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Zero-bias operation of polycrystalline chemically vapour deposited diamond films for Intensity Modulated Radiation Therapy

M. Bruzzi [a,*], C. De Angelis [b], M. Scaringella [a], C. Talamonti [c], D. Viscomi [b], M. Bucciolini [c]

[a] INFN-Firenze and Dip. Energetica, Università degli Studi di Firenze, Via S. Marta 3, 50139 Firenze, Italy
[b] Dipartimento di Tecnologie e Salute, Istituto Superiore di Sanita` and INFN, Viale Regina Elena 299, 00161 Rome, Italy
[c] Università degli Studi di Firenze, Dip. Fisiopatologia Clinica, and INFN-Firenze, Viale Morgagni 85, 50134 Firenze, Italy

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ABSTRACT

A detailed investigation of the performance as a dosimeter of state-of-art polycrystalline CVD (pCVD) diamond detectors operated in photovoltaic regime for applications in clinical radiotherapy has been carried out with conventional 6–10–18 MV X-photons, as well as with a 10 MV photon Intensity Modulated Radiation Therapy (IMRT) beam from a linear accelerator. Our results show that the performances of a pCVD diamond dosimeter improves dramatically when operated in null-bias conditions. Main improvements with respect to operation with an external voltage applied are: a reduced pre-irradiation dose; an excellent time stability, characterised by standard deviations less than 0.5%; a rise-time comparable to that of commercial reference dosimeters; a linearity with dose proven over three decades; a reduced deviation from linearity of the current vs. dose–rate curve, output factors comparable to that of commercial reference dosimeters. These results represent a significant step towards clinical applications as IMRT with synthetic polycrystalline CVD diamond films.

1. Introduction

Conformal radiotherapy, as X-photons Intensity-Modulated radiotherapy (IMRT) [1], represents a major development in clinical treatments. With IMRT it is possible to deliver a highly conformed dose to an irregularly shaped tumor volume while sparing the surrounding healthy tissue as much as possible. The delivered dose can be precisely modulated and a steep dose gradient between the tumour and the healthy organs at risk is obtained [2–5]. The effective implementation of intensity modulated radiotherapy represents a major challenge for the next generation of dosimeters used for real-time dose monitoring and dose pre-treatment verifications. Large fields (typically up to 20×20 cm^2) must be covered by bidimensional dosimeters as active matrixes monitoring point-to-point and in real-time the absorbed dose [6]. High-dose gradients, elevated variability in space and time of the dose rate and of the beam spectral composition will require detectors of high sensitivity, with fast and stable response, independent of energy and dose rate.

Many of the outstanding chemical and physical properties of diamond make this material potentially attractive in this respect. Due to its atomic and mass numbers, $Z=6$ and $A=12$, comparable to those of soft tissue, it is considered an almost tissue-equivalent material. This is an important advantage against e.g. silicon, being diamond response almost energy independent for a wide energy range of photons [7]. As a consequence, no energy dependent factor has to be applied to the diamond response when used with different energies, while Si dosimeters suffer of this drawback. Being non-toxic and chemically stable against all body fluids, diamond can be used *in vivo*. Dosimeters made with natural diamond are already commercially available [8] but extremely expensive and rare due to the difficulty in selecting stones with the proper dosimetric characteristics. Polycrystalline Chemical Vapor Deposited (pCVD) diamond films have been proposed in their stead, due to the potential low cost of this material and much effort has been devoted worldwide to produce pCVD diamond films with properties suitable for dosimetry in radiotherapy. At present, data reported in literature show for pCVD diamond a behavior similar to that of natural diamond concerning many dosimetric properties such as linearity and dose–rate dependence, nevertheless, a major drawback of pCVD detectors concerns the slow rise and decay times and the poor response stability [9]. It was shown that an improved stability can be obtained by pre-irradiating the pCVD diamond films with a high fluence of fast neutrons [10], the method was demonstrated capable of improving the performance of low-to-medium crystalline quality pCVD diamond dosimeters to the level of the highest-purity pCVD diamond films [11]. Nonetheless, it is well known that in this high-quality material there is still a significant amount of native deep and shallow defects, which can be active at room temperature for trapping charges, thus affecting the stability of response of the devices [12]. A study recently performed on pCVD detectors based on state of-the-art polycrystalline diamond films

* Corresponding author. Tel.: +39 0 554796449; fax: +39 0 554796342.
E-mail address: Mara.Bruzzi@unifi.it (M. Bruzzi).

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demonstrated that this high quality pCVD diamond is still unsuitable for IMRT applications [9]. The response dynamics of an on-line dosimetric device must be sufficiently fast in order to follow in real time the changes of intensity of the radiation beam. This requires a time of response of the order of 1 s for IMRT applications, where typical irradiation rates are of 200–400 MU/min (1 Monitor Unit = 1 MU usually corresponds to 1 cGy in standard irradiation conditions, one monitor unit is delivered approximately every 0.3 s) [13]. The dynamic response of high-quality pCVD diamond dosimeters, still affected by trapping dynamics, show instead rise and decay times typically of the order of a few seconds [9].

It is well known that defects in pCVD diamond films are mainly located at grain boundaries [14]. So, a possible way to reduce the effect of native defects on response stability is to use single-crystal diamond films. High electronic quality homoepitaxial diamond can be grown on high temperature high pressure (HPHT) diamond substrates by a CVD method, to produce synthetic single crystal CVD diamond films (scCVD) [15]. Recently, it has been demonstrated that dosimeters made with such scCVD diamond films exposed to X-photons are characterised by a fast, reproducible and stable dynamics, making them suitable for IMRT applications [16–18].

