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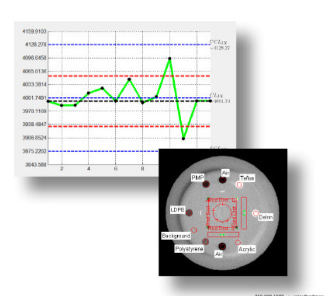

A novel synthetic single crystal diamond device for *in vivo* dosimetry

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Purpose: Aim of the present work is to evaluate the synthetic single crystal diamond Schottky photodiode developed at the laboratories of "Tor Vergata" University in Rome in a new dosimeter configuration specifically designed for offline wireless *in vivo* dosimetry (IVD) applications.
Methods: The new diamond based dosimeter, single crystal diamond detector (SCDD-iv), consists of a small unwired detector and a small external reading unit that can be connected to commercial electrometers for getting the detector readout after irradiation. Two nominally identical SCDD-iv dosimeter prototypes were fabricated and tested. A basic dosimetric characterization of detector performances relevant for IVD application was performed under irradiation with ^{60}Co and 6 MV photon beams. Preirradiation procedure, response stability, short and long term reproducibility, leakage charge, fading effect, linearity with dose, dose rate dependence, temperature dependence, and angular response were investigated.
Results: The SCDD-iv is simple, with no cables linked to the patient and the readout is immediate. The range of response with dose has been tested from 1 up to 12 Gy; the reading is independent of the accumulated dose and dose rate independent in the range between about 0.5 and 5 Gy/min; its temperature dependence is within 0.5% between 25 and 38 °C, and its directional dependence is within 2% from 0° to 90°. The combined relative standard uncertainty of absorbed dose to water measurements is estimated lower than the tolerance and action level of 5%.
Conclusions: The reported results indicate the proposed novel offline dosimeter based on a synthetic single crystal diamond Schottky photodiode as a promising candidate for *in vivo* dosimetry applications with photon beams. © *2015 American Association of Physicists in Medicine.* [http://dx.doi.org/10.1118/1.4926556]

Key words: synthetic diamond dosimeter, *in vivo* dosimetry, radiation therapy, semiconductor detectors

1. INTRODUCTION

Modern radiation therapy techniques have increased demands of additional tools for the verification of the dose actually delivered to the patient. The use of *in vivo* dosimetry (IVD), as a component of the radiation therapy process, is aimed at providing information necessary for assessment of accuracy and precision in dose planning and delivery. In a number of reports,[1–7] IVD has been indeed recognized as an essential quality assurance (QA) instrument to detect possible errors in dose delivery, to check the patient setup during the actual treatment, and to evaluate the dose to critical organs (gonads, eyes, etc.). Moreover, in some countries, IVD has become mandatory, which implies the fulfillment of highly demanding clinical and legal requirements.[5] An accurate evaluation of IVD detector systems is thus advisable for the clinical goals of IVD in a radiotherapy program. An accuracy of ±5% in the delivery of absorbed dose to tumors and normal tissues was recommended by International Commission of Radiological Units (ICRU report No. 24).[1] The figure of 5% can be interpreted as an interval of acceptable deviation between the prescribed dose and the dose delivered to the target volume (IAEA 398).[8]

Possible aims of IVD are the measurement of entrance and/or exit dose and, by combination of the two, the check of the target dose.[1,9,10] IVD is also of particular relevance in total body irradiation (TBI)[11] and in total skin irradiation (TSI) as well.[12] In those cases, IVD measurements are considered not only an independent check but also an integral part of the overall dosimetry program.[13]

Several types of detectors have been made available for IVD application, such as thermoluminescent dosimeters (TLDs), radiochromic films, metal oxide semiconductor field effect transistors (MOSFETs), alanine, and silicon diodes. Ion chambers are not commonly used because of their fragility and the need of the polarization voltage that may cause injury to the patients. Extensive reviews on the main dosimetric features, advantages, and drawbacks of state-of-the-art IVD detectors can be found in the literature.[5,13–15] It is worth to point out that among possible classifications, IVD detectors can be roughly divided into online detectors, and passive or offline detectors. Silicon diodes[3,4] and currently available MOSFETs (Ref. 16) (Best Medical Canada, Ltd) are real-time wired dosimeters, widely used for IVD. The main advantage of such kind of detectors is the immediate readout, i.e., dose information is instantaneously available, with the patient still in the treatment couch. If a deviation is observed, an immediate check of the dose delivery setup is thus potentially allowed, provided that the expected detector signal is known at the very moment of the IVD by means of preliminary calculations done before the irradiation session.[13] Among offline wireless IVD detectors, TLDs are the most commonly employed ones.[2,13] Portable patch-like MOSFETs and implantable MOSFETs were also developed (Sicel Technologies, Inc.) and used for IVD, but they are no more commercially available.[17,18]

In a patient-care centered approach to IVD, wireless dosimeters, with no need of unwieldy cables on patient skin during the irradiation session, are in principle recommendable. Due to some limitations in the use of TLDs and alanine, such as a long reading time and temperature dependence,[13] the development of new detector design and materials for wireless IVD is a challenging issue. Workload and costs involved by newly proposed dosimetric systems for IVD should be carefully taken into account too.

In this work, a novel prototype detector based on a synthetic single crystal diamond Schottky photodiode is evaluated for offline wireless IVD application. The results from a basic dosimetric characterization of two nominally identical detectors under irradiation with a ^{60}Co reference gamma-ray beam and a 6 MV clinical photon beam are reported and discussed.

2. MATERIALS AND METHODS

2.A. Detector assembly

The novel prototype synthetic single crystal diamond detector (SCDD-iv) tested in the present work was specifically designed for offline wireless IVD application. The diamond based IVD system consisted of a small unwired portable detector unit, into which photogenerated charge is kept stored with negligible leakage loss over time, and of a reading unit, where the detector unit must be plugged in. The reading unit is connected to a low noise triaxial cable terminating with a triaxial connector. Charge can be easily read by means of standard commercially available electrometers for low-level current measurements, with no need of dedicated read-out electronics.

Schematics and photos of the SCDD-iv system are shown in Fig. 1. Figure 1(a) shows a schematic diagram of detector and reading units, with indication of detector components and electrical connections. A photo of four detector units and of the reading unit is shown in Fig. 1(d).

As highlighted in Figs. 1(a) and 1(b), the two key components of the detector unit are (i) the SCDD plate, which acts as a photodiode, producing a photocurrent under irradiation, and (ii) a charge storage capacitor, where the photogenerated charge is kept stored after irradiation.

