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PIN diodes for radiation therapy use: Their construction, characterization, and implementation

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ABSTRACT

Purpose: Development and implementation of a PIN diode measurement system that can measure dose on the surface of a patient, which can be compared to dose calculated by a treatment planning system. Measurements are to be possible for static or rotational photon beams and electrons. Simple calibration procedures are to be devised that require a minimum set of correction factors.

Methods: Readily available PIN type photodiodes are fabricated into devices that can be used for detection of ionizing radiation. Single or dual PIN diodes are soldered onto flexible shielded cables for static and rotational irradiation use. Diode signals are measured with an electrometer with zero input bias voltage. Diode temperature is determined by operating it as a thermistor. Linac photon and electron dose is measured. A commercial treatment planning system is used for calculating dose expected in various test geometries.

Results: The scanning and surface diodes have an intrinsic buildup of 0.3 mm water equivalence and the IMRT device a buildup of 1.5 mm. Correction factors are determined for changes in diode sensitivity with the following factors: dose-per-pulse, dose rate, angle of radiation incidence, temperature, and field size. The surface or IMRT diode detector measurements agree within 1.0% with Eclipse calculations.

Conclusions: Clinically useable dose detection devices can be fabricated from PIN type photodiodes. These diodes are used to measure dose at the surface of a patient or under bolus. Correction factors are unnecessary if a surface or IMRT type device is chosen for use with static orthogonal or rotational IMRT delivered beams, respectively.

1. Introduction

The fundamental concepts and guidelines for clinical use of commercially available PN junction diodes have been presented in extensive review articles [1,2]. In these earlier works diodes were placed on the patient's skin surface and the dose at the depth of maximum dose was calculated from the diode measurements. This type of measurement required the application of many correction factors and the use of buildup caps on the diode to accomplish accuracy in predication of dose at depths other than where the diode was positioned.

In this work a PIN photodiodes versus a PN junction diode is used for detection of ionizing radiation in radiation therapy clinical use. PIN photodiodes are known to be able to detect ionizing radiation [3]. Construction methods are described that can be performed in the clinic for assembling useful diode systems from surface mount PIN diodes.

The PIN diodes are found to have different and preferable characteristics compared to PN junction diodes. The PIN diodes have intrinsic buildup of 0.3 mm water equivalent and are used to measure surface dose or dose under a bolus if it is used on the patient. Dose measured by the diode is compared to dose calculated by the treatment planning system for the diode position and depth under a bolus if it exists. The measurement of correction factors for changes in sensitivity to dose with dose-per-pulse, dose rate, accumulated dose, temperature, angle of incidence of radiation, radiation energy, and field size are presented.

Three types of diode systems are described: a scanning diode that is water proof and has intrinsic buildup of 0.3 mm water equivalence; a surface diode mounted on thin flexible cable that has 0.3 mm buildup; and an IMRT diode mounted on a thin flexible cable, with angular dependence of ± 1.7%, and intrinsic buildup of 1.5 mm. The thin flexible cable allows for precise, nonintrusive placement on the skin of a patient. With the proper choice of diode system, measurements can be made without the use of correction factors.

The diode systems have been used in our clinic for in vivo static field and IMRT dosimetry measurements, small field IMRT commissioning, and cone collimator SRS commissioning measurements. Procedures for using these diodes for clinical and small field measurements are discussed.

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Fig. 1. Measurement devices: Scanning detector, surface dose detector, IMRT detector shown from top to bottom. The white dot indicates the front surface of the diode.

2. Materials and methods

The diodes used in this work were silicon PIN photodiodes, model VEMD1060X01, (Vishay Semiconductor, Shelton, CT). The overall size of this surface mount device is $2 \times 1.25 \times 0.85\,mm^3$ and the radiant sensitive area is $0.48 \times 0.48\,mm^2$. While the VEMD1060X01 photodiode is designed for use in the 350–1070 nm range it is quite sensitive to X-ray and electron beam radiation.

