Accuracy of Dose Delivery Using Diodes in External Beam Radiotherapy (EBRT)

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Summary

Accurate prescribed dose delivery is very important for each patient undergoing radiation treatment in order to avoid over or under dosing. One way of verifying the delivered dose is through in-vivo dosimetry i.e. measuring the patient dose during treatment. In in-vivo dosimetry silicon diodes are used to measure the entrance dose in radiotherapy. Two diode detectors from IBA Dosimetry used in this study were tested, calibrated and corrected in order to be accepted for treatment verification. The corrections of varying field sizes, source-to-surface distances, temperature and angle of incidence have been reported. This work investigates the feasibility of performing routine quality control protocol using in vivo dosimetry for two dimensional (2D) and three dimensional conformal radiotherapy (3D-CRT) treatments. For each radiation field used in treatment a measured dose on the patient skin and calculated dose from treatment planning system were compared using a 5 % tolerance. The maximum entrance dose deviation was observed to be 4.1 % for all the considered 10 cases. It can detect potential errors in accurate dose delivery to the patient.

Keywords: In-vivo dosimetry, Diode, Radiotherapy

Introduction

The purpose of radiotherapy treatment is to deliver radiation dose to various malignant or non-malignant targets efficiently, accurately and safely. Normally the quality assurance program have been introduced to verify the accurate performance of all the components of radiotherapy together with treatment planning, imaging modality to treatment execution. As the number of radiation incidences reported due to human errors, separate patient dose verification (in-vivo) is required during the actual treatment delivery in external beam radiotherapy (EBRT).[1-4]

In early 1980s the silicon diodes were first introduced for in-vivo dosimetry with proper buildup cap based on type and beam quality. Silicon diode offers many advantages like instantaneous read out, high sensitivity, simple instrumentation and robustness. Though, silicon diode with buildup cap response to various influencing factors that need to be corrected for field size, source to surface distance, temperature and angle of incidence. The selection of proper build up cap is important for entrance dose measurements as the dosimeter below significantly attenuates the dose by the buildup cap material. The entrance dose measurement often performed for limited fractions.

Nowadays p-type diode is preferred over n-type because of accumulated dose over its life time. Over a period of time diode sensitivity decreases due to accumulated dose and is more evident in n-type diodes compared to p-type diodes.[5] During exposure the rate of electron hole pairs production increases with increase in dose rate. At higher dose rate, large charge carriers will escape recombination process compared to lower dose rate. Hence diode sensitivity depends on dose rate.[6] The p-type diodes have very limited dose rate dependence with a higher doping level.[7-9] The diode detector after significant use accumulates dose may show increase in response with increase in dose rate.[10]

In this paper, we will be discussing how we established in-vivo dosimetry technique using p-type diodes by measuring various correction factors at our center.

Material and Method

The measurements were carried out on a Elekta Synergy linac, providing a 6 MV photon beam. Two identical diodes from IBA Dosimetry [EDP-10/5143 and EDP-10/5144 detectors (p-type silicon diodes)] with hemispherical buildup cap with DPD-12 (emX) electrometer were tested for entrance dose measurements. The diode detectors for in-vivo dosimetry are available commercially in two forms, cylindrical and flat design with buildup cap of different thickness and nature. The detectors used in our study offers flat design for easy placement on skin, diameter of active area is 1.6 mm, sensitivity as 100 nC/Gy and sensitivity variation with temperature (SVWT) is 0.25%/^{o}C. High density materials like stainless steel and epoxy (thickness 1 g/cm^2) are used in order to reduce the physical thickness of the build-up cap. Buildup cap is accountable for optimized low field perturbation, minimized field size and directional dependencies, and low temperature dependence. Before using the diode in actual dosimetry, one should always measure the diode response as a function of energy in which it is used. These diodes were calibrated against an ion chamber (FC65) from Scanditronix Wellhofer in a 6 MV photon beam.