The diamond sensitive element of the investigated SCDD-iv system was based on a well assessed detector technology developed at the laboratories of "Tor Vergata" University in

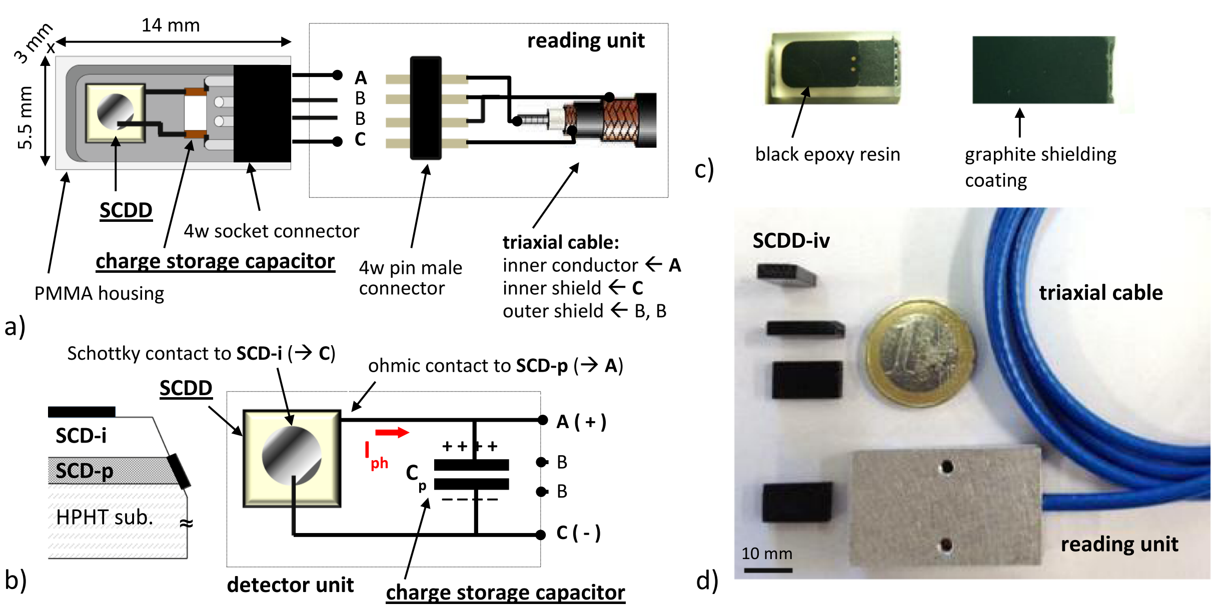

FIG. 1. Schematic diagrams and photos of the SCDD-iv system. (a) Sketch of the detector and reading units with indication of the detector system components and electrical connections. (b) Structure of the multilayered SCDD plate and sketch of the internal connections inside the detector unit. Current flow I_{ph} of photogenerated carriers and charging of the charge storage capacitor are schematically shown. (c) Photos of the SCDD-iv unit during the fabrication process: after application of a black epoxy resin to fill the detector slot in the PMMA housing (left) and after application of a graphite shielding coating (right). (d) Photo of the SCDD-iv system, showing four detector units and the reading unit.

Rome, successfully employed in the fabrication of the PTW-Freiburg microDiamond™ dosimeter (PTW-type 60019) for external beam radiotherapy dosimetry in clinical photon and electron beams. Details on such device technology, fabrication process, and detection mechanism, as well as dosimetric properties are reported elsewhere.[19–21] A multilayered *p*-type diamond (SCD-p)/intrinsic diamond (SCD-i) structure was fabricated by means of a two-step microwave plasma enhanced chemical vapor deposition (CVD) on a commercial low-cost $3.0\times3.0\times0.3$ mm^3 high-pressure high-temperature (HPHT) synthetic single crystal diamond substrate. Metal contacts were formed by thermal evaporation of a 2.2 mm in diameter rectifying metallic contact on the nominally intrinsic diamond surface and of an ohmic contact on the *p*-type diamond layer. The resulting device, whose structure and contacts/connections are sketched in Fig. 1(b) (left), acts as a Schottky photodiode, due to the formation of a built-in potential at the metal/intrinsic diamond interface, and can be operated with no need of an external voltage bias ($V_b = 0$ V). The sensitive volume is about 0.004 mm^3, determined by the thickness of the depletion region extending below the circular Schottky contact through the whole ~1 μm thick SCD-i top layer. An intrinsic capacitance C_j of ~200 pF is associated with the metal/SCD-i Schottky junction.

A charge storage capacitor C_P ($C_P \gg C_j$) was connected in parallel to the SCDD electrodes, as schematically shown in Figs. 1(a) and 1(b). During irradiation, the photogenerated charge Q_{ph} accumulates on the capacitor plates due to the photocurrent I_{ph}. A positive voltage drop $V_{\mathrm{AC}} = Q_{\mathrm{ph}}/C_P$ arises between the capacitor plates and it is kept ideally with no losses over time after irradiation. For this to happen, charge losses in the device have to be minimized, by using high capacitor insulation resistance, achieving negligible SCDD leakage currents, and by a good choice of insulating materials for device packaging.

A 10 nF surface mount ceramic chip capacitor was used for the two SCDD-iv units studied in the present work. Particular care was addressed to the choice of the capacitor, being its features critical for the overall IVD detector performance. A low noise, high thermal stability capacitor with a nominal insulation resistance IR $\geq 10^6$ MΩ in the 0 to 40 °C temperature range was used.

As shown in Fig. 1(a), in the SCDD-iv unit, the SCDD plate and the charge storage capacitor are embedded in a polymethylmethacrylate (PMMA) housing (cuboid shaped, 5.5 mm wide, 14 mm long, and 3 mm thick) provided with a 2 mm deep recess and connected to each other and to a commercial ultraminiature four-way PCB socket connector. The SCD-i surface was located ~1.7 mm below the top surface of the PMMA housing, with the HPHT substrate side being glued on the PMMA housing. The detector measurement point was assumed to be at such ~1.7 mm depth, at the center of the top circular metallic Schottky contact. As shown in Fig. 1(c), in the later stages of the detector manufacturing procedure, the bare detector assembly sketched in Fig. 1(a) was filled by a two-component black epoxy resin. A graphite coating was thus sprayed on the SCDD-iv unit in order to provide electromagnetic shield. Finally, mechanical strength against possible scratches was achieved by spraying on the SCDD-iv unit a transparent plastic coating.

