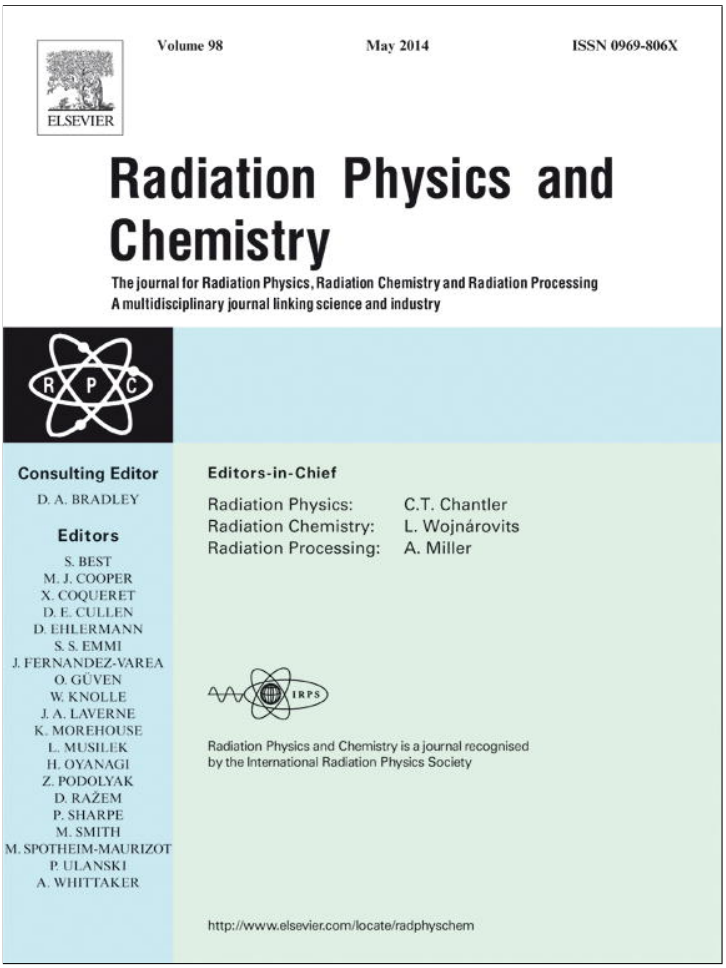

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The dose rate dependence of synthetic diamond detectors in the relative dosimetry of high-energy electron therapy beams

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HIGHLIGHTS

- We investigated the dose rate dependence of synthetic diamonds in electron beams.
- The linearity index Δ was evaluated for its dependence on electron energy.
- Values of Δ for crystals with low defect levels varied with electron energy within 2%.
- The dosimtric influence of Δ was evaluated for field sizes from 1×1 cm^2 to 10×10 cm^2.
- With crystal selection, the relative dose measured with the probe was within 1%.

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ABSTRACT

Evaluation of the linear response of a radiation detector with absorbed dose rate should be of paramount importance in clinical dosimetry. As modelled by Fowler, electrical conductivity, σ, of a solid-state detector and absorbed dose rate, D_r, are related by $\sigma \sim D_r^{\Delta}$ where Δ is the linearity index. The detector is thus independent of dose rate if Δ is unity. This contribution investigates and evaluates the dependence of Δ of synthetic diamond detectors of various types on therapy electron energy and its influence in relative electron dosimetry with the aim of selecting a suitable crystal. The study was conducted initially on one HPHT and eight CVD synthesised diamonds of optical grade (OG) and detector grade (DG) qualities using 6–14 MeV electron therapy beams. For quality control, the diamond specimens were characterised by Raman spectroscopy and electron spin resonance (ESR). Values of Δ ranging between 0.79 and 1.03 were obtained for all the nine diamond detectors at 1000 V/cm for 7 and 12 MeV electron beams. Whereas the Δ values of the HPHT diamond were found not to vary with the electron energies, those of three CVD samples of a given class varied with the electron energies within 2%. In addition, a very strong variation of about 9% was observed for two OG crystals of another class. The Δ values were found to decrease with increasing dose rate and there was a tendency for the Δ values to change with defect levels present within the crystals. Due to the independence of the HPHT diamond's Δ values on electron energy and its better stability of response to radiation, a small-size HPHT crystal was then evaluated of its potential applications in small radiation fields. Relative dose distributions measured with the diamond probe on exposure to 6, 12 and 14 MeV electron beams between 1×1 cm^2 and 10×10 cm^2 fields were compared with those obtained with reference ion chambers and a Dosimetry Diode E. The results showed that with careful selection of a suitable diamond crystal relative dose distributions taken with the diamond probe would compare favourably with those obtained with the reference detectors within or of the order of 1% with or without dose rate dependence corrections. In addition, the presented results have demonstrated for the CVD diamonds that Δ may vary with electron energy and it could be influenced by defect levels.

1. Introduction

Accurate determination of radiation dose is of paramount importance for clinical applications such as radiotherapy. Das (2009) noted that an ideal radiation dosimeter should have high

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sensitivity to radiation; be tissue equivalent and the response should be linear with absorbed dose and independent of beam energy, radiation type, dose rate and direction. In addition, the physical size should be suitable for the dosimetry of small fields and non-equilibrium (i.e. electronic equilibrium) conditions. Also, it should have a high signal-to-noise ratio, be rugged and resistant to radiation damage.

