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IN VIVO DIODE DOSIMETRY SYSTEM

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Abbreviations

(CF): Correction factor.

(CV): Coefficient of Variation.

(EPIDs): Electronic Portal Imaging Devices.

(ICRU): International Commission on Radiological Units and Measurement

(PFD): Photon Field Detector.

(PMMA): Poly methyl methacrylate.

(PSDs): Plastic Scintillation Detectors.

(SDD): Source-to- Detector Distance.

(SVWT): Sensitivity Variation with Temperature.

(TDP): Transversal Dose Profile.

(TLDs): Thermos luminescent Dosimeters.

(TPS): Treatment Planning System.

Contents

CHAPTER III: Results and Discussion

CHAPTER IV: Conclusion

CHAPTER V: Applications

List of figures

Aim of the Work

There are many factors which influence the diode response to radiation for in-vivo dosimetry. In order to achieve the best possible accuracy between the prescribed dose and the measured dose, the proper correction factors need to be applied to the diode in-vivo measurements.This can allow us to catch the human made errors more easily. In this study, the systematic study of dosimetric characteristics of some of the commercial radiation diodes (n-type and p-type) was performed, but not limited to temperature, dose rate and SDD, and energy dependence under different radiation beams.These studies can help us better quantify the dosimetric characteristics of different diode detectors and obtain greater accuracy for relative in-vivo dose measurements in the clinic.

ABSTRACT

In-vivo dosimetry with diode detectors is used in radiation therapy as a quality assurance tool. The diode sensitivity under radiation depends upon temperature, dose rate and SDD (source-to-detector distance), field size, beam angle, and energy. In the temperature dependence study, the systematic study on the dose rate dependence of SVWT (sensitivity variation with temperature) was performed. It was concluded that sufficient preirradiation can eliminate dose rate dependence of SVWT. However, preirradiation cannot eliminate dose rate dependence of the diode sensitivity, S, itself. In the dose rate and SDD dependence study, it was shown that the p-type diodes do not always show less dose rate dependence than the n-type diodes. Preirradiation does not always reduce diode dose rate dependence. SDD dependence of diode sensitivity can be explained by the instantaneous dose rate dependence if sufficient buildup is provided to eliminate electron contamination. In the energy dependence study, the energy dependence diode detectors are quantified.

CHAPTER (1)

Introduction and Literature Review

1. Introduction

After x-rays were discovered by Wilhelm Conrad Roentgen in 1895, the ionization radiation has been used for the treatment of cancer. Nowadays, surgery, radiotherapy and chemotherapy are the three main methods for treating cancer. The radiotherapy consists of teletherapy and brachytherapy. Teletherapy mainly applies 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 medical linear accelerator (Linac) is the most widely used device for External beam radiotherapy. The aim of radiotherapy is to deliver a high dose to the target while delivering the lowest possible dose to the surrounding healthy structures. Experimental and clinical evidence shows that small changes in the dose of 7% to 15% can reduce local tumor control significantly [1]. So the International Commission on Radiological Units and Measurements (ICRU) recommends that the dose delivered to a tumor be within 5.0% of the prescribed dose [2].

Each of the many steps in the treatment planning and execution will contribute to the overall uncertainty in the dose delivered. Therefore, some organizations (AAPM [3], ICRU [2]) recommend that in vivo dosimetry (i.e. assess the dose directly in the patient) should be made. In vivo treatment verification includes geometrical and diametrical verification.

The geometry, i.e. the patient anatomy and tumor location, can be obtained by using a simulator, CT or MRI. Usually the CT and/or MRI data (image fusion) are used to design the 3D treatment plan with a computer treatment planning system.

However, due to setup errors and internal organ motion, the planned high dose volume may not agree with the target very well. The laser alignment system, immobilization system etc. Can reduce the setup and motion errors effectively, and portal imaging and electronic portal imaging devices (EPIDs) can be used to check the position of a patient during the irradiation. However, internal organ motion is somewhat difficult to control and check. The typical examples are the lung and the prostate. Their movements are up to several centimeters. The dosimetric treatment verification is also very important. Each step can contribute to the final dose uncertainty, for example, geometry errors mentioned above, errors introduced by transferring treatment data from the treatment planning system or simulator to the accelerator, errors of beam setting,etc. The final accuracy of the dose delivered can only be checked directly by means of in vivo dosimetry.

