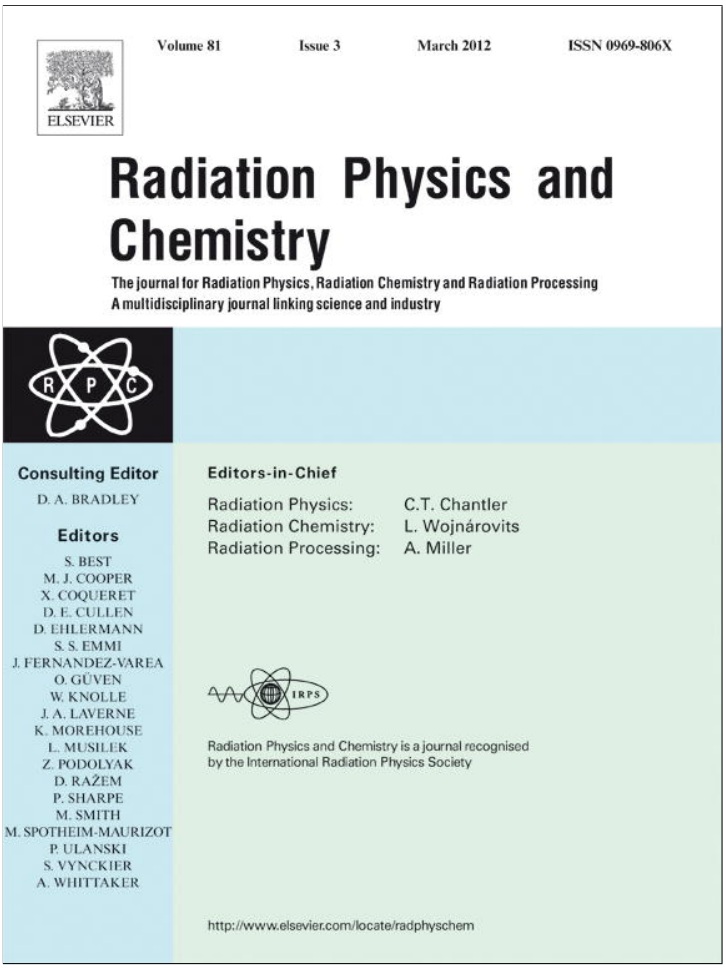

Radiation Physics and Chemistry 81 (2012) 232–239

Contents lists available at SciVerse ScienceDirect

Radiation Physics and Chemistry

journal homepage: www.elsevier.com/locate/radphyschem

A synthetic diamond probe for both low-energy mammography X-rays and high-energy electron therapy beams

N. Ade [a], T.L. Nam [a,*], M. Assiamah [b]

[a] School of Physics, Radiation and Health Physics Unit and DST/NRF Centre of Excellence in Strong Materials, University of the Witwatersrand, Private Bag 3, Wits 2050, Johannesburg, South Africa
[b] Radiological and Medical Sciences Research Institute, Ghana Atomic Energy Commission

ARTICLE INFO

Article history:
Received 16 May 2011
Accepted 17 November 2011
Available online 25 November 2011

Keywords:
Synthetic diamond probe
'Edge-on' orientation
'Flat-on' orientation
Dosimetry
Mammography X-rays
Megavoltage electron therapy beams

ABSTRACT

Although diamond has been studied for dosimetry principally due to its near-tissue equivalence, its use in both low-energy X-rays and high-energy electron beams has not been reported. This report is based on dosimetric studies of a synthetic diamond probe when subjected to diagnostic mammography X-ray photons and megavoltage electron therapy beams. The probe, constructed using entirely tissue-equivalent Perspex body, was configured for radiation dose measurement in either 'edge-on' or 'flat-on' exposure geometry without having first to re-orient the diamond within the body of the detector, and it was designed to be compatible with commercial electrometer systems. The radiation response of the diamond tested showed negligible energy dependence; its minimal background signal, high sensitivity (547.52 nC Gy^{-1} mm^{-3}) and suitability for measurements in small radiation fields of steep dose gradients due to its small size make it suitable for clinical dosimetry. The presented probe has the potential advantage of replacing conventional radiation dosimeters.

1. Introduction

To qualify as an ideal radiation dosimeter, the radiation sensing element and its housing should have attenuation properties similar to that of human tissue. It should also be possible to use the detector in small radiation fields where high spatial resolution is required. Conventional radiation dosimeters commonly used for photon and electron dosimetry, the air ionization chambers and silicon diodes have a range of applicability limited by the design and intrinsic properties of their respective detector elements (Heydarian et al., 1993). While ionization chambers are not suitable when high spatial resolution is required, silicon diodes have a significant disadvantage of being non-tissue equivalent ($Z_{Si}=14$), (Björk et al., 2000). Separate shielded and unshielded diodes are also required for photon and electron dosimetry, respectively (Heydarian et al., 1993). Two types of air ionization chambers are recommended for the dosimetry of megavoltage electron therapy beams, cylindrical chambers and plane-parallel chambers (Rikner, 1985; AAPM, 1983). However, the displacement and electron fluence corrections for cylindrical chambers, the size and angular dependence of plane-parallel chambers, and the energy dependence and air density corrections of air ionization chambers are some of the drawbacks.

Diamond, whether natural or synthetic, is attractive for use as a radiation dosimeter because of its desirable physical and dosimetric properties. The near-tissue equivalence of diamond ($Z_C=6$ is close to the effective atomic number of human tissue, $Z_{eff}=7.4$), its high sensitivity and high spatial resolution make diamond detectors suitable for clinical radiation dosimetry (Whitehead et al., 2001; Van der Merwe and Keddy, 1999; Heydarian et al., 1993; Cirrone et al., 2006; Bruzzi et al., 2000). With a high atomic number density of 1.75×10^{23} atoms cm^{-3} (Mainwood, 2000) corresponding to a molar volume of 3.44 cm^3 mol^{-1} and a mass density of 3.515 g cm^{-3}, diamond detectors with sizes of a few mm^3 can be fabricated for 'pin-point' radiation dose measurements, which is an advantage over large volume detectors. Although Heydarian et al. (1993) suggested that, unlike silicon diodes, a single diamond detector could be used for the dosimetry of both photon and electron beams, Heydarian et al. (1993) evaluated the dosimetric performance of a natural diamond detector in megavoltage electron beams only.

