CHAPTER 1

PRINCIPLES OF RADIATION DOSIMETRY AND RADIATION THERAPY

1.1 INTRODUCTION

Cancer is one of the dreadful diseases and it’s a burgeoning health problem globally. Cancer or malignancy has been defined as a “group of abnormal mass of tissue, the growth of which exceeds in an uncoordinated manner with that of normal tissues and persists in the same excessive manner even after cessation of the stimuli which evoked the change”. It has the potential to spread to distant sites of the body through the blood stream and lymphatic system. The incidence of cancer is expected to rise worldwide notably because of increase in life expectancy, changes in life style pattern and environmental factors. World Health Organization (WHO) has predicted that cancer in all forms causes about 12.5% (7 million) deaths throughout the world and more than 10 million cases diagnosed every year. This devastating disease poses considerable challenges for health care professionals and its impact can be reduced through basic research and improvements in treatment approaches.

Generally, cancer is treated by surgery, radiotherapy or chemotherapy either separately or in combination of any of the above treatment modalities. Recently immunotherapy, photodynamic therapy, monoclonal antibody therapy or other methods are used as an alternative or complementary technique in the treatment of cancer. The choice of therapy
depends upon the location and grade of the tumor and the stage of the disease, as well as the general state of the patient (performance status). However among various cancer therapeutic modalities, more than 65% of patients are primarily treated with radiation therapy.

The goal of radiotherapy is to cure or locally control the cancer diseases while minimizing complications in normal tissues using ionizing radiation viz., X-rays, gamma ray photon beams and electron beams etc. Radiotherapy is classified broadly in to two main categories viz., teletherapy and brachytherapy. “Tele” means “distance” and the term “teletherapy” stands for distant treatment i.e., treating cancer by radiation from long distance using x-rays, gamma rays, electron beams and other charged particles. “Brachy” in greek means “short” and the term “Brachytherapy” stands for “Short distance treatment” i.e., treatment of cancer by keeping radioactive sources at short distance, usually in contact with the tumor. However the precise treatment and selecting the radiation source and type of therapy depends on various factors. In this regard, the present thesis is aimed to provide the details of radiation therapy and its dosimetry.

1.2 INTERACTION OF IONIZING RADIATIONS WITH MATTER

Ionizing radiation is radiation that has enough energy to remove electrons from atoms or molecules when it passes through or collides with some material. The loss of an electron with its negative charge causes the atom (or molecule) to become positively charged. The loss (or gain) of an electron is called ionization and the resulting charged atom is called an ion. Some forms of ionizing radiation are Gamma rays, Alpha particles, Beta particles, Neutrons.
When photons (x or γ rays) pass through a medium, interactions between photons and matter can take place, with the result the energy of the photon is transferred to the medium. The relative prevalence of any particular type of interaction depends on two major factors, they are the photon energy and the atomic composition of the medium. In some interactions, a part of the energy of the photon is transferred to the electrons of molecules in the tissue. These electrons, because of their short range dissipate their energy locally around the sites of interactions. If a photon enters the medium and if it merely changes its direction and leaves the medium with its entire energy intact, during an interaction, no radiation is delivered (Jayaraman and Lanzl 1996).

Attenuation of a photon beam by an absorbing material is caused by five major types of interaction. Among them photo disintegration is the reaction between photon and nucleus which is important at very high photon energies (>10MeV). The other four processes are coherent scattering, Photoelectric effect, Compton effect and pair production. Each of them can be represented by its own attenuation coefficient, which varies in its particular way with the energy of the photon and with the atomic number of the absorbing material. The total attenuation coefficient is the sum of individual coefficients for these processes.

\[
\mu/\rho = \sigma_{\text{coh}}/\rho + \tau/\rho + \sigma_{\text{c}}/\rho + \pi/\rho
\]

(1.1)

where \(\sigma_{\text{coh}}, \tau, \sigma_{\text{c}}\) and \(\pi\) are attenuation coefficients for coherent scattering, Photoelectric effect, Compton effect and pair production.
1.2.1 **Photoelectric Effect**

The photoelectric effect is a phenomenon in which a photon interacts with an atom and ejects one of the orbital electron from the atom. In this process, the entire photon energy $h\nu$ is transferred to that of the atomic electron. The kinetic energy of the ejected electron is equal to $h\nu - E_B$, where $E_B$ is the binding energy of the electron. Interaction of this type can take place with electrons in the K, L, M or N shells. After the electron has been ejected from the atom, a vacancy is created in the shell, thus leaving the atom in an excited state. The vacancy can be filled by an outer orbital electron with the emission of characteristic x-rays. There is also the possibility of emission of auger electrons, which are monoenergetic electrons produced by the absorption of characteristic x-ray, internally by the atom. Since the K shell binding energy of soft tissue is only about 0.5KeV, the energy of the characteristic photons produced in biologic absorbers is very low and can be considered to be locally absorbed. For higher energy photons and higher atomic number materials, characteristic photons are of higher energy and may deposit energy at large distances compared with the range of photo electron. In such cases, the local energy absorption is reduced by the energy emitted as characteristic radiation.

![Figure 1.1 Schematic representation of Characteristics of photoelectric effect](image-url)
1.2.2 Compton Effect

In the Compton process, the photon interacts with an atomic electron as though it were a free electron. The terms free here means that the binding energy of the electron is much less than the energy of the bombarding photon. In this interaction the electron receives some energy from the photon and is emitted at an angle $\theta$. The photon with the reduced energy is scattered at an angle $\phi$. Compton scattering depends mainly on the electron density of the material and it slowly decreases with energy. It predominates with a material having low atomic number, except at very low energies.

![Compton Effect Diagram](image)

**Figure 1.2 Schematic representation of Characteristics of Compton effect**

1.2.3 Pair Production

Pair production occurs at photon energies greater than 1.02 MeV and results in a creation of electron-positron pair due to the interaction of an incident photon with an atomic nucleus. After losing its kinetic energy, the positron is annihilated, producing two photons traveling in opposite directions. The probability of pair production increases with higher energies and atomic number ($Z$). The pair production process is an example of an event in which energy is converted into mass, as predicted by Einstein’s equation
E=mc². The reverse process, namely the conversion of mass into energy, takes place when a positron combines with an electron to produce two photons, called annihilation radiation.

![Diagram of pair production](image)

**Figure 1.3 Schematic representation of Characteristics of pair production**

![Graph of interaction processes](image)

**Figure 1.4 Predominance of interaction processes: photo electric Effect (κ), Compton scattering (σ), Coherent scattering (σcoh), Pair Production (πn in a nuclear field, πe in an electric field)**
Radiation therapy delivered primarily with high energy photons (gamma rays and X-rays) and charged particles (electrons, protons). The distinction between gamma rays and X-rays lies in their origin; gamma rays originate from excited and unstable nuclei, while X-rays are produced by electron energy transition within the atom or through the deceleration of high kinetic energy electrons (bremsstrahlung) (Attix 1986 and Khan 2003).

1.3 ABSORBED DOSE

The quantification of radiation has evolved over time. In the early 20th century, units of skin erythema and roentgen, a measure of ionization produced in air, were used. Today the accepted unit of measurement is “absorbed dose”, which is the energy absorbed per unit mass. Physically, the absorbed dose represents the energy deposited by secondary charged particles in the medium.

The unit of absorbed dose is Gray (abbreviated to Gy) after the British radiation physicist Louis Harold Gray, which is defined as the absorption of 1 joule per kilogram (J/Kg).

\[ 1 \text{ Gy} = 100 \text{ cGy (or) 100 rads} \quad (1.2) \]

1.3.1 Kerma

The quantity kerma (K) (kinetic energy released in the medium) is defined as “the quotient of \( \text{d}E_{\text{tr}} \) by dm, where \( \text{d}E_{\text{tr}} \) is the sum of the initial kinetic energies of all the charged ionization particles (electrons and positrons) liberated by uncharged particles (photons in the materials of mass dm” (Khan 1994).
The unit of kerma is the same as for dose i.e., J/kg.

\[ K = \frac{dE}{dm} \]  \hspace{1cm} (1.3)

Figure 1.5  Relationship between absorbed dose \((D)\) and collision kerma \((K^\text{col})\) for a mega voltage photon beam. \(\beta\) is the ratio of absorbed dose to collision kerma. The point designated as c.e.p is the center of electron production.

The relationship between absorbed dose \((D)\) and the collision part of kerma \(K^\text{col}\) is shown in Figure 1.5. When a broad beam of photons enters a medium. Whereas kerma is maximum at the surface and decreases with depth, the dose initially builds up to a maximum value and then decreases at the same rate as kerma. Before the two curves meet, the electron build up is less than complete, and

\[ \beta = \frac{D}{K^\text{col}} < 1 \]  \hspace{1cm} (1.4)
where $\beta$ is the quotient of absorbed dose at a given point and collision part of kerma at the same point. Because of increasing range of electron, complete electronic equilibrium does not exist within mega voltage photon beams. However, conceptually electronic equilibrium would exist if it were assumed that photon attenuation is negligible throughout the region of interest. Then

$$\beta = \frac{D}{K_{col}} = 1$$  \hspace{1cm} (1.5)$$

At depths greater than the maximum range of electrons, there is a region quasi equilibrium called the transient electron equilibrium in which

$$\beta = \frac{D}{K_{col}} > 1$$  \hspace{1cm} (1.6)$$

In the transient equilibrium region, $\beta$ is greater than unity because of the combined effect of attenuation of the photon beam and the predominantly forward motion of the electrons. Since the dose is being deposited by electrons originating upstream, one can think of point somewhere upstream at the distance less than maximum electron range from where the energy is effectively transported by secondary electrons. This point has been called the “centre of electron production”. Since the effective center of electron production is located upstream relative to the point of interest, the dose is greater than kerma in the region of transient electronic equilibrium.

