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ORIGINAL PAPER

Dosimetric characteristics of active solid state detectors in a 60 MeV proton radiotherapy beam

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Abstract. Several solid state detectors, such as dosimetric diodes, MOSFET detectors or diamond detectors are used for quality control of radiotherapy beams. The goal of this work was to determine dosimetric properties of the PTW diamond detector (DD) and the PTW silicon diode in the 60 MeV therapeutic beam (practical range in water R_p = 29.17 mm) located at the Institute of Nuclear Physics, Polish Academy of Sciences (IFJ PAN, Kraków). A PTW Markus ionization chamber was used as a reference device. The empirical correction factor for diamond detector, $k^{DD}(R_{residual})$, introduced in the Technical Report Series of IAEA, TRS-398 [9] as a function of beam quality, $R_{residual}$, was found to decrease from 1.12 for $R_{residual}$ = 1.5 mm to 1.04 for $R_{residual}$ = 26 mm. The reproducibility of response of DD and PTW diodes in the proton filed did not exceed 0.11%. Our results show that diamond detectors and dosimetric diodes are useful tools for quality assurance (QA) of therapeutic proton beam, but each type of detectors has specific properties which should be taken into account when choosing particular application.

Key words: diamond dosimeter • diode Si • dosimetry • proton beam

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Introduction

The solid state active detectors, widely used for conventional radiotherapy, such as diamond detectors (DD) and diodes may be attractive tools also in dosimetry of therapeutic proton beams. Diodes are broadly applied in oncology clinics and routinely used for *in-vivo* dosimetry [1, 8]. These real-time detectors are rugged, relatively inexpensive and provide on-line readings, what allows for quick correction of errors encountered during dose delivery. Since a long time, diamonds have been considered as a promising tool for clinical dosimetry, but the dosimeters based on natural diamonds are expensive and hardly available. Synthetic diamond (CVD) is potentially one of the most interesting materials in the field of detectors applicable in dosimetry. Diamond detectors, working as solid state ionization chambers, are considered to be tissue equivalent which is an advantage for clinical dosimetry. Commercially available DD offered by PTW are currently produced from natural diamonds. It was reported that the sensitivity of PTW DD was independent of the photon beam quality, and in the electron beam energy in the therapeutic range [5]. Similarly, the CVD diamond, tested in photon, electron and proton beams, has sufficient sensitivity to be used as a dosimeter [3]. One of the problem is the effect of dose rate dependence which needs to be corrected, e.g. using Fowler's model, which introduce the Δ parameter, describing the deviation from the linearity [3, 5]. In proton beam dosimetry the knowledge of linear energy transfer (LET) response of DD is obligatory for their use in measurements of depth dose distribution. For natural DD it was reported that a sensitivity decrease with LET, underestimating the dose by about 18% at the Bragg peak, by 7% at the centre and by 17% at the distal end of the spread-out Bragg peak (SOBP) region [13, 16].

Second type of solid state detector, considered for relative proton dosimetry, is a silicon diode. Dosimetric diodes are excellent charged-particle detectors, because they are stable, show limited LET dependence, and can be made small in size. The key structure in the silicon diodes used for dosimetry is the pn junction, which is non-tissue equivalent and very thin (less than 0.01 mm) [1] but the entire structure is thicker, due to the casing with the buildup layer. The most popular silicon diodes used in dosimetry and described are n-type and p-type [6, 12, 13]. Some authors showed the sensitivity decrease of a diode used in a proton beam even up to 40 percent in 5 years, possibly to radiation damage [10]. In contrast, the diamond detector is considered to be highly resistant to radiation damage [5]. The data shows that the properties of the solid state detectors should be evaluated for each specific detector used [2, 5, 12].

The goal of this work was to determine dosimetric properties of a PTW natural diamond detector and a PTW silicon diode, in particular dose and dose rate dependence, energy response and repeatability of the detector in ^{60}Co gamma rays and the 60 MeV proton beam. One of the issue in using diamond detectors and diodes in proton beams is their LET dependence. The empirical correction factor for diamond detector, $k^{DD}(R_{residual})$ was introduced in the TRS-398 as a function of beam quality. In this paper the $k^{DD}(R_{residual})$ values was determined, by adapting the method described by Fidanzio *et al*. [4].

Materials and methods

The proton beam

The proton beam forming and monitoring system has been installed on the optical line located inside a proton radiotherapy room [11, 17].

The proton beam parameters have been controlled by a set of ionization chambers connected to eight dedicated electrometers. The beam forming and monitoring system has been controlled by a computer with a real time steering system used to set the working parameters and control the proton beam.

