Characterization of p-type silicon semiconductor detectors for use in in vivo dosimetry

Kamran Ul Haq

University of Nevada, Las Vegas

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CHARACTERIZATION OF \textit{p}-TYPE SILICON SEMICONDUCTOR DETECTORS FOR USE IN \textit{IN VIVO} DOSIMETRY

by

Kamran Ul Haq

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A thesis submitted in partial fulfillment of the requirements for the

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KAMRAN UL HAQ

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Examination Committee Chair

Dean of the Graduate College

Examination Committee Member

Examination Committee Member

Graduate College Faculty Representative
ABSTRACT

Characterization of p-type Silicon Semiconductor Detectors for Use in In Vivo Dosimetry

by

Kamran Ul Haq

Dr. Steen Madsen, Examination Committee Chair
Assistant Professor of Health Physics
University of Nevada, Las Vegas

The response of a p-type diode detector for use in patient dose monitoring in photon and electron fields was evaluated for a number of commonly-encountered clinical parameters. Effects of energy, field size, source-to-surface distance, presence of beam modifiers, beam orientation, and temperature were investigated. Shielding effects and batch variations were also examined. In most cases, diode performance was in good agreement with manufacturer's specifications and/or the results of others using similar detectors. Variations in excess of ±12% were observed in high-energy wedged fields over the source-to-surface distances investigated. Reductions in local dose directly behind the diode due to electron shielding ranged from 8 to 12%. Entrance and exit dose measurements were made on patients undergoing treatment for lung or prostate carcinoma. Differences in measured and calculated exit doses could be improved by the inclusion of tissue heterogeneity correction factors in the treatment planning algorithm.
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CHAPTER I

INTRODUCTION

1.1 Radiation Therapy

In the early days of radiation therapy, radiation-induced chemical and biologic effects provided a crude measure of radiation dose. Reddening of the human skin was related to the amount of radiation absorbed, and changes in color of various chemical compounds on photographic emulsions were indicative of radiation effects. When orthovoltage machines were used skin was the limiting organ to the delivery of tumor doses, skin erythema dose (SED) was used as a dosimetric parameter. The SED is the amount of x- or γ-radiation required to produce reddening of human skin (Khan 1994). For dosimetry purposes, this unit is somewhat inadequate since it depends on numerous parameters including dose fractionation, differences between early and delayed skin reactions, quality of radiation, and skin type. When megavoltage beams with skin-sparing properties became available for radiotherapy, the reliance on skin reaction for the assessment of radiation effects had to be abandoned in favor of more precisely measurable units, such as the Roentgen which is used for exposure only (Washington and Leaver 1996).

In present day radiotherapy, unexpected skin reaction is investigated carefully because it provides an indication of gross calibration errors and alerts the Radiation
Oncologist. It also gives a clear indication of the accuracy and reproducibility with which a treatment field has been set up. The most obvious way to check the accuracy of a patient's treatment is through direct measurement of the intended dose. Treatment doses need to be accurate to within ±5% of the calculated dose according to World Health Organization standards (World Health Organization 1988). In vivo dosimetry allows verification of external beam treatment fields that have been optimized for patient treatment according to a dose distribution plan (Bentel 1992). By implementing an in vivo dosimetry program in a clinical setting, setup errors, such as the omission of block trays, wedges, or other beam-modifying devices, incorrect energy selection, or patient positioning can be caught early in a course of treatment and corrected before a misadministration occurs. The principal detectors used for in vivo measurements are thermoluminescence dosimeters (TLDs), metal oxide semiconductor field effect transistors (MOSFETs), thimble ionization chambers, and silicon semiconductor detectors (Li and Tom 1995).

1.2 Thermoluminescence Dosimeter

In thermoluminescence dosimetry, electrons may be trapped in metastable states in certain materials when provided with sufficient energy (Hussey, Shi and Saw 1998). These metastable electrons escape from their unstable state by heating the material, and in reverting to a stable state, an optical photon is emitted which can be detected with a photomultiplier tube. After the process of annealing, to eliminate any residual thermoluminescent signal, dosimeters can be re-used. There are many physical and chemical forms of thermoluminescent materials. Lithium fluoride is the most common material used in radiotherapy since it is approximately tissue...
equivalent (Shani 1991). The two most common forms are chips of lithium fluoride mixed with Teflon and microrods made of extruded material. Advantages of TLDs include small size, low cost, no electrical cables, easily inserted into cavities, and record many simultaneous results. With these devices there is no permanent record, accurate results require care, and readout and calibration are time-consuming and have poor precision (Metcalfe, Kron, and Hoban 1997). TLDs have been used in numerous in vivo dosimetry applications; however, their main use is in the measurement of eye doses, total body irradiation doses, and in solid phantom measurements (Rikner and Grusell 1986).

1.3 Metal Oxide Semiconductor Field Effect Transistors (MOSFETs)

Metal oxide semiconductor field effect transistors (MOSFETs) are direct-reading semiconductor dosimeters (Bower and Hintenlang 1998). They can be applied to small-beam dosimetry measurements due to their miniature size - the active region is about $1 \mu m \times 200 \mu m \times 200 \mu m$ (Kaplan et al. 2000). The MOSFET detector consists of a $p$-type silicon semiconductor substrate separated from a metal gate by an insulating oxide layer forming a sandwich-type device. When exposed to ionizing radiation, electron hole pairs are formed in the oxide insulation layer. Due to the effect of applied bias, electrons travel to the gate while holes migrate to the silicon where they are trapped in a silicon-oxide interface. A negative voltage shift is produced by the trapped positive charges that allows current to pass through the MOSFET. This shift in voltage is directly related to the radiation dose delivered. MOSFETs and TLDs have similar principles of operation.
However, from a clinical perspective, the main advantage of MOSFETs is that they can be read electronically immediately after irradiation (Gladstone, Lu, Humm, Bowman and Chin 1994). Other advantages of MOSFETs include small size, lightweight, and accuracy at low doses (Metcalfe et al. 1997). The primary disadvantage of MOSFETs is the significant signal drift following irradiation. These dosimeters are also dependent on energy, temperature, and direction of the incident radiation (Heukelom, Lanson, and Mijnheer 1991). Furthermore, since MOSFETs have no build-up region, electronic equilibrium is not readily achieved, thus limiting their use to surface dose measurements. MOSFETs are used primarily in in vivo skin dose measurements with electron beams and in vivo rectal dose measurements with brachytherapy (Metcalfe et al. 1997).

1.4 Thimble Ionization Chamber

The thimble ionization chamber is an instrument employed in the measurement of the exposure (Knoll 1989). It is a spherical air wall compressed into a solid shell with an air cavity at the center. The distance between the outer sphere and inner cavity is equal to the maximum range of secondary electrons generated by photon interactions in the sensitive volume of the detector. If the number of these electrons entering the cavity is the same as that leaving the cavity, electronic equilibrium exists. By knowing the volume or mass of air inside the cavity, the charge per unit mass can be calculated. Since the density of a solid air equivalent wall is much greater than that of free air, the thimble ionization chamber considerably reduces the thickness required for electronic equilibrium. For the thimble ionization chamber to be air equivalent, the effective atomic number of the
wall material and central electrode must be such that the system behaves like a free-air chamber. Graphite is the most common wall material used. A thimble ionization chamber could be used to measure exposure "X" by (Khan 1994):

\[
X = \frac{Q}{p \cdot v} \cdot \frac{1}{A}
\]

where "A" is the fraction of the energy fluence transmitted through the air equivalent wall, "v" is the volume, "p" is the density of the cavity air, and "Q" is the ionization charge liberated.

For accurate dose determination in a clinical setting, thimble ionization chambers should have the following characteristics: minimal energy and direction dependence, suitable volume to allow measurements for the expected range of exposures, and minimal stem leakage and ion recombination losses (Johns and Cunningham 1983). The ionization chamber is connected to a direct integrating multichannel electrometer for the measurement of charge. The electrometer readout can be correlated to absorbed dose if the appropriate calibration procedures have been performed. However, care must be exercised as readings can be affected by changes in air temperature and pressure. According to gas law, the density of air depends on the temperature and pressure; therefore, the density of air inside the chamber volume will also depend on the atmospheric conditions. The reading for a given exposure will increase as the temperature decreases or as the pressure increases since the exposure is given by the ionization charge per unit mass of air. For other than reference conditions (1.01 X 10^5 Pa, 22°C), it can be corrected by

\[
C_{corr} = \frac{760}{P} \times \frac{273 + t}{295}
\]
where $P$ is pressure in mmHg and $t$ is temperature in °C. Thimble ionization chambers are precise, accurate, reliable, and provide an instant readout (Knoll 1989). All the necessary corrections are well understood. The disadvantages of ionization chambers include large measurement volume, fragility, a connection cable is needed, high voltages are required, and dependence on temperature and pressure (Johns and Cunningham 1983). Due to these drawbacks, ionization chambers are rarely used in routine clinical patient dosimetry. Their use is limited mainly to machine output calibrations.

