Sudan University of Science and Technology
College of Graduate Studies

Verification and Monitoring of Radiation using In-vivo and In-vitro Dosimetry in External Beam Radiotherapy in Libya

مطابقة ومراقبة الجرعة الاشعاعية باستخدام القياسات
داخل الجسم الحي وفي المختبر اثناء العلاج
بالاشعة الخارجية في ليبيا

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DEDICATION

To

My Father

To

My Mother

To

My Wife

To My Son, My **Daughter**

Abstract

Radiation therapy has been used in the treatment of cancer patients since its discovery in 1895, and these techniques were always in continuous scientific and technical development. It was necessary to introduce diode in-vivo dosimetry for radiation measurements to control the dose administered to patients. The patient simulating device, planning system and treatment machine are tested regularly according to set protocols developed by national and international organizations. Even though these individual systems are tested for errors which can be made in the transfer between the systems. The best quality assurance for the system is at the end of the treatment planning chain. In vivo dosimetry is used as a quality assurance tool for verifying dosimetry as either the entrance or exit surface of the patient undergoing external beam radiotherapy. It is a proven reliable method of checking overall treatment accuracy, allowing verification of dosimetry and dose calculation as well as patient treatment setup. The aim of the study was to calculate entrance dose obtained by the treatment planning system with measured dose using diode detectors and discover discrepancies lager than $\pm 5\%$ between the calculated dose and the measured. Calibration of the diode was done using Cobalt-60 teletherapy machine, linearity and calibration factors were determined. Measurements were preformed in Perspex phantom for calibration procedure. In vivo dosimetry represents a technique that has been widely employed to evaluate the entrance/exit dose to the patient mainly in radiotherapy. The analysis of all available measurements gave a mean percent deviation of 0.91% and average discrepancy 3.178 ± 0.507 ($\pm 15.97\%$). A great majority of measurements were found within the acceptable limit$\pm 5\%$. Diode dosimeters are considered the best methods for in vivo dosimetry is simple, cost effective, provides immediate results and is a useful quality assurance tool for verification of absorbed dose delivered during patient treatment on Co-60 machine, also diode need to be calibrated against an accurate dosimetric reference, such as an ionization chamber, to determine the calibration factor and we conclude that using diode for in vivo dosimetry requires careful attention.

الخلاصه

تـم اسـتخدام الاشـعاع فـي عـلاج مرضـى الاورام السـرطانية منـذ اكتشـافه عـام 1895 وهـذه التقنيـات في تطـور علمـي وتقنـي دائـم حيـث اصبـح مـن الضـروري ادخـال قيـاس ومراقبـة الجرعـات المعطـاة للمريض باستخدام الصمام الثنائي .ويتم اختبـار جهـاز محاكـاة المريـض ونظـام التخطيـط وآلـة العـلاج بانتظـام وفقًـا للبروتوكـولات المحـددة التـي طورتهـا المنظمـات الوطنيـة والدوليـة ويتـم اختبـار هـذه الأنظمـة بصـورة فرديـة بحثًـا عـن الأخطـاء التـي يمكـن أن تحـدث فـي نقـل المعلومـات بيـن الأنظمـة. القياسـات الداخليـة تسـتخدم كـاداة لضمـان الجـودة والتحقـق مـن الجرعـات الاشـعاعية امـا علـى مـدخل او مخـرج الاشـعاع للمريـض الـذي يخضـع للعـلاج بالاشـعاع الخـارجي وهواسـلوب موثـوق منـه للتحقـق مـن دقـة المعالجـة الاشـعاعية ممـا يتيـح التحقـق مـن حسـاب وقيـاس الجرعـات وكذلـك اعـداد المريـض للعـلاج وتـم قيـاس الجرعـات فـي الجسـم الحـي عـن طريـق الصـمامات الثنائيـة ومقـاييس كواشـف الـوميض الحـراري وعنـد اسـتخدام هـذه الانـواع فـي قيـاس الجرعـات فـي الجسـم الحـي يجـب عمـل معـايرة لهـذه الاجهـزة وتوخـذ العوامـل التـي تـوثر علـى حساسـية الاسـتجابة فـي الاعتبـار. الهـدف مـن الدراسـة هـو حسـاب جرعـة الدخـول التـي تـم الحصـول عليهـا مـن خـلال نظـام تخطيـط العـلاج بجرعـة مُقاسـة باسـتخدام كاشـفات الصـمام الثنـائي واكتشـاف تباينـات أكبـر مـن 5٪ بيـن الجرعـة المحسـوبة والجرعـة المقاسـة. الاجهـزة المسـتخدمة: تمـت معـايرة الصـمام الثنـائي باسـتخدام جهـاز المعالجـة الاشـعاعية عـن بعـد الكوبالـت -60 وتـم تحديـد عوامـل المعـايرة الخطيـة للصـمامات الثنائيـة ويمثـل قيـاس الجرعـات فـي الجسـم الحـي تقنيـة تـم اسـتخدامها علـى نطـاق واسـع لحسـاب الجرعـة التـي حصـل عليهـا مـن مـدخل / خـروج الاشـعاع للمريـض بشـكل رئيسـي فـي العـلاج الإشـعاعي.تحليـل جميـع القياسـات المتاحـة اعطـى متوسـط انحـراف بنسـبة 0.91% ومتوسـط تبايـن 0.507 ± 3.178 (15.97%±).الغالبيـة العظمـى مـن القياسـات ضمـن الحـد المقبـول 5% ± . تعتبـر مقاييس الجرعـات باسـتخدام الصـمام الثنـائي هـي أفضـل الطـرق لقيـاس الجرعـات فـي الجسـم الحـي ، فهـي بسـيطة وفعالـة مـن حيـث التكلفـة وتـوفر نتـائج فوريـة وهـي أداة مفيـدة لضمـان الجـودة للتحقـق مـن الجرعـة الممتصـة التـي يتم تسـليمها للمريـض أثنـاء العـلاج، كمـا يجب معـايرة الصـمام الثنـائي مقابـل مرجـع دقيق لقيـاس الجرعـات ، مثـل غرفـة التأيـن ، لتحديـد عامـل المعـايرة. نسـتنتج أن اسـتخدام الصـمامات الثنائيـة لحسـاب الجرعة في الجسم الحي يتطلب معالجة دقيقة.

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LIST OF ABBREVIATIONS

TLDs	Thermoluminescent dosimeters
IVD	In- vivo dosimetry
QA	Quality assurance
ICRU	International Commission on Radiation Units
MeV	Mega electron-volt
RT	Radiation Therapy (Radiotherapy)
NCIS	National Cancer Insatiate Sabratha
Ref	Reference
IMRT	Intensity modulated radiotherapy
WHO	World health organization
TPS	Treatment Planning System
SSD	Source to surface distance
SAD	Source to axis distance
PDD	Percentage depth dose
MU	Monitor units
TD	Target dose
cGy	Absorbed dose unit
CM	Centimeter
TMR	Tissue maximum ratio
ID	Isocenter dose
SCD	Chamber source to chamber distance
μm	Micro meter
SD	Standard deviation
IC	Ionization chamber
EBRT	External Beam Radiotherapy
F_{cal}	Factor of calculation
CT	Computer tomography
CF	Calibration factor
Sp	Phantom scatter factor
FS	Scatter factor
QED	Type of diode detector use for electron beam therapy
CSSD	Non-reference conditions
CFS	Reference conditions
DCF	Diode calibration factor
DOF	Output factor
SSF	Surface-scatter-factor

DWF	Surface-scatter-factor
TBF	Block tray factor
IEAE	International Atomic Energy Agency
H& N	Head and Neck
LL	Left lateral
RT	Right lateral
NY	New York
USA	American United State
TRS	Calibration Protocol An International Code of Practice for Dosimetry Based on Standards of Absorbed Dose to Water.

CHAPTER ONE

INTRODUCTION

1.1 Preface

The ultimate overall goal of radiotherapy is to deliver a specified radiation dose to the prescribed target volume with the least dose to healthy tissues. This means a sophisticated balance between the cure of the illness and the possibility of radiation induced complications. Therefore the demands for precision and accuracy in radiotherapy are high, because very often a small increase in radiation dose will have crucial influence on the probability of a cure but simultaneously the probability of induction of irreversible damage to the patient will increase. An "error" is any deviation between the given numerical value of a quantity, such as the dose at a point or the position of a point, and its "true" value. In radiotherapy, errors may arise from at least four main sources:

(i).Human mistakes caused by inattention, misunderstanding or misjudgment;

(ii).Instrumental mistakes caused by mechanical or electrical failure;

(iii).Random errors due to unknown and/or uncontrolled experimental conditions in the process involved in the planning and delivery of radiation; and

(iv).Systematic errors, i.e. biases, in the same set of processes. In the following discussion, mistakes will be considered separately from the random and systematic errors. In principle, mistakes can be eliminated completely by a proper system of cross-checks of both human and instrument performance (by quality assurance system), although, in practice this may prove very difficult and expensive. Random and systematic errors, on the other hand, cannot be eliminated but the magnitude of these uncertainties can be reduced by accumulation of better data and improved techniques of measurements and delivery of radiation (by improved quality control of all steps of radiotherapy process) (IAEA, 1997) . Regarding radiation safety, errors or poor performance in diagnosis can lead to a higher collective dose than necessary, leading to undue radiation detriment to the population.

Errors or poor performance in radiotherapy can lead to severe consequences to patients, hospital staff and general public which are different from radiological accidents in industrial irradiation facilities where only the last two groups of people can be involved.

The full benefit of radiotherapy treatment of cancer can only be achieved if the radiation doses to patients are accurate and reproducible. There are two fundamentally different but equally vital requirements for achieving this:-

Firstly: Accuracy and precision can be achieved by high quality measurements of the treatment beams and careful calculation of doses to target volumes, supported by a good preventive maintenance programme for the equipment, i.e. well implemented quality assurance programme.

Secondly: It is necessary to prevent a wide range of simple errors, which compromise safety. This second requirement has not always been acknowledged but its importance may be demonstrated by accidents at busy radiotherapy centers. Even if all recommendations for quality assurance, local rules and practical guidelines are followed the occurrence of misadministration and accidents in radiotherapy departments are still very common (IAEA, 1997).

1.2 Problem of the study

Radiation therapy (RT) is a complex and rapidly advancing technology that is used to treat cancer. It utilizes ionizing radiation to effectively kill cancer cells leading to cure or symptomatic relief. If delivered incorrectly the radiation can have debilitating side effects and even result in death. As such it is imperative that we are able to accurately measure that the dose is being delivered as intended. Therefore measuring of dose entrance at the point of each delivery (in vivo dosimetry) is considered the gold standard for dose verification. Following serious radiation incidents the use of in vivo dosimetry has become an objective for all international departments and mandatory in some countries.

1.3 General objective

The objective of this study was to calculate entrance dose obtained by the treatment planning system with measured dose using diode detectors and discover discrepancies lager than ±5% between the calculated dose and the measured dose. To achieve this goal, the delivery dose verification program was initiated by using the diode in vivo dosimetry (IVD) system for entrance dose.

1.3.1 Specific objectives

- To check the diode IVD system to measure the entrance dose in radiation therapy for head and neck, pelvic and breast malignancies during treatment and its implementation as a patient-specific QA tool for the verification of the dose delivery.

1.4 Significance of Study

Discovering and treating errors before radiation therapy process, so that it does not affect the quality of radiotherapy. Also the importance of the study lies in improving the safety of cancer patients during radiation therapy using in vivo dosimetry to check the radiation dose received by the patients. However, the use of IVD may prevent a serious radiation incident. And also provide a better understanding of the performance of diodes used in radiation therapy departments can use as national quality assurance protocol in future.

1.5 Thesis outline

The research is divided into five chapters. Chapter's one introduction includes the preface, problem of the study, general objective, specific objective and significance of study. Literature review is given in chapter two. A description of the material and method is given in chapter three. In chapter four results were presented. Finally In chapter five the discussion, conclusion and recommendations were given.

CHAPTER TWO

LITERATURE REVIEW

2.1 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 (Fonten *et al*, (1996). 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 Howie *et al*., (1999). The breakthrough of in vivo dosimetry occurred at the end of the sixties, when thermoluminescent dosimeters (TLDs) became available and more recently when semiconductor detectors were introduced as radiation dosimeters. For most of the in vivo dosimetry (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). In vivo dosimetry is the most direct method for monitoring the dose delivered to the patient receiving radiation therapy. It allows comparison of prescribed and delivered doses and thus provides a level of radiotherapy quality assurance that supplements port films and computational double checks. When performed early in treatment as a supplement to the clinical quality assurance (QA) program, simple in-vivo measurements are an additional safeguard against major setup errors and calculation or transcription errors that were missed during pre-treatment chart check (AAPM Report, 2005). In (ICRU report 24, 1976) it is also specified what in vivo dosimetry might include: Entrance dose measurements, exit dose measurements, transmission measurements and 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 variation of the body of the patient on the dose calculation procedure.

2.2 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 like diode and Metal Oxide Semiconductor Field Effect Transistor (MOSFETs), but are not yet in routine clinical use. In this study diode were used because it a valuable in Radiation NSIC.

2.3 Diode dosimetry systems

The diode is a good relative dosimeter for in-vivo dosimetry because it exhibits certain characteristics. Compared to ionization chamber, the diode has the advantages of high sensitivity (charge collected per unit dose to the diode) and quick response time. Other major advantages of semiconductor detectors are excellent reproducibility, good mechanical stability, absence of external bias, small size, and the energy independence of mass collision stopping power ratios (between silicon and water for clinically usable electron beams with energy between 4-20 MeV) (Rinker. G, Grusel. E, (1983). The real-time in-vivo dosimetry allows one to check the prescribed dose for dynamic beam immediately and make it possible to correct the treatment errors interactively Rinker.G, Grusel. E,1983, (1987).

Semiconductor diodes as detectors for in vivo dosimetry are considered as very useful tool in clinical practice. Their main advantage over other detectors such as TLDs is a possibility of immediate read out and detection of errors while a patient is still on a treatment cough. Moreover, diodes are known for their high sensitivity, small size, simplicity of operation and mechanical stability Loncol. T, Vynkier. S, (1969).

2.4 Patient dose verification in external beam therapy

The first paper that introduced the silicon diode detectors into radiotherapy is Ref Jones. A. R, (1963). In recent years, encouraged by the work of Riker. G, Grusell.E, (1987) the use of semiconductor diode detectors for in vivo dosimetry has been extensively investigated. The large number of steps and persons involved leads to a large probability of errors in the preparation and execution of a radiation treatment. Different ways can be used to assess the uncertainty in the dose delivery. The sequence of the different procedures involved in delivering a dose to a patient (dose calculation, treatment machine calibration, patient setup etc.) can be analyzed. The total uncertainty in all the steps may then be considered as the maximum attainable accuracy in dose delivery. In practice the actual accuracy in dose delivery may be less than desirable for any of the following reasons:' Errors in patient contours, patient mobility, in homogeneities, organ motion, transfer of treatment data from simulator to the treatment unit, machine settings, and positioning of modifiers, etc. The ultimate check of the individual dose delivered to a patient can only be performed at the patient level, by means of in vivo dosimetry. The method most commonly used for standard techniques consists of positioning point detectors on the patient's skin and measuring the entrance and/or exit doses in conditions adapted to the type and energy of the beam. By combination of the two, it is then possible to obtain the dose inside the patient (target or midline dose, for instance). The detectors usually employed are diodes and thermoluminescence dosimeters. Some authors have also used portal films or electronic portal imaging devices. Although more demanding to carry out, in vivo dosimetry can be used to detect errors or to check the dose delivered using intensity modulated therapy. Depending upon the objectives to be achieved dosimeters can either be put at the entrance of the different beams at points situated in a low dose gradient area or put within the target volume using, for example, a nasogastric tube. When this is not possible, in vivo measurements done in high dose gradient regions are subject to question Mayles. P, (2007).

2.5 In vivo dosimetry

In vivo dosimetry (IVD) began to be used at the same time as radiotherapy when skin erythema was the only form of dosimetry available. More than 30 years ago in vivo measurements were also recommended for brachytherapy of cancer of the uterine cervix because it was not possible to calculate the dose delivered to the bladder and rectum, which were limiting factors for the irradiation. Currently, in vivo dosimetry is most often considered a quality assurance tool useful to identify deviations in the delivery of standard or complex treatments such as intensity modulated radiotherapy (IMRT), to evaluate the dose to critical structures (lens, gonads, etc.) or when computer calculations are not possible or are questionable (limits of block shielding, junction of non-coplanar irradiation fields, dose at skin or within in homogeneities, etc.). In vivo dosimetry (1VD) can also be used to monitor the irradiation for special techniques such as total body irradiation or total skin electron irradiation etc. Dosimetric investigations in special or anthropomorphic phantoms loaded with dosimeters and irradiated in the same conditions as patients can also be useful to check the validity of special techniques prior to routine practice, to point out problems related to suboptimal treatment planning systems, errors in irradiation technique or in dose calculations, or simply to validate the method used for in vivo dosimetry Mayles.P(2007).