1.4 **TELEThERAPy**

In external beam therapy or teletherapy, the radiation beam which is coming from a source located at a distance from the patient is directed towards the patient. The source of the radiation emitted is enclosed in a heavily shielded source head. As stated earlier, the prefix “tele” in tele therapy signifies the use of a large source to patient distance. In most practical
cases, this distance is in the range of 30 to 150 cm from the surface of the patient to the source head.

The design of individual teletherapy machines vary widely. There are two broad categories of teletherapy machines: Those that make use of the nuclear gamma rays from a radioisotope source, and accelerators, in which electrons are first subjected to forces from which they gain enough energy. The energetic electrons can be spread out produce an electron beam, or they can be directed to strike a target and generate an x-ray beam, for irradiating a patient. Some of the modern therapy accelerators are so versatile that they can produce electron beams of several energies and photon beams of one or two energies. Accelerators have also been used for production of beams of neutrons, protons, helium ions, and other heavy charged particles for radiotherapy. However, it is expensive to produce these beams, and they are currently being used at research facilities for clinical trials only.

1.4.1 Cobalt-60 Therapy unit

The majority of the Co\textsuperscript{60} units functions are operated and can be controlled either from a hand control located near the unit or from a control console located outside the treatment room. Most isocentric cobalt-60 Machines use source to axis distance of 80 cm. Later models for 100 SAD with higher activity (10,000 Ci) the high specific activity of cobalt-60 permits fabrication of small high activity sources 6000Ci in 1.5 to 2.0 cm diameter sources given dose rate about 1.5 to 2 Gy/min at 80cm. The penetration energies are 1.17-1.33 MeV gamma ray sources. Disadvantage of this units include the need for source replacement after 4 to 5 years, poor field flatness and lower depth dose compared with high energy photons from the linac.
1.4.2 Linear Accelerator

The linac is a device that uses high frequency electromagnetic waves to accelerate charged particles such as electrons to high energies through a linear tube high energy electron beam itself can be used for treating superficial tumors or it can be made to strike a target to produce x ray for treating deep-seated tumors. There are several types of linear accelerator designs, but the ones used in radiation therapy accelerate electrons either by traveling or stationary electromagnetic waves of frequency in the microwave region ~3000 mega cycles/sec. The difference between the traveling wave and stationary wave accelerators is the design of the accelerator or dummy load to absorb the residual power at the end of the structure thus preventing a backward reflected wave. So that the combination of forward and reverse traveling waves will give rise to stationary waves. However it is more expensive and requires installation of a circulator or isolator between the power source and the structure to prevent reflections from reaching the power source.

1.4.3 Characteristics of External Beam Photon Fields

Energy is carried in to the irradiated tissue primarily by the radiation coming from the source. However, the energy is distributed in the tissue primarily by secondary electrons set in motion by this primary radiation. The dose distribution depends mostly on the range of these secondary electrons.

For soft radiation, such as grenz rays and superficial x-rays, these secondaries are immediately set in motion at the surface, and travel immediately in all directions, but their energies are quickly absorbed. Below the surface, there is less and less primary radiation due to lack of penetrating
ability of the original radiation. Thus for a soft radiation, the maximum absorbed dose for a single field occurs at a skin.

For higher energy radiation the number of secondaries tends to build up more slowly primarily due to longer range of electrons. In addition, the secondary particles have less tendency to “bounce away” sideways; they tend to be set in motion more or less in the same direction as the primary was traveling (Shahabi 1989).

1.4.4 Characteristics of Electron Beams

There is one major difference between the absorption of electrons and photons. When a beam of photons enters a absorbing medium, the photons penetrate to all depths within the medium. Although the number of photons is constantly reduced as the beam penetrates deeper, the beam is still there, regardless of depth. Electrons, on the other hand, penetrate only to a certain depth (depending on the initial electron energy) but no deeper; i.e., electrons have a definite maximum range, whereas photons do not. For example, the range for 18MeV electrons is 9cm in water. This means that all electrons will stop by 9cm depth, but none will reach 9.5cm. The advantage of using electron is obvious when you consider diseased volume at a depth of 5cm with a sensitive healthy tissue directly under it at 10cm depth.

An empirical formula (empirical means it cannot be derived from valid physical principles, but just happens to work by coincidence) for the range \( R \) of electrons having energies useful in teletherapy is

\[
R \text{ (cm)} = 0.521 \ E_0 \text{ (MeV)} - 0.376
\]  (1.7)
where $R$ is the range in cm of water or muscle and $E_0$ is the initial energy of the electrons in MeV before they enter the water or muscle.

1.5 BRACHYTHERAPY

Brachytherapy is a technique in which the sealed radioactive sources are kept in contact with the tumor. Brachytherapy is characterized by a high dose gradient to the tumor and steep dose gradient in the surrounding normal tissues. The dose is delivered continuously either over a short period of time (temporary implants) or over the lifetime of the source to a complete decay (permanent implants). Most of the commonly used brachytherapy sources emit photons. However, in a few specialized situations beta or neutron emitting sources are used. From a radiobiological point of view, brachytherapy could result in a complex dose rate effects that may also influence the therapeutic outcome. The continuous delivery of dose will influence the repair of sublethal and potentially lethal damage, cell proliferation and other cell kinetics, all of which could modify the radiation response of tumor and normal tissues.

1.6 RECENT DEVELOPMENTS IN RADIATION THERAPY

Radiation therapy is a loco regional treatment modality for many malignant diseases and majority of the cancer patients need radiation treatment as a concomitant or adjuvant therapy.

1.6.1 Conventional Radiotherapy

Conventional or two dimensional radiotherapy (2DRT) is usually practiced with planar radiographs or single CT slices taken in central plane. Treatment planning and dose calculation are performed from a single two
dimensional slice (contour) through a given treatment volume. Practitioners use bony landmarks on plain simulation radiographs to identify the tumors and normal structures to draw blocks and align treatment beams. Radiation therapy is one of the principle treatment modalities especially for localized tumors, which constitutes about 60% of the patients collectively. However with conventional techniques, approximately one third of these tumors (18% of all cancer patients) cannot be cured, because treatment fails to stop the tumor growth. The fraction of deaths due to local failure depends strongly on the tumor site as follows: brain (95%), prostate (61%), uterine cervix (60%), esophagus (59%), bladder (54%), head and neck (41%), breast (14%) and lung (10%) (Webb 2001). There are many possibilities that the radiotherapy may fail to achieve a cure in localized tumors: (i) there may be radio resistive cell clones (ii) the dose to nearby critical normal structures may compromise the tumor dose (iii) the diagnostic modality may have failed to yield the true extent of tumor; (iv) there may be inaccuracies in dose planning and delivery. Issue (i) may be insuperable, (ii) is a challenge to treatment planning and largely governed by the physics of photon tissue interactions, (iii) demands the use of multimodality imaging, (iv) demands good immobilization, position verification and quality assurance. For example, Figure. 1.6 below shows a simplified but roughly to scale diagram of the prostate in relation to the bladder and the rectum. One sketch shows a side view and the other is a transverse slice through the body. The basic problem in external beam radiotherapy is to be able to subject the prostate to beams of radiation consisting of streams of high energy particles (photons) so that the prostate receives a sufficiently high dose of radiation to destroy the cancerous tissue without serious damage being inflicted on the surrounding organs - the rectum being the most sensitive to damage. It is clear that this cannot be achieved with a single beam because the organs in front of and behind the prostate would receive more or less the same level of radiation as the prostate.
To avoid this problem, beams are directed at the prostate from several angles and the Figure 1.7 shows a simple 3-beam arrangement with a frontal beam and two lateral beams. Where the beams intersect is the region of highest radiation intensity and this is centered on the prostate as shown in Figure 1.7.

1.6.2 Three Dimensional Conformal Radiotherapy (3DCRT)

The term “three dimensional conformal radiotherapy” (3DCRT) is reserved for treatments for which 3D anatomical information is derived from
modern cross-sectional imaging and whose dose distribution are shaped, so that the region of high dose conforms as closely as possible to the target volume while minimizing dose to the surrounding normal tissues. Thus, it is known as 3D- conformal therapy. Conformal therapy has been classified by DY NARAD (DYNA mic RADIotherapy) consensus report (Kolitsi et al 1997) according to the methodology and tools associated with each step of procedure, as level 1, level 2, level 3 conformal techniques namely, 3D conformal radiotherapy (3DCRT), intensity modulated radiotherapy (IMRT), Stereotactic radiosurgery/radiotherapy (SRS/SRT) with intensity modulation respectively. All these forms of conformal therapy are principally the same, concerned to the conformation of prescribed dose to the target shape in three dimension.