1.5 Semiconductor Diodes

The use of semiconductors as radiation detectors dates back to the early 1960s. However, it was not until 1977 that they were first used clinically to verify patient doses during radiation therapy (Heukelom et al. 1991). Semiconductors have several properties that make them ideally suited for in vivo dosimetry, including good mechanical stability, absence of external bias, high sensitivity to radiation, robustness, negligible recombination effects, short-measuring time due to rapid signal rise time, and air pressure independence (Rikner and Grusell 1987). Furthermore, since the active volume of these detectors is very small, they may be used in dosimetric applications requiring fine spatial resolution (Becker 1973). Limitations of silicon semiconductors include their dependence on several clinically relevant parameters such as temperature, dose rate, beam direction, and energy. It is important to investigate the extent of the effects of these parameters on diode output prior to clinical use.
A silicon semiconductor is a substance that has electrical conducting properties between an insulator and a conductor. The periodic lattice of crystalline silicon establishes energy bands for electrons that exist within that solid. Gaps, or ranges of forbidden energies, separate the energy of electrons within pure materials. In other words, electrons are confined to specific bands within the crystal. The "valence band" corresponds to a lower band in which the electrons are bound to specific lattice sites within the crystal. Electrons in this band contribute to the interatomic forces within the crystal. A "conduction band," which is a higher lying band, represents electrons that are free to migrate through the crystal (Figure 1). This band contributes to the electrical conductivity of the material. The extent of the gap which separates the two bands determines whether the material is classified as a conductor, a semiconductor, or an insulator. In the absence of excitation, both insulators and semiconductors have a configuration in which the conduction band is completely empty and the valence band is completely full. In order to generate a current (i.e., produce a signal), the electron must be given sufficient energy to cross the band gap and enter the conduction band. When this occurs, a hole is created in the valence band by the vacated electron. The resultant electron-hole pair is analogous to the ion pairs formed in air ionization chambers. In the case of semiconductors, application of an electric field (bias voltage) will cause migration of the charge carriers; holes in the valence band will move in one direction, while electrons in the conduction band will move in the opposite direction. It is the migration of the charge carriers, in the presence of an applied field, that generates the signal, which is proportional to the dose of incident radiation. For
Fig. 1 Insulators and semiconductors band structure for electron energies.

Fig. 2 n-type silicon semiconductor band gap corresponding donor levels.

Fig. 3 p-type silicon semiconductor band gap illustrating acceptor levels.
semiconductors, the band gap is 1.12 eV-3.5 eV, whereas for insulators, the band gap is usually 5 eV or more.

In a pure semiconductor, the number of electrons in the conduction band is exactly equal to the number of holes in the valence band. In practice, this is virtually impossible to achieve due to the presence of impurities. In fact, the electrical properties of real materials are dominated by the very small levels of residual impurities. In practical applications, the electrical properties of semiconductor materials are manipulated through the addition of small amounts of known impurities. Depending on the nature of the impurity, silicon semiconductors are commonly classified as $n$-type or $p$-type.

1.5.1 The $n$-type Semiconductor

In normal crystalline structures, tetravalent silicon forms covalent bonds with its four nearest neighbors. Absorption of energy may cause one of the covalent electrons to break loose and leave behind an unsaturated bond or hole. Addition of an impurity from Group V of the periodic table (e.g., phosphorous, arsenic, antimony, or bismuth) results in an excess of electrons after the pentavalent impurity has formed covalent bonds with surrounding Si atoms. Since this extra electron remains very lightly bound to the original impurity site, very little energy is required to dislodge it to form a conduction electron without a hole (Figure 2). Because they contribute electrons to the conduction band, they are referred to as donor impurities. The extra electrons associated with donor impurities can occupy a position within the normally forbidden gap because they are not part of the regular lattice. In almost all cases, the concentration of impurity is large compared with the concentration of
electrons expected in the conduction band for the intrinsic material. Therefore, the
number of conductor electrons is dominated by the contribution from the donor
impurities. The end result is that the conductor electrons greatly outnumber the
holes. Thus, in n-type semiconductors, the conductivity is determined almost
exclusively by the flow of electrons; holes play only a minor role. The electrons and
holes are referred to as majority and minority carriers, respectively.

1.5.2 The p-type Semiconductor

In this type of semiconductor (Figure 3), an element such as boron,
aluminum, gadolinium or indium from Group III of the periodic table is added to a
silicon lattice. Since this impurity has one less valence electron than the surrounding
silicon atoms, one covalent bond is left unsaturated. This vacancy represents a hole
similar to that left behind when a normal valence electron is excited to the
conduction band. An electron filling this vacancy (hole) is less firmly attached than
a typical valence electron. As a result, these acceptor impurities also create electron
sites within the normally forbidden energy gap. Since the energy difference between
the acceptor sites and the valence bond is relatively small, electrons can easily be
excited into the acceptor site. This results in the creation of holes in the valence
band. If the concentration of acceptor impurities is large compared to the intrinsic
concentration of holes, then the number of holes is completely dominated by the
concentration of acceptors. In p-type semiconductors, the electrical conductivity is
determined by holes which are classified as the majority carriers.

1.5.3 The Semiconductor Junction

Silicon diode detectors are based on the properties that are formed near the
junction between p- and n-type semiconductor materials. When "p" and "n" regions
are brought together in thermodynamic contact, charge carriers are able to migrate across the junction. The junction is normally created by causing a change in the impurity content from one side of the junction to the other. First, a p-type crystal is doped with a uniform concentration of acceptor impurity on one side. Then a minute amount of n-type donor impurity is added to the surface of the crystal on the opposite side that diffuses some distance into the crystal. The donor impurities are made to outnumber the acceptors which are close to the surface converting the other side of the crystal to n-type material. A net diffusion from regions of high concentration to those of low concentration occurs when the density of conduction electrons is much higher in the n-type region than in the p-type. The n-type material leaves behind immobile positive charges after the diffusion of conduction electrons. Holes in the p-type material also diffuse across the junction. The net effect creates a positive space on the "n" side and a negative space charge on the "p" side of the junction. In the depletion region, the charge imbalance exists and expands into both the "n" and "p" sides of the junction. When the donor concentration in the n-type material is higher than that of acceptor atoms in the p-type, electrons will tend to travel a greater distance into the p-type material, diffusing across the junction before all have combined with holes.

A p-type p-n junction diode is used in this study. This type of junction will readily conduct current when voltage is applied in the "forward" direction. A positive voltage is applied to the "p" side of the junction with respect to the "n" side. The potential will attract conduction electrons from the "n" side as well as holes from the "p" side across the junction. This is the direction of forward biasing.
However, in the reverse bias, the "p" side is made negative with respect to the "n" side and it conducts very little current. The potential difference is enhanced from one side of the junction to the other. The p-n junction allows relatively free flow of current in one direction while presenting a large resistance to its flow in the opposite direction. Because its resistivity is much higher than that of normal n- and p-type material, virtually all the applied voltage will appear across the depletion region. When the ionizing radiation is incident on the detector, electron-hole pairs are created in the depletion region and are swept from the region by an electric field; their motion causes a current to flow. The resulting charge is monitored by an electrometer.

1.6 Clinical Rationale for p-type Detectors

There are numerous parameters that affect the characteristics of semiconductor diode detectors. Since the diode signal may be very sensitive to changes in these parameters, it is important to investigate the extent of these dependencies prior to clinical use.

Perhaps the most important parameter affecting diode reading is radiation dose. Semiconductor diodes are prone to damage from ionizing radiation due to the displacement of Si atoms from their lattice sites (Rikner and Grusell 1987). The amount of damage depends on total dose, radiation quality, and semiconductor type. As a result of radiation damage, diode sensitivity drops. Thus, as radiation damage accumulates with total dose over time, the diode signal decreases. Periodic recalibration is thus required due to the decrease in diode sensitivity with accumulated dose. Prior to clinical use, diodes are commonly pre-irradiated to 10
kGy to overcome the initial severe sensitivity decrease. The sensitivity dependence is especially severe for $n$-type semiconductors, particularly when used in pulsed radiation beams. Under these conditions, $n$-type detectors suffer significant drops in sensitivity and rapidly become non-linear with respect to dose rate as a result of radiation damage (Rikner and Grusell 1987).