1.١ Different Types of in-vivo dosimetry

The most commonly used detector types for in vivo dosimetry are diodes and thermos luminescence dosimeters (TLD). The diode is superior to TLD, since the diode measurements can be obtained on line and allow an immediate check. Other advantages of diodes include high sensitivity, good spatial resolution, small size, simple instrumentation, no bias voltage, ruggedness, and independence from changes in air pressure [4]. The sensitivity relative to the ionization volume is high for a semiconductor, about 18,000 times higher than for an air ionization chamber. The average energy required to produce an e--hole pair in silicon is only 3.5eV compared with 34eV in air. The sensitive volume can thus be small, and hence the diode detector has high spatial resolution [5]. However, there are many factors that can affect the response of the diode to radiation, and diodes are different from one to another, even from the same batch, same model and same manufacturer. So the commissioning or characterization of every diode individually is necessary for accurate dosimetry [6,7].

1.٢ Types of silicon diodes

The silicon diodes can be made of n-type or p-type silicon. A semiconductor. With an excess of electrons is called an n-type semiconductor, while one with an excess of holes (electron deficits) is called a p-type semiconductor. Normally a pure silicon crystal has an equal number of electrons and holes. To make an n-type or a p-type silicon, certain impurities need be added into the pure crystal [8]. Silicon is in group IV in the periodic table. If atoms in group V, each of which has five valence electrons, are added to the pure silicon, then there will be an excess number of electrons and finally results in n-type silicon. Similarly, a p-type silicon can be made by adding an impurity from group III to the pure silicon.

Generally, the impurities used are phosphorus from group IV and boron from group III. One of the crucial keys to semiconductor detectors is the nature of the P-N junction. When p-type and n-type materials are placed in contact with each other, the junction behaves very differently than it does with either type of material alone. Specifically, current will flow readily in one direction but not in the other, creating the basic diode. If the n- region is connected to the positive terminal and the p region to The negative, which is known as reverse bias, almost no current (except for a very small current due to thermally generated holes and electrons) flows across the junction. Under this condition, the resistance of the p-n junction is very high, and almost all potential difference falls on the p-n junction, thus creating a strong electronic field in the p-n junction. The region around the junction is swept free by the potential difference. This region in a semiconductor that has a lower-than-usual number of mobile charge carriers is called the depletion layer. The depletion layer is the sensitive volume of the semiconductor detector [9]. The diodes are used without bias voltage in radiotherapy.

1.٣ Theory of operation

The charge collection process is described in the following way [10,11]:

- When an ionizing particle passes through the depletion layer, primary or secondary particles from the radiation source are absorbed, generating electron hole pairs throughout the diode.
- By diffusion, those electrons and holes generated within one diffusion length from the junction will be able to reach the junction.
- The built-in potential across the p-n junction then sweeps the electrons and holes apart and to the opposite sides, giving rise to a pulse in the external circuit.

Some of the radiation generated electron-hole pairs will recombine through the recombination centers. When the instantaneous dose rate (dose per pulse) increases, the generated carrier concentration increase proportionally. Then the recombination centers are becoming saturated and recombination portion decreases. This portion, which is not recombined, will contribute to the signal, therefore the diode detector sensitivity increases. Generally, p type diodes have lower instantaneous dose rate (dose per pulse) dependence than n type [10,11].

Not only is the diode detector dependent on the dose per pulse, but also it Is dependents on the accumulated dose. Because radiation dose introduces defects in the semiconductor and thus forms more recombination centers and traps, the diode detector sensitivity decreases with the accumulated dose. Diode current generated by sources other than radiation, say, heat and light, is considered to be leakage current. The leakage current depends on the temperature. The diode current generated by radiation is also temperature dependent. The sensitivity of the diode detectors increases with the increase of temperature [12]. Ref [13] has shown that the sensitivity variation with temperature of a p type silicon detector increases linearly with increasing temperature. Since the buildup materials and the encapsulation materials are not water equivalent, there are interface phenomena. The shape and geometry of the diode and p-n junction also affect diode's response to radiation. Both of the above two factors give rise to directional dependences [5]. The aim of the thesis is to characterize an In Vivo Diode Dosimetry System for Clinical Use. A model will be made to find the total correction factors (Correction Factor = Dose at Diode/ (Diode reading)), for the diodes readings for given modality (photons or electrons), given energy, given SSD, given field size (cone size), given diode and machine, and given wedge.

