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A synthetic diamond detector as transfer dosimeter for D_w measurements in photon beams with small field sizes

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Abstract
A chemical vapour deposition diamond detector fabricated at Rome 'Tor Vergata' University was investigated for its applicability as transfer dosimeter in radiotherapy photon beams with small field sizes. The detector consists of a single crystal diamond with a very small sensitive volume (0.004 mm^3). The detector showed a measurement repeatability of 0.1% and a long term reproducibility of 0.4%. Monte Carlo simulations revealed a response dependence on the photon beam energy of about 2% from the ^{60}Co quality to 10 MV photon beam. The calculated detector response was found to be independent of field size within 0.5% from 10 cm × 10 cm to 2 cm × 2 cm beam size for both 6 MV and 10 MV photon beams, increasing in smaller field sizes. D_w values obtained by the diamond detector were found to be in agreement with D_w values obtained by a small volume ionization chamber in photon beams with field size down to 2 cm × 2 cm.

1. Introduction

Synthetic diamonds produced by chemical vapour deposition (CVD) have been considered for years potentially suitable for dosimetry in small field sizes due to their small volume and near water-equivalence. The CVD polycrystalline diamond detectors developed in the past were found to be inadequate for absolute dosimetry due to the poor stability and reproducibility of the detector response. More recently a new detector based on a single crystal diamond (SCD) developed at Rome 'Tor Vergata' University [1] showed a good signal stability and measurement repeatability, without the need for pre-irradiation with a high level of absorbed dose. Moreover, the detector has a very small sensitive volume that makes it potentially useful in radiotherapy dosimetry when high spatial resolution is required. Within the framework of the External Beam Cancer Therapy project, the SCD detector has been investigated for its applicability as transfer dosimeter in radiotherapy photon beams with small field sizes such as those used in intensity modulated radiation therapy. Firstly the detector was evaluated for its adequacy as calibrated dosimeter, then the energy and field size dependence of the detector response were studied both experimentally and by Monte Carlo calculations and appropriate correction factors were determined. This work describes the main aspects and results of the SCD detector investigation.

2. Materials and methods

The SCD detector prototype studied in this work (figure 1) consists of a SCD substrate about 500 µm thick on which firstly a conductive boron-doped diamond layer is deposited, then on the doped surface an intrinsic diamond layer acting as sensitive material (about 1 µm thick) is grown. The diameter of the sensitive volume as defined by an aluminium contact is 2.2 mm. Therefore, the detector has a sensitive volume of 0.004 mm^3. The diamond is embedded in a waterproof PMMA housing with about 8 mm external diameter and 3.5 cm length. The sensitive volume is positioned at about 1.5 mm from the top of the external housing. For measurements performed

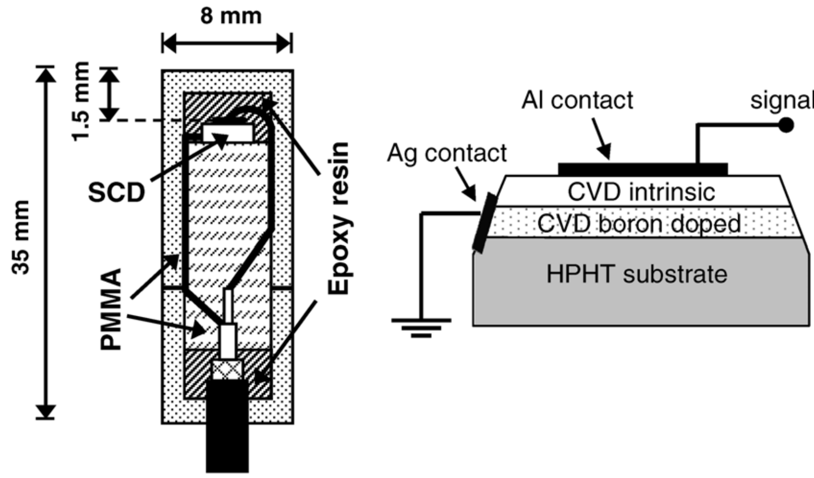

Figure 1. Schematic diagrams (not to scale) of the SCD dosimeter showing the longitudinal cross section of the dosimeter (left) and details of the SCD including the electrical contacts (right).

in this work the dosimeter was connected to a computer-controlled Keithley 6514 electrometer by means of a low noise triaxial cable. The detector was operated without applying any external polarizing voltage. The diamond dosimeter was irradiated with its symmetry axis parallel to the beam axis (axial orientation) and with the centre of the diamond sensitive volume positioned at the depth of measurement. Even though the detector has a lower spatial resolution (2.2 mm) when irradiated in the axial orientation with respect to the perpendicular radial orientation (1 µm), the axial orientation was considered the most appropriate for performing reference measurements in small field sizes as it minimizes possible stem effects on the dosimeter response.