The international commission of radiation units and measurements (ICRU) recommends that the dose be delivered within 5% of the prescribed dose. This means that the end of the planning and treatment "chain" the total error in dose delivered is less than 5%. Each stage in the planning process has an inherent error, so it is therefore to meet this requirement .As in vivo dosimetry (IVD) is at end of the planning chain any error made in treatment chain (such as patient position ,calculation , accessory in extremely useful in the detection of any error along the chain it is underused , and should be put into practice more often there should be feedback in all step in radiotherapy chain as any problems at one point will require a change at another point also recommend that in situations where higher than normal dose is given IVD is desirable and portal image is essential. In vivo dosimetry (IVD) is seldom used to estimate dose to the tumor volume, despite the fact that it was recommendation of world health organization (WHO).

Work done by the leavens group has found that considerable benefits can be achieved by the implementation of regular IVD. And that errors that would otherwise have

been missed have been found in the other hand, the inherent error in IVD make it difficult to identify source of error. It is therefore important to set realistic error boundaries so that time is not wasted looking into errors in the measurement chain in that do not exist. IVD is measure dose given to patient. Errors in the treatment chain have been found that could have been detected by phantom measurement, while some errors have been found that could only have been measured with IVD on the patient, any new technique should be checked on phantom before it is implemented clinically. Mortor.J, (2006). In the absence of errors, routine in vivo measurements uniquely document that treatment was delivered correctly within a user-specified tolerance. Unlike other QA methods, in vivo dosimetry checks the dose delivered to the patient rather than the individual components prior to treatment. Most treatments are without serious error—in a recent review from Europe, out of 10,300 patients at three institutions performing in vivo dosimetry for all new patients, 120 treatment errors exceeding 5% were found, and the estimated serious error (misadministration) rate in the United States is 0.002%. Although there is not universal agreement on the benefit of in vivo dosimetry, a strong argument in its favor is that preventing the severe consequences of major errors as illustrated by the recent overexposure of 28 patients in Panama—warrants the effort and expense of an in vivo dosimetry program. IVD is also helpful in supporting the high accuracy in dose delivery expected from complex and conformal therapy techniques AAPM. No, (2005).

2.6 Clinical application of in-vivo dosimetry

A first possible aim of in IVD is to compare the doses derived from the signal of the detectors placed on the skin with the theoretical values, as calculated by the Treatment Planning System (TPS). As however the accuracy of the calculation of the dose to the skin is questionable, and in many cases irrelevant, the signal of the detector is converted to the dose, at a point which is still close to the skin, but at a certain depth where the accuracy of the TPS is much more satisfactory. One point is close to the entrance, while the other is close to the exit surface of the beam. The corresponding doses are called entrance and exit doses, respectively. With regard to the exit dose, one should realize that in the real patient there is in most cases a considerable loss of backscatter, while the TPS calculations are valid for semi-infinite patients implying complete backscatter at the exit surface. A correction is then necessary Van. D, Rosenwled. J, (2006). A more ambitious aim of in vivo dosimetry is to check the target dose, in order to verify the correct delivery of irradiation. Except when detectors can be introduced in natural body cavities such as esophageal tube, rectum, vagina, etc, this is impossible. As a matter of fact, a check of the entrance and exit dose is also an indirect check of the target dose. However, if a deviation is observed between the computed and measured entrance or exit dose (under the assumption that the experimental value is correct) it may be because the target dose is wrong (due to a wrong in time, an error in the irradiation parameters, an incorrect patient set-up or an unexpected variation of the machine output), because the calculation of the entrance or exit doses, even from a correct target dose, is wrong, or because of a combination of both types of error. A more selective check of the target dose is then of high interest. A third possible aim of in vivo dosimetry can be the determination of the skin dose itself. This measurement is critical and requires a special methodology Van. D, Rosenwled. J, (2006).

2.7 Dose calculation

Several methods are available for calculating absorbed dose in a medium. Two methods commonly used are the source to surface distance (SSD) and the source to axis distance (SAD). The two have their advantages and disadvantages relative to each other.

2.7.1 The SSD method

In this method, the percentage depth dose (PDD) is used in the calculation of monitor units (MU) utilized in delivering the required dose. The PDD is defined according to Rinker. G , Grusell, (1983), as the absorbed dose at any depth, d to the absorbed dose at a reference depth, along the central axis of the beam.

$$PDD = \frac{TD \times 100}{D_{max}} \qquad (2.1)$$

Where TD is the absorbed dose at depth, d and is the dose at a reference depth. The monitor unit is defined as the time during which a particular amount of dose is delivered to a point. The high energy x-ray machines are calibrated to deliver 1cGy/MU at a reference depth of for a reference field size of 10 cm x 10 cm. The monitor unit necessary to deliver a target dose, *(TD)* at a depth, d for a field size r at the surface at SSD = 100 cm are given by:

$$MU = \frac{TD}{K \times PDD \times S_c(r) \times S_p\ (r) \times (SSD\ Factor)^2} \qquad (2.2)$$

where $k = 0.01$ Gy/MU, r is the collimator field size, is a factor related to field size at source axis distance, *(SAD)* and is a factor related to the field treating the patient. In this study it is necessary to measure the maximum dose for calculated monitor units. The maximum dose will be measured by the diodes. The theoretical maximum dose can be obtained by using the definition of PDD given in equation 2.3 and rearranging it gives the following equation:

$$D_{max} = \frac{TD \times 100}{PDD} \qquad (2.3)$$

2.7.2 The SAD method

In this method, also known as the isocentric method, the tissue maximum ratio *(TMR)* is used for the dosimetric calculation. The *MU* necessary to deliver Isocenter dose, *(ID)* at depth, *d* is given by:

$$MU = \frac{ID}{k \times TMR(dr_d) \times S_c\ (r) \times S_p\ (r) \times (SAD\ factor)} \qquad (2.4)$$

$$TMR(d, 0) = e^{-\mu(d - t_0\ )} \qquad (2.5)$$

Where is the effective linear attenuation coefficient and is the reference depth of maximum dose, and Where *SCD* is the chamber source to chamber distance.

$$SAD\ factor = \left(\frac{SCD}{SAD}\right)^2 \qquad (2.6)$$

The calculated doses can be complemented by direct measurement using irradiation measurement instruments Khan. F. M , (1994).

2.8 Radiation measurement devices

The radiation measurement devices also called dosimeters are instruments or systems that measure or evaluate either directly or indirectly the quantities highlighted in, which include kerma, absorbed dose and dose equivalence. Solid state dosimetry is the area of focus of this project though there are several other types of dosimeters available commercially. The solid state detectors can only be used as relative dosimeters because they need to be calibrated against a known standard for them to be used effectively.

2.9 Solid state devices

There are two types of solid state dosimeters:

(a) Integrating type dosimeters (thermoluminescent crystals, radiophoto-luminescent glass, optical density type dosimeters such as glass and film); and (b) Electrical conductivity dosimeters (semiconductor junction detectors, induced conductivity in insulating material). Films, thermoluminescent dosimeters (TLDs) and diodes are commonly used.

2.9.1 Silicon diode

Silicon p-n junction diodes are used for dosimetry and have been used as radiation detectors for over 30 years Dutreix. A, (1984); Chair. E, et al (2005). The physics of charge generation and collection in silicon semiconductor diodes provide characteristic features, which make them useful as radiation detectors. A cross section of a 6 MeV Isorad Sun Nuclear diode used in radiation measurement is shown in figure 2.2.
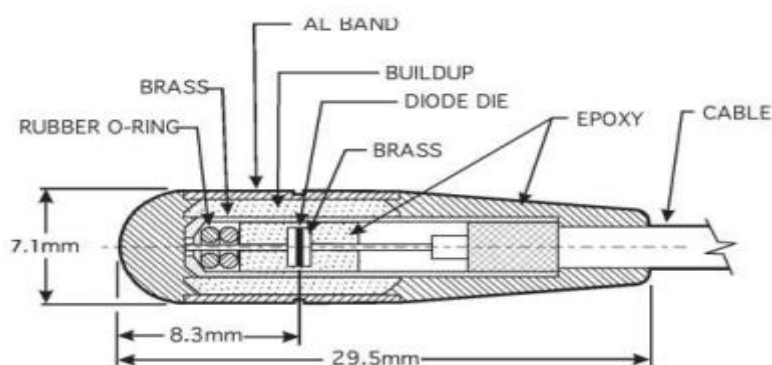
Figure 2.1 Cross sectional view of a diode used for in-vivo dosimetry.

The silicon die with an active detection thickness of 15 μm (Users Guide, (2005) is the location of electron hole pair generation caused by radiation York et al, (2005) describes the process of charge generation and measurement as the following: (a) Primary or secondary particles from the radiation source are absorbed thus generating electron hole pairs throughout the diode. (b) The electrons and holes generated within one diffusion length from the junction are able to reach the p-n junction. (c) The excess minority carriers (electrons on the p side and holes on the n side) are swept to the opposite sides of the build-in potential across the p-n junction giving rise to a pulse in the external circuit. (d) When the diode terminals are connected to the input of an operational amplifier, the charges generated by the irradiation are collected, amplified, measured and converted to dose. Diode detector sensitivity, *S* is approximately proportional to the minority carrier diffusion length, *L* and is given by:

$$S = aL = a\sqrt{\tau D} \quad (2.7)$$

Where = constant, = lifetime of radiation generated excess carriers, and *D* = minority carrier diffusion coefficient. The change in sensitivity of the diode affects the accuracy of the measurements hence the need for correction factors. Most investigations Rickener.G, Grusell. E, (1987); Sze. S. M, 1969; Mangili. *et al.,* (2000); Strojnik. A, (2007); Huyskens. *et al* ., (2001); Meiler.R, Podgorsak. M, (1997) for diode in-vivo dosimetry were done for detectors placed on a flat surface

2.10 Theory of Silicon diodes

Silicon diodes have very small dimensions hence the sensitivity relative to the ionization volume is high. The sensitive volume is small and well defined, the effective point of measurement can be placed less than 1 mm below the outer surface of the detector. Recent silicon detectors are surrounded by water equivalent material and special care is exercised to optimize their performance in radiation dosimetry to reduce the interface phenomena. The electrical properties (conductivity) of a semiconductor material can be changed by introducing impurities into the crystal also called doping. The n-type silicon is obtained by doping silicon with Group V (P, As, or Sb) elements called donors and p-type silicon by doping with Group III (B, Al, Ga, or in) elements called acceptors. The p-type silicon is joined with an n-type material to obtain a p-n junction.

2.10.1 P-N junction

A p-n junction is an internal boundary between the p-type and n-type regions in a single crystal. The n-type material has a large concentration of electrons and few holes, while the p-type material has a large concentration of holes and few electrons. When these two regions are joined together, diffusion of charge carriers takes place because of the large gradient of carrier concentration at the junction. Electrons in the n side diffuse to the p side, and holes in the p side diffuse to the n side due to the gradient. Holes diffusing from p region leave uncompensated acceptors while electrons diffusing from n to p region leaves behind uncompensated donor ions in the n region. There is positive space charge near the n side, and negative space charge near the p side of the material. The charged ions left on both sides form a depletion region (space charge) over which a built-in voltage drop of about 0.7 V is created over a distance of few micrometers Sze. S. M, (1969); Chair. E, (2005); User Guide, (2005); Meiler. R *et al* (1997). The p-n junction is formed when equilibrium is reached. There is no net current flow across the junction at equilibrium until it is irradiated. The current signal is due to charge carriers created in the depleted region and the minority carriers created in the base material that diffuse to the depletion region.

The thickness of the effective volume is determined by the lifetime of the carrier and is due to recombination centers and traps in the crystal. The traps consist of imperfection in the crystal lattice Meiler. *et al* (1997). Identified and described the following:-

(a) Type of doping;

(b) Doping level;

(c) Pre-irradiation level;

(d) Mechanical construction;

(e) Detector volume;

(f) Leakage current and connection to the connector, as parameters which also influence the behavior of the detectors.
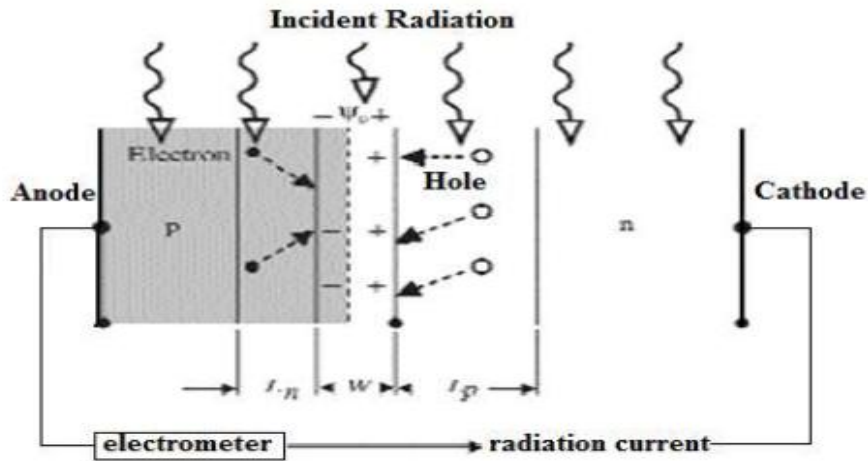
Figure 2.2 Schematics of a silicon p-n junction diode as a radiation detector.

A pictorial summary of the recombination process is as shown in figure 2.4
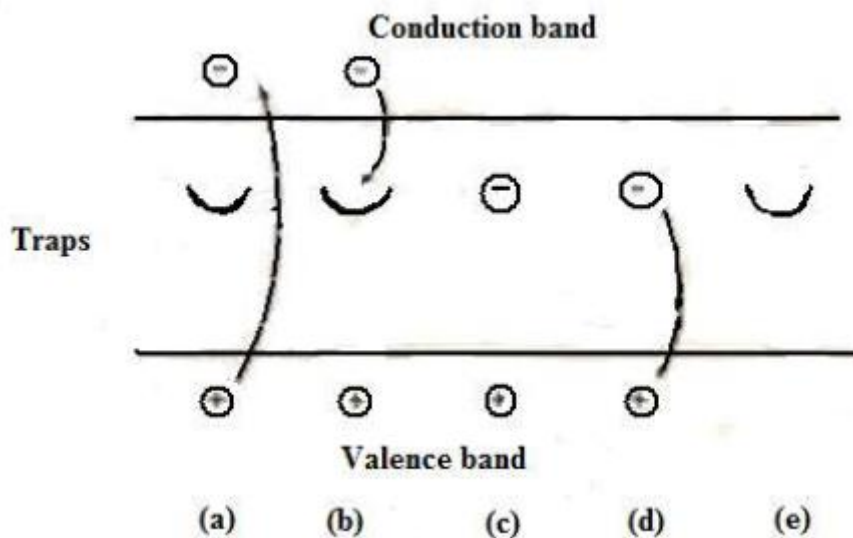
Figure 2.3 Recombination by traps in p-type silicon. When irradiated, electron–hole pairs are created. Minority carriers will move in the conduction band (a) until they are trapped (b) and occupy the traps (c) before they recombine with majority carriers (d). In (e) recombination is complete

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2.11 Electrometer

The diode should be connected to a dedicated electrometer with low input impedance and low offset voltage. Diode current generated by source other than radiation is considered to be leakage current and it's not desirable. The leakage current ideally should be zero. Due to the input offset voltage of the amplifier, However, there is always a small bias a cross the diode introducing a small leakage current. An electrometer used together with a diode therefore the offset voltage of the amplifier to be low, 10μ or less. The leakage current increases with temperature and accumulated dose due to detects in the diode and its essential that the electrometer has adequate zero drift and compensation and stabilization.