1.٤ Literature Review

- **G. VASILE, et al,** was investigated 160 breast treatments for 158 patients were screened using in vivo dosimetry were performed during one year. (131 of patients were treated before the surgical intervention, 27 patients were radiated after the surgical intervention) and they use Apollo 5 electrometer unidose E electrometer, framer chamber and water phantom), Calibration of each detector was (performed against the ionization chamber and in this case the calibration was adjustment of this factor was done weekly in order to avoid the influence of dose accumulation in response of diodes. As the numbers of fields involved in treatments depend on chosen radiation technique. Occasionally, two additionally photon beams for the excessively large mammary glands were added. For the measurement where the deviation is greater than tolerance level (5%). In the third verification action level of quality assurance program, the radiation field with a deviation greater that 5% was investigated. Statistical tests result showed that for 0.05 level the means are significantly different (t-test: t= -.6914, p= 4.87E-11 and One-way ANOVA-test F=1911.8, p= 0). [14]

- **Emma Kelly et, al. This** project was carried out in the Radiotherapy Department of St. Vincent's Private Hospital in Dublin. The aim of this project was to calibrate a diode and to investigate if this semiconductor detector gave an accurate indication of the radiation dose that the patient should receive during irradiation. The diodes performance under clinical conditions was within tolerance when compared with that of the ionization chamber and the TPS, that is to say that the dose was within the 5% ICRU limitation. These calibration factors obtained for the given set-up can be factored into the diode charge reading file, which converts it to dose and gives a more accurate result to within the 3% [15].

- **T.C.dos santos, et, al.** use for poly methyl methacrylate and silicon diodes. The diode was held between PMMA plates, placed at the reference depth and centered in a radiation field of 10 x 10 cm^2 depicts the transversal Dose profile (TDP) measurements performed in PMMA with the EPI diode Placed at the reference depth (2.5 cm) for 12 MeV electron beam Irradiation within the field size of 10 x 10 cm^2. The current response of the diode irradiated with 21 MeV electron beam, show that the diode has good. short term repeatability with coefficient of variation (CV) of 2.4% agree with the TDP simulation performed with the Concentra Master Plan Treatment Planning System (solid line), confirming that the expected good spatial resolution of the EPI diode. The overestimation of the penumbra is due to the finite size of the diode, which causes an averaging effect [15].

- **Cassiane Cardoso Bonato, et,al.** use Sixteen patients with a mean age of 6.99 years received therapy for a variety of non –thyroid cancer in several region of the body , the sample comprised 16 patient from whom 102 scattered dose measurements were obtained by TLDs and 72 paired dose estimates were calculated by TPS. Statistical analysis. This was a cross –sectional study on a sample of children and adolescent who underwent treatment. it uses TLDs: Thermos luminescent dosimeters. The results: Analysis of Bland-Altman plots for the TPS skin /TLD ratio at different levels of radiation. showed that the bias of the TPS/TLD ratio behaved differently in relation to TLD measured doses. Below 1 cGy, the TPS overestimated the actual dose, as measured by TLDs. Between 1 and 10 cGy, the bias ranged around 1, showing agreement between the two methods. Above 10 cGy, the TPS was more likely to underestimate the scattered dose as compared with actual TLD measurements. Although the overall bias was small. (bias = 1.02, the 95% limits of agreement were broad (95% LA: 0.05 to 21.09). Analysis by site of irradiation showed different TPS/TLD biases according to the proximity of the thy- roid gland to the treatment field. when the pelvis or abdomen were irradiated. The TPS overestimated the scattered dose, and there was little agreement between TPS and TLD measurements—bias of 15.01 (95% LA 9.16 to 24.61) and 5.12 (95% LA 3.04 to 8.63) respectively. When the lungs or mediastinum were irradiated, TPS estimates and TLD measurements behaved in a similar fashion, with broad limits of agreement and a bias of 1.13 (95% LA 0.03 to 40.90) and 0.39 (95% LA 0.02 to 7.14) respectively. At doses >10 cGy, the TPS underestimated the scattered dose up to tenfold as compared to TLD-measured values [16].

- **alessandro M. costa, et,al.** use head and neck cancers . A total of 45 thermos luminescent dosimeters (TLD) divided into 2 batches (one of 17 and other of 28 TLDs) were used. the thermos luminescent dosimeters LiF: Mg, Ti (TLD 100) in the form of extruded square ribbons (about **3** Â **3** Â **0.9** mm^3) manufactured by harshaw. Nitrogen flux was used. readouts were taken within 25 s and temperature between 50 C and 25 C. all TLDs of the 2 batches were annealed and irradiated to same dose. After readout, the procedure was repeated 3 times. A calibration was performed during each series of in vivo dose

measurements using 5 TLDs selected at random at random from each batch. A total of 49 treatment fields involving
11 patients randomly selected were included in the pilot study. these patients were patients treated for head and neck cancers. In vivo entrance dose measurements were performed during at least 2 treatment sessions on every patient in every treatment field. The goal was to discover discrepancies larger than 5% between the expected dose and measured dose. Each patient was treated with an immobilization mask with reference marks to the entrance points in each field. The result: The batch of 17 TLDs was found to have an intrinsic precision of 1.5%. The batch of 28. TLDs was found to have an intrinsic precision of 1.6%. The TLD dosimetric system allow individual dose measurements with an expected overall uncertainty lower than 3%. This overall uncertainty is less than 5% [17].