The dosimetric properties of the SCD detector were investigated in a ^{60}Co gamma beam and in 6 MV and 10 MV photon beams produced by a Varian DHX clinical accelerator.

The energy and field size dosimeter response in terms of absorbed dose to water, D_w, was determined by Monte Carlo simulations with the EGSnrc/egs_chamber user code [2, 3]. The EGSnrc C++ geometry package [4] was used for modelling the dosimeter according to its technical data. The dosimeter response was determined as a ratio of the average absorbed dose in the diamond sensitive volume, $\overline{D}_d$, and the average absorbed dose in the same volume when the dosimeter is replaced by water, $\overline{D}_w$. Simulations were performed for ^{60}Co, 6 MV and 10 MV photon beams. A cut-off energy of 1 keV was used for both photons and electrons. Dosimeter irradiations in a 30 cm × 30 cm × 30 cm water phantom at the reference depth of 10 cm were simulated for beam field sizes from 10 cm × 10 cm down to 1 cm × 1 cm. For each field size the $\overline{D}_d/\overline{D}_w$ ratio was calculated within the same simulation applying the correlated sampling technique implemented in the egs_chamber code [2]. It was assumed that the major photon beam characteristics affecting the detector response were the spatial and energy distributions of the photons. In the simulations the beam incident at the phantom surface was modelled as a divergent beam produced by a point source with a source-to-detector distance, SDD, of 100 cm. The literature photon spectra [5] distributed within the EGSnrc package were used as input. Some additional simulations were also performed using the beam spectra obtained by simulating with the BEAMnrc code [6] photon beams with different field sizes produced by the Varian DHX accelerator. The BEAMnrc simulations were used exclusively to determine the spectral fluence needed as input in the egs_chamber code. Therefore, the variation with field size of backscattering into the accelerator monitor chamber was not determined since the calculated $\overline{D}_d/\overline{D}_w$ value is not influenced by this.

To improve the simulation efficiency for the calculation of the SCD detector response, various variance reduction techniques available in the egs_chamber user code (i.e. photon cross section enhancement, intermediate phase-space storage and correlated sampling) [2] were applied and the simulations were run in parallel on a cluster formed by 156 CPU. Nevertheless, due to the very small volume of the scoring region (0.004 mm^3) the simulations were still extremely time consuming so that the $\overline{D}_d/\overline{D}_w$ ratios were determined with a statistical uncertainty typically of 0.3%.

The beam quality correction factors, $k_{Q,f}$, needed for D_w measurements were determined from the calculated SCD detector response as

$$k_{Q,f} = \frac{\overline{D}^{f}_{\mathrm{w},Q}}{\overline{D}^{f}_{\mathrm{d},Q}} \bigg/ \frac{\overline{D}^{10\times10}_{\mathrm{w},^{60}\mathrm{Co}}}{\overline{D}^{10\times10}_{\mathrm{d},^{60}\mathrm{Co}}} \qquad (1)$$

where $\overline{D}_d$ and $\overline{D}_w$ have the same meaning as above and the indexes Q and f refer to the beam quality and field size, respectively.

The calculated $k_{Q,f}$ factors were used for D_w measurements at the reference depth of 10 cm in water, z_{ref}, and SDD 100 cm in photon beams with field sizes from 10 cm × 10 cm to 2 cm × 2 cm. The results were compared with those obtained by a PTW 31014 small volume ionization chamber using the k_Q factors reported in the IAEA TRS-398 dosimetry protocol [7] and assuming that the chamber response is independent of field size.