The conformal dose distribution is always accomplished from more number of incident beam directions, from which the tumor is irradiated with uniform fluence over the lateral profile of beam aperture. The absolute fluence values for the individual beam including the beam directions that are employed to achieve the desired dose sparing for Organs at Risk (OARs) are constant. This technique however fails in achieving conformal dose distribution for patient geometries where OARs are located in close proximity to or are even embedded within a complicated tumor shapes. Due to lesser conformality of dose distribution in3DCRT, it may be insufficient to allow adequate escalation of tumor dose and there is a need for further improvement in conformality of dose distribution to make feasible further escalation of tumor doses required. It is possible only with intensity modulated radiotherapy.
1.6.3 Intensity Modulated Radiotherapy (IMRT)

One of the weaknesses of 3-D conformal radiotherapy is that it is not possible to have three dimensional indentations in the surfaces of equal radiation dose. This is because the intensity of the radiation across the beam is constant. A refinement in technique that allows these three dimensional concave surfaces to be developed is intensity modulated radiotherapy - IMRT. The term “Intensity Modulation” has been widely used in order to express the spatial variation of the fluence within the beam opening. IMRT is the higher version of 3DCRT, in which the prescribed dose highly conforms to the target volume while low dose to surrounding normal structure. It has been postulated that highly conformal treatments can be realized by means of cross firing the target with the beams of varying intensity in the treatment plane across the field. This enables to achieve a uniform dose in regularly or irregularly even concave shaped targets, while avoiding the deposition of high dose to the surrounding critical and normal structures. Clinical rationale for the use of intensity modulated beams are found in the following situations: (i) when the target volume features the concave shapes (head and neck, Para spinal tumors), (ii) for the delivery of intentionally inhomogeneous dose distribution (Prostate, brain tumors), (iii) with restricted spatial range of favorable and feasible beam directions (mediastinum and lung) and (iv) to improve dose homogeneity (breast). The requirements of dose escalation, re-irradiation and the multiple targets with differential dose delivery simultaneously are more favorable clinical indication for IMRT treatment.

Figure.1.8 below is an illustration of the sort of intensity profiles that would produce a closer fitting of the radiation contours to the prostate. Intuitively, it is clear that beams of constant intensity could not achieve the same sort of 'fit' as beams of variable intensity
Clinical implementation of modern methods of IMRT requires at least three important components namely computer controlled linear accelerator, multileaf collimator (MLC) (Figure 1.9) and an inverse treatment planning system. IMRT delivered with multileaf collimator is a promising resource to achieve higher degree of conformal radiotherapy. Using MLC, there are two methods of producing intensity modulation: (i) step-and-shoot IMRT (ii) dynamic IMRT. Several other IMRT treatments techniques such as compensator based IMRT, sequential and helical tomotherapy based IMRT and intensity modulated arc therapy (IMAT) are also available.
1.6.4 SRS and SRT

Stereotatic radiosurgery (SRS) is a single fraction radiation therapy procedure for treating intracranial lesions using a combination of a stereotatic apparatus and narrow multiple beams delivered through noncoplanar isocentric arcs. The same procedure when used for delivering multiple dose fractions is called stereotatic radiotherapy (SRT). Both techniques involve three-dimensional imaging to localize the lesion and delivering treatment that concentrates the dose in the target volume and spares as much as possible the normal brain. A high degree of dose conformity is a hallmark of SRS, which is generally achieved by using appropriate circular beams to fit the lesion, optimization arc angles and weights and using multiple isocenters or dynamically shaping the field during arc rotations with mini( or micro) multileaf collimators. Accuracy of beam delivery is another hallmark of SRS. It is strictly controlled by a specially designed stereotatic apparatus, which is used to all steps of the process: imaging, target localization, head immobilization, and treatment set-up. Because of the critical nature of brain tissue, elaborate quality assurance procedures are observed. The best achievable mechanical accuracy in terms of isocenter displacement from the defined centre of target image is $0.2\text{mm} \pm 0.1\text{mm}$, although a maximum error of $\pm 1.0\text{ mm}$ is commonly accepted in view of the unavoidable uncertainties in target location.

1.6.5 Image Guided Radiotherapy

Traditionally, imaging technology has been used to produce three-dimensional scans of the patient anatomy to identify the exact location of the cancer tumor prior to treatment. However, difficulty arises when trying to administer the radiation since cancer tumors are constantly moving within the body (for example, from movement caused by breathing). Hence, the exact
location of the tumor may have changed between the time of scan and actual treatment.

IGRT combines a new form of scanning technology, which allows planar or X-ray volume imaging (XVI), with IMRT. This enables physicians to adjust the radiation beam based on the position of the target tumor and critical organs, while the patient is in the treatment position.

Radiation measurements and investigations of radiation effects require various specification of the radiation field at the point of interest. Radiation dosimetry deals with methods for a quantitative determination of energy deposited in a given medium by directly or indirectly ionizing radiations. A higher degree of overall accuracy in the delivery of the prescribed dose to patient is necessary for success in radiotherapy. Radiation dosimetry concerns precise measurements of absorbed dose received by patient and it plays vital role in providing accurate dose delivery. Since advanced treatment techniques are becoming increasingly conformal, the opportunities for geometric misses of the target and normal structure also increase, if not accurately defined. Similarly, any error in dosimetry of beam calibration and dose modeling which is the starting point for the accurate dose delivery, can result in under dose and overdose from the prescribed dose as a systematic error if not handled correctly. It is generally believed that the small error in broad beam dosimetry do not lead a major sequelae, whereas in narrow beam dosimetry which is applicable in conformal techniques, even small errors can have significant impact on an individual patient.

The intensity modulated beams are generated from the summation of many small, irregular, asymmetric and off-axis fields and its delivered dose is mainly function of transmission, penumbra and head scatter contribution. The required overall accuracy in dose delivery can be achieved through
rigorous investigation on the chain of dosimetry and quality assurance procedures. Although there is drastic technical improvement in the field of radiotherapy to treat cancer viz., 3DCRT, SRS, SRT, IMRT, and IGRT to safely implement them clinically comprehensive dose verifications are essential, including *in vitro* measurement using phantoms and *in vivo* measurements during patient treatment using reliable radiation detectors.

### 1.7 Radiation Detectors

Radiation is detected by measurement of the effect of its interaction with target materials. The radiation-induced effect on a detector produces a signal which can be interpreted to give the radiation quantity of interest. Radiation can be measured in two ways: (i) in phantom measurements which are also called as *in vitro* dosimetry (ii) Patient dosimetry which is also called as *in vivo* measurements. The detectors used for *in vitro* dosimetry are ionization chamber and Film and the detectors used for *in vivo* measurements are Thermoluminescence dosimeter and Semiconductor detectors.

### 1.8 In-Vitro Dosimetry

*In vitro* in Latin called as (with) in glass refers to a technique of performing a given experiment in glass or generally in a controlled environment outside the living organism. The measurement of radiation using tissues equivalent material or phantom is referred to as *in vitro* dosimetry. The detectors used for this kind of measurement are as follows:

#### 1.8.1 Ionization Chambers

Figure shows a cylindrical ion chamber detector connected to a commercial electrometer of high sensitivity for measurement of nanocolumb
quantities of charge. The electrometer has a built-in power supply that provides the necessary polarizing voltage to the ion chamber detector. For routine use, the chamber and the electrometer are commercially available as an integrated system, with the polarizing voltage, $v$, preset for optimum operation. Let us assume that $V$ can be varied, the effect of the applied voltage on the electrometer reading for alpha and beta radiations, representing densely and sparsely ionizing radiations, respectively. The figure suggests that an alpha particle crossing the chamber can be expressed to cause more ion pairs than are produced by a relatively more penetrating beta particle.

When a low voltage is applied to the radiation detector, the ions are not collected quickly, and some may recombine prior to collection. The detector’s response to a given amount of radiation (i.e.; to a radiation pulse) will increase with increases in the applied voltage until a saturation level is reached. This is shown as plateau I in this plateau; all the ions are collected, without any recombination loss. An ionization chamber should operate at a voltage that corresponds to this plateau. Ion chambers used in practice operate at 150 to 300 V of applied potential. At such potentials, the ions are merely collected while they are drifting with no acceleration. If the voltage applied across the detector is increased further, the ions receive acceleration and thus gain energy. Such ions of increased energy can produce even more secondary ions in the process of dissipating their energy. Such an increase in ionization is called “gas multiplication”. This effect results in regions P and G in the curves of which are called proportional counter and Geiger- Muller (GM) counter regions, respectively.

Proportional counters and GM counters operate in pulsed mode. That is a single photon or beta particle produces an avalanche or pulse of ions. Any single interaction in the detector creates an initial group of ions. At moderately high voltages (corresponding to region P), the size of the final
pulse remains proportional to the size of the initial pulse. It is for this reason that the detector is called a proportional counter. A proportional counter can discriminate between a high-pulsed (i.e., a large number of ions) and a low-pulsed initializing event. Still higher potentials cause the detector to operate in the GM counter region G, where even a small initial ionizing event is amplified which such a degree as to ionize the entire gas volume. Here an avalanche of electrons is released which results in a large pulse, with no discrimination between alpha and beta particles and no m dependence on the amount of energy that they lost in the initializing event. The region G is called the GM plateau. Still higher potentials, corresponding to region d of figure 2 will contribute to a direct discharge between the electrodes (caused by the potential itself). The direct discharge principle is used in a device called a spark counter, which is used in high energy particle physics.

1.8.2 Photographic Film Detector
1.8.2.1 Photographic process

A photographic emulsion is made of a polyester basic photosensitive silver halide (predominantly silver bromide) grains. The radiation or the light energy absorbed in the grains forms a latent image. The grains can be developed to become pure silver by treatment of the emulsion in a chemical

Developer. The developing solution makes use of the latent absorbed energy in the grains for a chemical reaction that reduces the silver halide to silver. Next, the film is treated with a chemical solution that fixes some of the silver to the film base and washes away the silver halide. At places on the film that received a high absorbed energy from the radiation, the silver concentration will be high, and the film consequently will be dark. At sites on the film that received negligible radiation, the film will appear clear
and transparent. Hence, the film will give an overall image of the radiation energy absorption pattern.

