Dosimetric performance and array assessment of plastic scintillation detectors for stereotactic radiosurgery quality assurance

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(Received 20 July 2011; revised 14 November 2011; accepted for publication 14 November 2011; published 28 December 2011)

Purpose: To compare the performance of plastic scintillation detectors (PSD) for quality assurance (QA) in stereotactic radiosurgery conditions to a microion-chamber (IC), Gafchromic EBT2 films, 60 008 shielded photon diode (SD) and unshielded diodes (UD), and assess a new 2D crosshair array prototype adapted to small field dosimetry.
Methods: The PSD consists of a 1 mm diameter by 1 mm long scintillating fiber (BCF-60, Saint-Gobain, Inc.) coupled to a polymethyl-methacrylate optical fiber (Eska premier, Mitsubishi Rayon Co., Ltd., Tokyo, Japan). Output factors ($S_{c,p}$) for apertures used in radiosurgery ranging from 4 to 40 mm in diameter have been measured. The PSD crosshair array (PSDCA) is a water equivalent device made up of 49 PSDs contained in a 1.63 cm radius area. Dose profiles measurements were taken for radiosurgery fields using the PSDCA and were compared to other dosimeters. Moreover, a typical stereotactic radiosurgery treatment using four noncoplanar arcs was delivered on a spherical phantom in which UD, IC, or PSD was placed. Using the Xknife planning system (Integra Radionics Burlington, MA), 15 Gy was prescribed at the isocenter, where each detector was positioned.
Results: Output Factors measured by the PSD have a mean difference of 1.3% with Gafchromic EBT2 when normalized to a 10×10 cm^2 field, and 1.0% when compared with UD measurements normalized to the 35 mm diameter cone. Dose profiles taken with the PSD crosshair array agreed with other single detectors dose profiles in spite of the presence of the 49 PSDs. Gamma values comparing 1D dose profiles obtained with PSD crosshair array with Gafchromic EBT2 and UD measured profiles shows 98.3% and 100.0%, respectively, of detector passing the gamma acceptance criteria of 0.3 mm and 2%. The dose measured by the PSD for a complete stereotactic radiosurgery treatment is comparable to the planned dose corrected for its SD-based $S_{c,p}$ within 1.4% and 0.7% for 5 and 35 mm diameter cone, respectively. Furthermore, volume averaging of the IC can be observed for the 5 mm aperture where it differs by as much as 9.1% compared to the PSD measurement. The angular dependency of the UD is also observed, unveiled by an under-response around 2.5% of both 5 and 35 mm apertures.
Conclusions: Output Factors and dose profiles measurements performed, respectively, with the PSD and the PSDCA were in agreement with those obtained with the UD and EBT2 films. For stereotactic radiosurgery treatment verification, the PSD gives accurate results compared to the planning system and the IC once the latter is corrected to compensate for the averaging effect of the IC. The PSD provides precise results when used as a single detector or in a dense array, resulting in a great potential for stereotactic radiosurgery QA measurements. © *2012 American Association of Physicists in Medicine*. [DOI: 10.1118/1.3666765]

Key words: plastic scintillation detector, detector array, radiation detector, small field, stereotactic radiosurgery

I. INTRODUCTION

Stereotactic radiosurgery (SRS) is a noninvasive treatment that can be administered to treat intra and extra cranial lesions in a single fraction with a high spatial precision. Linac-based SRS treatment can treat tumors up to 4 cm diameter usually delivering a dose ranging from 12 to 24 Gy depending on the tumor type and size. In order to decrease the dose to healthy tissues, this modality delivers the ionizing radiation using noncoplanar arcs. To precisely locate the tumor, stereotactic localization is performed prior to the treatment when the planning computed tomography images are taken. The precise localization is very important, since the complete treatment dose is delivered at once using very steep dose gradients to spare the surrounding healthy tissues.[1]