- **Ben mijnheer , Sam Beddar , Joanna Izewaks , Chester Reft** For IMRT treatments of head and neck patients, MOSFET detectors were used to measure the dose in the oral cavity for patients fitted with a customized mouth plate. . And for the treatment of prostate cancer patients, MOSFET detectors with brass build-up caps were positioned on the patient’s skin on the central axis of each IMRT beam to measure the entrance dose. Diode dosimetry was also performed for treatments of head and neck cancer by Vinall et al. and during IMRT treatments of tumors in the head and neck and prostate regions by Kadesjö et al. Both groups achieved good agreement between measurements and calculations; 95% and 92.2% of the results were within the ±5% tolerance level [18].

- **Andrej Strojnik,**Institute of Oncology Ljubljana, Department of Radio physics, Ljubljana, Slovenia, describes the calibration of dosimetric diodes and presents the results of in vivo dosimetry in 209 rectal cancer patients. It's use Materials for Calibration Two dosimetric diodes EDP-20, manufactured by Scanditronix, In reference conditions, each diode was taped to a plastic water phantom (dimensions: 20 cm x 30 cm x 30 cm) at a distance of 100 cm from the accelerator focus, in the center of an open treatment field measuring 10 cm x 10 cm and with gantry angle set to 0°And also use for In vivo measurements: The two diodes were used in routine measurements in rectal cancer patients treated with four-field box technique with the isocenter in the center of the planning target volume. With this technique, the beams delivered the dose to the target from four directions, with the gantry angle values of 270°, 0°, 90° and 180°.Such configuration allowed the diode taped to the patient's skin on the 0°beam's axis to measure not only the 0° beam's entrance dose, but also the 180° beam's exit dose. The same principle applied to the 90° and 270° beams. It's Results: 421 entrance dose and 415 exit dose measurements were performed. The average difference from expected values was 0.9% for entrance dose (SD 2.1%) and -0.5% for exit dose (SD 3.3%). In 209 patients, the average absorbed dose in the isocenter differed from the planned values by 0.2% (SD 1.4%). Measurement results exceeded the tolerance levels in two patients. And The Conclusion, Smaller standard deviation of absorbed dose to the isocenter (1.4%), compared to those of entrance (2.1%) and exit dose measurements (3.3%), confirms a correlation between the entrance and exit dose deviations of pairs of opposed fields. The fact that during this study in vivo dosimetry exposed two cases of potentially inaccurate treatments proves its necessity [19].

CHAPTER (2)
Material and Methods

2. Introduction

In-vivo dosimetry with diode detectors is used in radiation therapy as a quality assurance tool. The diode sensitivity under radiation depends upon temperature, dose rate and SDD (source-to-detector distance), field size, beam angle, and energy. This dissertation presents the first systematic and quantitative study of dosimetric characteristics for most of the commercial radiation diodes (n-type and p-type) under different radiation beams.

2.1 Correction Factors Methodology for Diode Dosimetry

The diode sensitivity is defined as the ionization charge per unit radiation dose (usually in unit of nC/cGy). The sensitivity of semiconductor diode detectors depend on the construction of the diodes and the surrounding structures (including inherent buildup
material). The temperature [20,22-25], dose rate [20,22,26,30], and accumulated dose are caused by the die (bare diode) itself. The energy [20,21,31], field size, and directional dependency are mainly caused by the construction of the surrounding buildup structures. The absorbed dose measured at a particular point measured by the silicon diode detector can be calculated as:

$$D = \frac{M_{diode}}{S} \qquad (1)$$

M_{diode} is the charge collected in nC by the silicon diode detector and S is the total diode sensitivity. D is the absorbed dose at a particular point without the diode. One can also define the sensitivity of the bare diode, S_{diode}, as:

$$S_0 = M_{diode}/D_{diode} \qquad (2)$$

S_o is the ideal sensitivity with all correction factors taken out. D_{diode} is the dose measured with the silicon diode before all the correction factors are applied. Sensitivity of the diode detector, S, depends upon the temperature, dose rate, energy, field size, and beam angle i.e.

$$S = CF_{total} . S_0 \qquad (3)$$

And

$$CF_{total} = CF_{temp} x CF_{dose\ rate} x CF_{energy} x CF_{field\ size} x CF_{beam\ angle} x \ldots \ldots \ldots \qquad (4)$$

CF_{total} accounts for all the correction factors for the diode detector. CF_{temp}, $CF_{dose.rate}$, CF_{energy}, $CF_{field.size}$, and $CF_{beam.angle}$ are the correction factors for the diode detector for temperature, dose rate, energy, field size, and beam angle. Another term, S_{diode}, is the sensitivity of the bare diode defined in nC/cGy.

