDI_{vol}$) represents the scanner output. It is defined as $CTDI_w$ divided by the beam pitch factor (Equation 3.3). It is the most commonly used index for modern Multiple Detector Computed Tomography (MDCT) equipment [10-16].

The $CTDI_{vol}$ can be calculated as:

$$CTDI_{vol} = \frac{CTDI_w}{Pitch} \quad (5.5)$$

$$\text{Or, } CTDI_{vol} = \frac{n.T}{I} CTDI_w \quad (5.6)$$

Where, pitch is the Table movement per rotation/slice thickness, n is the number of simultaneous axial scans per x-ray source rotation, T is the thickness of one axial scan (mm) and I is the table increment per axial scan (mm) [11-14].

Since for an axial CT scan, there is no table movement, the $CTDI_{vol}$ is the same value as $CTDI_w$.

5.5 Estimation of External Dose

For the measurement of $CTDI_{vol}$, a cylindrical polymethylmethacrylate (PMMA) phantom representing a head with a diameter of 32 cm was used with a 10 cm active pencil-ionization chamber. The CTDI acquired by using this method is the $CTDI_{100}$.



Figure-5.2: CTDI measurement using Phantom

The phantom was placed in the CT scanner in a central position. The central position was aligned by using two perpendicular laser beams. The phantom has several holes, one in the center and others are evenly distributed throughout the periphery of the phantom, usually separated by 1 cm. To make the measurements, the pencil-ionization chamber was placed in the holes, and exposed to X-ray from the CT scanner. The ionization chamber placed at the center and periphery of the phantom, measured $CTDI_{100}$ separately by converting the X-ray exposure to an equivalent radiation dose.



(a)



(b)

Figure-5.3: (a) MagicMax Digitizer; (b) Pencil Ionization Chamber

The doses delivered by the CT scanner to the phantom is converted to mGy with the help of a digitizer. The digitizer employs MagicMax software to display the dose into the computer

screen measured by pencil ionization chamber. These doses are then expressed in terms of CTDI₁₀₀.

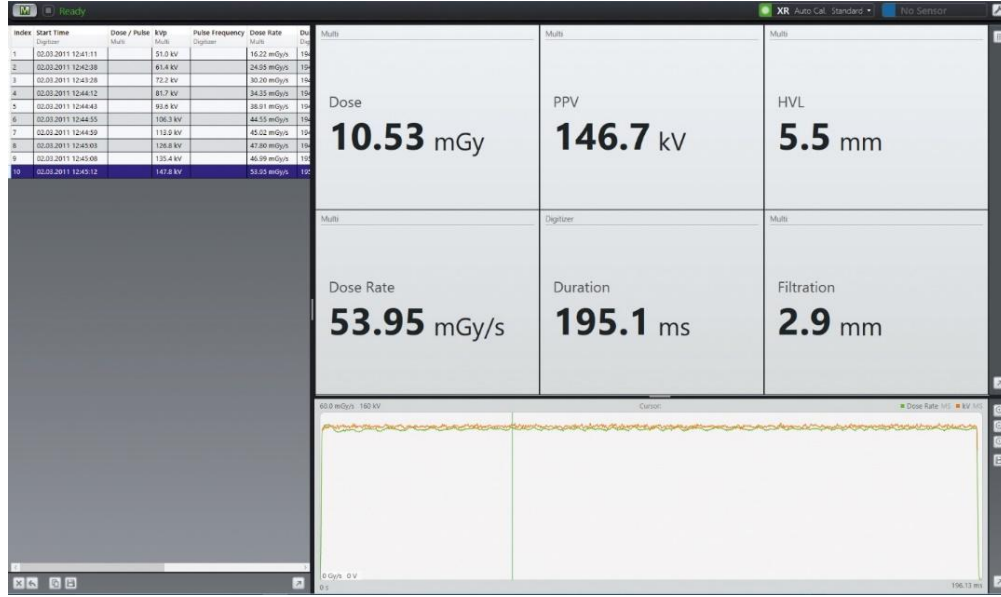


Figure-5.4: Dose Output at Computer Display

The rotation center of the X-ray beam was at the center of the phantom. The value of CTDI₁₀₀ changes when the values of tube voltage, tube current time and slice width are changed.

CTDI_w is termed as the weighted CTDI, which gives an estimation of the average CTDI over the whole field. CTDI_w, generally expressed in terms of radiation dose to air, is calculated using CTDI₁₀₀, as determined by the International Electrotechnical Commission (IEC). As the pencil-ionization chambers convert exposure to air KERMA to an equivalent dose in air, a conversion factor (f-factor) which is 0.87 rad/R is used to calculate the dose. The CTDI_w is calculated using the equation,

$$CTDI_w = \frac{1}{3} CTDI_{100}(\text{Centre}) + \frac{2}{3} CTDI_{100}(\text{Periphery}) \quad (5.4)$$

And with this CTDI_w, CTDI_{vol} is calculated by using the equation,

$$CTDI_{vol} = \frac{CTDI_w}{Pitch} \quad (5.5)$$

Pitch of a CT system is defined as the ratio of the distance of the table from the top, which has travelled 360° gantry rotation, to the beam collimator or width [68] It is simply is the table movement per rotation divided by the total width of the collimated beam.

5.5.1 Steps of Calculating CTDI Volume

- First a cylindrical PPMA phantom used that represents human tissue. There are typically two standard phantom sizes: one for head scans (16 cm diameter) and one for body scans (32 cm diameter). Here, we have used a 32 cm body phantom.
- Then dosimeter is placed (typically 10 cm pencil ionization chambers) in specific locations within the phantom: Once at the center and four times at the periphery (top, bottom, left, and right sides, each 1 cm from the surface).
- A CT scan of the phantom is performed using the specific protocol.
- After the scan, the dose measurements are recorded from the dosimeters placed in the phantom. These readings are generally given in units of milligray (mGy).
- Measured doses are then represented in terms of $CTDI_{100}$
- Then $CTDI_w$ is calculated using central and peripheral $CTDI_{100}$ values.
- Finally, $CTDI_{vol}$ is calculated by dividing the $CTDI_w$ by the pitch of the CT scan. For an axial CT, the value of the pitch is 1. Hence, $CTDI_{vol}$ is actually the $CTDI_w$ in the case of an axial CT scan.