Crucial disadvantages of scCVD diamonds for a large scale bidimensional application in clinical dosimetry are nonetheless the wafer maximum size achievable (approximately 1 cm diameter due to the limited size of the HPHT diamond seed), and the exceptionally high cost of the material. For these reasons, large-scale bidimensional dosimeters, as those needed for IMRT [6], cannot be foreseen with this kind of material. On the contrary, current state-of-art technology demonstrates that segmented devices can be processed on high quality polycrystalline CVD diamond 4–6″ wafers [19]. It is then of crucial importance to continue pursuing the study of pCVD diamond in view of an application in conformal radiotherapy techniques as IMRT.

To further improve the performance of pCVD diamond devices we focussed on possible changes of its operational conditions. In on-line dosimetric measurements, two electrodes are deposited on the diamond film surfaces and a constant external bias is applied: the incident radiation generates a current in the biased structure which is monitored during exposure to get real-time measurements of the absorbed dose. Contacts on diamond are usually manufactured through the vacuum deposition of metals such as Ti/Au, Cr/Au or Ti/W followed in cases by thermal annealing to ensure the formation of a carbide interface [20], measurements of the I–V characteristics in the typical operation ranges (10–1000 V) show the ohmic behaviour of such devices [21]. Diamond devices are generally operated using an electric field of 1 V/μm to saturate the drift velocity [19,22] and increase the sensitivity, which depends on the average internal electric field through the mobility [21,23]. On the other hand, it is known that the application of an external voltage promotes the trapping of charged carriers in defects within the bulk. This brings to unwanted polarization effects [24,25] which must influence both the stability of the dynamic response and the rise and decay times of the device.

Recently, it was pointed out that, in spite of the ohmic behaviour observed in the typical operational range of applied voltages, the electrode inner structure of a diamond-based on-line dosimeter is in fact characterised by a back-to-back Schottky barrier configuration [26,27]. The work function of diamond is very high and it depends on its surface treatment, with an oxygen terminated surface its value is as high as 5.8 eV. As the work function of titanium and chromium are respectively 4.3 eV and 4.5 eV, the metallization process creates at the metal-diamond interface a junction with a potential difference of the order of 1 V and an in-built electric field of around 1 V/nm, with the diamond surface negatively biased and the titanium/chromium surface positively biased [27]. This experimental evidence suggests that null-bias operation during exposure to radiation is viable. In fact, due to the settlement of the Schottky barriers at the interfaces between metal and diamond, an active volume is present in the diamond sample also at null-bias and the built-in electric field at electrodes can be used to collect the charge generated by the incident radiation in such active regions. In the absence of an applied external bias, and thus of an electric field driving the generated charges through the dosimeter volume, trapping mechanisms at defects in the diamond bulk are negligible. This in principle must render the response dynamics mostly unaffected by impurities and less dependent on the overall crystalline quality of the bulk.

This paper investigates the use of high-quality state-of-art polycrystalline diamond films in clinical dosimetry, both for conventional and intensity modulated radiotherapy, in photovoltaic regime, namely in zero-bias operation. The work has been carried out on the same pCVD diamond films studied in [9] to get a quantitative and direct comparison between the dosimetric characteristics of the sample with an applied external bias and in zero-voltage operation. Results show that the dosimetric performances of the pCVD diamond films increase significantly at null-bias operation. Our results represent a significant step towards applications as IMRT with synthetic polycrystalline CVD diamond films.

2. Experimental details

2.1. Detectors

We tested three polycrystalline CVD samples, undoped, 0.5×0.5 cm^2 size, 500 μm thick, produced by Element Six, U.K., in the framework of the RD42 CERN collaboration [28]. They are high detector-grade quality samples with charge collection distances of the order of 200–250 μm. Optimization of the electronic quality of the pCVD diamond material has been obtained using a material removal procedure, performed from the substrate side, where the density of grain boundaries is higher due to the typical columnar growth of the diamond microcrystals. The diamond surface is lapped on both sides: a scanning electron microscopy micrograph of the lapped surface of a pCVD diamond film of the same quality is shown in Ref. [29]. The diamond films are equipped with Cr/Au electrical contacts on the front and rear surfaces in transverse geometry. The electrical contacts of one detector (ES1) were produced at Ohio State University. The contact is a circular dot with a radius of 1.5 mm surrounded by a 254 μm wide guard ring, distant 235 μm from the central pad. The other two samples (ES5,ES9) were equipped with in-house made electrical contacts with a diameter of 2.8 mm. Before metallization diamond surfaces were cleaned in a highly acidic and corrosive environment leaving the diamond surface partially oxygen terminated. Chromium deposition on oxygen terminated diamond surface was used in order to achieve a good mechanical adhesion and, as discussed in [27], to create a built-in potential barrier at the metal-semiconductor interfaces. A complete dosimetric characterisation of the devices when an external voltage is applied has been reported in [9]. A detailed study of the microscopic properties of the material and of the electrode configuration of the same samples has been reported in [26].

A Farmer type Ionization Chamber NE 2571 (IC) connected to an UNIDOS electrometer was used as the reference dosimeter in order to take into account of beam instabilities and to determine the absolute dose. The chamber was calibrated at the Italian Primary Standard Dosimetry Laboratory (INMRI-ENEA), in terms of absorbed dose to water in a ^{60}Co source.

