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IN VIVO DOSIMETRY WITH DIODES

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CHAPTER 1

INTRODUCTION

Radiation therapy is one of the primary means for cancer management. Most of the external beam radiotherapy is carried out with photon beams, some with electron beams, and a very small fraction with more exotic particles, such as protons, heavier ions and neutrons. Chemotherapy is also another method and sometimes the combination of both is required for the treatment of cancer. The radiotherapy mainly consists of teletherapy and brachytherapy. Teletherapy mainly applies to high energy photons or electrons from a medical linear accelerator to treat the tumor from different directions, while brachytherapy mainly applies radioactive seeds to treat the tumor. Here only teletherapy is considered. The aim of the radiotherapy is to deliver a high dose to the target while delivering the lowest possible dose to the surrounding healthy structures. Conformal therapy and intensity modulated radiation therapy (IMRT) greatly improve the ability to reach this aim.

Experimental and clinical evidence shows that small changes in the dose of 7% to 15% can reduce local tumor control significantly. So the International Commission on Radiological Units and Measurements (ICRU) recommends that the dose delivered to a tumor should be within 5.0% of the prescribed dose. Each of many steps in the treatment planning and execution will contribute to the overall uncertainty in the dose delivered. Therefore, some Organizations recommend that in vivo dosimetry (i.e. Assess the dose directly in the patient) should be made.
Aim of study:
The basic aim was to briefly present a method for implementing an effective IVD program i.e., a program which would produce the maximum results in the minimum time with the minimum effort.

In this work the response of a commercially available diode dosimetry system was studied for two energy qualities, 6MV and 15 MV. Diodes were calibrated against ionization chambers. Signal stability post-irradiation, intrinsic precision, linearity of response with dose and dose decrease under the diode were studied. For each beam energy the response of the diode relative to the given dose as measured by an ionization chamber was evaluated. Diode was calibrated for every energy to give entrance dose, exit dose and eventually the midline dose. Entrance and exit correction factors for field size, tray, source to skin distance, angle and wedge were determined. It was found that diode response i.e. diode reading per cGy of given dose varies significantly with treatment beam set-up. Finally the effects of dose rate, temperature and accumulated dose on the diode's response were studied.

DOSIMETRY IN RADIOTHERAPY

It has become obvious in radiotherapy that quality assurance programs are essential if the best possible therapeutic results are to be obtained. Although the technical and physical aspects of quality assurance are well documented, no guidelines exist for the verification of the whole radiotherapy process at the individual patient level. Each step involved in the planning or accomplishing of a treatment is subject to a certain degree of uncertainty leading to cumulative discrepancy between prescribed and delivered dose. Because it is not possible to eliminate all possible errors with conventional quality assurance programs, it is increasingly recommended to perform verifications on individual patients to check the whole chain of radiotherapy.

In August 1988, a research project for patient related quality assurance measurements was set up in the radiotherapy department of Leuven. Besides portal imaging for the verification of the target volume, in vivo dosimetry was one of the main study topics. In vivo dosimetry is the measurement of the dose actually delivered to the patient during a treatment session. The breakthrough of in vivo dosimetry occurred at the end of the sixties, when thermoluminescent dosimeters (TLD) became available and more recently when semiconductor detectors were introduced as radiation dosimeters.

For most of the IVD measurements diodes proved to be the dosimeters of choice due to their advantages (real time read-out, high sensitivity, good spatial resolution, simple instrumentation, robustness and air pressure independence). When a discrepancy is found between the actually delivered dose and the expected
dose, the immediate availability of the results of the in vivo dose measurements makes it possible to check the actually applied patient set-up to detect the sources of the error enabling their correction. In vivo dose measurements can consist of entrance dose, exit dose, intracavitary dose, and the determination of the dose delivered to critical organs such as the eyes or the gonads.(15)

In ICRU report 24 it is also specified what in vivo dosimetry might include:

- Entrance dose measurements
- Exit dose measurements
- Transmission measurements
- Intracavitary absorbed dose measurements

Entrance dose measurements serve to check the output and performance of the treatment apparatus as well as the accuracy of the patient set-up. Exit dose measurements serve, in addition, to check the dose calculation algorithm and to determine the influence of shape, size and density variations of the body of the patient on the dose calculation procedure.

IN VIVO DOSIMETRY

A very common use of in vivo dosimetry is the determination of the dose to a point in a patient outside the target volume. The dose to the eye, spinal cord or rectum might be important if the target volume is situated close to these organs. Even if this distance is large, it is sometimes necessary to determine the dose delivered to the patient, e.g., the dose to the abdomen in the case of a pregnant woman.(26) Usually the accuracy for this type of in vivo dose measurements needs not to be very high. Most of in vivo dosimetry measurements are, however, performed to check the dose to the target volume. The philosophy behind these measurements is however, different. Several types of errors can be detected by means of in vivo dosimetry. These errors can be related to the overall accuracy, to equipment performance, to patient set-up or to inaccuracies in the dose calculation algorithm.
CHAPTER 2

DOSIMETERS

Radiation dosimeter is a device instrument or system that measures or evaluates, either directly or indirectly the quantities exposure, kerma, absorbed dose, or their time derivatives (rates) or related quantities of ionizing radiation. A dosimeter along with its meter is referred to as dosimetry system. In order to be useful, radiation dosimeters must exhibit several desirable characteristics. For example, in radiotherapy, the exact knowledge of both the absorbed dose to water at a specified point and its spatial resolution are of importance, as well as the possibility to derive the dose to an organ of interest in the patient. In this context the desirable dosimeter properties will be characterized by accuracy and precision, linearity, dose or dose-rate dependence, energy response, directional dependence and spatial resolution. Obviously, not all dosimeters can satisfy all the characteristics, therefore, the choice of radiation dosimeter and its reader must be made judiciously, taking into account the requirements of the measurement situation, e.g., in radiotherapy ionization chambers are recommended for beam calibrations, and other dosimeters are suitable for the evaluation of the dose distribution or dose verification.

PROPERTIES OF DOSIMETERS

ACCURACY AND PRECISION

In radiotherapy dosimetry the uncertainty associated with the measurement is often expressed in terms of accuracy and precision.

The precision of dosimetry measurements specifies the reproducibility of the measurements under similar conditions and can be estimated from the data obtained in repeated measurements. High precision is associated with a small standard deviation of the distribution of measurement results.

The accuracy of dosimeter measurements is the proximity of their expectation value ‘true value’ of the measured quantity. Results of measurement can not be absolutely accurate and the inaccuracy of the measurement is characterized as “uncertainty”.

The error of the measurement is the difference between the measured value of the quantity and the true value of that quantity. Typically the measurement errors are not shown exactly, but they are estimated in the best possible way and corrections are made for them. (38)

LINEARITY
Ideally, the dosimeter reading should be linearly proportional to the dosimetric quantity. However after a certain dose range a non-linearity sets in. The linearity range and the non-linearity behaviour depend on the type of dosimeter and its physical characteristics. (42) A dosimeter and its reader may both exhibit non-linear characteristics but their combined effect could produce linearity over a wider range.

**DOSE RATE DEPENDENCE**

Ideally the response of a dosimetry system, at two different dose rates should remain constant. In reality dose rate may influence the dosimeter readings and the appropriate corrections are necessary. (41)

**ENERGY DEPENDENCE**

The response of a dosimetry system is generally a function of radiation beam quality (energy). Since the dosimetry systems are calibrated at a specified radiation beam quality and used over a much wider energy range, the variation of the response of a dosimetry system with radiation quality should be corrected for.

Again, ideally the energy response should be flat, i.e., the system calibration should be independent of energy over a certain range of radiation qualities. In reality, the energy correction has to be included for most measurement situations. (44) In radiotherapy, the quantity of interest is the dose to water (or to tissue). As no dosimeter is water or tissue equivalent for all radiation beam qualities, the energy dependence is an important characteristic of a dosimetry system.

**DIRECTIONAL DEPENDENCE**

The variation in response of a dosimeter with the angle of incidence of radiation is known as directional, or angular, dependence of the dosimeter. Dosimeters usually exhibit directional dependence due to their constructional details, physical size, and the energy of the incident radiation. (2)

Directional dependence is important in certain applications, e.g., in in-vivo dosimetry while using semiconductor dosimeters. Therapy dosimeters are generally used in the same geometry as that in which they are calibrated.
READOUT CONVENIENCE

Direct-reading dosimeters are generally more convenient than passive dosimeters, i.e., the one that are read after due processing following the exposure e.g., films, TLDs, etc.

CONVENIENCE OF USE

Ionization chambers are reusable with no or little change in sensitivity.

Semiconductor dosimeters are reusable but with a gradual loss of sensitivity.

Some dosimeters are not reusable at all.

Some dosimeters measure dose distribution in a single exposure (e.g., films, gels).

Some dosimeters are quite rugged (i.e., handling will not influence sensitivity, e.g., ionization chambers, while others are sensitive to handling e.g., TLDs.)

TYPES OF DOSIMETERS

The most commonly used detector types for in vivo dosimetry are diodes and thermoluminescence dosimeters (TLDs). Some other detector types have also been tested for in vivo dosimetry purposes, but are not yet in routine clinical use. For accurate in vivo dosimetry, the characteristics of the employed detectors have to be known and the calibration of the detectors has to be performed frequently. (8)

DIODES

The diodes in use for in vivo dosimetry are silicon detectors. The base material can be n- or p-type silicon. The dependence of the diode on accumulated dose, dose rate, and temperature is related to several characteristics, for example the doping level. P-type diodes resist radiation much better than n-type detectors because (heavy) holes are more easily trapped than electrons. This means that the decrease in radiation is much smaller for p-type than for n-type diodes. After an initial fast sensitivity drop the decrease in sensitivity is almost linear: a value of about 0.1% per 100 Gy for p-type diodes while for n-type diodes a
A decrease of 4.5% per 100 Gy has been reported. The dependence of diode sensitivity on dose rate and temperature is also larger for n-type detectors than for p-type diodes. A number of properties of importance for clinical use of diodes are related to this dose rate dependence of the sensitivity, for example, the variation in response with source-to-skin distance (SSD), field size and the presence of a wedge. (13) These variations in response, due to differences in beam set-up during patient treatment and diode calibration are much larger for n-type diodes than for p-type diodes. Consequently it seems advantageous that diodes in use for in vivo dosimetry are pre-irradiated p-type detectors.