### 1.8.2.2 Optical density

The darkness (or response) of the film is evaluated in terms of optical density, which can be determined quantitatively with a densitometer. The densitometer shows in Figure 1.10 has a collimated light source below a light-sensing detector. Let us consider that the intensity measured initially (i.e., with no film in place) is \( I_0 \) when a film is placed between the light source and the detector, the intensity may decrease to a value \( I \). Then the optical density \( D \) is defined by

\[
D = \log_{10} \left( \frac{I}{I_0} \right)
\]

A density value of 1 means that the transmitted light intensity is \( I_0/10 \). A density of zero means 100% transmission of light by the film. However, this does not happen in practice. The film base filters some amount of light even without any darkening contributed by radiation. This darkness
inherent in the film base is referred to as “fog” density. The measured density minus the fog density is called the net density and is relevant for dosimetry.

1.8.2.3 Calibration of a film

Film is not an absolute dosimeter. The grain size, the developing time, developer concentration, temperature, and humidity all influence the outcome. Film should be considered to be merely a radiation sensor. For use of the film as a dosimeter, it is necessary to calibrate the response (i.e., the darkening) produced at the different doses. For such a calibration, the film should be irradiated by a therapy beam of known dose rate. The conditions of irradiating the film should closely match the conditions under which the calibration of the therapy beam holds. Photon beams are calibrated for conditions of full secondary-electron build-up. While the film is irradiated for calibration by photon beams, the film must be covered above and below with tissue-or air-equivalent materials of a thickness sufficient to give the needed secondary-electron build-up. Different irradiation times will match different dose or exposure levels for calibration purposes.

1.8.2.4 Film response curves

Figures 1.11 a and b presents the variation of the film density with exposure and with the logarithm of exposure, respectively. It can be seen from figure 1.6 that the density is proportional to exposures for low exposures and low densities. At higher density becomes less than proportional to exposure and finally saturates to a maximum value. When all grains have been developed. The shape of this curve should be known for any particular film for that film to be used as a dosimeter. Ideally, the exposure range should lie in the region of linear response some films have a low sensitivity and range, and others may saturate quickly.
Figure 1.11 Typical variation of net optical density with exposure for Photographic film (a) Linear-linear graph; (b) log-linear graph

The curve of Figure 1.11 is useful for the diagnostic application of films. This curve (which was first published in 1980 by Hurter and Driffield in England) is referred to as “H and D curve” or “characteristic curve”. It can be drawn for any film with or without intensifying screens. Diagnostic films are viewed with the help of a lighted box. Because the eye responds to light on a logarithmic scale, the level of exposure of the diagnostic should be in the linear region of the curve in Figure 1.11. The extent of the straight-line region is the “latitude” of the film. The slope of the curve is called “gamma” and is a measure of the film sensitivity. “Speed” is another term used in reference to sensitivity. “High speed” and “low speed” mean high and low sensitivities. At extremely high exposures (not shown in the figure), the density is known to decrease. (This phenomenon can be exploited for copying an exposed film onto another film by keeping them in contact and shining intense light through them).
1.9 IN-VIVO DOSIMETRY

In vivo dosimetry is the most direct method for monitoring the dose delivered to the patient receiving radiation therapy. When performed early in treatment as a supplement to the clinical quality assurance (QA) program, simple in vivo measurement are an additional safeguard against major setup errors and calculation or transcription errors that were missed during pretreatment chart check (Essers and Mijnheer 1996, Fontenla et al. 1996, Lanson 1999, Leunens et al. 1990) and In the absence of errors. Routine in vivo measurement uniquely document that treatment was delivered correctly within a user-specific tolerance. Unlike other QA method, in vivo dosimetry checks dose delivered to the patient rather than the individual components prior to treatment. Most treatments are without serious error-in recent review from Europe, out of 10,300 patients at three institutions performing in vivo dosimetry for all new patients, 120 treatment errors exceeding 5% were found and the estimated serious error (misadministration) rate in the United State is 0.002% (Feldman and Edwards 2001). Although there is not universal agreement on the benefit of in vivo dosimetry (Feldman and Edwards 2001), a strong argument in its favor is that preventing the severe consequences of major error-as illustrated by the recent overexposure of 28 patient in program In vivo dosimetry is also helpful in supporting the high accuracy in dose delivery expected from complex and conformal therapy technique. For these and other reasons, AAPM TG-40 recommends that clinics” should have access to TLD or other in vivo system” Kutcher et al. 1994.

Most in vivo dosimetry employs either silicon diodes or thermoluminescent dosimetry (TLDs). TLD dosimetry has been used for over 30 year. It is the method of choice in many large department, and the subject of much literature Essers and Mijnheer 1999, Mayles et al. 1993. However, diodes have gained popularity since the early 1980s due to their rapid
processing time (second vs. jour for TLD) and high sensitivity (over 18,000 times that of an air-filled ionization chamber of the same volume). (Van dam and Marinello 1994, Rikner and Grusell 1983, Rikner and Grusell 1987, Rikner 1983) with care, diode may equal or even surpass the accuracy of TLD (Loncol et al 1996) for in-field measurement.

TLD and diodes are used similarly for external beam in vivo measurements, although the underlying physics is very different. The dosimeter is placed on the patient’s skin, and dose to a point of interest inside the patient is inferred from surface measurement. In vivo entrance measurement, where dose at a point within the patient is inferred from the reading of a detector on the entrance surface compared with calculation, can detect numerous serious errors including an incorrect daily dose. Treatment with the wrong beam energy, omission or use of the wrong wedge, and setup errors such as a patient set up with SAD (source-to-axis distance) rather than SSD (source-to-phantom surface distance) technique (Lee et al 1994, Howlett et al 1999) changes in treatment machine output between calibration have also been reported (Essers and Mijnheer 1999) combined in vivo measurement at other entrance and exit point, while introducing extra complexity Meijer et al 2001, can detect additional errors, including large errors patient thickness and problem system for total body irradiation (TBI), combined entrance and exit measurement are used to assess the adequacy of missing tissue compensation.(Greig et al 1996 and Planskoy et al 1996).

The ranges of discrepancy between expectation and measurement beyond which clinical action must be taken are referred to as action or tolerance level—below, we shall call these simply “action level.” Regardless of the dosimeter used in an in vivo dosimetry program, the size of action level and the associated action are key decisions, requiring careful consideration of the clinical goals of the program and the accuracy that is reasonably
achievable with the available personnel and equipment. An interesting review of the “philosophies and result” of in vivo dosimetry has recently been published (Essers and Mijnheer 1999).

1.9.1 Thermoluminescent Dosimeters (TLD)

1.9.1.1 Thermoluminescence

Thermo luminescence is the phenomenon of light emission from solid crystals that are subjected to heating. The heat acts as an agent to shake off the excited electrons trapped in any metastable energy states in the solid. The electronic transition during the release of the trapped electrons results in the emission of light. The traps are metastable states created by lattice by the addition of the impurities. An example is the addition of manganese or magnesium as an impurity to an otherwise regular crystal lattice of LiF, with alternating lithium and fluorine atoms. The impurities sites have electron traps. Irradiation of the crystal lifts some of the electrons from normal energy levels to traps. These electrons contribute to the thermoluminescence observed on subsequent heating of the irradiated sample.

1.9.1.2 TLD Instrumentation

In practice, the thermo luminescence signal is rather week and has to be measured with an instrument called a TLD reader. Which consist of a dark enclosure with a heating pad a photomultiplier light sensor coupled to an amplifier, a meter for reading the integrated signal. A basic TLD reader system consists of a planchet for placing and heating the TLD, a PMT to detect the thermoluminescence light emission and convert it into an electrical signal linearly proportional to the detected photon fluence and an electrometer for recording the PMT signal as a charge or current.
Figure 1.12 A typical glow-curve (thermogra) of LiF: Mg, Ti measured with a TLD reader at a low heating rate

1.9.2 Semiconductor Diodes

A semiconductor is a substance that has electrical condition between that of a good conductor and that of an insulator. Silicon and germanium are well-known semiconductors that have found wide applications. Normally, a semiconducting substance will have an equal number of electrons and holes ("holes" refer to the presence of a local net positive charge caused by the because of an electron in a crystal site). Semiconductor materials can be modified or doped with electrons is referred to as “n-type” and that with excess holes as “p-type”. A semiconductor diode is a junction of p-type and n-type materials. If an electric field is applied across the p-n junction to have a forward bias with the positive terminal connected to the p-type a current can flow across the junction. If a reverse bias is applied as shown in Figure1.13 no current is observed; the excess electrons and holes are swept away from the junction, forming a thin charge "depletion layer" at the interface. When the charged particles cross the depletion layer, electron-hole pairs are produced.
These cause a current across the junction and produce a signal. Thus, the p-n junction functions like an ion chamber with a sensitive volume corresponding to the depletion layer.