2.12 Software

There is a range of electrometer for in vivo dosimetry has greater of lesser degree of sophistication. The simplest type of electrometer provides 5 to 10 channels with manual adjustment of the input offset and gain for each channel. This type of electrometer may allow only one gain setting for each channel. Thus one detector may calibrate to be used in several different irradiation conditions. Most of the electrometers offer the possibility to use interface software designed to run in widow's environment in conjunction with commercial available software or house made program loaded onto a personal computer. More advanced systems are incorporated with the department verifications simplifying the management system of the in vivo dosimetry procedure. Such system provides the possibility to store all calibration and correction factors for every diode in use. The measured diode signal is than automatically converted to dose using the treatment field parameter download from the patient's data in the verification system. This gives an immediate on line check of the preparation and treatment delivery in the radiotherapy process, thereby reducing the incidence of errors.

2.13 Use of Diodes in the in vivo dosimetry

Diodes are useful in radiation dosimetry because of their high radiation sensitivity relative to the ionization volume. Therefore, the measuring volume can be very small, leading to good spatial resolution. Semiconductor diodes offer many advantages for clinical dosimetry: high sensitivity, real time readout, simple instrumentation, robustness and air pressure independence. However, diodes are subject to influence from a number of factors, including temperature dependence and, for a rigorous system, consideration needs to be given to all of these European Society, (2004); AAPM, WI,(2005)

2.14 Validation before use

The signal stability of the diode influenced e.g by the leakage current without irradiation, should be checked after adequate warm-up time with the diode connected to the electrometer and compensated. Compared to the current obtained for the real measurement, the leakage current should be insignificant. It's advisable to measure the leakage current for a time period that is at last five times longer than the time period used in the clinical application. The leakage current should not exceed 1%.in one hour. A general test of the reialiblity and stability of the equipment before using it in clinical routine can be performed as follows. The diode positioned on top of a calibration phantom is irradiated for 10 -15 times with same reference field size, the SD of the resulting signals should be within 5% the measurements are repeated on different days during the weeks. The measurement procedure, including the measurement equipments the phantom set-up and diode positioning is reliable and stable if all measurements are within 1% (provided that the beam out-put of the treatment unit is stable).

2.15 Calibration of Diodes

In vivo dosimetry can be divided into real-time and passive detectors that need some finite time following irradiation for their analysis. Both types of dosimetry require a calibration generally obtained by comparing their response ageist a calibrated ionization chamber (IC) in a known radiation field. Most of these detectors have a response that is energy and/or dose rate dependent and consequently require adjustments of the response to account for changes in the actual radiation conditions compared to the calibration situation. Correction factors are therefore necessary to take, for instance, changes in field size, source-detector distance, temperature, pressure, and orientation, including the presence of a build-up cap, into account. The presence of a build-up cap is important for detectors used for entrance or exit IVD. Special attention should be paid to the selection of appropriate build-up cap material and thickness for entrance dose measurements during EBRT because the dose beneath the dosimeter may be significantly attenuated by the dosimeter build-up material for that reason entrance dose measurements are often limited to a few fractions . Furthermore, the material and thickness of the build-up cap has an effect on the magnitude of correction factors when moving away from reference conditions. However, these corrections are sometimes ignored when using commercially available detectors for patient measurements under conditions different from those used for their calibration, thus increasing the uncertainty in the measurement. It is worth mentioning that some detectors can be use both as real-time and passive detectors, depending on the specific methodology Mijinheer *et al*, (2013).

Since diode response for radiation dose rate is nonlinear, and diodes have many characteristics that are very different from the ion chambers, the commissioning (or characterization) of the diodes is essential before clinical use. There are many papers that address these aspects of diodes Huyskens *et al*, (2001); Jursinic.A, (2001).

2.15.1 Calibration of the diode for entrance dose measurements

The diode is calibration to measure the entrance dose i.e. when positioned of the skin of the patient the measured dose should correspond to the dose to tissue at the depth of maximum dose of the photon quality in use for particular beam geometry. The calibration procedure firstly involves the determination of the calibration factor F_{cal}. it is measured to calibrate the diode for each beam quality with which its intended to be used. Due to the variation of the diode signal with accumulated dose, calibration should be regularly repeated in time. Time intervals typically vary between weekly and monthly. The temperature dependence of the diode signal can be accounted for during calibration if this is performed of the same temperature as measurements with that particular diode in the clinical application. The entrance dose value in clinical situation is calculated from the diode measurement as the product of the diode reading, the calibration factors and the correction factors. Diode in vivo dose measurements can be made at three positions:

(1) Beam entrance: The diode is placed at the entrance points only. Entrance measurements give a check of correct settings of beam parameters such as energy, collimator jaw settings, monitor units given, source-to-distance (SSD), customer blocks, wedges used, and compensators. Entrance measurements minimize the extra workload for the staff and extra setup time. The basic idea is to calibrate the diode first and then use various calculation methods to obtain the target dose Jornet. N *et al*., (1998); Wolff, Carter et al (1998). Correction factors are needed. This method is the most popular. And is the part of my study.

(2) Beam exit: The diode can be placed at the exit point Yaparlvli, Fonlenla *et al*, (2000). Theoretically exit measurements can check all of the parameters mentioned above for entrance measurements, plus changes in patient thickness, contour errors, and problems with CT data transfer or CT mis-calibration (in homogeneities in tissue). However, there are some reasons for avoiding the exit position measurements. For example, there are much better more direct methods than in vivo diode measurements to provide quality assurance checks for CT and treatment planning system. These quality assurance methods should be applied long before an in vivo diode measurement is made Millwater. G *et al*., (1998) .

In addition, there is the problem of reduced backscattered radiation. Most computer treatment planning systems assume the exit dose as the dose on a depth dose curve without taking into account the finite extent of the patient. One way to solve this

problem is described in Ref Wierzbick. J, Waid. D, (1998). One can compare the readings of diode and ion chamber to get a calibration factor:

$$CF=D/R \qquad\qquad (2.8).$$

Where **D** is the absorbed dose measured with the ion chamber, **R** is the diode reading (the inverse square factor is not employed).
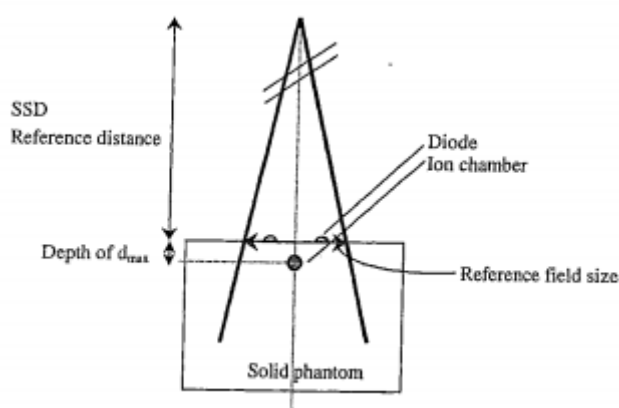
Figure 2.4 Diode calibration procedures for entrance dose measurements. The ionization chamber is positioned at the depth reference in the phantom and the diode at the entrance surface in the reference geometry.

(3) Both beam entrance and beam exit: Theoretically this way is the best method. However, practically, not many institutions employ a diode in vivo system in this manner Eveling. J, Morgan. A *et al*., (1999). The reason is evident: for a busy department, performing both entrance and exit measurements may increase the overall treatment time unacceptably. Since diode response for radiation dose rate is nonlinear, and diodes have many characteristics that are very different from the ion chambers, the commissioning (or characterization) of the diodes is essential before clinical use. There are many papers that address these aspects of diodes Verney. J.N, Morgan. A. M, (2001).

(1) **Linearity:** Under the conditions of fixed SSD and FS, diode measurements are taken with different numbers of monitor units. The linearity of diodes is very good: the standard error of the line is less than 0.1%.

(2) **Dose per pulse dependence** there is a relationship between diode response (or correction factor) and the dose-per-pulse. Dose-per-pulse is not the clinically used dose rate. The clinical dose rate is an average dose rate. For example, for 6 MV X rays with a pulse duration of 5μs, 1Gy at Source axis distance (SAD) = 100 cm was delivered with 3550 pulses, so the dose-per-pulse is 1Gy/3550pulses = 2.8 x 10-4 Gy / pulse. However, the clinical dose rate is about 1.0cGy/MU. The dose-per-pulse and clinical dose rate is a function of the

Source-to surface - distance (SSD). Sometimes the gun current can be adjusted on the Linear accelerator to deliver a different dose per pulse (especially for higher dose-per-pulse values). Grussell. E, Rickner. G, (1987) hypothesized that dose rate dependence is associated with pre-irradiated n-type Si diodes and no dose rate dependence would be expected for p-type diodes. However, actual measurements indicated that both n- and p-type of diodes have dose per pulse dependence, although the dependence for n-type diodes is greater Levenens *et al.,* (1990).

(3)-**Field size dependence**: For high energy photon beams, backscattering is negligible and almost all scattered photons come from the overlying layers Heukolom. S, Laanson *et al*., (1991). So as the diode is placed on the phantom surface, the reading of the diode is virtually independent of the phantom scatter and only sees the head scatter. Therefore, the phantom scatter factor Sp should not be included in the calculation of the dose to the diode. Because Sp increases when the FS increases, we would expect that the FS correction factor of diode to increase when the FS increases. However, both increases and decreases were found with changes in field size Jursinic.P. A, (2001).

(4)-**SSD dependence:** Generally the diode correction factor increases when SSD increases Millwaster. *et al*., (1998). That is, diodes tend to underestimate the dose when SSD increases.

(5)-**Energy dependence:** Diode response to radiation dependents on energy.
The calibration of the diode need be performed individually for each energy Wierzbick. J. C, Waid. D, (1998).

(6)-**Temperature dependence:** Depending on the amount of pre-irradiation, the temperature correction of the Scanditronix diodes can be up to 3.5% if the diode is positioned on the patient skin and calibrated at room temperature. For Sun Nuclear

Corporation QED and Isorad diodes, the temperature dependence is small, just 0.3% per degree Celsius Alecu. *et al* (2000).

(7)-Directional dependence: Just as what described in the Chapter one, both of interface phenomena and the shape and geometry of the diode give rise to directional dependences Lee. P.C, Sawicka. J, (1991); Wolff. T, Carter. *et al*, (1994). If the incident beam is not perpendicular to diode, the diode reading may be smaller or larger than that of perpendicular beam.

(8)-Wedge correction factors: The wedges decrease the dose per pulse and also change the beam quality; consequently, they change the diode response. So wedge correction factors must be considered Jursinic. P. A, (2001).

(9)-**Cumulative dose dependence:** as the cumulative dose to a diode increases, the diode sensitivity decreases. This will decide how often to re-calibrate the diode (Alecu. R, (2001); Leunens. G, (1990).

(10)-Tray correction factor: The use of trays to support blocks modifies the incident photon fluency by producing scattered electrons. This correction is usually within 2%.

(11)-**Off-axis correction**: Off-axis corrections are large for wedged fields and low energy photons Alecu. R, (2001); Leunens. G, (1990). There are primarily two published methods to obtain the actual dose from the diode reading. One method is to make measurements varying each of above conditions, and find various diode correction factors, Ci, for each of the non-reference conditions, e.g., CSSD, CFS , etc. The correction factors are obtained by comparing readings from the diode and from the ion chamber under various non-reference conditions.

That is Correction Factor =

$$\text{Dose at Diode}/(\text{Diode reading}) \qquad (2.9)$$

After obtaining all correction factors, for any actual clinical situation the "expected" diode reading R is calculated by Diode Expected:

$$\text{Rdg} = \text{Dose} * (\Pi\, Ci)^{-1} = \text{Dose} * (CSSD * CFS * \Pi\, Ci)^{-1} \qquad (2.10)$$

Another method, which requires the similar measurements but is conceptually different. The basic idea is to find all or most physical quantities (or physical parameters) for the diode itself, not for ion chamber. This skips the step of determining diode correction factors that were obtained by comparing the readings of the diode and an ion chamber, and directly uses the physical quantities measured

using the diode. One such example is detailed in Ref Millwater.J. G, Waid. D. S , (1998). Which used the following formula Diode:-

$$Rdg = MU \times DCF \times DWF \times TEMPF \times SSF \times DOF (FS_{coll}) \times [(100/SSD)2 \times TBF \times CF]^{n+1}$$

(2.11)

Where MU is the number of monitor units, DCF is the diode calibration factor, DWF is the surface-scatter-factor, SSF is the surface-scatter-factor, DOF is the output factor measured with the diode (Field size dependence), TBF is the block tray factor, and CF Is the compensator factor. The "n" in the above formula is the fitting parameter that arose from the dose-per-pulse dependence the author found:

$$Diode\ Rdg\ /\ dose\text{-}per\text{-}pulse = (dose\text{-}per\text{-}pulse)^{n}$$
(2.12)

Most of these quantities are for the diodes, and not applicable to ion chamber responses. In particular note that the DCF above is the "Diode Calibration Factor". However, in this thesis and in many publications the DCF also is used with a different meaning: "Diode Correction Factor".

Summary, the second method tends to use quantities measured with and for the diode itself directly, in a similar way ion chamber corrections are determined. All of above are for photons. There also are a few papers on diode in vivo electron dosimetry. Similar to diode in vivo photon dosimetry, diodes for electrons need be calibrated under a reference condition and commissioned Rikner.G, Grusell. E, (1987). The commissioning is similar to that of photons. One must determine the dose per pulse dependence, cumulative dose dependence, temperature dependence, directional dependence, field size dependence, energy dependence, the influence of the electron cut-out (insert), and the dose perturbation behind the diode detector. The dose reduction behind the diode detector for electrons can be as large as 25% Alecu. *et al* (2000) for some types of diodes, especially for low energies and small field size, say 6MeV and 3cm diameter circular field. Only entrance measurements are used for electron in vivo dosimetry.

2.16 Recording of in-vivo dosimetry

Recording of the in-vivo entrance dose may be done on a treatment chart, on a separate sheet for QA and / or in a database accessible in a network (possibly linked to the R& V system).

The results should be easily available (after the first session during chart rounds, etc.). It's important to record in vivo dosimetry data together with sufficient information, such as the data of measurements, the type of fields, the treatment unit, the anatomical location and so on. The more complete the database is, the more information can be derived when receiving in-vivo dosimetry.

2.17 Previous study

Radiotherapy is a multidisciplinary specialty using complex equipment and procedures for assessment, planning, and delivery of the treatment, and the main goal of radiotherapy is to treat cancer without health detriment to the patient, which requires delivery of prescribed dose to various types of tumor safely and accurately, as technology and use developed, many accidents in radiotherapy have been reported in several countries in Europe over the last years, to reduce these accidents, the monitoring technique of radiation doses given to the patient was used based on the recommendations of international organizations and the results of previous studies in this aspect..

Derreumaux *et al*., (2008) Studied the Lessons from recent accidents in radiation therapy in France their results showed that In fact, the importance of in vivo dosimetry has been recognized and implemented as a part of QA program in many countries like France, Sweden, Norway, Denmark, and the UK.

W. P. M. Mayles (2007) studied the Glasgow incident - a physicist's reflections. Their results showed that a number of radiation incidents in various countries have been reported. In addition to incidents caused by human errors, suboptimal patient treatments may also occur because one or more of the parameters involved in a patient irradiation may have a systematic error.

WHO, Geneva (1988), Studied Quality Assurance in radiotherapy. Their results showed that the To ensure that the delivered dose agrees with the prescribed dose at the end of the entire treatment process, it has been recommended by number of international organizations that an overall check of the entire process is carried out. One of the recommended methods is in- vivo dosimetry.

Mijnheer *et al*., (2013); Van Dam and Marinello, (1994). In vivo dosimetry in external beam radiotherapy. Their results showed that the in-vivo dosimetry is the most direct method of measuring the radiation dose delivered to the patient during radiation therapy.

Frontela, D. P., *et al*., (1996) Studied Customization of a Radiation Management System to Support *in-vivo* Patient Dosimetry Using Diodes, their results showed that the in vivo dosimetry has proved to be a useful tool for quality assurance in radiotherapy.

Rutonjsk, Petrovic *et al* (2014) Studied a clinical implementation of in-vivo dosimetry with n-type Isorad semiconductor diodes their results showed that systematic in vivo dosimetry proved to be very useful tool for quality assurance of patients plan and treatment , both in detecting systematic errors and estimating the accuracy of radiotherapy treatment delivery.

Aikins, Acquah *et al* (2015) Studied accuracy of using in-vivo dose verification with diodes for different sites their results showed that the in-vivo dosimetry is an effective method to detect errors in radiotherapy to assess clinically relevant differences between the prescribed and delivered doses to reduce potential harm to patients and to fulfill requirements set forth by national and international regulations.