A PTW natural diamond, 0.3 mm thick and with a sensitive area of 4.5 mm^2, coupled to a Keithley 6517A electrometer calibrated vs. the Farmer chamber in a 10 MV photon beam, was used as the gold standard in IMRT and in conventional applications when dynamic response was studied [30]. For the output factor analysis our reference for the smallest field 2.4×2.4 cm^2 was a Stereotactic Field Detector (SFD) diode of Scanditronix, while for larger fields, from 4×4 cm^2, we used the Farmer chamber. During exposure samples were placed in a PMMA slab phantom at a water-equivalent

depth of 5–10 cm. The dosimetric performance of the pCVD diamond has been investigated measuring both current and charge signals. Data have been directly compared with those of the PTW natural diamond dosimeter, irradiated along with the pCVD diamond. The devices were both connected to a Keithley 6517A electrometer; PTW natural diamond was polarized with a +100 V bias, the pCVD diamond dosimeter was kept unbiased. We observe that, being characterised by a back-to-back Schottky contact configuration, current polarity of the pCVD diamond samples during exposure to irradiation in null bias can be either plus or minus, depending on which of the two Schottky barriers is prevailing on the other. In the following we will therefore always show the absolute value of the current response. Dosimetric parameters investigated in this study are: time stability, dynamic response, linearity, dose–rate and energy dependence, output factors.

2.2. Experimental setup

The dosimetric characteristics of the pCVD diamond dosimeters have been investigated at the Radiotherapy Unit of the University of Florence by means of an Elekta Synergy LINAC, both with conventional 6–10–18 MV photon beams in a 10×10 cm^2 field size (sample placed at the isocentre) and with a 10 MV IMRT field. During irradiations with conventional beams the IC was placed 2.5 cm apart from the pCVD diamond at the same water-equivalent depth and irradiated simultaneously. In modulated intensity beams the photon fluence impinging on the patient varies point to point inside the radiation field. Such a modulation is obtained thanks to a Multi Leaf Collimator (MLC) mounted on the linear accelerator. At the Radiotherapy Unit of the University of Florence the step-and-shoot modality is used, i.e. the desired non uniform fluence distribution is obtained for each beam by a sequence of numerous static irradiations (segments) each characterised by a different MLC configuration; the beam is switched off during the MLC rearrangement. The width of the leaves (projected at the isocentre plane) of the Elekta Synergy LINAC is 4 mm. In IMRT measurements, a 10 MV photon beam for prostate treatment in step-and-shoot modality was released, obtained by a sequence of 26 segments with a nominal dose rate of 400 MU/min. Fig. 1 shows the dosimetric map of the 10 MV photon beam as given by the treatment planning system used for the IMRT application. During irradiations with the IMRT field the pCVD and the PTW diamond responses have been compared directly: the cross in Fig. 1 indicates where they have been positioned in the field map.

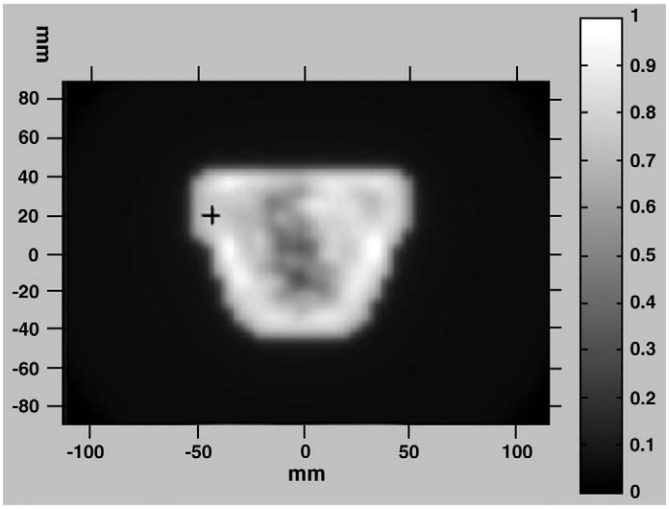

Fig. 1. Dosimetric map of a 10 MV photon beam for prostate treatment released in step-and-shoot modality, as obtained by a sequence of 26 segments by the treatment planning system used for the IMRT application. The cross indicates where pCVD and PTW diamonds have been positioned to get data shown in Figs. 8 and 9.

3. Experimental results and discussion

3.1. Pre-irradiation

One important effect generally observed with diamond dosimeters operated with an external voltage is the increase in current response, during the early stages of exposure, under steady irradiation conditions. Typically, after an X-photon beam irradiation up to about 10–50 Gy (depending on the diamond crystalline quality) a stabilization of the response is achieved, although slight changes of the signal are often observed also beyond this irradiation level [31]. Due to this effect, called priming, a diamond dosimeter must be pre-irradiated before first operation.

To investigate pre-irradiation, the pCVD diamond dosimeters were exposed, in null-bias operation, to a 6 MV photon beam with dose rate 400 MU/min at a water-equivalent depth of 10 cm. Fig. 2a shows the current response of sample ES1 to the first exposure: a cycle of ten consecutive irradiations. Fig. 2b compares the pCVD diamond signals during the first four irradiation steps. During the first irradiation step the current increases, achieving a value up to 92.5% of the stabilization one after approximately 1 Gy (1 MU ~ 1 cGy). Then, the current is still slowly increasing, reaching a 99.5% of the stabilization value, after ~14 Gy. Doses to achieve the 92.5% of the final signal are ~1 Gy and ~8 Gy for samples ES5–ES9, those to reach the 99.5% are ~20 Gy (ES5) and ~8 Gy (ES9). These values of pre-irradiation are comparable to those reported for the single crystal diamond dosimeter (10–15 Gy) [17] and for the PTW natural diamond (5–15 Gy) [29]. For comparison we show in Fig. 2c the current response of the same ES1 diamond film when an external voltage of 500 V is applied (inset shows a partial view of the response for the two irradiation steps, normalized to the value of current achieved when stability is reached). During the first irradiation step the current increases, achieving a value up to 92.5% of the stabilization one after approximately 5.6 Gy, the current is then increasing, reaching a 99.5% of the stabilization value, after $\approx$50 Gy. These results show that in null-bias operation the pre-irradiation dose is reduced to less than one half of that needed when an external voltage is applied.