The ionization chamber was irradiated with the chamber main axis parallel to the beam axis as the diamond detector. The effective point of measurement was considered to be 2 mm

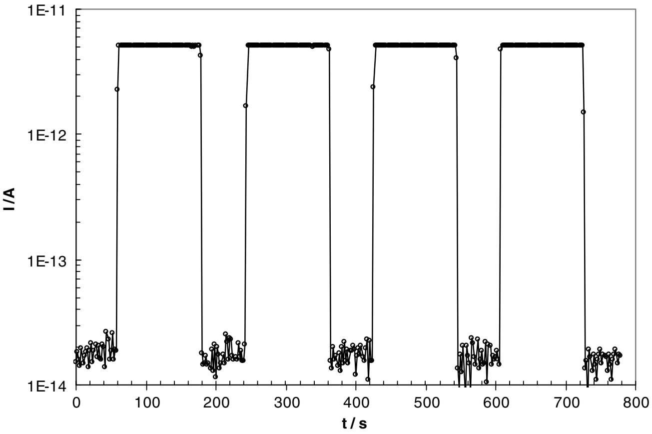

Figure 2. Diamond dosimeter current recorded during subsequent 2 min irradiations (delivered dose of 0.8 Gy) in a ^{60}Co beam followed by a pause of 1 min. The repeatability of the current average value obtained in each irradiation is 0.04%.

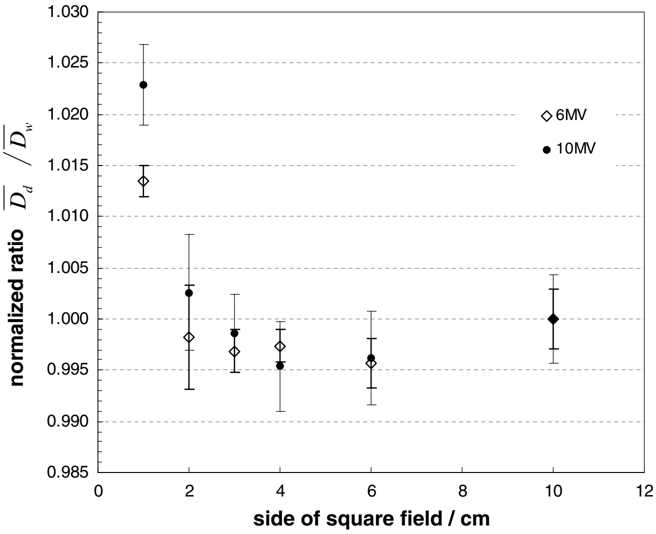

Figure 3. Monte Carlo calculation of field size dependence of the diamond detector response in terms of D_w for 6 MV and 10 MV radiotherapy photon beams. $\overline{D}_d$ is the calculated absorbed dose in the sensitive volume of the diamond detector and $\overline{D}_w$ is the calculated absorbed dose in the same volume when the detector is replaced by water. The $\overline{D}_d/\overline{D}_w$ ratios were normalized to the value obtained in the reference field size (10 cm × 10 cm). The error bars refer to the type A uncertainty.

from the chamber tip. Preliminary tests established that after a pre-irradiation with a dose of 5 Gy the chamber signal is stable within 0.2%. The tests also showed a linear dependence of $1/Q$ versus $1/V$, where Q is the measured ionization charge and V the polarizing voltage, for voltage values up to 100 V. Thus the chamber was operated with a polarizing voltage of 100 V. The ionization chamber signal was corrected for the ion recombination and polarity effects according to the IAEA TRS-398 dosimetry protocol [7].

For measurements in small fields, the positioning of the detectors in the centre of the beam was performed by measuring the beam profiles and positioning the detector at the centre of the profiles.

3. Results and discussion

3.1. Diamond detector characterization

Measurements in a ^{60}Co beam showed that after a pre-irradiation with an absorbed dose of 2 Gy the SCD detector signal is stable within 0.5%. In figure 2 the dosimeter current as a function of time obtained in subsequent irradiations in a ^{60}Co beam with a dose rate of 0.4 Gy min^{-1} is given. The data refer to irradiations lasting 2 min followed by a pause of 1 min. The dosimeter showed a short response time since the current reaches its stable value within 1 s when the beam is switched on and off. The dark current was typically 10^{-14} A and the dosimeter sensitivity about 0.8 nC Gy^{-1} so that the correction accounting for the dark current is within a few tenths of a per cent at typical radiotherapy dose rates. The repeatability of the average value of the current obtained in each irradiation was within 0.1% (1 σ) both in the ^{60}Co beam and in accelerator photon beams. The reproducibility of the dosimeter response in ^{60}Co beam measured over a period of one year was 0.4% (1 σ). The dosimeter response was found to be independent of dose rate within 0.2% in the range from 0.2 Gy min^{-1} to 5 Gy min^{-1}. No effects within 0.2% were observed on the dosimeter response due to the temperature during irradiation in the range from 18 °C to 25 °C.