Gadhi, *et al* (2016) Studied verification of absorbed dose using diodes in Cobalt-60 radiation therapy their results showed that the clinical dosimetry using diodes is simple, cost effective, provides immediate results and is a useful quality assurance tool for verification of absorbed dose delivered during patient treatment on Co-60 machine.

Ibrahim.M, Attia *et al* (2016) Studied evaluation of photon beam dose calculation accuracy of treatment planning systems using in-vivo dosimetry their results showed that in-vivo dosimetry is an effective method for dictating radiotherapy errors, assessing clinically relevant differences between the prescribed and delivered doses, reducing potential patient harm, and fulfilling requirements set forth by national and international regulations, recommended that a more accurate calculation of expected diode values be performed, especially for fields that was pass through the table.

Shawata., El Nimr *et al* (2015) Studied improving patient care and accuracy of given doses in radiation therapy using in-vivo dosimetry verification their results showed that the results indicate that the diodes exhibit excellent linearity, dose reproducibility and minimal anisotropy; that they can be used with confidence for patient dose verification.

Mijnheer, Beddar *et al* (2013) Studied in-vivo dosimetry in external beam radiotherapy their result showed in-vivo measurement is to provide an accurate and independent verification of the overall treatment procedure it will enable the identification of potential errors in dose calculation, data transfer, dose delivery, patient setup, and changes in patient anatomy.

Colussi, Beddar *et al* (2001) Studied in-vivo dosimetry using a single diode for megavoltage photon beam radiotherapy: Implementation and response characterization their results showed that high-energy buildup diodes Sun Nuclear Corporation can be used for in vivo dosimetry in the entire megavoltage energy range used in radiotherapy.

Rutonjdki, Petrrovi *et al* (2014) Studied clinical implementation of in vivo dosimetry within n-type semiconductor diodes their results showed that *in vivo* dosimetry has given the full confidence that patients are being treated with the prescribed and planned dose.

Ruiz, Beddar *et al* (2020) Studied In vivo dosimetry in external beam photon radiotherapy: Requirements and future directions for research, development, and clinical practice their results showed that In vivo dosimetry (IVD) is an essential element of modern radiation therapy because it provides the ability to catch treatment delivery errors, assist in treatment adaptation, and record the actual dose delivered to the patient.

Gadhi , Buzdar *et al* (2016) Studied *In-Vivo* Dosimetry with Diode for the Treatment of Pelvic Malagnancies their results showed that diodes clinical dosimetry system is a useful QA tool for verification of dose delivery and in identifying the systematic/random errors. It has enhanced the quality of radiation dose delivery and reliability of the system.

MacDougall *et al*. (2017) Studied guideline & recommendations In vivo dosimetry in UK external beam radiotherapy: current and future usage their results showed that owing to technological advances, such as electronic data transfer, independent monitor unit checking and daily image-guided radiotherapy, the overall risk of adverse treatment events in RT has been substantially reduced. However, the use of IVD may prevent a serious radiation incident. Point dose IVD is not considered suited to the requirements of verifying advanced RT techniques, leaving EPID dosimetry as the current modality likely to be developed as a future standard the use of IVD may prevent a serious radiation incident.

AAPM REPORT NO. 87 (2005) diode in vivo dosimtery for patients receiving external beam radiation therapy their results showed that In vivo dosimetry directly monitors the radiation dose delivered to a patient during radiation therapy. It allows

comparison of prescribed and delivered doses and thus provides a level of radiotherapy quality assurance that supplements port films and computational double checks. A well-devised in vivo dosimetry program provides additional safeguards without significantly extending treatment delivery time.

Gadhi *et al*. (2019) Studied Measurements of radiation dose for cancer patients their results showed that quantitative entrance and exit absorbed dose verification with diode dosimeter is beneficial for quality improvement in radiation therapy. Execution of entrance and exit dose measurement procedure has demonstrated to be very helpful for noticing potential mistakes and avoiding errors due to inaccurate positioning of patients.

Mrcela, Bokulic *et al* (2005) Studied calibration of p-type silicon diodes for in-vivo dosimetry in Cobalt-60 their results showed that the Results were within expected values for this type of diodes giving acceptable agreement in dose delivered and the expected dose.

Trujillo, Ibbott (2005) Studied pre- clinical evaluation of a diodes- based in-vivo dosimetry system their results showed that while care must be taken in choosing and handling diode detector systems they are able to provide an efficient and effective method of ensuring the dose delivered to the patient during treatment is within acceptable limits.

Allahverdi1. M *et al* (2008) Studied Diode calibration for dose determination in total body irradiation ,their results showed that the diode dosimetry is very useful as a check of midplane dose delivered to patients under TBI treatment.

DÖNMEZ KESEN. N. *et al* (2017). In Vivo Dosimetry in External Radiotherapy, their results showed that the measurement of the skin dose by IVD is an important part of current QA programs that use advanced radiotherapy techniques. While performing IVD, the characteristics of the measurement systems and their effective measurement depths should be known to evaluate the results correctly. The purpose of IVD programs is to increase the accuracy and quality of treatments, similar to other QA programs.

Lenunens.G *et al* (1994) Importance of in vivo dosimetry as part of a quality assurance program in tangential breast treatment. Their results showed that in vivo

dosimetry is an important tool in a department quality assurance program to detect systematic errors in dose delivery, to indentify inadequate treatment situations, to investigate weak point in the chain of treatment preparation and to ensure accurate dose delivery for individual patients.

IAEA Human Health Reports No. 8. (2013) It allows comparison of prescribed and delivered doses and thus provides a level of radiotherapy QA that supplements portal films and computational double checks. The ultimate check of the actual dose delivered to a patient in radiotherapy can only be achieved using *in vivo* dosimetry.

Essers. M *et. al* (1999) Studied in vivo dosimetry during external photon beam radiotherapy. Their results showed that is recommended techniquices are checked systematically for a few patients, and to perform in vivo dosimetry a few times for each patient for situations where errors in dose delivery should be minimized.

Fontenla. D. P, R. Yaparpalvi, *et al* (1996)., Studied the use of diode dosimetry in quality improvement of patient care in radiation therapy, their results indicate that the diodes exhibit excellent linearity, dose reproducibility, minimal anisotropy, and can be used with confidence for patients dose verification.

Huyskens. D. P ., *et al* (2001).Practical guideline for the implementation of in vivo dosimetry with diodes in external radiotherapy with photon beams (entrance dose),Their results showed that the in vivo dosimetry confirm that a number of serious systematic errors might escape the independent check of dose calculation and data transfer, which should be always performed before treatment delivery. Moreover, in vivo dosimetry permits detection of a number of minor errors (SSDs and thickness errors) which would be undetected by the independent check, thus improving the global quality of the treatment.

Leunens. G., *et al*., (1990). Studied the Quality assurance in radiotherapy by in vivo dosimetry. Entrance dose measurements, a reliable procedure. Their results that the study demonstrated the reliability of the use of semiconductor detectors for in vivo dosimetry and its usefulness as part of a departmental quality assurance program.

Petkovska.S. (2018), Treatment verification in Radiotherapy. Explained that treatment verification in radiotherapy doesn't mean that the exact predicted dose on the exact place will be delivered during treatment execution later on. But moreover it helps to avoid radiation accidents.

Adeyemi A, Lord J (1997) Studied an audit of radiotherapy patient doses measured with in vivo semiconductor detectors. Their results showed that the ntrance dose measurements serve to check the output and performance of the treatment apparatus as well as the accuracy of 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; a variety of detectors, including thermoluminescent dosimeters (TLD), silicon diodes, and new detectors such as metal oxide silicon field-effect transistors are currently available for in vivo dosimetry.

CHAPTER THREE

MATERIAL AND METHOD

3.1 Introduction

This chapter deals with the material and methodology of the study, this work is divided into two parts:

(i)   Materials used in this study

(ii)   Measurements procedures (phantom measurements and patients measurements).

3.2 External beam therapy equipment

Beam data were obtained from (Co60 (CIRUS, CIS BIO, French AEC SN 4248, ACCT) teletherapy Cobalt unit which is installed in special suite. The machine is designed to house Cobalt-60 source of maximum capacity 233TBq. A pneumatic air system, using compressed air, controls the source drawer. It is used to derive the source from fully shielding position to a fully exposed position. Emerging photons from the Cobalt-60 source have energies of 1.17 and 1.33 MeV, and effective energy of 1.25 MeV beam radiation reaches its maximum dose at 0.5 cm below the skin surface; therefore, it was especially well suited for radiation therapy of the head, neck and breast, and for tumor within 5 cm of the skin surface in other parts of the body.

 These devices undergo acceptance testing processes as well as calibration to ensure their suitability to apply treatment programs accurately, and this is done using phantoms.
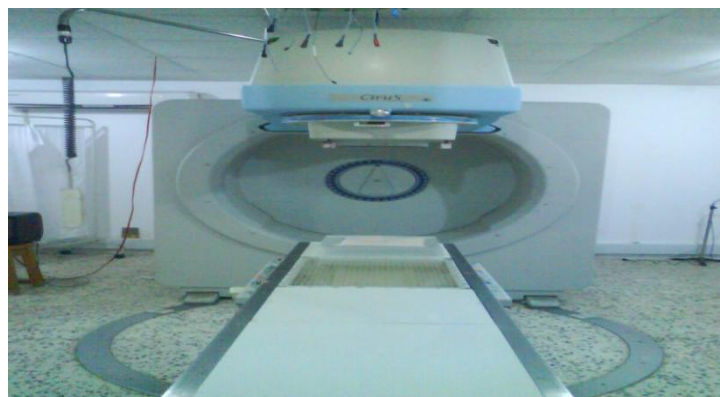
Figure 3.1 Cobalt- 60 CIRUS, CIS BIO

3.3 Phantoms

The Phantom made from Perspex slabs C5H8O2 of 1.15 gm/cc density the thickness of each slab is 1cm, the exception is the slab in which the ionization chamber is inserted, 2 cm^2 and 0.5cm^2) phantom with a sided window of 30 cm^2 x 30 cm^2 x 30 cm^2 size were used for this study along with a 0.6 cc Farmer ion chamber were used. Directional dependence of diodes, the effects of thickness correction factor and complete backscatter factor were studied as shown in Figure 3.2.
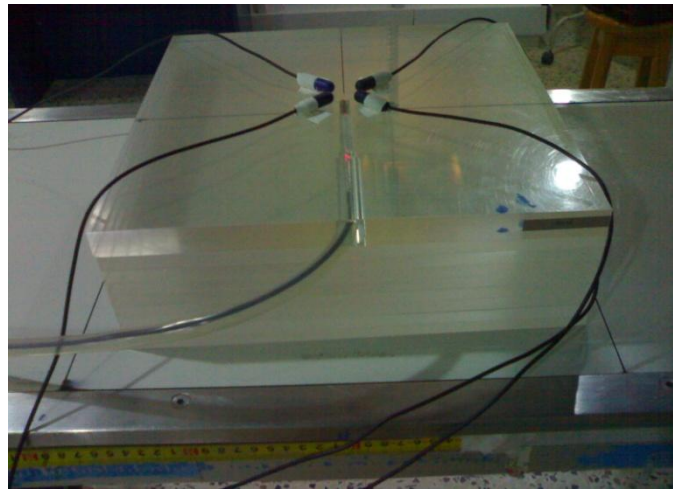
Figure 3.2 Perspex slabs Phantom

Diodes are installed on the outer surface of the phantom in the middle of the treatment field of the beam center during the measurements.

3.4 Description of detector system

The system employed in this study was IVD 1131 dosimetry from Sun Nuclear Corporation with a set of n-type especial semiconductor diode photon detectors Isorad (Sun Nuclear Corporation) for measurements on Cobalt-60 photon beams) and use at National Cancer Institute NCIS – Libya for in vivo dosimetry, the real time measurement of radiation dose administered to a patient during radiation therapy.

The diodes were connected to a 4 channel microprocessor-controlled mobile electrometer with provision for diodes. The electrometer was positioned in the treatment room and linked by concocted cables to a computer and a display unit placed at the console area. As seen in Figure (3.3) to (3.5).

Figure 3.3 Model 1131 wired detectors pods, control Module

Figure 3.4 ISORAD-p™ detectors for photon energy.

Control Module

Control module mounts to the wall, and removes the need for software as shows in Figure 3.5.

Figure 3.5 Shows Control Module.

PC Software

Software includes patient database, automatic correction factors, record-and-verify interface, and more measurement options. The system is operated by clicking buttons on the PC screen.

3.5 Ionization Chamber

One calibrated ionization chamber used for dose measurements in this study.

3.6 Design of the study (type) Principle of the method

The methodology used in this study, Analytical case study.

3.7 Population of the study

Oncology patients to be treated with radiation therapy, in different technique's (Radical, Palliative) for head, neck, breast and pelvis site).

3.8 Sample size of the study

A total of Forty nine patients with different types of cancer diseases were randomly selected and admitted for this study after obtaining due clearance from the ethical committee of the hospital. Admission of patients for this study was based on the cancer distributions reported for treatment in the center and each patient consent was sought before measurements were taken. The most common cancer types featuring in the center are: Breast, pelvis (prostate, rectum and cervical), and H and N (Brain and ngopharynx, Larynx, mouth cavity). The patient's measurement with diode was divided into three groups as shown in table 3.1.

Table 3.1 shows the distribution of tumor patients and field monitored in the sample.

Description Sites	Total Patients	Fields Monitored
Head, Neck	22	45(46.87%)
Pelvis	19	37(38.54%)
Breast	8	14(14.58%)

3.9 Area and duration of the Study

All the practical and analysis of the data was carried out at Medical Physic and Radiation therapy Department NCIS – Libya. the irradiation was done by Cobalt-60 unit.

3.10 Methods of data collection

This study was divided into two phase (i) and phase (ii):-

(i)   Phantom measurement (Calibration diode procedure as in-vitro study).

(ii)   Patient dose measurement procedures.

The first phase involves reviewing the current calibration procedures for diodes and making recommendations about the possible improvements in diode calibration and correction factors

3.10.1 Phantom measurement "calibration procedures"

Source of cobalt Co^{60}
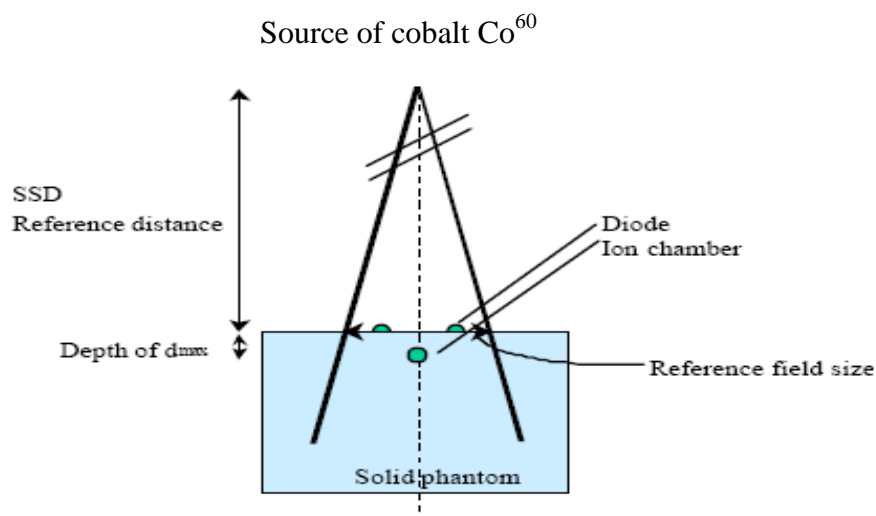
Figure.3.6 Shows involves all the measurement procedures in this study.

Calibration means the determinations of the calibration factors of each diode and the determination of the correction factors which are required to calculate the absorbed dose when measuring in clinical and calibrating conditions differences. And the calibration situation will be similar to clinical situation. For entrance calibration the diodes were placed on the front surface of the phantom. Measurements were performed using photon beams of Cobalt-60 machine. The beam was calibrated using iodination chamber placed at 5cm depth in the sold phantom according to IAEA TRS-398 protocol and also the ionization chamber was calibrated at the secondary standard dosimetry. The in-vivo dosimetry system used in this study consisted of four Isorad n-type diodes model 1137 wired IVD-2 connected with cable use for Cobalt-60 beam and diodes were taped onto the surface of the phantom for characterization as shown in Figure (3.1) the reproducibility of the system and the stability of the signal of the diodes after irradiation and leakage were the preliminary testing of the diodes. The diodes are relative dosimetry and therefore these need calibration to confer the dose at the depth maximum dose d max the calibration factors for entrance dose setup for all diodes were measured with a 10x10 cm field size at source to surface distance SSD 80 cm and 100cGy dose at d max.