3.2. Sensitivity

After pre-irradiation, samples sensitivity was measured under different beam conditions. Measured values of sensitivity per unit area are 4 nC/Gymm^2, 1 nC/Gymm^2 and 0.4 nC/Gymm^2 respectively for samples ES1, ES5, ES9. The different sensitivity of the three samples can be ascribed to a different value of the built-in potential settled at the metal-semiconductor interfaces. Sensitivities of the pCVD diamonds are of the same order of magnitude or higher than those obtained with single crystal diamond dosimeters (a value of 0.32 nC/Gymm^2 is e.g. reported in [17]), but definitely lower than that of PTW natural diamond (18.6 nC/Gymm^2 measured at 100 V) and of the pCVD diamond when kept under bias (e.g. 268.8 nC/Gymm^2 for sample ES1 at 100 V). Anyway, even with sample ES9, characterised by the lowest sensitivity and with a contact area of 6 mm^2, a current of 240 pA is measured during irradiation with a 400 Mu/min dose–rate: considering that noise is of the order of 1 pA or less, the signal to noise ratio is therefore still reasonably high for operation. We note that a realistic size for a pixel in a bidimensional dosimeter for IMRT radiotherapy measurements is close to 4 mm^2, so even in this worst case the signal-to-noise ratio would be sufficiently high to be monitored with the required precision.

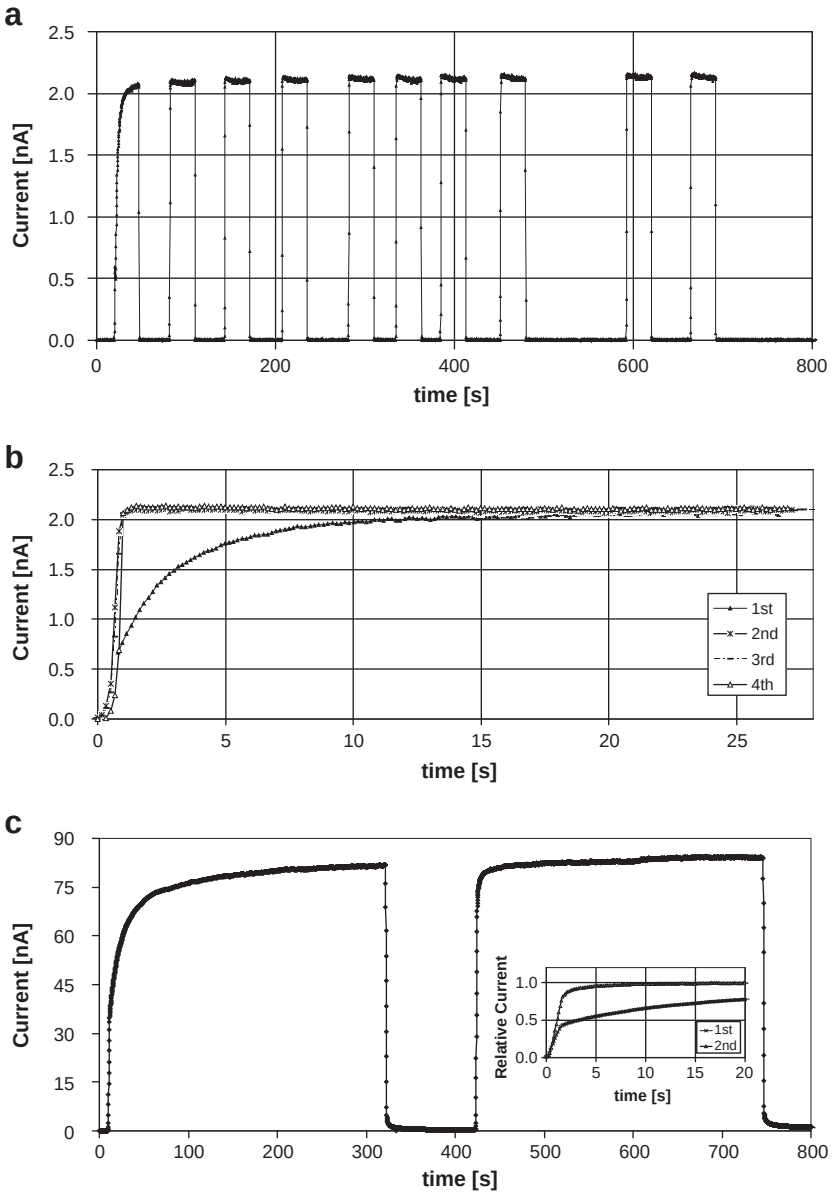

Fig. 2. Pre-irradiation responses of the pCVD diamond. (a) in null-bias operation, in a 6 MV photon beam 400 MU/min, water-equivalent depth 10 cm, (b) detail of the increase of the current signal in null-bias operation (relative to the maximum value achieved in stable conditions); (c) with a 500 V external voltage applied (inset shows the relative increase of the signal) in a 10 MV photon beam.