5.5.2 Flowchart of CTDI volume Calculation

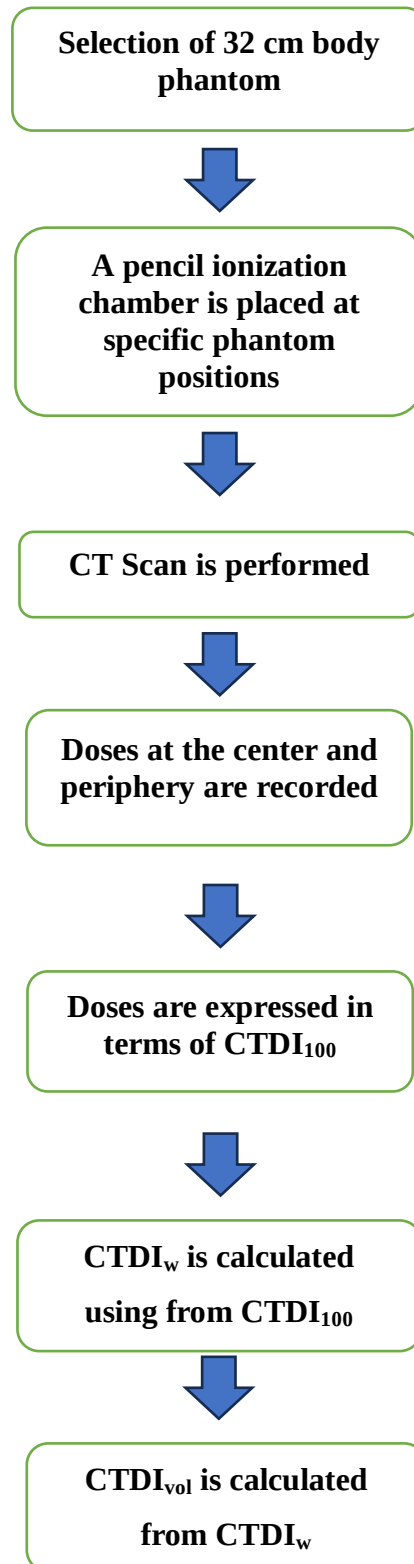


Figure-5.5: Flowchart of $CTDI_{vol}$ Calculation

5.6 Calculation of Effective Dose for CT

The effective dose for CT can be calculated with two different methods:

- a) The Size-Specific Dose Estimation (SSDE) method
- b) The Dose-Length Product (DLP) method
- c) Monte Carlo Simulation method
- d) MIRD method
- e) ICRP Tissue Weighting Factor Method

Here, we have used the Dose-Length Product (DLP) method.

5.6.1 Dose-Length Product (DLP) method

The DLP is a dosimetric quantity for CT that has a unit of mGy.cm that considers the amount of radiation the scanner has produced and absorbed by the patient's body. It indicates the total dose integrated over the entire scan and is dependent on various CT parameters. It's the product of two important parameters which are the Computed Tomography Dose Index (volume), measured in mGy, and the length of the patient, measured in cm, through which the scan was conducted [16-18]. For this certain calculation, the scanned length was considered from the crown of the head to the mid- thigh of the patients.

Therefore,

$$\text{DLP (mGy.cm)} = \text{CTDI}_{\text{vol}} \text{ (mGy)} \times \text{Scan Length (cm)} \quad (5.7)$$

Using this DLP, the effective dose for CT was calculated by the following equation,

$$E = \text{DLP} \times k \quad (5.8)$$

Where k is the conversion coefficient with a unit of $\text{mSv.mGy}^{-1}.\text{cm}^{-1}$. This coefficient is dependent on the anatomical region scanned and based on the empirical weighting factor according to the ICRP Publication 102 [19]. The values of k that are used for computing the effective doses of different anatomical regions were: $0.0021 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for head, $0.0031 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the head and neck and $0.0059 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the neck region, $0.014 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the chest, $0.015 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for abdomen, $0.015 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the pelvis and $0.015 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the trunk (chest, abdomen, pelvis) region. The diameter of the reference phantoms is 16 cm for the head and neck region and 32 cm for the chest, abdomen, pelvis, and whole trunk region [20-21]. In this study, the value of $k = 0.015 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ was used for the pelvis region.

5.6.2 Flowchart of Effective Dose Calculation Using DLP method

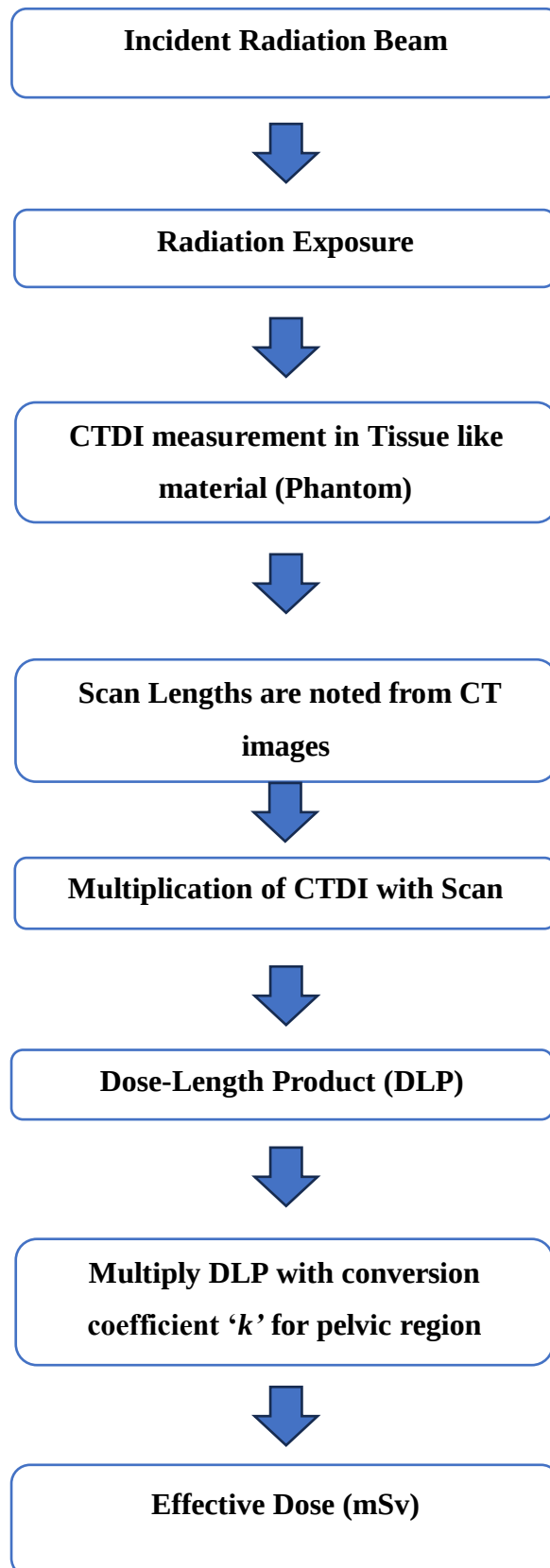


Figure-5.6: Flowchart of Effective Dose Calculation

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CHAPTER SIX

RESULTS AND DISCUSSIONS

6.1 Introduction

The effective radiation doses from pelvic CT examinations were estimated for 53 cancer patients of different ages, gender, and other physiological characteristics. The examinations were performed using a Philips Ingenuity Time-of-Flight (ToF) 64-slice PET/CT system (Philips Healthcare, Cleveland, OH, USA) installed at the Institute of Nuclear Medicine Physics, AERE, Savar, Dhaka. As the requirement the CT effective doses for cervix and prostate were estimated. In this chapter are presented, discussed and compared the present results with the reported series. Also are reported the general characteristics of the demographic data of the study subjects.