A. Intrinsic precision

The intrinsic precision is given by the reproducibility of the signal for at least 10 consecutive irradiations at the same dose and calculation the standard deviation for each diode. Ten reading for diode A, B, C and D, the average of ten readings and the standard deviation for each diode ranged 0.11% - 0.17%. The standard deviation should not exceed 1%. The standard deviation these values are within the recommended value of less than 1%.

B. Initial testing and reproducibility

The stability of the all diodes ISORAD diodes were determined by exposing these diodes to their corresponding energy, intermediate after irradiation with the display of the signal taken five minutes after the end of irradiation. All measurement done in the same conditions, field size 10 x10 cm, SSD 80 cm, angle 0° ' 100cGy.The stability of the ISORAD and This test is usually performed by comparing the display of the signal taken immediately after irradiation with the display of the signal taken five minutes after the end of irradiation. One minute readings for diodes taken immediately after irradiation and five minutes after the end of irradiation. It also displays the percentage difference between the two readings for the four diodes. The test was repeated five times to ensure consistency in the results. As can be seen, the percentage difference ranged between 0% and ±0.1 %. Such variation is within the expected value, which is usually less than 0.5%.

C. Dose perturbation

The diode is used for entrance dose determination; there is a decrease in dose (a dose shadow) below the diode which depends on the effective thickness of the diode, beam modality and energy, field size, and the depth of interest. This was done by repeating the measurements with and without diode. Within its designated range, a detector's dose shadow (10 cm^2 ×10 cm^2 field sizes, SSD 80 cm^2, 4 to 5 cm^2 depth) should not exceed 5% to 6%, but if it is used for lower-energy photon beams. The perturbation can exceed 10%.

D. Dose linearity

The dose linearity of the four diodes was tested using two different methods. In the first method, the readings for one exposure at the lowest time expected for in vivo dosimetry (Dose / minute) and the readings for one exposure of the same field at the maximum time expected for in vivo dosimetry (Dose / minutes) were recorded.

E. Source- surface distance dependence

The measurements were only performed on flat part of the phantom with field size range from 5x5 to 20x20 cm were used for four diodes. Each field size the distance from the source of the beam to the surface of the phantom at central axis was varied from 80 cm^2 to 90 cm^2 and calculations to deliver 100 cGy to the depth at d max for respective SSD and field size was made, the response of the diode for the varying SSD was noted and a comparison of the trend of the response for different field sizes was made.

F. Field size factors

The response for different field size (5x5 - 25x25 cm^2) was investigation by respective field sizes and noting corresponding changes in response of the diode. The four diodes were also used for the procedure. The diode was placed at 80 cm on the top phantom a dose 100cGy was delivered to a d max. the field size were varied and the response of the diodes was noted for all fields and the equivalent square field and comparison of the correction doctors for the fields and corresponding equivalent square field was done. All measurements and irradiation under standard condemnations for all fields and the signals were normalized relative to 10 x10 cm field signal.

G. Incident angle dependence

The phantom was laid on the treatment couch and positioned as a patient and the diode was placed transversely with respect to the axis of rotation of the gentry on the central part of the phantom and aligned using a sagital laser, the gentry angle was set 0° and this was taken as the nominal position Treatment fields such as 10 x10 cm^2 fields was used a distance 80 cm^2 from source of the beam, a dose 100cGy was delivered to the phantom at depth such that a dose at d max was measured. The fore

diodes were irradiated using 100cGy at angle 0° to 90° the measured readings were collected and recorded. The irradiations were repeated for gantry angle from -90° to 0° taken as negative.

3.10.2 Diode Correction factors

The response of the diodes was measured as a function of the following;

(i)   Gantry angle

(ii)   Field size

(iii)   Source to surface distance (SSD)

(iv)   Wedge angle

(v)   Tray

The methods of all measurements for all measurement stages will be studied separately later in the section.

A. Gantry Angle

The phantom was laid on the treatment couch and positioned as patients during treatment. A diode was taped onto the surface of the phantom for as shown in Figure (3.1). The gantry was set to 0° and this was taken as the nominal position. A field size of 10 x10 cm was used at a distance of 80cm from beam source. A dose 100 cGy was delivered to the phantom at 0.5cm depth such that a dost at d_{max} was measured, the four diodes were irradiated, the measured readings were collected and recorded and the irradiation were repeated for gantry angles rending from to this range of angles was the most ideal for measurements.

B. Fields size

The diode response for different field's size was investigated by varying the respective field sizes and noting the corresponding changes in response of the diode. The diode dosimetry reference point DDRP. the diodes was placed at 80 cm from the beam source on the surface of the phantom as shown in Figure (3.1) the setup for determining the field size correction factor are made at different field sizes 5x5, 10x10, 20x20 and 25x25 cm, SSD 80 cm and a dose of 100 cGy was delivered to a depth 0.5cm. A comparison of the correction factors for the fields and corresponding equivalent square fields was done.

C. Source to surface distance (SSD) dependence

Measurements were only performed on the flat surface of the phantom. The setup for determining the SSD correction factor FC_{SSD} is identical to that used for dose calibration except that measurements are made at different SSD, covering the range expected during treatment (70, 75, 80, 85 and 95). And calculations to deliver 100 cGy to a depth of 0.5cm for the respective SSD and field size were made and diode taped on the surface of the phantom at center axis of the beam. The response of the diode for the varying SSD was noted and a comparison of the trend of the response for different field sizes was made.

D. Wedge angle dependence

A wedge is a device which modifies the beam intensity profile by attenuation, thereby modifying its shape. The wedge affects the dose rate dependence of a diode. The phantom was positioned as front the beam directly and four types of physical wedge were selected for the measurements. They were 15°, 30°, 45° and 60°. The wedge correction factors are defined as the ratio between the wedge transmission factors for a 10x10 cm field, SSD 80 cm, a dose 100cGy. The four diodes were used in the investigation. Diode taped on flat surface of the phantom at the centre axis of the beam. (3.1) gives the possible combination of field size and wedge angles used in the measurements. In all cases the same dose was delivered to the phantom at depth 0.5cm.

No:	Field size (cm)	Wedge angles used
1	10x10	15°,30°,45°,60°

Table (3.2): Possible combination between field size and wedge angle.

F. Tray dependence

Tray a device attached to the treatment head front the central axis of the beam as modifying devices such as wedge. The tray affects the dose rate dependence of a diode and correction factors have to be applied. The tray transmissions for different fields size at d max are measured with ionization chamber and the diodes taped to the

surface of the phantom at SSD 80cm and a dose 100cGy was delivered to the surface of the phantom and the results were recorded, compared and plotted.

No:	Field Sizes (cm)	Tray used
1	(5x5),(10x10), (15x15),(20x20) and (25x25)	Yes

Table (3.3) gives the field's size and tray.

3.11 Patient measurements

The entrance dose was measured at the first or second treatment session and the patients were divided in categories according to the tumor localization and patient immobilization. Accordingly, the assessment of set up precision was done as well as the de termination of tolerance/ action levels for different tumor localizations. The patients were divided in next categories:-

(i)   (H& N) brain and head, and neck patients, and

(ii)   Breast patients, and

(iii)   Pelvis

A total of 45 treatment fields involving 22 patients randomly selected were included in the pilot study. The patients were patients treated for head and neck cancers. In-vivo entrance dose measurements were performed during at least two treatment sessions on every patient's treatment field. The goal was to discover discrepancies larger than 5% between the calculated dose and the measured dose by diode, the calculated dose was defined as dose at depth of dose maximum and was calculated by TPS from the prescribed tumor dose. Each patient was treated with an immobilization mask with reference marks at entrance points in each field. Diodes were positioned on these reference marks in the center of every treatment field.

The study sampled 2-D technique, SSD $80cm^2$ during treatment procedure, patients treated for head and neck and brain malignancies, were immobilized in the thermoplastic mask. The most often beam arrangement was lateral, ant/post field.

Breast patients treated with tangential technique positioning with breast board as immobilization device. The entrance dose measured at axis medial tangential field during the first or second treatment session. The majority of patients with pelvic malignancies were treated with either 2 fields or a 4 field technique, with or without wedges. Patients were in most cases positioned in a supine position. Diode were taped onto the patient's body or on the surface of the fixation device used during treatments so that it is in the middle of the treatment beam, well fixed during the treatment. All measurements are taken at axis medial field for all cases. All values of the radiation dose deposited to the patient are recorded and then compared to the dose calculated using the TPS system.

Figure 3.7 Different portals of pelvis patients with diode dosimeter fixed in the center of radiation treatment fields.

3.12 Variable of the study

International commission on radiation unites and measurements ICRP has recommended that radiation dose must be delivered to within 5% of the prescribed dose Alam R,I, Pourang. R (1997), ICRU Bethesda, MD, (1978). And in recent publication by IEAE (2013), an appropriate goal is to be able to use a tolerance level of 5% for simple treatment, with a level of 7% for situations such as breast treatments and other treatment where measurement complications exist. However, it is recommended that, although in the initial stage of the introduction of in vivo dosimetry the tolerance levels may need to be higher, every effort should be made to achieve tolerance level of about 5% by a process of progressive elimination of identified causes of dose differences Georg D, Hoornaert. B (1999). In my study we seeks to compare the entrance doses derived from the signal of the diode detectors place on the skin with the theoretical values (prescribed dose) as calculated by the TPS under set tolerance values.

3.13 Methods of data analysis

The expected doses, measured doses, dose deviations, and percentage mean deviations were recorded as means (standard deviation). Statistical analyses significant test for all the data were performed *using-t* test online.

CHAPTER FORE

RESULTS

4. Introduction

This chapter presents the results of the measurements for the diode response characteristics with different parameters. The methods of obtaining the results are clearly given in chapter three. The results are shown as tables and figures. The figures show the variation of the diode correction factors and dose with the investigated parameters namely gantry angle, field size, SSD, wedge angle and tray. The results obtained from the validation process are also given in this chapter. A selected number of measurements were done both for phase 1 and phase 2. Phase 1 was for the measurements for flat phantom surface and phase 2 is for the patients measurements. In this study four n-type semiconductor diodes were first commissioned prior to use them for in vivo dosimetry of patients, labeled as A, B, C and D. The commissioning procedure consisted of conducting acceptance tests; determination calibration and correction factors.

4.1 Result of Phantom Measurements (in-vitro)

Table 4.1 Shows results of ten readings for diodes A, B, C and D, after exposure.

SN	Diode A	Diode B	Diode C	Diode D
1	1.014	1.015	1.016	1.015
2	1.015	1.015	1.018	1.014
3	1.014	1.014	1.015	1.014
4	1.013	1.012	1.016	1.013
5	1.012	1.012	1.014	1.014
6	1.012	1.011	1.015	1.013
7	1.012	1.011	1.015	1.012
8	1.013	1.011	1.015	1.012
9	1.013	1.011	1.014	1.011
10	1.012	1.011	1.014	1.012

Table 4.2 Shows the mean of ten readings and their percentage standard deviation for all diodes

N0 Diode	Average of 10 times Reading After one minute's irradiation.		
	Min	Mean ± SEM	Max
A	1.012	1.013± 0.0003	1.015
B	1.011	1.012±0.0054	1.015
C	1.014	1.015±0.0004	1.018
D	1.011	1.013±0.0039	1.015

Table 4.3 Shows the results of the stability of the signal diodes which taken immediately and after irradiation for all diodes (mean reading and SD for all diodes).

SN	Diode Readings	A	B	C	D
1	Immediately	0.946	0.955	0.946	0.931
	After 5 min	0.946	0.955	0.945	0.931
2	Immediately	0.946	0.955	0.947	0.931
	After 5 min	0.945	0.955	0.946	0.931
3	Immediately	0.947	0.955	0.946	0.93
	After 5 min	0.947	0.955	0.945	0.931
4	Immediately	0.944	0.954	0.946	0.931
	After 5 min	0.945	0.954	0.946	0.931
5	Immediately	0.945	0.953	0.946	0.929
	After 5 min	0.945	0.953	0.945	0.929

Table 4.4 Shows summary results of the stability of the signal diodes which taken immediately and after irradiation for all diodes (mean reading and SD for all diodes).

N0 Diode	Immediately Reading			After 5 min Reading			Deviation %
	Min	Mean±SEM	Max	Min	Mean±SEM	Max	
A	0.944	0.945±0.0005	0.947	0.945	0.945±0.0004	0.947	0. 11
B	0.953	0.954±0.0004	0.955	0.953	0.954±0.0004	0.955	0.09
C	0.946	0.946±0.0002	0.947	0.945	0.945±0.0002	0.946	0. 05
D	0.929	0.931±0.0004	0.931	0.930	0.931±0.0004	0.931	0.09

Table 4.5 Shows results of Diode's readings at the minimum and maximum measured for in vivo dosimetry.

Dose Gy	Diode A reading	Diode B reading	Diode C reading	Diode D reading
0.5	0.502	0.5	0.502	0.502
8	8.093	8.049	8.088	8.073

Table 4.6 given the results for each diode together with calibration factors.

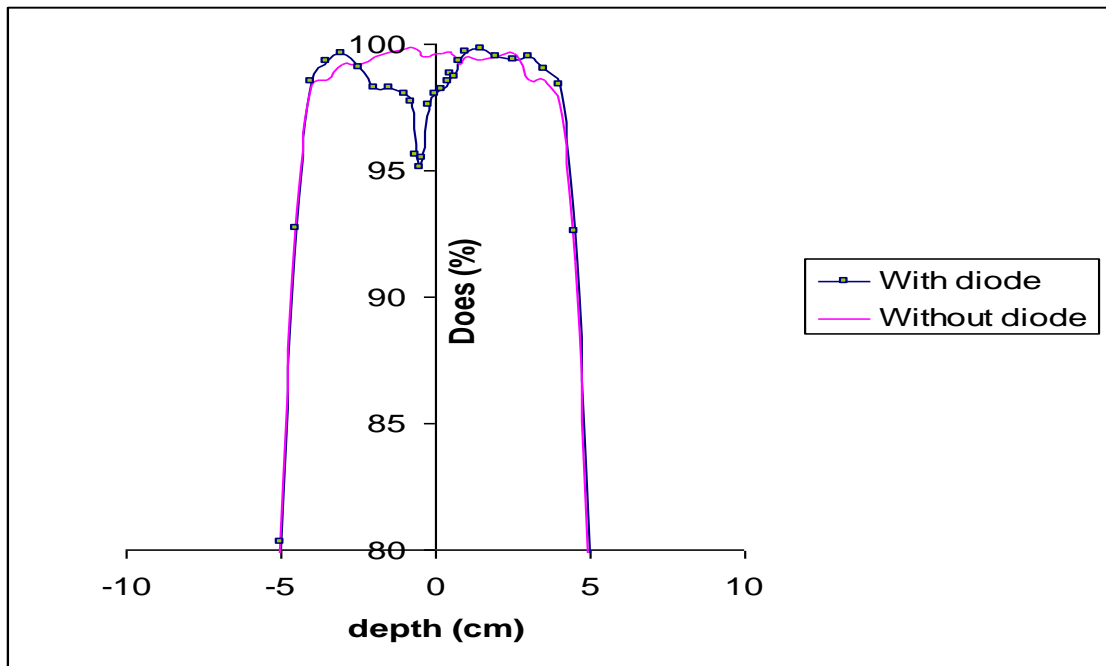| Acceptance test	Diode A	Diode B	Diode C	Diode D
Intrinsic precision standard deviation	0.10 %	0.17 %	0.12 %	0.12 %
Linearity- correlation coefficient	1.0075	1.0061	1.0069	1.0051
Signal stability after irradiation deviation.	0.10	0.09	0.10	0.10

Figure 4.1 Shows dose perturbation, normalized dose profiles produced at 5cm depth in water with and without the diode positioned at surface, the dose decreasing under diode 5%.