Thus, although the use of diamond detectors in radiotherapy and diagnostic radiology has been investigated, their use as a single probe for dosimetric measurements in low-energy diagnostic mammography X-rays and high-energy electron therapy beams has not been reported. This paper reports on the dosimetric response of a synthetic diamond probe to both beam types. The probe, constructed entirely of tissue-equivalent material, was designed to be compatible with commercial electrometer systems and it was configured to measure radiation in both 'edge-on' and 'flat-on' exposure geometries.

* Corresponding author. Tel.: +27 11 717 6878; +27 11 351 7001.
E-mail address: Tom.Nam@wits.ac.za (T.L. Nam).

0969-806X/$ - see front matter © 2011 Elsevier Ltd. All rights reserved.
doi:10.1016/j.radphyschem.2011.11.041

While natural diamond detectors are in use in a number of hospitals, the main drawbacks of using natural diamond are: no two natural diamonds can perform in an identical fashion (Yacoot et al., 1990); high cost and long delivery times due to the scarcity of suitable stones; poor reproducibility; and the lengthy selection and testing processes of diamond stones for suitable detection properties (Guerrero et al., 2004, 2005, 2006) as it has been established that the response of a diamond crystal to radiation is dependent on the types of impurities (most arguably nitrogen) and their concentrations within the crystal. Nam et al. (1991) established that single substitutional nitrogen in diamond is responsible for many performance characteristics of diamond radiation detectors. Diamonds with high nitrogen concentration tend to have much lower shallow trap concentrations than those synthesized with lower nitrogen levels (Nam, 1989). Yacoot et al. (1990) noted that the best quality diamond would give the highest photocurrent and that synthetic diamond with nitrogen content <100 ppm is good. It is now possible to synthesize diamond with controlled amount of impurities.

2. Experimental procedures and results

2.1. Diamond sample

A synthetic, single crystal, high-pressure high-temperature (HPHT) diamond crystal with a nominal volume of 48.63 mm^3 ($7.94 \times 6.38 \times 0.96$ mm^3) was investigated. The diamond sample has been prepared, characterized and metallized on both surfaces. Some of the standard analysis techniques used for diamond characterization were Raman spectroscopy, which evaluates the material quality of diamond crystals, showed the characteristic Raman line occurring at 1332.8 cm^{-1} with a Raman width of 2.27 ± 0.04 cm^{-1}, and electron spin resonance (ESR), which determines the concentration of single substitutional nitrogen impurities, gave a value of 130 ± 2 ppm. Ohmic contacts were produced by evaporating under vacuum, onto opposite surfaces of the diamond successive layers of Ti/Pt/Au metals. The metal thicknesses were 200, 200 and 2000 Å for Ti, Pt and Au, respectively, and the contacts cover the entire area of the two opposite surfaces except at the peripheral edges of the diamond. Extra precautions were taken to ensure that the peripheral edges are not contaminated by any of the metallic layers to avoid short-circuiting between the contacts.

2.2. Probe design and construction

Fig. 1 shows a schematic cross-section of the synthetic diamond probe. It is constructed using entirely tissue-equivalent material with four rods (electrodes) (Nam et al., 2004; Nam and Ade, 2011) providing electrical contacts. The diamond is sandwiched between two insulating boards each of size 10×10 mm^2 and each with conductive surfaces on opposite surfaces. Conductive adhesive provided electrical connections between the metallized diamond and the four electrodes. Electrical connections to external electronics were carried out through the use of a standard triaxial connector. Gold plated conductive clips with wires attached were used to connect onto the rods. The other ends of the wires were soldered directly onto the triaxial connector. The entire body of the probe is made up of Perspex and it is coated with a thin opaque layer (Nam and Ade, 2011) to eliminate the effects of ambient light. The probe is designed for radiation detection in either "edge-on" or "flat-on" exposure geometry. In the 'flat-on' orientation, the incident radiation is normal to the face of the diamond wafer, consequently a larger area but reduced attenuation depth. In the 'edge-on' orientation, the incident radiation is normal to a wafer edge, and thus a smaller area but greater attenuation depth as shown in Fig. 2.

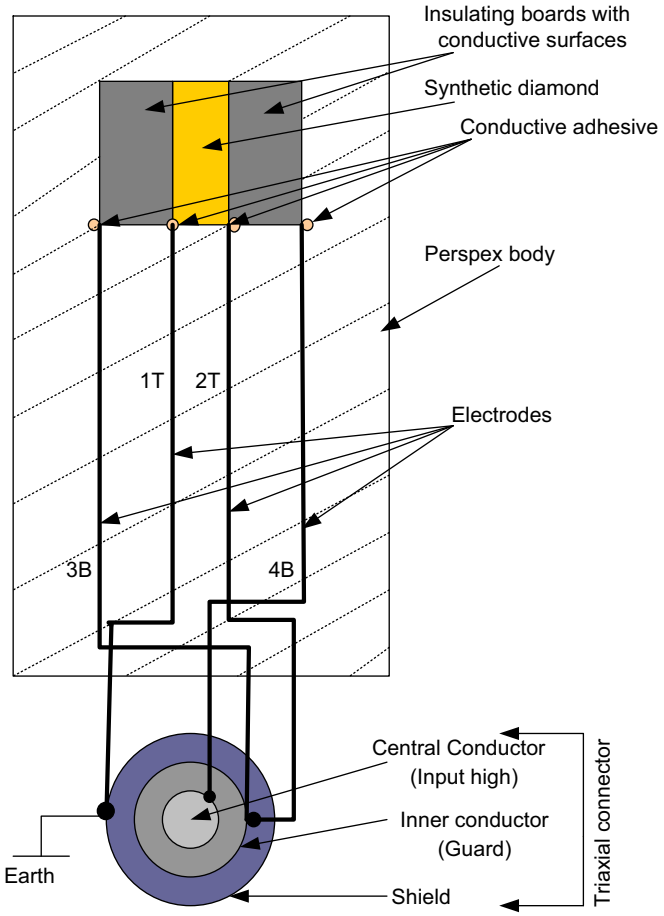

Fig. 1. Schematic cross-section of the synthetic diamond probe showing electrical connections. The top electrodes, designated 1T and 2T, provide shielding and guarding while the bottom electrodes, 3B and 4B, are connected to 'Input high' and 'Guard', respectively.