The main advantages of diodes are: high sensitivity to radiation, small size, good mechanical stability, absence of external voltage, and immediate availability of the measured dose. The sensitivity per unit volume of a diode is about 18,000 times higher than for an air-filled ionization chamber. On line dose verification has the advantage that reasons for large deviations can be investigated with the patient still in the treatment position. (19) If necessary the dose can even be adapted during the treatment session, as is done in some institutions during total body irradiation (TBI).

The diodes are usually calibrated in such a way that using the diode positioned at the entrance surface of the beam the dose at the depth of maximum dose, entrance dose is measured. The exit dose, the dose at the same distance upstream form the exit surface, is measured with the diode at the exit surface.

The calibration factor needs to be determined on a regular basis, because radiation damage affects the diode sensitivity. An uncertainty in calibration factor of about 0.5% is acceptable, which means that for p-type diodes, a recalibration will be necessary after about one kGy. Re-calibration has to be performed much more frequently for n-type diodes due to their faster decrease in sensitivity. (21)

Besides the calibration factor, determined under the reference conditions, correction factors have to be applied for accurate dosimetry. They originate from the variation in sensitivity of the diode with dose per pulse, the photon energy spectrum, the temperature, the field size and from directional effects. Most correction factors, correcting for the change in SSD, field size, phantom thickness, or presence of a wedge, are small for p-type diodes (corrections of less than 2%) but are up to 15% for n-type diodes. Without using correction factors large systematic errors in the measured dose can occur. (50) However for several standard treatment techniques, it is possible to use the same overall entrance or exit correction factors for each field direction for each individual patient. A final prerequisite for accurate diode dosimetry is accurate and reproducible diode positioning, especially for fields with large dose gradients, such as for large inhomogeneities and in beams with wedges.

The overall accuracy of the measurements is determined by the combined uncertainty in the calibration factor, the correction factors, and the accuracy and reproducibility in diode position.
**Different types of detectors for in vivo dosimetry**

**TLDs (Thermoluminescence dosimeters)**

TLDs have been used for in vivo dosimetry for a very long time. They are based on the principle that imperfect crystals can absorb and store the energy of ionizing radiation. Due to irradiation, free electrons and holes are formed. The electrons may be trapped at defects, where they can stay for a long period. When heated to a temperature which is typical for the detector material, electrons return to the conduction band and then may recombine with a hole, while emitting energy in the form of electromagnetic radiation. (43) This radiation mainly in the visible wavelength region is detected by a photomultiplier and correlated to the absorbed dose received by the material. After annealing, the TLD can be used again.

TL materials used for in vivo dosimetry are: lithium fluoride(LiF), Lithium Borate (Li₂B₄O₇), and calcium sulphate(CaSO₄). TL detectors can either be powders or solid dosimeters, in the form of rods chips or pellets. Their small dependence on dose rate, temperature, and energy in the therapeutic range, and their wide applicable dose range make TLDs suitable for in vivo dosimetry purposes. TLDs, as well as diodes, are small, and therefore have good spatial resolution. They also have very high sensitivity and again, no bias voltage has to be applied. TLDs have the main advantage against diodes, that they do not have to be connected to an electrometer with a cable and that they are easy to transport. Since, in addition, the dose information can be stored over a long period of time; TLDs are suitable detectors for mailing, which implies they can be used for intercomparison of dose values delivered in different institutions, such as in multicenter trial. (31) They can also be tissue or bone equivalent. Disadvantages are that they can not be used for on-line in vivo dosimetry, they are difficult to handle, and it is difficult to identify an individual dosimeter.

Like diodes, TLDs can be used for entrance and exit dose measurements when a build-up cap is applied. However, TLDs are also useful for entrance dose measurements. TLDs can also be used for intracavitary measurements, for example, during brachytherapy. Then accuracy and reproducibility of carefully calibrated TL dosimeters is about 2%, again provided that the detector is accurately positioned with respect to the patient anatomy and beam modifiers. However, a recent study comparing in vivo dosimetry with diodes and TLDs showed that the reproducibility was better for diodes than for TLDs, because of the complicated TL process and acquisition and annihilation procedure.
OTHER DETECTORS FOR IN VIVO DOSIMETRY

Other detectors such as plastic scintillators, alanine, and diamond detectors, have also been investigated for in vivo dosimetry purposes. *Diamond detectors and plastic scintillators* have the advantages of good stability, a high spatial resolution, nearly water equivalence, and linear response versus dose rate. They are much less energy and dose rate dependent than diodes, show less radiation damage, and are not affected by temperature variations. Alanine dosimeters offer the possibility of measuring the integrated dose during the overall treatment series. However, these detectors require expensive and complicated reading equipment and have therefore only been used under laboratory conditions and not on a routine base in a clinic. Metal oxide semiconductor field effect transistors (MOSFET) have not yet been tested for in vivo dosimetry in external beams, but might also be suitable for this purpose. Temperature and signal drift corrections are necessary for MOSFET detectors. Minimal high voltage is required and the detectors have very small size. The above described detectors are all point detectors, and are therefore not very suitable to determine spatial dose distributions.

Using point detectors for the thorax region and other regions with large inhomogeneities, accurate exit dosimetry is not possible because of the large dose gradient at the detector position. 2D detectors such as photographic film and EPIDs, are currently investigated in various centres to assess their suitability for in vivo dosimetry. These detectors can be used to determine the exit or midplane dose in the entire field, to check the dose homogeneity throughout the irradiation field, and to determine the dose delivery to organs at risk and the dose under wedges, shielding blocks or other beam modifiers. Further clinical testing of EPIDs for patient dose verification demonstrated their usefulness not only for checking the absolute dose delivery, for example, the monitor unit calculation, but also for evaluation of the deviations between the actual and calculated relative dose distribution caused by the differences between patient anatomy and planning CT data.

➤ IONISATION CHAMBER

Ionisation chambers are used in radiation therapy and in diagnostic radiology for the determination of radiation dose. The dose determination in reference irradiation conditions is also called *beam calibration*. Ionisation chambers have various shapes and sizes depending upon the specific requirements. An ionisation chamber is basically a cavity surrounded by a conductive outer wall and having a central collecting electrode. The wall and the collecting electrode are separated with a high quality insulator to reduce the leakage current when a polarizing voltage is applied to the chamber. A guard electrode is usually provided in the chamber to further reduce the chamber leakage. The guard electrode intercepts the leakage current and allows it to flow to ground bypassing the collecting electrode.
RADIOGRAPHIC FILM

Radiographic x-ray film performs several important functions in diagnostic radiology, radiation therapy, and radiation protection. It can serve as radiation detector, relative dosimeter, a display device, and archival medium. Unexposed x-ray film consists of a base of thin plastic with a radiation sensitive emulsion (silver bromide AgBr grains suspended in gelatin) coated uniformly on one or both sides of the base.(41)

Ionisation of AgBr in the grains, as a result of radiation interaction, forms the latent image in the film. Image becomes visible (film blackening) only after development. Light transmission is a function of the film opacity and can be measured in terms of optical density (OD) with special devices called densitometers. Optical density is defined as \( OD = \log_{10} \left( \frac{I_0}{I} \right) \) and is a function of dose. \( I_0 \) is the initial light intensity and \( I \) is the intensity transmitted through the film.

Film gives excellent 2D spatial resolution and, in a single exposure, it provides information about the spatial distribution of radiation in the area of interest or the attenuation of radiation by objects.

Useful dose range of film is limited; energy dependence is pronounced for lower energy photons, and the response depends on several, difficult to control, parameters. Typically, films are used for qualitative dosimetry but with proper calibration, careful use and analysis, film can also be used for dose evaluation.

Various types of films are available for radiotherapy work (e.g., direct exposure non-screen films for field size verification, phosphor screen films used with simulators, metallic screen films used in portal imaging, etc.). (48)

LUMINESCENCE DOSIMETERS

Some materials, upon absorption of radiation, retain part of the absorbed energy in metastable states. When this energy is subsequently released in the form of ultraviolet, visible or infrared light, the phenomenon is called as luminescence. Two types of luminescence: fluorescence and phosphorescence, are known depending on the time delay between the stimulation and the emission of light. Fluorescence occurs with a time delay between 10-10 to 10-8 seconds; phosphorescence with a time delay exceeding 10-8 seconds.(45)

The process of phosphorescence can be accelerated with a suitable excitation in the form of heat or light. If the exciting agent is heat, the phenomenon is known as thermoluminescence and the material is called a thermoluminescent (TL) material or a thermoluminescent dosimeter (TLD) when used for purposes of dosimetry. If the exciting agent is light, the phenomenon is referred to as optically stimulated luminescence (OSL). Thermoluminescence (TL) is thermally activated phosphorescence; the most spectacular and the most widely known of a number of different ionizing radiation induced thermally activated phenomena. Its practical applications range from archeological pottery dating to radiation dosimetry. In 1968 Cameron, Suntharalingam and Kenney published a book on the TL process that is still considered an excellent treatise on the practical aspects of the TL phenomenon.
TL dosimeters most commonly used in medical applications are LiF:Mg,Ti, LiF:Mg,Cu,P and Li2B4O7:Mn, because of their tissue equivalence. Other TLDs, used because of their high sensitivity, are CaSO4:Dy, Al2O3:C and CaF2:Mn. They are available in various forms (e.g., powder, chips, rods, ribbon, etc.). Before they are used, TLDs have to be annealed to erase the residual signal. Well-established reproducible annealing cycles should be used including the heating and cooling rates. TL dosimeters have to be calibrated before they are used (thus they serve as relative dosimeters). To derive the absorbed dose from the TL-reading a few correction factors have to be applied, such as energy correction, fading and dose-response non-linearity corrections. (49)Typical applications of TLD in radiotherapy are: in vivo dosimetry on patients (either as a routine QA procedure or for dose monitoring in special cases, e.g., complicated geometries, dose to critical organs, total body irradiation, in brachytherapy, etc.), verification of treatment techniques in various phantoms (e.g., Rando phantom), dosimetry audits (such as the IAEA/WHO TLD postal dose audit programme) and comparisons among hospitals.

