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Documentation of radiation therapy doese using a silicon solid state detector system

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DOCUMENTATION OF RADIATION THERAPY DOSES USING A SILICON
SOLID STATE DETECTOR SYSTEM

A Thesis
Presented to
The Faculty of the Nuclear Science Program
San Jose State University

In Partial Fulfillment
of the Requirements for the Degree
Master of Science

By
Trisha Edgerton
December 1995

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ABSTRACT

DOCUMENTATION OF RADIATION THERAPY DOSES USING A SILICON SOLID STATE DETECTOR SYSTEM

by Trisha Edgerton

This thesis investigates the use of solid state silicon diodes to document patient doses in a radiation therapy department. Methods were developed to test linearity; dose rate, angular and field size dependences; wedge effects; optimum calibration configuration; and response to high energy electrons. Finally, patient studies were conducted to determine if treatment setup errors could be identified during treatment administration.

Findings indicate that the diode responded well to all acceptance tests. Data collected on 32 patients, 114 measurements, confirmed that 93.0% of the treatments were within the acceptable radiation therapy accuracy limit of $\pm 5\%$. All measurements (7%) which exceeded the limit were performed in wedged fields. It is concluded that wedged fields present unique problems when documenting patient doses. In addition, all patients would benefit from daily treatment dose verifications. Any under- or over-treatments identified could be corrected before the end of the treatment regiment and improve patient outcome.

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INTRODUCTION

Accuracy in radiation therapy is of great importance. Proper planning and administration of therapy doses is imperative to ensure a consistent and safe fulfillment of the dose prescription to the target volume, with minimal dose to normal tissues and minimal exposure to personnel (American Association of Physicists in Medicine 1984). Success or failure of radiation therapy depends on the actual dose delivered. It has been shown that because many tumors exhibit a steep dose-response curve, dose errors of only 7-10% to the target volume may result in a significant change in the tumor control probability (American Association of Physicists in Medicine 1984; International Commission on Radiation Units and Measurements 1976; Lee, Sawicka, and Glasgow 1994; Mijnheer, Battermann and Wambersie 1987). Therefore, many institutions perform *in vivo* dose monitoring during each patient treatment in order to document and verify that the dose delivered matches the original prescription.

Besides appropriate tumor control, there are other reasons to monitor patient doses. One is to satisfy reporting requirements from federal and state regulating agencies regarding reportable events, and a second is to have a recorded confirmation measurement of the dose

delivered to a patient in the event of litigation (Lee, Sawicka, and Glasgow 1994).

The history of instrumentation used to verify patient doses in radiotherapy is described by Nilsson, Ruden and Sorcini (1988). The instrument preferred for linear accelerator quality control, the ion chamber, has significant problems when used on a patient. Specifically, ion chambers that are connected to a power source pose a risk of shock to the patient and may not be used. In addition, they are not sufficiently small to be used with small beam ports. Disconnected units of the capacitance discharge type are small enough, but are inaccurate due to loss of charge through leakage.

In 1932, Rolf Sievert used the small, self-contained discharge capacitance ionization chambers to measure patient doses. Other scientists have used them for intracavitary readings. However, the need for a smaller detector resulted in the introduction, around 1968, of thermoluminescence dosimeters (TLDs) for routine use. Even though thermoluminescent dosimeters improved the sensitive volume problem seen with larger ion chamber type dosimeters, they have their own significant disadvantage due to the unavoidable time delay between irradiation and obtaining final results. In addition, to obtain accurate results,

they require tedious attention to detail during processing and calibrating.

Finally, in the mid 1980's, silicon detectors were introduced to correct for the deficiencies of the other two methods. They offer the advantages of high sensitivity relative to an ion chamber, on-line readout, simple electrical measuring equipment, air pressure independence, mechanical strength, small size, good spatial resolution, the absence of bias voltage and, represent a lower cost per result from TLD monitoring (Rikner and Grusell 1987; Nilsson, Ruden and Sorcini 1988; Heukelom, Lanson and Mijneer 1991).

The basic theory of silicon detectors is described well by Rikner and Grusell (1987). Silicon detectors can be made of n-type or p-type silicon. n-type silicon is obtained when silicon is doped with a material from group V in the periodic table, usually phosphorus, and p-type when the doping material is a group III material, usually boron. All literature reviewed for this thesis stated that p-type silicon is most suitable for patient dosimetry in pulsed beams. The technology and research at the time demonstrated that p-type diodes had a more linear dose-rate dependence and a smaller sensitivity drop with increased absorbed dose (radiation damage) when compared to the n-type (Nilsson, Ruden and Sorcini 1988). However, in a personal interview

with Steve Szeglin, Director of Research and Development, Technical Services, Nuclear Associates (February 8, 1995), he stated that the Co-60 - 4 MV photon diode used in this thesis experimentation is manufactured using undoped special n-type silicon. He explained that by using this material they have reduced the temperature coefficient to 0.1%/°C from those values (1.5% - 2.5%/°C) published for p-type diodes. Temperature coefficients describe the change in the diode reading due to the differential temperature gradient when a room temperature diode (approximately 22°C) is placed on a patient with a body temperature of 36°C. Mr. Szeglin also stated that special processes in the manufacture of their n-type diodes have reduced their propensity to radiation damage over time. He quoted proprietary literature as saying that after 1500 treatments of 200 rads/treatment (total 300,000 cGy), the Co-60 - 4 MV diode lost less than 6% of its response. In fact, he now states that the n-type diode is more resistant to radiation damage than a p-type.

An integral part of the diode construction is the build-up cap which encases the silicon. This cap serves four purposes (Gray, Nuclear Associates Monograph, undated): 1) it provides mechanical support and protects the detector and lead wires; 2) it establishes electronic equilibrium for high-energy x-rays and filters out low-energy radiation; 3)

it provides a light-tight enclosure (diodes are very light sensitive); and 4) it facilitates a symmetrical electrical intermediary between the outer diode lead and the cable sheath.

Solid state silicon detectors operate in the following manner. During irradiation, electron-hole pairs are created in the diode. A current signal results when charge carriers are created in the depletion region and minority carriers (electrons in p-type material, holes in n-type material) diffuse to this area. Because of the electric field across the depletion zone, these migrating charges are collected. The sensitivity of this method can be increased by using a material with a long lifetime for minority carriers. In commercially available systems, the sensitivity is reported to be 18,000 times that of an air-filled ion chamber, due to the low energy needed to create an electron-hole pair (3.5 eV for a silicon diode versus 30 - 40 eV to produce an ion pair in most gasses) and the density of silicon compared to that of air at room temperature (International Commission on Radiation Units and Measurements 1982). The sensitive volume can therefore be small and well defined, thus increasing the spatial resolution (Grusell and Rikner 1986). The average sensitivity of a p-n junction dosimeter is 3 nA/rad/min, resulting in a dynamic range covering dose rates from 10^{-3} rad/min to 3×10^3 rad/min (Klevenhagen 1977). In

addition, for electron beam dosimetry, a diode depth ionization curve is equivalent to a depth dose curve because the mass collision stopping power ratio between water and silicon is fairly independent of electron energy (Rikner and Grusell 1987). Mr. Szeglin, of Nuclear Associates, stated that the electron diodes manufactured by his company, and used for the purposes of this thesis, are n-type doped with platinum, and have been shown to be equivalent in performance to an electron p-type diode.

All of the above basic characteristics make silicon diode systems ideal to document the accuracy of delivered radiotherapy doses. The accuracy that can be achieved in delivering a radiotherapy dose depends on two types of errors, systematic and random (Lee, Sawicka and Glasgow 1994; Mijnheer, Battermann and Wambersie 1987). Systematic errors are instrument-related and are transmitted to every patient. These errors can be reduced by careful calibration of associated ion chambers and beam output. In addition, by participating in various mailed TLD programs, an institution can compare in-house results with those from other facilities and against an absolute standard (Lee, Sawicka and Glasgow 1994).

Random errors are those errors that are associated with patient dose delivery, including erroneous setups and misapplication of beam-shaping devices. These errors are

operator dependent and while they are normally thought to result in only small errors in dose delivery, they occasionally can result in a gross under- or over-treatment of the selected target volume, greatly varying the patient's clinical outcome.