A schematic of the reading unit is also shown in Fig. 1(a) (right). It consists of an aluminum housing, (cuboid shaped, 25 mm wide, 40 mm long, and 6 mm thick) equipped with a straight through hole male four-way connector, soldered to a screened low noise triaxial cable. The correspondences among the electrical connections between the four-way connector and the triaxial cable are indicated by the letters A, B, C. The two central pins in the reading unit were shorted and connected to the outer shield of the triaxial cable. Similarly, the two central pins of the four-way socket connector embedded in the detector unit were shorted by means of the outer graphite shielding coating. Once the detection unit is plugged into the reading assembly, the outer shielding graphite coating is thus directly connected to the outer shield of the triaxial cable. The stored photogenerated charge produces a positive voltage drop V_{AC} between the A and C poles, connected to the central conductor and to the inner shield of the triaxial cable, respectively.

2.B. Irradiation setup and charge measurements

The dosimetric properties of the tested SCDDs-iv were investigated in a ^{60}Co reference gamma-ray beam at the National Institute of Ionizing Radiation Metrology of ENEA (ENEA-INMRI C.R. Casaccia, Rome, Italy) and in a 6 MV clinical photon beam produced by an Elekta Precise linear accelerator (Elekta Crawley, UK) at the Tor Vergata University General Hospital in Rome.

Irradiations in the ^{60}Co gamma-ray beam were performed in a 7×7 cm^2 field size. The SCDDs-iv were placed in a PMMA holder attached to the outer surface of a $30\times30\times30$ cm^3 water phantom with a 4 mm thick PMMA buildup at a source to surface distance (SSD) of 70 cm. In such irradiation conditions, the dose rate was about 0.5 Gy/min. A dose of 1 Gy was delivered to the detector in each irradiation. Charge measurements were carried out by connecting the SCDD-iv reading unit to a Keithley 6514 electrometer after each irradiation. The two tested detectors were irradiated one-by-one and at least 3 min delay occurred between end of irradiation and charge reading. A different experimental setup was used for evaluating the SCDD-iv angular response. In this case, the detectors were placed in the middle plane of a cylindrical PMMA mini-phantom ensuring the build-up conditions all around the SCDD-iv centered along the cylinder axis. Irradiations were performed in a 4×4 cm^2 beam size, and the incident angle of the beam on the SCDD-iv was varied rotating the PMMA mini-phantom, and thus the detector, around the cylinder axis.

Irradiations in the 6 MV clinical photon beam were performed at a nominal dose rate of 300 monitor units (MU)/min. The accelerator beam was calibrated to give an output of 10^{-2} Gy/MU at the depth of maximum dose in water, with a SSD of 100 cm and a field size of 10×10 cm^2 at the water surface. The two SCDDs-iv were placed in two slots in a PMMA holder attached on the top of a plastic phantom composed of ten 1.5 cm thick water-equivalent RW3 slabs (PTW-Freiburg, Germany). An additional slab, of the same thickness, was superimposed above the detectors during irradiation, so that the SSD was 100 cm and the SCDD-iv measurement point was placed at the position of the maximum dose. Simultaneous irradiations of both detectors were performed. For these measurements, the SCDD-iv reading unit was connected to a PTW Unidos E Universal Dosimeter.

The above mentioned PMMA holders were designed with the aim of ensuring the best repeatability in positioning the detectors in subsequent irradiations, thus reducing as much as possible positioning errors. Unless otherwise specified, the detectors were oriented with the top metallized SCD-i surface perpendicular to the beam axis.

2.C. Measurements details

A basic dosimetric characterization was carried out in order to evaluate the suitability of the tested detector system for IVD applications. Two SCDD-iv units were measured, referred to as SCDD-1 and SCDD-2 in Secs. 3.A–3.E. Measurements of leakage charge, preirradiation dose, repeatability and response stability over time, linearity with dose, dose rate dependence, temperature dependence, and angular response were performed. The analysis of the response stability included the evaluation of the fading effect, i.e., the possible variation of detector signal depending on the delay occurring between irradiation and reading.

Except for the warm-up irradiation and the evaluation of detector signal repeatability, three charge readings (related to three subsequent irradiations of the same detector) were recorded for each measurement point. The experimental standard deviations (1σ) of experimental data sets were calculated and reported in the graphs as error bars. The reproducibility of detector response was estimated from the experimental relative standard deviation (RSD) of measured data in different measurement sessions. The uncertainty budget of absorbed dose to water (D_w) measurements by the IVD detectors was finally evaluated.

The following procedure was adopted for irradiation and reading of the SCDD-iv. First, the residual charge stored in the SCDD unit was reset by the electrometer, with the detector unit plugged into the reader unit. When the Keithley 6514 electrometer was used, such a reset was done by passing from charge measurement mode to current measurement mode, so that the pin terminals A and C were shorted [see Fig. 1(b)]. After charge reset, the detector unit was ready to be irradiated, and after irradiation, it was again plugged into the reader unit for charge reading and further charge reset procedure. No specific precautions were needed in handling the SCDDs-iv, neither before nor after irradiation.

Due to the requirement of storing photogenerated charge with no loss over time, a measurement of leakage charge accumulated in dark conditions, i.e., in absence of radiation, after charge reset and prior detector irradiation, was also performed. A high value of leakage current would determine the accumulation of a leakage charge before irradiation, as well as loss of stored photogenerated charge after irradiation, thus making the device response unstable. The measurement of leakage charge was carried out using the Keithley 6514 electrometer. The two SCDDs-iv were in turn left plugged into the reading unit for 30 min, and charge was measured at intervals of 5 min after the initial charge reset procedure.

The studied SCDD-iv may need a preirradiation procedure to settle the detector response to a stable level, as previously reported on our similar cabled SCDDs.[19,20,22] Thus, the warm-up irradiation was evaluated for the two tested SCDDs-iv both in the ^{60}Co and in the 6 MV photon beams. In 6 MV clinical photon beam, the SCDD-iv was irradiated up to an overall dose of 1200 MU (about 12 Gy), delivered in 12 steps of 100 MU (about 1 Gy) each. In the ^{60}Co beam, a maximum dose of 8 Gy was delivered in steps of 1 Gy.

Short-term stability of the detector response was evaluated from the warm-up data, by calculating the relative variation of charge readings after the fourth irradiation as well as in terms of measurement repeatability (i.e., standard deviation of detector signal) in each measurement condition considered in this work. Long-term stability was evaluated in the ^{60}Co beam by measuring the relative difference in the detector response under the same irradiation conditions over a five days period.

Fading effect was investigated in the ^{60}Co beam, up to a 30 min delay between end of irradiation and charge reading. A dose to water of about 1 Gy was delivered for each irradiation. It is worth to point out that the fading effect can be a serious concern for offline wireless IVD detectors, in which dose information must be stored in before signal is read, for times ranging from few minutes to tens of minutes.[2,13]

SCDD-iv linearity with dose was investigated in the dose range from 10 to 500 MU in the 6 MV photon beam (from about 0.1 to 5 Gy) and from 2 to 5 Gy in the ^{60}Co beam.