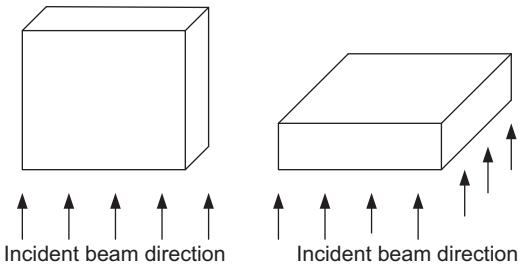

Fig. 2. 'Edge-on' (left) and 'flat-on' (right) exposure geometries of the diamond probe.

A customized Perspex phantom and a circular Perspex build-up cap were also constructed for the probe for in phantom and free-in air angular dependence measurements, respectively. A Perspex probe cover has a window, which could be used for the direct measurement of incident radiation, that is, at zero depth. When the probe cover is in position, the window is exactly above the diamond, and it allows for zero depth measurement in the 'edge-on' orientation only. Should the need be to avoid the window acting as an air cavity when probe is used in the solid phantom, the cover could be replaced with another one that has no window. It should be pointed out that there are no air cavities surrounding the diamond—the

diamond is completely encapsulated with Perspex and surrounded by the two insulating boards. Any unwanted signal generated other than that within the diamond by impinging radiation is directed away to the guard input (Fig. 1). Also, the probe's design is such that when placed in the solid phantom, the gold-plated conductive clips and wires are positioned out of the phantom/radiation field and are not irradiated.

A cylindrical prototype version of a CVD diamond probe that allows for radiation detection in both 'edge-on' and 'flat-on' exposure geometries has been designed and constructed by Assiamah (2004) and Assiamah et al. (2007) for mammography radiation dose measurements. From the results of the dosimetric studies, it was established and concluded that the 'edge-on' exposure geometry had higher attenuation levels than the 'flat-on' geometry and ought to be the exposure geometry of choice for radiation dose measurements in the mammography X-ray energy range. In addition, because both 'edge-on' and 'flat-on' geometries exhibited comparable attenuation levels at lower X-ray energies, 'flat-on' orientations would be appropriate for use at lower beam energies (<10 keV).

2.3. Apparatus

The response of the synthetic diamond probe was characterized in both mammography X-ray and megavoltage electron therapy beams. A Senographe 500 T mammography X-ray machine was used for measurements in X-rays in the energy range 25–32 kVp. Preliminary studies in mammography X-ray beams included measurements of the signal-to-noise (SNR) ratio of the diamond probe, response of the probe with varying applied voltage for polarizing voltage selection, response of the probe as a function of X-ray tube loading (mAs) and nominal X-ray tube voltage (kVp). The measurements were done using the 'edge-on' exposure geometry with the diamond probe in its customized Perspex phantom at the nominal distances used in routine mammography.

The response of the probe was also characterized in megavoltage electron therapy beams from a clinical linear accelerator (CLINAC) (Siemens Primus) in the energy range 6–14 MeV at the Charlotte Maxeke Johannesburg Academic Hospital (CMJAH). Dosimetric characterization included response of the probe with varying applied voltage for polarizing voltage selection, linearity of the probe's response with absorbed dose, energy dependence, percent depth-ionization and angular response measurements, and measurement of the output of the CLINAC with the probe. For all the measurements in electron beams, except for angular dependence measurement done free-in air, measurements were done at the depth of dose maximum d_{max}, with the probe in its customized Perspex phantom using a dose rate of 3 Gy/min, a 100 cm source-to-surface distance (SSD), a 10×10 cm^2 field size and the dose (determined by number of monitor units (MU) where 100 MU = 1 Gy) was either fixed or varied where necessary.

For all the measurements, the probe was connected to a PTW-Freiburg UNIDOS E electrometer system. Unless otherwise specified, the charge or current measured by the electrometer is defined as the response of the probe. UNIDOS E, operated manually in the 'charge' mode (except for SNR measurements where it was operated in 'current' mode), was set to ranges "Med" for X-rays measurements and "High" for electron beam measurements except for measurements involving polarizing voltage selection where UNIDOS E was set to range "Med" for both radiation types. For X-rays, the response was measured over a time interval of 10 s. For electron beams, the charge integration time, depending on the number of MU used, corresponded to the irradiation time of the CLINAC. In all the measurements, the incident radiation was perpendicular to the face or edge of the probe, and the effective point of measurement was assumed to be the physical center of the probe.

2.4. Dosimetric response of the probe to mammography X-rays

2.4.1. Signal-to-noise ratio (SNR) of the synthetic diamond probe

An important factor to consider in the dosimetric characterization of a solid state detector, especially a diamond detector with its relatively high impedance, is minimization and stabilization of the background signal since its effect may obscure the measured signal. Kania et al. (1993) pointed out that leakage currents and their fluctuations are an undesirable noise source in radiation dose measurements and can limit the magnitude of the applied electric field. In this study the background signal as a function of applied voltage of the diamond probe was investigated in light and dark conditions in the absence of irradiation. When exposed to ambient light the magnitude of the background signal, I_o was in the range $50\ \text{pA} \le I_o \le 100\ \text{pA}$ at all positive polarizing voltages. It was observed that exposing the probe repeatedly to mammography X-rays only marginally reduced the background signal. With the Perspex body of the diamond probe coated with a thin opaque layer to eliminate the effects of ambient light, the background signal reduced to <3 pA and it was found to vary linearly with applied voltage (Fig. 3).

The SNR (calculated as signal-to-background) as a function of applied voltage (0–400 V) of the probe was measured in mammography X-ray beams at 200 mAs and 30 kVp. With the probe placed in its customized Perspex phantom and connected to Unidos *E*, the noise was recorded and the response (signal plus noise) of the probe as a function of applied voltage when irradiated was measured. Fig. 4 shows the variation in the SNR with applied voltage.

2.4.2. Selection of polarizing voltage

The response of the synthetic probe as a function of positive polarizing voltage in the range 0–400 V was measured for

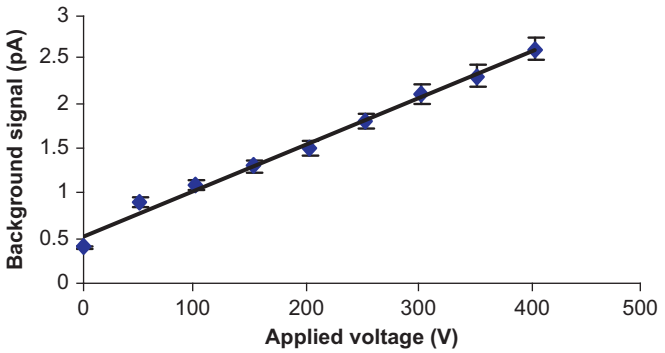

Fig. 3. Background signal of the diamond probe as a function of applied voltage.