OSL SYSTEMS

Optically-stimulated luminescence (OSL) is based on a principle similar to that of the TLD. Instead of heat, light (from a laser) is used to release the trapped energy in the form of luminescence. OSL is a novel technique offering a potential for in vivo dosimetry in radiotherapy. The integrated dose measured during irradiation can be evaluated using OSL directly afterwards. (39)

MOSFET DOSIMETER

The Metal-Oxide Semiconductor Field Effect Transistor (MOSFET), a miniature silicon transistor, seems to be a promising candidate for medical dosimetry. MOSFETs are small in size even compared to diodes, offering very little attenuation of the beam when used for in in-vivo dosimetry. They require a special read-out facility. A single dosimeter can cover the full energy range of photons and electrons, although the energy response should be examined, since it varies with radiation quality. Similarly to diodes, MOSFETs exhibit temperature dependence. As they show non-linearity of response with total absorbed dose, regular sensitivity checks are required. MOSFETs are also sensitive to changes in the bias voltage during irradiation (it must be stable) and their response drifts slightly after the irradiation (the reading must be taken in a specified time after exposure). They have a limited life-span. MOSFETs have been in use for the past few years in radiotherapy applications, such as surface dose measurements, radiosurgery, in vivo dosimetry, and brachy-therapy measurements. (32)
SCINTILLATION DOSIMETERS

Over the last decade, there has been an increased interest in scintillation dosimetry using small water-equivalent plastic scintillators, because of their favourable characteristics when compared with other more commonly used detector systems. Although plastic scintillators have been shown to have many desirable dosimetric properties, as yet there is no successful commercial detector system of this type available for routine clinical use in radiation oncology.(26) The main factor preventing this new technology from realizing its full potential in commercial applications is the maximization of signal coupling efficiency and the minimization of noise capture. A principal constituent of noise is Cerenkov radiation.

This study reports the calculated capture of Cerenkov radiation by an optical fibre in the special case where the radiation is generated by a relativistic particle on the fibre axis and the fibre axis is parallel to the Cerenkov cone.

The fraction of radiation captured is calculated as a function of the fibre core refractive index and the refractive index difference between the core and the cladding of the fibre for relativistic particles. This is then used to deduce the relative intensity captured for a range of fibre core refractive indices and fibre core–cladding refractive index differences. It is shown that the core refractive index has little effect on the amount of radiation captured compared to the refractive index difference. The implications of this result for the design of radiation therapy plastic scintillation dosimeters are considered.(51)

DIAMOND DETECTORS

Diamond has an extremely high resilience to radiation - three orders of magnitude higher than silicon - making it an ideal material for detectors that monitor radioactive emissions inside the hostile environments found in nuclear energy plants. But because of the high price of real diamond, synthetic diamond is needed. (54) The latest developments in making and improving this synthetic diamond are presented today by Dr Bergonzo of CEA Saclay (the French Atomic Energy Commissariat) at the Institute of Physics Congress 2002.

Diamond has a lot of enviable properties that make it desirable to reproduce synthetically. As well as being highly resilient to radiation, unlike its semiconductor rivals it hardly reacts with any chemicals, particularly acids. As radioactive waste is reprocessed using the separation of the various constituting elements using acids, diamond radiation monitoring devices would be the perfect detectors.

But the problem with synthetic diamond is that it is a long way from the quality of the real thing, so the researchers are concentrating their efforts on finding uses for this imperfect substitute. Dr Bergonzo and colleagues are currently tailoring and improving the synthetic diamond so that it is ideal for monitoring radiation in the core of a nuclear reactor. Such measurements are usually performed using gaseous ionisation chambers, because solid-state devices cannot withstand the high radioactivity levels. But these chambers are very large making a smaller synthetic diamond device better.
CHEMICAL DOSIMETERS

Chemical dosimeters are systems in which measurable chemical changes are produced by ionizing radiation. Radiation produces acids in the system, the amount of which can be determined from visible color changes, or, more accurately, by titration or pH readings. Most chemical systems of practical size are useful only for gamma doses of hundreds to millions of cGy. The major problem with this method was that the dosimeters exhibited a large dependence on the radiation quality due to heavy metal contents of the chemical compound. Today chemical dosimetry is not used. (51)

MAIN ADVANTAGES AND DISADVANTAGES OF FOUR COMMONLY USED DOSIMETRIC SYSTEMS (45)

<table>
<thead>
<tr>
<th>Dosimeter</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
</table>
| Ionisation chamber | Accurate and precise  
Recommended for beam calibration  
Necessary corrections well understood  
Instant readout | Connecting cables required  
High voltage supply required  
Many corrections required for high energy dosimetry |
| Film            | 2D spatial resolution  
Very thin - does not perturb the beam | Dark room, processing facilities required  
Processing difficult to control  
Variation between films and batches  
Needs proper calibration against ion chamber measurements  
Energy dependence problems  
Cannot be used for beam calibration |
| TLD             | Small in size - point dose measurements possible  
Many TLDs can be exposed in single exposure  
Available in various forms  
Some are reasonably tissue equivalent  
Not expensive | Signal erased during readout  
Easy to lose reading  
No instant readout  
Accurate results require care  
Readout and calibration time consuming  
Not recommended for beam calibration |
### Diode

| Small size | High sensitivity | Instant readout | No external bias voltage | Simple instrumentation | Requires connecting cables | Variability of calibration with temperature | Change in sensitivity with accumulated dose | Special care needed to ensure constancy of response | Cannot be used for beam calibration |

### HISTORY

Solid state materials can be grouped into three classes—insulators, semi-conductors and conductors. Insulators have very low conductivities. Semiconductors have conductivities between those of insulators and conductors. The conductivity of a semiconductor is generally sensitive to temperature, illumination, magnetic field and minute amount of impurity atoms. The sensitivity in conductivity makes the semiconductor one of the most important materials for electronic applications. (31)

The study of semiconductor materials began in the early nineteenth century. Over the years many semiconductors have been investigated. Table 1 shows a portion of the periodic table related to semiconductors. The element semiconductors, those composed of single species of atoms, such as Silicon (Si) and germanium (Ge) can be found in Column IV. However, numerous compound semiconductors are composed of two or more elements. For example gallium arsenide (GaAs) is a III-V compound that is a combination of gallium (Ga) from Column III and arsenic (As) from Column V. (33)

Prior of the invention of the bipolar transistor in 1947 semiconductors were used only as two terminal devices, such as rectifiers and photodiodes. In the early 1950s, germanium was the major semiconductor material. However germanium proved unsuitable in many applications because germanium devices exhibited high leakage currents at only moderately elevated temperatures. In addition germanium oxide is water soluble and unsuited for device fabrication. Since the early 1960s silicon has become a practical substitute and has now virtually supplanted germanium as a material for semiconductor fabrication. (42) The main reasons we now use silicon are that silicon devices exhibit much lower leakage currents, and high quality silicon dioxide can be grown thermally. There is also an economic consideration. Device grade silicon costs much less than any other semiconductor material. Silicon in the form of silica and silicates comprises 25% of the Earth’s crust, and silicon is second only to oxygen in abundance. At present, silicon is one of the most studied elements in the periodic table; and silicon technology is by far the most advanced among all semiconductor technologies.
Many of the compound semiconductors have electrical and optical properties that are absent to silicon. These semiconductors especially gallium arsenide (GaAs), are used mainly for microwave and photonic applications. (52)

**BASIC THEORY OF SEMICONDUCTOR DIODES**

All materials have electrical properties that allow them to be organized into three broad categories: conductors, insulators and semiconductors. Metals (pure elements and alloys) are typically conductors of electricity. Thousands of miles of aluminium and copper wires crisscross the country bringing electricity into our homes and places of work. A relatively small number of nonmetallic substances can also be classified as conductors. Also, a very few ceramic compounds have exhibited the unusual property of superconductivity at the frigid temperature of liquid nitrogen or below. The nonmetallic elements and their compounds fall into the class of electrical insulators. Most ceramics and plastics do not conduct electricity under ordinary circumstances. Plastic coatings are frequently found covering copper wires to protect the user from shock and keep devices from short circuiting. Ceramic knobs are used where electrical wires are attached to utility poles or to the back of a house. (47) The third group of materials, the semiconductors, can be understood from their name, to fall somewhere midway between conductors and insulators.

Although pure elements such as silicon play an important role in many semiconductor devices, it is most often utilized by adding very small but controlled amounts of impurities in order to alter its properties. Silicon-based materials dominate in the semiconductor industry and in electronic devices like computers and calculators, but a number of other compounds are also used extensively— including GaAs (or gallium arsenide) which is the material used in the laser of a CD player. Some other combinations of elements that exhibit semiconductor properties are indicated on the periodic table below (See Figure 1). In the readings and lab activities that follow, the emphasis is on what semiconductor materials are, how they are used, what properties they possess, and why they behave as they do. (43)
apart, as in a gas, they have very little influence upon each other, and are very much like lone atoms. But if a sufficient amount of energy is absorbed by an electron, it is possible for that electron to be completely removed from the influence of the atom. This is called IONIZATION. When an atom loses electrons or gains electrons in this process of electron exchange, it is said to be ionized. For ionization to take place there must be a transfer of energy that results in a change in the internal energy of the atom. An atom having more than its normal amount of electrons acquires a negative charge, and is called a NEGATIVE ION. The atom that gives up some of its normal electrons is left with fewer negative charges than positive charges and is called a POSITIVE ION.(32) Thus, we can define ionization as the process by which an atom loses or gains electrons.
greatly modify the behaviour of the other electrons. One consequence of this close proximity of atoms is to cause the individual energy levels of an atom to break up and form bands of energy. Discrete (separate and complete) energy levels still exist within these energy bands, but there are many more energy levels than there were with the isolated atom. In some cases, energy levels will have disappeared. Figure 1-5 shows the difference in the energy arrangement between an isolated atom and the atom in a solid. Notice that the isolated atom (such as in gas) has energy levels, whereas the atom in a solid has energy levels grouped into ENERGY BANDS.

![Energy Arrangement in Atoms](image)

**Figure 2:** The energy arrangement in atoms

The upper band in the solid lines in figure 1-5 is called the CONDUCTION BAND because electrons in this band are easily removed by the application of external electric fields. Materials that have a large number of electrons in the conduction band act as good conductors of electricity.