The dose-rate dependence of the two tested SCDDs-iv was measured in the 6 MV photon beam by varying the nominal accelerator pulse rate frequency from 50 MU/min to 500 MU/min.

Temperature dependence was evaluated in the ^{60}Co gamma-ray beam. The SCDD-iv in a PMMA holder realizing the build-up conditions was placed on the external wall of a thermostatically controlled water bath. Irradiations were performed after waiting 5 min for thermal equilibrium. As an additional check, temperature was measured both in water (T_w) and near the SCDD-iv slot (T_d). Temperature was varied in the 25–38 °C range (T_d). At each irradiation, a dose of ~1 Gy was delivered to the detector.

The angular dependence of the SCDD-iv response was measured in the ^{60}Co beam, by changing the angle of incidence of the impinging radiation beam with respect to the normal to the SCDD-iv plate surface, passing through the center of the top metallic contact (i.e., passing through the SCDD-iv measurement point). Both polar and azimuthal angular dependence were considered. Again, a dose of about 1 Gy was delivered to the detector in each irradiation.

3. RESULTS AND DISCUSSION

3.A. Preirradiation, response stability, fading effect

The preirradiation results for the two tested SCDDs-iv in the 6 MV and ^{60}Co beams are shown in Fig. 2. Charge

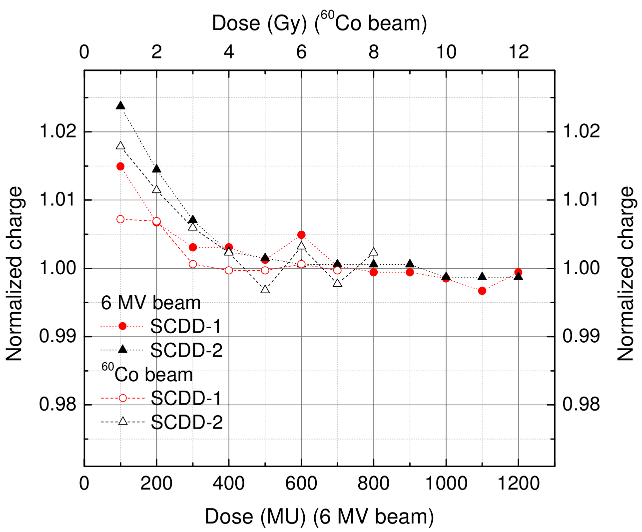

FIG. 2. Behavior of the two tested SCDDs-iv under preirradiation. Measurements in both the ^{60}Co beam (open symbols) and the 6 MV photon beam (full symbols) are reported. Charge readings for the two detectors in the two irradiation conditions are normalized to their respective plateau values.

readings are plotted vs delivered dose. The reported data are normalized to their respective plateau value, calculated as the mean value of the measured charges after the fourth irradiation (i.e., after delivering a dose of 4 Gy). Very similar trends with respect to the response of the two detectors were obtained in the two different irradiation conditions, with the SCDD-iv named SCDD-2 showing a slightly more pronounced charge decay before reaching response stability. Maximum variations of ∼2% and ∼3% can be observed for SCDD-1 and SCDD-2, respectively. All curves are within ±0.5% deviation with respect to the plateau level after a $3 \div 4$ Gy delivered dose. Such a signal stability was maintained during daily use and further characterization tests, with no need to repeat the preirradiation.

A preirradiation dose is generally required in the case of cabled SCDDs designed for external beam radiotherapy dosimetry. Indeed, a nominal preirradiation dose of 5 Gy is recommended for the PTW-Freiburg microDiamond™ dosimeter. However, it should be pointed out that the need of such a preirradiation procedure is clearly an issue for a SCDD-iv. The need of a preirradiation would imply an increase of the workload for such a dosimetric system, in view of an actual IVD clinical application. Therefore, a primingless behavior would be preferred, and a more stringent preliminary selection of the SCDDs to be embedded in the SCDD-iv system is in principle required. On the other hand, if the preirradiation is omitted, an over response (maximum 3%) can arise in the first irradiation until a dose of 3–4 Gy has been delivered. Such inaccuracy must be accounted for and compared to the accuracy level accepted for the IVD program.

A first evaluation of short-term reproducibility of the two SCDDs-iv was done from the same preirradiation data shown in Fig. 2, by calculating the RSD of the measured charges at the plateau. The response stability of SCDD-1 and SCDD-2 was, respectively, 0.2% and 0.1% (1σ) in 6 MV photon beam and 0.1% and 0.3% (1σ) in the ^{60}Co beam. These figures were consistent with the relative standard deviation of the detector signal obtained in the various irradiation conditions considered in this work. Indeed, the RSD for both detectors was in the range from 0.1% to 0.8% with a modal value of 0.3%. Long-term stability was evaluated in the ^{60}Co beam and variations in detector response of 1% for SCDD-1 and 0.4% for SCDD-2 were found over a five days period.

The results of leakage charge measurements are shown in Fig. 3. Preirradiation leakage charge vs time from charge reset, i.e., measurement of charge accumulation due to leakage dark current (no irradiation) in the detector, is shown in Fig. 3(a). Negligible leakage currents of +0.1 and −0.5 fA were calculated from the measured charges for SCDD-1 and SCDD-2, respectively. The maximum leakage charges measured in the 30 min period were 0.3% and 0.4% of the charges measured after a 1 Gy irradiation in ^{60}Co beam for SCDD-1 and SCDD-2, respectively. Postirradiation charge loss measurement after ^{60}Co irradiation, i.e., the fading effect measurement, is shown in Fig. 3(b) as a function of delay time between end of irradiation and detector reading. A similar behavior was observed for both measured SCDDs-iv, with charge losses below 0.5% up to a delay of 30 min before detector reading.

