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Synthetic diamond X-ray dosimeter for radiotherapy

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A B S T R A C T

The performance of a synthetic diamond X-ray detector during typical clinical beam characterisation procedures was compared to the performance of standard clinical detectors; this diamond detector used a single crystal diamond film synthesised using chemical vapour deposition as its sensitive element. Measurements were performed using 6 MV photons from a Varian 600C linear accelerator. The procedures measured the dose profile with depth along the central axis in a phantom (tissue maximum ratio) for a 10×10 cm^2 field, variation in dose at the isocentre with field size (output factor), and dose profile across and beyond the X-ray beam (off-axis ratio) for 10×10 cm^2 and 1×1 cm^2 fields. Tissue maximum ratio values were within 0.8% of the values from a standard ion chamber, over a depth range of 1–15 cm. Output factors were measured for field sizes from 0.6×0.6 cm^2 to 15×15 cm^2 and agreed well (<1.3% difference) with available ion chamber data for field sizes down to 3×3 cm^2. Off-axis measurements showed reduced penumbral width when the lateral size of the detector was reduced by holding the diamond detector in an edge-on orientation; values were comparable to those obtained using a diode detector. Overall, these results demonstrate the potential of synthetic diamond detectors for clinical beam characterisation, particularly for small beam sizes.

1. Introduction

Modern clinical radiotherapy techniques demand advanced dosimeters for beam characterisation. Desirable characteristics include: small dimensions, high-spatial resolution; high sensitivity, as the dose deposited in a small volume is likely to be small; energy and dose rate independence; fast response; and tissue equivalence, so dose can be obtained without further corrections.

Diamond has been considered as a detector material for many years, e.g. (Cotty, 1956; Kozlov et al., 1974; Planskoy, 1980; Burgemeister, 1981), for reasons including: high atomic density, increasing density of interaction with radiation beam making small detectors viable; near-tissue equivalence, as the atomic number of diamond ($Z = 6$) is close to the effective atomic number of biological tissue ($Z_{eff} \approx 7.4$); being chemically inert; and having high radiation tolerance, meaning long detector lifetimes. Natural diamond detectors are commercially-available (PTW, 2008). However, the disadvantages of high cost and low availability arising from the scarcity and variability of suitable high quality natural diamonds means that, even though they have highly desirable characteristics, they are not commonly used in radiotherapy dosimetry.

The improvement in quality of synthetic diamond in recent years, particularly through chemical vapour deposition (CVD), has increased the interest in diamond as a radiation detection material; many reviews of the CVD synthesis of diamond are available, e.g. (Butler et al., 2009; May, 1995; Gracio et al., 2010). We have recently reported on the suitability of a range of inexpensive commercially-available synthetic diamond films for radiotherapy detectors (Lansley et al., 2009a,b,c) and further analysis of the operating parameters of the best performing of these detectors (Betzel et al., 2010). In this work, we present results from typical clinical beam characterisation procedures (tissue maximum ratio, output factor, and off-axis ratio measurements) using this detector, and compare its performance to that of detectors commonly used for such procedures.

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2. Materials & methods

2.1. Detector material

The sensitive material used in the fabrication of this detector was single crystal CVD diamond purchased from Element Six Ltd. (2010). It was in the form of a transparent 3-mm-by-3-mm square tile of thickness 0.5 mm. It was nominally undoped; the concentration of nitrogen atoms was quoted as being less than 1 ppm and boron atoms as less than 0.05 ppm. The top and bottom surfaces had both been polished to an average surface roughness (R_a) of less than 30 nm.

2.2. Detector fabrication

A sandwich detector structure was fabricated by thermally-evaporating a 1-mm-diameter, ~150-nm-thick silver contact on each side of the diamond film. The diamond film was then attached, both physically and electrically, to the end of a narrow strip of printed circuit board (PCB) and housed within a custom-designed Perspex enclosure (Fig. 1). The external dimensions of this enclosure were made to be the same as a Perspex protective sheath used for a thimble ionisation chamber used at Christchurch Hospital; this permitted use of the hospital's solid water phantom with minimal air cavity around the detector. Further details of the detector construction are reported elsewhere (Lansley et al., 2009b).

2.3. Detector characterisation

All measurements reported here were obtained using a Varian 600C treatment linear accelerator (linac) at Christchurch Hospital. This linac was operated at 6 MV and was able to provide dose rates of 50, 100, 150, 200, and 250 monitor units (MU) per minute. A dose rate of 250 monitor units per minute was used to obtain the results reported here.