The primary reason for using $p$-type silicon semiconductor detectors is that they are relatively insensitive to the effects of pulsed radiation beams. Furthermore, they exhibit a much slower sensitivity decrease after pre-irradiation than $n$-type detectors. Thus, from a clinical point of view, $p$-type detectors are convenient for a number of reasons: they can be used in pulsed beams; they have a longer lifetime than $n$-type detectors; and they do not have to be calibrated as often.

$p$-type detectors can also be tailored to specific applications since the amount of radiation damage can be controlled, to some extent, by the doping level. For example, a high-doped $p$-type detector shows linearity for a high dose per pulse, whereas, a low-doped $p$-type detector remains non-linear in high energy beams (Rikner 1985).

1.7 Dose Verification with $p$-type Diodes

Since water is the main constituent of human tissue, dosimetric data acquisition for use in computerized planning systems is performed in water phantoms. Dosimetry parameters of clinical interest include depth dose profiles, beam flatness, and symmetry. From a treatment planning perspective, patients are essentially treated as homogeneous water-like objects (Khan 1994). This is not a bad approximation because, at megavoltage energies, Compton interactions dominate.
The probability of a Compton interaction is strongly dependent on the electron density of the material (Khan 1994). For most tissues, electron densities do not vary by more than a few percent (Khan 1994). There are a few exceptions however; air-filled tissues such as lungs have low electron densities, while bone has high electron densities. Failure to account for these discrepancies in the treatment planning calculations will result in erroneous dose distributions (Metcalfe et al. 1997). The magnitude of the discrepancy will depend on the volume of the heterogeneity (e.g., lung, bone) encompassed by the treatment fields (Fletcher 1973). Accurate localization of these inhomogeneities is accomplished by Computed Tomography (CT). Lung, bone and other inhomogeneities are contoured on CT images and in some cases, inhomogeneity correction factors are assigned. Typical correction factors for lung and bone tissue are 0.25 and 1.25, respectively (Khan 1996). By comparison, soft tissue has a correction factor of unity (i.e., no correction is performed). In order to achieve dose uniformity to target volumes in lung treatments, parallel opposed fields, such as anterior-posterior (AP) and posterior-anterior (PA), are commonly employed. A four-field technique (AP, PA, right lateral, and left lateral) is typically used in the treatment of prostate patients, a treatment in which significant volumes of bone tissue are irradiated.

In addition to the discrepancies resulting from a failure to account for tissue inhomogeneities in the treatment field, differences between measured doses and doses calculated near the patient surfaces may occur due to electron contamination from field-shaping blocks and beam-modifying devices such as wedges. The magnitude of
this contamination depends strongly on the source-to-surface distance (SSD); surface doses increasing with decreasing SSD.

In order to verify intended dose to the patient at different depths, the dose verification procedure is needed which consists of entrance and exit dose measurements. In general, the diodes are placed in the field at the central axis on the surface of the patient to determine the entrance and exit doses. The entrance and exit doses are defined as the doses at points found at distances equal to the depth of maximum dose \( d_{\text{max}} \) from the patient's entrance and exit surface, respectively. The depths of maximum dose for 6 and 15 MV photons are 1.5 and 2.8 cm, respectively. Entrance and exit doses obtained from diode measurements on the patient's surface should correspond to the calculated entrance and exit doses obtained with the treatment planning computer. Large discrepancies between measured and calculated doses may occur if the treatment planning system does not account for inhomogeneities in the treatment field. In addition, discrepancies may result due to the dependence of the diode response on a number of treatment parameters such as: energy, field size, temperature, beam orientation, source-to-surface distance, and the presence of beam-modifying and shaping devices.

In summary, significant discrepancies between measured diode doses and doses calculated by the computer planning algorithm may not be indicative of a dose misadministration. It may be due simply to a failure to adequately account for tissue inhomogeneities and/or a lack of understanding of how diode response is affected by various treatment parameters. In both cases, application of appropriate correction factors will make dose misadministration more apparent. In the vast majority of
treatments, a ±5% agreement between measured and calculated doses is readily achievable using carefully determined correction factors.

1.8 Thesis Statement

This thesis is comprised of two distinct projects: (i) evaluation of the response of a $p$-type silicon semiconductor diode detector to various treatment conditions typically encountered in the clinic, and (ii) determination of the effects of tissue inhomogeneities on diode detector response during treatments of lung and prostate patients.

The purpose of the first part of this thesis is to evaluate diode response as a function of various parameters including: source-to-surface distance, energy, field size, temperature, beam orientation, and the presence of beam-modifying devices. Furthermore, as a check on consistency, the response of identical diodes from two different batches is compared. Finally, the potential of underdosage to tissues directly beneath the diode is investigated using film dosimetry.

In the second part of the thesis, measured entrance and exit doses are compared to predictions of the treatment planning algorithm for representative lung and prostate patients. These patients were chosen because air in lungs and the thick bones in the pelvis are known to cause deviations in entrance and exit doses. In all cases, patients were treated according to calculations assuming tissue homogeneity. Sometime after treatment, dose distributions were re-calculated using appropriate tissue correction factors for lung or bone. The extent of the discrepancy between measured and calculated doses are investigated with and without tissue heterogeneity correction factors.
CHAPTER II

MATERIALS AND METHODS

2.1 Materials

2.1.1 p-type Si Semiconductor Diodes

Newly-designed QED (Sun Nuclear Corporation) model 1112 (electrons), 1115 (6 MV x-rays), and 1116 (15 MV x-rays) p-type silicon semiconductor diode detectors were evaluated since they cover most of the needs in the radiation therapy clinic. Each diode is specifically designed to serve as an in vivo dose verification tool for external photon and electron beam therapy.

For 6 and 15 MV photons, two diodes of total brass buildup of 1.85 and 3.05 g/cm$^2$ respectively were used. A single type of diode made of acrylic material with total buildup of 0.30 g/cm$^2$ was utilized for all electron energies (6, 8, 9, 10, 12, and 14 MeV). All three diode models have the same proprietary p-type silicon p-n junction where diode die is surface-mounted in FR4 glass reinforced epoxy resin (Figure 4). The die plane is parallel to the circular area and located in the center. According to specifications, all QED diodes have an effective detection thickness of 50 $\mu$m and are 2.3 mm thick at the measurement point. All diodes have a flat bottom for easy placement on the patient.
Fig 4. Schematic top view and cross section of QED detector (Sun Nuclear Corp, 1997). ‘D’ and ‘h’ are the height and bottom surface diameter of the buildup, respectively.
2.1.2 Ionization Chamber

A Farmer-type ionization chamber (Capintec PR-06G) was used to verify diode response at \( d_{\text{max}} \) in a water-equivalent phantom.

2.1.3 Electrometer

Victoreen (37-721) and (37-720) multichannel electrometers by Nuclear Associates were used to measure all diode readings. Ionization chamber measurements were obtained using a Capintec 192A electrometer.

2.1.4 Radiation Source

A Siemens Mevatron (MD 80) at Nevada Radiation Oncology Centers in Las Vegas, Nevada, was used to produce 6 and 15 MV x-rays and electrons of energies 6, 8, 9, 10, 12, and 14 MeV for all irradiation measurements.

2.1.5 Phantoms

A 15 cm thick (30 cm X 30 cm) solid-water Poly (Methyl Methacrylate) phantom was used in all measurements except those involving temperature dependence. Solid water is composed of an epoxy resin compound that is approximately water-equivalent. A 15 cm X 15 cm X 15 cm liquid-water phantom was used to investigate the effect of temperature on diode response.