Measurements in ^{60}Co gamma radiation

The Theratron 780E unit, containing a ^{60}Co radioactive source is used at the IFJ PAN for cross-calibration according to the protocol TRS-398. The activity of ^{60}Co source is currently 68.0 TBq, which is enough to obtain a 0.54 Gy/min dose rate at the isocenter of this unit (source-to-chamber distance SCD = 80 cm).

The repeatability of the diamond detector and the silicon diode connected to the PTW UNIDOS electrometer has been tested under the ^{60}Co gamma field. The measurements were performed in the water phantom in the reference geometry (SCD = 80 cm, d = 5 cm depth in water).

Measurements in 60 MeV proton beam

Measurements were performed with a 60 MeV proton beam produced by isochronous cyclotron AIC-144 at the IFJ PAN. Uniform lateral distribution at the isocenter in a therapy room is realized by passive scattering with a single scattering, 25 μm thick tantalum foil [11, 17]. The scattering system is located about 12 m from isocenter of the facility. The beam forming system in the therapy room consists of a proton beam range shifter, range modulator and a set of collimators. The range shifter consists of a PMMA wheel setting up by a step motor for changing PMMA thickness with a precision better than 0.01 mm. The energy modulator is used to produce SOBP.

The measurements of proton beam depth dose distribution were conducted in an in-house developed 10 × 10 × 10 cm^3 PMMA water phantom with a thin entrance window. In order to determine the proton range in water it is necessary to know the water equivalent thickness of the water phantom window, the thickness of the detector and the dimensions of detector sensitive area (Table 1). The dedicated holders for the Markus IC, diamond detector, and silicon diode enable positioning the detectors in the central axis of the beam and moving detectors in X, Y, Z coordinates. Each detector was positioned close to the entrance window of the water

Table 1. Important parameters of tested detectors. The signals of DD and diode were tested according to the PTW Markus ionization chamber

	Detector type		
	Markus chamber PTW 23343	Diamond PTW	Diode PTW
Active volume	Radius = 2.65 mm Thickness = 2 mm	Radius = 2.18 mm Thickness = 0.24 mm	Radius = 0.56 mm Thickness = 2.5 μm
Detector material	Air (under air-pressure)	Natural monocrystal of diamond	Silicon
Energy for pair production	34 eV	12.8 eV	< 1 eV
Water equivalent thickness (μm)	2	750	4.6
Entrance window water equivalent (mm)	1.06	1.15	0.7
High voltage (V)	300	100	0
Dark current (Pa)	< 0.16	0.7	15
Predose (Gy)	–	5–15 Gy	about 1 Gy

phantom and moved with a 0.2 mm step. Detectors were stopped in each measuring point for 10 s for collecting the dose, which corresponded with the dose of about 5 Gy at the entrance part of the Bragg peak.

The repeatability tests were performed for the same value of monitor units, in the position of $d = 5$ mm water equivalent depth. The dose dependence of the signal was determined in the range from 1 to 30 Gy. The typical value of dose delivered to the tumor in ocular radiotherapy is 13.64 Gy.

The same procedure was repeated for the dose rate dependence measurement of the diamond detector. The effect of dose rate on detector response was measured in the dose rate range from 6 to 40 Gy/min. Dose rate values were determined with Markus ionization chamber and the dose rate variation was obtained by changing the proton current. Fowler's model, predicts a dose rate $\dot{D}$ dependence of the detector current, I as:

$$I = c \cdot \dot{D}^{\Delta} \tag{1}$$

where c is a constant and Δ is an exponential parameter which describes the deviation from the linearity [4, 12].

The TRS-398 IAEA code of practice [9] provides recommendations for ionization chamber dosimetry to be used in dosimetry of the proton beams. They are based on calibrations in a Co-60 beam in terms of absorbed dose to water. The protocol mentioned above provides the calculated values for chamber specific factors, k_Q, as a function of beam quality index, R_{residual} for cylindrical and plane-parallel ionization chambers. According to the TRS-398, the residual range, R_{residual}, is chosen as the beam quality index. The residual range (in g/cm^2) at a measurement depth d is defined as

$$R_{\mathrm{residual}} = R_p - d \tag{2}$$

where d is the depth of measurement and R_p is the practical range (both expressed in g/cm^2), which is defined as the depth at which the absorbed dose beyond the Bragg peak or SOBP falls to 10% of its maximum value.

The empirical values of k_Q for diamond detector were determined by applying the Markus ionization chamber as a reference dosimeter, and adapting the method which takes into account the LET dependence of this detector [4]. The calibration factor, N_w^{DD} of the DD was obtained by using a non-modulated proton beam.

$$N_w^{\mathrm{DD}} = \frac{D_{w,Q}(d)}{M(d)} \tag{3}$$

where $D_{w,Q}(d)$ is dose to water, as measured by the Markus ionization chamber and $M(d)$ is the DD signal, corrected for the dose rate dependence [4, 13], obtained in the same irradiation conditions at the depth $d = 5$ mm.