3.2. Monte Carlo results

The Monte Carlo calculations showed that the ratio $\overline{D}_d/\overline{D}_w$ in the reference field size (10 cm × 10 cm) varies from 0.893 at ^{60}Co quality to 0.882 at 6 MV and 0.877 at 10 MV photon beams. Results obtained using the literature photon beam spectra ($TPR_{20,10}$ 0.67 and 0.73) and the specific spectra derived from the simulations of the Varian photon beams ($TPR_{20,10}$ 0.668 and 0.738 for 6 MV and 10 MV photon beams, respectively) agreed with each other within the Monte Carlo statistical uncertainty of 0.3%.

The Varian beam simulations showed that when the field size decreases, the low energy components in the spectral fluence decrease, leading to an increase in the photon mean energy. However, these variations in the beam spectral fluence were found to not significantly affect the calculated dosimeter response. In figure 3 the SCD detector response normalized to the response in the reference field size is shown as a function of the field size. For both 6 MV and 10 MV photon beams the calculated detector response is independent of the field size within 0.5% in the range from 10 cm × 10 cm to 2 cm × 2 cm, while it tends to increase for smaller field sizes. On the basis of these results, values of $k_{Q,f}$ factors determined in the reference field size can be applied for D_w measurements down to 2 cm × 2 cm beam size while in smaller field sizes $k_{Q,f}$ values specifically calculated using equation (1) are required.

3.3. Measurements in accelerator beams

Under reference conditions, D_w values obtained by the SCD detector were in agreement with those obtained by the ionization chamber, resulting in differences of −0.8% and −0.3% at 6 MV and 10 MV, respectively. These differences

Table 1. Output factors measured with the SCD detector, OF_{dia}, and the PTW 31014 ionization chamber, OF_{IC}, in 6 MV and 10 MV accelerator photon beams as a function of field size f. Measurements were performed at 10 cm depth in water with a source-to-detector distance of 100 cm.

Side of square field, f/cm	$TPR_{20,10} = 0.668$			$TPR_{20,10} = 0.738$		
	OF_{dia}	OF_{IC}	$\frac{OF_{\mathrm{dia}}}{OF_{\mathrm{IC}}}$	OF_{dia}	OF_{IC}	$\frac{OF_{\mathrm{dia}}}{OF_{\mathrm{IC}}}$
10	1	1	1	1	1	1
6	0.917	0.918	0.999	0.934	0.929	1.005
4	0.858	0.860	0.998	0.883	0.881	1.002
3	0.823	0.827	0.995	0.847	0.850	0.996
2	0.781	0.785	0.995	0.787	0.788	0.999

are well within the uncertainties considering the uncertainty on the ion chamber k_Q of 1% (1 σ) [7].

Output factors, $OF = D_{\mathrm{w}}(z_{\mathrm{ref}}, f)/D_{\mathrm{w}}(z_{\mathrm{ref}}, 10 \times 10)$, measured down to a field size 2 cm × 2 cm with the diamond detector and the PTW 31014 chamber are summarized in table 1. The values of k_Q factors were considered not to vary in this field size range for both the diamond detector and the ion chamber. The relative combined standard uncertainty on the OF values was estimated to be 0.3% (1σ) for both the SCD detector and the ionization chamber, without including the uncertainty on the ratio $k_{Q,f}/k_{Q,10\times10}$ equal to unity. An agreement within 0.5% was found between diamond dosimeter and ionization chamber measurements.

4. Conclusions

The single crystal diamond dosimeter investigated in this work showed dosimetric characteristics adequate for absolute dosimetry in radiotherapy. The measurement repeatability was comparable to that observed for ionization chambers and the long term reproducibility was 0.4%. The dosimeter response in terms of D_{w} varies by about 2% in the energy range from the ^{60}Co quality to 10 MV photon beam. Thus, beam quality correction factors are needed for D_{w} measurements with low uncertainty (around 1%) in radiotherapy beams. The detector response was found to be independent of the field size from 10 cm × 10 cm down to 2 cm × 2 cm. In smaller field sizes the Monte Carlo simulations showed a rapid increase in the detector response requiring the application of appropriate correction factors to the detector signal. D_{w} values obtained with the diamond detector using the Monte Carlo calculated beam quality factors were found to be in agreement with those obtained by a PTW 31014 ionization chamber down to 2 cm × 2 cm field sizes. All these results demonstrate that the SCD detector can be considered reliable for use as a calibrated dosimeter in photon beams with small field sizes.

Acknowledgments

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