![Diagram of a p-n junction semiconductor detector](image)

**Figure 1.13 Diagram of a p-n junction semiconductor detector**

### 1.10 PHYSICS OF THE SILICON DIODE USED AS A RADIATION DETECTOR

For over 30 years, the silicon semiconductor diode has been used as a radiation detector. The density of silicon and the low average energy required to form a carrier pair in silicon result in a radiation current density which is about 18,000 times that of air, allowing a small volume (approximately $10^{-2} - 10^{-1}$ mm$^3$) of silicon diode to produce an easily measured current. As a result, diodes have a high sensitivity (defined as change collected per unit dose to the diode). Their small volumes, mechanical ruggedness, and real-time readout make diodes attractive for in vivo dosimetry. However, the physics of charge generation and collection in semiconductor diodes introduces characteristic features that are relevant to their accurate use in this application. The key structure in the silicon diodes used for in vivo dosimetry is the p-n junction. N-type silicon is doped with impurities of a free of pentavalent element (e.g., phosphorous) called a "donor." Each donor can contribute a
free electron to the silicon. Therefore the majority carriers in n-type silicon are electron, and holes are the minority carriers. P-type silicon is doped with impurities of a trivalent element (e.g., boron) called n “acceptor” Each acceptor can accept an electron, resulting in a mobile hole in silicon that is equivalent to appositively charged carrier. In p-type silicon, holes are the majority carrier while electrons are the minority carriers.

Both n-type and p-type diodes are commercially available. An n-type diode is formed by doping acceptor impurities into of n-type silicon. A p-type diode is formed by doping donor impurities into substrate. In either case, a spatially doping creates a region where p- and n-type silicon is in direct contact. The majority carriers from each diffuse to the opposite side, i.e., electrons on the n side diffuse to the p side, leaving positively charged donor ions behind, while holes on the p side diffuse to the n side leaving behind negatively charged acceptor ions. These oppositely charged ions establish an electric field (the “built-in potential”) that, at equilibrium, prevents further diffusion of the majority carriers. This spatially charged region is the pn junction, also called the “depletion region”. For diode used for in vivo dosimetry, the typical width of the depletion region is than several microns. Although the typical built-in potential is less than 1 volt, the electric field across the pn junction is very high (greater than $10^3$ V/cm). If the diode were connected to an idealized, leakage-free electrometer, no current would flow unless excess carriers were injected by external, source, such as bias voltage, light, heat, or ionizing radiation (Ellen Yorke 2005).

Charge collection in a semiconductor diode is very different than in an ionization chamber. While an ionization chamber requires a high voltage supply, the high electric field across the pn junction makes charge collection possible for the diode without external bias. As schematically illustrated in figure 1, the incident ionizing radiation generates electron-hole throughout the
diode. The minority carriers (electron on the p side and holes on the n side) diffuse towards the pn junction. Those carriers within approximately one diffusion length from the junction edge are able to reach it before they recombine. They are then swept across the junction by the built-in potential and measured by the electrometer. The total current consist of the radiation-induced photocurrent (below, called “radiation current”) and the electrical leakage current due to the offset voltage from the electrometer.

The processes that determine how many of the mobile of the charge generated radiations are collected are also very different from those in an ionization chamber. Direct recombination, which dominates in an ionization chamber, is highly improbable in silicon. The dominates mode in a silicon diode is direct recombination. This is a function of material defect, which facilitates recombination, and also of the density of radiation-generated electrons and holes. Indirect recombination determines the lifetime of radiation-generated carriers and thus fraction of carriers that diffuses to the pn junction and collected. Thus, the carrier lifetime controls the diode sensitivity (the charge collected per unit dose to the diode). Exposure to large (>kGy) dose from a high energy beam (>2MeV) produce radiation damage defects which shorten the minority carriers lifetime and reduce the diode’s sensitivity. Indirect recombination is responsible for sensitivity change with instantaneous dose rate (for linear accelerators, dose per pulse), which is a major cause of the variation of the diode sensitivity with SSD. The magnitude of these sensitivity changes depends upon the material characteristics of the diode. (e.g., n- or p-type, doping levels (resistivity), and pre-damage of the material) (Ellen yorke 2005).
1.11 FEATURES OF IN-VIVO DOSIMETRY

The aim of radiotherapy treatment is to deliver an accurate dose to the target volume. In order to achieve this goal, great care must be taken at every step of prescription, preparation, calculation and delivery of the daily treatment. In order to limit the errors arising during the treatment course, some international organizations such as World Health Organization (WHO), International commission Radiation Unit (ICRU), American Association of Physicist in Medicine (AAPM), have recommended the implementation of quality assurance programmes which consist of making use of the necessary means to attain the aim designed to ensure achievement of the intended result (Noel et al. 1995). There may be many steps involved in the dose delivery process in radiotherapy. An overall check of the whole procedure is therefore recommended and can be performed only by means of in vivo dosimetry.

In vivo or patient dosimetry is used as quality assurance tool to measure the radiation dose to patients during radiotherapy. The assessment of the final uncertainty between the prescribed dose and dose actually delivered to the patient is an effective way of checking the entire dosimetric procedure. The only real link possible between treatment planning and dose delivery to the patient is in vivo dosimetry. However, in vivo dosimetry can be a very time and labor intensive process. This aspect of radiation therapy QC for the most part, remains neglected. Several methods are currently available for in vivo radiation dose measurements, e.g. Thermoluminescent dosimeters and semiconductor diodes. TLDs have been the more commonly used method of measuring the radiation dose, but they are very labor intensive and impractical for the use of every patient. Generally, just a few of the uncommon cases receive in vivo dosimetry in this process. Furthermore TLD is not a real time dosimeter. Silicon detectors, in contrast provide a more convenient way of measuring the patient dose, in real time, due to instantaneous response. The initial calibrations against a standard such as a calibrated ionization chamber.
are tedious, but the silicon detectors require relatively little additional effort thereafter and are easy to use for therapist. (Fontenla et al 1996).

1.12 SCOPE OF THE PRESENT INVESTIGATIONS

In vivo dosimetry is considered to be important part of quality assurance in advanced radiation therapy modalities. MOSFET dosimeter has been considered for measurement of in vivo dose. However, its use still under primitive stage as it needs more characterization under various physical and treatment conditions. Based on these,

The present work investigates the following objectives:

1. Study and comparison of calibration factor for MOSFET at various energies.
2. In vivo estimation and comparison of entrance dose using MOSFET and Diode dosimeters.
3. Study of dose perturbation and dependency of temperature on in vivo dosimetry using MOSFET.
4. Study of depth dependency of off-axis and peripheral dose using different detectors.
5. Comparison of the response of MOSFET with IC and film.
CHAPTER 2

MATERIALS AND METHODS

The present thesis work involves the dosimetric characteristic of MOSFET, Entrance dose calibration of MOSFET, measuring the various dosimetric parameters of and comparing the same with other detectors. The following are the instruments used in the present work

2.1 MOSFET DOSIMETER

The radiation dosimetry using Metal Oxide Semiconductor Field Effect Transistor (MOSFET) was first proposed by Holmes-Siedle as an space charge dosimeter.

Identical TN-502RD standard MOSFET detectors (Thomson Neilson electronics limited, Ottawa, Canada) were used for all the measurements. This dual bias dual MOSFET is designed for the dosimetry of photons, electrons, protons and x-rays of different energies used in both diagnostic and therapeutic applications. The detector has an active area of less than 0.04 mm$^2$ and generally designed for the absorbed doses greater than 1 Gy. The dosimeters are encapsulated by approximately 2 mm of black epoxy.

The dual bias dual MOSFET was connected to the bias supply (model TN-RD-22), which can accommodate up to 5 dosimeters was set to standard nominal sensitivity of 1 mV/cGy. The bias supply box has two different optional sensitivity modes, called standard and high sensitivity
modes. These two sensitivity modes (1 mV/cGy or 3 mV/cGy) can operate a
dual MOSFET at different dual bias voltages. Choosing a different sensitivity
mode results in a different recombination effects in the MOSFET during
irradiation. Hence, the same MOSFET can be operated at two different
sensitivities (Jornet et al 2004). In all the measurements, the orientation of
MOSFET was such that flat surface faces the beam. A custom built reader for
the MOSFET dosimeter (model TN-RD-10) was used to measure threshold
voltage difference before and after irradiation. Table 2.1 (a) and (b) shows the
nominal sensitivity of standard MOSFET dosimeter and high sensitivity
dosimeter.

2.1.1 Structure

MOSFET is an sandwich type device consisting of a P-type silicon
semiconductor substrate separated from a metal gate by an insulating oxide
layer.