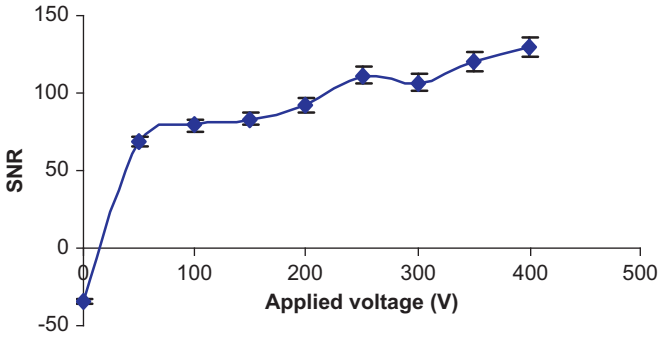

Fig. 4. Signal-to-noise ratio (SNR) of diamond probe as a function of applied voltage.

mammography X-rays to determine the operating voltage to be used. The response was measured in both the 'edge-on' and 'flat-on' orientations at tube settings of 30 kVp and 200 mAs. Saturation in response occurred at polarizing voltage in the range 100–400 V (Fig. 5). This saturation region corresponds to the ion saturation region in which ionization chambers are conventionally operated. +200 V was arbitrarily chosen as the bias voltage for all the measurements. The lower response of the 'flat-on' orientation in X-rays compared to the 'edge-on' orientation confirms the fact that the 'edge-on' exposure orientation is suitable for dose measurements in mammography rather than the 'flat-on' orientation (Assiamah et al., 2007).

2.4.3. Response of the probe to X-ray tube loading and tube voltage variations

The response of the probe as a function of X-ray tube loading (mAs) and tube voltage (kVp) variations was measured. This was carried out for the former, at constant tube voltages of 26, 28 and 30 kVp and the latter at constant tube loadings of 100 and 200 mAs, the results of which are shown in Figs. 6 and 7. The measurements were intended to assess the response characteristics of the probe with dose since mAs and kVp are two determinants of patient dose and image quality (contrast) in mammography, with patient dose increasing either with an increase in mAs or kVp. It is well known that scatter radiation increases the patient dose and degrades the image contrast. As Compton scatter accounts for most interactions above 28 kVp, tube setting between 25 and 27 kVp is generally recommended as it gives the highest possible image contrast while maintaining an acceptable radiation dose to the patient. In general the lower the kVp in a Mo/Mo target/filter combination in screen-film mammography, the better the image contrast.

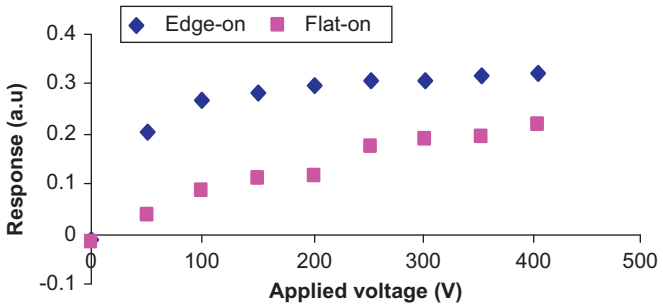

Fig. 5. Response of probe as a function of applied voltage at 200 mAs and 30 kVp.

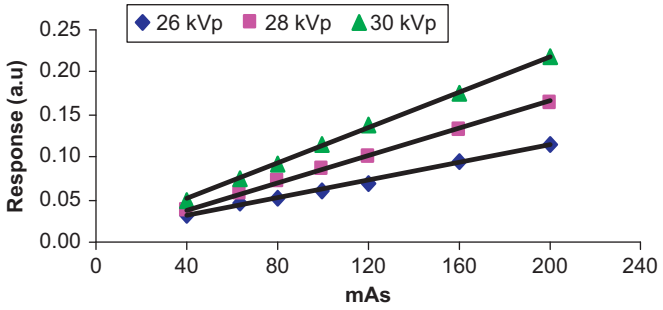

Fig. 6. Response of the diamond probe to changes in mAs at 26, 28 and 30 kVp settings.

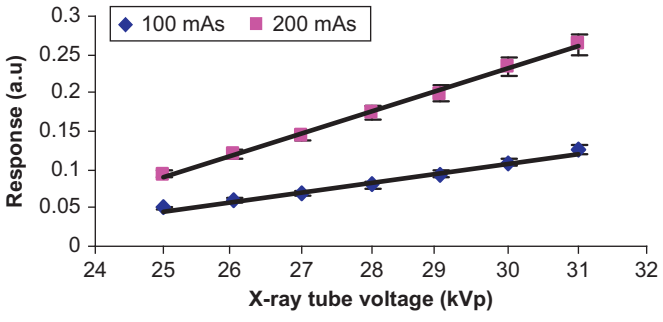

Fig. 7. Response of the diamond probe to changes in kVp at 100 and 200 mAs tube loadings.

2.5. Dosimetric response of the probe to megavoltage electron therapy beams

2.5.1. Selection of polarizing voltage

The response of the synthetic diamond probe as a function of positive polarizing voltage in the range 0–400 V, measured in the 'flat-on' exposure geometry, was studied also in a 7 MeV electron therapy beam and is shown in Fig. 8. Similar to the response measured in X-rays (Fig. 5), saturation in response also occurred at polarizing voltage in the range 100–400 V, and the probe was again biased at +200 V.

2.5.2. Dose linearity, sensitivity and energy independence

The response of the diamond probe with absorbed dose on exposure to 7 and 14 MeV electron beams was studied in order to evaluate its linearity and energy dependence, and to determine its sensitivity. The delivered dose was varied from 1 to 7 Gy (by varying monitor units from 100 to 700). Measurements were done at d_{max}, with the diamond probe in its customized Perspex phantom. Fig. 9 shows a linear absorbed dose response in the energy range, 7–14 MeV and negligible energy dependence as

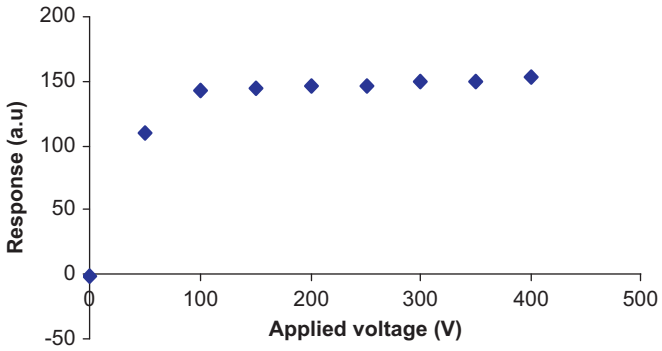

Fig. 8. Response of probe as a function of applied voltage for a 7 MeV electron therapy beam.