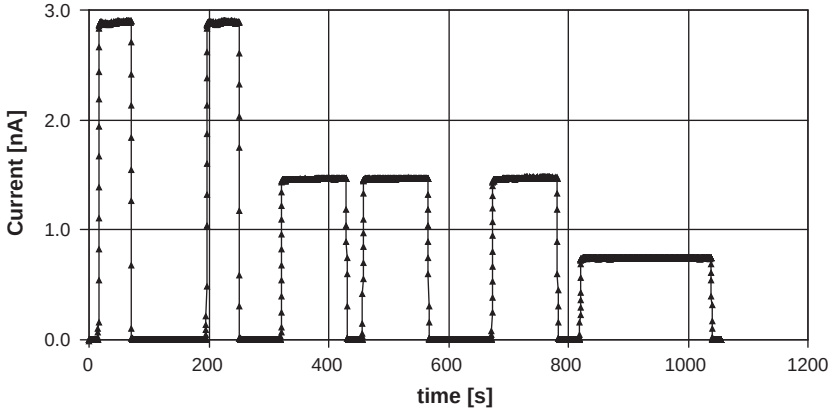

Fig. 3. pCVD diamond current signal at null-bias during different cycles of exposure with a 6 MV Linac beam at nominal dose rates of 400 MU/min (first two), 200 MU/min (third to fifth) and 100 MU/min (last), 10 cm × 10 cm field size and a water-equivalent depth of 5 cm.

3.3. Time stability

After pre-irradiation, the response of the pCVD diamond dosimeters operated in zero-bias is quite stable even when dose rate is changed. Fig. 3 shows the current signal of the ES1 sample as a function of time for three dose–rates: 400 Mu/min (first two irradiation steps), 200 Mu/min (third to fifth) and 100 Mu/min (last). We evaluated the stability of the current response as a function of time during several other irradiation cycles with the three samples: measured current changes were characterised by standard deviations always less than 0.5%, of the same order of those measured for the reference PTW natural diamond detector. Repeatability of the pCVD diamond dosimeters in null-bias, monitored in integration mode, has been evaluated with 11 repeated charge measurements under a 6 MV photon beam at 1 Gy with a dose rate of 400 MU/min. Results give a repeatability of approximately 0.4%.

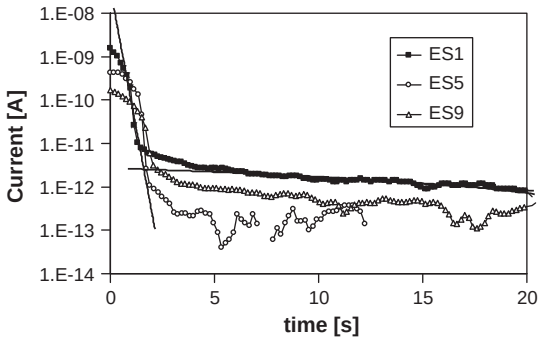

Fig. 5. Decay of the pCVD diamond current responses after the switch off of the 10 MV photon beam (same experimental conditions as Fig. 4). For sample ES1 the decay has been fitted considering two exponential components with different time constants.

3.4. Dynamic response

Fig. 4 shows the current response of the ES1 pCVD diamond compared with the PTW natural diamond during irradiation with a 10 MV photons LINAC beam. A scaling factor is applied to the PTW diamond signal in order to make an easier comparison between the two signals. The pCVD diamond device measured in photovoltaic regime shows a fast response, comparable to that of the PTW natural diamond. The inset shows the signal rise during the setting of the beam: rise time (evaluated up to 90% of the maximum current value) for the two devices is the same, in this measurements resulting of the order of 1 s, although a specific value of the diamond devices time response cannot be accurately assessed because it depends on the read-out electronics adopted in the measurement and on the time structure of the beam. The inset shows as a detail of the measurement how precisely the rise time of the pCVD diamond follows the PTW response during the setting of the LINAC beam. Concerning the rise times, a similar behaviour is shown by the two other samples ES5 and ES9. On the contrary, a different behaviour is observed for the signal decay of the three samples, after the LINAC beam is set off. With the ES1 sample, after a fast decrease, developed in approximately 150 ms, the decay of the pCVD diamond current shows a long signal tail from approximately 10pA to 1pA developed in about 20 s, with time constant of the order of 13.7s (Fig. 5). This long decay tail is probably related to trapping–detrapping mechanisms at defect states characterised by energy levels active at room temperature and occurring in the proximities of the diamond–metal contact interfaces. On the contrary, the other two samples, ES5 and ES9, show a faster decay behaviour, reaching the 1 pA level respectively in 1.9 s and 3.7 s.

3.5. Linearity

The linearity in the low dose range is a relevant parameter for IMRT application where segments of a few Monitor Units (MU) are often delivered. We evaluated the linearity of the pCVD diamond response in integration mode in the 1–1000 MU range (1 MU ~ 1 cGy) with a dose rate of 400 MU/min: results are shown in Fig. 6. A good linearity is proven over three dose decades, results are characterised by a mean value of residuals of 0.3% and a maximum deviation of 1.7%. To directly compare our results with the single crystal diamond dosimeter reported in Ref. [17] we calculated the linear behaviour index α as given by fitting experimental data with the function $y = ax^{\alpha}$ (see inset on the right): we obtain $\alpha = 1.0085$, a value very close to unity which proves the good linearity of the response over three decades. Further, to directly compare our results with those reported in Ref. [9], where linearity of the same pCVD diamond films with an external voltage applied is discussed, in the same figure we show a plot of the residuals against the corresponding dose values (inset on the left). We obtain values below 5% from a dose of 4 cGy, while with external voltage applied the deviations from linearity were quite large, reaching almost 20% for doses close to 0.1 Gy [9]. The good result obtained is a consequence of the fast response dynamics of the device when operated in null-bias, in fact when the same device is kept under bias (100–500 V) a deviation below 5% is observed only for doses up to about 0.2 Gy [9].