6.2 Demographics of the Study Subject

The present study included a total of 53 patients (6 male and 47 female) having cancer in prostate and cervix. A percentage distribution among the study subjects is shown in Figure-6.1.

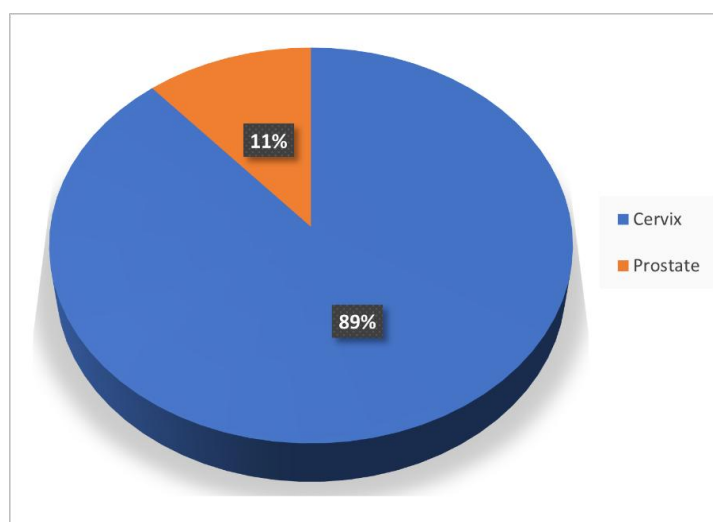


Figure-6.1: Percentage distribution among the study subjects

The age of the patients ranged from 35 to 75 years (50.698 ± 8.538 year), with a mean age of 61.5 ± 7.136 year for male patients and 49.319 ± 7.672 years for female patients.

Table-6.1: Demographic characteristics of the patients

Categories	Mean Age (year)	Mean Height (m)	Mean Weight (Kg)	Mean BMI (Kg/m ²)
Male, n=6	61.5±7.136	1.598±0.048	64±17.039	24.33±6.37
Female, n=47	49.319±7.672	1.498±0.058	56.298±9.895	25.045±3.96
Total, n=53	50.698±8.538	1.51±0.065	57.169±11.209	25.015±4.183

The mean height for male and female groups are 1.598±0.048 m and 1.498±0.058 m respectively, while the mean weights are 64±17.039 kg and 56.298±9.895 kg respectively. The mean BMI for male patients is 24.783±5.626 Kg/m² and for female patients is 26.384±4.88 Kg/m². All of these characteristics of the study subjects are shown in Table 6.1.

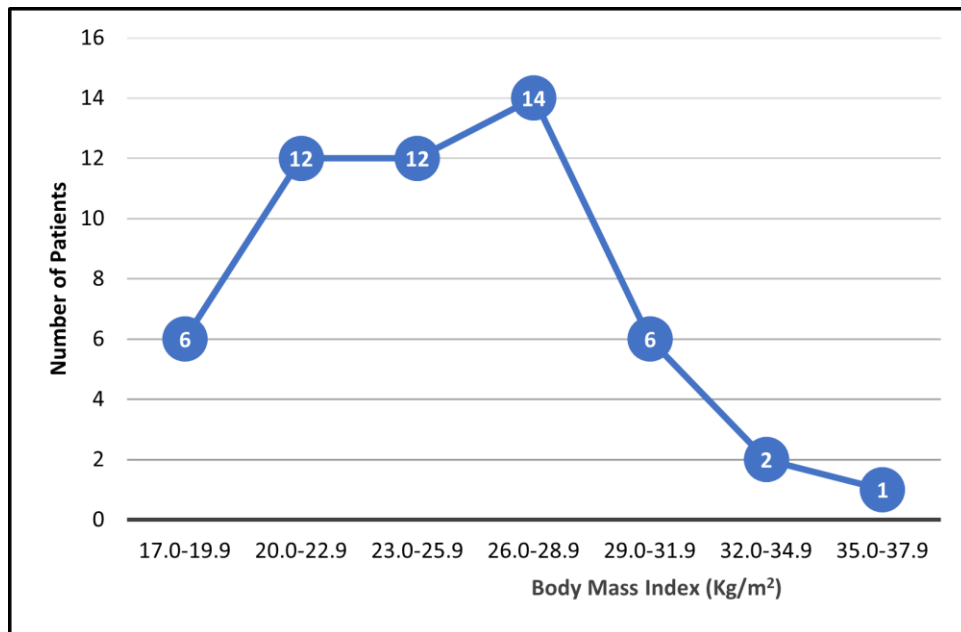


Figure-6.2: Representation of BMI for both cervical and prostate cancer patients

6.3 CTDI Volume (CTDI_{vol})

CTDI₁₀₀ values for center and periphery of the phantom are recorded corresponding to mAs values. Average value of periphery doses are taken. Then CTDI_w is calculated using central and peripheral CTDI₁₀₀ values. Since we have taken axial CT scan on the phantom, the pitch is 1. Therefore, CTDI_{vol} values are the same as CTDI_w values. The values of CTDI_{vol} were in the range 8.952 mSv 37.327 mSv. The mean±SD value of CTDI_{vol} was obtained as 18.979±5.914 mSv. Table 6.2 lists the measured values of CTDI_{vol}.

Table-6.2: Values of CTDI_{vol}

mAs	CTDI₁₀₀ (C) (mGy)	CTDI₁₀₀ (P) Avg. (mGy)	CTDI_w (mGy)	CTDI_{vol} (mGy)
99	5.38	10.738	8.952	8.952
123	6.65	13.123	10.966	10.966
128	6.92	13.565	11.350	11.350
142	7.64	15.025	12.564	12.564
148	8.04	15.768	13.192	13.192
150	8.15	15.968	13.362	13.362
152	8.26	16.168	13.532	13.532
161	8.73	17.003	14.245	14.245
168	9.16	17.860	14.957	14.957
173	9.42	18.653	15.575	15.575
174	9.47	18.910	15.764	15.764
180	9.73	19.313	16.118	16.118
186	10.25	20.248	16.916	16.916
194	10.29	20.763	17.272	17.272
208	11.25	22.280	18.603	18.603
211	11.43	22.578	18.862	18.862
221	12.05	23.453	19.653	19.653
228	12.51	24.615	20.580	20.580
229	12.53	24.865	20.754	20.754
241	13.18	25.573	21.442	21.442
248	13.40	26.925	22.417	22.417
253	13.68	27.468	22.872	22.872
257	14.00	27.523	23.016	23.016
261	14.16	27.863	23.295	23.295
262	14.20	27.993	23.395	23.395
264	14.22	28.095	23.470	23.470
274	15.12	29.710	24.847	24.847
297	16.28	32.235	26.917	26.917
301	16.44	32.490	27.140	27.140
421	22.92	44.53	37.327	37.327

The CTDI_{vol} values are represented as a function of mAs in Figure-6.3.