The PIN diode is a variation of the PN-junction type diode [3,4]. The PIN diode has a thick intrinsic region, I, which is undoped and has no net charge carriers, that is between a P-type and N-type semiconductor regions. The P-type and N-type regions are heavily doped as they are used for signal contact points. It is in this large intrinsic region where photons are absorbed and cause electron hole pairs and a photocurrent to be generated. The large volume of the intrinsic region allows the PIN photodiode to have high sensitivity.

Three types of devices have been configured as shown in Fig. 1. They have been fabricated differently to optimize the characteristics of low intrinsic build up, small change in sensitivity with angle of incident radiation (angular dependence), flexibility of the connection cable, water resistance, and water proofing.

The PIN diodes are soldered to the end of a shielded cable. The scanning device is a diode which is mounted on coax cable, model 8216 RG-174/U (Belden, Inc. Indianapolis, IN), covered with black heat-shrink tubing, and water proofed with silicon sealant. The surface dose and IMRT detectors are mounted on 8 cm long, flexible shielded cables, model W2912 (Mogami Cable, Torrance, CA), are covered with black heat-shrink tubing that is heat sealed at its end, and are water resistance not water proofed. The flexible cable is spliced onto a 3 m long piece of RG-174/U coax cable. The flexible shield cable allows for easy taping of the detector on the skin of the patient without the diode twisting out of alignment from torque in the cable. The black heat-shrink tubing is needed to avoid visible light and dirt and moisture from reaching the diode.

The IMRT detector is fabricated from two PIN diodes mounted front-to-front with anode to cathode adjacent to each other as shown in Fig. 2. The two diodes are cemented together with cyanoacrylate glue. The two diode arrangement [5,6] greatly reduces the dependence of diode sensitivity on the angle of incident radiation, which is important when measuring IMRT dose that is delivered from many angles. Having the anode to cathode adjacent is shown in this work to give very small angular dependence. The two diodes are connected in parallel to the shielded cable.

All diode outputs were measured with a clinic built amplifier that integrated charge during radiation exposures or with a Max4000 electrometer (Standard Imaging, Madison, WI) operated in the zero-bias mode.

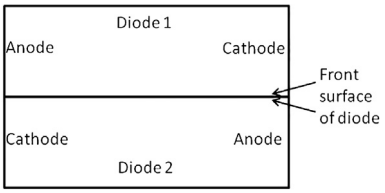

Fig. 2. Diagram of the arrangement of two diode that comprise the IMRT type device.

The X-ray beams used in this work had nominal energies of 6 and 15 MV. For these energies respectively, the percent depth dose of X-rays at depth 10 cm, $\%dd(10)_x$, was 66.4 and 76.5, which was measured at source-to-surface distance of 100 cm, according to the TG-51 protocol [7]. The electron beam that was used had a nominal energy 6 MeV, which had depths of 50% maximum dose, R_{50}, of 2.26 cm that was measured at source-to-surface distance of 100 cm, according to the TG-51 protocol [7]. The X-ray beams were generated by a Varian Trilogy (Varian Medical Systems, Milpitas, CA) linear accelerator.

Absolute dose measurements were made with a cylindrical ion chamber, model N30001 (PTW Hicksville, NY), which had been calibrated at the University of Wisconsin Dosimetry Calibration Laboratory. All doses delivered by the accelerators were compared against ion chamber measurements that were traceable to TG-51 [7] calibrations.

Dose measured by a diode is compared to dose calculated by the treatment planning system for the detector position and depth under a bolus if it exists. The treatment planning system used was Eclipse (Varian Medical Systems, Milpitas, CA). For X-rays the AAA and Acuros algorithms were used and for electrons the Monte Carlo algorithm was used.

To test for changes in device sensitivity with accumulated dose, devices were irradiated to tens of kGy of dose with Ir-192 gammas from a GammaMed Plus (Varian Medical Systems, Milpitas, CA) source and with 15 MV X-rays.