Although diamond exhibits a number of unique properties necessary to meet most of the above mentioned criteria that are desirable for dosimetric applications, diamond detectors, like silicon diodes are often reported to show dose rate dependence (Planskoy, 1980; Laub et al., 1997, 1999; Buttar et al., 2000; Cirrone et al., 2006; Schirru et al., 2008) in accordance with Fowler (1966) theory of induced conductivity in insulators. As modelled by Fowler (1966), electrical conductivity, σ, of a solid-state detector and absorbed dose rate, D_r, are related by:

$$\sigma \sim D_r^{\Delta} \tag{1}$$

where Δ is a parameter accounting for the dose rate dependence and can take a range of values around unity (usually $0.5 \leq \Delta \leq 1.0$) depending on the material characteristics. It is known that conductivity in equilibrium is proportional to recombination time and that this time is inversely proportional to the number of holes, which is equal to the sum of the equilibrium density of both free and trapped electrons (Fowler, 1966). The change in conductivity with dose rate as depicted in Eq. (1) is therefore influenced by the difference in the recombination time between electrons and holes. If many electron traps are present within the material, then the number of trapped electrons (m) relative to the number of free electrons (n), dictates the number of vacant holes (i.e. for $m \gg n$, there will be $m+n$ ($\approx m$) vacant holes), hence the change in conductivity with dose rate (Fowler, 1966; Hoban et al., 1994). According to Fowler (1966), Δ can take on values from less than one (sub linear) to one (linear) and greater than one (supra linear) depending on the purity, the capture cross sections and the distribution of traps within the detector material. In particular, for a pure crystal or when the excitation rate is so high that traps present are unimportant, $\Delta = 0.5$; if the carrier lifetime is dominated by traps with the same capture cross sections, $0.5 < \Delta < 1.0$; uniform and quasi-uniform trap distributions over a wide range of depths in the forbidden gap lead to $\Delta \approx 1$. The value of Δ may also exceed unity if traps in the crystal have differing capture cross sections (Fowler, 1966) or are distributed non-uniformly (Mohamed Abdel-Rahman et al., 2012).

Although theory predicts that conductivity is not strongly dependent on dose rate since in general $n \leq m$ and the proportional increase in m with dose rate is much less than n (Fowler, 1966), Hoban et al. (1994) however noted that the change in conductivity with dose rate is significant. Compared to silicon diodes which show both energy and dose rate dependence, Hoban et al. (1994) noted that as the diamond detector is only dose rate dependent, a correction based on dose rate alone may be applied to its response to radiation. In a recent study presented by this group (Ade et al., 2013), the results demonstrated that the performance (in terms of stability of response) of a HPHT diamond was better compared to its polycrystalline CVD counterparts, and this was related to its fewer defect levels. Since Fowler's theory of dose rate dependence is also influenced by electron traps, the relevance of the present study was therefore to systematically investigate and evaluate the dependence of Δ of synthetic diamond detectors of various types on therapy electron energy and its influence in relative electron dosimetry with the aim of selecting a crystal for dosimetry.

Although Planskoy (1980) examined three natural diamond detectors and concluded from his study that Δ is independent of variables such as bias voltage, beam energy and modality for the tested diamonds, his conclusion was based on a composite curve drawn after averaging data for all beam energies and bias voltages used. In addition, many studies based on this aspect have examined only one to about three diamonds of the same type such as Planskoy (1980), Laub et al. (1997,1999), Björk et al. (2000), Fidanzio et al. (2004), Cirrone et al. (2006) and Górka et al. (2008). Thus the amount of experimental data available in literature might be limited for the influence of beam energy on Δ. This study therefore considers up to nine synthetic diamond crystals of various types and Δ was computed for different electron energies unlike the work by Planskoy (1980).

2. Experimental

Initially nine commercially available synthetic diamond crystals of various types with sizes ranging between 5×5 mm^2 and 10×10 mm^2 and thickness of either 0.5 mm or 1.0 mm were investigated. These included one single crystal high-pressure, high-temperature (HPHT) sample (HPHT1) and eight polycrystalline CVD (chemical vapour deposition) diamond crystals of detector grade (DG) and optical grade (OG) qualities (Table 1). The opposite surfaces of each of the diamonds were metallised as reported in a previous study (Ade et al., 2012) to provide the necessary ohmic contacts for voltage biasing and acquisition of the ionization signal. For quality control the diamond crystals were characterised using Raman spectroscopy and electron spin resonance (ESR) in similar procedures as reported by Assiamah (2004). Subsequent to the results on the influence of beam energy on the Δ values and stability of response, a small-size HPHT diamond (HPHT2) of dimensions $\approx 3 \times 3 \times 1$ mm^3 was deliberately selected for evaluation of its potential application in small radiation fields.

Dosimetric measurements were conducted with 6–14 MeV electron therapy beams produced by a clinical linear accelerator (Siemens Primus) at the Charlotte Maxeke Johannesburg Academic Hospital. Each diamond crystal was encapsulated in a probe

Table 1
Synthetic diamond crystals and their parameters.

Diamond Specimen	Crystal dimensions (mm^3)	Raman FWHM (cm^{-1})	ESR [N_s] ($\pm$2 ppm/ppb)	Δ Values ($\pm$0.02)		Percent difference	Mean Δ values ($\pm$0.02)	Percent (%) deviation from unity of mean values
				7 MeV	12 MeV			
HPHT1	$8.0 \times 6.4 \times 1.0$	2.27 ± 0.04	130 Ppm	0.95	0.95	0.0	0.95	5.0
OG A1	$5.0 \times 5.0 \times 1.0$	2.56 ± 0.09	43 ppm	0.94	0.86	8.9	0.90	10.0
OG A2		2.42 ± 0.09	71	0.92	0.84	9.1	0.88	12.0
DG A1	$5.0 \times 5.0 \times 1.0$	2.60 ± 0.13	3.5 ppm	0.91	0.87	4.5	0.89	11.0
DG A2		2.55 ± 0.09	5.0	0.82	0.79	3.7	0.81	19.0
DG A	$10 \times 10 \times 0.5$	2.34 ± 0.08	20 ppb	1.02	1.01	1.0	1.02	2.0
DG B1		2.37 ± 0.11	10	1.01	0.99	2.0	1.00	0.0
OG A	$10 \times 10 \times 1.0$	2.42 ± 0.11	50 ppb	1.01	0.96	5.1	0.99	1.0
OG D		2.46 ± 0.25	46	1.03	1.01	2.0	1.02	2.0