Below the conduction band is the FORBIDDEN BAND or energy gap. Electrons are never found in this band, but may travel back and forth through it, provided they do not come to rest in the band.

The last band or VALENCE BAND is composed of a series of energy levels containing valence electrons. Electrons in this band are more tightly bound to the individual atom than the electrons in the conduction band. However, the electrons in the valence band can still be moved to the conduction band with the
application of energy, usually thermal energy. There are more bands below the valence band, but they are not important to the understanding of semiconductor theory and will not be discussed.

The concept of energy bands is particularly important in classifying materials as conductors, semiconductors, and insulators. An electron can exist in either of two energy bands, the conduction band or the valence band. (26)

All that is necessary to move an electron from the valence band to the conduction band so it can be used for electric current, is enough energy to carry the electron through the forbidden band.

The width of the forbidden band or the separation between the conduction and valence bands determines whether a substance is an insulator, semiconductor, or conductor. Figure 1-6 uses energy level diagrams to show the difference between insulators, semiconductors, and conductors.

![Energy Level Diagram](image)

**Figure 3**: Energy level diagram.

The energy diagram for the insulator shows the insulator with a very wide energy gap. The wider this gap, the greater the amount of energy required to move the electron from the valence band to the conduction band. Therefore, an insulator requires a large amount of energy to obtain a small amount of current. The insulator "insulates" because of the wide forbidden band or energy gap. (36)

The semiconductor, on the other hand, has a smaller forbidden band and requires less energy to move an electron from the valence band to the conduction band. Therefore, for a certain amount of applied voltage, more current will flow in the semiconductor than in the insulator.

The last energy level diagram in figure 3 is that of a conductor. Notice, there is no forbidden band or energy gap and the valence and conduction bands overlap. With no energy gap, it takes a small amount of energy to move electrons into the conduction band; consequently, conductors pass electrons very easily. (31)
DOPING OF SEMICONDUCTORS

DOPING PROCESS

The pure semiconductor mentioned earlier is basically neutral. It contains no free electrons in its conduction bands. Even with the application of thermal energy, only a few covalent bonds are broken, yielding a relatively small current flow. A much more efficient method of increasing current flow in semiconductors is by adding very small amounts of selected additives to them, generally no more than a few parts per million. These additives are called impurities and the process of adding them to crystals is referred to as DOPING. The purpose of semiconductor doping is to increase the number of free charges that can be moved by an external applied voltage. When an impurity increases the number of free electrons, the doped semiconductor is NEGATIVE or N TYPE, and the impurity that is added is known as an N-type impurity. However, an impurity that reduces the number of free electrons, causing more holes, creates a POSITIVE or P-TYPE semiconductor, and the impurity that was added to it is known as a P-type impurity.(45) Semiconductors which are doped in this manner - either with N- or P-type impurities - are referred to as EXTRINSIC semiconductors.

N-Type Semiconductor

The N-type impurity loses its extra valence electron easily when added to a semiconductor material, and in so doing, increases the conductivity of the material by contributing a free electron. This type of impurity has 5 valence electrons and is called a PENTAVALENT impurity. Arsenic, antimony, bismuth, and phosphorous are pentavalent impurities. Because these materials give or donate one electron to the doped material, they are also called DONOR impurities.

When a pentavalent (donor) impurity, like arsenic, is added to germanium, it will form covalent bonds with the germanium atoms. Figure 1-10 illustrates this by showing an arsenic atom (AS) in a germanium (GE) lattice structure. Notice the arsenic atom in the center of the lattice. It has 5 valence electrons in its outer shell but uses only 4 of them to form covalent bonds with the germanium atoms, leaving 1 electron relatively free in the crystal structure.(41)

Pure germanium may be converted into an N-type semiconductor by "doping" it with any donor impurity having 5 valence electrons in its outer shell. Since this type of semiconductor (N-type) has a surplus of electrons, the electrons are considered MAJORITY carriers, while the holes, being few in number, are the MINORITY carriers.
P-Type Semiconductor

The second type of impurity, when added to a semiconductor material, tends to compensate for its deficiency of 1 valence electron by acquiring an electron from its neighbour. Impurities of this type have only 3 valence electrons and are called TRIVALENT impurities. Aluminium, indium, gallium, and boron are trivalent impurities. Because these materials accept 1 electron from the doped material, they are also called ACCEPTOR impurities.

A trivalent (acceptor) impurity element can also be used to dope germanium. In this case, the impurity is 1 electron short of the required amount of electrons needed to establish covalent bonds with 4 neighbouring atoms. Thus, in a single covalent bond, there will be only 1 electron instead of 2. This arrangement leaves a hole in that covalent bond. Figure 1-11 illustrates this theory by showing what happens when germanium is doped with an indium (In) atom. Notice, the indium atom in the figure is 1 electron short of the required amount of electrons needed to form covalent bonds with 4 neighbouring atoms and, therefore, creates a hole in the structure. Gallium and boron, which are also trivalent impurities, exhibit these same characteristics when added to germanium. The holes can only be present in this type semiconductor when a trivalent impurity is used. Note that a hole carrier is not created by the removal of an electron from a neutral atom, but is created when a trivalent impurity enters into covalent bonds with a tetravalent (4 valence electrons) crystal structure.

The holes in this type of semiconductor (P-type) are considered the MAJORITY carriers since they are present in the material in the greatest quantity. The electrons, on the other hand, are the MINORITY carriers.
The fifth valence electron of the n-type dopant can easily jump to the conduction band and carry current. In the p-type semiconductor, electrons are easily promoted to the vacant level in the dopant. This creates a hole in the valence band which can carry current by travelling in the opposite direction of electron flow.

**Comparison between n-type and p-type semiconductors**

Among the various solid-state detectors which can be used to measure “in vivo dose” at a point, silicon diodes and thermoluminescent dosimeters (TLD) have featured prominently. Silicon diodes have gained popularity due to their short processing time and the improvement of the dosimetric properties of models commercially available.
During the last fifteen year the use of silicon semiconductor detectors has increased rapidly for relative dose determination and \textit{in vivo} dosimetry. It has also been pointed out in several publications that silicon semiconductor detectors can be used in clinical dosimetry if some precautions are considered.

The main limitation when using silicon semiconductor detectors are some effects caused by radiation damage. Recently Grusell and Rikner showed that detectors based on n-type silicon, when used in pulsed radiation beams, not only suffer a sensitivity decrease, they also become non-linear with respect to the dose rate as a result of the radiation damage. These detectors are, therefore, recommended for clinical use in pulsed radiation beams. (51) Fortunately detectors based on p-type silicon do not show this drawback (Rikner and Grusell 1983). Such p-type silicon detectors also show a much slower sensitivity decrease after irradiation than those of n-type, and they are thus well suited not only for relative measurements in water phantoms but also for patient dosimetry (Rikner 1983).

Another effect is the sensitivity increase with increasing temperature in the detector. This effect is of similar magnitude for n- and p-type silicon detectors.

Based on different researches it was shown that by comparing the different correction factors for both types of diodes (p-type and n-type) no conclusion about which type is better can be driven. On the other hand, it was shown that n-type diodes have the highest sensitivity dependence on dose rate. Moreover p-type diodes showed higher sensitivity dependence with temperature and a greater sensitivity with accumulated dose than n-type diodes. Therefore, whether to choose one or the other will depend on the priorities of the user.

In a recent publication, Van Dam et al (1990) showed that also p-type detectors can show a non-linear response after preirradiation with high energy photons.

For a change of the dose rate of a factor of about 65 they reported a 1-10\% deviation from a linear response, after preirradiation with electrons up to 20 MeV and bremsstrahlung photons produced by an 18 MV linear accelerator, with the highest deviation after preirradiation with photons. Although this is much smaller effect than that observed with n-type detectors, where deviations of 10\% for a dose rate factor of about 6 are typical after preirradiation with 20 MeV electrons, it can be of clinical significance in some situations.(35)

The n-type and p-type detectors behave differently because the minority carriers are holes and electrons respectively. After experiments, p-type detectors resist radiation better than those of n-type.

A detector that does not show a linear relation between signal and dose per pulse will of course show an incorrect depth dose distribution in relative dosimetry.
In in-vivo dosimetry this phenomenon is important even if the detector is calibrated at reference situation, i.e. at a reference temperature and at a nominal SSD. If measurements are performed at other SSDs or generally at other conditions different from the reference ones, the dose per pulse can vary over a wide range. This can lead to a substantial error.

**IMPORTANT TECHNICAL BACKGROUND ABOUT DIODE DETECTOR**

Silicon diode radiation detectors have been in use for many years for the measurement of both photons and electrons used in radiation therapy. These detectors, when coupled with a good electrometer, offer the unique combination of small size, immediate readout, simplicity of operation, no external bias voltage, high accuracy and ruggedness.

Just as ion chamber characteristics are subject to environmental aspects (e.g., temperature, atmospheric pressure, saturation and others), silicon detectors are also responsive to their operating environment. Therefore, as with ion chambers, it is important that diode characteristics are well understood in order to use them properly and most efficiently. (53)

1. **Charge collection process in the diode**

A p-type pn junction diode is formed when n type material is fused into the p type material substance.

In the application of in vivo dosimetry, the electron-hole pairs (e-h) generated by radiation in the diode are measured. The charge collection process is described by the following:

i) primary and secondary particles form the radiation source are absorbed, generating electron-hole pairs throughout the diode

ii) by diffusion, those electrons and holes generated within one diffusion length form the junction will reach the junction

iii) the excess of minority carriers will be swept to the opposite sides by the built-in potential across the pn junction. When the diode terminals are connected to the input of an operational amplifier, the charges generated by the irradiation will be collected. The maximum signal to noise ratio is obtained when the offset voltage of the operational amplifier is less than 1µV. This is the so-called zero–voltage bias application of diodes which is used in in-vivo dosimetry.

Diode detector sensitivity $S$, i.e., charge collected per unit radiation dose is approximately proportional to the minority carrier diffusion length:

$$S = a_L = a\sqrt{D\tau}$$
Where $\alpha$ is a constant and $\tau$ is the lifetime of the radiation generated excess carriers. $L$ and $D$ are the minority carrier diffusion length and the diffusion coefficient respectively.