These random errors associated with setting up an individual machine and patient appear to occur at a rate of 2 - 3 per 100 operations. This agrees with known human norms for repetitive mechanical tasks (ICRU 1976; Hendrickson 1978). The ICRU (1976) states that each patient setup, because it involves a great number of operations, may exhibit an error rate of up to 14% per setup. However, it acknowledges that these errors may or may not create a clinically significant change in the dose delivered to the critical tumor volume. Hendrickson (1978) confirms that a treatment setup error frequency of 2 - 3% would rarely lead to a delivered dose variance from the prescribed dose of greater than $\pm 5\%$. For example, the ICRU (1976), states that maximum error in the total tumor dose from missetting the field size is 25 rads (i.e., less than 1% in 4000 rads). A more detailed table of technologist errors and associated differences in total tumor dose are presented in Lee, Sawicka and Glasgow (1994). They state that among the errors likely to be detected clinically, are: 1) setting the incorrect daily dose (accumulation of 2 Gy instead of

1.8 Gy over a series of treatments results in an error of 11%); 2) wedge errors (wedge factors differ by more than 7% for all wedged and open fields for both 6 and 18 MV beams); 3) incorrect energy (sensitivity of a diode differs by 15% between 6 and 18 MV beams); and, 4) SSD (source to skin distance) versus SAD (source to axial, tumor center, distance) technique confusion (100 SSD versus 100 SAD for spine treatment will result in a dose difference of 11%). They also list treatment errors resulting from improper patient setup that are not likely to be detected clinically. These include incorrect field size, misapplication of a block tray and misapplication of bolus material. These errors result in dose errors of approximately 2 - 3%.

Hendrickson (1978) looked for a pattern in patient setup errors in an attempt to correct them. He states that these errors are directly related to work pressure, and do not vary significantly with the day of the week. His data shows individual day error rates ranging from 2.8 to 3.9%. He provides a solution by clearly demonstrating that error rates when one technologist works alone (3.5% in 17,174 setups) can be dramatically reduced when two technologists are assigned to the same room (0.82% in 7065 setups).

This thesis investigates the ability of an *in vivo* diode monitoring system to identify random errors in the setup and administration of patient radiotherapy, thus allowing

identification of these errors, ultimately increasing the accuracy of each delivered dose.

CHAPTER 1

ACCURACY THAT CAN BE OBTAINED IN RADIATION THERAPY: A REVIEW OF THE LITERATURE

Many authors have attempted to define the accuracy that may be achieved in radiation therapy. Concern for accuracy of absorbed dose within the target volume is based on two main assumptions: 1) that normal tissue should be assumed to be excluded and thereby protected from radiation injury as patients may develop unexpected complications if the treatment plan approaches normal tissue tolerances, and, 2) that a geometric tumor miss may lead to the tumor receiving less dose than planned, producing an unexplained "resistant" tumor (Hendrickson 1978).

In attempting to create the perfect treatment field for any given tumor and maximize normal tissue sparing, Hendrickson (1978) warns that possible gains in excessively complex irregular shielding blocks may be lost in the precision with which they can be produced. In addition, treatment plans that prescribe excessively tight margins around grossly defined disease may improve normal tissue sparing, but run the risk of occasionally shielding the tumor. Thus, the correct size and shape of blocks is very important; however, there are several other factors that influence dose to the intended target and sparing of normal tissue.

A therapy setup can include beam conditioning devices such as wedges, boluses or compensators. If any of these are omitted or incorrectly used, the dose will differ from that prescribed. In addition, total dose is also directly dependent on collimated field size, gantry and collimator angles, timer settings, energy selection and proper interpretation of the treatment distance (SSD versus SAD). Therefore, medical physicists and technologists must address the problem of quality control in daily treatments with the same diligence as that applied to machine calibration and treatment planning. This paper attempts to determine if, by using solid-state detectors during patient radiation therapy, these common setup errors can be detected.

In order for these to be detected, the delivered diode-measured dose must differ from the prescribed dose by a margin greater than accepted limits for accuracy in radiation therapy. The following is a review of the literature to identify currently accepted levels of accuracy.

In a 1984 report, the American Association of Physicists in Medicine (AAPM) indicate a target accuracy value of $\pm 5\%$. They define the uncertainty in a patient dose as a combination of the cumulative uncertainty of a dose delivered to a tissue-phantom (acceptable level of uncertainty 2.5%) and the uncertainty in the treatment

planning computation of a patient dose (acceptable level 4.2%). They further state that the later consists of three components: 1) central axis factors for blocks, wedges and compensators, 2) computation of relative dose distribution (heterogeneity of the tumor), and 3) dose computation effects of tumor localization and contour delineation. The figure of 5% represents two standard deviations and thus indicates a confidence interval of 95%. The AAPM states that this level is both clinically acceptable and technically achievable.

Mijnheer, Battermann and Wambersie (1987) investigated the degree of accuracy that could be achieved in photon therapy. They analyzed dose-response curves and detrimental effects to normal tissues to determine an acceptable level of accuracy necessary for adequate tumor control and normal tissue sparing. The figure they propose is $\pm 3.5\%$. They admit that this value may be hard to achieve and in many cases it may be unnecessarily low; however, they state that for some tumors an even smaller value should be used.

They present relative gradients of dose-effect curves for local tumor control after photon irradiation, expressed as $\Delta_{75/50}$. These values represent the relative increase in absorbed dose (in %) producing a change in tumor control probability from 50 to 75%. Therefore, small values will denote tumors that have a critically narrow margin for

changes in total dose in order to be eradicated. The larger values represent tumors that are more radioresistant and demand a greater increase in dose to result in an increase in tumor control probability. They report the results from many clinicians, so some tumor types are listed more than once, with different $\Delta_{75/50}$ values. Specifically, the tumors they show to have the most narrow tolerance for dose accuracy are supraglottic and laryngeal tumors, having relative $\Delta_{75/50}$ values of 5 and 6 percent, respectively. Other clinicians quoted in Mijnheer, Battermann and Wambersie's paper show values for supraglottic, larynx, bladder, skin, lip and nasopharynx tumors having $\Delta_{75/50}$ values ranging from 11 to 19 percent. More radioresistant tumors include lymphoma, and those at the base of the tongue, which have values of $\Delta_{75/50}$ of 21 to 50%.

All of these values have been clinically determined; however, they are fundamentally based on homogeneous cell population survival curves. The curve shape provides an upper limit to the absolute steepness of the dose-effect tumor control probability. When considering tumors in patients, the steepness is reduced due to the heterogeneity of the tumors with respect to their size (cell number), cell radiosensitivity, distribution over the cell cycle, percentage of hypoxic cells, growth rates, etc. (Mijnheer, Battermann and Wambersie 1987). From the data presented, it

can be seen that the majority of tumors demand a narrow tolerance of dose delivery, approximately 10 - 20%, for adequate tumor control. Depending on tumor type, the accuracy of the delivered dose may have to be within 5% in order to produce the desired clinical effect.

To investigate the dose tolerances needed to spare normal tissue, the same authors also presented data showing the relative gradient expressed as $\Delta_{50/25}$, which represents the relative increase in absorbed dose (in %) producing a change in the probability for normal tissue reactions from 25 to 50%. In this case, effects with associated small $\Delta_{50/25}$ values will be manifested more easily than those effects with high values. They show that major complications of the larynx, peripheral neuropathy, late skin damage, late intestinal damage, and brachial plexus changes all occur at a $\Delta_{50/25}$ value of 5% or less. Radiation pneumonitis, skin and lip reactions, and major complications of the intestine and bladder occur at $\Delta_{50/25}$ values of between 6 and 10 percent. Myelitis and acute complications of the larynx can occur at relative gradients of 15 and 17 percent respectively. It can therefore be seen that non-acute late effects occur at lower dose variations than acute effects; however, all effects occur at relatively small dose changes. Again, the authors present data from multiple sources.

Mijnheer, Battermann and Wambersie (1987) state that on adjacent fields in the same patient, a difference in dose of 10% could be detected in 80% of the cases and a difference of 20% could be detected in 90% of the cases. In addition, in symmetrically irradiated supraclavicular fields, differences of only 5% could be detected clinically. They state that from clinical observations of normal tissue reactions, a difference in absorbed dose of about 7% could be detected in most situations. Therefore, they conclude that any accuracy value of greater than 7% represents an unacceptable risk of treatment complications. They then assume that this value represents at least twice the combined uncertainty, given as one standard deviation, resulting in a standard requirement for the combined uncertainty in absorbed dose delivery to be 3.5%..