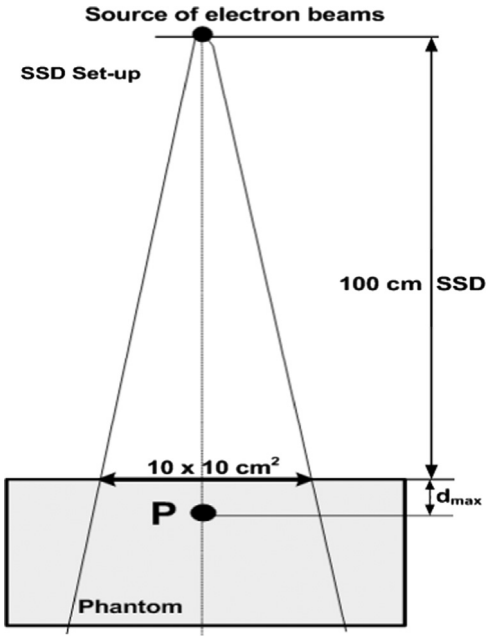

Fig. 1. Irradiation geometry for dose rate dependence and relative dose distribution measurements. *P* is the effective point of measurement of the detector.

housing as described in a previous study (Ade et al., 2012) and the probe, placed in its customised Perspex phantom was connected to a PTW-Freiburg UNIDOS E electrometer system operated manually in the 'charge' mode. An applied electric field of 1000 V/cm was used. The probe, constructed of entirely tissue-equivalent material with a Perspex body can hold diamond crystals of various sizes up to 10×10 mm^2 in size, has different exposure geometries[1] ('edge-on' and 'flat-on') for impinging radiation and is compatible with commercially available electrometer systems.

The linearity between detector's response and absorbed dose rate was investigated by positioning the diamond probe in the Perspex phantom at the depth of dose maximum, d_{max}, equal to 1.3 cm and 2.5 cm for 7 MeV and 12 MeV electron beams, respectively as illustrated in Fig. 1. The dose rate was varied by changing the source-to-phantom surface distance (SSD) from 100 cm to 145 cm, while keeping the radiation field size, defined by an electron applicator at the phantom surface, constant at 10×10 cm^2. The average absorbed dose rates at each position of the diamond probe at the depths of measurement were measured with a 0.6 cm^3 Farmer-type chamber connected to Unidose E electrometer biased at +400 V. The measured dose rates varied between 1.0 Gy/min and 3.07 Gy/min. Depth-dose distributions taken with the diamond probe on exposure to 6 MeV, 12 MeV and 14 MeV electron beams were measured in similar procedures as reported in a recent study (Ade et al., 2013). For all the measurements, 50 monitor units (corresponding to an absorbed dose value of 0.5 Gy) were used to irradiate each detector. The reference conditions were as follows: a dose rate of 3.0 Gy/min defined at d_{max} in water phantom at 100 cm SSD with an applicator defined field of 10×10 cm^2. Reference detectors (a Dosimetry Diode E (Type T60017), 0.055 cm^3 Markus (Type 23343) and 0.6 cm^3 Farmer-type ion chambers) recommended for the dosimetry of high-energy electron therapy beams were used for comparative measurements.

[1] In the 'flat-on' geometry, the incident radiation is normal to the face of the diamond wafer, which has a larger area but with reduced attenuation depth. In the 'edge-on' geometry (suitable for low-energy X-rays), the incident radiation is normal to a wafer edge which has a smaller area but with greater attenuation depth.

The Diode E, with a sensitive volume of 0.03 mm^3 was used for depth-dose and output (or cut-out) factor measurements in small field sizes due to its higher accuracy compared to large volume ion chambers. The diode was connected to the Unidos E electrometer and operated at 0 V bias. Small square fields ranging from 1×1 cm^2 to 5×5 cm^2 were defined by lead cut-outs inserted in a 10×10 cm^2 open applicator defined field. The output factor was then defined as the ratio of the charge reading at d_{max} for the small field of interest to that at d_{max} for the 10×10 cm^2 reference field.

3. Results and discussion

3.1. Raman spectroscopy and ESR

For quality control, Raman spectroscopy is a technique commonly used for the determination of the phase (sp^2 or sp^3) and crystalline qualities of diamond crystals. A single sharp line (the diamond Raman peak) at 1332 cm^{-1} (which is a feature characteristic of the diamond structure (sp^3 phase)) having a full-width-at-half-maximum (FWHM) of about 2 cm^{-1} is observed in natural diamond (Faggio et al., 1999). In synthetic diamond, the line can be shifted due to internal stress and the width is generally much broader than natural diamond due to crystal defects or non-homogeneous distribution of stress (Faggio et al., 1999). An additional feature commonly present in the Raman spectrum of synthetic diamond is a broad band around 1500 cm^{-1} due to the presence of graphitic inclusions (sp^2-bonded carbon). Three quality factors are thus introduced for the quantitative comparison of different diamond crystals (Faggio et al., 1999): the Raman FWHM of the diamond Raman peak for crystalline quality, the shift of the peak position for internal stress state and the ratio of the non-diamond to the diamond component for phase purity.

The Raman spectra recorded for all the crystals showed only the characteristic diamond Raman peak at 1332 cm^{-1} over a linear background with no evidence of non-diamond component confirming phase purity. The measured Raman line widths (FWHM) ranged from 2.27 ± 0.04 cm^{-1} to 2.60 ± 0.13 cm^{-1} (Table 1). Although the diamond Raman line position and broadening is known to be related to both strain and defect concentration, the results (to be reported in the future) of our investigation suggest that the broadening in the examined crystals could be related to defects alone. This is also reflected in the diamond Raman peak position at 1332 cm^{-1} which is not shifted from that of pure natural diamond. Hence, as the Raman line width is an indication of the crystalline quality of diamond, i.e. a measure of the concentration of defects within the crystal, this implies the HPHT sample having the smallest width (2.27 cm^{-1}) is the least defective crystal whereas DGA1 having the broadest width (2.60 cm^{-1}) is the most defective crystal.