2. **Instantaneous dose rate dependence (SSD and dose per pulse dependence)**

The excess carrier lifetime $\tau$ depends upon both the accumulated dose and the instantaneous dose rate.

It is well known that direct ion recombination occurs in a gas filled ionization chamber and that this reduces the amount of charge collection as the ion density increases. In semiconductor silicon, however, radiation generated electron-hole pairs recombine indirectly through the energy levels in the energy gap between the conduction and valence band of Silicon. These energy levels are formed by the defects and impurities in silicon. They are called recombination – generation (R-G) centers because they can capture and emit carriers.

The R-G centers act as intermediaries in this indirect recombination process. The empty R-G center first captures an electron (or a hole). This carrier stays trapped until the R-G center captures an opposite carrier hole (or an electron), resulting in recombination.

During radiation, most of the generated electron-hole pairs in the diode are swept across the junction and contribute to the signal. However, a portion of them will be trapped in the R-G centers, where they eventually recombine. The recombination portion is proportional to the number of the carriers to be captured and the number of empty R-G centers. (43)

When the generated carrier concentration is small, most of the R-G centers are available and all the recombination portion can be promptly recombined. As the carrier concentration increases, the empty R-G centers are increasingly becoming saturated. The recombination portion decreases. This portion will contribute to the signal. Therefore the Sensitivity $S$ of a diode detector increases with an increase in the carrier concentration which is proportional to the instantaneous dose rate $r$.

The instantaneous dose rate will be affected by SSD and wedges. The change of beam scatter conditions could also affect the detector response. Therefore, the SSD dependence of the diode detector is the combination effect of the instantaneous dose rate dependence and the beam scatter variation with SSD. It is important at nominal treatment beam conditions.(49)
Note that the selected console dose rate, or average dose rate, has no affect on the diode sensitivity, because the instantaneous dose rate is the same.

3. **Sensitivity variation with the accumulated dose (svwad)**

Radiation dose introduces defects in the semiconductor which form minority carrier R-G centers and traps. These R-G centers and traps will reduce the minority carrier lifetime $\tau$. Therefore, the sensitivity of the diode detector decreases with the accumulated dose.
- $svwad$ increases with the increase of the beam energy. Higher energy beam causes more radiation damage to the diode.
- $svwad$ decreases with an increase of accumulated dose. In other words, the sensitivity degradation will slow down with accumulated radiation.
- Considering the average recoil electron energy, high energy photon beams cause higher sensitivity degradation to the diode than expected. This is considered to be due to the much higher damage effect of the presented neutrons in the beam, even though the percentage of the neutrons is small.
- Typically, p type diode detector has lower $svwad$ due to the fact that hole capture cross section is larger than electron capture cross section. This means that the traps generated by radiation can capture more conductive minority carrier holes in n type detector than electrons in p-type detector.

4. **Leakage current**

Diode current generated by sources other than radiation is considered to be leakage current. Such a source could be heat, light electrical bias, etc. In the application of in-vivo dosimetry the major leakage current comes from the recombination of the thermal and electrical current. When a bias voltage is applied across the pn junction, electrical current will flow through the diode. (32)This current is well known as:

$$I_c = I_{rs}(e^{qV/kT} - 1)$$  \hspace{1cm} (2)

Where $V$ is the external bias voltage. $I_{rs}$ is the current of the diode under large reverse bias voltage. $I_{rs}$ increases with the concentration of the defects in the semiconductor. Therefore, leakage current will increase with the accumulated dose.

$I_{rs}$ strongly depends on the temperature. The diode current generated by radiation is also temperature dependent. Therefore the current or sensitivity variation with temperature (svwt) of the detector is the combination effect from the leakage current and the radiation current. The sensitivity of the diode detectors have been observed to increase with the increase of temperature.
From equation (2) \( I_e \) exponentially increases with \( V \). If \( V = 0 \), \( I_e \) zero. Since diode detector works at zero bias, ideally, the ideally, the leakage currents should be zero. However, due to the offset voltage of the amplifier, there is always a small bias across the diode. A good diode electrometer requires the offset voltage of the amplifier to be 10µV or less.

5. **Mechanical construction**

The mechanical construction of the detector is important and must be described in detail. The connecting material should have an atomic number which is close to silicon, such as aluminium, and the encapsulation material should be as water equivalent as possible to suppress interface phenomena, which otherwise could give rise to serious directional dependences.\(^{(56)}\)

Based on the mechanical construction, the effective measuring point should be given to an accuracy of better than 0.5mm for a detector to be used in a water phantom and within 1mm when used for patient dosimetry.

6. **Detector volume**

The effective measuring volume and the detector area should be stated since the statistical noise in the reading is correlated with the detector size (Rikner et al 1984). The smaller the detector, the better the spatial resolution, but the statistical noise increases and so does the measuring time for the same precision. Silicon semiconductor detectors may be so small that microdosimetry aspects have to be considered. Recalculation of results from Rikner et al 1984 for a p type detector with an effective measuring depth of about 60µm shows the number of pulses that have to be collected in order to have a precision of 0.2% at a 95% confidence level as a function of detector area.\(^{(52)}\) The detector area was assumed to be directed perpendicular to the beam axis. Figure 7 is valid for dose per pulse corresponding to a dose rate of 3 Gy min\(^{-1}\) and 300 pulses/s. A decrease in area from 4mm\(^2\), a commonly used detector size, to 1mm\(^2\) increases the measuring time four times for the same relative variance.

Semiconductor detectors must be designed and specified to meet the requirements of the dosimetric application. Misleading results have been reported for unspecified diodes connected to unspecified electrometers.

It is important to choose a p-type detector to avoid a non-linear response as effect of radiation damage when used in pulsed radiation. A doping level of \( 10^{15} \) atoms / cm\(^3\) seems reasonable in most clinical situations to obtain a linear reading in pulsed scatter-homogenized radiation beams from clinical accelerators. In special measurements in scanned beams with very high instantaneous dose rates a higher doping level can be necessary.
The sensitivity variation with temperature and pre-irradiation level may vary between different detector types, so this parameter should also be specified for each detector type.

In clinical water-phantom dosimetry there is a need for different detector volumes: one of an area less than 1mm², to be used in special cases when extremely good spatial resolution is needed, and one of about 4 mm², which in most clinical situations is a good compromise between good spatial resolution, statistical noise, and measuring time. In measurements of total body irradiation (TBI) where low dose rates are used, there is also need for a detector with a large area, as the measuring time will otherwise be long or the statistical noise too high. A high spatial resolution is normally not necessary in TBI measurements. In patient dosimetry the detector size can be small, in the range of 1-2 mm², as the integrated dose will be quite high. The thickness of the detector is assumed to be less than 100µm.

Pre-irradiated p-doped silicon semiconductor detectors designed and specified in accordance with the requirements mentioned should be good detectors both in relative dose determination and patient dosimetry. Of course in dose determination dosimetry, the semiconductor has to be supported by an ionization chamber in order to establish the calibration factor. If the silicon semiconductor detector is pre-irradiated the calibration factor does no change much with time of use, which means that constancy checks at time intervals comparable with those for ionization chambers are adequate.
CHAPTER 3

Materials and Methods

3. Materials

3A. Semiconductor detectors

The detectors used for in vivo dosimetry were silicon diodes. Specifically ISORAD-p detectors with model number 1163000-0. ISORAD –p detectors are designed with cylindrical symmetry (cylindrical build-up cap), which can be beneficial in some applications, such as tangential treatments. They are designed to serve as in-vivo dose verification tools. Besides promoting equilibrium and filtering low energy radiation, each cap is engineered to respond most effectively at one of three different energy range materials. Each energy range uses a different build-up material designed to give a reading at $d_{\text{max}}$, therefore no additional build-up is required. (7) The integral build-up cap of the 6-12 MV energy range is made of brass (1.36 g/cm²). ISORAD – p diode detector is an improved version of the original ISORAD detectors. The cylindrical configuration and the build-up remain the same; however the detector element has been replaced with new proprietary p-type silicon on junction diode. They are constructed with lower Z material and with a thinner wall surrounding the detection element. They also have a highly visible color band to easily identify the different photon models. A machined ring indicates the location of the sensitive volume of the detector.(14) The compensated detector has a white strip to indicate the sensitive volume. For all measurements diodes are connected to commercially available electrometers with low input impedance.

3A1. Lifetime

As with any diode, ISORAD-p sensitivity will gradually change over time. This rate of change will depend on the energies used and the frequency of use. For normal use, plan a re-calibration once per year. Note that ISORAD-p life expectancy is not limited by radiation exposure- in fact the more an ISORAD-p diode detector is used the less its response characteristics will change. Life expectancy is an issue of proper handling and care. When handled under normal stress loads, the ISORAD-p should be expected to last $> 1,000,000\text{cGy}$.(9)

3A2. Construction

ISORAD-p construction is an advanced second generation design. Build-up consists of material that is best suited for each energy range.
3A3. Features

ISORAD-p is the only available diode with Cylindrical Symmetry, angular corrections are not required. Advanced second generation p-type design is accurate and responsive. Offers compatibility with any diode based dosimetry system and has excellent reproducibility at >0.5% for 1cGy. It should be noted that ISORAD – p detectors are not waterproof.

3A4. Specifications

ISORAD-p detector (p-type) connected to a dedicated electrometer with low input impedance and low offset voltage.
The effective detection area of the diode used was 1.65x1.65 mm², the effective detection thickness was 50 µm and the active volume was 0.14 mm³ (effective detection area x effective detection thickness).

Figure 8: Cross sectional view of ISORAD-p detector photon series

The main advantage of the semiconductor detectors is the direct reading which allows the following:
- To detect on-line treatment errors in individual patients and correct them
- To evaluate the quality of a specific treatment technique at different tumor locations
- To estimate the overall accuracy of a department or specific treatment units.
- To measure the dose in situations when the dose calculation may be inaccurate.

3B. Ionization Chamber

The calibration of the semiconductor detector was based on an ionization chamber. The ionization chamber used was a PTW 31003 FLEX chamber. The flex (and semiflex) chambers are designed for therapy dosimetry. They have a short stem for mounting and a flexible connection cable. The nominal energy range is from 30 keV to 50 keV photons. The wall material is graphite with a protective acrylic cover. The guard
ring borders the measuring volume. (21) An acrylic build-up cap for in-air measurement in Co-60 beams is included with each chamber, as well as a calibration certificate for calibration in absorbed dose to water or in air kerma. The chamber is shaped cylindrically with an inner diameter of 5.5 mm; it has a cavity volume of 0.3 cm³. It also is a waterproof thimble chamber for measuring high-energy photon and electron radiation in air, water and phantom material.