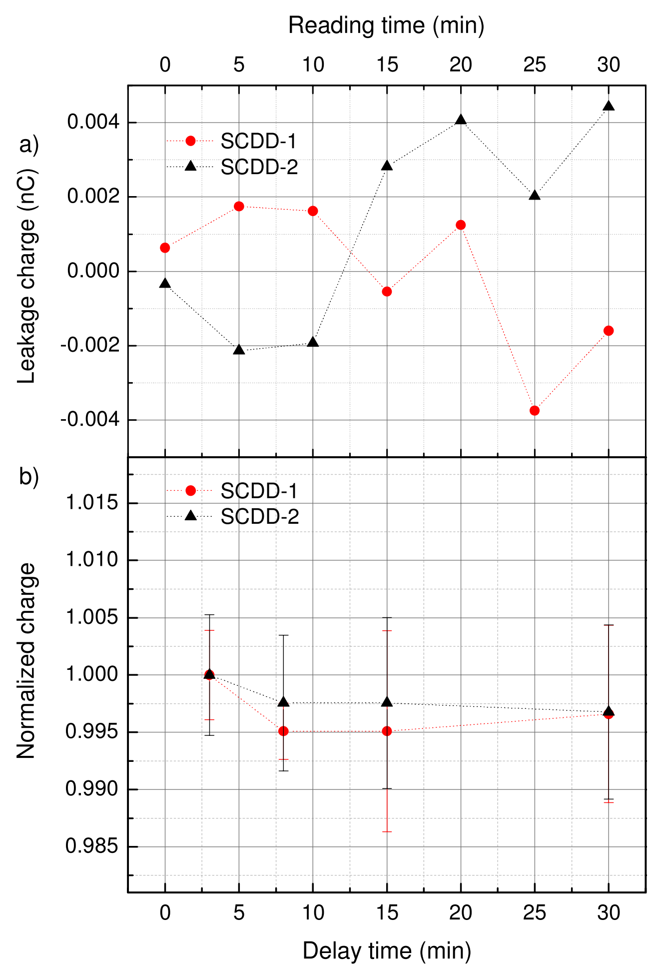

FIG. 3. Preirradiation and postirradiation leakage charge measurements: (a) accumulation of leakage charge in dark condition (no irradiation) vs time from detector charge reset, (b) charge loss by fading effect vs delay time between end of irradiation and detector reading.

As previously pointed out, fading of photogenerated charge stored in the detector unit of an offline wireless IVD system is a severe concern. Regardless the particular IVD detector used, the time delay between end of irradiation and signal reading should always be kept approximately constant.[7] For the studied SCDD-iv prototype, fading is due to the loss of photogenerated charges due to undesired leakage currents flowing during charge storage time. Such a current can be possibly attributed to insufficiently high insulation resistance of SCDD Schottky photodiode in off state, i.e., after irradiation. Indeed, once irradiated, the SCDD is forwardly biased by the voltage drop present on the charge storage capacitor plates. If the insulation resistance is not high enough, a forward current might flow, thus allowing a photogenerated charge loss. Other leak paths related to the device fabrication process and to the chosen detector elements, as well as to external causes, such as environment humidity, cannot be ruled out.

3.B. Linearity with dose and dose rate dependence

Measurements of linearity with dose for the two tested detectors in the 6 MV and ^{60}Co beams are reported in Figs. 4(a) and 4(b), respectively. In particular, Figs. 4(a.1) and 4(b.1) show the charges measured by the SCDDs-iv as a function of delivered dose. Linear best fit curves of the experimental data are also reported. In both the investigated irradiation conditions, the two measured detectors exhibit a good linear behavior, with the R^2 parameter of the linear best fit equal to 1 with a precision of 10^{-4}. Very similar sensitivities were found for both detectors in the two irradiation beams, within the error obtained by the fitting procedure.

The deviation from linearity of the detector response was evaluated as $(r_i/r_0 - 1) \times 100$, representing the percentage deviation of the ratio of measured-charge to delivered-dose ($r_i = Q_i/D_i$ or $r_i = Q_i/\mathrm{MU}_i$) with respect to the ratio obtained at the lowest delivered dose ($r_0 = Q_0/D_0$ or $r_0 = Q_0/\mathrm{MU}_0$). The results for the 6 MV and the ^{60}Co beams are shown in Figs. 4(a.2) and 4(b.2), respectively. An increasing deviation from linearity can be clearly observed as delivered dose increases. Deviations within ±2% can be observed up to 300 MU and 3 Gy doses for both detectors, whereas higher values up to 6%–8% are calculated as the delivered dose increases. Such a nonlinear behavior, which is markedly different from what reported in previous studies for cabled SCDDs,[19,20,22] is unavoidably related to the specific operation principle of the studied SCDD-iv. As a matter of fact, storing of photogenerated charge on the capacitor electrodes produces a bias voltage with opposite sign with respect to the diamond junction built-in potential. As a consequence, the higher is the accumulated charge, the lower is the electric field across the diamond junction, the depletion layer thickness, and hence the device sensitivity. This leads to an intrinsic nonlinearity of the diamond diode response. An optimization of the device performance relies on a trade-off between the need of accumulating charge and keeping the device response in the linear range, so to minimize eventual correction factors to be applied for dose determination. The range of linearity of SCDD-iv response would be enlarged by using a charge storage capacitor with