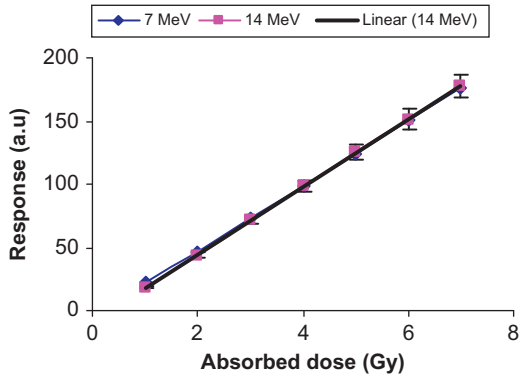

Fig. 9. Linearity of response of the diamond probe with absorbed dose for 7 and 14 MeV electron therapy beams.

depicted by the superimposition of the two graphs. This gives an advantage over air ionization chambers, which show energy dependence, needing energy corrections for dose measurements. It is well known that high atomic number metals used to form contacts (Laub et al. 1999; GÓrka et al., 2008) and air cavities could introduce some energy dependence. As pointed out in Sections 2.1 and 2.2, the metallic contact layer is an angstrom layer with negligible absorption and scatter and there are no air cavities surrounding the diamond. The observed small offset below 2 Gy could therefore be attributed to the high atomic number of the metallic conductive adhesive used to provide electrical contacts onto the diamond surfaces. This could be improved by replacing the metallic conductive adhesive with carbon based adhesive.

The sensitivity of the detector is defined as the collected charge, corrected for leakage, per unit absorbed dose and sensing volume (GÓrka et al., 2008). Based on the assumption that ionization takes place in the diamond only and taking the detector volume of 48.63 mm^{-3} as the active volume, the slope of Fig. 9 (26.626 $\mu C\ Gy^{-1}$) gave a figure of 547.52 $nC\ Gy^{-1}\ mm^{-3}$ sensitivity per unit detector sensing volume at 14 MeV electron therapy beam.

2.5.3. Depth–dose curves

Fig. 10(a) and (b) shows the depth ionization data taken with the diamond probe in both the 'edge-on' and 'flat-on' exposure geometries compared with depth dose data generated from measurements with a calibrated 0.055 cc Markus chamber (PTW-Freiburg—Type 23343) for exposures to 7 and 14 MeV therapy electron beams using a fixed dose of 0.5 Gy (50 MU). A 100 cm SSD was used with an applicator defined field of $10 \times 10\ cm^2$ for the measurements, which were taken in a Perspex phantom with the diamond probe, and a water phantom with the Markus chamber. Since the Markus chamber (acting as a reference for this study) shows energy dependence according to the Bragg–Gray cavity theory, its response has been corrected with the water to air restricted mass collision stopping power ratios, taken from Table V of the AAPM TG 21 Protocol (AAPM, 1983). The results show that, with the diamond probe, unlike exposure to mammography X-rays, the 'flat-on' exposure geometry is more suitable for relative electron dosimetry as the depth ionization data agreed favorably with the energy-corrected depth-dose data from the Markus chamber. The reason for difference in response between the 'edge-on' and 'flat-on' orientations may be attributed to an area effect termed the 'differential effect' of energy deposition as a result of a difference in surface area. With its larger surface area, more electron energy is deposited on the flat surface, resulting in creation and collection of more charge carriers by the electrodes. On the other hand, the diamond edge with a smaller surface area results in less electron energy deposition leading to a smaller creation of charge carriers and lower collection efficiency.

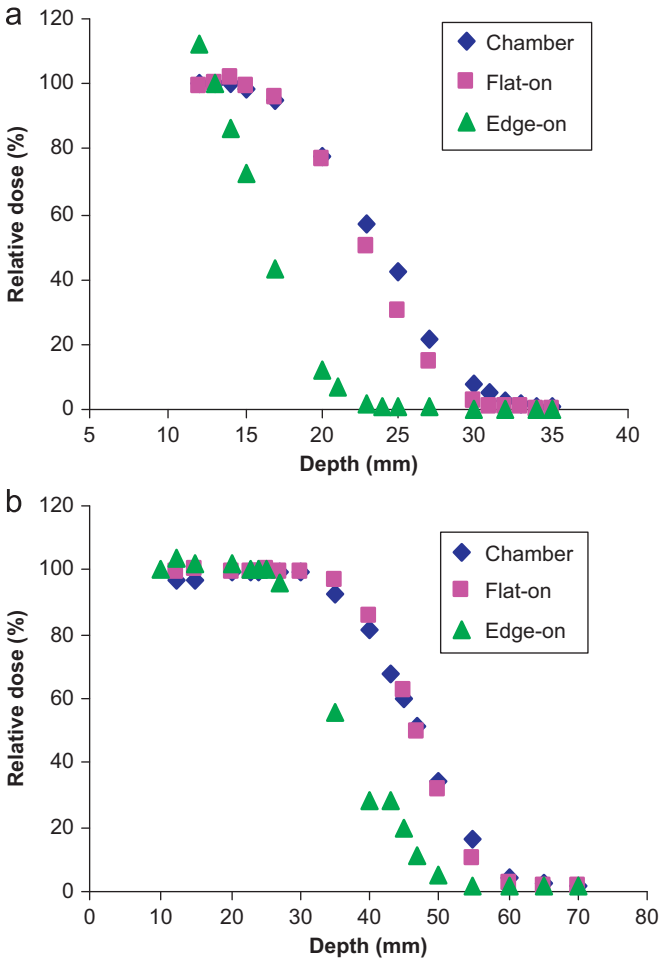

Fig. 10. Normalized depth–dose ionization in Perspex obtained with the synthetic diamond probe compared with depth–dose curves (energy-corrected ionization) as measured with the calibrated Markus chamber for (a) 7 MeV electron beam and (b) 14 MeV electron beam.