Nitrogen (N), which is a commonly observed impurity in diamond, affects its radiation detection properties. It is known that *N* in diamond can occur in different forms, the dominant form in synthetic diamond being single substitutional *N* (N_s) atoms. Nam et al. (1991) established that N_s is responsible for many performance characteristics of diamond radiation detectors. ESR is a useful and non-destructive tool for the accurate determination of the N_s levels. The concentrations of N_s ($[N_s]$) for the diamond crystals determined by ESR are displayed in Table 1. The reason for the much higher levels of N_s impurities of the HPHT and OG crystals compared to the DG samples has been explained in a previous study (Ade et al., 2013).

3.2. Dose rate dependence

The linear response of a diamond detector with absorbed dose and dose rate are principal requirements in radiotherapy (Marczewska et al., 2007). As detector current *I* is proportional to conductivity σ for

plane-parallel geometry where a voltage V is applied across a specimen of length L and cross-sectional area A (Fowler, 1966), Eq. (1) can be expressed as (Laub et al., 1997; Bruzzi et al., 2000; Cirrone et al., 2006):

$$I \sim D_r^{\Delta} \tag{2}$$

where I is the ionization current (corrected for leakage) measured by a diamond detector. The current values determined (given by the measured charge divided by the irradiation time) as a function of dose rate with the diamond detectors were then fitted by Eq. (2) using logarithmic (or power law) plots to obtain the Δ values which ranged between 0.79 and 1.03 for all the diamond crystals tested in this study. In comparison, Δ values between 0.49 and 1.16 have been reported by other researchers (Planskoy, 1980; Ramkumar et al., 2001; Fidanzio et al., 2004; Laub et al., 1997, 1999; Cirrone et al., 2006; Björk et al., 2000; Hoban et al., 1994; Marczewska et al., 2007; Górka et al., 2008; Lansley et al., 2009; Tranchant et al., 2008; Betzel et al., 2010; Schirru et al., 2010; Mohamed Abdel-Rahman et al., 2012) for both natural and synthetic diamonds, and a value of 1.20 has been reported for a silicon diode (Buttar et al., 2000). In the present study, a Δ value of 1.02 was obtained for the Dosimetry Diode E for a 12 MeV electron beam. The large difference (16.2%) between this value and the value of 1.2 reported by Buttar et al. (2000) could be related to different doping levels of Si diodes as the doping level varies for different manufacturers (Haryanto et al., 2002).

3.2.1. Dependence of Δ on electron energy

Displayed in Table 1 are values of the dose rate dependence correction factors, Δ, determined for 7 MeV and 12 MeV electron beams for the nine diamond detectors. It can be observed for the CVD diamonds that Δ decreases with increasing electron energy. A similar trend has also been observed by Górka et al. (2008) for 6 MV and 18 MV photon beams for a CVD diamond detector. In order to quantify and evaluate the variation of Δ with electron energy, the percentage differences between the Δ values determined for the two energies were calculated for each diamond detector. Although the AAPM TG 40 (American Association of Physicists in Medicine, AAPM TG 40, 1994) has set a tolerance value within 3% for the dosimetry of both electron and photon beams in order to attain the ICRU overall 5% uncertainty of absorbed dose delivery in radiotherapy (International Atomic Energy Agency, 1997; International Commission on Radiation Units and Measurements, 1976), $\pm 2\%$ tolerance was considered for the present study. With this figure and considering a Δ value of 1.02, the ICRU $\pm 5\%$ margin can be achieved even when the dose rate is changed by a factor of 10.

As Table 1 shows while the Δ values of HPHT diamond (HPHT1) are completely independent of the two electron energies, those of three CVD diamond detectors (DGA, DGB1 and OGD) of a given class (i.t.o. [N_s]—ppb level) were found to vary between the electron energies within 2%. In addition a very strong variation ($\approx 9\%$) was observed for two OG crystals (OGA1 and OGA2) of another class ([N_s] in ppm). The small dependence of Δ ($\leq 2\%$) on electron energy observed with some of the crystals could be attributed to the negligible energy dependence of diamond to radiation (this however depends on crystal purity or type) within a given energy range (Laub et al., 1997; van der Merwe and Keddy, 1999; Schirru et al., 2010; Ade et al., 2012). In comparison, while Planskoy (1980) tested three selected natural diamond detectors and concluded that Δ is independent of beam energy and modality, Górka et al. (2008) on the other hand found a difference of 8% between the Δ values obtained for different photon beam qualities. This study has also found a significant variation of Δ between 7 MeV and 12 MeV electrons for some diamond crystals and the variation could be as high as 9% depending upon crystal type and class. The reason for the variation of Δ with electron energy observed for the CVD diamond crystals in general and for OGA1 and OGA2 in particular, being OG crystals of a given class (i.t.o [N_s] —ppm level) is a subject for further investigation.

The results of Raman width and [N_s] presented in Table 1 serve to provide an indication of how defect levels influence the Δ values of the diamond detectors where it is demonstrated for each class of the CVD diamond crystals that there is a tendency for Δ to decrease with increasing [N_s] and increase with increasing Raman broadening which reflects defect concentration as explained in Section 3.1. Although the relationship between nitrogen level and trap concentration is not yet known for CVD diamonds unlike HPHT diamonds, the observed inverse link between [N_s] and Raman width could be related to the observation made by Nam et al. (1991) for HPHT diamonds that crystals with high [N_s] tend to have much lower trap concentrations than those synthesized with lower nitrogen levels. It can be seen that whereas the HPHT diamond has the highest [N_s] of 130 ppm and lowest Raman width of $2.27 \pm 0.04\ \mathrm{cm}^{-1}$, DGA1 with the lowest [$N_s$] of 3.5 ppm has the broadest Raman width of $2.60 \pm 0.13\ \mathrm{cm}^{-1}$.