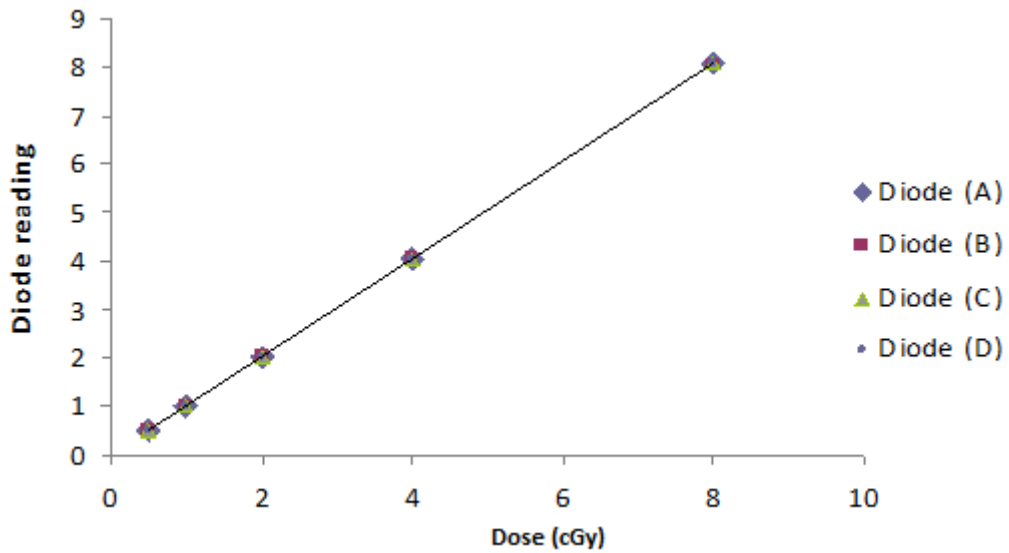Figure 4.2 Shows results of diodes signal as a function of dose, the signal increase in a linear way with the beam energy.
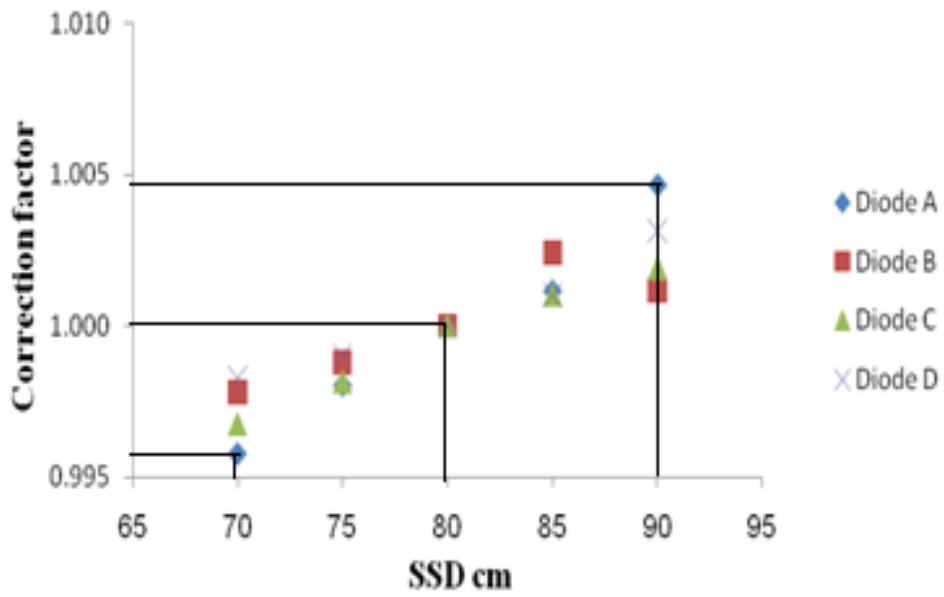
Figure 4.3 Show source- distance dependence of diodes for a photon.
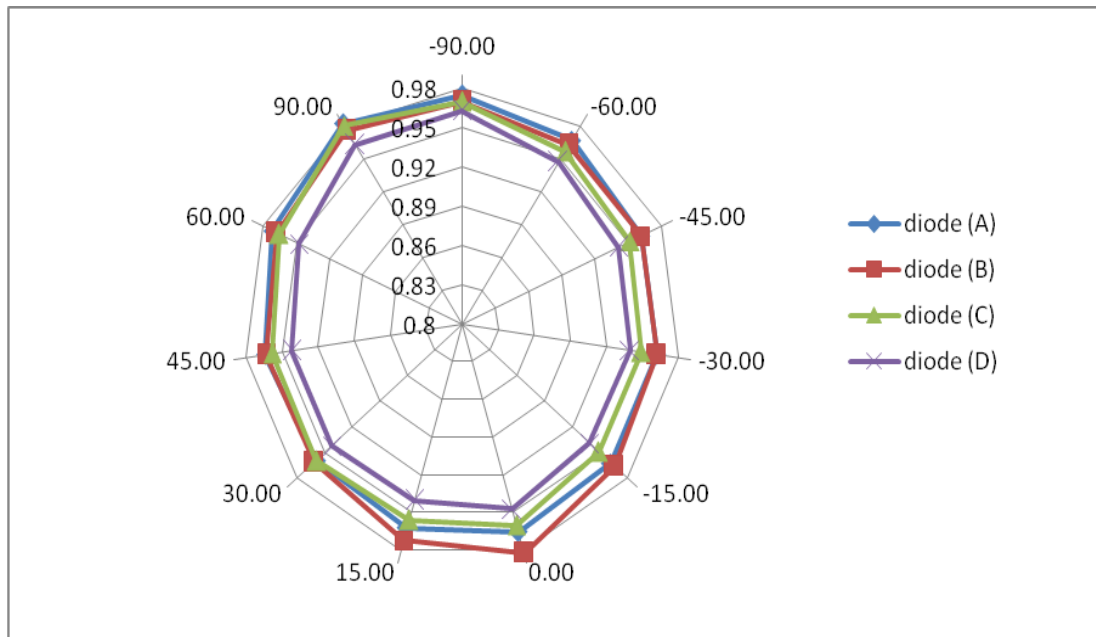
Figure 4.4 Shows diodes correction factors as a function of gantry angle on a flat surface.
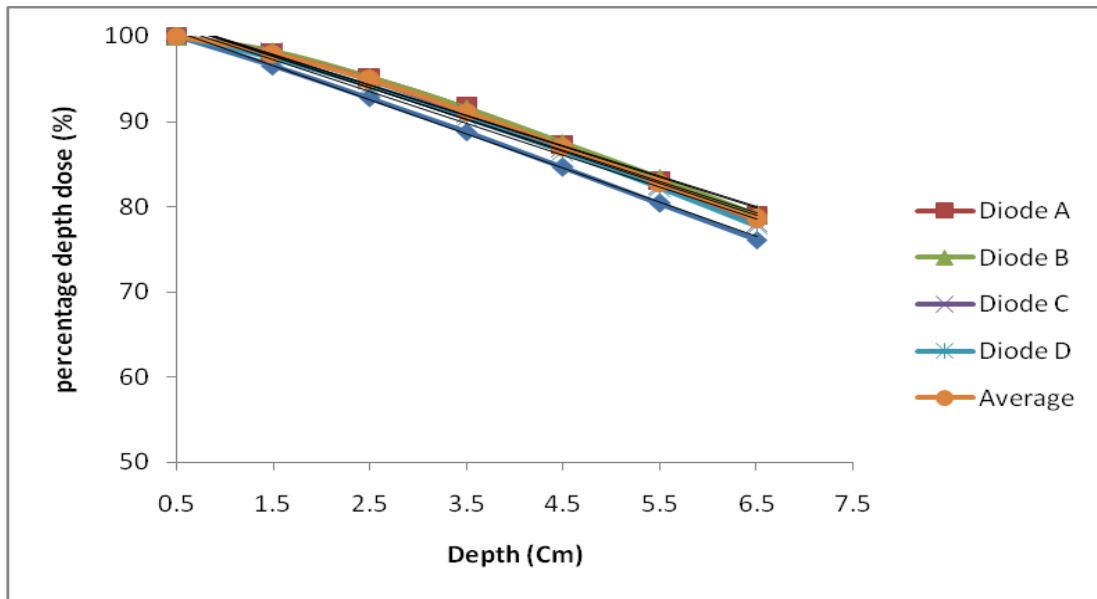
Figure 4.5 Shows the diode correction factors variation with buildup
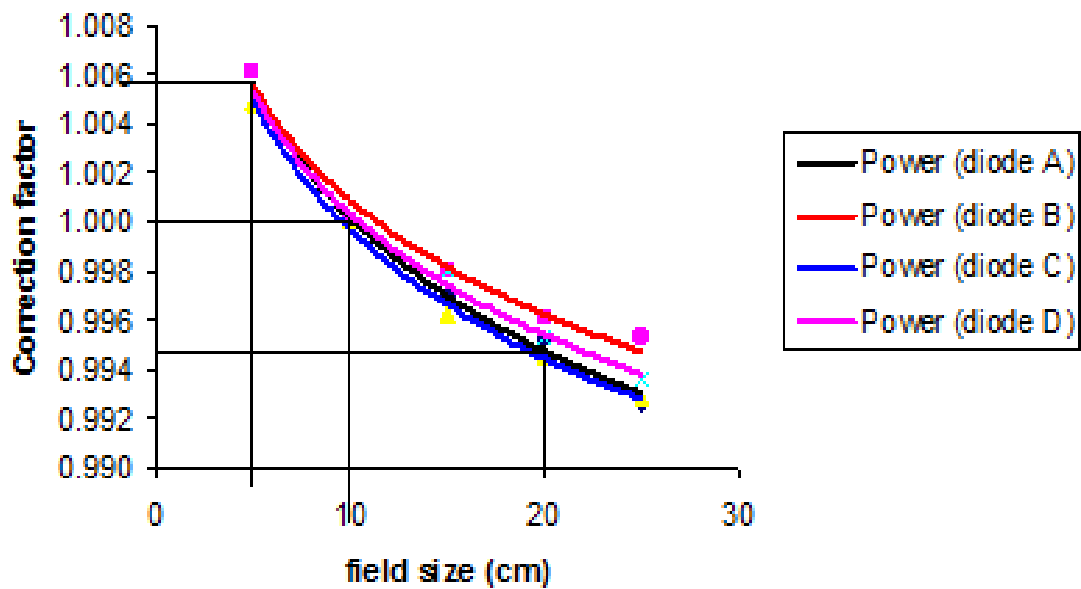
Figure 4.6 Shows field size dependence of diodes for the beam.
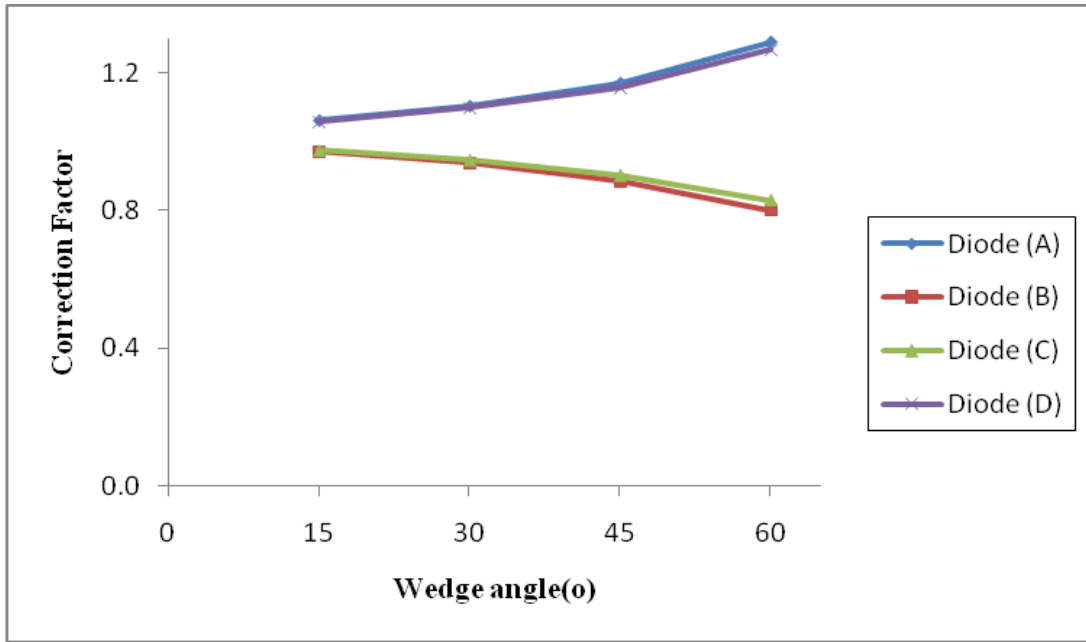
Figure 4.7 Shows effect of wedge falter on diodes response, and relation between inserting wedge in the beam with reading of each diode.
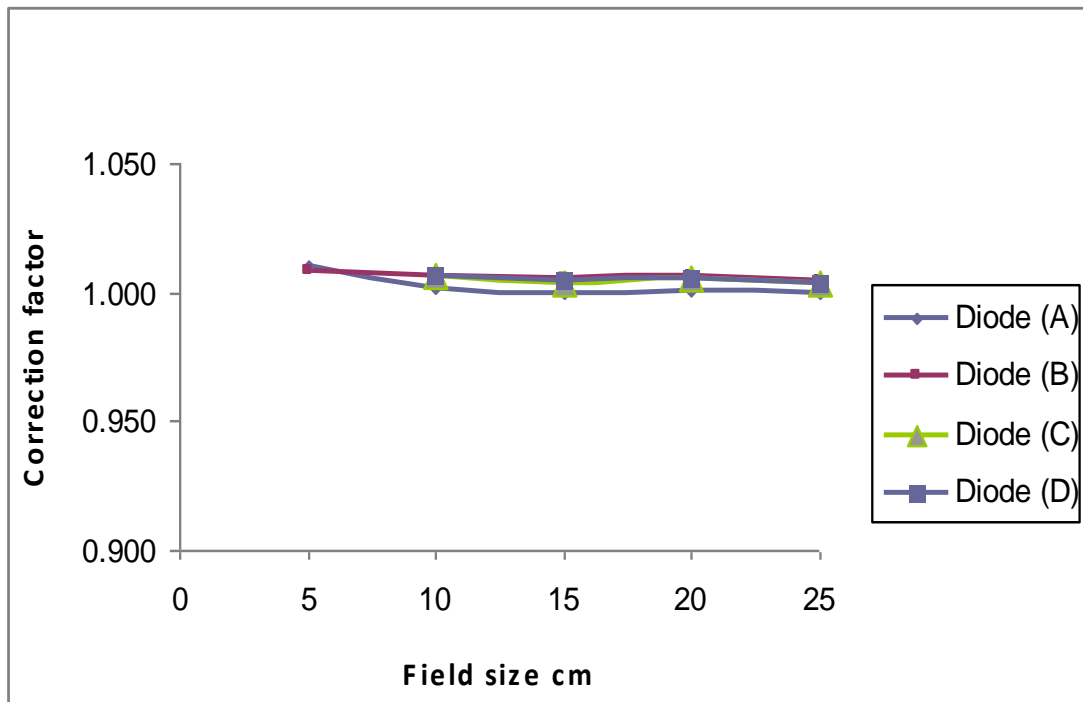
Figure 4.8 Shows the tray correction factor with diodes reading in different field size.

Diode A	Diode B	Diode C	Diode D
0.0109715	0.0106934	0.0103341	0.0099597

Table 4.7 Shows calibration factors value for entrance dose of each diode, value of calibration factors varied from 0.0099597 to 0.0109715.

4.2 Result of Patient measurements

The current investigation was deliberated to explore the difference between the calculated dose planned by TPS and measured doses by diode for cancer patients in different sites treated on Cobalt-60 teletherapy machine as shown in the figure 4.8.
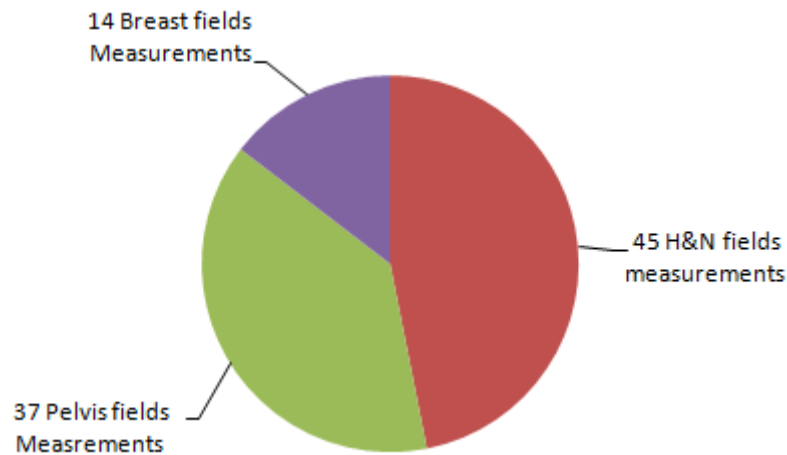
Figure 4.9 Shows the distribution of number of fields measurements in different sites in the sample.