Figure 2.1 shows the block diagram of TN502 RD MOSFET. The
typical size of the standard MOSFET (TN502 RD) used is
3 mm x 3 mm x 0.5 mm with an active volume of 0.2 mm x 0.2 mm. The
micro MOSFET which is a new design MOSFET is constructed with reduced
anisotropy and added tissue equivalent impurities (Ramani et al 2004). The
physical size of the micro MOSFET is half the size of the standard MOSFET.
The new micro MOSFET has small physical size of 1 mm x 1 mm and 3.5
mm long. The active area is 0.2 mm x 0.2 mm and it was about 0.5 μm thick
(200 μm width and 12 μm length). The MOSFET dosimeter mounted on the
laminated polyamide cable under a 1 mm layer of black epoxy.
2.1.2 Working Principle

The type used is a P-MOS type. The working of MOSFET dosimeter is based on the generation of electron-hole pairs in the oxide of MOSFET structure due to ionizing radiation and the energy production of one electron-hole pair in silicon oxide is about 18eV. The positive charges move in the direction of Si/SiO$_2$ interface, where they are captured on traps and hence create a positive build up charge $Q_T$. The positive charge sheet formed effectively changes the current in the channel of MOSFET and leads to the corresponding change of the gate bias voltage $\Delta V_{th}$ (i.e. a shift in the threshold voltage) to ensure a constant current flow through the channel. The current in the channel is very sensitive to the charge $Q_T$ as it is physically located very close to the channel. Thus, the shift in the threshold voltage is a measure of an absorbed dose in the gate oxide. The amount of absorbed dose is directionally proportional to the change in the threshold voltage before and after irradiation. Figure 2.2 (a) shows the schematic diagram of MOSFET and (b) schematic diagram of dual bias dual MOSFET.
The MOSFET sensor should be connected to the bias supply which is coupled to the reader. The dual bias dual MOSFET can be connected to the bias supply (model TN-RD-22), which can accommodate up to 5 dosimeters. The bias supply box has two different optional sensitivity modes, called standard and high sensitivity modes. These two sensitivity modes (1 mV/cGy or 3 mV/cGy) can operate a dual MOSFET at different dual bias voltages. Choosing a different sensitivity mode results a different recombination effects.
at the MOSFET during irradiation. Hence, the same MOSFET can be operated at two different sensitivities (Jornet et al 2004).

As per the manufacturer’s recommendations, the new MOSFET should be connected to the bias supply 1 hour before use and MOSFET sensor should be connected to the bias supply at all times in order to achieve stable readings. A custom built reader (model TN-RD-10) for the MOSFET dosimeter was used to measure threshold voltage difference before and after irradiation. The reader provides the MOSFET response in mV, cGy or in R by entering a calibration factor for each MOSFET (mV cGy$^{-1}$).

The MOSFET has two different states described as ‘bias’ and ‘read’ states during its operation. A positive bias of a few volts is applied between gate and the source terminals during ‘bias’ state. This positive bias is used to enhance electron collection at the gate and to avoid their recombination with the holes created by the radiation. The accumulation of holes at the gate-substrate interface, creating a change in the interfacial change of the MOSFET. After irradiation and during the ‘read’ state, a negative bias is applied between the gate and source terminals. Once a current of few μA from source to drain has been established by the current source, the corresponding gate voltage with reference to the source is measured and designated as the threshold voltage ($V_{TH}$) before and after irradiation.

2.1.3 Orientation of the MOSFET Dosimeter

The size of a commercially available MOSFET dosimeter size is less than 3 mm x 3 mm x 0.5 mm. It is encapsulated under approximately 1 mm of epoxy. So, it is perfectly flat on one side and epoxy coating placed on the other side. The dosimeter with the flat side of the tip in contact with the skin as the “flat sided dosimeter” and the dosimeter with the round side of the
tip in contact with the skin as the “round sided dosimeter”. The manufacturer recommends the “round sided dosimeter”. Figure 2.3 shows the cross sectional diagram of MOSFET.

When the round side of the tip is in contact with the skin, the epoxy coating acts as a thermal insulator and reduces the response of the MOSFET due to the influence of the body temperature on MOSFET. But, if the epoxy coating is removed i.e. unencapsulated MOSFET may be suitable for phantom surface dosimetry. However, the unencapsulated MOSFET may cause several disadvantages that include:

(i) The unencapsulated MOSFET sensors are not reliable for clinical use, since there is a possibility that this may cause skin injury if the device is connected to a bias supply (Paolo Scalchi et al 2005).

(ii) An uncovered sensor could be too fragile for applications in routine patient dosimetry (Scalchi et al 2005).
(iii) Short circuits can occur on the bare MOSFET in contact with the skin, when current is passed through the source and drain to measure the threshold voltage (Scalchi et al 2005).

(iv) Reducing covering material on the top of these dosimeters increase the probability of sensor cracks or failure due to handling, as well as patient movements (Scalchi et al 2005).

The water equivalent depth of the encapsulated MOSFET dosimeter at the flat side is 0.8 mm and 1.8 mm for the round side that was 35.6% and 56.1% instead of 16% in relation to Dmax.

Table 2.1a MOSFET Dosimeter Sensitivities – Standard Dosimeter (Model TN-502RD)

<table>
<thead>
<tr>
<th>Dosimeter</th>
<th>Bias Supply Setting</th>
<th>Nominal Sensitivity</th>
<th>Energy Source</th>
</tr>
</thead>
<tbody>
<tr>
<td>TN502</td>
<td>Standard</td>
<td>1 mV/cGy</td>
<td>Co$^{60}$ with buildup</td>
</tr>
<tr>
<td>TN502</td>
<td>High</td>
<td>3 mV/cGy</td>
<td>Co$^{60}$ with buildup</td>
</tr>
<tr>
<td>TN502</td>
<td>Standard</td>
<td>3 mV/R</td>
<td>Diagnostic X-ray</td>
</tr>
<tr>
<td>TN502</td>
<td>High</td>
<td>9 mV/R</td>
<td>Diagnostic X-ray</td>
</tr>
</tbody>
</table>

Table 2.1b MOSFET Dosimeter Sensitivities High Sensitivity Dosimeter (Model TN-1002RD)

<table>
<thead>
<tr>
<th>Dosimeter</th>
<th>Bias Supply Setting</th>
<th>Nominal Sensitivity</th>
<th>Energy Source</th>
</tr>
</thead>
<tbody>
<tr>
<td>TN1002</td>
<td>Standard</td>
<td>3 mV/cGy</td>
<td>Co$^{60}$ with buildup</td>
</tr>
<tr>
<td>TN1002</td>
<td>High</td>
<td>9 mV/cGy</td>
<td>Co$^{60}$ with buildup</td>
</tr>
<tr>
<td>TN1002</td>
<td>Standard</td>
<td>10 mV/R</td>
<td>Diagnostic X-ray</td>
</tr>
<tr>
<td>TN1002</td>
<td>High</td>
<td>30 mV/R</td>
<td>Diagnostic X-ray</td>
</tr>
</tbody>
</table>
2.1.4 Reader (TN-RD-10)

The model number of the reader used was TN-RD-10. The function of the reader is controlled by the “START” and “READ” switches to measure the dosimeter before and after irradiation and display the data. It also stores the user entered calibration factor, maintains a date and time clock for quality control records, provides power and data distribution. The reader will support up to four separate bias supplies at a time. These can be configured as A, B, C and D if more than one bias supply was to be used at a time.

2.1.5 Bias Supply

The bias supply provides a regulated bias voltage to the dosimeters as well as the interface from dosimeters to reader. Connection to the reader is necessary during “Starting” and “Reading” dosimeters. The bias supply provides two choices of sensitivity to cover a wide range of doses with optimum reproducibility. Each bias supply has its own “Signature”, recognized by the reader. The reader can support up to four independent bias supplies at a time. These are electronically labelled A, B, C, and D via setting the switches.

2.1.6 Dose Reporter Software

TABULA Dose Reporter Software provides a user friendly graphic interface to the MOSFET 20 Reader. It incorporates, displays, records the messages sent by the reader when connected to computer and records the patients’ treatment information up to 4 patients can be handled at a time. It graphically indicates the dosimeters’ positions on patients, records/calculates the measurement results and calculates deviations from target doses, sets/Modifies calibration factors (CFs) and Correction Factors(CRs), provides
password protection for system settings and calibration parameters, supports the use of custom images (such as scanned photos or digital camera photos) and views/prints/saves/opens the measurement reports.

2.1.7 Overview of the Program

This functions of the program is divided into five groups:

- System Setup
- Pre-Irradiation (Step 1)
- Treatment Information (Step 2)
- Measuring Dose (Step 3)
- Viewing/Printing Reports (Step 4)

Five display/control screens let users access these five function groups. They are straightforward and easy to use. Every screen except the last one has a yellow text box to show On-Screen Prompt. The Figure 2.4 is an overview of these screens.

Figure 2.4 Screenshot showing to setup the MOSFET system
2.1.8 Setting up the System

Designate the communication port; set up the title of measurement reports; set or change the password and determine its protection scope; input the lists of available radiation machines and MOSFET 20 readers.

2.1.8.1 Step 1 Pre-Irradiation

In step one Calibration Factors (Cfs) and Correction Factors (CRs) are modified, dosimeters was checked, system settings is modified and existing reports is viewed. If parameters need to be changed or a MOSFET needs to be replaced, this is an opportunity to verify settings before proceeding as shown in Figure 2.5.

![Step 1. Pre-Irradiation](image)

Figure 2.5 Screenshot showing the option before irradiation

2.1.8.2 Step 2 Treatment Info

In this step, patients data is entered and assign dosimeters and assigning of dosimeters is done as shown in Figure 2.6.
Figure 2.6 Screenshot of treatment information for patients

Figure 2.7 is the example which shows how to assign the dosimeter sites graphically.

Figure 2.7  Screenshot of image-Selection Window
2.1.8.3 Step 3 Measuring Dose

In this step, after irradiation, click “Record” on-screen to record data from the MOSFET 20 Reader. Dose data will be recorded and displayed on-screen (if the Reader’s output is set to “mV”) and deviation from Target Dose will be calculated as shown in Figure 2.8.