The more defective nature (i.t.o. Raman width) of the CVD diamonds could be due also to their polycrystalline structure compared to the HPHT diamond as polycrystalline materials have high defect densities due to their non-homogeneous structure (Bergonzo et al., 2007). In particular, it is well known that polycrystalline CVD diamond films have a high concentration of grain boundaries (Balducci et al., 2005) where electron traps are concentrated or in more defective regions (Manfredotti et al., 2006). It is observed that DGA1 being the most defective crystal does not have an overall better performance (Δ values are further from unity) compared to the other crystals. It is likely that a large concentration of active traps could be present in this crystal that could cause it to suffer from polarization effects as it has been pointed out that diamond crystals suitable for dosimetry need a certain concentration of impurities, but that too much would cause the crystal to suffer from polarization effects (Laub et al., 1999).

The results of this study have therefore demonstrated for the CVD diamonds that Δ may vary with electron energy and it could be influenced by defect levels. The influence of Δ in relative electron dosimetry is examined in Section 3.3.

3.2.2. Dependence of the linearity index Δ on dose rate

As noted by Planskoy (1980), theory predicts that Δ will decrease with increasing dose rate since conductivity is governed by dose rate as explained in Section 1 although Laub et al. (1999) could not observed this effect. In the present study, Table 2 demonstrates this effect within the measuring accuracy of 2% for 12 MeV electrons. For instance, in the range of dose rates considered in this study, Δ changes from 0.95 to 1.01 when the dose rate was varied from 3.07 to 2.25 Gy/min for the HPHT diamond. Since the energy of electrons strongly depends on depth in a phantom material, the variation of Δ with dose rate implies that a single value of Δ should not be applied for the correction of percent depth-dose data. In particular, the dose rates were also measured as a function of depth in the Perspex

Table 2
Change in Δ values as a function of dose rate for a 12 MeV electron beam.

Δ at dose rate									
Values up to:	**HPHT1**	**OGA1**	**OGA2**	**DGA1**	**DGA2**	**DGB1**	**DGA**	**OGD**	**OGA**
2.25 Gy/min	1.01	0.94	0.88	0.92	0.84	1.01	1.02	1.01	1.00
3.07 Gy/min	0.95	0.86	0.84	0.87	0.79	0.99	1.01	1.01	0.96
Percentage difference	6.1	8.9	4.7	5.6	6.1	2.0	1.0	0.0	4.1

phantom at a constant SSD of 100 cm using 12 MeV electrons. The dose rate varied from 0.77 Gy/min at 50 mm depth to 3.07 Gy/min at d_{max} of 25 mm.

3.3. Influence of Δ in relative electron dosimetry

3.3.1. Depth-dose distributions measurements in a 10×10 cm^2 field size

Since dose rate from a point source decreases with distance according to the inverse square law, the dose rate dependence of a diamond detector should manifest itself in a percent depth-dose (PDD) curve which would show a shallower fall-off, appearing to give an over response as depth increases (Hoban et al., 1994). It has been pointed out that the shallower fall-off (i.e., an over-estimation) in a PDD curve could be removed by the application of the following (Laub et al., 1997) expression:

$$D_{rel}(d) \rightarrow [D_{rel}(d)]^{1/\Delta} \quad (3)$$

where $D_{rel}(d)$ is the relative dose measured with a diamond detector at the depth of measurement d, and Δ is the dose rate correction factor referred to in Eqs. (1) and (2). Depth dose distributions are usually renormalized to dose maximum after performing the operation in Eq. (3) (Björk et al., 2000).

To evaluate the influence of the linearity index Δ in relative dosimetry, percent depth-ionization data were measured with the diamond probe and comparisons were made with reference ion chambers. The comparison considered the relative dose data measured with the tested diamond detectors with and without dose rate dependence corrections using the values of Δ obtained for the 7 MeV and 12 MeV electron beams (Table 1). The influence of Δ in depth-dose measurements in this section was investigated using DGA1 and OGD which have mean Δ values of 0.89 and 1.02, respectively. The variation of Δ with dose rate was also considered for the case of DGA1 (Table 2).

Fig. 2(a) and (b) shows depth ionization data taken with DGA1 and OGD for exposure to a 14 MeV electron beam, respectively, in a 10×10 cm^2 field compared with depth dose data generated from measurements with a Markus chamber (with its response corrected for stopping power ratios and other effects). Shown in the figures are the uncorrected and corrected data based on dose rate dependence for the diamond detectors. While the uncorrected data for DGA1 shows a very pronounced shallower fall-off with depth (clearly visible in Fig. 3(a)), such is not the case with OGD.

The deviations (relative differences) between the percent depth-doses measured with the diamond detectors and the ion chamber were analysed, as illustrated in Fig. 3(a) and (b) for DGA1 and OGD, respectively. The figures illustrate that while the dose rate dependence correction significantly improves the performance of DGA1 (Fig. 3(a)), it has a smaller effect on the performance of OGD (Fig. 3(b)). This could be attributed to the very strong dose rate dependence of DGA1 (mean $\Delta = 0.89$; percent deviation from linearity = 11%) compared to OGD (mean $\Delta = 1.02$;

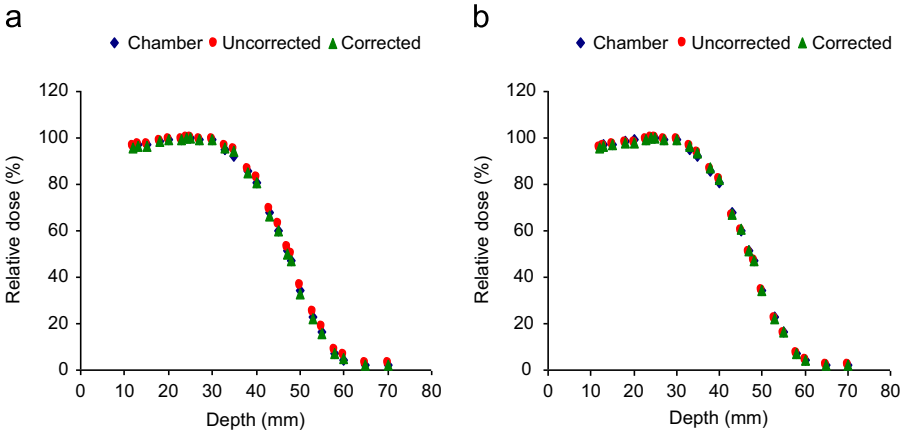

Fig. 2. Normalised depth-dose ionization in Perspex obtained with (a) DGA1 and (b) OGD CVD diamond detectors compared with energy-corrected depth-dose curves generated from measurements with a 0.055 cm^3 Markus ion chamber, for exposure to a 14 MeV electron beam in a 10×10 cm^2 field. The figures show the data before and after correction for dose rate dependence.