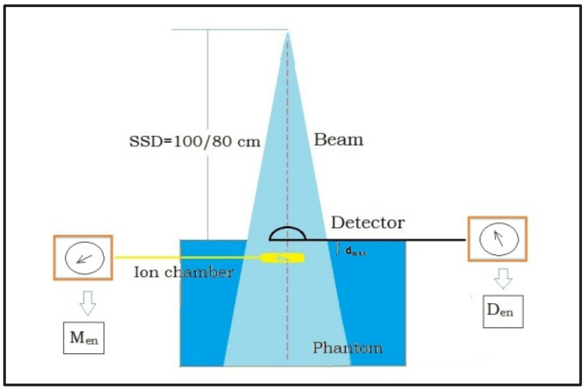

Figure 1: Schematic diagram for entrance dose calibration

As a reference condition, the diode was fixed on a solid water phantom having dimensions 30 x 40 x 40 cm^3. The diode was set at a distance of 100 cm from linac focus. The field size was set to 10 x 10 cm^2 at 0° gantry angle. The ion chamber was kept and exposed below the phantom surface at depth of maximum dose as shown in Figure 1. Apart from calibration factor, other correction factors for various field sizes, source to surface distances, temperature and gantry angles were determined.[11,12] The diode was connected to devoted channel on DPD-12 (emX) electrometer from Scanditronix Wellhofer. The electrometer was then connected to computer having DPD12-pc software. Before starting the actual measurement, dark current drift and any offset were measured and corrected.

Diode Calibration:

The diode calibration factor (F) is the ratio of adsorbed dose measured with the ion chamber D_{en}, and the meter reading of the diode M_{en} under reference conditions (Figure 1).

$$F = \frac{D_{en}}{M_{en}}$$

Field size dependence:

The diode was kept on a solid phantom at SSD=100 cm. For a number of square fields ranging from 3 x 3 cm^2 to 40 x 40 cm^2 meter-readings (response) were measured. The field size dependence correction factor (F_{FS}) was calculated using the following formula.

$$F_{FS} = \frac{\dfrac{D_{en}(FS)}{M_{en}(FS)}}{\dfrac{D_{en}}{M_{en}}}$$

Where M_{en} is the entrance dosimeter reading for reference field size. D_{en} is the dose at depth of maximum (1.5 cm) for reference field size. The dose variation is not measured directly but calculated from previously measured output ratios.

SSD dependence:

The diode was kept on the surface of phantom. The SSD correction factor F_{SSD} was measured at different SSD's covering the range from 80 cm to 120 cm.

The SSD dependence correction factor is given by

$$F_{SSD} = \frac{\dfrac{D_{en}(SSD)}{M_{en}(SSD)}}{\dfrac{D_{en}}{M_{en}}}$$

F_{SSD} accounts for dose rate dependency of diode, insufficient buildup and effect of source to diode distance and source to chamber distance.

Angular dependence:

The angular dependency of diode is because of diode buildup and phantom scatter. F_{gantry} is the ratio of reading at beam axis during calibration to central beam axis. In axial dependency of beam axis rotates in plane perpendicular to the cable and during tilt the beam axis rotates in the plane of the cable.

The angular dependence correction factor is given by

$$F_{gantry} = \frac{R\ (\Theta=0)}{R\ (\Theta)}$$

Temperature correction factor:

The temperature correction factor is given by

$$F_{temp} = \frac{R\ (Tcal)}{R(T)}$$

Where T is the room temperature and Tcal is the diode calibration temperature

If F_{temp} is less than 0.4 % per °C then temperature correction factor is not required.

After determining correction factors to account for these effects, the measured signal can be converted to measured dose using the following formula.

Measured dose (Gy) = Diode reading x F x F_{FS} x F_{SSD} x F_{gantry} x F_{temp}

In-vivo measurements:

The diode was used in regular measurements to verify the entrance dose with treatment planning system (TPS) calculated in patients treated with two fields (breast treatment), three fields (oesophagaus) and box technique (cervix) with the isocenter at the surface of the patient.

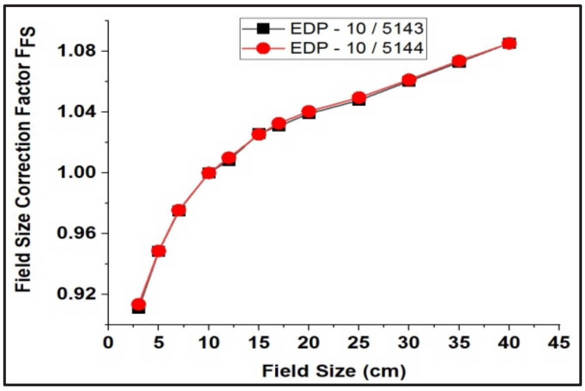

Figure 2 : Comparison of field size correction factors for two diode detectors as a function of field size

Table 1: Field size correction factors as a function field size

Field Size (cm2)	EDP-10/ 5143	EDP-10/ 5144
3 x 3	0.9108	0.9136
5 x 5	0.9485	0.9487
7 x 7	0.9749	0.9754
10 x 10	1.0000	1.0000
12 x 12	1.0083	1.0100
15 x 15	1.0259	1.0254
17 x 17	1.0310	1.0327
20 x 20	1.0390	1.0405
25 x 25	1.0479	1.0496
30 x 30	1.0604	1.0614
35 x 35	1.0729	1.0738
40 x 40	1.0854	1.0853

In clinical routine the patient was set-up in the proper treatment position as planned on treatment planning system. The diode was fixed on patient skin at the central beam axis. The reading was then multiplied by calibration factor and other suitable correction factors to get the measured dose. The measured dose and treatment planning system calculated dose was compared for deviation.