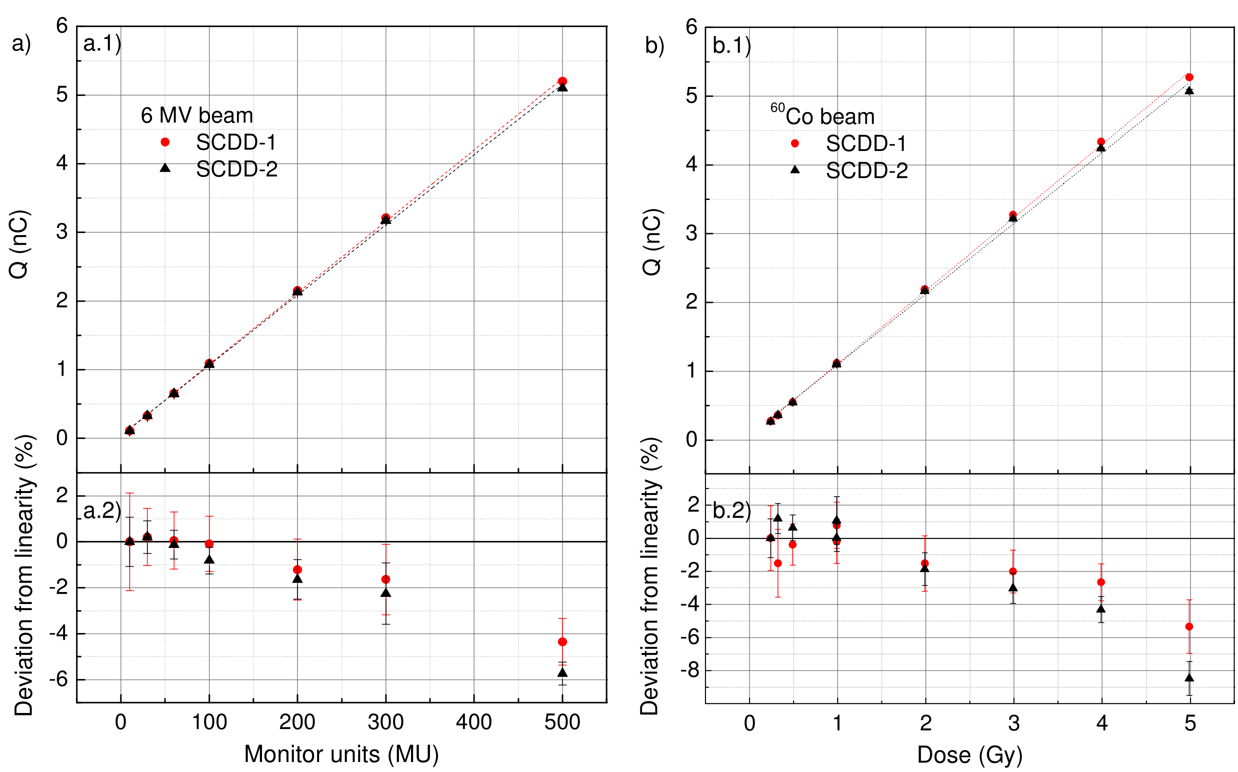

FIG. 4. Measurement of linearity with dose for the two SCDDs-iv in the 6 MV beam, (a), and in the ^{60}Co beam, (b). In the upper curves (a.1) and (b.1), measured charge vs delivered dose are reported, in which the dotted lines are the linear best fit of the measured values. The percentage deviations from linearity are reported in (a.2) and (b.2).

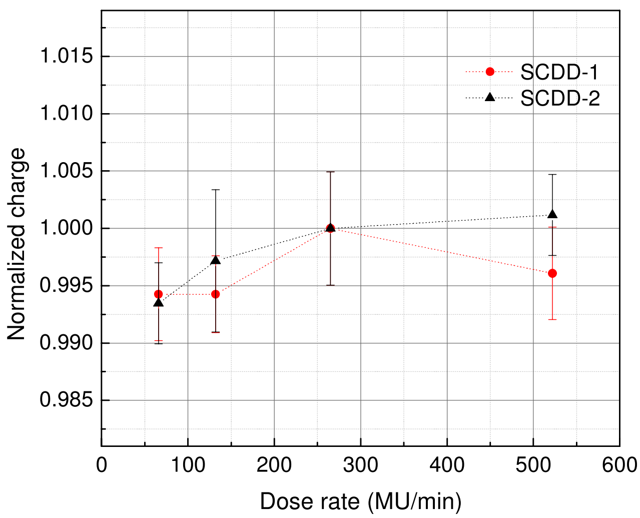

FIG. 5. Dose rate dependence of the SCDD-iv response in the 6 MV photon beam. Charge values are normalized to the value obtained at the nominal 300 MU/min (3 Gy/min) dose rate.

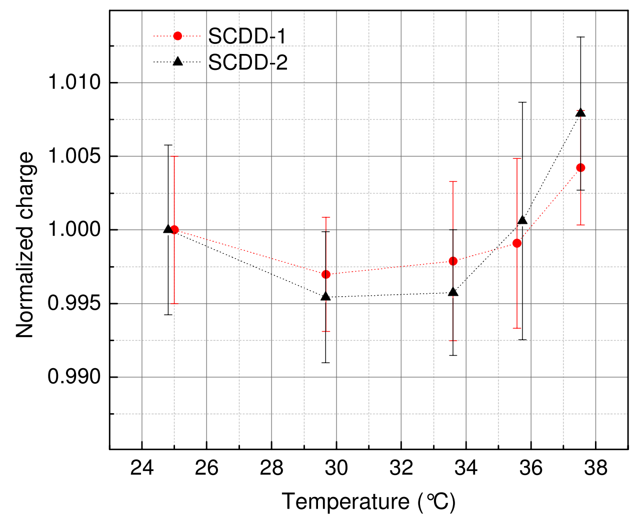

FIG. 6. Temperature dependence of the SCDD-iv response. Measured charges are normalized to the charge value obtained at 25 °C.

a higher capacitance value, which would result in a lower forward bias V_{AC} for a given delivered dose or photogenerated charge Q_{ph}. As a consequence, the reported response with dose of the tested SCDDs-iv ($C_P = 10$ nF in both cases) is to be considered a representative general behavior of any SCDD-iv, provided that a scale factor is set on the dose axis, depending on the specific storage capacitor adopted for the prototype fabrication.

As for the dose rate dependence of the SCDD-iv response, results obtained in the 6 MV photon beam are shown in Fig. 5. Measured data are normalized to those recorded at the nominal 300 MU/min (3 Gy/min) dose rate. Overall variations within 1% can be observed for both detectors.

3.C. Temperature dependence

The temperature dependence of a detector designed for IVD application is a key feature for effective clinical operation. When the detector is placed on the patient skin, its temperature can rise to skin temperature within a few minutes (~10 °C in room temperature from 2 to 3 min). Variations in detector response with temperature could be therefore a serious issue, due to the possible introduction of unacceptable errors in dose determination. The temperature dependence of the SCDD-iv response is shown in Fig. 6 in the case of ^{60}Co beam irradiation.

Charges measured from both tested detectors are normalized to their respective 25 °C average values. A similar behavior was found for both detectors with a tendency of the detector response to increase at temperatures higher than 36 °C. However, variations are within ±0.5% over all the considered temperature range from 25 to 38 °C. This SCDD-iv temperature dependence is somewhat higher than that observed for a cabled SCDD studied in a previous report (±0.2% in the 18–40 °C range).[19] Such a difference was systematically investigated in the framework of the present work and was mainly ascribed to the temperature dependence of the insulation resistance of the charge storage capacitor. Indeed, the temperature dependence was found to significantly decrease by using high performance capacitors, as previously specified in Sec. 2.A.