2.5.4. Directional response

"When performing dose profile measurements, the obliqueness of incidence of electrons on the detector increases with off-axis distance so the directional response of the detector is important. Directional response is also important in depth-dose measurements because the angular distribution of electrons broadens with depth," (Heydarin et al., 1993). The angular dependence of the diamond probe was measured for a 14 MeV electron beam using 50 MU. The probe was positioned at the machine isocenter and irradiated free-in air with its build-up cap attached. A $10 \times 10\ cm^2$ applicator was used and the response of the probe was measured as a function of gantry rotation at intervals of 15° from 0° to 105° to avoid collision between the gantry and the experimental set-up. The experiment was also repeated for a calibrated 0.6 cc cylindrical chamber (Farmer-Type), also irradiated with its build-up cap attached. Each detector, irradiated with its axis perpendicular to the plane of gantry rotation was suspended at isocenter with a retort stand clamped to its rigid body. The angular response of the diamond probe was measured in both the 'edge-on' and 'flat-on' exposure geometries. Fig. 11 shows the configuration for angular dependence measurement taken with the diamond probe.

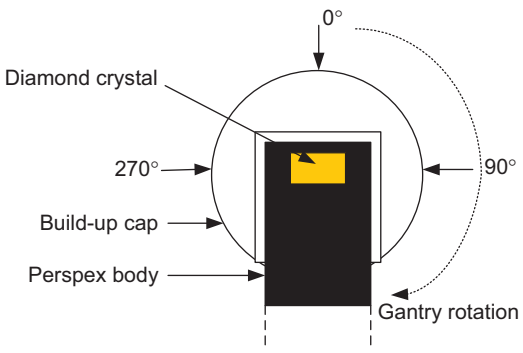

Fig. 11. Configuration for angular response measurements in the 'edge-on' orientation with the diamond probe suspended at isocentre and irradiated with its build-up cap attached.

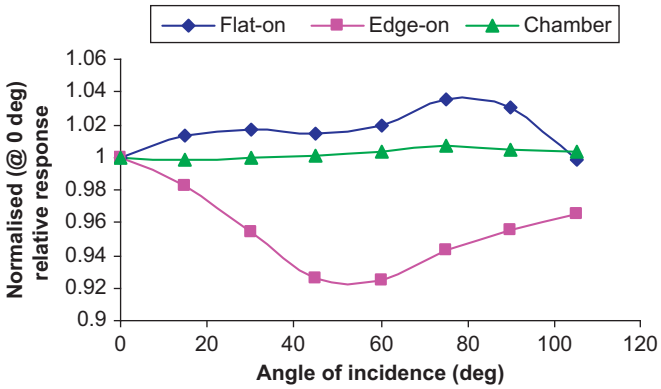

Fig. 12. Directional responses of the diamond probe and a calibrated cylindrical chamber.

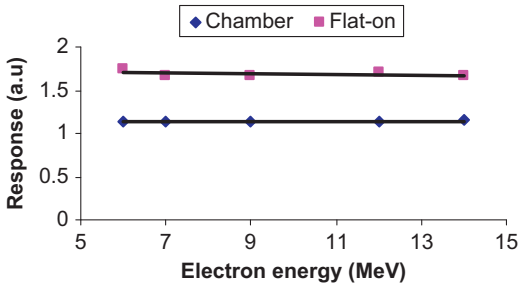

Fig. 13. Output of the CLINAC measured with both a thimble chamber and the diamond probe.

Fig. 12 shows the relative response, normalized to the response at normal incidence (i.e. at 0°) for both detectors. Owing to its construction, the cylindrical chamber (acting as a reference) has no structural angular dependence and its response remains flat to within 0.7%. The diamond probe, however, shows some angular dependence, with the response of the two geometries (relative to the cylindrical chamber) opposite to each other. Whereas the response of the diamond probe in the flat-on geometry increases slightly with its curve above that of the cylindrical chamber, the response in the edge-on geometry decreases with its curve below that of the chamber. In the 'flat-on' geometry, the change in angular response is within 2% between 0° and 60°. The stronger dependence of about 3.6% shown between 75° and 90° (also shown by the cylindrical chamber) is probably due to instability with the machine dose rate with gantry angle, a finite tolerance with mechanical isocentre (van der Merwe and Keddy, 1999) and scattering in the thin conductive layers used to form contacts. In the 'edge-on' exposure geometry, a stronger dependence of 7.5% is shown between 45° and 60°. Similar to depth dose profiles, the lower response and stronger angular dependence shown by the 'edge-on' geometry compared to the 'flat-on' geometry is due to the 'differential effect' of energy deposition, resulting in the creation of a smaller number of charge carriers and lower collection efficiency in the 'edge-on' geometry.

2.5.5. Determination of machine output with the diamond probe

Before clinical use, the output of photon and electron beams produced by teletherapy machines is calibrated. Beam calibration is the determination of radiation dose in reference irradiation conditions (Podgorsak, 2005). Beam calibration ensures optimization and accurate dose delivery to patients. As teletherapy machines are usually calibrated to deliver a dose of 1.0 Gy per 100 MU, the output of the CLINAC was measured with both a calibrated thimble (cylindrical) ion chamber (acting as a reference) and the diamond probe in the energy range 6–14 MeV. 100 MU was used to irradiate both detectors. Although the diamond probe shows a higher response compared to the chamber (Fig. 13), the response remains flat for both detectors in the energy range 7–14 MeV, except at 6 MeV where both detectors gave a higher response (probably due to more scattering at lower electron energies).

3. Discussion

Operationally for ease of use in clinical radiation dosimetry, an ideal dosimeter should have minimal leakage current or background signal, high sensitivity to the incident radiation, a wide range of linearity of response, energy and angular independence.