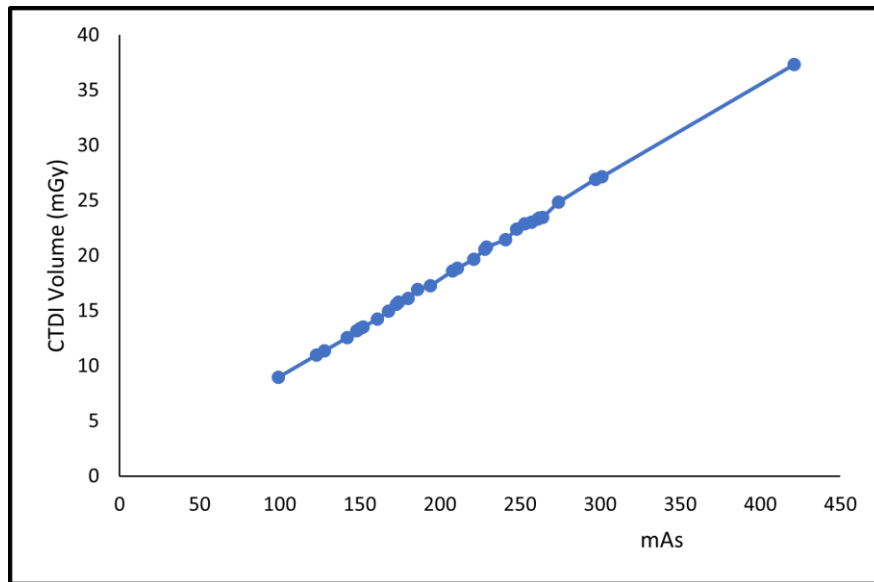


Figure-6.3: Presentation of CTDI volume as a function of mAs

6.4 CT Effective Dose

Table 6.3 lists the effective doses for CT scanning calculated for the pelvic regions. As mentioned earlier, the effective doses for CT procedure have been estimated using DLP values that represent radiation exposure to the entire scan length. The value of coefficient k used in this estimation was $0.015 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the pelvic region. The mean effective dose for the said region was obtained as $16.275 \pm 7.180 \text{ mSv}$. The doses for male and female patients were $15.125 \pm 8.815 \text{ mSv}$ and $16.417 \pm 8.930 \text{ mSv}$ respectively.

Table-6.3: CT effective doses for pelvic region

Anatomic Region	k (mSv. $\text{mGy}^{-1}.\text{cm}^{-1}$)	Effective Dose (mSv)			Phantom Diameter (cm)
		Male	Female	Total	
Pelvis	0.015	15.125 ± 8.815	16.417 ± 8.930	16.275 ± 7.180	32

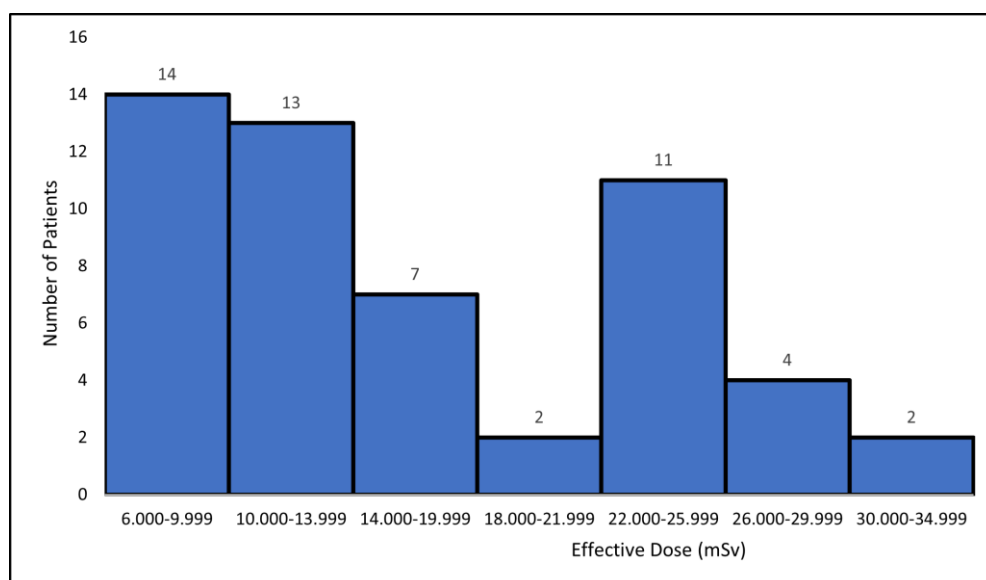


Figure-6.4: Demonstration of CT effective dose for pelvic region

Table-6.4 lists the CT effective doses for the cervical cancer patients. The values of DLP were in the range 411.225 mGy.cm to 1932.419 mGy.cm with an average of 1094.478 ± 462.029 mGy.cm. The mean effective dose for the cervix patients were 16.417 ± 7.180 mSv in the range of 6.168 mSv to 30.464 mSv.

Table-6.4: Effective Dose for Cervix Patients

Patient ID	mAs	CTDI _{vol} (mGy)	Scan Length (cm)	DLP (mGy*cm)	k (mSv/(mGy*cm))	Effective Dose (mSv)
18-00027	241	21.442	32.66	700.296	0.015	10.504
18-00029	180	16.118	38.96	627.957	0.015	9.419
18-00052	128	11.350	45.76	519.376	0.015	7.791
18-00062	229	20.754	49.82	1033.964	0.015	15.509
18-00066	194	17.272	37.44	646.664	0.015	9.699
18-00090	208	18.603	38.37	713.797	0.015	10.707
18-00095	262	23.395	44.55	1042.247	0.015	15.634
18-00104	301	27.140	33.65	913.261	0.015	13.699
18-00118	173	15.575	41.43	645.272	0.015	9.679
18-00120	241	21.442	39.27	842.027	0.015	12.630
18-00148	263	23.470	42.43	995.832	0.015	14.937
19-00048	261	23.295	35.24	820.916	0.015	12.314
19-00115	248	22.417	43.41	973.123	0.015	14.597
19-00182	297	26.917	41.27	1110.865	0.015	16.663