A thorough investigation of treatment plan need to be performed provided the deviation exceeds the acceptance level of 5%. At next treatment session, the in-vivo dosimetry was then repeated by carefully measuring SSD and verifying the correct position of dosimeter. If we repeat the procedure for more sessions, skin dose increases due to beam attenuation by diode and reduction in dose at depths.[13]

Results

Field Size Correction factor:

Figure 2 shows the comparison of field size correction factors for various square field sizes. For 6MV photons, the field size correction factors for two identical diodes are found to be comparable for all considered field sizes. Table 1 lists the field size correction factors for various field sizes for two diodes. For the 6 MV photon beam, the F_{FS} for the EDP-10/5143 diode are 0.9108 and 1.0854 for the field size of 3 x 3 cm^2 and 40 x 40 cm^2 respectively, while the F_{FS} for the EDP-10/5144 diode are 0.9136 and 1.0853 for the field size of 3 x 3 cm^2 and 40 x 40 cm^2 respectively.

Table 2: SSD correction factors for two types of identical diode detectors measured for field size 10 x 10 cm^2

SSD (cm)	FSSD	
	EDP -10 / 5143	EDP-10 / 5144
80	0.95013	0.95438
81	0.95327	0.9627
82	0.95575	0.95993
83	0.95912	0.96776
84	0.9617	0.96578
85	0.96514	0.97357
86	0.96661	0.96983
87	0.96956	0.97843
88	0.97108	0.97409
89	0.97487	0.98265
90	0.97673	0.9803
91	0.98012	0.98831
92	0.98138	0.98388
93	0.98331	0.99188
94	0.98591	0.98766
95	0.98922	0.99733
96	0.99143	0.99326
97	0.99429	1.00184
98	0.99639	0.99783
99	0.99912	1.00604
100	0.99999	0.99999
101	1.00347	1.0107
102	1.00656	1.01078
103	1.00813	1.01678
104	1.00916	1.01466
105	1.01183	1.01972
106	1.01452	1.01969
107	1.01778	1.02485
108	1.01865	1.02464
109	1.02243	1.0298
110	1.02432	1.03121
111	1.02671	1.03439
112	1.02704	1.03354
113	1.03043	1.03841
114	1.03032	1.03708
115	1.03473	1.03131
116	1.03691	1.04396
117	1.03956	1.04676
118	1.04124	1.04784
119	1.04188	1.04114
120	1.04366	1.05125

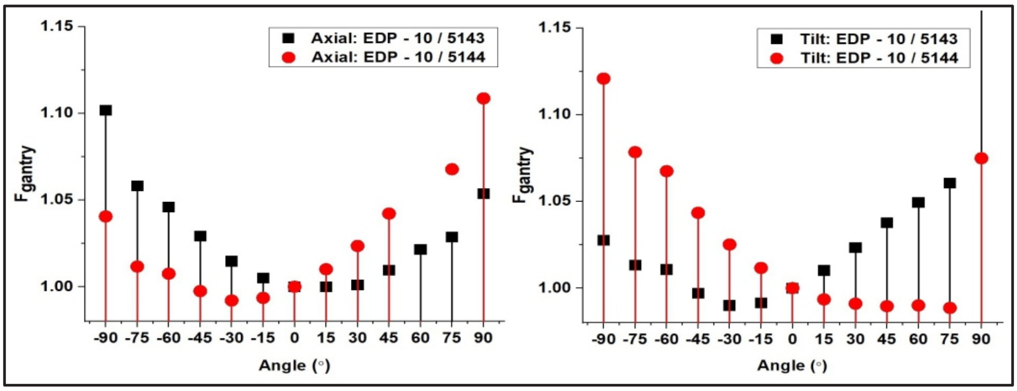

Figure 3: Angular dependence of two diode detectors

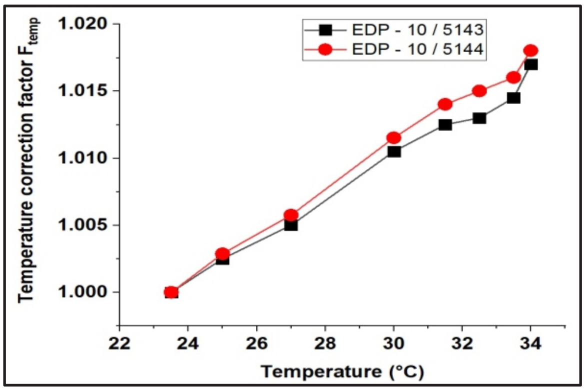

Figure 4: Temperature dependence for two diode detectors

SSD Correction factor:

Table 2 lists the SSD correction factors for change in SSD for two identical silicon diode detectors. For 6-MV photons, the F_{SSD} for the EDP-10/5143 diode are 0.95013 at SSD of 80 cm and 1.04366 at SSD of 120 cm, while F_{SSD} for the EDP-10/5144 diode are 0.95438 at SSD of 80 cm and 1.05125 at SSD of 120 cm.

Angular dependence correction factor:

Figure 3 shows the dependence of diode for various angles. Directional response of diode from angles -45°to +45°for both axial and tilt is less than 3 %. Table 3 provides the angular dependence factors.

Temperature correction factor:

The diode response due to change in temperature is shown in figure 4. The diode response is linearly increasing with temperature.

In-vivo dosimetry was conducted in 10 patients and the maximum entrance dose deviation was observed to be 4.1 % for all the considered cases.

Discussion

The discrepancies between the planned and measured dose should be analyzed and reported to the radiation oncologist and radiation technologist who treat the patient.[14] After measuring the entrance dose (D_m), the expected dose (D_c) computed from treatment planning system (TPS) was compared using the formula $(D_m-D_c) \times 100 / D_c$.[15,16]

In in-vivo dosimetry the accepted deviation level considered to be 5%. For the results to be effective and more realistic, we used 5% and 10% two sets of tolerances.[17] The radiation technologist will check for setup error, SSD, any treatment parameter etc, if the deviation is greater than 5% but less than 10%. The physicist will observe for the reading consistency during the next treatment and investigate the cause of deviation. As the reported error for wedged field is more than 8%, we have not included the wedged fields in our study.[17] Some observations are comparable to the literature.[16,18-20]

The diode sensitivity decreases as function of angle between the symmetry axis of diode and the beam axis. The angular dependence of diode may be of importance when oblique radiation beams are used. It has been observed for the first time with no supporting document that if the angle is more than 45°, the dose comparison will show more than 5% deviation. Hence, oblique radiation with more than 45° angle is not included in our study.

Table 3: Angular dependence factors

Angle (°)	Axial	Tilt		
	EDP-10 / 5143	EDP-10 / 5144	EDP-10 / 5143	EDP-10 / 5144
-90	1.1018	1.0406	1.0275	1.1208
-75	1.0581	1.0116	1.0133	1.0784
-60	1.0460	1.0076	1.0107	1.0674
-45	1.0293	0.9975	0.9970	1.0434
-30	1.0147	0.9921	0.9900	1.0252
-15	1.0050	0.9935	0.9915	1.0117
0	1.0000	1.0000	1.0000	1.0000
15	1.0000	1.0101	1.0102	0.9935
30	1.0010	1.0235	1.0233	0.9911
45	1.0096	1.0422	1.0378	0.9896
60	1.0214	0.0021	1.0493	0.9901
75	1.0288	1.0678	1.0606	0.9886
90	1.0537	1.1086	1.4406	1.0749

It has been observed that the temperature can affect the diode response.[10,21-23] The sensitivity of p-type diodes increases with increase in temperature after an accumulated dose.[22] But the temperature correction factor variation with temperature is linear in our study. As the treatment time is short. The diode kept on the patient skin can not reach the thermal equilibrium. Hence, in in-vivo dosimetry, the temperature correction factors measured at room temperature are used.

Conclusion

In-vivo entrance dose measurements have been proved to be a very useful tool for the verification of dose delivered to a given patient. During the patient treatment serious errors like incorrect selection of daily dose, selection of wrong beam energy, error in wedge selection, setup errors and changes in machine output can be rectified during the subsequent fractions.

Accurate absorbed dose can be obtained from diode signal by applying calibration and proper correction factors. There is a need to implement it on phantom first and verify whether diode system provide accurate dose.

The in-vivo dosimetry results using diodes are available in real time. In-vivo dosimetry is a useful technique for quality control in radiotherapy and increasing treatment accuracy.

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