2.1.6 Beam Modifiers

Acrylic block trays (0.8 cm thickness) and metallic wedge filters (15°, 30°, 45°, and 60°) were used as beam modifiers. Block trays are used to support customized field-shaping blocks, while wedge filters compensate for missing tissue or overlapping beams.
2.2 Methods

2.2.1 Energy Dependence

The diode response with energy was determined for all six electron energies (6, 8, 9, 10, 12, 14 MeV). The diode was taped to the surface of a solid water phantom at the central axis and measurements were taken in a 10 cm X 10 cm field at a SSD of 100 cm. Readings were repeated with an ionization chamber inserted in the solid-water phantom at $d_{\text{max}}$ for each specific energy using the same parameters. For each electron energy, the diode was connected to the 6 MeV channel of the electrometer. This channel was chosen randomly in order to determine energy response of the diode. A total of three separate measurements were taken to confirm consistency. Diode measurements were corrected at each energy for the difference in the position of the effective point of measurement on the depth dose curve. The irradiation geometry is illustrated in Figure 5.

2.2.2 Field Size Dependence

Output factors were measured by taping each diode separately to a (30 cm X 30 cm X 15 cm) PMMA solid water phantom. Output factors are defined as the ratio of the maximum absorbed dose on the central axis for a 10 cm X 10 cm field to an (X cm X Y cm) field for the same given dose. At 100 cm SSD, output factors relative to a 10 cm X 10 cm field were measured for photon (6 and 15 MV) and electron (6, 8, 9, 10, 12, 14 MeV) beams for field sizes of 4 cm X 4 cm, 5 cm X 5 cm, 10 cm X 10 cm, 15 cm X 15 cm, 20 cm X 20 cm, and 25 cm X 25 cm. For comparison purposes, the field size dependence of an ionization chamber was also investigated. A Capintec PR-06G air ionization chamber was placed in a solid water
Fig. 5 Cut away view of the head of a typical medical accelerator and standard setup for silicon diode detector and ionization chamber.
phantom at the depth of maximum dose and readings were recorded as a function of field size. Each measurement was repeated two to three times in order to check output consistency.

2.2.3 Directional Dependence

The diode was placed on the surface of a 15 cm X 15 cm X 15 cm water phantom to determine its directional response (Figure 6). Measurements were taken at 100 cm SSD in a 10 cm X 10 cm field with the diode’s long axis perpendicular to and parallel to the direction of gantry rotation. Photon energies of 6 and 15 MV, and electron energies of 6, 9 and 14 MeV were studied. To determine the angular response of the diode, the gantry was rotated clockwise and counterclockwise at 15° intervals from -45° to +45°. A 0.6 cm³ Farmer-type ionization chamber was taped to the collimator head to monitor variations in machine output with gantry angle. All readings were normalized to those at zero degree angle.

2.2.4 Source-to-Surface Distance Dependence

To determine the diode sensitivity with SSD for 6 and 15 MV photon beams, both diode and ionization chamber readings were evaluated as a function of SSD for a constant field size of 10 cm X 10 cm. Measurements were taken at clinically relevant SSDs ranging from 65 to 130 cm. An ionization chamber was placed in a solid water phantom at the depth of maximum dose for each beam energy and measurements were taken at SSDs of 65, 80, 90, 100, 110, 120, and 130 cm by moving the treatment table. A diode detector was taped to the surface of the solid water phantom and measurements were taken at identical SSDs.
Fig. 6 Schematic representation of setup used in the measurements of angular dependence. Parallel vs perpendicular position of the detector is also shown in relation to the gantry.
The inverse square law was used to calculate the expected readings at specified SSDs for both diode and ionization chamber detectors. In both cases, measured to calculated ratios were evaluated and normalized to 100 cm SSD.

2.2.5 Effects of Beam Modifiers on Diode Response

The influence of beam modifiers on diode and ionization chamber responses was investigated as a function of SSD. Diode detectors were taped to the surface of a solid-water phantom while the ionization chamber was placed at $d_{\text{max}}$. Readings were taken with and without beam modifiers using 6 and 15 MV x-rays at SSDs of 65, 80, 90, 100, 110, 120, and 130 cm (10 cm X 10 cm field size). The beam modifiers investigated included an acrylic block tray and four commonly used wedges ($15^\circ$, $30^\circ$, $45^\circ$, and $60^\circ$). All readings were normalized to the 100 cm source-to-surface distance open field readings. In order to ensure that the diode and ionization chamber were under identical wedge thicknesses, measurements were also taken by rotating the solid water phantom to $180^\circ$. When the initial and final readings, rotating the solid water phantom to $180^\circ$ from original, agreed to within $\pm 1\%$, it was assumed that the diode and ionization chamber were adequately aligned.

2.2.6 Temperature Dependence

Initially, diode readings were taken at ambient water temperature (22°C). The water was then drained from the phantom and filled with warm water (45°C). At this point, measurements were taken at approximately 5°C intervals as the temperature of the water phantom gradually decreased.
The phantom temperature was monitored with a thermometer (± 0.5°C) inserted into the ionization chamber slot. As a given measurement temperature was reached, ten minutes elapsed before measurements were taken. This allowed the diode to reach thermal equilibrium with the phantom surface. Two sets of readings were taken on separate days. In each case, measurements were initiated at 6 MeV and continued in sequence to 8, 9, 10, 12, 14 MeV, 6 and 15 MV. The 6 MeV measurement was repeated at the end of each sequence to evaluate any deviation in water temperature during the measurement sequence. There was no change of temperature during each temperature setting for all electrons and photons. In all cases, readings were taken at 100 cm SSD using a 10 cm X 10 cm field size. All readings were normalized to those at ambient water temperature.

2.2.7 Dose Perturbations of Field Behind the Diode

Attenuation of the electron beam due to the presence of the diode was investigated for different electron energies using film dosimetry. The film was placed vertical to the beam in the middle of two solid water phantoms (30 cm X 30 cm X 15 cm) for each electron energy as shown in Figure 7. The top edge of the solid water was at 100 cm from the source. The diode was taped to the top of the solid water at the central axis of the electron beam using a 15 cm X 15 cm cone. Electron energies of 6, 8, 9, 10, 12, and 14 MeV were evaluated. An optical dosimeter was used to scan the developed films at a sampling interval of .5 cm across the field to obtain beam profiles and evaluate the shadowing effect of the diode at depths of \(d_{\text{max}}\), ±90% and ±80% isodose lines.
Fig. 7 Perturbation setup using unexposed film in the middle of the vertical solid water phantom.
2.2.8 Batch Comparison for Two 6 MV Diodes

A comparison was made between two identical 6 MV diodes from the same manufacturer, but of different batches. Diode responses were compared as functions of field size, SSD, angular dependence, and beam modifiers.

First, the diode's response relating to field size was measured by setting up the same parameters for both diodes using field sizes of 4, 5, 10, 15, 20, 25 cm² and a SSD of 100 cm. In each case, the diode was placed on the surface of a solid water phantom at the central axis. Comparisons of diode responses as functions of SSD and modifiers were investigated using a 10 cm X 10 cm field size. SSDs of 130, 100, and 65 cm, and wedge angles of 15°, 30°, 45°, and 60° were considered. The directional dependence was investigated using a 15 cm X 15 cm X 15 cm solid water phantom at 10 cm X 10 cm field size, 100 cm SSD. Measurements were taken at 15° increments from -45° to 45° and normalized to 0°.

2.2.9 Patient Dose Measurements

The purpose of the last part of the project was to investigate the usefulness of tissue heterogeneity corrections in lung and pelvis treatments where significant air and bone volumes are encountered. Lung density correction factors ranging from 0.25 to 0.33 gcm⁻³ are commonly used depending on the amount of air in the lung, while bone density correction factors typically range from 1.2 to 1.8 gcm⁻³ depending on bone thickness. In this project, heterogeneity correction factors of 0.30 and 1.3 gcm⁻³ were chosen for air and bone, respectively.

Entrance and exit doses (Figures 8 and 9) were calculated by a treatment planning computer for selected patients under treatment for lung or prostate
Fig. 8 Opposed lateral radiation beams traversing a patient in a low density lung. Dose is calculated at isocenter, (dmax) from anterior posterior field and exit from posterior anterior field.