3C. Electrometer and Linacs
The ionization chamber was connected with an electrometer. The electrometer used was a UNIDOS electrometer. UNIDOS is well known and accepted worldwide as the therapy dosimeter of choice with the best performance available on the market. (23) Ion chambers and solid-state detectors can be connected to UNIDOS. It displays the measured values of dose and dose rate in Gy, Sv, R, Gy/min, Sv/min, R/min or Gy·m. In our case R/min was used. The electrical values of charge and current are displayed in C or A. It features both mains and battery operation. Before the ionization chamber-electrometer was used enough time should be allowed for the ionization chamber to reach thermal equilibrium.

Two linacs were used. Firstly the PHILLIPS SL 18 linear accelerator which gives 6/15 photon energies and of energies 4-18 MeV and secondly the PHILLIPS SL 75-5 linear accelerator which gives 6 MV photons and no electrons. The ISORAD – p diode was connected to a DPD-5 electrometer. The reading of the electrometer was the entrance and exit dose in the 6 MV and 15 MV photon beam produced by the Phillips Linacs. So the diode characteristics are investigated in 15 MV and 6 MV photon beams provided by the two linear accelerators.

3D. Phantom
A parallelepiped plastic water phantom PPMA (ρ = 1.19 g cm⁻³) with slices of area 30x30 cm² and thickness ranging from 0.1 to 2 cm were used including a specially shaped insert for the ionization chamber. (43)
CHAPTER 4
METHODS

Our aim was to deduce the midline dose from measuring entrance and exit doses. To do that calibration of diodes was necessary. The diode was calibrated against an ionization chamber/electrometer system. Diode was placed at the top of plastic water phantom and the ionization chamber at the $D_{\text{max}}$ for corresponding energy i.e., at 1.5 cm for 6 MV and 2.5 cm for 15 MV. The diode signal is influenced by different factors (physical and geometrical). Due to these dependencies, the “true” entrance dose will be incorrectly reflected by the diode signal. Calibration procedure consists of two parts: 1) calibration in reference set-up to establish the diode calibration factor, $CF$, for the absorbed dose to tissue, and 2) the evaluation of a series of correction factors, $C_i$, to account for calibration differences when measurements are performed under non-reference conditions. The subscript “$i$” refers to the actual irradiation condition, e.g., $C_{\text{SSD}}$, $C_{\text{FS}}$ etc.

4.1 INITIAL MEASUREMENTS

Diode was subjected to a set of tests that were performed whenever a new diode is received. These initial tests consist of the measurement of:

(a) **Signal stability after irradiation**: The display of the signal taken immediately after irradiation was compared with the display of the signal 5 min after the end of the irradiation.

(b) **Intrinsic precision**: The mean and the standard deviation of ten measures were calculated.

(c) **Linearity of the response with dose**.

The linear relation between the response and the absorbed dose was verified (MU-dose). Linearity of the diode-electrometer was also verified between 15-600 MU.

(d) **Perturbation of the radiation field behind the diode**. (17)

The dose perturbation by the diodes in the photon beams was measured in the Solid water phantom using the ionization chamber mentioned before. Dose perturbation was defined as the percent of deviation of the dose by the presence of the diode compared to an open field. The relative dose variation as well as the absolute dose variation values. The measurements were performed at the depth of dose maximum with phantom thickness of 12 cm and under reference conditions. (28)
**DIODE CALIBRATION (entrance and exit calibration factors)**

Measurements were made for two photon beam qualities (6 MV and 15 MV) using ISORAD-p diode. The calibration was performed against an ionisation chamber system. The ionization chamber has a calibration factor traceable to the National Standard Dosimetry Laboratory in Spain and was used to determine absolute dose values. The diodes were taped on the surface of the plastic water phantom (PPMA) near the field center, in such a way that the response of the ionization chamber was not perturbed. The reading \( R \) of the diode, positioned on the plastic water phantom (entrance or exit surface) was compared with the dose measured at the entrance or exit point. The ionization chamber was placed at depth of dose maximum (1.5cm for 6 MV and 2.5cm for 15 MV) and is thus probing the depth dose curve at its maximum, and not at its subsequent fall-off. The *entrance point* is defined as the point along the central axis of the beam at distance \( d_{\text{max}} \) from the entrance surface of the phantom. \( D_{\text{max}} \) is equal to the depth of dose maximum in the phantom. The *exit point* is defined as the point along the central axis of the beam positioned at distance \( d_{\text{max}} \) from the exit surface of the phantom.

The calibration was performed under reference irradiation conditions. These reference irradiation conditions were: a field size of 10x10 cm\(^2\) at the isocenter, a source to skin distance (SSD) of 100 cm, a thickness of polystyrene phantom of 12 cm and the room temperature was always between 17º C and 20º C. First, the reading in plastic was converted to a reading in water by multiplying the reading in plastic with an experimentally determined factor (in the case of plastic water, this factor is 1).

The calibration factor \( F_{\text{cal}} \) was determined as the ratio of the absorbed dose determined with the ionization chamber and the diode reading.

\[
F_{\text{cal}} = \frac{D_{\text{ic}}}{R_{\text{diode}}}
\]

Where \( D_{\text{ic}} \) is the dose at the ionization chamber and \( R_{\text{diode}} \) is the diode reading at reference conditions.
In order to calculate the exit calibration factor of the diode the gantry was rotated 180° and everything else remained the same (diode at the surface of the plastic water phantom, ion chamber at D_{max}, SSD=100cm, Field size 10x10 cm²).

The calibration factor needs to be verified on a regular basis, because radiation damage affects diode sensitivity. For p-type diodes, a recalibration will be necessary after about one kGy. Recalibration has to be performed much more frequently for n-type diodes due to their faster decrease in sensitivity. Besides a calibration factor, determined under reference conditions, correction factors have to be applied in order to have an accurate dosimetry. As it is already mentioned the correction factors originate from the variation in the sensitivity of the diode with dose per pulse, the photon energy spectrum, the temperature and from directional effects.

**DIODE CORRECTION FACTORS**

For irradiation conditions deviating from the reference conditions, a change in the calibration factor can be expected. Such a change will be related to change in the dose per pulse value and changes in the energy spectrum of the photon beam. For each diode, the variation of entrance and exit calibration factors were investigated systematically as a function of phantom thickness, field size at the isocentre, SSD, presence of wedges and temperature. The influence of these variations in irradiation conditions on the calibration factor can be expressed as a correction factor, C_i. The subscript i refers to the actual irradiation condition, e.g. C_{thickness}, C_{field size}, C_{SSD}, C_{wedge} and C_{temp}. These correction factors will in general be different for entrance and
exit points. The dose value at the entrance or exit point, determined from the diode reading, can now be expressed as

\[ D_{\text{diode}} = R_{\text{diode}} N_{D_i} C_i \]

In which the \( C_i \) values are independent of each other. Correction factors accounting for the variations in response were determined as the ratio of the reading of the ionization chamber and the reading of the diode for a clinical irradiation situation normalized to the same ratio for the reference situation. (23)

\[ \text{CF}_i = \text{CF}_{FS}, \text{CF}_{SSD}, \text{CF}_{\text{wedge}}, \text{CF}_{\text{tray}}, \text{CF}_{\text{block}}, \text{CF}_{\text{angle}} \text{ and } \frac{R_{ic} / R_{\text{diode}} \text{ clinical condition}}{R_{ic} / R_{\text{diode}} \text{ ref. condition}} \]

If the dosimetric characteristics of a diode are not (well) known it is recommended to check its response extensively at different irradiation conditions to establish the range where no correction factors are needed. As the type of diode is the major determinant of the magnitude and the behavior of most of the correction factors, diodes of the same type will require similar correction factors, thus showing similar tendencies. However when high accuracy is required it is advisable to check also the correction factors for each individual diode. (52)

To determine these correction factors:

- The ionization chamber was placed at the depth of dose maximum in the plastic water phantom for the beam energy and the field size being used.

- The diode was fixed on the surface of the plastic water phantom on the field axis. Because of measurable beam attenuation it was not possible to place the diode collinear with the ionization chamber on the beam central axis. Instead both the diode and the chamber were moved a few centimeters away from the central beam axis.

- Next, a few groups of measurements were made. First, the diode reading and the charge collected from the ionization chamber were recorded for each beam energy as a function of the collimator setting at the nominal machine SSD of 100 cm.
In order to get accurate SSDs, the couch height readings on the console monitor should be used, since the optical distance indicator (ODI) is not accurate for SSDs far from 100 cm. The couch table readings are more accurate. This is very important for wedge fields, since, a 1 mm deviation of the diode’s position could lead to a deviation as large as 1% in diode readings in a 60° wedged field.

The field sizes at the surface of the solid water phantom ranged from 4 x 4 cm² to 40 x 40 cm².

Next both diode and ionization chamber readings were evaluated as a function of SSD for constant field size of 10 x 10 cm².

The SSD varied within the useful range of 85-105 cm.

Next diode and ic readings were obtained for varying field size, constant SSD of 100 cm and wedge angle of 60°.

Also the same readings were obtained using a tray for SSD 100 cm and varying field size. In all groups of measurements so far, a constant monitor set of 100 MU was delivered for the 6 MV and for 15 MV.

Another set of measurements took place to check for the linearity of the detectors. The monitor units varied from 100 to 600 MU while diode and ionisation chamber readings were noted.

After the determination of correction factors, the semiconductor signal can be converted in measured dose for the different treatment conditions met in clinical practice. The correction factors for the temperature the directional response ad the dose rate response were neglected.