Irradiations of the diodes for various response tests and for calibration were done with the central axis of the beam orthogonal to the front surface of the diode with a source-to-detector distance of 100 cm. An 8-cm thick block of solid water was covered with a 0.5-cm thick piece of Superflab (Radiation Products Design, Inc., Albertville, MN). The diode was placed on top of the Superflab and various thicknesses of slabs of solid water were placed on top of the diode for desired depths of irradiation geometry. The Superflab conformed to the irregular shape of the diode and avoided large air gaps and the 8-cm thick block of solid water provided consistent back scatter of radiation. This same geometry was used for diode calibration. Photon and electron calibrations of diodes were carried out at a depth of 1 cm and an SSD of 99 cm and 100 cm for photons and electrons, respectively. Calibration doses for these geometries were calculated with the Eclipse treatment planning system.

For the measurement of angular dependence, a 10-cm diameter cylindrical phantom [8] was used. The diode is held at the end of a cylindrical hole bored through a cylindrical-rod made of solid water (Standard Imaging, Madison, WI) that fits snuggly into the cylindrical phantom. The rod is rotated to alter the incident angle of the beam with respect to the front surface of the diode. For all angular measurements, an incident angle of 0° corresponds to the central axis of the radiation beam being orthogonal to the front surface of the diode, which is indicated by a white dot shown in Fig. 1.

For dose-per-pulse dependence, a cylindrical ion chamber and diode were mounted adjacent to each other in a slab of solid water. The detectors are at a depth of 5 cm, irradiated with 6 MV X-rays at a dose rate of 600 MU/min. This assembly was mounted on a cart that could be

moved to various distances from the linear accelerator source position. This arrangement avoided large changes in scatter to the detectors while allowing for change in dose-per-pulse with distance. The linear accelerator gantry was at 90° with a 5 cm × 5 cm field size. It has been previously determined [9] that for this operation mode the pulse rate is 360 pulses/s and pulse duration is 3.9 μs. Additionally, the data point at the lowest dose-per-pulse of 0.0073 mGy/pulse was measured under the closed solid jaw of the accelerator at SSD 100 cm.

For diode response versus dose, the dose delivered was set with the accelerator console. For diode response versus dose rate, the rate was changed with the dose rate function on the linear accelerator console.

To determine the temperature of the diode it was operated as a thermistor, as previously published [10], by forward biasing the diode with a volt-ohm meter operated in the diode-measurement mode. In this mode, the voltage drop across the forward biased diode is measured by the digital volt-ohm meter, a model 77 (Fluke, Everett, WA). The dependence of diode voltage drop versus temperature was measured by allowing the diode to equilibrate to various temperatures in a small water bath whose temperature was measured with a laboratory thermometer. The water bath temperature was varied and the diode could be irradiated in this water bath.

3. Results

To test for changes in device sensitivity with accumulated dose, devices were irradiated with Ir-192 gammas and 15 MV X-rays. Fig. 3 shows how the sensitivity changes with up to 10.3 kGy of accumulated dose with Ir-192 gammas and then an additional 0.55 kGy with 15 MV X-rays. Overall there is about a 0.8% decrease in detector sensitivity, which has occurred by 5 kGy of accumulated dose. For accumulated dose above 5 kGy changes in sensitivity are smaller than measurement uncertainty. The 15 MV X-rays should create much greater damage than the Ir-192 gammas [11] and this is confirmed in Fig. 3, which shows a 2.2%/kGy decrease in sensitivity from 15 MV X-rays. Other characteristics that are presented in this section were measured on devices that had been given 10 kGy or more of accumulated dose.

Dependence of device sensitivity on dose-per-pulse was determined by comparing device sensitivity against that of an ion chamber. Dose-per-pulse was changed by measuring at different source to detector distance, SDD, from 70 to 340 cm. Additionally, the data point at the lowest dose-per-pulse of 0.0073 mGy/pulse was measured under closed solid jaws at SSD 100 cm. Fig. 4 shows that the device sensitivity decreases by 4% when the dose-per-pulse decreases by 59.4 fold from 0.436 to 0.0073 mGy/pulse.