Fig. 9 Isocentric four field technique using equal weighting in prostate patient. Dose is calculated at isocenter, entrance position (dmax) from AP and right lateral field, and at exit position from PA and left lateral.
carcinoma. In each case, plans were generated with and without heterogeneity correction factors. The effects of heterogeneity corrections were evaluated by comparing calculated entrance and exit doses to actual diode measurements during patient treatment. In total, 10 prostate and 8 lung patients were evaluated. Prostate patients were treated with 15 MV x-rays using a four-field technique consisting of anterior-posterior (AP), posterior-anterior (PA), right and left lateral fields. Lung patients were treated with 6 MV x-rays using a standard two-field (AP, PA) technique. In all measurements, the diode was placed as close to the central axis as possible and taped to the surface of the patient's skin using hypo-allergic tape.
CHAPTER III

RESULTS

3.1 Results

3.1.1 Energy Dependence

 Variation in diode response with electron energy is illustrated in Figure 10. All measurements were taken with the diode connected to the 6 MeV electrometer channel. The reading at each energy was corrected for the difference in the position of the effective point of measurement on the depth dose curve and then normalized to the reading at 10 MeV. With the exception of the 6 MeV data, diode response increases in an approximately linear fashion with increasing energy. The variation in diode response across the range of energies is approximately ±6%. By comparison, the variation in ionization chamber response across identical energies is approximately ±4% (Figure 11).

3.1.2 Field Size Dependence

 Diode and ionization chamber response as a function of field size presented in Table 1. Output factors were measured for 6 and 15 MV photons, and for all electron energies (6, 8, 9, 10, 12, and 14 MeV). The results show that there is excellent agreement between diode and ionization chamber response for fields larger than 10 cm X 10 cm; in all cases, discrepancies are within ±0.5%. At smaller field
Fig. 10. Variation in response of the QED silicon diode detector with electron energies.
Fig. 11. Variation in response of the ionization chamber with electron energies
sizes, however, discrepancies between diode and ionization chamber readings

Table 1. Output factor with QED p-type silicon diode detector and ionization chamber in solid water phantom

<table>
<thead>
<tr>
<th>Energy</th>
<th>Field Size (cm x cm)</th>
<th>Ion Chamber</th>
<th>Diode</th>
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<td>4 x 4   5 x 5  6 x 6  10 x 10  15 x 15  20 x 20  25 x 25</td>
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<td>6 MV</td>
<td>0.929   0.962  1.000  1.032  1.049  1.059</td>
<td>0.932  0.962  1.000  1.030  1.050  1.059</td>
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<tr>
<td>15 MV</td>
<td>0.911   0.958  1.000  1.032  1.048  1.059</td>
<td>0.933  0.962  1.000  1.032  1.049  1.058</td>
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</tr>
<tr>
<td>6 MeV</td>
<td>0.815   1.000  0.997  0.976  0.964</td>
<td>0.840  1.000  0.991  0.973  0.960</td>
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<td>8 MeV</td>
<td>0.886   1.000  0.989  0.972  0.960</td>
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<tr>
<td>9 MeV</td>
<td>0.905   1.000  0.989  0.972  0.960</td>
<td>0.934  1.000  0.991  0.973  0.960</td>
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</tr>
<tr>
<td>10 MeV</td>
<td>0.909   1.000  0.992  0.968  0.964</td>
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<tr>
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<td>14 MeV</td>
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<td>0.971  1.000  0.992  0.963  0.962</td>
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approach ±3% in some instances. In the case of photons, both types of detectors showed an increase in response with increasing field size for the two energies investigated.

The field size dependence is not as straightforward for electrons. In general, diode and ionization chamber response appears to peak at a field size of 10 cm X 10 cm and then gradually decrease with increasing field size.
3.1.3 Directional Dependence

The directional dependence of the diode detector is illustrated in Figures 12-16 for various representative energies. Variation in diode response across all energies is minimal for both photon energies investigated (±0.5 and ±2.0% for 6 and 15 MV, respectively). In the case of electrons, all diode response curves have a characteristic parabolic shape suggestive of an over-response at oblique angles. This over-response is particularly severe at 6 MeV where it approaches ±9% at 40° (Figure 14). The directional dependence of diode response diminishes with increasing electron energy. For example, at 14 MeV, variations in diode response are within ±3% across all angles investigated. The difference in diode response as a function of diode orientation (perpendicular vs. parallel) was within ±1%. At all energies considered, no significant differences were observed in diode response as a function of the direction of gantry rotation (clockwise vs. counter-clockwise).

3.1.4 Source-to-Surface Distance Dependence

The SSD response of a photon diode and an ionization chamber at 6 and 15 MV are shown in Table 2. Each entry in Table 2 is a ratio of the measured reading normalized to 100 cm SSD divided by the expected reading based on the inverse square law. The results in Table 2 (open field) show that there is good agreement between measured diode response and that expected from inverse square dependence. In most cases, agreement is within ±2% over the clinically relevant SSDs. A small over-response (±2-3%) is observed at small SSDs with the 15 MV diode detector. There is excellent agreement between the normalized diode and ionization chamber responses at 6 MV. At 15 MV, however, deviations of up to ±5% are observed at the extremes of clinically relevant SSDs.
Fig. 12. Directional dependence of the QED silicon diode detector as a function of 6MV photon energy.
Fig. 13. Directional dependence of the QED silicon diode detector as a function of 15MV photon energy.
Fig. 14. Directional dependence of the QED silicon diode detector as a function of 6MeV electron.
Fig. 15. Directional dependence of the QED silicon diode detector as a function of 9MeV electron.
Fig. 16. Directional dependence of the QED silicon diode detector as a function of 14MeV electron.
Table 2. Response of diode detectors and ionization chambers in photon fields with and without beam-modifying devices.

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<th>Energy</th>
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3.1.5 Effects of Beam Modifiers on Diode Response

In most cases, the response of the 6 MV diode was within ±2% of that expected from inverse square law. Over-responses of up to ±5% were observed for the two highest wedge angles (45° and 60°) at the smallest SSD (65 cm). In general, the dependence of diode response on wedge angle decreased with increasing SSD. The dependence of ionization chamber response on wedge angles was less severe than the diode response. In almost all cases, 6 MV ionization chamber response was within ±2% of expected values. Significant fluctuations in diode response were observed in the high energy fields (15 MV). Significant over- and under-responses were observed at small and large SSDs in the case of large wedge angles. Again, fluctuations in ionization chamber response was less severe than those observed for the 15 MV diode detector.

The effect of the acrylic tray was more significant in the high energy field (±4.5% over-response at 65 cm SSD) compared to the low energy field (±2.8% over-response at 65 cm SSD).

3.1.6 Temperature Dependence

Figures 17-24 show the sensitivity of the diode as a function of surface temperature of the water phantom. In all cases, diode sensitivity increased linearly with increasing temperature. For 6 and 15 MV photons, the observed response was approximately ±0.15 and ±0.13%°C⁻¹, respectively. The corresponding diode response in electron fields ranged from ±0.13 to ±0.21%°C⁻¹.
Fig. 17. Sensitivity variation with surface temperature for the QED silicon diode detector measured in 6MV photon beam.
Fig. 18. Sensitivity variation with surface temperature for the QED silicon diode detector measured in 15MV photon beam.
Fig. 19. Sensitivity variation with surface temperature for the QED silicon diode detector in 6MeV electron energy.
Fig. 20. Sensitivity variation with surface temperature for the QED silicon diode detector in 8MeV electron energy.
Fig. 21. Sensitivity variation with surface temperature for the QED silicon diode detector measured in 9MeV electron.
Fig. 22. Sensitivity variation with surface temperature for the QED silicon diode detector measured in 10 MeV electron.
Fig. 23. Sensitivity variation with surface temperature for the QED silicon diode detector measured in 12MeV electron energy.
Fig. 24. Sensitivity variation with surface temperature for the QED silicon diode detector measured in 14 MeV electron energy.
3.1.7 Dose Perturbations of Field Behind the Diode

Dose perturbation effects are shown in Figures 25-30 for all electron energies. The maximum beam reduction was approximately ±16% for 6 MeV. For electron energies between 8 and 10 MeV, the average dose reduction was found to be approximately ±12%. The smallest shadowing effect (±9.8% dose reduction) was observed for 14 MeV electrons (Figure 30). The results are summarized in Table 3. Figure 31 illustrates the shadowing effect at $d_{\text{max}}$, ±90% and ±80% clinical depths for all energies. The results indicate that as the depth increases, shadowing effect decreases.

<table>
<thead>
<tr>
<th>Energy</th>
<th>% Reduction</th>
</tr>
</thead>
<tbody>
<tr>
<td>6 MeV</td>
<td>15.6%</td>
</tr>
<tr>
<td>8 MeV</td>
<td>12.1%</td>
</tr>
<tr>
<td>9 MeV</td>
<td>11.2%</td>
</tr>
<tr>
<td>10 MeV</td>
<td>12.2%</td>
</tr>
<tr>
<td>12 MeV</td>
<td>10.7%</td>
</tr>
<tr>
<td>14 MeV</td>
<td>9.8%</td>
</tr>
</tbody>
</table>

3.1.8 Batch Comparison for Two 6 MV Diodes

The angular dependence of two QED diodes from different batches is illustrated in Figure 32. Each data point represents the average of readings acquired during clockwise and counterclockwise rotation. A maximum deviation of approximately ±0.7% was observed at 40°.