19-00185	262	23.395	34.39	804.554	0.015	12.068
19-00196	211	18.862	45.18	852.185	0.015	12.783
19-00203	174	16.764	44.71	749.518	0.015	11.243
19-00239	148	13.192	38.19	503.802	0.015	7.557
19-00365	173	15.575	38.62	601.507	0.015	9.023
19-00390	274	24.847	38.87	965.803	0.015	14.487
19-00431	257	23.016	39.28	904.068	0.015	13.561
19-00521	228	20.580	37.38	769.280	0.015	11.539
20-00013	150	13.362	37.36	499.204	0.015	7.488
20-00039	421	37.327	39.29	1466.578	0.015	21.999
20-00040	421	37.327	39.03	1456.873	0.015	21.853
20-00041	421	37.327	47.05	1756.235	0.015	26.343
20-00081	257	23.016	40.27	926.86	0.015	13.903
20-00097	229	20.754	50.78	1053.888	0.015	15.808
20-00134	186	16.916	37.91	641.286	0.015	9.619
20-00154	253	22.872	33.90	775.361	0.015	11.630
20-00166	161	14.245	42.07	599.287	0.015	8.989
20-00168	123	10.966	37.50	411.225	0.015	6.168
20-00251	173	15.575	38.05	592.628	0.015	8.889
20-00382	421	37.327	42.75	1595.729	0.015	23.936
21-RT020	421	37.327	50.36	1627.988	0.015	24.419
21-RT042	421	37.327	51.77	1932.419	0.015	28.986
21-RT064	421	37.327	42.64	1591.623	0.015	23.874
21-RT104	421	37.327	44.42	1658.065	0.015	24.871
21-RT108	421	37.327	47.87	1786.843	0.015	26.802
21-RT162	421	37.327	50.88	1899.198	0.015	28.488
21-RT171	421	37.327	39.87	1488.227	0.015	22.323
21-RT172	421	37.327	39.44	1472.177	0.015	22.083
22-RT027	421	37.327	44.56	1663.291	0.015	24.949
22-RT028	421	37.327	42.63	1591.250	0.015	23.869
22-RT065	421	37.327	39.32	1467.698	0.015	22.015
22-RT074	421	37.327	46.06	1719.282	0.015	25.789
22-RT098	421	37.327	54.41	2030.96	0.015	30.464

The effective doses for the prostate cancer patients listed on Table-6.4. The average value of DLP were 1008.347 ± 587.668 mGy.cm in the range 434.918 mGy.cm to 2004.087 mGy.cm. The mean effective dose for the cervix patients were 16.417 ± 7.180 mSv in the range of 6.168 mSv to 30.464 mSv.

Table-6.5: Effective Dose for Prostate Patients

Patient ID	mAs	CTDI _{vol} (mGy)	Scan Length (cm)	DLP (mGy*cm)	k (mSv/(mGy*cm))	Effective Dose (mSv)
19-00135	152	13.532	32.14	434.918	0.015	6.524
19-00486	99	8.952	88.41	791.446	0.015	11.872
20-00006	168	14.957	42.62	637.467	0.015	9.562
20-00109	221	19.652	82.29	1617.163	0.015	24.257
20-00167	142	12.564	44.97	565.003	0.015	8.475
21-RT161	421	37.327	53.69	2004.087	0.015	30.061

6.5 Discussion on CTDI_{vol} result

The present result of CTDI_{vol} were compared with those measured by Jian Xu et al. [1], Issahaku Shirazu et al. [3], Faruk Husremović et al. [4], Hamid Osman et al. [5] and ICRP Publication 103 [2]. The measured values of Jian Xu et al. [1] was 55.10±3.06 mGy, Issahaku Shirazu et al. [3] was 6.33 mGy, Faruk Husremović et al. [4] was 12.44 mGy, Hamid Osman et al. [5] was 10.8 mGy and ICRP Publication 103 [2] was 43.05 mGy respectively. The CTDI_{vol} measured in the present study is 18.979±5.914 mGy, which is not the lowest but smaller than that of Jian Xu et al. [1] and ICRP Publication 103 [2].

Table-6.6: A comparison of the present “CTDI_{vol}” results with reported series

Study	CTDI _{vol} (mGy)
Jian Xu et al. [1]	55.10±3.06
ICRP Publication 103 [2]	43.05
Issahaku Shirazu et al. [3]	6.33
Faruk Husremović et al. [4]	12.44
Hamid Osman et al. [5]	10.8
Present Study	18.979±5.914

The reference value of CTDI volume is 25 mGy according to American College of Radiology (ACR) [10] and European Commission [11]. This suggests that the measured value of CTDI_{vol} in the present study is within reference value.

6.6 Discussion on CT effective dose result

The present result of Effective doses were compared with those measured by J. Aldrich A et al. [7], Daniela Origgi et al. [8], H. Y. Tsai et al. [6], G Brix et al. [9], Hamid Osman et al. [5]. Their measured value of effective dose were 9 mSv, 8.8 mSv, 7.7 ± 5.7 mSv, 7.2 mSv, 7.2 mSv respectively, whereas, the measured value of effective dose is 16.275 ± 7.180 mSv, which is significantly higher than these reports.

Table-6.7: A comparison of the present “Effective Dose” results with reported series

Study	Effective Dose (mSv)
J. Aldrich A et al. [7]	9
Daniela Origgi et al. [8]	8.8
H. Y. Tsai et al. [6]	7.7 ± 5.7
G Brix et al. [9]	7.2
Hamid Osman et al. [5]	7.2
Present Study	16.275 ± 7.180

The reference value of CT effective dose for pelvis is 6-10 mSv according to American Association of Physicists in Medicine (AAPM) [12] and 7-8 according to International Commission on Radiological Protection (ICRP) [13]. This indicates that the result of CT effective dose for pelvis is above the reference value.

The reason of the higher effective dose is that, the patients underwent CT scan for radiotherapy treatment planning procedure via the process of CT simulation. The scan region of the patients exceeded the pelvic region causing higher scan length. Since effective dose is directly proportional to the scan length, the effective dose were higher for larger scan length.

One limitation of this study is the small sample size. Conducting a larger-scale study with a more diverse group of patients could yield more robust results that are representative of the entire population of Bangladesh.

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<https://w3.aapm.org/media/index.php>
- [13] International Commission on Radiological Protection (ICRP) <https://www.icrp.org/>

CHAPTER SEVEN

CONCLUSION

CT imaging in Bangladesh have grown to be essential in the medical field, offering crucial diagnostic capabilities for various conditions. The healthcare system has made notable progress with the adoption of advanced CT imaging technology in both cities and rural regions, aiding in the diagnosis of diseases like cancer, cardiovascular issues, and injuries related to trauma. During a CT scan, a patient gets exposed to ionizing radiation which carries inherent risks. Because higher radiation exposure from the machine may trigger secondary malignancies. In this study we determined CTDI volume that illustrates the amount of radiation delivered to the patient by the CT scanner, which ensures patient safety and quality of the scanner, and CT effective dose that unravel the biological risk of radiation exposure by estimating the amount of risk a patient is facing to induce secondary cancer.