The linearity of the diode response versus dose is shown in Fig. 5. As can be seen for a dose range of 10 cGy to 1000 cGy the diode response is 0.0977 nC/cGy with a coefficient of variation of 0.2%.

Dependence of sensitivity on dose rate was determined by comparing device sensitivity against that of an ion chamber. Dose rate was changed with the dose rate function on the linear accelerator console. Fig. 6 shows that the device sensitivity decreases by 0.3% when the dose rate is changed from 1400 MU/min to 10 MU/min.

The dependence of device sensitivity on the angle of incidence of radiation was measured with 6 MV X-rays and the device mounted on

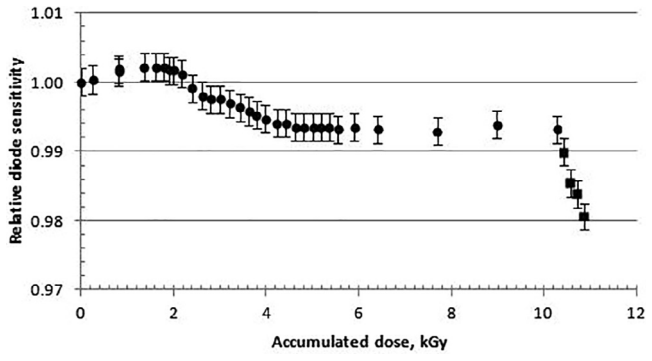

Fig. 3. Diode sensitivity versus accumulated dose. Dose was delivered with Ir-192 gamma rays except for the 4 data points from 10.3 to 10.85 kGy, which were delivered with 15 MV X-rays. Error bars indicate the response of three devices.

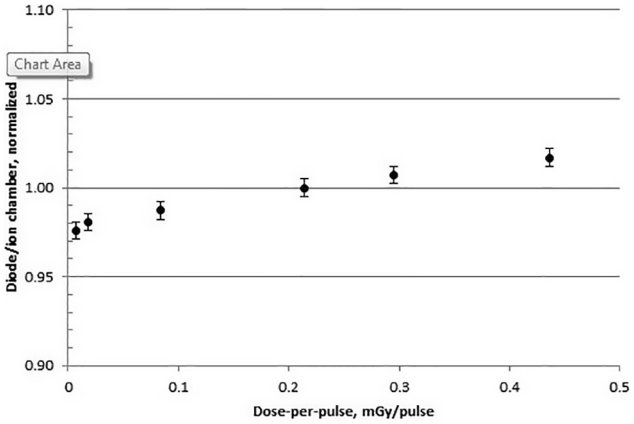

Fig. 4. Device sensitivity versus dose per pulse. The data are normalized at dose-per-pulse of 0.214 mGy/pulse, which is at SDD of 100 cm. These measurements were made with 6 MV X-rays and the detector was at depth 5 cm of solid water. Error bars indicate the standard deviation of three repeats of the measurement.

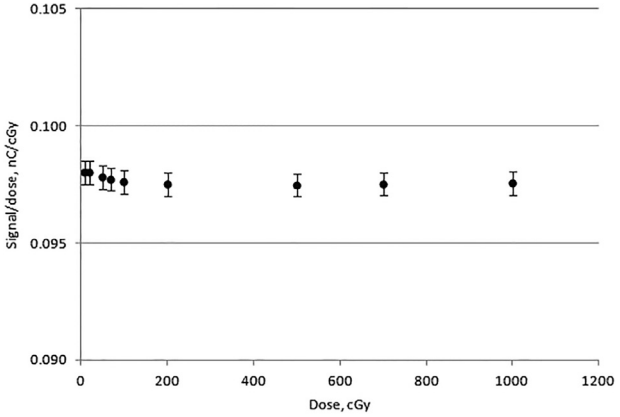

Fig. 5. Device response per unit dose versus delivered dose. These data are for 6 MV X-ray dose. Error bars indicate the standard deviation of three repeats of the measurement.