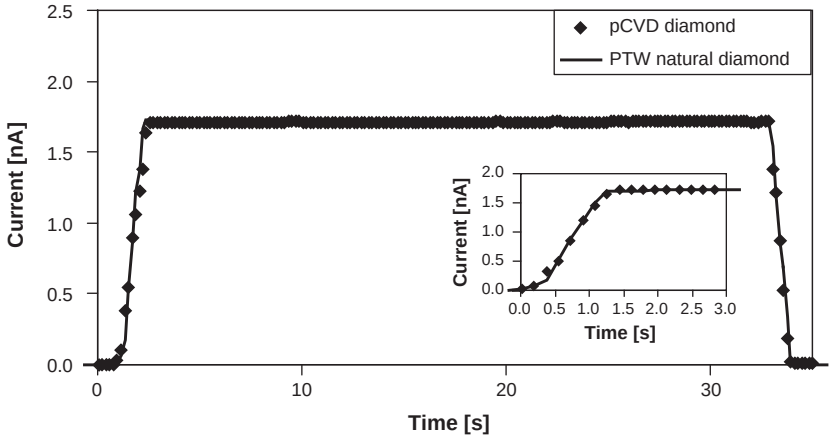

Fig. 4. Current response of the pCVD and PTW natural diamonds during an exposure to a 10 MV photon beam, 10×10 cm^2 field size, water equivalent depth 10 cm, dose rate 400 Mu/min. The PTW diamond current response was scaled to the pCVD diamond value to directly compare the dynamic of the two signals (multiplication factor 0.23). In the inset a detail of the measurement is shown to evidence that the rise time of the pCVD diamond follows the PTW response during the setting of the LINAC beam.

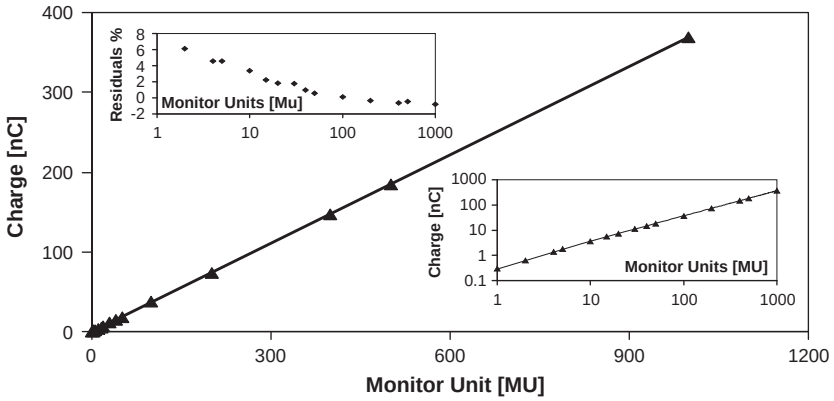

Fig. 6. Dose–response of the pCVD diamond film as a function of the delivered dose (1 Monitor Unit ~ 1 cGy). The log–log scale in the inset on the right shows the good linearity of the curve over three decades. The inset on the left shows the deviation from linearity against the corresponding dose values.

3.6. Dose–rate dependence

Dose rate dependence has been investigated in the range 0.4–10.1 Gy/min by changing both the Pulse Repetition Frequency (PRF) of the accelerator and the dose per pulse (i.e. varying the source-detector distance). Measurements have been performed in integration mode and repeated, for dose evaluation, with an ionization chamber in the same position of the pCVD diamond. The current signal (I) has been fitted according to the Fowler model [32]:

$$I = I_{dark} + RD_r^{\Delta}$$

with D_r dose rate, I_{dark} dark current, R and Δ fitting parameters (Δ describes the deviation from linearity). In case of null-bias operation we get Δ values of 0.955, 0.919, and 0.964 respectively for ES1, ES5 and ES9: a definite improvement with respect to the case of operation with external bias, characterised by Δ values ranging from 0.880 to 0.907 [9].

3.7. Energy dependence and output factors

Energy dependence has been investigated with the ES1 detector under 6–10–18 MV photon beams using a $10\times10\ cm^2$ field with a dose rate of 400 MU/min. According to the IAEA 398, a Farmer ionization chamber was used as the reference dosimeter. The sample has been placed at the isocentre at a water equivalent depth of 10 cm. Sensitivity values obtained for 6–10–18 MV beams are respectively of 30.27 nC/Gy, 27.67 nC/Gy and 25.17 nC/Gy. At a first sight these results could seem surprising as the diamond detector response is expected to be independent of the beam energy. Actually, the packaging of the sensor affects its response in terms of energy dependence as this particular device is not embedded in epoxy resin: a thin layer of air remains over the sensor in the PMMA holder. Energy dependence of the device could be removed performing a more accurate and suitable encapsulation. Anyhow, in the clinical application, it is more important that the energy independence holds within the same nominal energy when different field sizes are considered. Changing the field dimensions in fact the scattered radiation percentage varies, affecting the beam spectral composition. In order to investigate this issue, output factors, defined as the ratio between the dose absorbed with a certain field size and the one absorbed with a reference field when the same number of MU is delivered, have been measured under a 10 MV photon beam. Data obtained in integration mode are reported in Fig. 7 and compared with the ones previously measured with an ionization chamber. The maximum difference is about 3%, a value found for the smallest field size, where detector positioning is critical. The good result obtained even for small field sizes is particularly important in view of IMRT applications, where small field sizes are frequently present.