Another researcher, Rassow (1988), approached the problem of achievable radiotherapy accuracy by analyzing geometrical and dosimetric tolerances presented in the International Electrotechnical Commission's standard for quality control of radiation therapy equipment. He concludes that equipment which complies with the standard is capable of achieving an accuracy of dose delivery of approximately 4%. His analysis is based on physical measurements of the following tolerance allowances: light field indication of the center and the edges of the

radiation field, indication of the radiation beam axis, radiation beam axis relative to the isocenter, indication of the isocenter position, and indication of the distance from the isocenter along the radiation beam axis. He also takes into account parameters of the dose monitoring system including reproducibility, proportionality, angular dependence of the beam, stability of the machine throughout the day and the stability of the machine from day to day. Therefore his figure of 4% represents what he states is the most technically achievable accuracy limit.

Finally, the International Commission on Radiation Units and Measurements (ICRU) in a 1976 report reviewed the literature at that time. It is interesting to note that their recommendations in 1976 do not differ from accepted values today. The ICRU states that for some tumors, the probability of tumor control is a very steep function of dose. For example, a dose decrease of 10% reduces the probability of control of supraglottic squamous cell carcinoma from 70 to 10%. They go on to say that a dose deviation of 7 - 10% can be detected clinically through the outcome of the patient. They conclude that the evidence available at the time suggested that accuracies of $\pm 5\%$ were needed in the delivery of an absorbed dose to a target volume if the primary tumor was to be eradicated. They quote other clinicians as calling for limits of only $\pm 2\%$,

but state that, at that time, it was impossible to achieve such a standard.

After analysis of the literature, a value of 5% was accepted as the outside limit of unavoidable error in delivered patient treatment doses. This was taken to mean that measured doses within the work of this thesis that differed from prescribed doses by 5% or less would not be interpreted as being a result of errant patient setup. In this case, the patient setup error could not be separated from accepted levels of errors seen in today's clinical practices. However, conversely it means that measured doses differing from the prescribed dose in excess of 5% would most likely be the result of a patient setup error.

CHAPTER 2

MATERIALS AND METHODS

All photon measurements were made on a Varian Clinac 4 (4 MV) linear accelerator. All electron measurements were made on a Varian Clinac 18, possessing the ability to produce electrons of 6, 9, 12, 15, and 18 MeV, and 10 MV photons. The diode system used was a Nuclear Associates therapy silicon detector system with a digital model 37-720 electrometer, using an electron detector diode (model 30-496) and a Co-60 to 4 MV photon detector diode (model 30-490). For comparison measurements, a Keithley model 616 digital electrometer with a model 6169 ion chamber interface was used with both parallel plate and thimble chamber ion chamber probes.

The majority of measurements (reproducibility, electrometer/diode leakage, dose rate dependence, angular dependence, beam uniformity in x and y directions, and wedged readings) were performed using full backscatter material consisting of polystyrene sheets. All other parameters are explained in detail below.

Value leakage measurements were performed by accumulating 200 cGy and observing the value changing with time. Also, changes in a zero reading were observed in a similar fashion.

Dose rate dependence was measured by setting the Clinac 4 to accumulate 200 cGy, and then acquiring this amount at dose rates of 100, 200 and 300 cGy/minute.

Field size dependence was calculated from results obtained after calibrating the diode to a 10 cm x 10 cm field, and recording readings obtained from field sizes ranging from 4 cm x 4 cm to 32 cm x 32 cm. A 7-7/8" Lucite block phantom was used, placing the diode on top of the block. Concurrent ion chamber readings were taken at a depth of 1.2 cm overlying phantom material. This depth is where, for 4 MV photons, the maximum dose is achieved due to electronic equilibrium.

Linearity was evaluated by using a 10 cm x 10 cm field and a dose rate of 300 cGy/minute. The Clinac 4 was then set to accumulate totals of 100, 200, 300, 400, 500, 600 and 700 cGy.

Angular dependence was determined separately in two orientations - with the diode placed in a perpendicular (axial) orientation, then a parallel (transverse) orientation to the gantry. Considering 180° the neutral position (with the x-ray head pointing at the floor), readings were obtained from 120° to 240°, representing a variation of 60° on either side of the neutral position.

Evaluating the effects of a worst-case treatment plan on diode placement utilized a 15 cm x 15 cm breast beam

splitter and both 15° and 30° wedges. The diode was placed 1 cm from the penumbra to determine variations in output detection.

Determining the optimum calibration factor for everyday use was accomplished using a typical patient dose rate of 300 cGy/min and calibrating for a total absorbed dose of 200 cGy, then accumulating readings of 100, 200, 300 and 400 cGy. The measurements were repeated, calibrating the diode for a total absorbed dose of 400 cGy and accumulating the same readings of 100, 200, 300, and 400 cGy.

Checks to see if a calibration factor would have to be applied with varying source to skin distances was done using the 7-7/8" Lucite phantom. The diode was placed on top of the block and comparison Farmer ion chamber readings were obtained at a depth of dose maximum, 1.2 cm. A 10 cm by 10 cm field size was used and the diode was calibrated for a total of 200 cGy at 80 cm SSD. SSDs varied from 60 cm to 100 cm. These measurements were then repeated acquiring 600 cGy to determine if the statistical variation could be reduced.

Comparison of the tray-effect measured with an ion chamber versus the effect measured with the diode was performed by comparing ion chamber readings with and without a plastic tray in the beam with similar readings from the diode, at an SSD of 80 cm. Again, the 7-7/8" Lucite block

was used with the diode on top, and the ion chamber at a depth of 1.2 cm of overlying phantom material. This experiment was repeated at an SSD of 65 cm to determine if there was electron contamination within the diode at closer treatment distances.

The effect of adding backscatter material when performing diode measurements was then investigated, taking readings with full backscatter and comparing them to a diode reading at the same SSD, but in air. This was performed with a 10 cm x 10 cm field size at an SSD of 80 cm.

Beam uniformity in both the x and y directions was done with a 15 cm x 15 cm field size, at 80 cm SSD. The diode was calibrated for 200 cGy and readings were taken in both the radial and transverse directions every centimeter. These readings were then repeated in the radial direction with 15°, 30°, 45° and 60° wedges to determine the margin of error that could be seen if the diode was not placed in the center of the wedged field during the verification of a patient treatment. After completing each set of radial wedged readings, the wedge was reversed to verify proper wedge placement within the beam.

With the variance of readings conducted in air and with full backscattering material, the difference in readings that would be obtained on a patient with varying densities of underlying body parts was considered. The diode was

placed on an anthropomorphic phantom and readings were taken over the lung, clavicle, anterior aspect of the skull, hip, lateral skull (in the area of the mastoid bone), mid-abdomen off the spine and mid-abdomen over the spine. The diode was calibrated to 200 cGy with full backscatter polystyrene in a 10 cm x 10 cm field at an SSD of 80 cm. The calibration was checked again with the polystyrene phantom after all humanoid measurements were taken to verify the calibration had remained stable over the course of the measurements.

Patients may have multiple fields on any one treatment and radiation therapy departments may want to monitor them simultaneously. Therefore, the reproducibility between two diodes connected to different channels of one electrometer was examined. Both diodes were calibrated independently for 400 cGy at a dose rate of 300 cGy/min. Readings were then obtained for a total of 100, 200, 300, 400, 500, 600, and 700 cGy, and readings compared.

The research then turned to electron work with the diodes. All electron work was performed using a 14.8 cm x 14.8 cm field size and the standard 10 cm x 10 cm cone, at an SSD of 100 cm. An 11" x 11" x 3" polystyrene phantom was used as backscatter material.

First, linearity of the electron diode was considered. The machine was set for 6 MeV electrons and the diode was calibrated at 500 cGy/min for a total of 300 cGy. Readings

were then taken at total absorbed dose settings ranging from 100 cGy to 900 cGy at 100 cGy intervals.

The effect of energy dependence was investigated by using the same 6 MeV calibration and accumulating 300 cGy at a dose rate of 500 cGy/min. Readings were obtained at 9, 12, 15 and 18 MeV. Because of the strong energy dependence demonstrated, calibration factors were determined for all energies, with the machine calibrated at each individual energy. To accomplish this, the diode was calibrated for a total of 200 cGy at 6 MeV. Readings were then taken at 9, 12, 15, 18 and finally 6 MeV (for calibration verification). The value obtained at the new energy was divided by the reading at the calibration energy to determine an appropriate factor. The unit was then set to all succeeding energies and tested for a total of 200 cGy at all other available energies. The above mentioned manipulation was applied to all resultant values in order to obtain correction factors for all energies and settings.

A test to determine the effect of off-center placement of the diode in electron patient work was evaluated by taking readings in both the x and y directions, at 0 cm, 0.5 cm and 1.0 cm from the central axis.