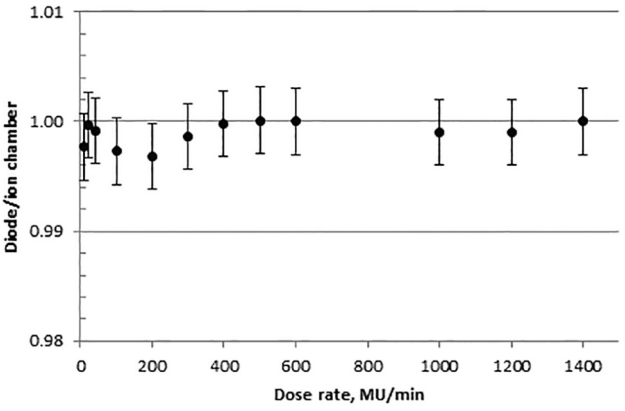

Fig. 6. Device sensitivity versus dose rate. The data are normalized at dose rate of 600 MU/min. Error bars indicate the standard deviation of three repeats of the measurement.

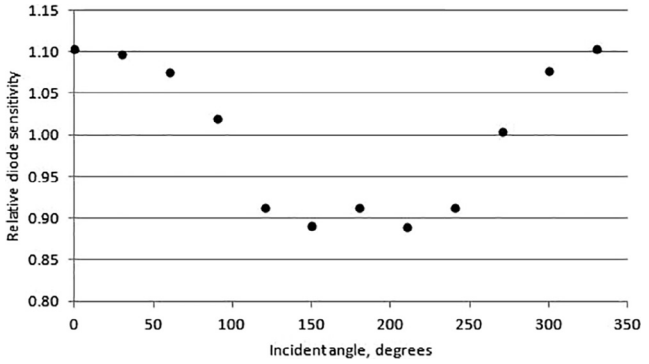

Fig. 7. Scanning and surface dose device sensitivity versus incident angle of radiation. The zero degree position corresponds to an incident angle perpendicular to the front surface of the diode.

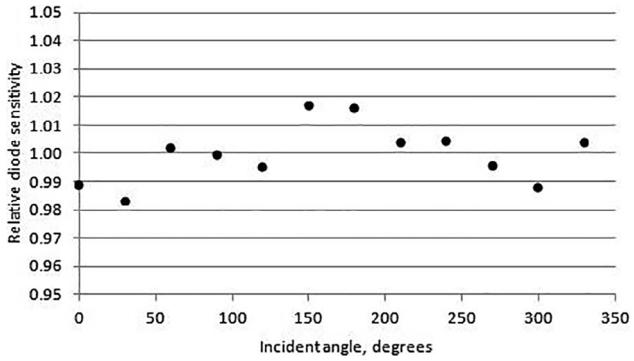

Fig. 8. IMRT device sensitivity versus incident angle of radiation. The zero degree position corresponds to an incident angle perpendicular to the front surface of the diode.

the central axis of a cylindrical phantom with 10 cm diameter. The scanning and surface dose devices have an angular dependence shown in Fig. 7 of ± 10%. The IMRT device is fabricated to reduce angular dependence, as shown in Fig. 2, and the response is shown in Fig. 8 and is found to be ± 1.7%.

To test the intrinsic build up of the surface detector, 6 MeV electron depth dose was measured and compared to Eclipse calculated depth dose. These data are shown in Fig. 9. As can be seen, the measured and calculated depth dose curves are in agreement when an intrinsic build up of 0.3 mm is included in the diode data.