3.8. Application in an IMRT field

As an example of IMRT application, the ES1, ES5 and ES9 pCVD diamonds and the PTW detector have been placed in the position indicated in Fig. 1 inside an IMRT field for prostate cancer treatment. The current signals have been acquired during the IMRT delivery and compared: results are shown in Figs. 8 and 9. The dose is delivered in 26 segments in an overall time window of 160 s, in the first 8 segments the

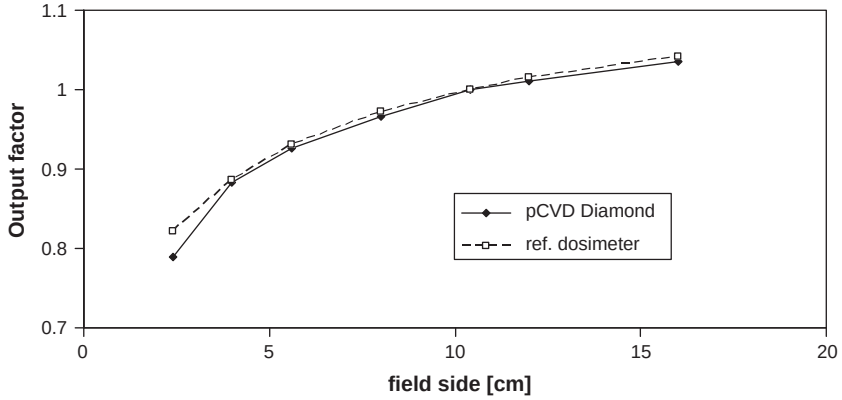

Fig. 7. Output factors for squared fields of a 10 MV photon beam measured with the pCVD diamond compared with ionimetric data (ion chamber used as standard reference).

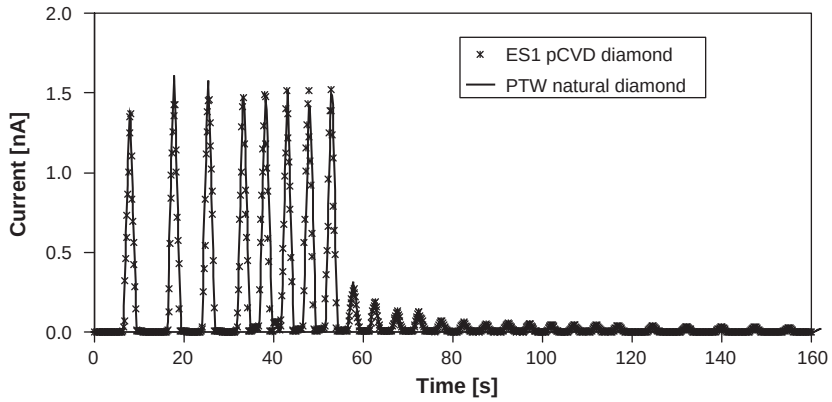

Fig. 8. Current signals of the pCVD and PTW diamonds placed in the position indicated in Fig. 1 inside an IMRT field for prostate cancer treatment. The dose is delivered in 26 segments in an overall time window of 160 s, in the first 8 segments the detectors are placed inside the segment treatment field, afterwards the detectors are outside and the current signals are due only to scattered radiation.

detectors are placed inside the segment treatment field, afterwards the detectors are outside and the current signal is due only to the scattered radiation. Fig. 8 compares the current signal of the PTW diamond and of the ES1 diamond (the PTW signal has been rescaled to fit the ES1 one). A good agreement is observed between the two signals, showing that both devices are able to follow the dynamics of the IMRT beam. This is an