The present study has estimated effective patient dose from CT scan during CT simulation process on 53 cancer patients of different ages, health conditions, and gender. The dose estimations were carried out determining CTDI volume on cylindrical PMMA phantom that mimic human tissues and employing simple Dose-Length-Product (DLP) method for CT. The results have been compared with other studies carried out in different countries under different protocols.

The average CTDI volume is obtained as 18.879 ± 5.914 mGy in the range of 8.952 to 37.327 mGy, which is not the lowest but significantly lower compared to some studies.

The effective doses for CT procedure have been estimated using DLP values that represent radiation exposure to the entire scan length. The value of dose conversion coefficient k used in this estimation is $0.015 \text{ mSv.mGy}^{-1}.\text{cm}^{-1}$ for the pelvic region. The mean effective dose for the pelvic region is obtained as 16.271 ± 7.180 mSv. The effective doses for cervix and prostate patients were 16.417 ± 8.930 mSv and 15.125 ± 8.815 mSv respectively. The present result of effective dose is moderately higher than the other studies compared.

The reason for the higher effective dose is that the scan length is extended beyond the regular scan region. This extension is necessary because the scans are performed on patients for radiotherapy treatment planning through the process of CT simulation, ensuring a comprehensive view of the surrounding anatomy and clear capture of the targeted area.

Therefore, this study will hopefully provide the necessary information about the radiation dose delivered by a CT scanner and the effective dose calculation for cervical and prostate cancer patients using other CT scanners throughout the country. It aims to help carry out proper justification of radiation doses.

APPENDIX

Table A1: Data of Cervix Patients

SN	Patient ID	Age (years)	Height (m)	Weight (kg)	BMI (kgm ²)
01	18-00027	55	1.47	56	25.9
02	18-00029	40	1.54	52	21.9
03	18-00052	35	1.41	50	25.1
04	18-00062	45	1.49	65	29.3
05	18-00066	50	1.57	67	27.2
06	18-00090	45	1.58	72	28.8
07	18-00095	65	1.49	58	26.1
08	18-00104	40	1.56	54	22.2
09	18-00118	50	1.45	50	23.8
10	18-00120	65	1.37	43	22.9
11	18-00148	55	1.59	64	26.3
12	19-00048	50	1.52	42	18.2
13	19-00115	48	1.41	60	30.2
14	19-00182	39	1.55	50	20.8
15	19-00185	50	1.52	60	26.0
16	19-00196	60	1.53	72	30.8
17	19-00203	45	1.42	54	26.8
18	19-00239	45	1.41	35	17.6
19	19-00365	58	1.45	57	27.1
20	19-00390	55	1.48	50	22.8
21	19-00431	45	1.51	68	29.8
22	19-00521	50	1.47	43	19.9
23	20-00013	61	1.41	48	24.1
24	20-00039	55	1.46	53	24.9
25	20-00040	50	1.57	80	32.5
26	20-00041	38	1.47	74	34.2
27	20-00081	51	1.49	68	30.6
28	20-00097	55	1.45	52	24.7
29	20-00134	35	1.48	60	27.4
30	20-00154	45	1.47	55	25.5
31	20-00166	50	1.55	48	20.0
32	20-00168	60	1.47	65	30.1
33	20-00251	43	1.52	48	20.8
34	20-00382	48	1.50	60	26.7
35	21-RT020	40	1.45	50	23.8
36	21-RT042	52	1.51	39	17.1

37	21-RT064	54	1.45	36	17.1
38	21-RT104	50	1.58	56	22.4
39	21-RT108	55	1.55	63	26.2
40	21-RT162	48	1.52	54	23.4
41	21-RT171	51	1.60	65	25.4
42	21-RT172	55	1.65	58	21.3
43	22-RT027	45	1.45	52	24.7
44	22-RT028	44	1.55	69	28.7
45	22-RT065	38	1.50	50	22.2
46	22-RT074	40	1.51	62	27.2
47	22-RT098	65	1.49	59	26.6

Table A2: Data for Prostate Patients

SN	Patient ID	Age (years)	Height (m)	Weight (kg)	BMI (kgm²)
1	19-00135	58	1.63	65	24.5
2	19-00486	65	1.58	53	21.2
3	20-00006	55	1.57	55	22.3
4	20-00109	75	1.62	93	35.4
5	20-00167	54	1.67	77	27.6
6	21-RT161	62	1.52	41	17.7

Table A3: Readings for 120 kV and 99 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI ₁₀₀) ₁ (mGy)	(CTDI ₁₀₀) ₂ (mGy)	(CTDI ₁₀₀) ₃ (mGy)	(CTDI ₁₀₀) _{av} (mGy)
Centre	0.553	0.532	0.547	0.538	5.53	5.32	5.47	5.38
P 1	1.118	1.154	1.162	1.147	11.18	11.54	11.62	11.47
P 2	1.062	1.109	1.066	1.079	10.62	11.09	10.66	10.79
P 3	0.975	0.985	0.976	0.979	9.75	9.85	9.76	9.79
P 4	1.089	1.069	1.112	1.090	10.89	10.69	11.12	10.90

Table A4: Readings for 120 kV and 123 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI ₁₀₀) ₁ (mGy)	(CTDI ₁₀₀) ₂ (mGy)	(CTDI ₁₀₀) ₃ (mGy)	(CTDI ₁₀₀) _{av} (mGy)
Centre	0.662	0.669	0.664	0.665	6.62	6.69	6.64	6.65
P 1	1.427	1.416	1.411	1.418	14.27	14.16	14.11	14.18
P 2	1.348	1.330	1.352	1.334	13.48	13.30	13.52	13.34
P 3	1.138	1.115	1.175	1.143	11.38	11.15	11.75	11.43
P 4	1.350	1.347	1.364	1.354	13.50	13.47	13.64	13.54

Table A5: Readings for 120 kV and 128 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI ₁₀₀) ₁ (mGy)	(CTDI ₁₀₀) ₂ (mGy)	(CTDI ₁₀₀) ₃ (mGy)	(CTDI ₁₀₀) _{av} (mGy)
Centre	0.689	0.697	0.691	0.692	6.89	6.97	6.91	6.92
P 1	1.446	1.442	1.453	1.447	14.46	14.42	14.53	14.47
P 2	1.368	1.362	1.357	1.362	13.68	13.62	13.57	13.62
P 3	1.224	1.235	1.243	1.234	12.24	12.35	12.43	12.34
P 4	1.382	1.379	1.387	1.383	13.82	13.79	13.87	13.83