The total output factor was calculated with Eclipse and measured

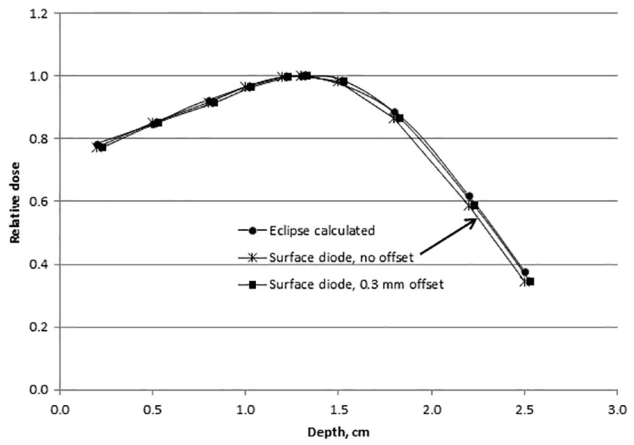

Fig. 9. Comparison of surface dose detector to Eclipse calculated electron dose versus depth. These measurements and calculations are made in water at SSD 100 cm and a 10 cm × 10 cm cone applicator.

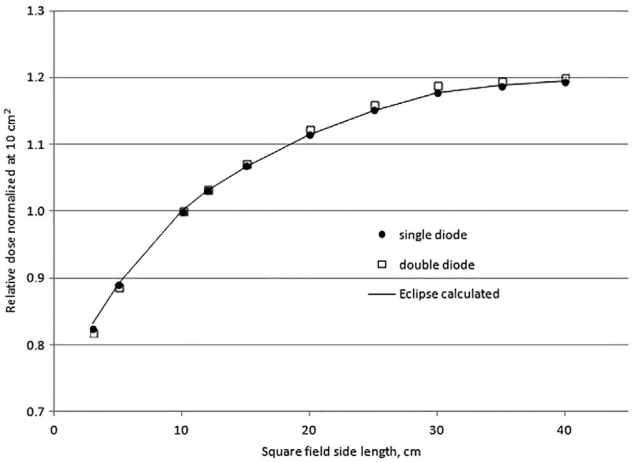

Fig. 10. Comparison of surface dose and IMRT type detector to Eclipse calculated total output factor in water versus square field size. These measurements and calculations are made in water at a depth of 10 cm and SSD of 90 cm.

with a surface dose and IMRT detector versus square field size. Data are for a 6 MV beam and are shown in Fig. 10 for 3 × 3 to 40 × 40 cm^2 fields. Over this range of field sizes, the surface and IMRT type PIN diodes are in agreement with the Eclipse calculated change in dose.

The total output factor for different stereotactic radiosurgery cones is shown in Fig. 11. As can be seen the scanning diode gives results that are very close to the Edge detector as well as the Varian published data [12].

Diode sensitivity is known to change with temperature. Fig. 12 shows this temperature effect for two PIN diodes. The temperature coefficient is 0.027%/°C for these two diodes. When using the diode as a thermistor, the typical temperature change when placing the diode on the patient skin was determined to be 8–12 °C.

4. Discussion

All of the measured characteristics of these diodes used in this work are summarized in Table 1 and are compared to some other commercially available PN junction diodes.

Diodes change sensitivity with accumulated dose [10,11,13–15]. Fig. 3 shows how the sensitivity changes with up to 10.3 kGy of accumulated dose with Ir-192 gammas and then an additional 0.55 kGy with 15 MV X-rays. The sensitivity change is 2.2%/kGy of 15 MV X-rays. If