Finally an examination of angular dependence on electron readings was performed. Considering the shallow angles used in electron radiotherapy, readings were obtained from 174°

to 186° through the neutral position of 180°. An exaggerated, and clinically non-applicable measurement was taken at 163° to set outside limits for accuracy.

Patient data were obtained during the course of normal treatments by Kurt Blodgett, a physicist at the Radiation Oncology Center. He performed all diode placements and recorded the resultant readings. All diodes were placed in the center of the treatment field with the top of the diode facing the beam. In cases where beam central axis was blocked, the diodes were placed in the center of the treatment field. None of these procedures were experimental. All treatment plans were calculated normally for the individual patient and tumor type.

Patient data is reported by % standard deviation only to preserve patient anonymity and to determine if the target accuracy of +/- 5% could be obtained. No long term follow-up studies were done to determine if those patients who received doses outside of the +/- 5% accuracy value had a poor outcome with respect to tumor control.

The electrometer readout was in whole numbers only; however, the manufacturer states that it is accurate to one more unit than is presented on the electrometer face. Therefore, it was assumed that when the last whole number flipped back and forth between two adjacent whole numbers, it should be represented as ".5".

Because the diodes are designed with integral filter/buildup caps, they are in equilibrium and exclude low-energy radiation. Therefore, when calibrated, the reading represents the dose at depth maximum. For 4 MV photons, D_{max} in tissue is at a depth of 1.2 cm. As a result, all diode readings were taken on the surface of any phantom, and ion chamber readings were taken with 1.2 cm of overlying phantom material. Both readings are therefore equivalent.

CHAPTER 3

RESULTS

Diode Linearity

The diode was first tested for linearity when subjected to varying total doses of photons. After calibration for a total of 200 cGy, all readings obtained from 100 through 700 cGy, in 100 cGy intervals, were within less than 0.2% of the intended value (Table 1). The diode system was then concluded to be linear with total accumulated dose.

Table 1. Diode linearity.

Accelerator Setting (cGy)	Diode Reading (cGy)
100	100
200	200
300	300
400	400.5
500	501
600	600
700	700

Reading Leakage from Zero and Accumulated Dose

The leakage rate from a reading of zero is presented in Figure 1. It can be seen that the diode should be zeroed immediately prior to initiating an irradiation.

Similarly, Figure 2 demonstrates that readings must be recorded immediately after the irradiation is terminated.

Otherwise, one can expect the electronics to bleed and false readings to result.

Figure 1. Diode leakage from zero.

Dose Rate Dependence

The dose rate dependence of diode readings was tested at clinically significant dose rates, accumulating a total of 200 cGy. Readings do not vary from the expected by more than 0.5% (Table 2).

Figure 2. Diode leakage from clinical reading.

Table 2. Dose rate dependence.

Dose rate (cGy/min)	Reading 1	Reading 2
100	200	200
200	199	199
300	199	199

Angular Dependence

The diode has a designated top that should be placed towards the beam. The manufacturer suggests that when the diode is taped to a patient, the tape should not be placed

over the cross-hairs that delineate the effective center of the diode. Therefore, it was not unexpected to find directional dependence at extreme angles. This is shown in Figure 3.

Figure 3. Directional dependence of diode in axial and transverse orientations.

As can be seen above, directional dependence is not pronounced in the axial direction. This orientation occurs when the gantry rotates around the short axis of the diode, essentially around its circular center. Accurate readings are obtained from gantry angles ranging from 140° to 210°.

This range is represented on Figure 3 by a pair of dashed vertical lines. The maximum percent difference is 2.0%, at an angle of 240° . It is concluded, that for an axial orientation, correction factors are not necessary except at extreme angles.

The transverse direction shows a more severe dependence. In a transverse orientation, the gantry rotates across the long axis, or length, of the diode. The diode was situated such that the lower angles ($<90^\circ$) are irradiating the cord end of the diode, and the higher angles ($>90^\circ$) are over the tip end. As can be seen in Figure 3, transverse measurements are linear over a shorter range of 155° to 190° than those for axial data points. This range lies within the pair of vertical straight lines shown in Figure 3. The maximum percent difference in this orientation is 5.5% at a gantry angle of 240° . This non-linearity demonstrates that if additional fields will be irradiated in the same patient, the diode must be replaced in the center of each field in the proper orientation. This should remove all reading errors due to positional dependence. The diode was not tested with a dynamic rotating field.

The percent differences found in the angular dependency tests are consistent with those values presented in the manufacturer's operating instruction booklet for 6 - 8 and 10 MV detectors (Nuclear Associates 1987).

Field Size Dependence

Correction factors for field sizes deviating from a standard size are known for ion chamber readings. These factors must be taken into account when calculating a patient treatment plan. Since the factors are known to correctly adjust ion chamber readings based on field size, a test was done to see if the same correction factors could apply to diode readings. All duplicate results were averaged, then the resultant diode value was divided by the ion chamber value to obtain a ratio. If the ratio remained constant around the reference field size (10 cm x 10 cm), then it could be postulated that the diode responds the same as the "gold standard" ion chamber. This in fact was found to be true, and the results can be seen in Table 3. The percent difference value was calculated comparing ratio values at varying field sizes with the standard 10 cm x 10 cm value. All readings are indicated in the table in the order they were taken. The 10 cm x 10 cm field size reading was taken three times. Once at the beginning, in the middle and at the end. These three values were averaged to determine the 10 cm x 10 cm ratio value used in the percent difference calculation. The percent differences shown, represent statistical variations in the readings. It can therefore be concluded that diodes respond to field size changes in a similar fashion to ion chambers.

Table 3. Field size dependence.

Field- Size (cm)	Ion Chamber (10^{-8} Coulombs)		Diode Readings (cGy)		Statistics	
	#1	#2	#1	#2	Ratio	%Diff
10x10	2.70	2.70	200	200	74.07*	-
32x32	2.88	2.88	213	213	73.96	-0.03
30x30	2.87	2.88	213	213	73.91	-0.09
25x25	2.86	2.85	211	211	73.90	-0.11
20x20	2.83	2.83	209	209	73.85	-0.18
15x15	2.78	2.78	206	205	73.92	-0.08
12x12	2.74	2.74	202	202	73.72	-0.35
10x10	2.71	2.71	200	200	73.80*	-
9X9	2.69	2.69	198	198	73.60	-0.51
8X8	2.66	2.66	197	197	74.06	+0.11
7x7	2.64	2.64	195	194	73.77	-0.28
6x6	2.61	2.61	193	193	73.95	-0.04
5x5	2.57	2.57	191	190	74.12	+0.19
4x4	2.53	2.53	189	188	74.50	+0.70
10x10	2.70	2.70	200	200	74.07*	-

* Average of three 10x10 reference values is 73.98.

Effects of Worst-Case Treatment Plan on Diode Placement

In order to find out whether diode measurements would be accurate for difficult treatment plans, doses were measured with a typical breast irradiation setup. This uses a breast beam splitter to spare non-breast tissue and either a 15° or 30° wedge to correct for tissue thinning at the edge of the breast. This setup is seen in Figure 4.

Figure 4. Typical breast irradiation setup.

The breast beam splitter is a 15 cm x 15 cm shield. It has lead on one-half of it to eliminate radiation to the contralateral breast and chest wall, and plastic on the other half to allow the beam to reach the target area. The central beam axis is at the junction of lead and plastic. The steep angle of the beam allows irradiation only of the affected breast, and not the lung or chest wall beneath. The wedge is used to shape the beam flux in order to create an equal absorbed dose per volume tissue across the thicker breast center and out to the thinner edges laterally. Measurements were taken 1 cm from the penumbra to be as close to the central ray as possible. The results can be observed in Table 4.

From this table, it can be seen that the readings 1 cm from the penumbra do not vary when the 15° wedge is replaced with the 30° wedge. This implies that the wedge thickness

is the same close to the central ray. It then makes it clear that if the beam splitter is in place, but a wedge is left out, there will be a 6.0% difference in the expected reading. In addition, if both the beam splitter and any wedge is left out (open beam) the reading will be 9.4% higher than expected. Both of these errors should then be detected as they are greater than the +/- 5% accepted error.

Table 4. Worst case treatment measurement effects.

Beam Shaping Device in Beam -----		Diode Reading (cGy)	Percent Difference if beam shaping device left out
Beam Splitter	Wedge		
+	-	398	+6.0%
+	15° wedge	374	-
+	30° wedge	374	-
-	-	413	+9.4%

Calibration Optimization

When the diode was calibrated at 200 cGy, it was more linear over the clinically useful range (150 - 200 cGy/treatment), than when it was calibrated at 400 cGy. Therefore, it should be calibrated for 200 cGy when used in clinical practice. Results of calibration optimization testing is seen in Table 5.