Figure 2.8 Screenshot of recording the measured dose

2.1.8.4 Step 4 Viewing/Printing Reports

This is the last step in which the measurement procedure, view/print/save reports, and open existing reports and type in comments on reports is done as shown in Figure 2.9.
2.2 IONIZATION CHAMBER

The chamber used was 0.6cm³ farmer type chamber of model FC-65-G. It is the standard reference detector for reference dosimetry and scientific application, manufactured by scanditronix, Sweden. It is a waterproof detector vented with waterproof sleeve fully guarded with a wall material of graphite connected to a electrometer model Dose 1 type capintec WK-92. Figure 2.10 shows the diagram of FC-65G IC.

Figure 2.9 Screenshot showing to print reports

Figure 2.10 Diagram of FC-65 ionization chamber
2.2.1 Electrometer

The electrometer used was DOSE 1 which was intuitive easy to use, soft keys and pop-up menus. It is having a large, high contrast graphic electro luminescent display with a wide viewing angle for complete and comprehensive display of all measured values, chamber and correction factors on the main screen - dose, dose rate, average rate, charge, current and dose per monitor unit are all measured and displayed simultaneously. Figure 2.11 shows the diagram of electrometer. It can be used with ionization chambers, semiconductor detectors and diamond probes - storage of up to 40 detector specific data sets in a sensor library, including physical and geometrical parameters – built-in electrical check source as well as leakage and bias voltage testing to ensure the maximum reliability and accuracy.

Figure 2.11 Diagram of Electrometer
2.3 FILM

2.3.1 Extended Dose Range (EDR-2)

The film used was EDR-2 (Kodak Extended dose range-2, Eastman Kodak Co., Rochester, NY) film, previously referred to as EC or EC-L film. EDR-2 film is coated with an emulsion containing very fine cubic microcrystal grains, highly uniform in both size and shape and about one tenth the grain size of XV film. These physical characteristics of EDR-2 film result in high contrast and wide dose range. Since the saturation doses for EDR-2 are approximately 700 cGy and 200 cGy respectively, it is better suited for the verification of dose distributions during patient treatment. It has the smaller dependence on depth, field size and energy. The use of EDR-2 avoids the necessity for rescaling the dose per beam to avoid the saturation of the film, thus enabling delivery of the total prescribed dose.

2.3.2 MD-55-2 Gafchromic Film

Radiochromic film used as a GAFCHROMIC film MD-55 is manufactured by ISP technologies by Wayne. Radiochromic film consists of double layer radio chromic sensors dispersion coated on both sides of a polyester base. It is a colorless transparent film responds to ultra violet light by turning blue color. It has wavelength Maximum (~610-670 nm) the radiation induced the color change is formed without the thermal optical or chemical development and the original blue image is stable at temperatures up to (60º degree Celsius). The color of the image changes from blue to red above this temperature. Gafchromic films are read at a wavelength of 670 nm since its maximum wavelength is ~610-670 nm. Its response increases up to 5% response when the temperature is increased from 20ºC to 40ºC. at higher range between (40ºC and 50ºC) and the temperature increases rise due to rise in irradiation in irradiation temperature ~1.6ºC %. The γ ray and electron
response shows negligible dose rate independence. When the films are irradiated with photons the dose rate between 1 and 30 Gy/min. in electron beams at an average dose rate ~10^6 Gy/min. High dose levels can be measured by spectrophotometer wavelength. If we use DM-1260 film it shows very high spatial resolution images. For ionizing photons and electrons in the energy region 0.1-10 MeV. However in experimental studies dependence of the film response on the photon energy shows that in terms of absorbed dose to water. And it response to photons is 0.03-0.04 MeV. Range is about 60% of photon energies >0.1 MeV.

The various types of Radiochromic films are

1. HD 810.
2. DM 1260.
3. DM 100.
4. MD 55.
5. NMD 55.

The sensitive structures of the Gafchromic film differ with each other.

1. Gafchromic film HD 810 film sheets was only available in 12.5 cm x 15 cm. This film can be purchased from the ISP available film clear sheets in 20cm x 25cm.
2. Double layer Gafchromic film MD-55 are currently available clear sheets 12.5cm x 12.5 cm. Suppose if we use a single layer Gafchromic film, it is very less sensitive which designated as MD-55 in model no (37-041).
2.3.3 Dose Range for MD-55-2 and HD 810 Films

Radiochromic MD-55-2 films are suitable for a measurement in the dose range a 3-100 cGy. Whereas in HD 810 films dose mapping from 50-2500 Gy.

2.3.4 Radio Chromic Densitometer Model 37-443

The Radiochromic densitometer is a highly accurate reliable and rugged test tool. That allows precise and repeatable measurements to be made quickly and easily. This model incorporates a narrow band pass filter. The filter matches the principle peak of the dose absorption spectrum of our Gafchromic dosimetry. The maximum sensitivity and calibration stability.

The two dimensional film transport system is supplied with the radiochromic densitometer. It’s like a micrometer design which provides a precise method of holding and moving a piece of Gafchromic dosimetry make over the aperture of the densitometer in both x and y dimensions. The micrometer movement and the devices vernier scale provides an x -y axis precision of +/- 0.1 mm.

2.3.4.1 Specifications for radiochromic densitometer

The technical specifications of Radiochromic densitometer is tabulated in Table 2.2
Table 2.2 Specification of radiochromic densitometer

<table>
<thead>
<tr>
<th>Specification</th>
<th>Details</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dose range</td>
<td>0-400</td>
</tr>
<tr>
<td>Accuracy</td>
<td>±0.02 over specified range</td>
</tr>
<tr>
<td>Reproducibility</td>
<td>±0.01 D</td>
</tr>
</tbody>
</table>
| Operating Temperature Range | 50 degree F to 104 degree F  
                          | (10 degree C to 40 degree C)               |
| Aperture            | 2 mm                                         |
| Measuring length (throat) | 5.3 in (135 mm)                             |
| Display             | 3-digit 5 inch LCD with polarity and low battery indicator. |
| Zero range          | Auto zeros to density 0.0  
                          | (for densities up to 1.00)                 |
| Sensor              | High efficiency Silicon photo diode          |

2.3.4.2 Specification for the film transport system

The specification for the film transport system is its range of film movement 6 cm x 3 cm, Precision is about ±0.1 mm and the dimension of the transport system is 57/8” x 10.7/8” L x ½” thick.

2.3.4.3 Optimized for use the Gafchromic dosimetry data

It is designed to measure the optical density of self developing GAFCHROMIC Dosimetry data.

Optimized to measure the principle (671 mm) absorption spectrum peak of Gafchromic dosimetry media.

High precision instant read out easy to use.
In features a high quality 2D film transport mechanism with ± 0.1 mm precision and it provides three times the sensitivity of HE-NE laser densitometer. Table 2.3 shows the dimensions of HD 810, MD-55 and MD-55-2 radiochromic films.

Table 2.3 Dimensions of Radiochromic Films

<table>
<thead>
<tr>
<th>Film type</th>
<th>Nuclear Associates model no (ISP)</th>
<th>HD 810 (DM 1260)</th>
<th>MD-55-1</th>
<th>MD-55-2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Model no</td>
<td>-----</td>
<td>37-041</td>
<td></td>
<td>37-041</td>
</tr>
<tr>
<td>Standard Size film</td>
<td>20x20 cm²</td>
<td>12.5x12.5 cm²</td>
<td>12.5x12.5 cm²</td>
<td></td>
</tr>
<tr>
<td>Sensitive layer (µm)</td>
<td>7±1</td>
<td>15±1</td>
<td>30±1</td>
<td></td>
</tr>
<tr>
<td>Base material (µm)</td>
<td>99</td>
<td>67</td>
<td>159</td>
<td></td>
</tr>
<tr>
<td>Adhesive layer (µm)</td>
<td>1.5</td>
<td>-----</td>
<td>-----</td>
<td></td>
</tr>
<tr>
<td>Polyester base (µm)</td>
<td>99</td>
<td>67</td>
<td>25</td>
<td></td>
</tr>
<tr>
<td>Conductive layer (µm)</td>
<td>0.05</td>
<td>-----</td>
<td>-----</td>
<td></td>
</tr>
<tr>
<td>Pressure sensitive Adhesive (µm)</td>
<td>-----</td>
<td>-----</td>
<td>44.5</td>
<td></td>
</tr>
<tr>
<td>Nominal thickness (µm)</td>
<td>107</td>
<td>82</td>
<td>278</td>
<td></td>
</tr>
</tbody>
</table>

2.3.5 Radiochromic Film Calibration and Sensitivity

Radiochromic film should be calibrated using a large well characterized uniform radiation field. The film should be placed on the central position. Of a large photon beam such as 40 x 40 cm² at a depth ≥ 5 cm. The characteristic of the calibration beam should be determined by some other dosimeter (such as ionization chamber). This would allow direct film calibration in terms of absolute dose within the dose range of interest. The relationship between absorbed dose and the film response should be determined. The slope of the calibration curve decreases as the dose increases.
The calibration curve can provide information for the conversion of film response dose and vice versa. The change in film response per unit absorbed dose can be represented by a single number for a net optical density up to 1.0.

2.3.6 Radiochromic Films used in Medical Applications

The Radiochromic films are relative in sensitivity to radiation compared to commonly available detectors used in medical applications. There will be a lack of sensitivity makes them ideal for dosimetry. Where high doses are utilized. Another important characteristic of radiochromic film is that have an elemental composition dose to that of water which reduces their sensitivity to photon energy for applications dealing with determination of dose delivered to water. Finally the radiochromic films are available in the form of two dimensional dosimeter. The dosimeter is needed to be calibrated to use in some medical applications.