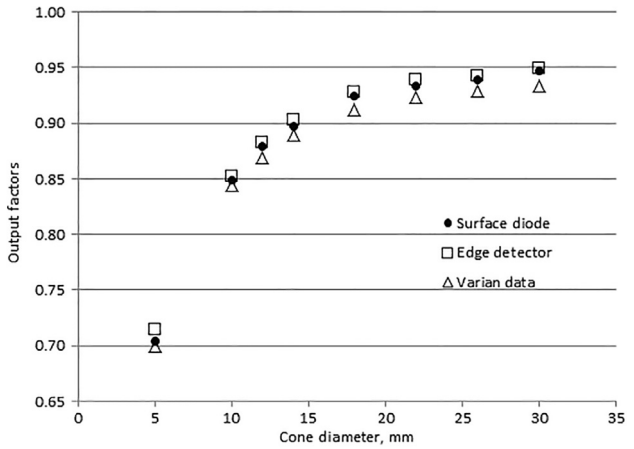

Fig. 11. Measurement of relative output of cone stereotactic radiosurgery cone collimators. These measurements are for a 6 MV stereotactic beam and are made in water at a depth of 1.5 cm and SSD of 98.5 cm. The output factors are normalized to an open 10 cm × 10 cm field.

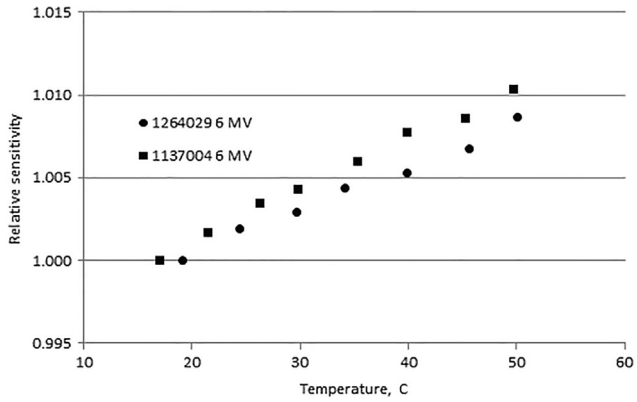

Fig. 12. Variation in diode sensitivity with a change in temperature.

one accepts a 0.5% change in sensitivity due to accumulated dose, then this change would occur with an accumulated dose of 227 Gy of 15 MV X-rays. The typical measured dose on a patient is 2 Gy, so 114 measurements could be made with a 0.5% change from a calibration point due to changes in sensitivity from accumulated dose. Making 1 patient measurement per day, the diode would need calibration after 23 weeks of use. In our clinic we update the diode calibration on a monthly basis.

The change in diode sensitivity versus a 59 fold change in dose-per-pulse is 3.6%, which is better than most other diodes reported in the literature [1,2,9] and Table 1. It is advisable that the diodes be calibrated at a dose-per-pulse level that will be in use for patient measurements to avoid the need for a dose-per-pulse correction factor.

The change in diode sensitivity versus a 140 fold change in dose rate is −0.3%. Other diodes have been found [9,16,17] to have as large as a 2% change in sensitivity with dose rate. Due to the small change on the diodes presented in this work, no correction is made for sensitivity changes due to dose rate.

PN junction diodes are known [1,2] to have a linear response to dose over many orders of magnitude. This linearity is confirmed for these PIN diodes in Fig. 5 from 10 to 1000 cGy. Interestingly, thermoluminescent [18] and Al_2O_3:C optical-luminescent dosimeters [19] have a limited range of linear response to dose, which becomes supralinear above 100 cGy. Complicated calibration procedures to accommodate non-linear response to dose are not needed for these diodes.

Earlier results [1,2,8] indicated that PN type diodes changed their sensitivity with field size. Also, a double PN junction diode [5] over responded to low energy scattered photons in large fields. The data in Fig. 10 indicate that this is not a problem with the PIN surface and IMRT diode devices described here and no corrections for field size are needed.

The typical temperature change when placing a diode on a patient is 8–12 °C. With a temperature coefficient of 0.027%/°C found in this work for the PIN diode this would result in a sensitivity change of < 0.3%. In our clinic we have chosen to not make any temperature corrections for patient measurements.

The traditional method for using PN junction diodes [1,2] has been to place the diode, with a build up cap that is appropriate for the beam energy, on the surface of the patient. Correction factors are applied to the diode reading and the dose at the depth of maximum dose is calculated. This measured dose is then compared to the dose prescribed at the depth of maximum dose.