3.D. Angular response

Figure 7 shows the polar angular dependence of the SCDD-iv response in the whole −180° to +180° angular range. Measured charges are plotted in terms of percentage difference with respect to the response at 0°. A sketch of detector unit and impinging radiation beam direction for 0°, 90°, and 180° polar angle values is also reported. An overall variation of ~4% in the SCDD-iv response was found, with a good symmetrical behavior in the evaluated angular range. Similar results were

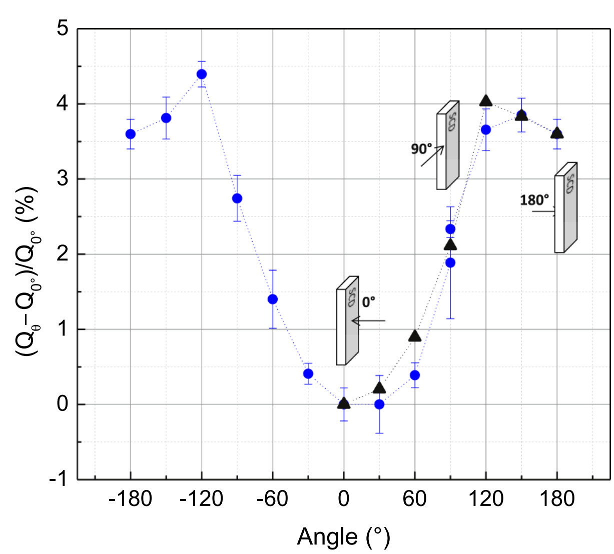

FIG. 7. Angular response of the SCDD-iv in the ^{60}Co beam. Percentage deviation of measured charges with respect to that measured at 0° and sketch of the irradiation procedure of the SCDD-iv. Average values between charge readings at corresponding angles of the 0°–180° and 0° to −180° branches are reported as black triangles.

obtained by studying the azimuthal angular dependence, and the intrinsic asymmetry in the detector structure caused by the presence of the capacitor was found to not significantly affect the angular response. These results are similar to those previously reported for a cabled SCDD,[19] which was based on the very same SCD plate, but embedded in a completely different encapsulating housing structure. Such an agreement possibly confirms the previously reported explanation of the observed behavior of the detector response vs angle of incidence. The observed angular dependence is mainly attributed to the intrinsic asymmetry of the multilayered diamond plate, with a ~1 μm thick active layer on the top of a ~400 μm thick substrate, even though a contribution from the surrounding encapsulating materials cannot be excluded. Even though the observed angular dependence is found to be relatively small, it should be accounted for in dose estimation by SCDD-iv readings with a specific angle-of-incidence correction factor, even more so in the case of complex irradiation techniques involving multidirectional beams.[23,24] However, in most IVD applications, the detector would be placed on a surface perpendicular to the radiation beam axis, so that the needed correction factors would be minimized.[7]

3.E. Uncertainty budget

Based on the above results, the uncertainty of absorbed dose measurements by means of a SCDD-iv dosimeter was estimated in the scenario of a dosimeter calibration performed in the clinical beam against a reference ionization chamber. The uncertainty budget of D_w measurements by the SCDD-iv dosimeter is reported in Table I where all the uncertainty components are given as standard deviations. In accordance with the IAEA TRS 398 dosimetry protocol,[8] the standard uncertainty of the reference absorbed dose to water measured in photon beams by means of an ionization chamber calibrated in ^{60}Co beam is estimated 1.5% (1σ). The uncertainty components related to fading, leakage, temperature, linearity, dose rate, and angular response were estimated first quantifying the corresponding largest effect on the dosimeter response then assuming a rectangular distribution for the correction. Moreover, it was assumed that the SCDD-iv dosimeter is used for D_w measurements up to 5 Gy and calibrated in the same range of D_w values. As shown in Table I, the combined relative standard uncertainty of D_w measured by the SCDD-iv dosimeter is 3.4% therefore lower than the tolerance and action level of 5%.

TABLE I. Uncertainty budget in D_w measurements by the SCDD-iv calibrated in the clinical beam against a reference ionization chamber.

Type of physical quantity	Relative standard uncertainty (%)
Step 1: SCDD-iv calibration	
$D_w(z_{\max})$ determined by a reference ionization chamber	1.5
Dosimeter reading relative to beam monitor	0.3
Fading, temperature, and leakage effects	0.2
Step 2: Dose measurement by means of SCDD-iv	
Reproducibility of dosimeter response	0.5
Dosimeter reading relative to beam monitor	0.3
Fading and leakage effects	0.2
Temperature dependence	0.3
Linearity	1.7
Dose rate effects	0.6
Directional response	2.3
Combined relative standard uncertainty of D_w	**3.4**

4. CONCLUSIONS

In this paper, a novel synthetic single crystal diamond Schottky photodiode prototype developed at the laboratories of Tor Vergata University in Rome was evaluated for *in vivo* dosimetry applications. A compact, low cost detector assembly, consisting of a small unwired detector unit and an external reading unit, was developed and tested. If compared to other available wireless IVDs , the SCDD-iv is not disposable and requires negligible workload for reading. A basic dosimetric characterization of detector performance relevant for IVD application was performed on two nominally identical SCDDs-iv under irradiation with ^{60}Co and 6 MV photon beams. Preirradiation procedure, response stability, leakage charge, fading effect, linearity with dose, dose rate dependence, temperature dependence, and angular response were investigated.

In both ^{60}Co and 6 MV photon beams, the tested detectors needed a 3 ÷ 4 Gy preirradiation dose to settle their response within ±0.5% deviation with respect to the plateau value. A maximum variation of about 3% was observed in the preirradiation behavior in the 6 MV photon beam. The variation became smaller in size (between 0.8% and 1.8%) at the ^{60}Co gamma-ray beam. Good response stability, below 0.5% (1σ), was found for both detectors in the two irradiation setups. Variations in detector response up to 1% were found in a five-day measurement period in the Co-60 beam.

Negligible leakage current, below 1 fA, was measured for both detectors. No significant fading effect occurred during charge storage, with charge losses below 0.5% up to a delay of 30 min between the end of irradiation and the charge reading. Good linear behavior was found for both SCDDs-iv, with deviations from linearity within ±2% up to ~3 Gy delivered dose. Higher deviations up to 8.5% were found as dose increased up to 5 Gy, due to intrinsic nonlinear behavior of the device response. A negligible dose-rate dependence was found in the 6 MV photon beam, with both detectors showing overall variations below 1%. A variation within ±0.5% was observed when evaluating the temperature dependence of the detector response in the 25–38 °C range. An overall variation of ~4% in the SCDD-iv response was found in angular dependence measurements from 0° to 180°, as expected from the asymmetrical device geometry.

Some dosimetric features of the new SCDD-iv unwired detector design were found to be slightly worse if compared to previously reported results from the analogous standard cabled SCDD dosimeter in similar characterization tests. These differences were basically attributed to the different operation mechanism of such a detectors, as well as to the novel device design intended for IVD operation. More specifically, the use of a commercial charge storage capacitor resulted to be a critical issue for the overall detector operation. Work is in progress to overcome the device performance limitation connected with this issue.

From the uncertainty budget of D_w measurements by the SCDD-iv, it appears that the combined relative standard uncertainty of D_w resulted 3.4%, lower than the tolerance and action level of 5%.