Radiochromic film used in medical applications:

- Ophthalmic applicator dosimetry.
- Brachytherapy dosimetry.
- Interface dosimetry.
- Stereo static radio surgery dosimetry.
- Dosimetry in the penumbra region of radio surgery beams.
- Hot particle dosimetry.
- Photon dosimetry.

2.4 DIODE

The Scanditronix Medical DPD-12pc, Direct Patient Dosimeter is designed to monitor radiation dosage to a patient during treatment. The device
provides multi-channel dose or dose rate measurement utilizing up to 12 semiconductor detectors. The Scanditronix Medical DPD-12 pc, Direct Patient Dosimeter, consists of

- emX, a 12 – channel electrometer
- Semiconductor detectors
- PC with display software – DPDpc

Semiconductor detector detects the radiation and multi channel electrometer, collects the data and transfers to PC. Based on the inherent build up, the diodes are calibrated either for Photon or electron dose measurements. Figure 2.8(a) and (b) shows the diagram of EDP diodes and DPD12pc system respectively

Figure 2.12 (a) Photograph of EDP diode 2.8(b) Diagram of DPD-12 pc system
2.4.1 DPD12 pc In vivo Dosimetry System Specification

2.4.1.1 Versatility

With 12 channels, the system is ideal for a wide variety of applications such as 'entrance' and 'exit' dose measurements, 'risk organ' monitoring, Total Body Irradiation (TBI) in external therapy and intracavitary dose measurements in brachytherapy.

2.4.2 Advantages of DPD pc

The advantages of the DPD pc are

- It has a user-friendly software.
- Measuring mode can be selected with one mouse click.
- (Dose, dose rate, difference or ratio)
- Easy access buttons for frequently used functions.
- (Save/print, alarm on/off, reset electrometer etc.,)
- Quick toggle between display of active channels or all channels.
- Individual alarm levels for each channel.
- Unlimited number of correction sets and alarm sets.
- Password protected calibration data.
- Separate handling of corrections and sensitivity calibration.
- Wizard for calibration process:
  a) Use MU or reference detector as dose reference.
  b) Calibrate one or several detectors at the same time.
  c) Unlimited numbers of cycles for mean value calculation.
The DPD12pc software works with windows 95/98 or NT operating system, also supporting a palm top PC with touch screen for a reduced footprint.

2.4.3 Technical Specification of DPD-12 Electrometer

The diodes should be connected to a dedicated electrometer with a low input impedance and low offset voltage. Diode current generated by sources other than radiation is considered to be leakage current and is not desirable. The leakage current ideally should be zero. Due to the input offset voltage of the amplifier, however there is always a small bias across the diode introducing a small leakage current. An electrometer used together with a diode requires the offset voltage of the amplifier to be low in the range of $10^{-6}$V or less. The leakage current increases with temperature and accumulated dose due to the defects in the diode. It is essential that the electrometer has adequate zero drift compensation and stabilization. Table 2.4 shows the technical specification of the electrometer.

2.4.4 Safety Recommendations for Electrometer

The DPD-12pc system should be kept out of direct sunlight, high or low temperature, chemicals, high humidity, high pressure and power shocks.

2.4.5 Radiation Stress

The electrometer - emX, along with extension cables and cable support should be kept out of the primary beam.
Table 2.4 Technical specification of electrometer

<table>
<thead>
<tr>
<th>Input channels</th>
<th>12</th>
</tr>
</thead>
<tbody>
<tr>
<td>Input sensitivity in dose mode</td>
<td>10 - 70 nC/Gy</td>
</tr>
<tr>
<td>Input sensitivity in dose rate mode</td>
<td>0.16 - 1.2 nA/Gy/min</td>
</tr>
<tr>
<td>Range for dose rate measurements with 2% resolution</td>
<td>0.7 - 20 Gy/min</td>
</tr>
<tr>
<td>With 5% resolution</td>
<td>0.1 - 0.7 Gy/min</td>
</tr>
<tr>
<td>Range* for dose measurements with 0.005% resolution</td>
<td>0.3 – 14 Gy</td>
</tr>
<tr>
<td>Time constant for dose rate measurements</td>
<td>Typically averaged over 5 sec</td>
</tr>
<tr>
<td>Accuracy</td>
<td>1% (in specified range)</td>
</tr>
<tr>
<td>Leakage current</td>
<td>&lt;0.04 pA (zero drift compensated by software)</td>
</tr>
<tr>
<td>Warm-up time</td>
<td>20 min</td>
</tr>
<tr>
<td>Operating temperature</td>
<td>18 ºC to 27 ºC (64 F to 80 F)</td>
</tr>
<tr>
<td>Relative humidity</td>
<td>30 - 70%, non-condensing.</td>
</tr>
<tr>
<td>Interface</td>
<td>RS232</td>
</tr>
</tbody>
</table>

* The range is dependent on the detector sensitivity and the selected operating range.

2.4.6 In vivo Diodes

Many types of diodes are commonly available having various properties with regard to pre irradiation level, doping type, design and thickness of build up cap to accommodate a large photon energy range. For accurate in-vivo dosimetry it is essential that each diode characteristics be
well understood in order to utilize it properly and efficiently\(^2\). For instance in high energy beams the diode reading is nearly independent of photon scatter. In addition, the diode may experience a different amount of head scatter electrons due to physical conditions.

2.4.6.1 Typical specifications of in vivo diode detectors

The technical specifications of DPD12pc detectors are tabulated in Table 2.5.

Table 2.5 Specification of in vivo detectors

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resistivity</td>
<td>0.2 ohm cm.</td>
</tr>
<tr>
<td>Depletion layer width</td>
<td>5µm</td>
</tr>
<tr>
<td>Substrate thickness</td>
<td>0.5mm</td>
</tr>
<tr>
<td>Detector position within capsule</td>
<td>centered within 0.5mm</td>
</tr>
<tr>
<td>Pre irradiation level</td>
<td>8kcGy/10 MeV electrons.</td>
</tr>
<tr>
<td>Effective thickness of sensitive volume</td>
<td>60µm typical.</td>
</tr>
<tr>
<td>Effective detection area</td>
<td>1.5mm</td>
</tr>
<tr>
<td>Impedance at zero bias</td>
<td>300 mega ohms typical</td>
</tr>
<tr>
<td>Sensitivity</td>
<td>40nC/Gy typical</td>
</tr>
<tr>
<td>SVWT</td>
<td>0.35% per °C</td>
</tr>
<tr>
<td>Collection time</td>
<td>5 µs typical</td>
</tr>
</tbody>
</table>

The calibration of diodes is typically performed by comparison of readings against an ion chamber under standard setup to establish a calibration factor for absorbed dose to water and establishment of a series of correction factors to account for calibration differences when measurements
are performed under various experimental conditions. Typical conditions are energy dependence, temperature, sensitivity, directional dependence and radiation damage.

2.5 WIDE ENERGY HEMISPHERICAL BUILD-UP CAP (WEHSBC)

In order to avoid the initial steep dose gradient in the depth dose curve for radiotherapy beams, the measurement of entrance dose must be carried out near the depth of dose maximum. Figure 2.13 shows the wide energy hemispherical brass build-up cap placed over the MOSFET. It is designed to cover the entire range of photon energies (4-18 MV x-rays) and for the electron beams of 15 to 18 MeV energies. The brass hemisphere is specially grooved for the precise placement of the MOSFET. Lightweight (4g) and small radius (0.635 cm) of brass makes it an ideal for placement of MOSFET on patients. To fix the MOSFET exactly on the groove, circular adhesive patches provided with the build-up caps were used.

Figure 2.13 Wide Energy Hemi-Spherical Brass Build-up Cap (WEHSBC)
2.6 BOLUS

Another build up material used for the measurement was tissue equivalent bolus material which was sliced to 1.5cm and 3cm for 6MV and 15MV photon beams.

2.7 PMMA PHANTOM

Poly methyl metha acrylate phantom is used to calibrate the detectors to measure the entrance dose, to measure the peripheral dose using MOSFET, MD-55, ionization chamber, EDR-2 film and to measure dose perturbation of MOSFET dosimeter. Slabs of 0.5cm thickness and 30x30cm² were used for the studies as shown in Figure 2.14.

Figure 2.14 Photograph of PMMA phantom
2.8 SOLID WATER PHANTOM

The phantom used to characterize the MOSFET dosimeter is a virtual solid water equivalent phantom (Med Tec Inc, Orange City, IA.). Each slab has the dimension of 40x40\(^2\) and 5cm thicknesses as shown in Figure 2.15.

Figure 2.15 Photograph of MED-TEC solid water phantom

2.9 SPHERICAL PHANTOM

Spherical phantom designed for stereotactic application was used as shown in Figure 2.16.

Figure 2.16 Photograph of spherical radiosurgical phantom
2.10 TELECOBALT UNIT (\textsuperscript{60}Co)

Theratron 780C and Alcyon telecobalt units were used for thesis work. A schematic diagram of the same is shown in Figure 2.17.

![Telecobalt unit](image)

Figure 2.17 Photograph of Telecobalt unit

2.11 LINEAR ACCELERATOR (LINAC)

Varian Linear accelerators of different model were used for the study. Varian 2300CD was used for characterizing the MOSFET and for calibrating the MOSFET to measure entrance dose. Varian 2100 were used to measure the peripheral dose, Dose perturbation and accumulated dose. Figure 2.18 shows the schematic diagram of the VARIAN linear accelerator.
Figure 2.18 Schematic diagram of Varian Linear Accelerator