Table 4.8 Discrepancies between the measured and TPS entrance Dose.

Treatment site /technique	Cases	Number of Measurements	%Average discrepancy	%SD
Head, Neck	22	45	3.23 ±0.892 (±27.62%)	0.72
Pelvis	19	37	2.8 ±0.595 (±21.58%)	1.52
Breast	8	14	4.12 ±0.945 (±22.93%)	2.45
All measurements	49	96	3.178 ±0.507 (±15.97%)	1.28
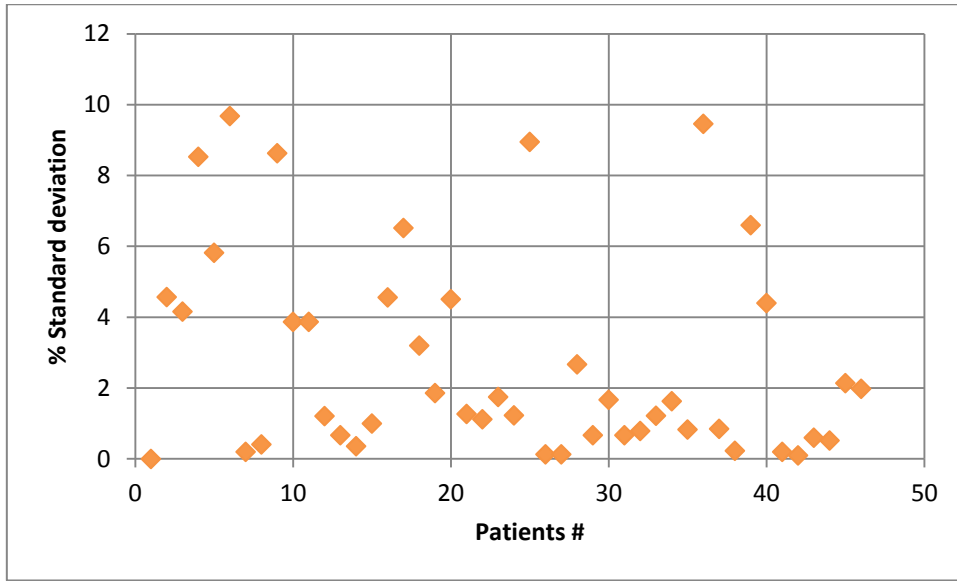
Figure 4.10 Frequency distribution of deviations from expected dose for measurements for head and neck site.
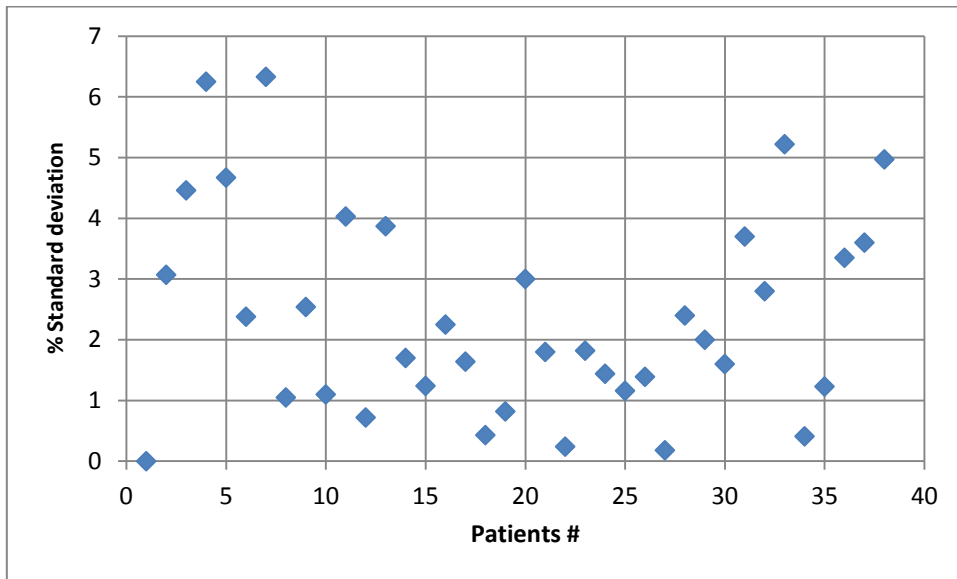
Figure 4.11 Frequency distributions of deviations from expected dose for measurements for pelvic site.
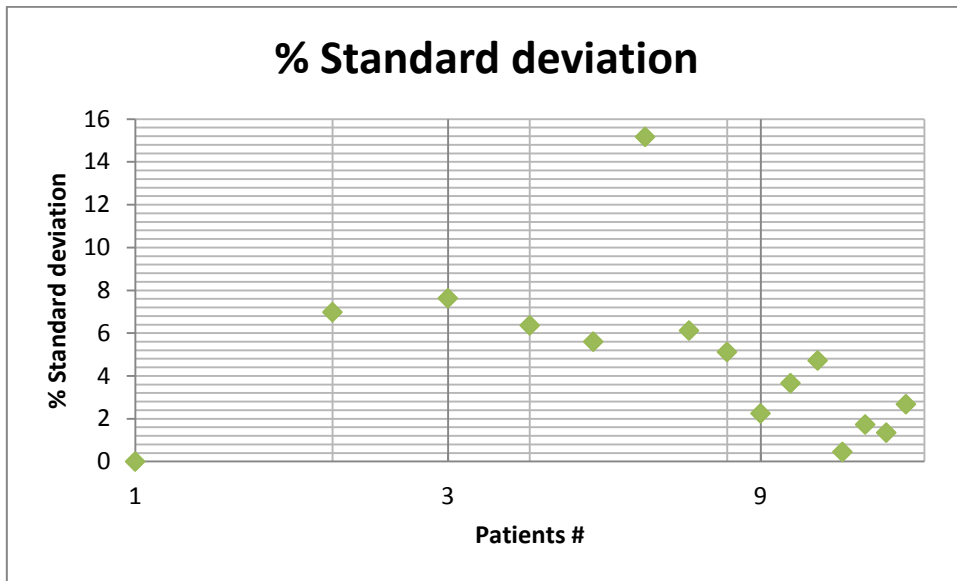
Figure 4.12 Frequency distribution of deviations from expected dose for measurements for breast site.
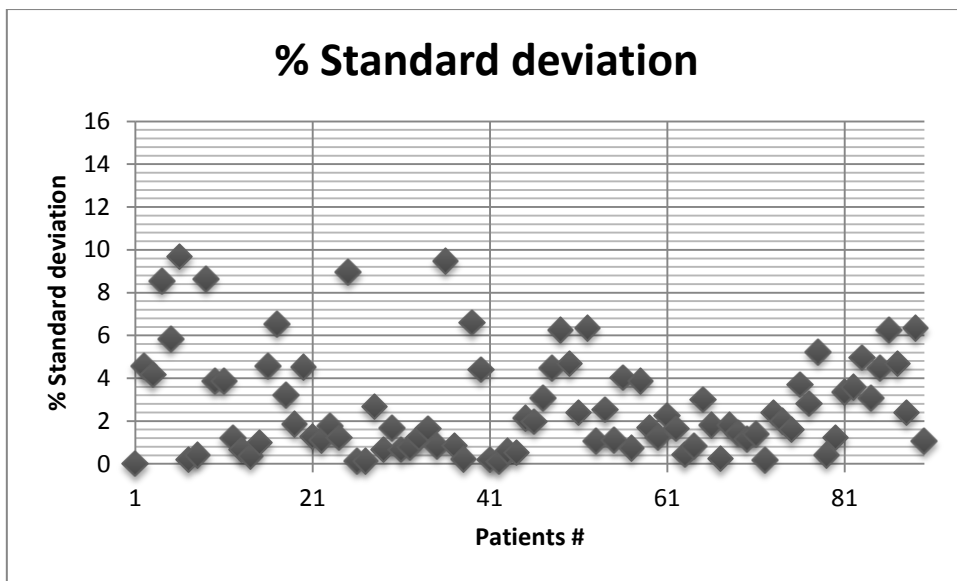
Figure 4.13.   Frequency distribution of deviations from expected dose for measurements for all treatment sites.

CHAPTER FIVE

Discussion, Conclusion and Recommendations

5. 1 Discussion

The calibration for diodes detector products is very stable. The calibration factor can be obtained by finding the ratio of the readings from the ion chamber and diode; this calculation is done automatically by the IVD2 model 1137 system (Sun Nuclear Corporation, USA) software. The calibration factors were verified on a regular basis, because radiation damage affects the diode sensitivity. Besides the calibration factor, which was determined under reference conditions (an SSD of 80 cm , a field size of 10x10cm, and dose of 100cGy at D_{max}, correction factors must be applied for accurate dosimetry.

The results of entrance calibration factors for each diode were presented in separate tables and figures as following. Table (4.1) Shows the results of one minute exposure readings for diodes, and reading take ten times, the recommended value of less than 1%, mean value of ten reading and their percentage standard deviation for all diodes shows in Table (4.2). The stability of the four diodes which taken immediately and after irradiation with mean and standard deviation for all measurements done with the diodes, usually value recommended less than 0.5% as seen in Table (4.3). Linearity of each diode response to dose interval that is typical used in patient's treatment is given in Table (4.5) and Figure (4.2), usually ranges between 0.98 and 1.02., The results obtained do not contradict the results reported in previous literature (Van Dam and Marinello, 1994; Mayles et al., 2000). The originate from the variations in diode sensitivity with the dose per pulse, photon energy spectrum and direction. The responses for gantry angles of 0°, 90°, -90 and 0° were equal, as in all cases the diode was in the same position facing the incident beam. Figure (4.3) showed the response of the diode with respect to the SSD. The diode over responded at SSDs of 70, 80, and 90 cm and exhibited significant perturbations in the response at these SSDs, which decreased with increasing SSD. For the SSDs dependence of open 10 x10 cm fields, the range for CFs dependence from   0.996 to 1.005, in other words, within 4%. Figure (4.4) showed the normalized diode response with respect to gantry angle; the response was normalized with respect to that obtained at a gantry angle of 0°. The maximum and minimum variation found in the angular response with respect to an arbitrary angle of 0° to 90° and -90 to 0° was 2% and 2.14 %, respectively. The dose measured by the diodes was compared with calculated dose, and all doses were

normalized to the dose at a measured field size of 10 x10 cm. From Figure (4.5) it can seen that the maximum reading was at a depth 0.5 cm; as a result buildup cap is suitable for user energy. The normalized diode response with respect to field size was shown in Figure (4.6) for entrance field in-vivo dosimetry, the diode reading increased almost line early with increasing field size, with no significant variation in response, but beyond a field size of 10 x10 cm, the diode reading increased to its maximum variation for 25 x25 cm field size. The minimum and maximum variations between the measured dose from the diode and the calculated dose were -0.6% 9 for a 5 x5 cm field size) and 5.8% 9 for a 20 x20 cm field size), respectively. Before a diode may be used in clinical applications, it was necessary to compare its response with that of a reference detector. The DCF of the Co-60 Isorad diode as a function of field size for open and different standard wedged fields was shown in Figure (4.7), The correction factor (CF) did not change much when the field size changed. For open fields with an SSD 80 cm, the range for CFs was generally within 1%, specifically, 0.8 to 1.2. The change was up to 2% when field size changed. It was found that the DCF did not always increase with increasing wedge angle. The field size dependences for open and 15° and 30° wedges fields were almost the same, but those for 45° and 60° wedges fields larger up to 6%. The diode response was recorded and compared with the calculated dose. It can seen from Figure (4.8) DCF as a function of the tray for entrance measurements with different field sizes (5x5cm to 25x 25 cm), with no significant variation. The entrance calibration factors for each diode in entrance condition are summarized in Table (4.7).

Each diode was individually calibrated and corrected for the entrance dose measurement. We have evaluated stability, linearity calibration and correction factors the results within expected value for this type diodes giving acceptable agreement in dose delivered and the expected dose. The correction factors for every diode were in close agreement with each other and also with correction factors reported in the literature with respect to both magnitude and trend.

Calibration and determination correction factors. The calibration procedures included signal stability after irradiation, which was found to be raining from 0 and 0.1%; intrinsic precision, where the standard deviation of ten readings of each diode was found to be between 0.1 and 0.16%. Dose decreasing under diode at depth of 5% was found to be 5%. The linearity of the diodes was also checked and found to be (1.008, 1.006, 1.007 and 1.005 for diodes. Diode response at different angles was also

checked and found to be from 2.04% to 2.14% with clockwise 0° to 90°, and with opposite clockwise -90° to 0° respectively. The results obtained from this study are similar to the results founded by Shawata. A et al (2015). The correction factor for the set of diodes used in this study is as follows 0.0109715, 0.0106934, 0.0103341 and 0.0099597 respectively. Correction factors CFs for different field size, SSDs, wedges, tray and gantry angles were determined, correction factors were measured as a ratio of chamber and diode reading given condition normalized the reference conditions.

In dosimetry using diodes, many factors can affect the response to radiation. For any diode detector, the sensitivity, reproducibility, correction factors due to the SSD, field size, wedge, radiation damage, and incident beam direction need to be considered. A ddiantiol, the diode sensitivity decreases with increasing cumulative dose. Scattered radiation from both overlying and underlying material might reach the sensitive part of the diode, contributing to the diode readings; the dose could be overestimated or underestimated as these complicating factors are dependent on field size and/or SSD. Therefore, the commissioning or characterization of every diode individually is necessary for accurate dosimetry.

Entrance dose for patients undergoing H& N a number combination of treatment fields such as anterior posterior (AP), Left lateral (LL), Right lateral (R L) open field or with wedge fields such as anterior posterior (AP), Left lateral (LL), Right lateral (R L) open field or with wedge fields, Pelvic and Breast (cancers) radiation therapy on (Co^{60} and SSD technique suet-up) photons beam has been measured using diode IVD system. The Co-60 photons beam has been calibrated according to IAEA TRS- 398 protocol. The IVD system used in this study consisted of four ISORAD n-type diode Model No.1137 (Nuclear Associates, NY, USA). The diode has been positioned onto the skin of the patient in center of the radiation beam in all measurements. All fields have been monitored in first or second treatment fraction. Entrance dose has been calculated at a depth of 0.5cm from entrance surface. Forty nine patients and 96 measurements (22 patients—45 fields H & N, 19 patients—37 fields pelvis, 8 patients—14 fields Breast) were monitored during the period of 3 months. The analysis results of all available measurements expected doses, measured doses, dose deviations, and percentage mean deviations were recorded as means (standard deviation) (SD). Statistical package for Social Sciences t-test online was used for data analysis. The action level is set ±5% for entrance dose as recommended.

A Detailed presented in figure (4.9) shows the distribution of number of field's measurements in different sites in the sample. The values of mean and standard deviation of the distribution of discrepancies between the measured and expected entrance doses are presented and summarized in Table (4.8) together with the percentage of measurements for which the discrepancy was within ± 5% tolerance level.

Table (4.9), (4.10) and (4.11) of appendix (A) Show the average discrepancy for each patients, averaging the values on all the fields used in treating the patients.

The entrance dose measurements were performed for total of 96 treatment fields, on 49 patients over three months period. During the treatment measurements period, the tolerance/action level of ±5% was applied for all fields. Patient diversity was in the sample 22 head and neck cancer patients, 45 (46.87%) fields monitored, 19 pelvic patients., 37 (38.54%) field monitored and 8 breast cancer patients ., 16 (14.58%) fields monitored.

The mean deviations value of the distribution for all measurement was 0.91%, standard deviation was 1.28% and average discrepancy 3.2 ±0.503 (±15.90%), the histogram plotted of frequency distribution of deviation from expected dose for all measurements spread of deviations for breast treatment site was the highest of all treatment sites. The deviations larger than ±5% were detected in 1 cases (1.04%) of all measurement were detected for the breast treatment site.

The mean deviation for head and neck site/technique was 0.51%, standard deviation was 0.72% and average discrepancy was 3.23 ±0.892 (±27.62%). The mean deviation value of the distributions for head and neck site was 0.51% and the standard deviation was 0.72%. This standard deviation was the lowest of all treatment sites and reflects a smaller number of random errors in treatment set up for this site. The histogram plotted of frequency distribution of the deviation of expected dose for measurements on head and neck site shown in Figure 4.9. it was noticed that the histogram distributions for measurements on head and neck site was approximately normal with a narrow spread close to the action level set up., Preparing the patient for the treatment and using good mobilization fixation devices that contribute to reducing the patients movement, which contributes to reducing the percentage of errors that affect the quality of the therapeutic dose given to the patient.