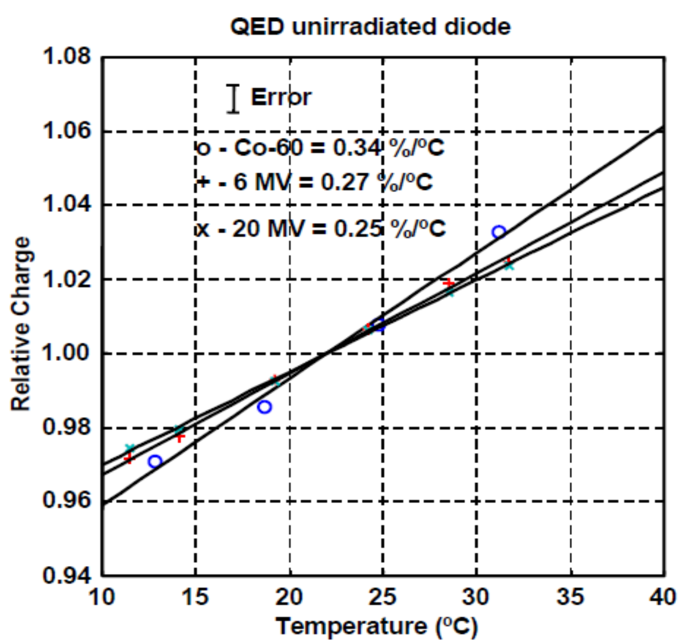
$$S_{diode} = CF_{temp} x CF_{dose\ rate} x S_0 \qquad (5)$$

2.٢ temperature Dependence of Commercially Available Diode Detectors

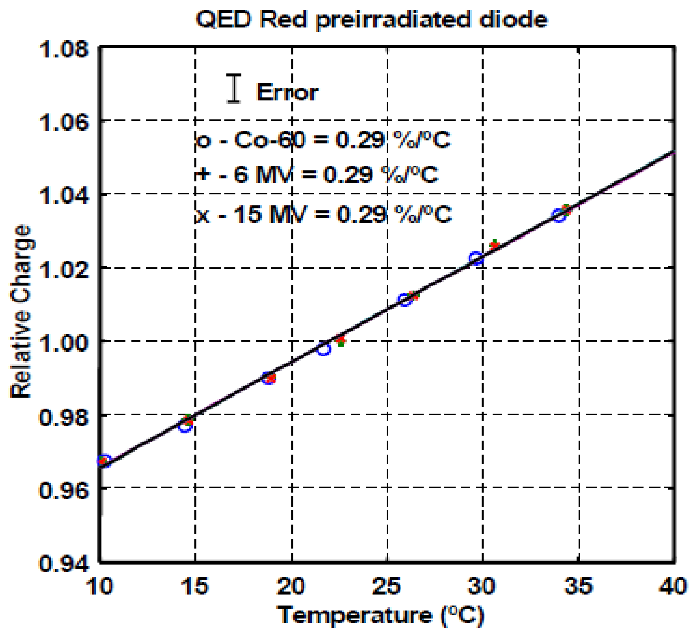
Temperature is an important factor in calculating the dose measured by the diode because when the diode is placed on the patient, its temperature can rise up to skin temperature within a few minutes. This can produce an inaccuracy in dose measurement up to 9% depending on the type of diode used, photon energy, dose rate, and pre-irradiation.