Taking note of the conclusions of Fidanzio et al. (2002, 2004) and Kania et al. (1993) concerning the possible effects of leakage current being a limiting parameter for efficient collection of charge carriers, the background signal and signal-to-noise ratio (SNR) of the synthetic diamond probe were investigated meticulously as a function of applied voltage in this study. Optical shielding of the probe was found to be a necessary procedure to improve its performance by reducing and stabilizing the otherwise noisy background. The background signal was found to vary linearly with applied voltage. A value of <3 pA was found at 400 V. Because the probe gave a higher response in electron beams compared to X-rays, its background signal was found to be about 0.003% of its response in 7 and 14 MeV electron beams at 3.0 Gy min^{-1} while it was about 1.08% in X-rays at 200 mAs and 30 kVp. At $+200$ V bias with a background signal of 1.5 pA, a SNR value of 92 was measured under X-rays exposure at 200 mAs and 30 kVp tube settings. In comparison, Fidanzio et al. (2004) reported a SNR value of 10 at 100 V measured in a 6 MV photon beam using a dose rate of 3.5 Gy min^{-1}, while Bruzzi et al. (2000) reported SNR values of 10 and 100 at 1000 V bias under 20 MeV electron beam exposure using a dose rate of 4 Gy min^{-1} for CVD diamond detectors.

Although the probe was pre-irradiated with a dose of 7.0 Gy, subsequent investigation showed that such a pre-requisite step is not required. For the presented probe this serves as an advantage over other commercial diamond detectors. Although Van der Merwe and Keddy (1999) also reported that pre-irradiation was not required for their synthetic diamond detector, no investigation as to the reason was carried out. In this study through systematic investigations, the cause of pre-irradiation has been isolated and attributed to the presence and effects of ambient light. Thus as long as the probe is properly shielded from light, pre-irradiation is not required.

Linearity measurements were verified for both mammography X-rays and electron beams. The linear response of the diamond probe with X-ray energy (kVp) and tube loading (mAs) was measured to assess the response characteristics with dose since kVp and mAs are two determinants of patient dose in mammography. Linearity of response with absorbed dose to 7 and 14 MeV electron therapy beams showed that the probe has negligible energy dependence. It has been suggested that since diamond is near tissue-equivalent, the detector signal is directly proportional to the absorbed dose rate to tissue and can be used without any correction (Vatnisky and Järvinen, 1993), and that relative ionization measurements taken with diamond could be used directly as dose if the ionization is assumed to take place inside the diamond alone (Van der Merwe and Keddy, 1999), which enables the straightforward measurements of dose for radiotherapy applications (Guerrero et al., 2004). A diamond detector thus has the potential advantage of becoming a reference and standard dosimeter.

An advantage of using solid state detectors is that they are more sensitive than air ion chambers. The 547.52 nC Gy^{-1} mm^{-3}

sensitivity per unit detector sensing volume of the diamond probe is much higher when compared to the values of 0.04, 370 and 50–135 nC Gy^{-1} mm^{-3} reported for a Farmer-type ion chamber at +300 V bias, a p-type silicon diode and commercial natural diamond detectors, and it compares favorably with the values of 4–599 nC Gy^{-1} mm^{-3} reported for other synthetic diamond detectors (Fidanzio et al., 2002 and 2004; Bruzzi et al., 2000; Marczewska et al., 2007; GÓrka et al., 2008; Lansley et al., 2009; Tromson et al., 2010; Betzel et al., 2010). It should however be pointed out that unlike most of the values reported in literature where the detector volume (active volume) has been assumed to be the product of the contact area (in most cases a fraction of the diamond surface) and thickness of the diamond (Betzel et al., 2010; Lansley et al., 2009; Marczewska et al., 2007), the contact area used to obtain the sensitivity in this study was an exposure area equal to the entire diamond surface.

Percent depth ionization data measured with the diamond probe in the 'flat-on' orientation exposed to 7 and 14 MeV electron therapy beams showed close agreement with energy-corrected depth-dose data generated from measurements with a parallel-plate Markus chamber. This is attributed to the homogeneity of the probe in terms of atomic composition, which is near tissue equivalent and the minimal background signal of the probe. Laub et al. (1997, 1999) and Fidanzio et al. (2004), who used diamond detectors in both megavoltage photon and electron beams, established that a sub-linearity correction (based on dose rate dependence) for depth dose curves measured with diamond was necessary before an agreement could be reached with depth dose values measured with an ionization chamber. The diamond probe used in this study showed no need for such correction. Similar results without correction for dose rate dependence have been reported (Van der Merwe and Keddy, 1999; Almaviva et al., 2010) where direct proportionality of the detector signal with absorbed dose rate was found (Van der Merwe and Keddy, 1999).

In an electron beam the distribution in energy and direction of the electrons changes rapidly with the depth of penetration, necessitating the use of detectors that are independent of energy and beam direction (Brahme and Svensson, 1976). The directional dependence of the probe was measured in a 14 MeV electron therapy beam. In the flat-on exposure geometry, the angular response remained flat to within 2% between 0° and 60°. A stronger angular dependence of about 3.6% was shown between 75° and 90° in the flat-on geometry and 7.5% dependence at 60° in the edge-on geometry. The results show that in the dosimetry of megavoltage electron therapy beams, the flat-on exposure geometry has a greater charge collection efficiency of diamond compared to the edge-on geometry as evidenced by both depth-dose and angular response measurements. As pointed out by Brahme (1985), the angular dependence of most detectors used for measurements in electron and photon beams may be due to the mechanical design of the detector or to the intrinsic angular sensitivity of the detection process. Brahme (1985) also noted that for some detectors such as semiconductor diodes, a 10−15% angular dependence is not uncommon. The results of this study show that the angular response of the diamond probe is better than such diode detectors. However, to improve the angular dependence of the presented diamond probe, the metal based conductive adhesive used to form contacts could be replaced with carbon based adhesive to reduce scattering of the incident electrons.

As beam calibration ensures accurate dose delivery to patients, the output of the CLINAC was measured with both the synthetic diamond probe and a calibrated thimble chamber. The response of both detectors remains flat in the energy range 7–14 MeV. However, the probe showed a response about 1.5 times higher than the thimble chamber in this energy range, but both detectors showed a higher response at 6 MeV.

4. Conclusions

From the presented dosimetric characterization of the response of the synthetic diamond probe to electron beams in the range 6–14 MeV and low-energy X-ray photons in the range 25–32 kVp, it can be concluded that a single well shielded low noise level probe with synthetic diamond as sensor could find applications in both low-energy diagnostic X-rays and high-energy electron therapy beams. The presented probe can hold diamond crystals of various sizes as large as $10 \times 10\ mm^2$. It is constructed using entirely tissue equivalent material, is compatible with commercial electrometer systems and is configured to allow for radiation detection in both 'edge-on' and 'flat-on' exposure geometries without having first to re-orient the diamond crystal within the body of the probe. This study not only confirms that the 'edge-on' exposure geometry is suitable for mammography dosimetry (Assiamah et al., 2007), but also concludes that because of its large surface area, the 'flat-on' exposure geometry with improved charge collection efficiency of the diamond probe ought to be the exposure geometry of choice for the dosimetry of megavoltage electron therapy beams rather than the 'edge-on' exposure geometry.