Unless otherwise stated, the detector was placed at a depth of 10 cm in a solid water phantom ($30 \times 30 \times 20$ cm^3) and positioned at the isocentre of the linac (source–detector distance of 100 cm); the position of the sensitive volume within the detector casing and hence its position within the phantom were known, meaning that positioning of the detector at the isocentre could be done using the linac alignment lasers. The default field size was 10×10 cm^2. Using these parameters, the dose at the detector was ~0.77 cGy per monitor unit (cGy/MU). The axis of the detector was coincident with the rotation axis of linac gantry.

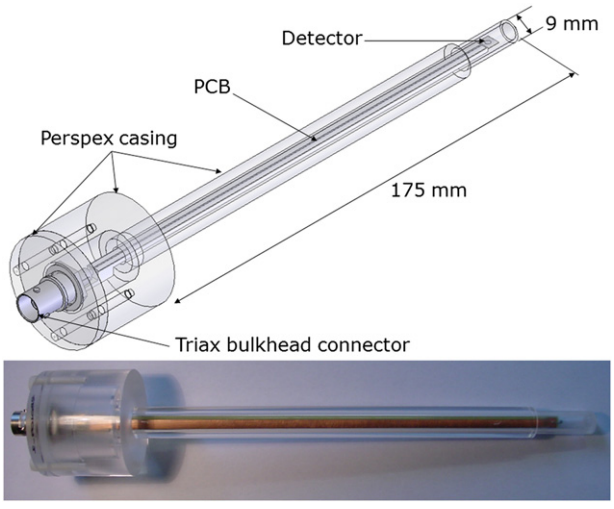

Fig. 1. Schematic and photograph of the detector.

Triaxial cabling was used between the detector and the electrometer, a Farmer 2570/1 Dosemeter, which was located outside the shielded treatment bunker. A bias of ~248 V was applied across the detector. The measurement reported here were obtained using fixed dose deliveries, rather than continuous exposure. The photo-induced charge was measured by the electrometer by integrating the current through the detector during a time interval long enough to span the duration of the dose delivery, e.g. for a dose of 100 MU delivered at 250 MU/min a time interval of 30 s was used. The charge through the detector without exposure to the X-rays (i.e. due to the leakage or dark current) was measured over the same time interval, and was subtracted from the data obtained during X-ray exposure. However, the dark current was typically <1 pA, compared to several nA during exposure, and so subtraction of this had minimal affect on the data.

Prior to measurements being taken, the detector was primed (pre-irradiated) with a dose of 1000 MU (~7.7 Gy). This was to stabilise the short-term detector response, through the filling of electronic trap states arising from defects in the diamond (Bergonzo et al., 2007). This dose had previously been shown to be sufficient for this detector (Lansley et al., 2009a,b). It is also comparable to the priming dose range (5–10 Gy) specified by PTW for their natural diamond detectors (PTW, 2008).

2.3.1. Tissue maximum ratio

The Tissue Maximum Ratio (TMR) displays the dose variation with depth (d) in water (or a water-equivalent phantom) when the detector is kept at a fixed distance from the source, i.e. the source–detector distance (SDD) is constant; the detector is usually held at the isocentre. The dose measured at each depth is normalised to the maximum dose measured, which occurs at a depth d_{max}.

The TMR was measured with the detector at the isocentre, i.e. an SDD of 100 cm, and slabs of Solid Water were added to increase the measurement depth, d, in the range 1–15 cm. At each depth, five exposures of 100 MU were recorded but the first measurement was ignored, to remove any short-term change in sensitivity; variation in the measured doses was less than 0.7%. The depth at which the maximum dose was measured was 1.5 cm, as expected for 6 MV photons.

In order to compensate for dose rate dependence of the detector, a power law correction factor $1/\Delta$ was used, as per (Fidanzio et al., 2005). During previous characterisation of this detector, it was found to have a dose rate dependence power law exponent (Δ) value of 0.92 (Betzel et al., in press). Hence this value was used in the dose rate dependence correction.

2.3.2. Output factor

The Output Factor (OF) shows how the dose at a point (the isocentre of the linac) varies as the field size is varied; SSD, SDD, and d were all kept constant. The linac jaws were used to change the field size. It is normal to use a square field, and measured doses were normalised to the dose measured for a 10×10 cm^2 field. The Output Factor is machine specific as it depends on the spectral output of the machine and the scatter from the jaws.

Output Factor measurements were taken with the detector in 'edge-on' orientation at the isocentre (SDD = 100 cm) and a depth d of 10 cm. For each field size, five exposures of 75 MU were recorded, with the first measurement ignored; this was a smaller dose than that used for the TMR measurements, but it did not affect the repeatability (<0.5% variation). As with TMR measurements, OF