The DD calibration factor $k^{\mathrm{DD}}(R_{\mathrm{residual}})$ changes with water depth as a function of proton beam quality $R_{\mathrm{residuals}}$ (formula (4)).

$$k^{\mathrm{DD}}(R_{res}) = \frac{D_{w,Q}(R_{\mathrm{residual}})}{N_w^{\mathrm{DD}} \cdot M(R_{\mathrm{residual}})} \tag{4}$$

where $D_{w,Q}(R_{\mathrm{residual}})$ is the dose to water at R_{residual}, $M(R_{\mathrm{residual}})$ is DD signal at R_{residual}.

Tested detectors

The PTW diamond detector (DD) consists of a natural diamond crystal, nearly tissue equivalent, with contacts of the radius 2.18 mm and a thickness of 0.24 mm (Table 1). The water equivalent of entrance window is equal to 1.15 mm. A bias voltage of 100 V was applied and the DD was connected to a PTW Unidos electrometer. In order to stabilize the detector signal, a preirradiation dose of about 10 Gy was applied each time before the DD use.

Second detector tested was the waterproof p-type Si diode detector PTW type 60012, with an extremely small sensitive volume shaped as a disc with an area of 1 mm^2 and a thickness of only 2.5 μm (Table 1). The size makes it possible to use the dosimetry diodes in small beams and to perform data acquisition with a very good spatial resolution. Since the diodes are waterproof, they can be used in water phantoms without additional protective sleeves.

Results

The detector response normalized to the charge, measured by transmission ionization chamber in function of the depth, d, is presented in Fig. 1. The response of the Markus IC, the diamond detector and the Si diode were normalized to the depth $d = 5$ mm for the non--modulated proton beam. The relative signal of DD is 2% lower and the relative signal of diode is about 13% higher than the Markus chamber.

The beam practical range $R_p = 29.17$ mm and the depth of the Bragg peak maximum dose $R_{\max} = 28.16$ mm were measured using the Markus IC. The range R_p and Bragg peak maximum $R_{\max}$ were determined using DD. For detectors tested, the values of $R_{\max}$ are comparable (within 0.1 mm below the uncertainty level) with the values obtained with the use of Markus chamber. Therefore, the uncertainty of the detector effective points of measurements is masked within the positioning uncertainties. Full width at half maximum (FWHM) of the Bragg peak measured by Markus and DD are consistent up to 0.05 mm, FWHM measured

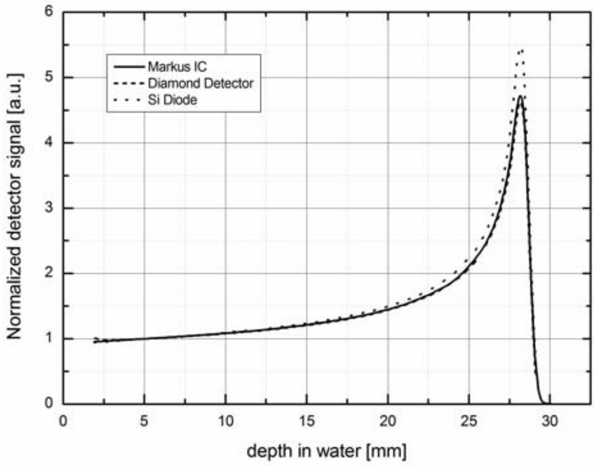

Fig. 1. Depth dose curves of proton beam in water measured with Markus IC (solid line), diamond detector (dashed line), Si diode (dot line).

Table 2. The results of signal repeatability measurements

	Farmer IC	Diamond detector	Silicon diode	Markus IC	Diamond detector	Silicon diode
	^{60}Co gamma field			Proton beam		
Average dose	2.00 Gy	2.00 Gy	2.01 Gy	13.64 Gy	13.67 Gy	13.70 Gy
SD[a] (repeatability of the signal)	0.02%	0.08%	0.16%	0.07%	0.11%	0.08%

[a] SD – standard deviation of mean value, calculated from 10 samples.

by a diode is 0.4 mm narrower than that obtained by Markus. This may be explained by a better resolution of the silicon diode comparing to the ionization chamber.

The repeatability of the detector signal allows to determine wherever a detector can be used as a radiation dosimeter. Results of signal repeatability measurements, both in ^{60}Co gamma field and proton beam, are presented in Table 2. The measurements were repeated 10 times with unchanged position and for the same values of monitor units (MU). The standard deviation (SD) of mean value of all measurements indicates the level of repeatability test.