**FIELD SIZE CORRECTION FACTORS**

The field size correction factor is defined as

\[
CF_{FS} = \frac{OF_{ic}(c)}{OF_{diode}(c)}
\]

where \(OF\) is

\[
OF(c) = \frac{R(c)}{R(10\,\text{cm})}
\]

with \(c\) the side of the square fielding cm and \(R\) the reading. If the measurements of \(OF_{diode}\) are performed at the same time as the measurements of \(OF_{ic}\) using a plastic phantom, attention should be paid to \(OF_{ic}\) because it may differ from \(OF_{ic}\) measured in water, so a factor to convert the reading in plastic to reading in water should be applied. (28) This factor would be probably depending on the field size. To simplify things, one can measure \(OF_{diode}\) and compare it with \(OF_{ic}\) measured in water at depth of dose maximum.
In order for the field size correction factors to be determined, the diode was positioned at the surface of the plastic water phantom, almost 1 cm away from the center of the field so it will not perturb the reading of the ionization chamber. The ionization chamber was placed at \( D_{\text{max}} \) (1.4 cm and 2.5 cm depth for 6 MV and 15 MV respectively).

Irradiation of 100 MU took place and the field size varied (4x4, 6x6, 8x8, 10x10) while SSD was 100 cm. Readings from the diode and from the ionization chamber were taken. (38)

Two correction factors were found: entrance correction factor and exit correction factor.

For the calculation of the exit correction factor the diode was placed at the surface of the phantom, the gantry was rotated 180\(^\circ\), the ionization chamber was placed at the depth of dose max but from the exit surface and irradiation (100 MU) took place. (42)

All the data points that were found were normalized to the ratio of ionization chamber and diode readings at reference conditions.

**TRAY CORRECTION FACTOR (\( CF_{\text{tray}} \))**

Shielding blocks are positioned on a tray to the treatment head. In our hospital the tray for the Linear Accelerator was made of PPMA of 0.5 cm thickness. Inserting a tray between the source and the patient changes the amount of the electrons that reaches the patient’s skin. Therefore if the diode does not have the appropriate build-up cap, the tray correction factor varies with field size. (46)

To measure the tray correction factors the tray transmission for different field sizes at the depth of dose maximum was measured first with an ionization chamber and then with the diode taped to the surface of the plastic water phantom.

The transmission factors measured with the ionization chamber and with the diodes are compared, and the \( CF_{\text{tray}} \) as a function of field size is obtained.

\[
CF_{\text{tray}} = \frac{\text{transmission}_{ic}(c)}{\text{transmission}_{\text{diode}}(c)}
\]
Where the transmission is defined as:

\[
\text{transmission}(c) = \frac{R(c,\text{tray})}{R(c)}
\]

Where \(c\) is the side of the square field in cm, and \(R\) the reading. The same experimental set-up was used as before but now the field size was varied (4x4, 6x6, 8x8...40x40 cm²).

**WEDGE CORRECTION FACTOR (\(\text{CF}_{\text{wedge}}\))**

Wedge filter: beam modifying device (most linear accelerators are provided with a selection of wedged filters mounted on the treatment head of the machine). Phillips 75/5 linear accelerator has a wedge (called auto wedge) with nominal angle of 60°.(21)

Inserting a wedge in the beam results in a decrease of the dose rate and a hardening of the spectrum of the beam. Therefore, as the sensitivity of the diode depends on dose rate and energy, a correction factor different from 1 is expected when wedges are used.(5)

The wedge correction factor is defined as the ratio between the wedge transmission factor for a 10 x 10 cm² field, measured with the ionization chamber placed at the depth of dose maximum, and the wedge transmission factor for the same field size, measured with the diode placed at the field center taped on the surface of the plastic water phantom.

\[
\text{CF}_{\text{wedge}} = \frac{\text{transmission}(w,10 \times 10\text{cm}^2, z_{\text{max}}, \text{ic})}{\text{transmission}(w,10 \times 10\text{cm}^2, \text{diode})}
\]

With \(w\) the wedge angle.

\(\text{CF}_{\text{wedge}}\) is considered to be independent of field size and \(\text{CF}_{\text{wedge}}\) determined for a field size of 10 x 10 cm² was used. (9)
The source to skin distance was constant at 100 cm, the field size was variable, the diode was placed at the surface of the phantom and the ionization chamber was placed at $D_{\text{max}}$ from the entrance and from the exit surface in order to determine the entrance and exit wedge correction factors respectively.

Figure 13: Experimental set-up used for the determination of wedge correction factors (entrance and exit).

SSD CORRECTION FACTORS (ENTRANCE AND EXIT)

When the SSD is changed, the dose per pulse and the electronic contamination change. First, the sensitivity of the diodes depends on dose per pulse.

Secondly if the build-up cap of the diode is not thick enough an overestimation of the dose at short SSD can be due to electronic contamination that would be seen by the diode but not by the ionization chamber placed at the depth of dose maximum. Therefore, a SSD correction factor different from one is expected. (11)

Again, the experimental se-up is the same with the one used before (no beam modifiers included). The field size remained constant at 10 x10 cm$^2$ while the SSD varied. For exit SSD correction factors the gantry was rotated 180 degrees.

The correction factor for SSD is defined as
Linearity correction factors (entrance and exit)

The linearity correction factors were calibrated by measuring the absorbed doses of 6 and 15 MV photon beams in the dose range of 100 to 600 cGy.

The measurements were performed by placing the diodes on the surface of the plastic water phantom, with 100 cm SSD, in a 10x10 cm² radiation field size for photons.

The ionization chamber was again placed at the depth of dose maximum for each energy quality.

Irradiation with variable monitor units took place and the following formula was used for the calculation of the correction factors.(19)

For the calculation of exit linearity correction factors the gantry was rotated 180 degrees and everything else remained the same. To prevent shadowing of the ionization chamber by the diode, the diode was placed 2cm off axis.

\[
CF_{SSD} = \frac{R_{ic}(z_{max}, 10 \times 10 \text{cm}^2, SSD)}{R_{ic}(z_{max}, 10 \times 10 \text{cm}^2, SSD = 100)} \times \frac{R_{diodes}(10 \times 10 \text{cm}^2, SSD)}{R_{diodes}(10 \times 10 \text{cm}^2, SSD = 100\text{cm})}
\]

TEMPERATURE CORRECTION FACTOR

The temperature dependence should be accounted if the particular diode is used at different temperatures. This may be done by applying a constant temperature correction factor or by using a thermostatically controlled calibration phantom. In our case the diodes were placed on the surface of a Plastic Water Phantom so the temperature was not changing as it would have if the diode was taped on the patient’s skin. The temperature was always between 17 and 20 ºC so no correction factors since the variation of temperature was not significant.(25)
CHAPTER 5
RESULTS-CONCLUSIONS
INITIAL MEASUREMENTS

Graph of the response / dose linearity

Graph showing the dose linearity of the diode 6MV.

Graph showing them dose linearity of the diode for 15MV
RESULTS FOR 6MV
The calibration factor obtained for the entrance dose measurements of 6 MV was 2.29 cGy/reading and for the exit dose measurements was 2.41 cGy/reading.

After the calculation of correction factors for 6 MV the following graphs were obtained showing the variation of the correction factors with field size, SSD, wedge and tray.

ENTRANCE MEASUREMENTS- 6MV

5.1 FIELD SIZE CORRECTION FACTORS AND THEIR DEPENDENCE ON FIELD SIZE

![Graph showing the entrance diode reading for different field sizes](image)

Figure 14: Graph showing the entrance diode reading for different field sizes
Figure 15: Graph showing the entrance IC reading for different field sizes.

Graph 16: Graph showing the CF field size for different field sizes.

Figure 16 shows the CF field size for the diode at SSD 100cm. Generally the field size effect is due to the different irradiation condition between the diodes and the ionization chamber. Since the diode is at the surface, and lacks an overlying layer, its reading is less dependent upon the phantom scatter, and his heavily dependent on the head scatter. Therefore CF increases as the diode under responds with increase in the field size. So for the Isorad-P diode a normal behaviour for the field size correction factors is to increase with field size.
The large variation of the 4x4 cm field size was due to the underexposure of the ionization chamber. For small sizes (<5x5 cm²) most of the photons came from the primary beam because the scatter phenomena (low energy photons, contaminating electrons) are low. As the field size increases, the number of the scatter photon increases compare to photons coming from the primary beam.

From the graphs listed above (figure 14, figure 15) it was observed that as the field size increased the entrance reading obtained from the ionization chamber and from the diode was also increasing. This was due to the fact that as the amount of low energy contamination increases as the field size increases. So the diode reading at entrance measurements follows the behaviour of the ic reading. This is explained by the fact that the diode is not as sensitive to radiation as the ionization chamber. Due to this lack to radiation, diode follows abrupt behaviour to its values as the field size increases.

**EXIT DOSE MEASUREMENTS-6 MV**

![Figure 17: CF exit field size vs Field size](image)

Figure 17 shows a plot of the normalized ratio of the ionization chamber to diode reading for exit dose measurements for constant SSD OF 100cm with energy of 6MV. Again it was observed that the response curves were quite steep for small fields. As the field size was increasing a more gradual decrease was observed. So the behaviour is the same as the entrance CF but nw we have larger variations due to the absorption of photons inside the phantom. (53)

Generally the field size effect is due to the different irradiation conditions between the diode and the ionization chamber. Diode measurements are taken at the surface of the plastic water phantom while ionization chambers are taken at a specific depth. Fir low energy photon beams, scattered radiation form
both overlaying and unerlaying material contributes to the diode and ionization chamber readings. For high energy photon beams, however the backscatter is negligible and only scattered photons from the overlaying layer contribute to the diode and ic reading. (31)Since the diode is at the surface and it lacks overlying layer its reading is less dependent upon the phantom scatter and dependent heavily on the head scatter. Therefore CF increases as the diode under responds with increase of field size.

SSD CORRECTION FACTOR DEPENDENCIES

ENTRANCE MEASUREMENTS

Figure 18: Graph showing the variation of diode R with SSD (6MV)
When SSD is changed, the dose per pulse and the electronic contamination changes. The sensitivity of the diode depends on the dose per pulse. If the build-up cap of the diode is not thick enough an overestimation of the dose at short SSD can be due to the electronic contamination that would be seen by the diode but not by the ionization chamber placed at the depth of dose maximum. The variation of SSD with CF ssd is basically determined by the variation of the relative sensitivity with dose per pulse.
During our measurements the couch of the linear accelerator could not go below 85 cm so no measurements were taken for SSD less than 85 cm. From 85 until 105 cm it was observed that the CFssd was increasing with increasing SSD. It was also observed in the previous graphs 18 and 19 that while SSD was increasing the diode reading and the ic reading was also increasing. This was because when SSD increases, the number of contaminating electrons and low energy photons which reach the diode at the surface of the phantom and the ionization chamber at Dmax, decreases. Electrons generated in the collimators have a wide angular distribution and, therefore, are more pronounced at short collimator to phantom distances compared to large distances where they spread out. Also, from figure 20 we observed that CF ssd was very close to unity. That happens because measurements made with the ISORAD-p diode were independent of SSD, so the decision of using or not a correction factor for SSD changes depends on the physicist.