The synthetic diamond probe has the potential advantage of replacing conventional dosimeters such as silicon diodes and air ionization chambers. Its suitability in clinical radiation dosimetry lies in the use of the near-tissue equivalence of diamond in a probe housing that gave rise to linear response characteristics with dose, energy independence and minimal angular dependence, high sensitivity, minimal background signal and high SNR. Because diamond is near-tissue equivalent and is not susceptible to changes in ambient conditions, the number of variables used in the evaluation of dosimetric quantities is reduced, and thus reduction in inaccuracies or uncertainties in dose determination.

Acknowledgments

This project was generously funded by the DST/NRF Centre of Excellence in Strong Materials at the University of the Witwatersrand. Prof. D. Van der Merwe is acknowledged for her academic assistance and guidance. Our sincere thanks go to Mr. H. Sikhumbuzo Mhlanga for assisting with the experimental setups, Mr. R. Erasmus for Raman spectroscopy measurements, Mr. S. Riekert and Mr. A. Voorvelt for their technical support.

References

AAPM American Association of Physics in Medicine, 1983. Med. Phys. 10, 741.
Almaviva, S., Ciacaglioni, I., Consorti, R., De Notaristefani, F., Manfredotti, C., Marco, Marinelli, Milani, E., Petrucci, A., Prestopino, G., Verona, C., Verona-Rinati, G., 2010. Diamond Relat. Mater. 19, 217.
Assiamah, M., 2004. Dosimetric techniques for mammography mass screening programs. Ph.D. Thesis, University of the Witwatersrand, Johannesburg, South Africa.
Assiamah, M., Nam, T.L., Keddy, R.J., 2007. Appl. Radiat. Isot. 65, 545.
Betzel, G.T., Lansley, S.P., Baluti, F., Reinisch, L., Meyer, J., 2010. Nucl. Instrum. Methods Phys. Res. A 614, 130.
Björk, P., Knöös, T., Nilsson, P., 2000. Med. Phys. 27, 2580.
Brahme, A., Svensson, H., 1976. Phys. Med. Biol. 21, 304.
Brahme, A., 1985. Acta Radiol. Oncol. 24, 301.
Bruzzi, M., Bucciolini, M., Cirrone, G.A.P., Cuttone, G., Mazzocchi, S., Pirollo, S., Sciortino, S., 2000. Nucl. Instrum. Methods Phys. Res. A 454, 142.
Cirrone, G.A.P., Cuttone, G., Lo Nigro, S., Mongelli, V., Raffaele, L., Sabini, M.G., 2006. Nucl. Phys. B (Proc. Suppl.) 150, 330.
Fidanzio, A., Azario, L., Venanzi, C., Pinzari, F., Piermattei, A., 2002. Nucl. Instrum. Methods Phys. Res. A 479, 661.
Fidanzio, A., Azario, L., Viola, P., Ascarelli, P., Capelli, E., Conte, G., Piermattei, A., 2004. Nucl. Instrum. Methods Phys. Res. A 524, 115.
GÓrka, B., Fernandez-Varea, J.M., Panettieri, V., Nilsson, B., 2008. Nucl. Instrum. Methods Phys. Res. A 593, 578.

Guerrero, M.G., Tromson, D., Rebisz, M., Mer, C., Bazin, B., Bergonzo, P., 2004. Diamond Relat. Mater. 13, 2046.
Guerrero, M.G., Tromson, D., Bergonzo, P., Barrett, R., 2005. Nucl. Instrum. Methods Phys. Res. A 552, 105.
Guerrero, M.G., Tromson, D., Descamps, C., 2006. Diamond Relat. Mater. 15, 811.
Heydarian, M., Hoban, P.W., Beckham, W.A., Borchardt, I.M., Beddoe, A.H., 1993. Phys. Med. Biol. 38, 1035.
Kania, D.R., Landstrass, M.I., Plano, M.A., 1993. Diamond Relat. Mater. 2, 1012.
Lansley, S.P., Betzel, G.T., Baluti, F., Reinisch, L., Meyer, J., 2009. Nucl. Instrum. Methods Phys. Res. A 607, 659.
Laub, W.U., Kaulich, T.W., Nusslin, F., 1997. Phys. Med. Biol. 24, 535.
Laub, W.U., Kaulich, T.W., Nusslin, F., 1999. Phys. Med. Biol. 44, 2183.
Mainwood, A., 2000. Semicond. Sci. Technol. 15, R55.
Marczewska, B., Kupriyanov, I., Pal'yanov, Yu., Nowak, T., Olko, P., Rębisz, M., Waligórski, M.P.R., 2007. Diamond Relat. Mater. 16, 191.
Nam, T.L., 1989. Nuclear radiation detection properties of diamond. Ph.D. Thesis, University of the Witwatersrand, Johannesburg, South Africa.
Nam, T.L., Karfunkel, U., Keddy, R.J., Every, A.G., 1991. Radiat. Eff. Def. Solids 116, 233.
Nam, T.L., Keddy, R.J., Assiamah, M., 2004. Improvement in Radiation Detectors. SA Patent 2005/10110.
Nam, T.L., Ade, N., 2011. Improved Multi-purpose Ionizing Radiation Detector. Patent pending.
Podgorsak, E.B., 2005. Radiation Oncology Physics: A Handbook for Teachers and Students. IAEA, Vienna, pp. 451–458.
Rikner, G., 1985. Acta Radiol. Oncol. 24, 71.
Tromson, D., Rebisz-Pomorska, M., Tranchant, N., Isambert, A., Moignau, F., Moussier, A., Marczewska, B., Bergonzo, P., 2010. Diamond Relat. Mater. 19, 1012.
Van der Merwe, D.G., Keddy, R.J., 1999. Radiat. Phys. Chem. 54, 325.
Vatnisky, S., Järvinen, H., 1993. Phys. Med. Biol. 38, 173.
Whitehead, A.J., Airey, R., Buttar, C.M., Conway, J., Hill, G., Ramkumar, S., et al., 2001. Nucl. Instrum. Methods Phys. Res. A 460, 20.
Yacoot, A., Moore, M., Makepeace, A., 1990. Phys. Med. Biol. 35, 1409.