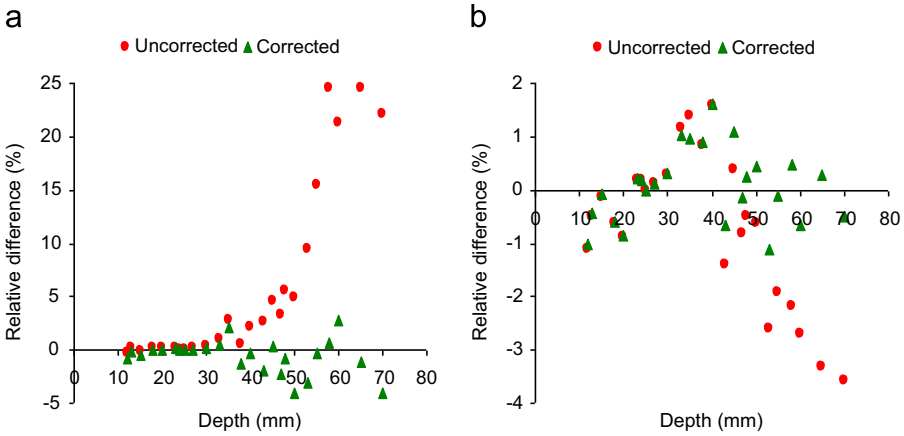

Fig. 3. Relative difference between the depth-doses measured with (a) DGA1 and (b) OGD CVD diamond detectors and an ion chamber, for exposure a 14 MeV electron beam in a 10×10 cm^2 field. The figures show the data before and after correction for dose rate dependence.

percent deviation from linearity=2%—Table 1). It could be observed that without dose rate dependence corrections, the relative dose difference between DGA1 and the ion chamber is $>2\%$ at depths beyond 40 mm and could reach 25% in the region of the bremsstrahlung tail. When the dose rate dependence corrections are applied, the relative dose difference is within 3% at depths between the build-up region and 50 mm. For the case of OGD, without the correction, the relative dose difference is only $>2\%$ at depths beyond 50 mm in the region of the bremsstrahlung tail with a maximum of $<4\%$. With the application of the correction factor, the dose difference is of the order of 1%. Similar results for a 6 MeV electron beam were also obtained for DGA1 (using a 0.6 cm^3 Farmer-type ion chamber as reference) where the relative dose difference without a correction could reach 20% at depths between 25 mm and 30 mm.

A systematic evaluation of the influence of each of the correction factors (mean Δ, 7 and 12 MeV's Δs) in combination with the variation of Δ with dose rate on the measured data showed that while those for OGD had a marginal dose variation ($\leq 0.5\%$) from each other, a significant variation ($>2\%$) was observed for those of DGA1 between depths 35 mm and 40 mm. Such an observation is probably due to the strong dependence of DGA1's Δ values on electron energy compared to those of OGD (Table 1).

The presented results have therefore highlighted that (i) a very strong dose rate dependence of a radiation dosimeter would result to a very large deviation of the measured dose if the necessary corrections are not applied and (ii) a single correction factor could be used for different electron energies at a given depth of measurement if the variation of Δ with beam energy is within 2%.

3.3.2. Measurements in small electron fields

Based on the results (related to defect levels) of a systematic study of the instability of response (Ade et al., 2013) and the influence of electron energy on the Δ values of the synthetic diamonds of various types, a small-size HPHT diamond (HPHT2) was selected and evaluated for its potential application in small radiation fields. A crystal of dimensions $\approx 3\times3\times1$ mm^3 was considered due to the relationship between detector and field size in small fields (Wuerfel, 2013). Two CVD diamond crystals (DGB1 and OGD) were also tested in a 5×5 cm^2. In all the measurements, 12 MeV electrons were used. Fig. 4 illustrates the deviations between the percent depth-doses measured with the diamond detectors (DGB1 and OGD) and the Diode *E* for the measurements in a 5×5 cm^2 field. The figures show that without corrections based on dose rate dependence, the relative dose difference for DGB1 (Fig. 4(a)) is only $>2\%$ at depths beyond 45 mm with a maximum difference of 3%, whereas for OGD (Fig. 4(b) the difference is $>2\%$ from 35 mm depth. When the dose rate corrections are applied, the relative difference for DGB1 is within 1% while that for OGD is of the order of 2%.

Illustrated in Fig. 5(a) is the percent depth ionization data measured with the HPHT2 diamond in a 3×3 cm^2 field compared with the data obtained with Diode E. The relative differences between the depth-doses measured with the two detectors are reported in Fig. 5(b). The figure demonstrates that the data measured with the diamond probe are in excellent agreement when compared to measurements taken with the diode. Without a dose rate dependence correction, maximum dose differences of the order of