Table A6: Readings for 120 kV and 142 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.761	0.767	0.763	0.764	7.61	7.67	7.63	7.64
P 1	1.596	1.571	1.577	1.581	15.96	15.71	15.77	15.81
P 2	1.551	1.528	1.519	1.533	15.51	15.28	15.19	15.33
P 3	1.347	1.389	1.376	1.371	13.47	13.89	1.376	13.71
P 4	1.543	1.512	1.512	1.525	15.43	15.12	15.12	15.25

Table A7: Readings for 120 kV and 148 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.807	0.802	0.803	0.804	8.07	8.02	8.03	8.04
P 1	1.663	1.664	1.672	1.666	16.63	16.64	16.72	16.66
P 2	1.570	1.608	1.611	1.596	15.70	16.08	16.11	15.96
P 3	1.453	1.427	1.433	1.432	14.53	14.27	14.33	14.32
P 4	1.613	1.622	1.604	1.613	16.13	16.22	16.04	16.13

Table A8: Readings for 120 kV and 150 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.815	0.813	0.818	0.815	8.15	8.13	8.18	8.15
P 1	1.685	1.684	1.688	1.686	16.85	16.84	16.88	16.86
P 2	1.592	1.630	1.597	1.606	15.92	16.30	15.97	16.06
P 3	1.457	1.454	1.457	1.456	14.57	14.54	14.57	14.56
P 4	1.667	1.629	1.623	1.639	16.67	16.29	16.23	16.39

Table A9: Readings for 120 kV and 152 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.829	0.823	0.826	0.826	8.29	8.23	8.26	8.26
P 1	1.707	1.703	1.710	1.707	17.07	17.03	17.10	17.07
P 2	1.613	1.652	1.618	1.627	16.13	16.52	16.18	16.27
P 3	1.476	1.473	1.468	1.472	14.76	14.73	14.68	14.72
P 4	1.689	1.650	1.644	1.661	16.89	16.50	16.44	16.61

Table A10: Readings for 120 kV and 161 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.875	0.874	0.869	0.873	8.75	8.74	8.69	8.73
P 1	1.798	1.792	1.803	1.798	17.98	17.92	18.03	17.98
P 2	1.702	1.723	1.718	1.714	17.02	17.23	17.18	17.14
P 3	1.550	1.568	1.525	1.548	15.50	15.68	15.25	15.48
P 4	1.748	1.733	1.741	1.741	17.48	17.33	17.41	17.41

Table A11: Readings for 120 kV and 168 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.916	0.915	0.917	0.915	9.16	9.15	9.17	9.15
P 1	1.886	1.887	1.892	1.888	18.86	18.87	18.92	18.88
P 2	1.849	1.828	1.854	1.844	18.49	18.28	18.54	18.44
P 3	1.576	1.547	1.547	1.557	15.76	15.47	15.47	15.57
P 4	1.862	1.853	1.849	1.855	18.62	18.53	18.49	18.55

Table A12: Readings for 120 kV and 173 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.934	0.942	0.949	0.942	9.34	9.42	9.49	9.42
P 1	1.947	2.116	2.123	2.062	19.47	21.16	21.23	20.62
P 2	1.922	1.842	1.843	1.867	19.22	18.42	18.43	18.67
P 3	1.612	1.673	1.626	1.637	16.12	16.73	16.26	16.37
P 4	1.885	1.894	1.906	1.895	18.85	18.94	19.06	18.95

Table A13: Readings for 120 kV and 174 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.938	0.948	0.954	0.947	9.38	9.48	9.54	9.47
P 1	2.213	2.128	2.116	2.152	22.13	21.28	21.16	21.52
P 2	1.913	1.863	1.835	1.870	19.13	18.63	18.35	18.70
P 3	1.621	1.676	1.624	1.643	16.21	16.76	16.24	16.43
P 4	1.917	1.895	1.885	1.899	19.17	18.95	18.85	18.99

Table A14: Readings for 120 kV and 180 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.971	0.974	0.975	0.973	9.71	9.74	9.75	9.73
P 1	2.027	2.122	2.079	2.076	20.27	21.22	20.79	20.76
P 2	1.975	1.968	1.963	1.965	19.75	19.68	19.63	19.65
P 3	1.743	1.727	1.747	1.739	17.43	17.27	17.47	17.39
P 4	1.957	1.949	1.928	1.945	19.57	19.49	19.28	19.45

Table A15: Readings for 120 kV and 186 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.025	1.023	1.026	1.025	10.25	10.23	10.26	10.25
P 1	2.129	2.206	2.160	2.165	21.29	22.06	21.60	21.65
P 2	2.055	2.078	2.106	2.079	20.55	20.78	21.06	20.79
P 3	1.836	1.824	1.833	1.831	18.36	18.24	18.33	18.31
P 4	2.064	2.011	1.998	2.024	20.64	20.11	19.98	20.24

Table A16: Readings for 120 kV and 194 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	0.998	1.047	1.044	1.029	9.98	10.47	10.44	10.29
P 1	2.212	2.176	2.184	2.191	22.12	21.76	21.84	21.91
P 2	2.065	2.151	2.116	2.111	20.65	21.51	21.16	21.11
P 3	1.904	1.924	1.932	1.920	19.04	19.24	19.32	19.20
P 4	2.091	2.012	2.147	2.083	20.91	20.12	21.47	20.83

Table A17: Readings for 120 kV and 208 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.122	1.121	1.132	1.125	11.22	11.21	11.32	11.25
P 1	2.332	2.371	2.342	2.348	23.32	23.71	23.42	23.48
P 2	2.214	2.266	2.307	2.262	22.14	22.66	23.07	22.62
P 3	2.072	2.041	2.063	2.059	20.72	20.41	20.63	20.59
P 4	2.242	2.254	2.233	2.243	22.42	22.54	22.33	22.43

Table A18: Readings for 120 kV and 211 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.138	1.149	1.141	1.143	11.38	11.49	11.41	11.43
P 1	2.365	2.424	2.392	2.394	23.65	24.24	23.92	23.94
P 2	2.343	2.243	2.238	2.275	23.43	22.43	22.38	22.75
P 3	2.055	2.105	2.089	2.083	20.55	21.05	20.89	20.83
P 4	2.277	2.284	2.275	2.279	22.77	22.84	22.75	22.79

Table A19: Readings for 120 kV and 221 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.202	1.208	1.204	1.205	12.02	12.08	12.04	12.05
P 1	2.457	2.539	2.482	2.493	24.57	25.39	24.82	24.93
P 2	2.352	2.348	2.351	2.350	23.52	23.48	23.51	23.50
P 3	2.153	2.178	2.181	2.171	21.53	21.78	21.81	21.71
P 4	2.368	2.387	2.346	2.367	23.68	23.87	23.46	23.67