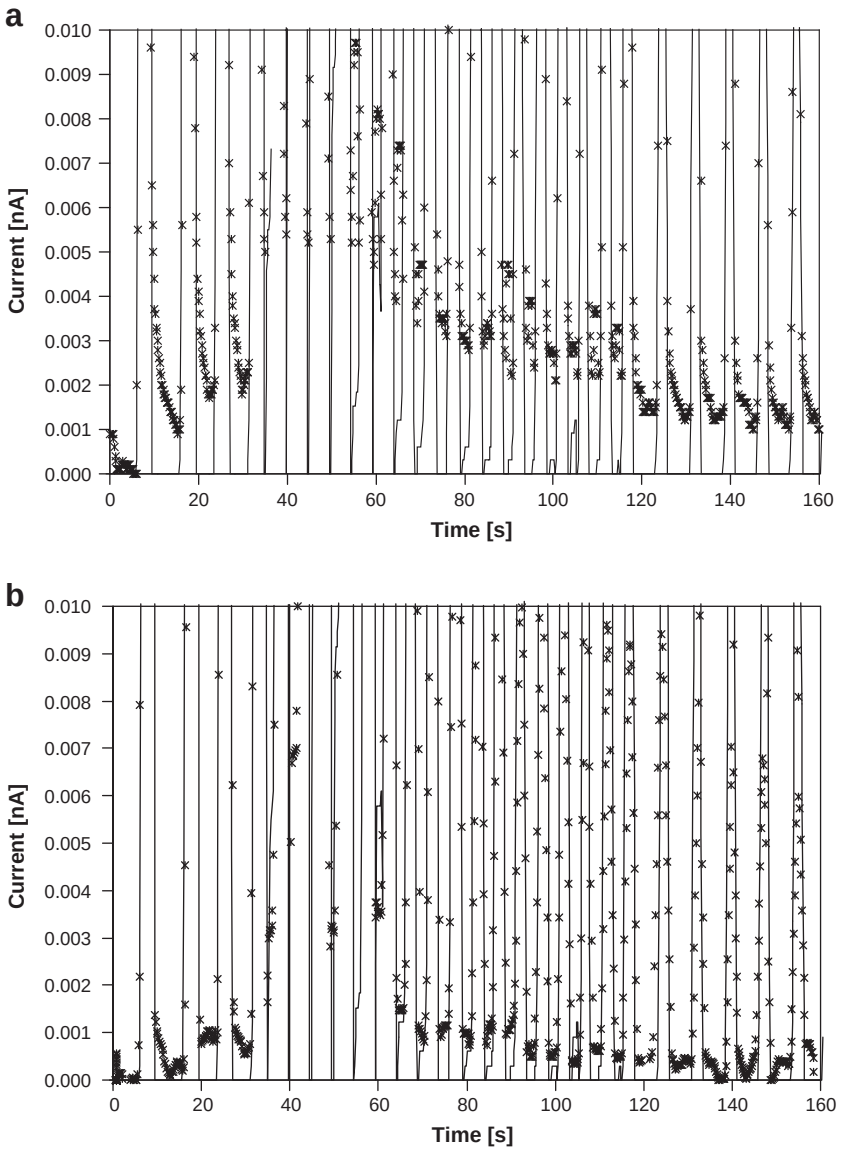

Fig. 9. Current signal of the PTW diamond in the pAs range during the IMRT test shown in Fig. 8 as compared to the current response of the (a) ES1 and (b) ES9 pCVD diamonds.

important result, as up to now polycrystalline CVD diamond were seen to be inadequate to follow the fast time structure typical of an IMRT beam: see e.g. Fig. 7 in Ref. [9], where the same pCVD diamond device is operated with an external voltage of 500 V.

Fig. 9a and b show the response of the ES1 and ES9 pCVD diamond samples against the PTW one in the low current range (0–10 pA), to evidence the small differences occurring between the two signals during the IMRT application. In the case of ES1 (Fig. 9a) the two signals differ due to the slow-decay of the pCVD diamond (see Fig. 5), occurring immediately after each segment of the IMRT application is set off. The tail of the pCVD diamond current is of the order of 5 pA, very low if compared to the ~1.5 nA maximum current signal observed when the segments are delivered (0.3%). Therefore, a definite improvement is obtained in null-bias operation with respect to the case reported in [9]. Nonetheless, the signal tail observed at null-bias can still be significant when considering the overall time window of the IMRT beam delivery, representing a potential source of uncertainty in the total delivered dose. In fact, the error to the total dose evaluation due to the current tail can bring to deviations up to 5–10% along the whole IMRT treatment plan, especially when large switch-off time fractions over the total treatment time are present in the segment distribution. On the contrary, in the case of sample ES9 (Fig. 9b) characterised by a faster decay time (see Fig. 5), the signal tail observed when the IMRT segment is set off is negligible and errors in the total dose evaluation are within the 2%.

The results presented above indicate that in null-bias the performances of the pCVD diamond as a dosimeter increase significantly with respect to the case of operation with an external bias. This experimental evidence is mainly to be ascribed to the fact that, being null the electric field within the bulk, as diffusion length in polycrystalline diamond is negligible due to the high concentration of recombination centres, defects in the bulk are not contributing to the current response and signal is thus only due to the charge collected at interfaces. Trapping–detrapping mechanisms in the bulk, which are believed to give the main contribution to priming and signal instability when the external electric field is applied, are therefore now negligible.

To conclude, we note that the performance of the device will surely further improve if, instead of being made with two back-to-back Schottky contacts, as in the present case, a front Schottky contact would be coupled to a really ohmic back contact. Recently a novel metallization technique using diamond-like carbon tunnelling junction and Pt/Au as electrical contacts has been developed in view to get ohmic contacts at metal semiconductor interfaces [27]. The manufacturing and test of pCVD diamond dosimeters made with such electrode structure will be the subject of our studies in forthcoming works.

4. Conclusions

A detailed investigation of the performance as a dosimeter of state-of -art polycrystalline CVD (pCVD) diamond detectors operated in photovoltaic regime for applications in clinical radiotherapy has been carried out. Results have been compared with those obtained with a natural PTW diamond detector and an ionisation chamber, used as reference detectors. Applications with conventional X-photon beams from LINAC as well as with a 10 MV photon Intensity Modulated Radiotherapy beam have been analysed experimentally.

Our results show that pCVD diamond detectors improve dramatically their dosimetric performances when operated in null-bias. Most important results are:

1) A pre-irradiation dose reduced to almost half of that needed when external voltage is applied;
2) excellent time stability after pre-irradiation, characterised by standard deviations less than 0.5% and a repeatability of about 0.4%;
3) fast rise times of the current signal, comparable to that of commercial reference dosimeters;
4) linearity with dose of the collected charge proven over three decades, with mean value of residuals of 0.3% and a maximum deviation of 1.7%;
5) a smaller deviation from linearity of the current vs. dose–rate curve;
6) Output factors comparable to that of commercial reference dosimeters even for small fields.

The significant improvement in the dosimetric behaviour is obtained thanks to the null-bias operation of the device. Of note is the fact that, concerning the above listed dosimetric characteristics, pCVD diamond in null-bias operation have performances comparable to those recently reported for synthetic single crystal diamond (scCVD) dosimeters [16,17], which are operated in null-bias as well. Authors in [16,17] ascribe the proper dosimetric behaviour of scCVD diamond detectors as mainly due to the high crystalline quality of the single crystal material: our experimental results suggest that a significant contribution to the beneficial performances could be as well attributed to the null-bias operation of the device.

Results obtained in this work stimulate further studies in the near future, in order to improve the dosimetric performances of the device. Certainly, more investigations are needed to optimize the metal-semiconductor interface barriers on the pCVD diamond for the IMRT application. Further, research could be focussed on the development of a device where the front Schottky contact is coupled to an ohmic back contact, this latter e.g. obtained through a diamond-like-carbon layer as discussed in [27]: this research program will be the subject of our studies in forthcoming works.

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