2.٣ Experiment Setup

The temperature dependence of the diodes was measured under both high instantaneous linear accelerator pulsed beam and low constant Co-60 dose rate radiations. Other than changing radiation machines, the same physical set-up was used throughout the study.The diodes were placed in a 30×30×30 cm^3 water phantom at a depth of 5 cm along the central axis of the beam, so the point of measurement was beyond the range of contamination electrons. The water surface was 100 cm from the x-ray target and the field size was set to 10×10 cm^2 at isocenter, defined as 100 cm from the x-ray source. The diodes were placed in a very thin rubber sleeve (0.33 mm thick) to make them waterproof.The temperature was cooled to 10°C using ice, and then the temperature was slowly increased from 10°C to 40°C at a rate of approximately 0.25°C/minute. After the water temperature was raised, the water pump was turned on to circulate the water for at least 10 to 15 minutes. Once the temperature in the phantom was completely stabilized, the radiation was applied. Normally one hundred-monitor units for pulsed radiation and one minute exposure for CO-60 was used. Three to four readings were taken and the average value was used for analyzing the data.



[Figure 1]



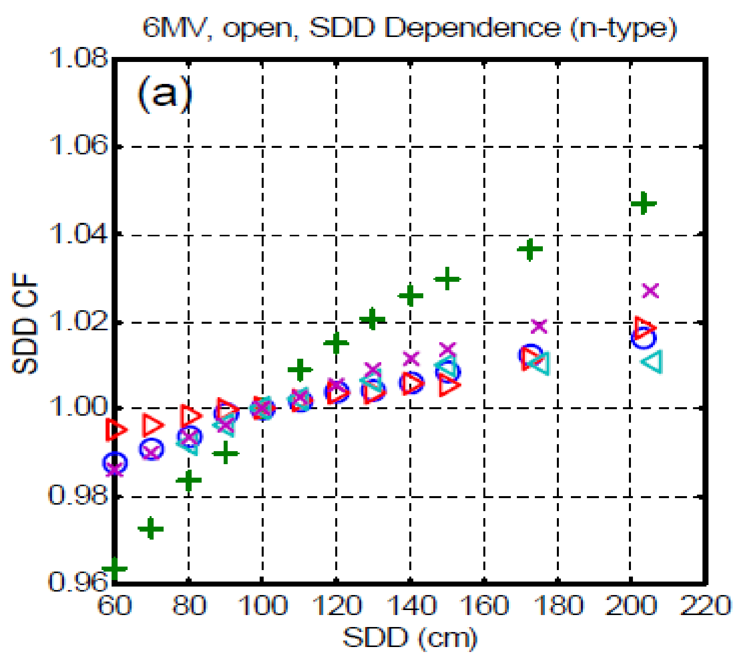
[Figure 2]

2.٤ Dose Rate and SDD Dependence of Commercially Avialable Diode Detectors

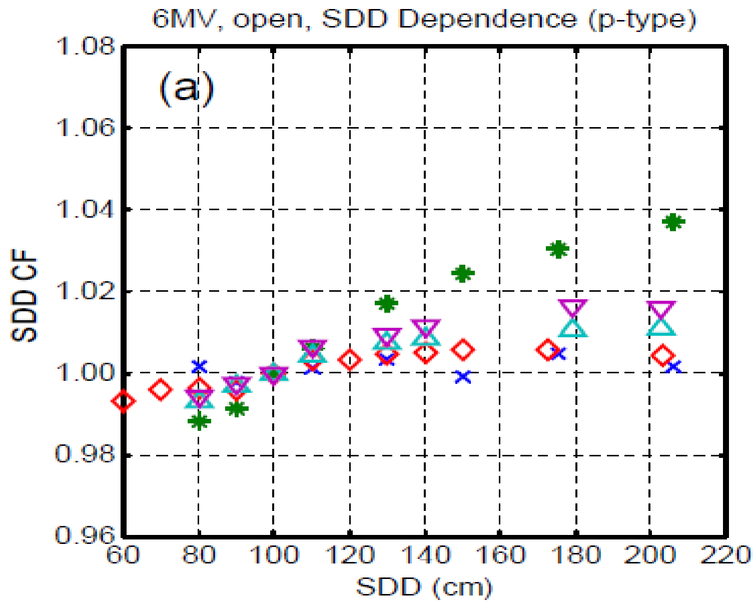
In clinical applications, dose rate dependence is the most essential dosimetric parameter for diode dosimetry since unlike ionization chamber ion recombination is inherited in a diode detector. As a result, the diode sensitivity changes greatly with the instantaneous dose rate especially for pulsed radiation beam. The dose rate could vary due to source– to-detector distance (SDD) changes, placement of transmission blocks or wedges, or transmission through the patient. SDD (or SSD (source-to-surface distance)) and dose rate dependence of diode sensitivity for different commercially available diode detectors were measured under high instantaneous dose rate (pulsed) and low dose rate (continuous) radiation. The dose rate dependence measured by adjusting radiation pulse height directly was compared to that measured by changing SDD.