Table 5. Optimum calibration test.

Target Reading	Actual Reading when calibrated at:	
	200 cGy	400 cGy
100	100	101
200	200	201
300	299	300
400	398	400

Inverse Square Law Verification

To ensure that the diode responded the same as an ion chamber with varying source to detector distances, readings were taken with both detectors from 60 cm to 100 cm SSD. Figure 5 shows the geometry of the measurements. Specifically the top of the Lucite block is 80 cm from the x-ray source. The diode is placed on top of the block such that its center of detection is $80 \text{ cm} - r_d$ (radius of the diode), or $80 - 0.39 \text{ cm} = 79.61 \text{ cm}$. The ion chamber is placed at depth of dose max, for 4 MV photons, that is 1.2 cm of Lucite. Therefore, the ion chamber is $80 \text{ cm} + 1.2 \text{ cm}$ from the radiation source, or 81.2 cm. Both detectors were offset to eliminate any possible attenuation from the diode overlying the ion chamber.

The resultant readings were tested with the formula shown in Figure 6 to ensure that the ratio of the diode reading at any distance, divided by the chamber reading at the same distance, inverse square corrected for both

Figure 5. Geometry of inverse square verification experiment

Figure 6. Inverse square verification formula.

$$\frac{(R_{D_x})(R_{C_{80}})}{(R_{C_x})(R_{D_{80}})} \frac{\left[\frac{81.2}{x + 1.2} \right]^2}{\left[\frac{79.61}{x - 0.39} \right]^2}$$

chambers to 80 cm, would produce a cluster of values centered around 1.0000. The raw data, along with the resultant values obtained after applying the formula, can be seen in Table 6.

Table 6. SSD verification measurements.

SSD (cm)	Observed Readings		Applied Formula Result
	Ion Chamber	Diode	
60	7.015	363	1.0061
62	6.56	339	1.0065
64	6.16	318	1.0070
66	5.795	298	1.0046
68	5.46	280	1.0033
70	5.14	263	1.0023
72	4.865	249	1.0039
74	4.60	235	1.0032
76	4.36	223	1.0055
78	4.15	211	1.0006
80	3.94	200	1.0000
82	3.75	190.5	1.0017
84	3.57	181	1.0007
86	3.405	172.5	1.0008
88	3.255	165	1.0022
90	3.11	157	0.9989
92	2.97	150	1.0000
94	2.84	144	1.0047
96	2.73	138	1.0024
98	2.62	132	0.9997
100	2.51	127	1.00047

It should be noted that none of the relative values obtained from the formula test exceed an absolute value of 0.0070 from the expected value of 1.0000. The extremely small variance seen in the numbers is clearly statistical variation in the data acquisition.

The inverse square experiment was repeated using a total accumulated dose of 600 rads, to see if the resultant formula values would be more tightly clustered around 1.0000, demonstrating that the variance seen was purely statistical. This proved to be the case, as the maximum absolute deviation from 1.0000 was calculated to be 0.0038 with the higher dose.

Tray Effect

Plastic trays are used in the beam to position various beam shaping devices. The attenuation effect for photon beam measurements with an ion chamber is well known (accepted to be between 3 - 4%) and taken into account when calculating patient treatment plans. Using an SSD of 80 cm, measurements were made with both the ion chamber and the diode to verify that diode readings would demonstrate a similar attenuation factor. Ion chamber readings taken with the tray in place and then removed show an attenuation of 3.76%. Similar readings for the diode yield a tray effect of 3.72%, a difference of only 0.04%.

The readings were repeated at an SSD of 65 cm, bringing the measuring instruments closer to the x-ray source to determine if the diode had enough buildup material. If adequate material was not present, it was expected that electron contamination would take place and readings would vary from the expected difference of 0.04% demonstrated in the earlier experiment. The results showed a tray effect of 3.21% for the ion chamber at 60 cm SSD, and one of 3.25% for the diode at the same distance. The 0.04% difference is preserved and thus verifies that the diode has enough buildup material for the 4 MV beam.

Contribution of Backscatter - The Peak Scatter Factor

The peak scatter factor for ion chamber measurements in a 4 MV beam using a 10 cm x 10 cm field size is 1.035. This means that backscatter from the absorbing medium contributes an excess dose of 3.5% at the surface of the medium. This was tested for the diode by performing measurements in air at an SSD of 80 cm and then on top of the State Phantom (7-7/8" Lucite) at the same distance. The excess dose due to scatter was measured to be 3.49%, agreeing very well with established values for the ion chamber.

Field Uniformity

Before undertaking an evaluation of the effects wedges have when placed in the beam, it was important to measure the beam uniformity in the x and y directions with an open

beam. The pattern mapped by the diode was expected and follows norms for these types of readings. The results can be seen in Figure 7.

Figure 7. Field uniformity in the x and y directions.

The inverted shape is a result of a "flattening filter" inserted into the head of the x-ray tube. This filter is shaped concave on one side, and flat on the other. It is used to "flatten" the beam across the entire field. This is needed because of the intense forward scattering of high energy photons from the target. The filter is designed to "flatten" the beam at a depth of from 4 - 10 cm, and

therefore it will not be flat at the surface. The curves shown in Figure 7 show the effect of beam hardening as it passes through the thicker concave center of the filter. The beam is softer at the field edges, and therefore is less penetrating, leading to a higher dose rate at the surface. The transverse and radial curves are equal to each other, and equivalent to curves plotted using data taken from an ion chamber.

Wedge Effects

With this result in mind, measurements were taken in the radial direction across four different wedges: 15° , 30° , 45° and 60° . Wedges are special beam-shaping devices placed in the beam and shaped exactly like a wedge, with a thick end tapering to a thin end. They are used to attenuate the beam in a continuous fashion when the treatment area contains non-uniformly thick tissue. For example, if the treatment area is the neck, then, at the center of the neck, the beam is passing through tissue approximately 17 cm thick. However, at the edge of the neck, that same flux of photons is traveling through a few centimeters. It is obvious that the absorbed dose per volume tissue would be excessive at this point. Other treatment setups which require the use of a wedge include breast, lung, multiple fields, or non-opposed fields. When the thickest part of

the wedge is placed over the thinning tissue volume, it reduces the flux appropriately.

An important reason for describing the attenuation across all wedges is to see what effect misplacement of the diode (off the center of the wedge) would have on perceived patient dose values. The results are seen in Figure 8.

Figure 8. Effect of wedges on dose

It can be seen that doses are reduced by as much as 73% at the thick end of the 60° wedge. In order to determine the effect of a missing wedge, or insertion of the wrong wedge, values at the centerline of each wedge were compared

with the value obtained with no wedge in the beam (200 cGy). The results are seen in Table 7.

Table 7. Wedge effects

Type of Wedge	Reading (cGy)	% Reduction in Reading from Open Field Reading
Open Beam	200	-
15°	161	19.5
30°	154	23.0
45°	115	42.5
60°	97	51.5

From the numbers in the table above, the error for mistaking one wedge for another can be calculated. This assumes that the diode is placed in the center of the treatment field, along the midline of the wedge. A mistaken wedge setup error could be distinguished by the total dose diode reading (i.e., exceeding the 5% acceptable error), except for the difference of only 3.5% between the midline measurements for the 15° and 30° wedges.

Another important factor which can be derived from the wedge data is the error associated with misplacement of the diode off-center in the treatment field. From the steep dose curves seen in Figure 8, it is apparent that if the diode were not placed exactly in the center of the field, errant readings could be obtained, and attributed to treatment error. Values taken from 1 cm and 2 cm on both sides of the centerline (0) value are shown below. The

wedge is oriented such that the thicker portion is towards the positive position values.

Table 8. Percentage error in diode placement for wedged fields

Type of Wedge	Raw Data (cGy)/% of Centerline Value				
	-2cm	-1cm	0	+1cm	+2cm
15°	172 +6.3%	165 +2.5%	161	158 -1.9%	158 -1.9%
30°	170 +10.4%	161 +4.5%	154	149 -3.2%	145 -5.8%
45°	136 +18.3	125 +8.7%	115	108 -6.1%	104 -9.6%
60°	125 +28.9%	109 +12.4%	97	89 -8.2%	81.5 -16%

It can be seen that diode placement is critical, especially in the thicker wedges. Improper placement could lead to the false assumption that there was a serious over- or under-treatment of the patient.