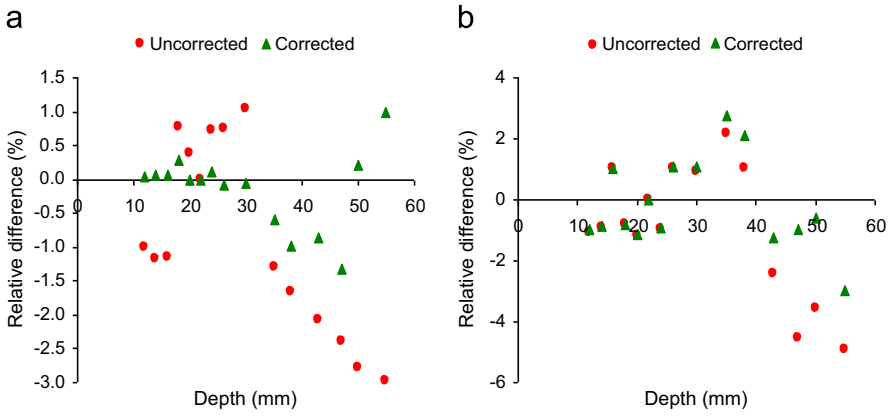

Fig. 4. Relative difference between percent depth-doses measured with (a) DGB1 and (b) OGD CVD diamond detectors and Diode E, for exposure to a 12 MeV electron beam in a 5×5 cm^2 field. The figures show the data before and after corrections for dose rate dependence.

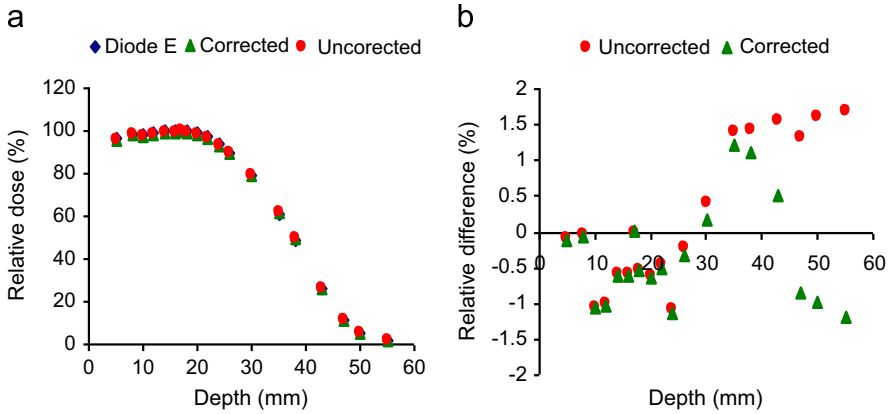

Fig. 5. (a) Normalised depth-dose ionization taken in Perspex with the HPHT2 diamond compared with depth-dose data as measured with Diode E for exposure to a 12 MeV electron beam in a 3×3 cm^2 field. (b) The relative difference between the depth-doses measured with the LSD and Diode E. The figures show the data before and after correction for dose rate dependence.

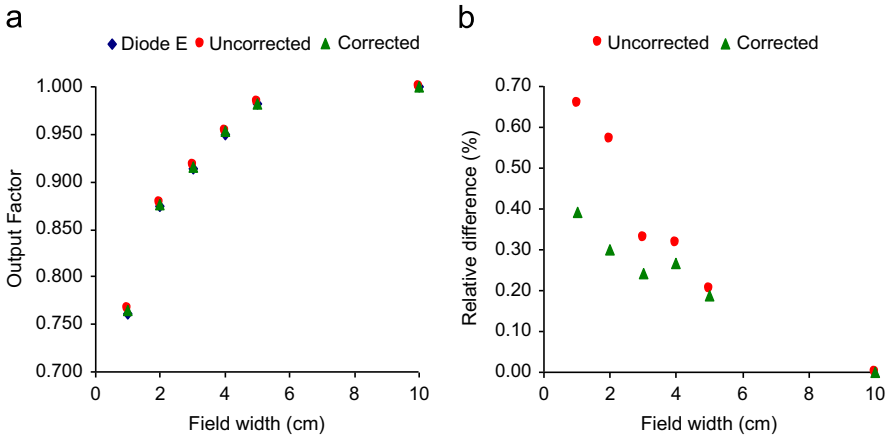

Fig. 6. (a) Output factors measured with HPHT2 diamond and Diode E for exposure to a 12 MeV electron beam for square field sizes between 1×1 cm^2 and 10×10 cm^2. (b) The relative difference between the output factors measured with the two detectors. The figures show the data before and after correction for dose rate dependence.

1.5% could be observed. With a dose rate dependence correction (using a Δ value of 0.99 ± 0.02 obtained in a 10×10 cm^2 with a 1% variation with dose rate), the dose difference is of the order of 1%.

The output factors measured with HPHT2 detector and Diode E are reported in Fig. 6(a) for the square field sizes from 1×1 cm^2 to 10×10 cm^2. Maximum deviations (Fig. 6(b)) of the order of 0.5% could be observed even without dose rate dependence corrections.

A synthetic single crystal diamond detector (SCDD) has recently been fabricated and described as a radiotherapy dosimeter (Pimpinella et al., 2012; Ciancaglioni et al., 2012). The fabrication technique consisted of the deposition of an intrinsic diamond film on a commercial $4 \times 4 \times 0.4$ mm^3 HPHT Ib single crystal diamond. In the present study, it has been demonstrated that with careful selection of a suitable diamond specimen, results (i.t.o. relative dose measurements) comparable to those reported for the SCDD could be achieved, but with a higher sensitivity value of 313 nC Gy^{-1} mm^{-3} determined in a smaller field under a 6 MV photon irradiation. In addition, unlike most commercial diamond detectors which require daily pre-irradiation, the results of our recent study showed however that once the response of our diamond probe has been stabilised and properly shielded from ambient light, pre-irradiation prior to each measurement is not required (Ade et al., 2013).

4. Conclusions

In this study, the dose rate dependence of nine synthetic diamond detectors of various types was investigated. The linearity index, Δ, was determined for each diamond detector and evaluated for its dependence on therapy electron energy. Within a 2% tolerance Δ was found to be independent of electron energy within the clinical range for the HPHT diamond and three CVD diamond detectors of a given class. On the other hand a very strong dependence of Δ with electron energy was observed for two OG CVD diamonds of another class. The Δ values were found to vary with dose rate in accordance with theory. The results of this study have therefore highlighted that depending on crystal type or class Δ could vary strongly with electron energy. Therefore for such diamond crystals Δ should be determined for all electron energies of clinical interest.