The mean deviation for pelvic site/technique was 1.10%, standard deviation was 1.52% and average discrepancy was 2.8 ±0.595 (±21.58%), while the corresponding standard deviation was 1.52 %. It seen that 98.9% of our results remained within action levels, i.e. within ±5%, our similar to results founded by Gadhi, M, A et al (2016), (2019). The histogram plotted of frequency distribution of the deviation of expected dose for measurements on pelvic cancer patients were plotted in Figue 4.10. It was noticed that the histogram distributions for measurements on pelvic site was approximately normal with a narrow spread.

The smallest group of patients of the three analyzed treatment sites categories was for breast treatment site. The mean deviation for breast site/technique was 1.73% and standard deviation was 2.45% with an average discrepancy 4.12 ±0.945 (±22.93%). The histogram plotted of frequency distribution of the deviation of expected dose for measurements on pelvic cancer patients were plotted in Figure 4.11., the spread of deviations for this measurement site was the highest of all treatment sites. The mean deviations value of the distribution for all measurements was 0.91%, standard deviation was 1.28% and average discrepancy 3.2 ±0.503 (±15.90%), the histogram plot of frequency distribution of deviation from expected dose for all measurements spread of deviations for breast treatment site was the highest of all treatment sites.

The deviations larger than ±5% were detected in 1 cases (1.04%) of all measurements and most of them, 1 (14) cases (7.14%) (14) Were also detected for the breast treatment site. The frequency distribution of deviations from expected dose, for measurements on head and neck site is shown in Figure 4.12.

 The theoretical uncertainty in measuring the entrance dose with diodes, taking into consideration the uncertified in the calibration factor and the correction factors determination and the positioning of the diode. Other sources of uncertainty, which should be taking into account when choosing the tolerance /action levels are output, field size, patient movement during treatment due to berating, possible movement of the patient during the treatment, the use of fixation devices equipment to set-up the patient and the uncertainty in the entrance dose calculation. So, that is the reason why the majority of the radiation therapy center has a ±5% tolerance level for most treatments. In this study the patients have been divided in groups, according to treatment site/ technique, in order to monitoring, investigation and detect the groups for which the uncertainty was larger or for which a systematic error occur.

During treating the patients breast cancer with half-block field technique there was no real field central axis to place the diode so it was decided to place the diode in the position along the beam profile. In position which shifted approximately 2 cm off axis inside the irradiation field. Due to all these facts, it was difficult to place the diode in accurate position for in-vivo measurements. Therefore, any misplacement of the diode caused wrong reading of the diode, and a larger spread of the results. In 1 out of 96 (1.04%) measurements which exceeded the ±5% tolerance was for breast treatment site. The source of the error identified in the most cases was incorrect position of the diode or incorrect SSD%. The (% SD of 2.45%) for breast irradiation was the larger of the three treatment site categories analyzed and reflected a number of errors in both treatments set up and dose measurement technique. Most publications regarding in-vivo dosimtery in tangential irradiation of breast, reported similar standard deviations between the measured and the expected doses Heukelom .S. *et al* .(1991), Shakeshaft *et al* (1999)., in reporting 2 years worth of measurements on 278 breast patients found a mean deviation equal to -2.9% and standard deviation equal to 3.5%. Data similar to ours for breast patient were found by Cozzi *et al* (1998). Which found a mean deviation on 421 measurements equal t0 -1.33% and with standard deviation 0f 2.7%. Appleyard *et al*. (2005) reported a mean deviation on 1073 measurements on breast patient's fields equal to 1.15% and standard deviation was 3.04% (1SD). Fiorino *et al* .(2000) found a mean deviation on 506 measurements equal to 0.1% and standard deviation was 3.5%. Also, it was found that the rate of second checks was significantly higher for breast patients (16/205, 7.8%) against non-breast patients (3/246, 1.2%).

The % SD of 0.72% for head and neck site was smaller than the one seen for other categories and indicates a high level of reproducibility. A number of paper inspection the treatment accuracy by in-vivo dosimetry for patients treated for head and neck cancer and the reported the results similar to our Appleyard *et al*. (2005). Leunens *et al* (1990) reported the data concerning 364 measurements of 47 patients during the brain and head and neck irradiation with a mean deviation around 0% with a SD equal to 2.3% Shakeshaft *et al* (1999) found a mean deviation equal to -0.6% and standard deviation equal to 2.8% for in-vivo measurements on 246 head and neck patient. Our data are similar with the results reported by Appleyard *et al*. (2005), which found a mean deviation equal to 0.35%(2.20%(1SD)) on 326 measurements for brain and

head and neck patient irradiation. Fiorino *et al* .(2000) found a mean deviation for head and neck paitent irradiation equal to 1.0% and standard deviation was 2.8%.

The largest deviations were attributed to a measurement i. a highly oblique wedged fields and difficulty to place the diode into the correct position for in-vivo measurement, because to the patient's body contour line.

In most cases of pelvic cancer patient, the diode is placed in the anterior field of the box technique, in some cases the reason for the inaccuracy of the measurements is due to the patient's lack of cooperation due to culture.

Our results for pelvic cancer patients (N = 37) indicate mean standard deviation was 1.10%., and standard deviation of 1.52% the results within the tolerance levels in all cases. Appleyard *et al*. (2003), (2005), reported a mean deviation on 712 measurements on pelvic irradiation patient's fields equal to 0.52% and standard deviation was 2.75%. Fiorino *et al* (2000)., reported a mean deviation for pelvic irradiation patients equal to 0.8% and standard deviation was 3.0%. Strojnik (2007) found a mean deviation between 0.0 – 1.0% and standard deviation between 2.7 – 3.0 % for in-vivo measurements for radiotherapy patients treated with box field technique during the rectal cancer irradiation.

The overall measurement results for all sites indicated good agreement with the results reported in number of papers Heukelom *et al* (1992), Shakeshaft *et al* (1999) and Fiorino *et al* (2000). The tolerance/ action level of ±5% was put in order to check the possibility to change the initial tolerance/action level for some treatment sites. The results for head and neck treatment site indicated that all results are within tolerance action level ±5%. The out-comes of this study not only provided self confidence that the absorbed dose of radiation was delivered as planned (patients were being treated as per prescribed dose); at the same time other mistakes /errors were noticed as well and were corrected. The main goal of radiation therapy a safe and accurate dose delivery using external beam therapy is to give the patient a carefully controlled dose of radiation to kill the cancer cells inside the affected organ without causing damage to normal tissues. Thus, the increase and decrease in the amount of radiation dose has several risks to the patients recovery rate, as the increase leads to the inability of normal cells to restore themselves and continue to live, and the decrease in the amount of the radiation dose leads to the re-growth and spreads of cancer cells, their a activity again and impedes the achievement of the main goal of radiation therapy

recovery or improve the patients health level. And based on the technical advantage of diode includes small size, bias less, cost- effectives and the immediate results that can facilities the rectification of variation observed (if patients and necessary too), even though the patient was on the treatment couch or during following fractions. The analysis of all available (96) measurements showed a mean standard deviation of (average discrepancy 3.2% ±0.504 (±15.87%) with SD of 1.28%. it was seen that 98.95% of our results remained within action level (±5%) (ICRU, 1976) were detected. This indicates that the combined uncertainty of treatment deliver and in-vivo dosimetry at NCIS- Libya is 1.28%.

5.2 Conclusion

In summary, in vivo dosimetry is an effective method for detecting radiotherapy errors, assessing clinically relevant differences between the prescribed and delivered doses, reducing potential patient harm, and fulfilling requirements set forth by national and international regulations. The pilot study to test the applicability of a diode dosimetric system for performing in vivo entrance dose measurements in external photon beam radiotherapy presented good results. These measurements demonstrated the value of diode dosimetry as a treatment verification method and its applicability as a part of a quality assurance program in radiotherapy.

To summarize in-vivo dosimetry has given the full confidence that patients treated with the prescribed and planned dose.

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5.3 Recommendations

- In vivo dosimetry is useful tool in quality assurance program of radiotherapy department thus the recommendation is to perform it for each patient at least in first fraction. Treatment error discovered through in vivo dosimetry should be discussed with therapist team, and, in special, handled as are any other treatment error at the clinic, discrepancies exceeding 5% are immediately should be reviewed.

- Encouraging and helping radiotherapy departments in Libyan Medical Centers and hospitals to perform and implement in-vivo dosimetry as part of their quality assurance.

- Conducting iv-vivo measurements that include both entrance and exit surface dose as more information can be inferred compared with those obtained using entrance dose only.

- Their main advantage over other detectors, such as TLDs, is a possibility of immediate readout and detection of errors while patient is still on a treatment couch. Moreover, diodes are known for their high sensitivity, small size, simplicity of operation and mechanical stability, Therefore, it is the best option we recommend their use.

- Implementation of specialized training programs for staff in the quality assurance program within the departments of radiation oncology.

- Conducting more studies on the subject of study for its importance.

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APPENDIX A

AVERAGE DISCREPANCIES IN DOSE AND CORRESPONDING STADARD DEVIATIONS

Table 4.9 Shows Average discrepancies in dose and corresponding % standard deviation and % mean standard deviation for head and neck cancer treatments

No: Fields	Average Discrepancy %	Standard deviation SD%	Mean SD %
1	- 6.9	4.57	3.23
2	- 6.3	4.16	2.94
3	-7.5	8.53	6.03
4	5.1	5.82	4.11
5	- 8.4	9.68	6.84
6	0.3	0.2	0.14
7	- 0.7	0.41	0.29
8	-7.6	8.63	6.1
9	1.68	3.87	2.74
10	-7	3.87	2.74
11	2.1	1.21	0.85
12	-1.2	0.67	0.48
13	-0.5	0.36	0.25
14	-1.4	1	0.71
15	-4.5	4.56	3.22
16	-6.4	6.52	4.61
17	5	3.2	1.62
18	2.1	1.86	1.31
19	6.9	4.51	3.19
20	1.9	1.27	0.89
21	1.8	1.12	0.79
22	-2.5	1.75	1.24
23	1.8	1.23	0.86
24	14.2	8.95	6.33
25	-0.2	0.13	0.1
26	-0.2	0.13	0.1
27	4	2.67	1.89
28	-1	0.67	0.48
29	-2	1.67	1.18
30	1	0.67	0.47
31	-0.1	0.79	0.05
32	1.3	1.22	0.86
33	3.5	1.63	1.15
34	0.8	0.83	0.59
35	-8.7	9.46	6.68
36	-1.2	0.85	0.60
37	-0.3	0.23	0.16
38	6	6.6	4.64
39	4.1	4.4	3.12
40	0.3	0.2	0.14
41	-0.6	0.1	0.05
42	0.1	0.6	0.05
43	0.5	0.52	0.36
44	3.35	2.14	1.51
45	3.1	1.98	1.40

Table 4.10 shows average discrepancies in dose and corresponding % standard deviation and % mean standard deviation for pelvic cancer treatments.

No: Fields	Average Discrepancy %	Standard deviation SD%	Mean SD%
1	-3.6	3.07	2.17
2	-5	4.46	3.15
3	-3.7	6.25	4.41
4	-2.8	4.67	3.30
5	3.4	2.38	1.68
6	8.9	6.33	4.5
7	1.2	1.05	0.74
8	-3	2.54	1.8
9	-1.2	1.1	0.74
10	-4.5	4.03	2.84
11	1	0.72	0.51
12	-4.9	3.87	2.73
13	-2.1	1.70	1.20
14	-1.6	1.24	0.87
15	-2.5	2.25	1.60
16	-1.8	1.64	1.16
17	-0.5	0.43	0.30
18	-1	0.82	0.58
19	-3.7	3	2.13
20	-2.2	1.8	1.26
21	-0.3	0.24	0.12
22	-2	1.82	1.30
23	2-	1.44	1
24	-1.6	1.16	0.82
25	1.6	1.39	0.98
26	0.2	0.18	0.13
27	-3	2.4	1.7
28	-2.6	2	1.43
29	-2	1.6	1.11
30	-4.6	3.7	2.6
31	-3	2.8	1.7
32	-7	5.22	3.69
33	0.3	0.41	0.3
34	-1	1.23	0.9
35	-4	3.35	2.37
36	-4.6	3.6	2.53
37	3	4.97	3.51

Table 4.11 shows average discrepancies in dose, corresponding % standard deviation and % mean standard deviation for breast cancer treatments.

No: Fields	Average Discrepancy %	Standard deviation SD%	Mean SD%
1	-6.2	6.98	4.93
2	-6.76	7.63	5.39
3	5.2	6.36	4.5
4	6.4	5.6	3.57
5	-5.3	15.17	10.72
6	-2.1	6.12	4.32
7	-4.6	5.12	3.62
8	2.7	2.25	1.59
9	4	3.66	2.59
10	5	4.72	3.34
11	0.7	0.45	0.31
12	-2.6	1.73	1.22
13	2	1.35	0.95
14	4	2.68	1.89

APPENDIX B

International Journal of Scientific and Research Publications
ISSN 2250-3153

Certificate of Publication

This is to certify paper titled **"Calibration and Correction Factors of ISO-Rad Diodes P-Type for In-Vivo Dosimetry in External Beam Therapy with Co60"** submitted by Author(s) **Nureddin. A. S. Musa, Pro. Mohamed Elfadil. M** has been published for January 2021, Volume 11, Issue 1 publication under ISSN 2250-3153.

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APPENDIX C

Certificate of Publication

This is to certify paper titled **"Verification of Absorbed Dose using Diodes in External Beam Radiation Therapy Cobalt-60"** submitted by Author(s) **Nureddin. A. S. Musa, Pro. Mohamed Elfadil. M** has been published for January 2021, Volume 11, Issue 1 publication under ISSN 2250-3153.

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APPENDIX D

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الموافق: / /

السيد/ مدير المعهد القومي لعلاج الأورام/ صبراته

بعد التحية ،،

نفيدكم الشؤون الثقافية والأكاديمية بالسفارة الليبية الخرطوم بان الطالب/نور الدين احمد سعد موسى

والمقيد ببرنامج الدكتوراه تخصص الفيزياء الطبية بجامعة السودان للعلوم والتكنولوجيا ،وبناء، علي كتاب

مسجل الكلية والتي تتطلب بأن الطالب المذكور أعلاه يتواجد داخل ليبيا من أجل تجميع المعلومات

المتعلقة بموضوع دراسته.

عليه.. تأمل منكم التعاون مع الطالب وتسهيل كافة إجراءاته.

وذالاوام مطلبكم دوحة والثه تبارك وتعالى

د. فتحية عبدالحميد دلاف

الملحق الأكاديمي

صورة

السيد / القائم بالاعمال

الملف الســـــــــــدوري

بر / فتحية   ت: رانلدا

APPENDIX E

المعهد القومي للأورام صبراته
وحدة البحوث

طلب الموافقة على إجراء بحث

- اسم الباحث: نور الدين أحمد سعد
- المؤهل العلمي: ساجستير تمريض باطني جراحة ..... الوظيفة: اخصائي جراحة
- الجهة التابع لها: جامعة أطلس للعلوم .. القسم: التمريض العلمي
- الهاتف: 0.9.2/.519.4.76.1 .... البريد الالكتروني: nureddine.2001@yahoo.com
  0911 519 4761

أرجو الموافقة و المساعدة على إجراء البحث والذي هو:

☐مشروع تخرج   ☐ استكمال رسالة دبلوم   ☐ماجستير   ☑دكتوراه

- عنوان البحث: تأثير برنامج تثقيفي على مرضى سرطان ....
- اسم المشرف على البحث: P.د.محمد بلال محمد ..... الوظيفة: استاذ جامعي
- الجهة المطلوب الحصول منها على البيانات (مكان إجراء البحث): المعهد القومي للأورام ....
- المشاركين في البحث :

المؤهل العلمي	الاسم
/	/

تعهد

أتعهد أنا مقدم الطلب نور الدين أحمد سعد ..... الموقع أدناه بتسليم نسخة من النتائج النهائية للبحث فور الانتهاء منها إلى إدارة وحدة البحوث خلال .... مع التزامي بالقوانين والتوقيت المحدد للعمل داخل المعهد.

قسم العلاج الاشعاعي

التوقيع والتاريخ ....

نأمل مساعدة المعني لاجراء كشر سرطان المريء واستبيان

ملاحظة: الرجاء إحضار صورة من الرسالة الرسمية من الجهة المشرفة المكلفة بإجراء البحث للحفظ.

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