Here a non-traditional method for using diodes is employed. Our physicians are interested in confirming the dose at the patient's skin surface and not at the depth of maximum dose. For static beams, the surface type diode is used, which has an intrinsic buildup of 0.3 mm water equivalent, and no build up cap is added to the diode. Measurements are made with the beam orthogonal to the front surface of the diode and the diode is positioned at the surface of the patient. Measurements are made with the diode being at the skin surface or

Table 1
Comparison of various PN junction diodes. Data for diodes is from manufacturer literature or data in this paper.

Diode	Active area, mm	Distance of active area from the top surface, mm	Water equivalent depth, mm	Sensitivity decrease %/kGy	Sensitivity dose-rate dependence for 1400 to 100 cGy/min,%	Sensitivity increase, dose per pulse increased by 59 fold,%	Sensitivity, nC/Gy	Waterproof
SunNuclear EDGE	0.8 × 0.8	0.3	0.5	Not given	Not given	Not given	Not given	yes
Iba RAZOR	0.6 (diameter)	0.8	Not given	< 2	Not given	Not given	4.1	yes
PTW E-diode	1.1 (diameter)	1.33	0.7	1	Not given	Not given	9	yes
PTW SRS diode	0.56 (diameter)	1.31	Not given	0.8	Not given	Not given	175	yes
Exradin D1V-D1H	0.56 (diameter)	Not given	0.8	1	−0.1	4.5	5	yes
New scanning device	0.5 × 0.5	0.4	0.3	2.2(15 MV X-rays)	−0.3	3.6	**10**	**yes**
New surface dose device, PIN diode	0.5 × 0.5	0.4	0.3	2.2(15 MV X-rays)	−0.3	3.6	**10**	**no**
New IMRT device, dual PIN diodes	0.5 × 0.5	0.8	1.4	2.2(15 MV X-rays)	−0.3	3.6	**20**	**no**

under the bolus if one is used on the patient. As explained earlier in this discussion, no correction factors are required and the diode measurement is converted to dose with the dose calibration factor. This measured dose is compared to the treatment planning system calculation of dose at the measuring point of the diode.

When beams are not orthogonal to the surface of the diode or if the treatment delivery is rotational IMRT, then the IMRT type diode is used, which avoids corrections for changes in sensitivity due to angle of incidence of the beam. No correction is made for the ± 1.7% angular dependence shown in Fig. 8 since it will average out to 0% when radiation impinges from all angles with rotational IMRT delivery method. Measurements are made with the diode being at the skin surface or under the bolus. The diode dose measurement is compared to the treatment planning system calculation of dose to the measuring point of the diode.

We have been using these surface and IMRT type diodes and methods described in this work for a couple of years in our clinic. We have an action level of ± 3% difference between diode measurements and treatment plan calculations. This is achieved most of the time when a 0.5 or 1 cm thick bolus is used. When there is no bolus, then the calculated dose varies by 20–30% with a change in depth of the calculation point of 1 mm. Again, we find our diode surface dose measurements are routinely within this large range of calculated dose. When comparisons of measured and calculated dose are outside the action level, then measurements are repeated and the treatment plan is thoroughly checked for errors in order to resolve the discrepancy.

The scanning type PIN diode is water proofed and can be mounted on a water tank scanning arm. Because of this diode's 0.5 mm size and 0.3 mm build up, see Table 1, it is ideal for profile and depth dose scanning of beams smaller than 2 cm in cross section. As shown in Fig. 10, results with this diode compare well with other trusted measures of cone collimators with diameters as small as 5 mm.

5. Conclusions

Clinically useable dose detection devices can be fabricated from PIN type photodiodes. These diodes can be used to measure dose at the surface of a patient or under bolus if one is used on the patient. Correction factors are unnecessary if a surface or IMRT type device is chosen for use with static orthogonal or rotational IMRT delivered beams, respectively.

Conflict of interest

The author has no conflict of interest.

Acknowledgements

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