The SCDD-iv is simple, with no cables linked to the patient, the readout is immediate, the range of response with dose has been tested from 1 up to 12 Gy, the reading is independent of the accumulated dose, independent of dose rate between 0.5 Gy/min and about 5 Gy/min, independent of temperature within 0.5% between 25 and 38 °C, and directionally independent within 2% from 0° to 90°.

The reported results indicate the proposed novel synthetic single crystal diamond Schottky photodiode as a promising candidate for *in vivo* dosimetry applications with photon beams.

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[1]ICRU (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 No. 24 (ICRU, Bethesda, MD, 1976).

[2]M. Essers and B. J. Mijnheer, "*In vivo* dosimetry during external beam photon beam radiotherapy," Int. J. Radiat. Oncol., Biol., Phys. **43**, 245–259 (1999).

[3]European Society of Radiation Oncology, "Practical guidelines for the implementation of *in vivo* dosimetry with diodes in external radiotherapy with photon beams (entrance dose)," ESTRO Booklet No. 5 (European Society of Radiation Oncology, ESTRO, Brussels, Belgium, 2001).

[4]Report of TG 62 of the Radiation Therapy Committee, "Diode *in vivo* dosimetry for patients receiving external beam radiation therapy," AAPM Report No. 87 (Medical Physics Publishing, Madison, WI, 2005).

[5]B. Mijnheer, S. Beddar, J. Izewska, and C. Reft, "*In vivo* dosimetry in external beam radiotherapy," Med. Phys. **40**, 070903 (9pp.) (2013).

[6]G. J. Millwater, A. S. Macledo, and D. I. Thwaites, "*In vivo* semiconductor dosimetry as part of routine quality assurance," Br. J. Radiol. **71**, 661–668 (1998).

[7]International Atomic Energy Agency, "Development of procedures for *in vivo* dosimetry in radiotherapy," IAEA Human Health Report No. 8. (IAEA, Vienna, Austria, 2013).

[8]International Atomic Energy Agency, "Absorbed dose determination in external beam radiotherapy: An international code of practice for dosimetry based on standards of absorbed dose to water," Technical Report Series No. 398 (IAEA, Vienna, Austria, 2000).

[9]Th. Loncol, J. L. Greffe, S. Vynckier, and P. Scalliet, "Entrance and exit dose measurements with semiconductors and thermoluminescent dosimeters: A comparison of methods and *in vivo* results," Radiother. Oncol. **41**, 179–187 (1996).

[10]G. Leunens, J. Van Dam, A. Dutreix, and E. van der Schueren, "Quality assurance in radiotherapy by *in vivo* dosimetry. 2. Determination of the target absorbed dose," Radiother. Oncol. **19**, 73–87 (1990).

[11]R. P. Patel, A. J. Warry, D. J. Eaton, C. H. Collis, and I. Rosenberg, "*In vivo* dosimetry for total body irradiation: Five-year results and technique comparison," J. Appl. Clin. Med. Phys. **4**, 306–315 (2014); available at http://www.jacmp.org/index.php/jacmp/article/view/4939.

[12]G. Guidi, G. Gottardi, P. Ceroni, and T. Costi, "Review of the results of the *in vivo* dosimetry during total skin electron beam therapy," Rep. Pract. Oncol. Radiother. **19**, 144–150 (2014).

[13]J. Van Dam and G. Marinello, *Methods for In Vivo Dosimetry in External Radiotherapy, ESTRO Booklet No.1*, 2nd ed. (European Society for Radiation Oncology, ESTRO, Brussels, Belgium, 2006).

[14]B. Mijnheer, "State of the art of *in vivo* dosimetry," Radiat. Prot. Dosim. **131**, 117–122 (2008).

[15]A. Ismail, J. Y. Giraud, G. N. Luc, R. Sihanath, P. Pittet, J. M. Galvan, and J. Balosso, "Radiotherapy quality insurance by individualized *in vivo* dosimetry: State of the art," Cancer/Radiothérapie **13**, 182–189 (2009).

[16]M. Soubra, J. Cygler, and G. Mackay, "Evaluation of a dual bias dual metal oxide-silicon semiconductor field effect transistor detector as radiation dosimeter," Med. Phys. **21**, 567–572 (1994).

[17]P. H. Halvorsen, "Dosimetric evaluation of a new design MOSFET *in vivo* dosimeter," Med. Phys. **32**, 110–117 (2005).

[18]C. W. Scarantino, D. M. Ruslander, C. J. Rini, G. G. Mann, H. T. Nagle, and R. D. Black, "An implantable radiation dosimeter for use in external beam radiation therapy," Med. Phys. **31**, 2658–2671 (2004).

[19]I. Ciancaglioni, M. Marinelli, E. Milani, G. Prestopino, C. Verona, G. Verona-Rinati, R. Consorti, A. Petrucci, and F. De Notaristefani, "Dosimetric characterization of a synthetic single crystal diamond detector in clinical radiation therapy small photon beams," Med. Phys. **39**, 4493–4502 (2012).

[20]C. Di Venanzio, M. Marinelli, E. Milani, G. Prestopino, C. Verona, G. Verona-Rinati, M. D. Falco, P. Bagalà, R. Santoni, and M. Pimpinella, "Characterization of a synthetic single crystal diamond schottky diode for radiotherapy electron beam dosimetry," Med. Phys. **40**, 021712 (9pp.) (2013).

[21]S. Almaviva, M. Marinelli, E. Milani, G. Prestopino, A. Tucciarone, C. Verona, G. Verona-Rinati, M. Angelone, M. Pillon, I. Dolbnya, K. Sawhney, and N. Tartoni, "Chemical vapor deposition diamond based multilayered radiation detector: Physical analysis of detection properties," J. Appl. Phys. **107**, 014511 (2010).

[22]W. U. Laub and R. Crilly, "Clinical radiation therapy measurements with a new commercial synthetic single crystal diamond detector," J. Appl. Clin. Med. Phys. **15**, 92–102 (2014); available at http://www.jacmp.org/index.php/jacmp/article/view/4890.

[23]S. J. Alnaghy, S. Deshpande, D. L. Cutajar, K. Berk, P. Metcalfe, and A. B. Rosenfeld, "*In vivo* endorectal dosimetry of prostate tomotherapy using dual MOSkin detectors," J. Appl. Clin. Med. Phys. **16**, 107–117 (2015); available at http://www.jacmp.org/index.php/jacmp/article/view/5113.

[24]I. S. Kwan, A. B. Rosenfeld, Z. Y. Qi, D. Wilkinson, M. L. F. Lerch, D. L. Cutajar, M. Safavi-Naeni, M. Buston, J. A. Bucci, Y. Chin, and V. L. Perevertaylo, "Skin dosimetry with new MOSFET detectors," Radiat. Meas. **43**, 929–932 (2008).