**EXIT MEASUREMENTS**

![Graph of exit diode reading vs SSD](image)

Figure 21: Graph of exit diode reading vs SSD
From figures 21 and 22 it was observed that the diode reading and the ionization chamber reading for exit dose measurements was decreasing as the SSD distance was increasing. As far as the CF ssd exit, it was observed that the readings obtained were close to unity and this tells us that again the diode used had very small dependence from SSD changes.
**WEDGE CORRECTION FACTOR DEPENDENCIES**

The following figure shows the “wedge factors” for diode. They are not the ordinary wedge factors measured using ion chamber. The wedge factor for diodes used here is the ratio of diode reading with wedge over that without wedge.

**ENTRANCE MEASUREMENTS-6MV**

The graph shows the relationship between entrance IC reading and field size with a wedge angle of 60 degrees. The equation that describes this relationship is:

\[ y = 2\times10^{-5}x^3 - 0.0025x^2 + 0.1119x + 4.1875 \]

with a good fit of 

\[ R^2 = 0.9994 \]

Figure 24: Graph of entrance IC reading vs field size (wedge)
Our linear accelerator was provided by a 60° wedge angle. The term wedge angle refers to the angle through which the isodose curves are tilted relative to their normal position perpendicular to the beam axis. Inserting a wedge in the beam results in a decrease of the dose rate and the hardening of the spectrum of the beam. Therefore as the sensitivity of the diode depends on both dose rate and energy, a correction factor different than one has to be applied. So when a wedge is inserted the photon energy spectrum changes and the dose per pulse vary with the wedge factor.
Due to the differences in shape and on energy dependence of the diodes at the entrance and exit sides, it can be expected that the beam hardening due to the presence of wedge will result in an increase of sensitivity at the exit side. So we observed that for entrance dose measurements the CF wedge had a small variation with the field size and the values of CF wedge were very close to 1. For exit dose measurements though, the variation of CF wedge with the field size was greater and the CF wedge was decreasing with increasing field size.
Figure 27: Graph showing linearity correction factor entrance vs MU

From figure 27 we can see that the variation of the entrance correction factor against MU is not very high. Actually the Entrance linearity correction factors do not change a lot while the Monitor Units varied from 100 to 600 MU. The values of CF lin are very close to unity so the correction factor is not of a great importance.
The linearity correction factor was measured for SSD of 100cm and field size of 10x10 cm² and for different monitor units. From the figure above it was observed that we had very small differences from unity. This was due to the absorption of photons from the diode. The decision of using or not this correction factor depends only on the physicist that will do the calibration of the diode.

Figure 28: Graph showing linearity correction factor exit vs MU.
MEASUREMENTS WITH TRAY

ENTRANCE-6MV

Figure 29: Entrance diode reading vs Field size (tray, 6MV)

Figure 30: Entrance ionization chamber reading vs Field size (tray, 6MV)
Figure 31: CF tray entrance vs field size (tray)

EXIT-6MV

Figure 32: Graph showing exit diode reading vs field size (tray)
Shielding blocks are positioned on a tray attached to the treatment head. The tray used was made of PPMA of 0.5cm thickness. Inserting a tray between the source and the patient changes the amount of electrons that reaches the patient’s skin. Therefore, if the diode does not have the appropriate build-up cap, the tray correction factor varies with field size. From the graphs we could conclude that the CFtray entrance and exit did not vary very much with the field size. The range of their value was very close to unity.
6. RESULTS FOR 15MV

The calibration factor of the diode for the entrance dose measurement and for the energy of 15MV was found to be 2.13 cGy/reading and for the exit dose measurements was found to be 2.47 cGy/reading.

FIELD SIZE CORRECTION FACTORS AND THEIR DEPENDENCE ON FIELD SIZE

Figure 34: Graph showing entrance ionization reading against field size

Figure 35: Graph showing entrance diode reading against field size
Figure 36: Graph showing CF fs entrance vs field size

Figure 37: Graph showing the exit diode reading vs Field size
Figures 36 and 39 show them field size correction factors for the diode detectors for the energy of 15MV. It was observed that for both entrance and exit dose measurements the diode correction factors decrease with increasing field size. For small field sizes, many photons come from the primary beam because the scatter phenomena are low. This underexposure is due to the presence of the diode which perturbs the reading. As the field size increases gradually, the number of scatter photons increases compared to photons coming from the primary beam. Due to the increase in scatter photons as the field size increases, we have an over
exposure of the diode so the diode reading increases. This means that as the field size increases, CF decreases.

It can also be noted that range the variation of CFx was between 0.965 and 1.001 for entrance and 0.965 and 1.008 for exit dose measurements.

SSD CORRECTION FACTORS

**ENTRANCE DIODE READING VS SSD**

\[ y = -0.002x^3 + 0.5929x^2 - 59.843x + 2119.5 \]

\[ R^2 = 0.9998 \]

![Graph showing the entrance diode reading vs SSD](image)

**ENTRANCE IONISATION CHAMBER READING VS SSD**

\[ y = -2E-05x^3 + 0.0105x^2 - 1.5824x + 87.185 \]

\[ R^2 = 1 \]

![Graph showing the entrance ic reading vs SSD](image)
Figure 42: Graph showing CF ssd ENTRANCE versus SSD

Figure 43: Graph showing EXIT diode reading vs SSD
From figures 42 an 45 show the variation of SSD correction factor with distance SSD. When the SSD is changed the dose per pulse and the electronic contamination change. The variation of the CF ssd entrance is basically determined from the variation of the relative sensitivity with dose per pulse. CF ssd exit is only determined by the variations in dose per pulse and the inverse square law and is independent of any other correction factors. It was observed that as SSD increases the diode reading decreases. This is because when SSD increases the number of contaminating electrons and low energy photons able to reach the diode at the surface of the phantom decreases. Electrons generated in the collimators have an angular distribution.
and therefore, are more pronounced at short collimator to phantom distances compared to larger distances where they spread out.

So, since it was observed that with increasing SSD the diode reading decreases for the reasons mentioned above, the CFssd increased with increasing SSD. It should also be mentioned that the variation of CF ssd with SSD for entrance and exit dose measurements is between 1,000 and 1,021, 0.985 and 1,021 respectively.

![Graph showing entrance diode/ic reading vs SSD for wedge angle 60 degrees.](image1)

Figure 46: Graph showing entrance diode/ic reading vs SSD for wedge angle 60 degrees.

![Graph showing CF wedge entrance vs Field size](image2)

Figure 47: Graph showing CF wedge entrance vs Field size
Exit diode/ic reading vs field size (wedge angle 60 degrees)

\[ y = 0.0008x^3 - 0.0503x^2 + 1.0581x + 5.7821 \]

\[ R^2 = 0.982 \]

\[ y = 8E-05x^3 - 0.0024x^2 + 0.0559x + 1.6955 \]

\[ R^2 = 0.9661 \]

Figures 48: Graph showing exit diode/ic reading vs field size

Field size vs CF wedge EXIT

\[ y = -2E-05x^3 + 0.0018x^2 - 0.0392x + 1.1591 \]

\[ R^2 = 0.7766 \]

Figure 49: Graph showing CF wedge Exit vs Field size

Figures 48 and 49 show the variation of CF\textsubscript{wedge} entrance and exit with Field size. From the data obtained it was obtained that the CF for both entrance and exit dose measurements did not vary much and their values for different field sizes was very close to unity.
Figures 50 and 51 show the variation of CF<sub>linearity</sub> with increasing Monitor Units for entrance and exit dose measurements. From the graphs it was obtained that for entrance and for exit dose measurements the deviation from unity was very small.
Figure 52: Graph showing entrance diode reading vs Field size (tray)

Figure 53: Graph showing the entrance ic reading vs field size (tray)
Figure 54: Graph showing CF tray entrance vs Field size

Figure 55: Graph showing Exit diode reading vs Field size (tray)
From figures 54 and 57 it was observed the the CF tray entrance and CF tray exit diverge from the value of unity for small field sizes while for larger field sizes CF (entrance and exit) approach unity.
**CONCLUSIONS**

The delivery of a treatment in radiotherapy requires many sequential, complex steps of prescription, imaging, calculation and patient positioning. Every step can contribute to the total uncertainty of delivered dose. So it is necessary to check each step. In vivo dosimetry is the only check that is performed during the patient treatment, and since it is independent of the calculation method. It is the only method that can trace a number of errors. In vivo dosimetry with diodes is relatively easy and accurate with results immediately available. There is an increasing trend to use diode in vivo dosimetry.

However, just as ion chamber responses are subject to designs and environmental say, temperature, atmospheric pressure, etc., silicon diode detector responses are also subject to their designs and operating environment. Diodes of different brands must be characterized individually due to different materials and designs. For accurate dosimetry, this characterization needs to be done individually, since even diodes from same batch can be very different. Additionally, diodes at different Linacs also need to be characterized individually, because the spectra from different Linacs might be different even with the same nominal energy. For any one photon diode detector, the correction factors due to SSD, field size, wedge, temperature, beam incident direction, radiation damage, off-axis distance, etc., need be considered.

For Isorad-p photon diodes, it is generally not necessary to consider the incident beam angle correction up to 60° [22], since they are designed with cylindrical symmetry.

A number of errors in the dose delivery to the patients can be detected by means of in vivo dosimetry. Parts of these errors are human mistakes in the set-up of the patient, of the beam modifying devices and of wrong machine settings. Systematic errors related to the machine performance, the dose calculation procedure and the patient set-up have also been detected with in vivo dose measurements. For all these applications it has been shown that in vivo dosimetry is a useful tool in the quality assurance program of a radiotherapy department.
References