The detector response in the proton beam as a function of dose to water was measured using DD and Si diode. Measurements were performed at d = 5 mm depth, in the range from 1 to 30 Gy. These tests indicated that the signal of the tested detectors was proportional to the accumulated dose.

The dose rate dependence was evaluated using formula (1). The Δ value of 0.996 ± 0.005 was obtained for the diamond detector. This is in agreement with the values obtained at the same energy proton beam [3, 4, 13].

Finally, using the formula (4) the values of the correction factor $k^{DD}(R_{residual})$ for the DD in 60 MeV proton beam were calculated. The empirical values of $k^{DD}(R_{residual})$ as a function of $R_{residual}$ are compared to results presented by Fidanzio [4], and shown in Fig. 2. This data are essential for accurate dosimetry of a modulated proton beam, and are obtained with uncertainty not exceeding 2.5%.

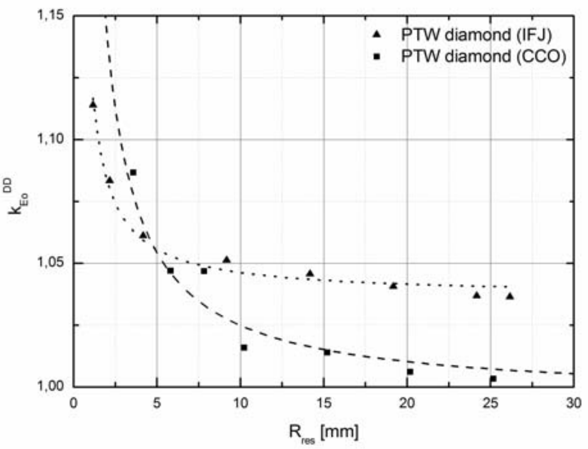

Fig. 2. The empirical coefficient $k^{DD}(R_{residual})$ for PTW natural diamond detector in the proton beam as a function of residual range, $R_{residual}$, tested in ocular proton beam at IFJ Kraków (values obtained with uncertainty not exceeding 2.5%) and in ocular proton beam at the Clatterbridge Center of Oncology (CCO) [4].

Discussion and conclusion

The performance of a natural diamond detector and the Si diode has been studied in the 60 MeV ocular proton beam at the IFJ PAN. Our results confirmed promising properties of the diamond detector for its applications in the dosimetry of proton beam.

The depth dose distribution can be measured using DD, however during measurements in phantom certain parameters (e.g. the values of water equivalent of the entrance windows, the reference point location and differences in holder geometry) have to be taken into consideration. Figure 1 shows that the Si diode gives a signal which, by Co-60 calibration, gives efficiency relative to Markus IC exceeding unity in all parts of the Bragg curve. After preirradiation, the detectors based on n- or p-type silicon, exhibit recombination effects that are related to the proton stopping power, yielding an increase in the detector signal per unit dose with increasing LET. Only for Hi-p Si diode the shape of Bragg curve correlates with those measured with ionization chamber [6, 10]. Despite the LET dependence, diodes are useful to measure dose distribution at constant energy (e.g. lateral beam profiles).

The stability of the signal, for both tested detectors, is only slightly higher than that achieved by ionization chamber (Table 2). The dose dependence is linear in a wide dose range.

The Δ value reported in this study is consistent with those reported in the literature by other authors [4, 7, 12, 13]. Moreover, it is comparable with those obtained with other beam qualities (X-ray and electron beams) [3, 5]. In order to adapt the detector signal for the dose rate effect the correction factor should not exceed 1.005.

The diamond detector showed stable calibration factor, not changing with accumulated dose [18]. Additionally, Fidanzio *et al.* [4] reported that the DD calibration factor is constant (with signal precision 0.3%) after the accumulated proton dose reaches 1 kGy. The stability of the calibration factor for diodes is not so stable. It was reported [6] that the sensitivity of diodes decreases with the accumulated proton dose more than for photon and electron beams. Decreasing sensitivity suggests that radiation damage effects are increased for protons. This may be induced by direct collision of primary heavy charged particles with Si nuclei which contribute to their displacement in the crystal lattice. The damage mechanism increases recombination effects and thereby significantly affects the collecting charge efficiency. Therefore, diodes should not be used for beam calibration, since their sensitivity changes after repeated use, due to radiation damage [14, 15].

Our results show that the diamond detectors and dosimetric diodes are useful tools for QA of therapeutic proton beam, but each type of detectors has specific properties which should be taken into account when choosing particular application.

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