Table A20: Readings for 120 kV and 228 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.245	1.255	1.254	1.251	12.45	12.55	12.54	12.51
P 1	2.576	2.543	2.558	2.559	25.76	25.43	25.58	25.59
P 2	2.531	2.529	2.517	2.526	25.31	25.29	25.17	25.26
P 3	2.214	2.222	2.231	2.222	22.14	22.22	22.31	22.22
P 4	2.553	2.527	2.536	2.539	25.53	25.27	25.36	25.39

Table A21: Readings for 120 kV and 229 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.254	1.256	1.248	1.253	12.54	12.56	12.48	12.53
P 1	2.589	2.588	2.588	2.588	25.89	25.88	25.88	25.88
P 2	2.532	2.551	2.572	2.552	25.32	25.51	25.72	25.52
P 3	2.224	2.238	2.241	2.241	22.24	22.38	22.41	22.41
P 4	2.590	2.541	2.564	2.565	25.90	25.41	25.64	25.65

Table A22: Readings for 120 kV and 241 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.317	1.314	1.323	1.318	13.17	13.14	13.23	13.18
P 1	2.715	2.727	2.688	2.710	27.15	27.27	26.88	27.10
P 2	2.555	2.596	2.569	2.573	25.55	25.96	25.69	25.73
P 3	2.348	2.349	2.351	2.349	23.48	23.49	23.51	23.49
P 4	2.578	2.603	2.609	2.597	25.78	26.03	26.09	25.97

Table A23: Readings for 120 kV and 248 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.338	1.339	1.344	1.340	13.38	13.39	13.44	13.40
P 1	2.774	2.779	2.795	2.783	27.74	27.79	27.95	27.83
P 2	2.682	2.874	2.757	2.771	26.82	28.74	27.57	27.71
P 3	2.405	2.447	2.443	2.432	24.05	24.47	24.43	24.32
P 4	2.788	2.667	2.896	2.784	27.88	26.67	28.96	27.84

Table A24: Readings for 120 kV and 253 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.365	1.371	1.368	1.368	13.65	13.71	13.68	13.68
P 1	2.830	2.836	2.852	2.839	28.30	28.36	28.52	28.39
P 2	2.736	2.812	2.832	2.793	27.36	28.12	28.32	27.93
P 3	2.453	2.497	2.492	2.481	24.53	24.97	24.92	24.81
P 4	2.822	2.955	2.845	2.874	28.22	29.55	28.45	28.74

Table A25: Readings for 120 kV and 257 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.397	1.399	1.404	1.400	13.97	13.99	14.04	14.00
P 1	2.893	2.830	2.925	2.883	28.93	28.30	29.25	28.83
P 2	2.778	2.759	2.768	2.763	27.78	27.59	27.68	27.63
P 3	2.671	2.528	2.504	2.568	26.71	25.28	25.04	25.68
P 4	2.786	2.792	2.807	2.795	27.86	27.92	28.07	27.95

Table A26: Readings for 120 kV and 261 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.413	1.415	1.420	1.416	14.13	14.15	14.20	14.16
P 1	2.927	2.862	2.958	2.916	29.27	28.62	29.58	29.16
P 2	2.812	2.790	2.806	2.803	28.12	27.90	28.06	28.03
P 3	2.731	2.548	2.533	2.604	27.31	25.48	25.33	26.04
P 4	2.818	2.825	2.822	2.822	28.18	28.25	28.22	28.22

Table A27: Readings for 120 kV and 262 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.427	1.415	1.418	1.420	14.27	14.15	14.18	14.20
P 1	2.953	2.884	2.940	2.926	29.53	28.84	29.40	29.26
P 2	2.846	2.811	2.828	2.828	28.46	28.11	28.28	28.28
P 3	2.652	2.567	2.552	2.591	26.52	25.67	25.52	25.91
P 4	2.861	2.859	2.835	2.852	28.61	28.59	28.35	28.52

Table A28: Readings for 120 kV and 263 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.423	1.416	1.428	1.422	14.23	14.16	14.28	14.22
P 1	2.895	2.956	2.962	2.938	28.95	29.56	29.62	29.38
P 2	2.838	2.854	2.827	2.839	28.38	28.54	28.27	28.39
P 3	2.573	2.588	2.636	2.599	25.73	25.88	26.36	25.99
P 4	2.877	2.861	2.849	2.862	28.77	28.61	28.49	28.62

Table A29: Readings for 120 kV and 274 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.511	1.496	1.529	1.512	15.11	14.96	15.29	15.12
P 1	3.192	3.333	3.203	3.243	31.92	33.33	32.03	32.43
P 2	2.957	2.906	2.912	2.925	29.57	29.06	29.12	29.25
P 3	2.655	2.635	2.689	2.659	26.55	26.35	26.89	26.59
P 4	2.955	3.130	3.087	3.057	29.55	31.30	30.87	30.57

Table A30: Readings for 120 kV and 297 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.631	1.626	1.627	1.628	16.31	16.26	16.27	16.28
P 1	3.432	3.339	3.334	3.368	34.32	33.39	33.34	33.68
P 2	3.165	3.155	3.161	3.160	31.65	31.55	31.61	31.60
P 3	3.079	3.012	2.989	3.027	30.79	30.12	29.89	30.27
P 4	3.468	3.265	3.258	3.339	34.68	32.65	32.58	33.39

Table A31: Readings for 120 kV and 301 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	1.647	1.642	1.643	1.644	16.47	16.42	16.43	16.44
P 1	3.467	3.373	3.398	3.403	34.67	33.73	33.98	34.03
P 2	3.197	3.187	3.192	3.192	31.97	31.87	31.92	31.92
P 3	3.111	3.042	3.019	3.057	31.11	30.42	30.19	30.57
P 4	3.503	3.237	3.291	3.344	35.03	32.37	32.91	33.44

Table A32: Readings for 120 kV and 421 mAs

Position	Dose 1 (mGy)	Dose 2 (mGy)	Dose 3 (mGy)	Average (mGy)	(CTDI₁₀₀)₁ (mGy)	(CTDI₁₀₀)₂ (mGy)	(CTDI₁₀₀)₃ (mGy)	(CTDI₁₀₀) av (mGy)
Centre	2.296	2.288	2.292	2.292	22.96	22.88	22.92	22.92
P 1	4.719	4.708	4.758	4.728	47.19	47.08	47.58	47.28
P 2	4.446	4.463	4.457	4.455	44.46	44.63	44.57	44.55
P 3	4.082	4.084	4.082	4.083	40.82	40.84	40.82	40.83
P 4	4.456	4.453	4.549	4.546	44.56	44.53	45.49	45.46