The data presented in Table 4 show no significant batch variation with field size. A maximum discrepancy of ±0.3% was observed between the two diodes.
Fig. 25. Beam profile underneath an QED silicon diode detector for 10cm x 10cm field using 6MeV electron beam.
Fig. 26. Beam profile underneath an QED silicon diode detector for a 10cm x 10cm field using 8MeV electron beam.
Fig. 27. Beam profile underneath an QED silicon diode detector for a 10cm x 10cm field using 9MeV electron energy.
Fig. 28. Beam profile underneath an OED silicon diode detector for a 10 cm x 10 cm using 10 MeV electron beam.
Fig. 29. Beam profile underneath an QED silicon diode detector for a 10cm x 10cm field using 12MeV electron beam.
Fig. 30. Beam profile underneath an QED silicon diode detector for a 10cm x 10cm field using 14MeV electron beam.
Fig. 31. Depth dose characteristics of the electron energies as a function of perturbation effect using QED silicon diode detector.
Fig. 32. Directional dependence of two QED silicon diode detector as a function of 6MV photon energy
The response of the two diodes was investigated as functions of SSD and beam-modifying devices. In all cases, diode response agreed to within ±0.5% as shown in Table 5.

Table 4. Batch comparison between two 6 MV semiconductor silicon diodes in different field sizes

<table>
<thead>
<tr>
<th>Field Size (cm)</th>
<th>5 x 5</th>
<th>10 x 10</th>
<th>15 x 15</th>
<th>20 x 20</th>
<th>25 x 25</th>
</tr>
</thead>
<tbody>
<tr>
<td>Primary Diode</td>
<td>0.948</td>
<td>1.000</td>
<td>1.030</td>
<td>1.047</td>
<td>1.057</td>
</tr>
<tr>
<td>Secondary Diode</td>
<td>0.949</td>
<td>1.000</td>
<td>1.027</td>
<td>1.047</td>
<td>1.057</td>
</tr>
</tbody>
</table>

Table 5. SSD and beam modifiers batch comparison between two 6 MV diodes

<table>
<thead>
<tr>
<th>SSD(cm)</th>
<th>Open Field</th>
<th>Acrylic Tray</th>
<th>Wedges</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>15</td>
<td>30 45 60</td>
<td></td>
</tr>
<tr>
<td>Primary</td>
<td>130</td>
<td>0.996 0.992 0.992 0.992 0.988 0.991</td>
<td></td>
</tr>
<tr>
<td></td>
<td>100</td>
<td>1.000 1.000 1.000 1.000 1.000 1.000</td>
<td></td>
</tr>
<tr>
<td></td>
<td>65</td>
<td>1.004 1.016 1.009 1.019 1.034 1.032</td>
<td></td>
</tr>
<tr>
<td>Secondary</td>
<td>130</td>
<td>0.996 0.994 0.993 0.992 0.989 0.989</td>
<td></td>
</tr>
<tr>
<td></td>
<td>100</td>
<td>1.000 1.000 1.000 1.000 1.000 1.000</td>
<td></td>
</tr>
<tr>
<td></td>
<td>65</td>
<td>1.000 1.012 1.007 1.015 1.033 1.027</td>
<td></td>
</tr>
</tbody>
</table>

3.1.9 Patient Dose Measurements

The ratio of measured to calculated entrance and exit doses for two opposing lung fields is illustrated in Figures 33 (AP) and 34 (PA). In all patients evaluated, the measured and calculated AP doses agreed to within ±4% (Figure 33). In the case of the PA field, however, significant discrepancies were observed in the absence of heterogeneity corrections (Figure 34). In one patient, measured and calculated...
Fig. 33. Relative entrance lung dose is shown using Effective Pathlength heterogeneity correction method.

<table>
<thead>
<tr>
<th>Patient</th>
<th>A</th>
<th>B</th>
<th>C</th>
<th>D</th>
<th>E</th>
<th>F</th>
<th>G</th>
</tr>
</thead>
<tbody>
<tr>
<td>Relative reading</td>
<td>0.98</td>
<td>0.99</td>
<td>1.01</td>
<td>1.02</td>
<td>1.03</td>
<td>1.04</td>
<td>1.05</td>
</tr>
</tbody>
</table>
Fig. 34. Relative exit lung dose as a function of heterogeneity correction.
doses differed by approximately ±70%. Significant improvements in these discrepancies were realized following heterogeneity corrections. In all cases, calculated and measured values agreed to within ±25%.

Ratios of measured to calculated entrance and exit doses for fields typically used in the treatment of prostate cancer are illustrated in Figures 35 and 36. Good agreement (±5-10%) was found between measured and calculated AP and right lateral entrance doses (Figure 35), however, discrepancies of up to ±75% were found in the case of the corresponding exit doses (Figure 36). Some improvement was observed following heterogeneity corrections.
Fig. 35. Relative entrance pelvis dose from anterior posterior and right lateral fields.
Fig. 36. Relative exit pelvis dose from posterior anterior (PA) and left lateral fields.
CHAPTER IV

DISCUSSION

For each radiotherapy field used to treat a patient, an in vivo measurement of delivered dose is recommended (ICRU, Report 29, 1978). Both entrance and exit doses are measured and used to deduce the dose delivered to some point within a patient. A comparison is made between the dose measured with a diode dosimeter and the dose calculated with the treatment planning computer. A difference in prescribed and measured doses larger than ±5% may indicate an incorrect beam or patient setup and should be investigated further (Alew, Alew, and Ochran 1998). Since there are numerous parameters that affect the sensitivity of diode detectors, accurate patient dosimetry requires an assessment of the variations in diode response as a function of these parameters prior to clinical use. Parameters such as temperature, beam orientation, diode shielding, energy, source-to-surface distance, field size, and presence of beam modifiers should be investigated since they affect the response of QED p-type silicon detector in high energy electron and photon beams. Knowledge of how these parameters affect diode response allows the application of appropriate correction factors for accurate in vivo patient dosimetry.

The origin of the ±6% variation in diode response as a function of energy dependence across electron energies (Figure 10) is not known. It may be due to an
inability to adequately account for the amount of buildup material surrounding the sensitive volume of the diode. The manufacturer does not specify the energy to which the diode buildup is optimized. The results in Figure 10 suggest that the buildup is likely optimized to energies around 10 MeV since the diode under-responds at lower energies and over-responds at higher energies. It is interesting to note that the ionization chamber also shows significant variations in response (±4%) across the electron energies examined. In this case, the variations may be due to difficulties in making measurements at the appropriate d_max in the solid water phantom and correcting it with the stopping power ratios between air and solid material. Resolution of this issue will require fine depth dose measurements in a water phantom.

The manufacturer does not specify expected diode sensitivity as a function of electron energy, however, it should be noted that Eveling et al. (1999) observed a ±0.8% variation across similar electron energies using an EDD-2 p-type diode detector system. The rather large variation in diode response observed in this study is not a major concern from a clinical perspective, as long as the appropriate correction factors are applied.