2.◦ Experiment Setup

The diode SDD dependence was measured for the diodes with SDD ranging from at least 80 to 200 cm for the pulsed radiation and CO-60 radiation. All the measurements were taken by using the 10 x 10 cm^2 collimator setting. The diode was placed at 5-cm depth in a 4-cm diameter Lucite mini-phantom. A thimble-type ionization chamber was used for the inter-comparison, under the same geometric conditions. One hundred monitor units or one-minute time exposures were given for pulsed or CO-60 radiation respectively. Each set of data was completed within about 3 hours, with each diode or ionization chamber measurement completed within an hour, on the same day.



[Figure 3]



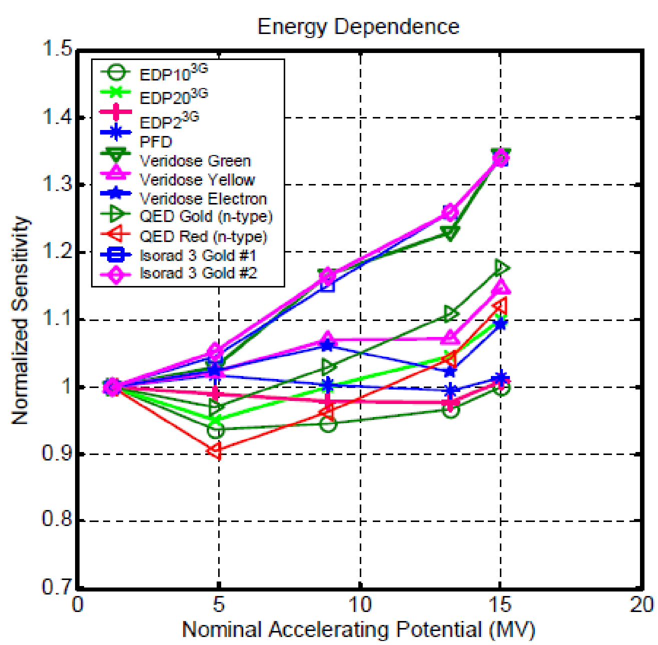
[Figure 4]

2.٦ Energy Dependence of Commercially Available Diode Detectors for In-Vivo Dosimetry

The diode detector used for radiation diode dosimetry is designed for specified energy range. Different high Z buildup materials are placed around the diode detectors, so that the dose is measured is close to the depth of maximum dose. Significant energy dependence is observed when the inherent buildup is too thick. Energy dependence of different commercially available diode detectors were measured for energies ranging between CO-60 and 20 MV pulsed radiation.

2.٧ Experimental Setup

The diode energy dependence was measured for the diodes using SDD (source-to-detector- distance) setup (SDD=100 cm) for the pulsed and CO-60 radiations. All the measurements were taken by using the 10 x 10 cm^2 collimator setting. The diode was placed at 5-cm depth in a full scatter liquid water phantom. A thin rubber protective sleeve was used to prevent water from reaching the diode. One hundred monitor units or one-minute time exposures were given for pulsed or CO-60 radiation respectively. Charges from all diodes were measured using an electrometer under zero-bias.



[Figure 5]

CHAPTER (3)
Results and Discussion

3. Results and Discussion

Prior to the implementation of diode in-vivo dosimetry, it is necessary to assess the effect of different parameters which affects the diode sensitivity. Generally, in clinic, the entrance dose is calculated from the different correction factors which affect the diode sensitivity.

3.١ Temperature Dependence of commercially available diode detectors

Temperature dependence of commercially available n- and p-type diodes were studied experimentally under both high instantaneous dose rate (pulsed) and continuous radiation. The sensitivity vs. temperature was measured at SSD = 80 or 100 cm, 10×10 cm^2, and 5-cm depth in a 30×30×30 cm3 water phantom between 10 and 35°C. The response was linear for all the diode detectors.

The temperature coefficient (or sensitivity variation with temperature, svwt) was dose rate independent for preirradiated diodes. They were (0.30±0.01)%/°C, (0.36±0.03)%/°C, and (0.29±0.08)%/°C for QED p-type. The temperature coefficient for unirradiated n-type diodes was different under low dose rate ((0.16 to 0.45)%/°C,34 continuous, cobalt) and high instantaneous dose rate ((0.07±0.02)%/°C, pulsed radiation).

Moreover, the temperature coefficient varies among individual diodes. Similarly, the temperature coefficient for a special unirradiated QED p-type diode was different under low dose rate (0.34%/°C, cobalt) and high instantaneous dose rate ((0.26±0.01)%/°C, pulsed radiation). Sufficient pre-irradiation can eliminate dose rate dependence of temperature coefficient.