Effects of Patient Backscatter

Next, the effect of the backscatter material used for calibration was assessed, along with an attempt to quantify the differences when the diode is placed over varying densities of patient tissue. Backscatter, and therefore increased diode readings, would be expected if the diode

were placed over bone versus air (lungs). The following Table 9 identifies the results of using a humanoid phantom to test different body areas. The diode was calibrated to 200 cGy with 10 cm x 10 cm of polystyrene backscattering material at an SSD of 80 cm.

Table 9. Effect of patient treatment density on diode backscatter readings

Underlying Tissue	Result (cGy)
10cm x 10cm polystyrene	200
Lung	197.5, 197
Clavicle	199, 199
Skull (over sinuses)	198, 198
Hip	200, 199.5
Lateral Skull (mastoid bone)	201.5, 201
Mid-Abdomen (off spine)	197, 197
Mid-Abdomen (over spine)	200, 200

From these values, it can be seen which organs duplicate the backscattering ability of the standard calibrating phantom. The values obtained are expected, such as

decreased backscattering for the lungs and increased backscattering for the densest bone in the body, the mastoid. The largest percent difference is for the lung value at 1.5%.

Dual Diode Mode

The electrometer used for this thesis work had two BNC input ports to acquire readings from two diodes simultaneously. It also is equipped with two readouts, one for port A and one for port B. There are times when two diodes may be needed to characterize a treatment dose, therefore, two diodes were calibrated and subjected to concurrent radiation doses of 100 to 700 cGy. The results in Table 10 show that both responded similarly and intra-diode variability in the diodes used was negligible.

Table 10. Dual diode linearity

Set to Acquire (cGy)	Diode Reading (cGy)	
	A	B
100	100	100
200	200	199.5
300	299.5	300
400	399	399.5
500	499	499
600	599	599
700	699	699.5

Electron Investigation

The possibility of using the diodes to perform routine quality control tests on electron producing machines was investigated. First a basic linearity test was performed using 6 MeV electrons. With the diode calibrated at 500 cGy/min for a total of 300 cGy, the accelerator was set to accumulate from 100 to 900 cGy in 100 cGy increments. The results can be seen below in Table 11.

Table 11. Linearity of diode response to 6 MeV electrons

Accelerator Set to Deliver (cGy)	Diode Reading (cGy)
100	100
200	200
300	300
400	400
500	500.5
600	600.5
700	700.5
800	800.5
900	900

It can be seen that the diode responded quite well to increasing amounts of total dose.

The 4 MV linear accelerator used for photon work only allowed 4 MV photons to be produced. However, the Clinac 18 used for electron work allows the user to select 6, 9, 12, 15 or 18 MeV electrons. It therefore became important to see how the diode would respond to other energies if it was calibrated at a single energy. The results show that the diode's response is very energy-dependent. When calibrated for 300 cGy at 6 MeV, values ranged from 299 cGy of accumulated dose for the 9 MeV setting, up to 321 cGy for the 18 MeV setting. Because the diode would be calibrated at one energy each morning, calibration factors were devised for each possible combination of calibration energy and patient dose energy. For this part of the investigation, the accelerator was set to accumulate 200 cGy per reading. The results can be seen in Table 12.

These calibration factors are designed to be used in the following manner. If the diode was calibrated in the morning at 12 MeV but the patient was treated with the 9 MeV electron beam; the resultant dose read from the diode electrometer would need to be multiplied by the proper calibration factor to obtain the true patient dose at D_{max} . In this case, the calibration factor is 0.95.

These conversion factors were calculated for each day that experimental readings were taken, and a slight variation was noted. It is therefore recommended that if

the diode is used for the purpose of monitoring electrons, calibration factors should be determined each day of use. It is only necessary to calibrate at one energy, then take readings at all other energies. This is not an excessively time-intensive protocol.

Table 12. Experimentally derived conversion factors for varying electron energies

Calibration Energy (MeV)	New Energy Setting (MeV)	Diode Reading (cGy)	Calibration Factor
6	9	195.5	0.98
	12	206.5	1.03
	15	211	1.05
	18	213	1.06
9	6	204	1.02
	12	211	1.06
	15	216	1.08
	18	218	1.09
12	6	193	0.96
	9	190	0.95
	15	201	1.02
	18	206.5	1.03
15	6	189	0.94
	9	185	0.92
	12	195	0.98
	18	202	1.01
18	6	187.5	0.98
	9	184	0.92
	12	194	0.97
	15	198	0.99

Electron treatment fields are usually smaller than photon fields. Diode placement was proposed to be very critical. Therefore, the effect of off-center placement was evaluated by moving the diode 1 cm and 2 cm at 90°, 180°,

270° and 360° from the central point. This investigator was surprised to see that there was no fluctuation from the intended value of 200 cGy in any direction up to the maximum distance of 2 cm. It was then concluded that to obtain accurate results, diode placement was not as critical as in a wedged photon beam. The diode must simply be placed within an inch of the beam axis.

Finally, angular dependence of an electron beam was considered. Steep treatment angles are not generally used in electron treatments. As a result, angular dependence data was obtained in the radial rotation, from 186° to 174° (assuming that 180° is the neutral position, with the machine head pointing down to the floor). No variation from the intended dose of 200 cGy was seen. To measure the results obtained from an angle beyond those used in clinical practice, a reading was taken with the head in a position of 163° to the diode. Again, the reading was 200 cGy. It is then concluded that within the clinically useful angles, the electron diode does not demonstrate angular dependence. No similar measurements of these types were seen in the literature for comparison.

Patient Study

The clinical readings consisted of 114 measurements from a population of 32 patients. Of these readings, 106 (93%) fell within the acceptable accuracy range of +/- 5%. The

standard deviation of the sample set is 2.98% error. The number of values falling within one standard deviation is 92 out of the 114, yielding a percentage of 80.7%. A total of 107 readings were within two standard deviations, or 93.8%. These statistics indicate a tightly clustered set of values around a mean of -0.24%, and a median of -0.71%.

The mean of -0.24% indicates that the projected treatment ordered by the Radiation Oncologist was successfully calculated by the treatment plan software. In other words, there is a lack of a systematic error shown. Leunens, Van Dam, Dutreix and van der Schueren (1990) performed a similar analysis on a patient population of 68, obtaining a total of 554 measurements. Their data showed a systematic error of -2.2%. After investigation, they concluded that it was due to inaccuracies in the algorithm used for the dose calculation with the treatment plan system and to an inaccuracy of the absorbed dose determination with the ionization chamber in the region of the maximum build-up depth. Specifically, they discovered that scatter effects due to the use of shielding blocks, or when part of the field was in air, were not taken into account by the computer software. They were using a 6 MV beam and found, after taking phantom measurements, that the dose was being overestimated by 0.5% to 1.0%. In addition, they had applied a displacement correction factor to correct for the

displacement between the effective measurement and the ionization chamber center. They concluded that the factor should not have been applied at depth of dose maximum due to the shape of the depth-dose curve. They estimated that the application of the displacement correction factor led to an underestimation of the measured dose of 1.25%, and resulted when they failed to calibrate the diodes against the ion chamber reference depth of 5 cm. After accounting for these errors, their data centered around the expected 0.0% error value. For the data accumulated for the purposes of this thesis, the mean of -0.24% therefore indicates that diode calibration procedures and assumptions made in the treatment software are valid.

The patient setups were reviewed to ascertain which used wedges or compensators. It was interesting to discover that all values exceeding 4.0% error used one or both of these devices. For example, the largest error recorded was +13.41%. This particular treatment field setup was very complex. Therefore, the physicist performing the diode placement and recording resultant values repeated the measurement on succeeding treatments. Even with careful diode placement, a percent error of no less than +6.80% was obtained. The field was subsequently discontinued due to the fact that it could not be adequately reproduced in the clinical setting.

A total of 50 out of 114 measurements were taken in wedged or compensated fields. The results of both wedged, and unwedged fields may be seen in Figure 9.

Figure 9. Percent error of patient measurements

The excessive error seen in wedged fields may be due to diode placement off of the wedge center, or to other unidentified setup irregularities. It is clear that special care must be taken when documenting patient doses employing wedged fields.

CHAPTER 4

DISCUSSION

Are Correction Factors Necessary When Transcribing Diode Readings?

Documenting patient doses with a silicon diode system can be accurate and simple. In the experimentation performed for the purposes of this thesis, all acceptance testing criteria (linearity, angular dependence, wedge effects, etc.) were verified to be accurate without the application of a "correction" factor for a 4 MV photon beam. This allowed readings to be taken directly from the electrometer after completion of a patient treatment. This was further verified by the patient results percent error mean (-0.24%). It is expected that, if correction factors needed to be applied, a systematic error would have been identified, and the mean would vary more from a 0.0% error. From acceptance testing performed by other authors, this is not always the case.