Variations in diode and ionization chamber response as a function of field size in photon fields can be explained by changes in scattered photon contributions. The dose at any point in the field is due to contributions from primary and scattered photons. As field size is increased, the dose increases due to an increased number of scattered photons from overlying and underlying material. As a result, diode response also increases. Conversely, diode response decreases with decreasing field size since the scatter contribution decreases.
Variations in diode and ionization chamber response with changing electron fields are not as straightforward as those observed in photon fields. This is due to the fact that electron beam parameters (including field size) are somewhat specific to each linear accelerator. As a result, these parameters must be investigated carefully prior to clinical use. For example, the effects of field size on output and position of $d_{\text{max}}$ is significant when the distance between the point of measurement and the edge of the field is shorter than the range of laterally scattered electrons. When the field size is reduced below that required for lateral scatter equilibrium, the output decreases rapidly and $d_{\text{max}}$ moves closer to the surface. This is especially the case for lower energies due to the greater lateral spread of electrons. Failure to account for this effect could result in an underestimate of detector response. This may explain the significant discrepancies observed in the small fields (4 cm X 4 cm and 5 cm X 5 cm), especially at lower energies. As field size increases, lateral equilibrium is established, and field size effects become less significant. Variations in diode response observed in this study were more severe than those obtained by others using similar detector systems. For example, Eveling et al. (1999) observed a $\pm 5.5\%$ under-response at 6 MeV (4 cm X 4 cm field size). By comparison, an underestimate of $\pm 16\%$ was observed under identical conditions (Table 1) in this study. The reasons for the discrepancies between the two studies are not known. However, it is likely due to differences in the electron collimation system. It is interesting to note that similar under-responses were observed for the ionization chamber (Table 1).
The response of the diode in electron fields exhibits a pronounced angular dependence, especially at low energies (Figure 14). This is due to the fact that, at oblique angles, the path length of the beam through the detector increases compared to that at 0° incidence. This effect will be more severe at lower electron energies due to the steeper dose gradient at the point of measurement of the diode. The angular dependence of the diode may cause problems in the clinic since application of appropriate correction factors may be difficult. This is due to the fact that the angle of incidence of the beam on the detector must be known to a fairly high degree of accuracy before a correction can be applied. With care in positioning the detector on the patient, the variation in angle between the incident beam and the detector axis can be minimized. Even so, the angular dependence of this detector is likely to result in enhanced readings for the majority of patients, especially at low electron energies.

The results obtained in this study are in good agreement with those obtained by others using similar diode detectors. For example, Eveling et al. (1999) observed a ±9.5% over-response at a 45° angle using 6 MeV electrons, and Lee et al. (1994) found angular dependence of less than ±2% in photon fields across the range of angles investigated in this study. This is in excellent agreement with the results obtained in this study for 6 and 15 MV photons (Figures 12 and 13). According to the manufacturer’s specifications, variations in diode response in 6 MeV electron fields should not exceed ±4% across the angles investigated in this study. In photon beams, the maximum deviation quoted by the manufacturer is ±2.5% (Sun Nuclear Corp. 1991).
The photon fluence and, hence, the exposure rate varies inversely as the square of the distance from the source. Strictly speaking, the inverse square law is only valid in primary photon fields. Intrinsic materials in the treatment head (e.g., collimators) and externally-introduced materials, such as block trays or wedges, cause photon scatter which may result in deviations from the inverse square law. Deviations may also result from the contribution of low energy electrons originating in the treatment head. Finally, it is important to note that the changes in SSD affects the instantaneous dose rate of the beam. This is significant because, according to the manufacturer, diode response depends on instantaneous dose rate; increasing the dose rate results in an increase in diode sensitivity (Sun Nuclear Corp. 1991). Since dose rates are higher for 15 MV photons, sensitivity variations of the corresponding diodes are expected to be higher than those of the 6 MV diodes.

The slight increase in diode response with decreasing SSD (Table 2) was attributed mostly to photon scatter and low energy electron contamination. As expected, this effect diminished with increasing SSD as fewer scattered photons and electrons reached the diode on the surface of the phantom.

In general, deviations from inverse square law observed in this study were insignificant; in almost all cases, measured diode readings were within ±2% of expected values. Normalized readings were in excellent agreement with those specified by the manufacturer over comparable SSDs. Sensitivity variations with SSD for this particular diode system were less severe than those observed for other diode models. For example, Meiler and Podgorsak (1997) observed deviations of ±5% and ±6% at SSDs of 70 and 130 cm, respectively, in a 6 MV photon beam, and ±5% and ±8% using 15 MV photons over the same SSD range.

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Absorbing blocks and/or special filters are commonly placed in the path of photon beams to modify isodose distributions. Heavy metallic-absorbing blocks are used in treatments requiring irregularly-shaped fields. These blocks are mounted on an acrylic tray close to the treatment head. Part of the photon beam will be absorbed by the blocks, while the remainder passes through the tray and is incident on the patient. A wedge filter is a wedge-shaped absorber which causes a progressive decrease in the intensity across the beam. It is made of a dense material, such as lead or steel, and mounted on a transparent plastic tray. The wedge angle is defined as the angle through which an isodose curve is tilted at the central ray of a beam at a particular depth.

There are basically three effects that cause variation in diode response with beam modifiers: photon scatter, electron contamination and change in beam quality. Electron contamination occurs when modifiers (e.g., trays, wedges) are introduced into the photon field. The degree of electron contamination and, hence, diode response depends on the type of accessory in the beam path, photon energy, and SSD. The shorter the SSD the higher the dose due to electron contamination and head-scattered photons. As the SSD increases, the angular spread of electrons and scattered photons increase and dose decreases. Increased diode response at higher energies may be due, in part, to increased forward scatter of higher energy electrons and photons. A wedge filter alters beam quality by preferentially attenuating lower energy photons (beam hardening) and, to a lesser extent, by Compton scattering which results in energy degradation (beam softening). Since these are competing effects, the net effect on diode response is minimal. The main effect is due to electron contamination and scattered photons.
The variations of diode response with SSD and wedge insertion are shown in Table 2. In the 6 MV photon field, the sensitivity of the diode increased at lower SSD and decreased at higher SSD across the field. The highest deviations between the measured and calculated readings were seen at the lower SSD due to the number of contaminating electrons and head-scattered low energy photons. At higher SSD, under-response was seen due to fewer scattered electrons and photons reaching the solid angle of the diode. For 15 MV photons, deviations were higher at both smaller and larger SSDs as compared to the 6 MV readings. Output values were consistently ±5-8% higher at small SSDs. In high energy beams, backscattering is negligible and almost all scattered photons originate from overlying layers. Due to the absence of overlying matter, the diode readings in high energy photon beams are independent of the phantom scatter. Lack of backscatter contribution at higher energies could explain the observed diode under-response in a 15 MV field at large SSD. At lower SSDs over-response is seen due to an increased forward component of electron scatter and greater contribution of head-scattered photons.

Sensitivity variations with temperature are important since diode temperature can increase significantly during patient measurement. Rikner and Grusell (1986) have shown that the temperature in a silicon diode detector will reach about 27°C after 1 minute on the patient's skin, and 31°C within 2-3 minutes. Based on the data obtained in this study, such a rise in temperature, compared to a calibration temperature of 22°C, suggests an increase in diode response of between ±1% and ±2% when irradiated on the skin surface. It is somewhat difficult to apply a correction factor since the temperature in the diode at the time of measurement is not
known, however, if a correction for a 7°C temperature rise (compared to calibration temperature) is applied to all measurements, the error will be less than ±1% for all diodes investigated in this study.

Although the manufacturer did not specify the temperature dependence for this particular type of diode, the results obtained here are in good agreement with those observed by others. For example, Eveling et al. (1999) observed an increase in diode response of approximately 0.26% °C\textsuperscript{-1} for 10 MeV photons using an EDD 2, \textit{p}-type diode detector.

The shadowing effect of the diode is significant at all energies investigated in this study. The approximate ±12% reduction in dose behind the diode (at \(d_{\text{max}}\)) is due to absorption of electrons by the acrylic buildup material which surrounds the sensitive volume of the detector. The perturbative effect of the diode was investigated at various depths of clinical significance (\(d_{\text{max}}\), 80% and 90% depth dose). The shielding effect decreased in an approximately linear fashion with increasing electron energies (Figure 31). The results suggest that the perturbative effects of the diode decrease with increasing tissue depth.

The reduction in dose due to the presence of the diode could have significant implications on short fractionation boost treatments, particularly those involving the breast. In such treatments, it has been recommended that dose monitoring should only be performed during one of the five fractions (Eveling et al. 1999).

The degree of shadowing is, to some extent, dependent on diode design, specifically on the type of buildup material used. The results obtained with the QED diode are in good agreement with those obtained for a similar diode by Eveling et al.
(1999), and are much better than the results of Sen et al. (1996) who observed dose perturbations of up to ±25% with an ISORAD™ diode employing an aluminum buildup cap.

Batch comparisons were made under a number of conditions. In all cases, diode responses were in agreement to within ±1%. Admittedly, this was a somewhat limited survey involving diodes from only two different batches. It is strongly recommended that the response of new diodes be evaluated prior to clinical use.