In contrary, pre-irradiation cannot eliminate dose rate dependence of the diode sensitivity itself.

3.٢ Dose Rate and SDD Dependence of commercially available Diode

SDD CF was measured for commercial n-type diodes using pulsed radiation from an accelerator with low and high photon energies, respectively, in a miniphantom. The SDD CF of n-type diodes increased with decreased dose rate (by increasing SDD). Both unirradiated and pre-irradiated n-type Isorad diodes showed SDD dependence under pulsed radiation. For example, at SDD = 150 cm, the SDD CF, normalized to 100 cm, was 1.008 and 1.004 for n-type unirradiated Isorad Gold #1, 1.030 and 1.028 for n-type pre-irradiated Isorad Red (n-type), 1.014 and 1.014 for unirradiated QED Red (n-type), 1.005 and 1.005 for Isorad-3 Gold, and 1.010 and 1.011 for n-type pre-irradiated Veridose Green diode under pulsed radiation for low (6 MV) and high (18 MV) energies, respectively.

3.٣ Energy Dependence of commercially available Diode

The energy dependence was calculated by taking the percent difference, (maxmin)/min*100, between the maximum and minimum normalized sensitivity. It varied by 34% for Isorad-3 Gold #1 and #2, 35% for Veridose Green, 15% for Veridose Yellow, 9% for Veridose Electron, 21% for n-type QED Gold, 24% for n-type QED Red, 3% for EDP23G, 2% for PFD (Photon Field Detector), 7% for EDP103G, and 16% for EDP203G for nominal accelerating potential between CO-60 and 15 MV.

CHAPTER (4)
CONCLUTION

4.CONCLUTION

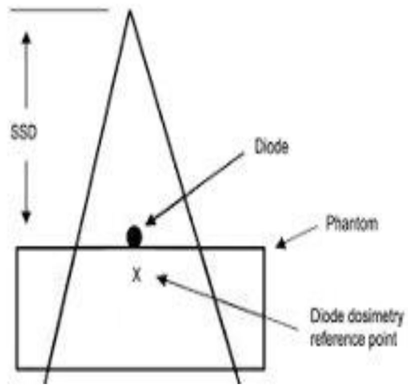
In order to achieve the best possible accuracy between the prescribed dose and the measured dose, proper correction factors for temperature, dose rate or SDD, and energy need to be applied to the diode measurements.

CHAPTER (5)

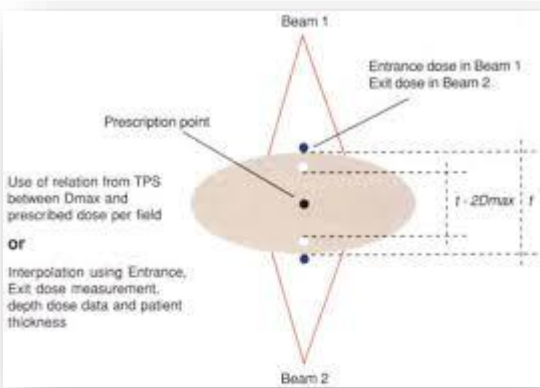
Applications

5. Applications

1. [Calibration Phantom]



2. [Entrance and Exit dose measurement]



3. [Diodes measurements placed on the patient's skin]

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الهدف من العمل

هناك العديد من العوامل التي تؤثر في استجابة (الصمام الثنائي diode) لقياس جرعات الإشعاع في الجسم الحي. من أجل تحقيق أفضل دقة ممكنة بين الجرعة الموصوفة والجرعة المحسوبة ،لا بد من ضبط مناسب حتى يُستعمل الصمام الثنائي للقياسات في الجسم الحي، وعوامل تصحيح مناسبه. وهذا يمكن أن يسمح لنا بالتعرف على الأخطاء البشرية بأكثر سهولة . في هذه الدراسة، تم إجراء دراسة منهجية لخصائص قياس الجرعة dosimetric من بعض إشعاع الثنائيات diode (نوع n ونوع p) المطبقه ، وقد وجد أن قراءة الديود تتغير بتغير درجة الحرارة، و مع معدل الجرعة، وايضا مع المسافة من المصدر SDD إلى الديود، و مع الطاقه المستخدمة. الدراسة يمكن أن تساعدنا أكثر على تقدير أفضل لخصائص مقياس الجرعات لأجهزة الكشف بالصمام الثنائي المختلفة والحصول على قدر أكبر من الدقة في قياس جرعة الجسم الحي في العيادة.