Temperature

Even though temperature sensitivity of the diode system was not investigated within the scope of this thesis, the literature contains several interesting views on the perceived problem. Reinstein, Kalend, Verga and Meek (1986) performed an exhaustive comparison of two types of diode

systems. They report no dependence on temperature from 15° C to 33° C. Gray (Nuclear Associate Monograph, no date given) also reported no temperature dependence when the diode he tested was placed in a water bath varying from 20° to 37° C. However, Heukelom, Lanson and Mijneer (1991) noted some of their diodes increased in sensitivity with temperature, some showed a decrease, and some were unaffected. They used seven p-type diodes with 8 MV and 4 MV beams. The diode system they tested was the same as one of the two systems used by Reinstein, Kalend, Verga and Meek, which showed no temperature effects. Another group of authors, Nilsson, Ruden, and Sorcini (1988), report when the diode is placed on a medium of a different temperature, they found an increase in sensitivity of 2 - 3% over the period of 3 - 4 minutes that it took the diode to reach steady state equilibrium.

Nilsson, Ruden and Sorcini (1988) also state that measurements should not be taken on a patient unless the diode has had 3 - 4 minutes to come to temperature equilibrium. As an alternative solution, they suggest that the diode could be calibrated on a phantom with a raised temperature, or a calculated calibration factor could be applied. These last two options are not as accurate, they warn, because individual diodes differ, especially after

accumulated radiation damage, and it would be time consuming to continually calibrate each diode repeatedly.

A discussion with Steve Szeglin (personal communication February 8, 1995) stated that the undoped n-type diodes used for this thesis experimentation have a temperature coefficient of only 0.1%/°C, as opposed to those values (1.5% - 2.5%/°C) published for p-type diodes. All other authors described above used p-type diodes.

Field Size

For field size investigations, Reinstein, Kalend, Verga and Meek (1986) report that diode measurements agreed with ion chamber results with a maximum discrepancy of 1.7% for one system and 0.7% for the other system tested. These numbers exceed those found in this thesis investigation - an average of 0.22% with an outside value of 0.70%. They conclude that there is diminished backscatter seen by the diode detector element (which is above the surface of the phantom) as compared with the ion chamber (placed at $D_{\max}$ beneath the surface). They conducted further measurements in air to verify this hypothesis and the results showed closer agreement, thereby confirming their supposition.

Nilsson, Ruden and Sorcini (1988) also report that the diode functioned linearly over varying field sizes (ratios of ion chamber readings to diode readings varied by less

than 1%). They cite another author as stating that correction factors were needed, and they conclude that it could be related to an inadequate amount of build-up material on the diode used.

Angular Dependence

Angular dependence is another area where literature results varied. Different types of diodes showed a stronger angular dependence than other brands. Reinstein, Kalend, Verga and Meek (1986) were able to show that the two diode systems they tested had very dissimilar results. One brand of diode showed a marked decrease in sensitivity (2 - 5%) at angles varying from a neutral position of only 20° or greater in the transverse direction. Similar readings in the axial direction remained fairly stable. Conversely, for the other brand of diode, the axial angular dependence was more pronounced (1.5 - 2%) at outside angles (+/- 30°) and the transverse direction varied up to 5% when the gantry was rotated over the cord end.

Less angular dependence was demonstrated in this thesis, with a maximum of 2% difference in the axial direction at a full +60° from the neutral position, and 5.5% in the transverse direction, also at +60°. It is interesting to note that the thesis data show a greater decrease in diode sensitivity in the transverse direction when the gantry is

rotated over the diode tip. Measurements taken by Reinstein, Kalend, Verga and Meek (1986) show a greater drop when rotating over the cord end of the diode.

Nilsson, Ruden and Sorcini (1988) demonstrated a variance of 10% depending on the angle of variance. Their raw data were not presented to derive a more precise analysis of the diode orientation and exact angles. It is noted that they took measurements up to 90° to either side of the neutral position. They suggest that angular dependence must be considered at "oblique" treatment angles.

Wedges

The last area where authors had a significant variety of opinions was whether a wedge factor should be applied or not. Nilsson, Ruden, and Sorcini (1988) worked with Co-60 gamma radiation, and 4 MV, 6 MV and 21 MV x-ray beams. They describe an interesting effect for all but the 4 MV x-ray beam. They state that when placing wedges in the beam, a large difference was obtained for all energies and fields, except the 4 MV diode. The measured values were 3 - 6% lower than expected. The results were obtained by taking the mean value of measurements with the wedge in two opposite directions, thus minimizing the effect of displacing the diode. The main reason for these lower values was postulated to be a dependence on dose rate for

those diodes. Specifically, measurements of the diodes used in their investigation showed that decreasing the dose rate by a factor of two may decrease the response by up to 4%. This change in dose rate was obtained with a wedge angle of 45° . They further state that the dose rate dependence varied with each diode and with accumulated dose (radiation damage).

Finally, they state that the diode used with 4 MV x-rays did not demonstrate a dose rate dependence; therefore, they did not observe any unexpected difference between an open field and a wedged field in a 4 MV beam. This result is verified by the data obtained in this thesis with a beam of the same energy. In addition, Reinstein, Kalend, Verga and Meek (1986) also agree that diode measurements, when compared with ion chamber measurements, did not vary more than the experimental uncertainty of the diodes when placed in wedged fields, and therefore they do not suggest use of wedge correction factors.

Heukelom, Lanson and Mijneer (1991) further investigated wedge effects by plotting relative diode sensitivity against beam energy. For exit measurements, they observed a decrease in sensitivity from a value of 150 kV to 1 MV, then a slight rise when tested to a maximum value of 16 MV. Entrance measurements show a more shallow decrease along the entire curve, with no upslope. They compare their

wedged results to these curves, and explain that wedge factors are necessary. Specifically, they state that if a wedge is inserted in a photon beam, the photon energy spectrum changes due to beam hardening while the dose per pulse value varies with the wedge factor. From their sensitivity versus energy graph, they deduce that for an entrance measurement, a small reduction in diode sensitivity is expected with increasing wedge angle due to specific absorption of lower-energy photons in the wedge. They state that for four of their seven diodes, the sensitivity was only slightly influenced by the decrease of the dose per pulse value, and conclude that their wedge correction factor is mainly determined by the beam hardening due to the wedge.

Radiation Damage

Radiation damage over time is another area that was not investigated within the parameters of this thesis, but is discussed in the literature. The measurements taken for this thesis occurred over a period of four months, making it difficult to assess long term radiation damage. Work with the electron diode took place over the period of only one week. All papers reviewed stated that p-type diodes are superior to n-type with regards to minimizing radiation damage. However, the manufacturer of the undoped n-type diodes used in this thesis states that they are now superior

to p-type diodes, using new manufacturing techniques. The following researchers all used p-type diodes.

Heukelom, Lanson and Mijnheer (1991) designed their experiments to take ratios of a dose value, D , at entrance and exit points with a calibrated ionization chamber, to the reading, R , of the diode, irradiated in a reference field. Their reference field consisted of a 15 cm x 15 cm field size, an SSD of 100 cm, a 15 cm thick polystyrene phantom and a diode temperature of 22° C. On a weekly basis they performed reference readings and calculated calibration factors for different field conditions, including changes in target thickness, field size, SSD, temperature, and whether a wedge is used in the field or not. By plotting the basic ratios of ion chamber readings divided by diode readings for the reference set up, they report a small and linear increase in the value of the calibration factor for six of the seven diodes tested. The increase amounts to approximately 4.6 and 3.5% in 500 days for the two oldest diodes. They report that the calibration factor of one diode initially decreased 18% in the first 120 days, then increased after that. According to these authors, for two diodes followed over a period of 2.5 years, calibration factors showed an increase of approximately 2.2%/kGy of accumulated radiation dose.

The calibration method used for the purposes of this thesis involved setting the accelerator to accumulate a certain cGy value, then adjusting the calibration potentiometer to read that value after irradiation was completed. Because the calibration was performed each day of use, the absolute amount of sensitivity lost would be difficult to assess. All authors who have investigated this phenomenon, with electrons and photons, agree that radiation damage is inevitable. However, with proper calibration, it should not affect the accuracy of results.