In most clinical situations, dose distributions are calculated assuming homogeneous unit density medium. However, beams commonly traverse tissues of varying densities (e.g., fat, bone, muscle, lung, and air). These inhomogeneities produce changes in the dose distribution depending on the quality of radiation and the amount and type of tissue present. The presence of an inhomogeneity may cause significant changes in: (1) the absorption of the primary beam and the associated pattern of scattered photons, and (2) the secondary electron fluence. The relative importance of each depends on the distance of the point of interest from the inhomogeneity. For tissues that lie well beyond the inhomogeneity, the predominant effect is the attenuation of the primary beam. Changes in the photon scatter distribution alters the dose distribution more strongly near the inhomogeneity than further beyond it. The changes in secondary electron fluence affects the tissues within the inhomogeneity and at the boundaries.

For x-ray beams in the megavoltage range where Compton interactions dominate, the attenuation of the beam in any medium is determined by electron
density. When megavoltage x-ray beams traverse low density regions, such as lung tissue, doses to tissue beyond the inhomogeneity will be comparatively higher due to decreased attenuation of the beam in the low density region. As illustrated in Table 6 (Khan 1994), the degree of attenuation is strongly dependent on energy and the amount of lung in the field. For example, the exit dose for 6 MV photons passing through 7 cm of lung tissue is approximately ±21% higher compared to that expected based on unit density tissue. In other words, neglecting lung correction factors in the treatment planning calculations will result in significant discrepancies between observed diode readings and those expected based on the treatment plan.

The main effect of bone is to decrease the dose to points beyond the inhomogeneity. As illustrated in Table 7 (Khan 1994), the shielding effect of bone diminishes rapidly with increasing x-ray energy. The shielding effect of bone for x-rays between 500 KV and 4 MV is due to its greater electron density as all attenuation is due to the Compton process. However, as energies increase beyond 10 MV, the shielding effect begins to increase as pair production becomes significant.

In general, the corrections for bone attenuation in most clinical situations are small and are usually neglected.

Table 6. Increase in Dose to Tissues Beyond Healthy Lung (Khan 1996)

<table>
<thead>
<tr>
<th>Beam Quality</th>
<th>Correction Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Othovoltage</td>
<td>+10%/cm of lung</td>
</tr>
<tr>
<td>$^{60}$Co $\gamma$ rays</td>
<td>+4%/cm of lung</td>
</tr>
<tr>
<td>4-MV x-rays</td>
<td>+3%/cm of lung</td>
</tr>
<tr>
<td>10-MV x-rays</td>
<td>+2%/cm of lung</td>
</tr>
<tr>
<td>20-MV x-rays</td>
<td>+2%/cm of lung</td>
</tr>
</tbody>
</table>
At Nevada Radiation Oncology Centers in Las Vegas, Nevada, the treatment planning systems use two different types of inhomogeneity correction algorithms: Effective Pathlength and Batho Power Law. The Effective Pathlength method is mostly used in 2D treatment planning, whereas the Batho Power Law method is utilized in 3D planning.

Table 7. Reduction in Dose Beyond 1 cm of Hard Bone (Khan 1996)

<table>
<thead>
<tr>
<th>Beam Quality</th>
<th>Correction Factor (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 mm Cu HVL</td>
<td>-15</td>
</tr>
<tr>
<td>3 mm Cu HVL</td>
<td>-7</td>
</tr>
<tr>
<td>$^{60}$Co</td>
<td>-3.5</td>
</tr>
<tr>
<td>4 MV</td>
<td>-3</td>
</tr>
<tr>
<td>10 MV</td>
<td>-2</td>
</tr>
</tbody>
</table>

The Effective Pathlength correction treats all dose as primary dose and only takes into account the variation in medium density along a path from the radiation source to the point of calculation. This method provides an accurate and efficient way of predicting dose distributions in situations where the source-to-surface distance is not that used for data collection and when the patient surface is curved. However, this method does not take into account the effects of heterogeneities on scattered photons or secondary electrons.

A more complex algorithm is the Batho Power Law method that considers the position of inhomogeneities along a path as well as their thickness and density. This method takes some account of the perturbation in scattered dose. The main difference between this and the Effective Pathlength method is that the position of the
inhomogeneity above the dose calculation point is considered, however, the
inhomogeneity is assumed to be a slab extending well beyond the beam boundaries.
Furthermore, perturbations in backscatter dose due to underlying inhomogeneities are
not considered. The accuracies of the Batho Power Law and Effective Pathlength
algorithms are typically greater than \( \pm 5\% \) for most irradiation conditions
encountered in radiotherapy. In general, Nevada Radiation Oncology Centers do not
account for tissue inhomogeneities when planning 2D or 3D photon treatments.

In this study, the Effective Pathlength algorithm was used to correct for lung
and bone inhomogeneities. As expected, the ratios of measured to calculated lung
entrance doses were in good agreement (Figure 33). In almost all cases, the
application of appropriate correction factors resulted in significant improvements in
the ratio of measured to calculated lung exit doses (Figure 34). The \( \pm 25\% \)
discrepancy noted for patient 'D' may have been due to inaccurate positioning of the
diode. Comparisons between measurements and calculations assume diode placement
on the central axis. Errors in positioning could result in significant discrepancies
between measured and calculated exit doses if the thickness of lung beneath the
mispositioned diode is significantly different than that at the central axis.
Discrepancies between measured and calculated doses due to diode misalignment may
be especially severe in treatments involving modifiers such as wedges.

In most cases, measured and calculated pelvis entrance doses agreed to within
\( \pm 10\% \) (Figure 35). The greater discrepancies observed for the right lateral fields
are probably due to difficulties associated with accurate positioning of the diode on
the right hip of the patient. In this case, the use of correction factors did not result
in significant improvement in discrepancies between measured and calculated exit
doses (Figure 36). This is not unexpected since shielding by bone is not significant at the high energies used for prostate treatments. As expected, correction factors have more effect when applied to lateral fields due to the greater thickness of bone in the field. The origin of the large discrepancy observed for patient 'E' is unknown. It is most likely due to error in diode positioning.

In most clinical situations, inhomogeneity correction algorithms are not used in the treatment planning process. This is due to the fact that, until recently, correction algorithms were too simplistic to yield accurate results for the majority of complex treatment geometries commonly encountered in the clinic. This will likely change in the near future as increased computational speed has provided the impetus for the development of more sophisticated correction algorithms.
CHAPTER V

CONCLUSIONS

Factors affecting the response of a new type of silicon semiconductor diode detector have been investigated. In the majority of cases, the detector's performance was in good agreement with manufacturer's specifications and/or the results obtained by others using similar p-type diode detectors. Significant sensitivity variations in diode response were observed in photon fields with beam modifiers, and in low energy electron fields at large angles. These studies emphasize the importance of evaluating diode response in commonly encountered clinical situations so that appropriate correction factors can be established prior to clinical use. Knowledge of these correction factors is important since it will make dose misadministrations more apparent.

The shadowing effect of this particular type of diode in electron fields was found to be significant. In order to avoid underdosage to underlying tissues, it is recommended that the diode be used in only a small fraction of electron boost treatments for each patient.

Significant discrepancies between measured and calculated exit doses were observed in patients treated for lung and prostate carcinoma. In some cases, significant improvements were realized following the application of correction factors.
to account for bone or lung inhomogeneities. Unfortunately, diode response is very dependent on its location in the treatment field. This is especially true if beam modifiers, such as wedges, are used in the treatment. It is important to realize that large discrepancies between measured and calculated exit doses may not be due to intrinsic variations in diode response, or to a failure to adequately account for the presence of tissue inhomogeneities in the treatment field. The patient measurements performed in this study suggest that inaccurate diode placement is an important contributor to discrepancies between measured and calculated exit doses. This could be due to difficulty placing the diode on the irregular lateral shape of the pelvis. Misplacement of the diode with respect to lung could also give higher readings. Resolution of these discrepancies can only be effected through a more systematic approach of diode placement on the patient.
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VITA

Graduate College
University of Nevada, Las Vegas

Kamran Ul Haq

Local Address:
2403 Brockton Way
Henderson, NV 89014

Degrees:
Bachelor of Science, Radiological Science, 1990
University of Nevada, Las Vegas

Thesis Title: Characterization of $p$-type Silicon Semiconductor Detectors for Use in
In Vivo Dosimetry

Thesis Examination Committee:
Chairperson, Dr. Steen Madsen, Ph.D.
Committee Member, Dr. Mark Rudin, Ph.D.
Committee Member, Dr. William Johnson, Ph.D.
Graduate Faculty Representative, Dr. Cheryl Bowles, Ph.D.
Committee Member, Charles Spencer, M.S.