The influence of Δ in relative electron dosimetry was also evaluated for field sizes between 1×1 cm^2 and 10×10 cm^2. The results show that with careful selection of a suitable diamond specimen, relative dose distributions measured with the diamond probe would compare favourably with those obtained with reference dosimeters within or of the order of 1% even without dose rate dependence corrections for the probe. The results demonstrated also that with a crystal that shows a significant dose rate dependence, appropriate corrections can be applied to its response to obtain good results.

A study aimed at evaluating the dependence of Δ on photon energy and radiation type and the influence of defect/impurity levels on the Δ values of the diamond crystals is under investigation. In addition, another study aimed at characterising the performance of the diamond probe using various diamond crystals in both small and very small photon fields ($< 1 \times 1$ cm^2) is also under investigation.

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References

American Association of Physicists in Medicine, AAPM TG 40, 1994. Med. Phys. 21, 581.
Ade, N., Nam, T.L., Assiamah, M., 2012. Radiat. Phys. Chem. 81, 232.
Ade, N., Nam, T.L., Mhlanga, S.H., 2013. Radiat. Phys. Chem. 86, 42.
Assiamah, M., 2004. PhD Thesis. University of the Witwatersrand, Johannesburg.
Balducci, A., Marinelli, M., Milani, E., Morgada, M.E., Prestopino, G., Scoccia, M., Tucciarone, A., Verona-Rinati, G., 2005. Appl. Phys. Lett. 87, 222101.
Bergonzo, P., Tromson, D., Descamps, C., Hamrita, H., Mer, C., Tranchant, N., Nesladek, M., 2007. Diamond Relat. Mater. 16, 1038.
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.
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.
Buttar, C.M., Airey, R., Conway, J., Hill, G., Ramkumar, S., Scarsbrook, G., Sussmann, R.S., Walker, S., Whitehead, A., 2000. Diamond Relat. Mater. 9, 965.
Ciancaglioni, I., Marinelli, Marco, Milano, E., Prestopino, G., Verona, C., Verona-Rinati, G, 2012. Med. Phys. 39, 4493.
Cirrone, G.A.P., Cuttone, G., Lo Nigro, S., Mongelli, V., Raffaele, L., Sabini, M.G., 2006. Nucl. Phys. B (Proc. Suppl.) 150, 330.
Das, I.J., 2009. Diamond detector. In: Rogers, D.W.O., Cyglar, J.E. (Eds.), Clinical Dosimetry Measurements in Radiotherapy. Medical Physics Publishing, Madison, WI, pp. 891–912
Faggio, G., Marinelli, M., Messina, G., Milani, E., Paoletti, A., Santangelo, S., Verona-Rinati, G., 1999. Microsyst. Technol. 6, 23.
Fidanzio, A., Azario, L., Viola, P., Ascarelli, P., Capelli, E., Conte, G., Piermattei, A., 2004. Nucl. Instrum. Methods Phys. Res. A 524, 115.
Fowler, J.F., 1966. Solid state electrical conductivity dosimeters. Radiation Dosimetry, vol. II. Academic, New York p. 291.
Górka, B., Fernandez-Varea, J.M., Panettieri, V., Nilsson, B., 2008. Nucl. Instrum. Methods Phys. Res. A 593, 578.

Haryanto, F., Fippel, M., Laub, W., Dohm, O., Nüsslin, F., 2002. Phys. Med. Biol 47, N133.
Hoban, P.W., Heydarian, M., Beckham, W.A., Beddoe, A.H., 1994. Phys. Med. Biol. 39, 1219.
International Atomic Energy Agency, Quality Assurance in Radiotherapy, IAEA-TECDOC-989, IAEA, Vienna, 1997.
International Commission on Radiation Units and Measurements, Determination of Absorbed Dose in a Patient Irradiated by Beams of X or Gamma Rays in Radiotherapy Procedures, ICRU Report 24, ICRU, Bethesda, 1976.
Lansley, S.P., Betzel, G.T., Meyer, J., 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.
Manfredotti, C., Fizzotti, F., Garino, Y., 2006. Diamond Relat. Mater. 15, 1362.
Marczewska, B., Kupriyanov, I., Pal'Yanov, Yu., Nowak, T., Olko, P., Rębisz, M., Waligórski, M.P.R., 2007. Diamond Relat. Mater 16, 191.
Mohamed Abdel-Rahman, A.E., Lohstroh, A., Jayawardena, I., Henley, S.J., 2012. Diamond Relat. Mater 22, 70.
Nam, T.L., Karfunkel, U., Keddy, R.J., Every, A.G., 1991. Radiat. Eff. Def. Solids 116, 233.
Pimpinella, M., Ciancaglioni, I., Consorti, R., Di Venanzio, C., Guerra, A.S., Petrucci, A., Stravato, A., Verona-Rinati, G., 2012. Metrologia 49, S207.
Planskoy, B., 1980. Phys. Med. Biol. 25, 519.
Ramkumar, S., Buttar, C.M., Conway, J., Whitehead, A.J., Sussman, R.S., Hill, G., Walker, S., 2001. Nucl. Instrum. Methods Phys. Res. A 460, 401.
Schirru, F., Kupriyanov, I., Marczewska, B., Nowak, T., 2008. Phys. Status Solidi A 205, 2216.
Schirru, F., Kisielewicz, K., Nowak, T., Marczewska, B., 2010. J. Phys. D: Appl. Phys 43, 265101.
Tranchant, N., Tromson, D., Descamps, C., Isambert, A., Hamrita, H., Bergonzo, P., Nesladek, M., 2008. Diamond Relat. Mater 17, 1297.
van der Merwe, D.G., Keddy, R.J., 1999. Radiat. Phys. Chem. 54, 325.
Wuerfel, J.U., 2013. Med. Phys. Int. J. 1, 82.