Patient Data

In this thesis investigation, 114 measurements were made on 32 patients. Fifty of the measurements were made in a wedged field (43.8%). The overall standard deviation of the percent error calculated from comparing the prescribed absorbed dose to the actual delivered absorbed dose (diode reading) for all measurements was 2.98%. The patient percent error values ranged from -7.55% to +13.41%, with a mean of -0.24% and a median of -0.71%. Eight values (7.0%) exceeded the acceptable accuracy of $\pm 5\%$. All readings which fall into this range were taken from wedged fields. The standard deviation of just the wedged readings was 4.03%, with a mean of -0.06% and a median of -0.87%. The 95% confidence interval (1.15) for wedged readings was nearly twice that of

the 95% confidence interval calculated for all readings (0.55).

Leunens, Van Dam, Dutreix and van der Schuren (1990) present statistics from a total of 554 measurements. They separated their data into similar field types. The overall standard deviation was reported to be 2.75%. The systematic error leading to a mean of -2.2% was discussed in a previous section of this paper, and was due to errors in the treatment planning algorithm. They report standard deviations of 3.34% for treatment setups without a wedge, and 2.31% for wedged setups. They acknowledge that this is contrary to the literature, and state that after reviewing studies by Nilsson, Ruden, and Sorcini (1988) showing the severe inaccuracy of a reading taken from a diode misplaced in a wedged field, they paid special attention to diode placement. They conclude, and this author agrees, that the positioning of the diodes in wedged fields is very critical and that interpretation of the results is impossible when the diodes are misaligned. They conclude that their care in positioning the diodes resulted in their lower standard deviation for wedged fields.

Data from Nilsson, Ruden and Sorcini (1988) calculated standard deviations for individual patients over the course of their treatment. Numbers of measurements vary from a low of eight readings per patient to a high of 111. Patient

measurements were performed during treatment with Co-60 gamma rays, and 6 and 21 MV x-rays. They found that the smallest standard deviations were performed with Co-60 treatments and conclude that this is due to the fact that irradiation conditions are more unstable in accelerator beams than in Co-60 beams.

For wedged fields, they report a mean of 3.0% (they report as 0.97) for Co-60 gamma radiation and 6 MV x-rays, along with a mean of 2.0% (they report as 0.98) for 21 MV x-rays. They attribute this to a non-corrected dose-rate dependence of the diodes. This dependence is not seen in the work of this thesis (wedged mean: -0.06%). They also averaged standard deviations for all patients treated with the same energy and relate average standard deviations of 3% for 6 MV beams, 4% for 21 MV beams and 2% for Co-60 beams.

Reinstein, Kalend, Verga and Meek (1986) performed a similar analysis on their clinical data; however, they separated the statistics into "simple" and "complex" treatment fields. Wedged fields were categorized as complex. They report individual patient standard deviations of less than 3.0% for 12 of 14 patients. The other two were 3.1% and 3.3%. The average standard deviation across the entire population was 2.0%. On complex fields (including placing the diode on a severely sloping surface, near the shadow of a block, or in a wedged beam), only 7 of 11

patients had individual standard deviations of 3.0% or less. The average across the entire patient sample was 2.7%. The greatest variations from prescribed individual doses were underdoses of 9.9% and 14.4%.

All standard deviations in the literature demonstrate that the 2.98% seen within the scope of this work is within acceptable limits.

The purpose of this thesis was to investigate if diodes could be used with each patient treatment to determine if the accelerator was properly set up and that the prescribed dose was delivered. Other authors have conducted clinical trials using diode systems and have identified just these types of problems.

Lee, Sawicka and Glasgow (1994) used an acceptable error level of $\pm 7\%$. This value was chosen because 7% is the lower bound of dose error that could result in a significant change in tumor control probability. Within their clinical trials, they investigated all treatments that showed a percent error of seven or greater. Repeat measurements, with careful attention to detail, brought all values within a $\pm 7\%$ error. They conclude it is not difficult to achieve a diode dosimetry accuracy of $\pm 7\%$ in radiation therapy in a well-managed department.

They did not specify what errors were found in their institution, but did provide theoretical causes. They state

that errors which could cause an error of 7% or greater, include setting the wrong daily dose on the accelerator, misuse of wedges, setting the wrong energy on multi-mode machines, or setting the treatment distance to either SSD or SAD when the other is indicated. They state that errors which would probably go undetected include improper bolus material, field size errors or misuse of a block tray.

Reinstein, Kalend, Verga and Meek (1986) identified specific reasons for irregular results. One patient's overall percent error was 10% lower than expected. The diode reading was taken 3 mm from the edge of a block shadow. Therefore, they conclude that this value is the true dose given to that area in the beam axis, next to the block, and the large variation was due to small daily changes in patient setup with lateral patient movement. They point out that this proves variations exist across a field which is assumed to be homogeneous. In addition, they indicate that diode readings near the edge of a block shadow can become an excellent indicator of patient setup reproducibility.

Another interesting patient reviewed by Reinstein, Kalend, Verga and Meek (1986), showed the first two readings of seven on the same patient falling within normal limits. The next five were decreased systematically. Investigation revealed that on the day of reading number three, a small

transmission filter was inserted into the beam. A further study of the chart and the timing of the diode readings brought out the fact that the transmission filter may have been inserted several days after the date indicated on the therapy plan.

Among the other errors identified, two were due to the technologist placing the diode in such a manner that angular dependence was maximized. These setups included treating a sloping thigh, and a breast treatment at an angle which exceeded 45° .

All of the investigations performed by Reinstein, Kalend, Verga and Meek (1986) were performed by chart review, after the patient's treatment had been completed. Leunens, Van Dam, Dutreix, and van der Schueren (1990) reviewed each patient setup when an unacceptable result was obtained real-time, before the patient was removed from the treatment couch. In this manner they identified discreet errors. These included: patient positioning errors; inaccuracies in the measurement of patient contour (true value 6 cm, treatment planning value 8.5 cm); repeated automatic interruption of the treatment resulting in an unsteady dose rate; the use of an isocentric (SAD) technique instead of the prescribed SSD technique; large shielding blocks, covering 45% of the field area; and treating one field at the appropriate SSD of 100 cm, then

rotating the field isocentrically and treating at 86 cm instead of the prescribed 100 cm. Their clinical study was composed of head and neck, and pancranial irradiations. Each error was immediately investigated, as stated previously, and rectified before the next treatment took place.

This appears to be the most efficacious and helpful way in which to use these measurements. Ideally, with time permitting, diodes will allow real-time identification of errors. If these errors are not corrected, but left to perpetuate throughout the treatment plan, gross over- or under-treatments will occur.

Individual investigations into each unacceptable reading were not performed at the time of each patient's treatment in this author's study. Due to time constraints, the physicist who placed the diodes, and took the reading, simply recorded the value. It is interesting to note that all of the unacceptable values were recorded from wedged fields. With the warnings of other authors on the critical nature of wedged field diode placement, it is most probable that the majority of these errors are a result of diode misplacement. As the setup was not verified in real-time, it is impossible to determine if other factors played a part in the error calculation.

CHAPTER 5

CONCLUSIONS

Verification of administered radiation therapy doses is imperative to affirm that the dose prescribed for appropriate tumor control was given. Over- or under-treatments of only 5% can reduce the tumor control probability significantly while causing serious harm to normal tissue meant to be spared.

Current methods of dose delivery verification include verification films and the use of thermoluminescent dosimeters or diodes to record the dose the patient has received (Blanco, Lopez-Bote, and Desco 1987). The diodes studied in this investigation showed excellent linearity, reproducibility, and a lack of significant angular and field size dependence. They are simple to use and provide a direct readout without correction factors.

As a result, errors in patient setup or changes in a patient's contour from the treatment plan measurements can be identified during the course of the treatment and corrected. This allows the treatment plan to be modified in succeeding doses in order to achieve an accurate total prescribed dose. This should be able to significantly increase the accuracy of treatment and avoid "unexplained" poor clinical outcomes.

The standard deviation seen in this patient population (2.98%) compares favorably with those values quoted in the literature. Ninety-three percent of patient readings demonstrated an error of 5% or less. Of the remaining 7% that exceeded the acceptable range of error associated with radiation therapy, all fields were wedged. This points out the critical nature of correct diode placement within a wedged field.

Patient monitoring was able to identify a complex treatment setup that was difficult to reproduce. Readings varied from +13.41% to just under 7% and this dose uncertainty led to this field being discontinued.

Therefore, it is concluded that diodes should be used to document all patient doses. For optimum results, time should be allowed for investigation of values that exceed 5%, immediately after the patient is treated. This would ensure that errors are corrected before the end of the treatment course.

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