If positioning precision is of a great importance in SRS, dosimetry accuracy is as much crucial in order to deliver the treatment as planned. One of the major aspect that makes SRS dosimetry difficult is the presence of small fields used during treatment. A good definition of the concept of small field has been used in the literature[2,3] and consisted in establishing the threshold when the lateral charged particle equilibrium (CPE) on the central axis is lost. For a 6 MV beam, it occurs for fields of 3×3 cm^2 and smaller. In lateral charged particle disequilibrium, dosimetry becomes more complex for several reasons. Due to loss of CPE, as a field gets smaller, the flat region of the dose profiles will decrease in size to finally become inexistent. The output factor ($S_{c,p}$) will also decrease significantly due to a smaller contribution of secondary photons on the central axis. Furthermore, if the detector placed in the irradiation field is not water equivalent, its response will change relative to what a water equivalent detector would measure. This perturbation effect will depend on the photon and electron energy spectrum, which can vary simply by changing the depth of measurement or the size of the field. This phenomenon is more important for small fields, because the perturbed area is larger compared to the whole field area. Furthermore, the presence of a nonwater equivalent dosimeter in a small field such as an ion chamber or a diode having a smaller or greater electronic density than water will decrease or increase the lateral CPE artificially and affect the dose readout.

Due to those multiple complications, radiosurgery field dosimetry remains a challenging task, and there is still no detector that can be considered ideal for this purpose and consequently for SRS quality assurance (QA). Indeed, in order to do adequate measurement in these conditions, more than one detector is often required to overcome specific drawbacks of each dosimeter in particular situations. As a possible solution, we propose a plastic scintillation detector (PSD), which contains numerous advantages for small field dosimetry and linac-based SRS QA. Characterization of the response of the PSD has been described previously in different papers[4–7] for conventional radiotherapy beams. Other studies have been published[8–11] regarding radiosurgery beams measurements with a PSD, and many of them used a background fiber in order to subtract Cerenkov radiation next to the fiber collecting the scintillation light emitted. However, our prototype is composed of a single optical fiber and Cerenkov radiation produced in the detector is removed using spectral discrimination[12–15] resulting in a higher spatial resolution which is critical for the dose measurement in small field conditions. Likewise, a 2D-matrix benefits from this improvement allowing a higher detector density. To depict its feasibility applied for stereotactic radiosurgery, a 2D crosshair array has been built and evaluated taking profile measurements. Furthermore, a complete SRS treatment has been planned on the planning system Xknife (Integra Radionics Burlington, MA) and measured with various detectors inserted in a phantom, including the PSD, in order to assess the efficiency of each detector to perform a SRS QA.

II. MATERIAL AND METHODS

II.A. Detectors

In small field dosimetry, a number of different factors may influence the final output data of a given detector. Therefore, conception differences between detectors need to be properly considered in order to analyze adequately the readings obtained. Five different detectors have been used in this study and are shown in Fig. 1.

II.A.1. PSD

The PSD used in this study is composed of a 1 mm diameter by 1.0 ($\pm$0.1) mm long polystyrene scintillating fiber (BCF-60, Saint-Gobain Crystals, Paris, France) for a total sensitive volume of 7.8×10^{-4} cm^3, coupled with a polymethylmethacrylate Eska premier optical fiber[16] (Mitsubishi, Rayon Co., Ltd., Tokyo, Japan). Each single PSD is coated by a polyethylene jacket to increase the toughness of the detector. The light produced is collected by a RGB CCD camera (the Alta U2000c from Apogee Instruments, Inc., Auburn, CA) in order to subtract the Cerenkov radiation using a spectral discrimination technique.[12–15]

II.A.2. A16 Exradin microionization chamber (IC)

The A16 Exradin microionization chamber (IC) (Standard imaging, Middleton, WI) has been used with a polarization high voltage of −300 V. This ionization chamber's shell, collector and guard material are made of Shonka air-equivalent

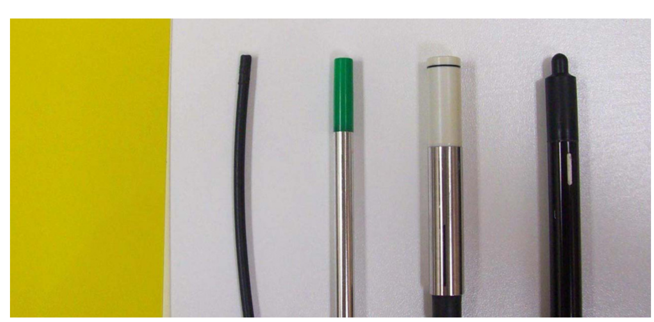

FIG. 1. Dosimeters used in the study, from left to right: Gafchromic EBT2, PSD, SFD stereotactic UD, 60 008 shielded photon diode (SD), and A16 Exradin microionization chamber (IC).

plastic and its stem is made of aluminum. Its collecting volume is 0.007 cm^3 and the length of it cross-section facing the beam measures 2.5 mm.

II.A.3. 60 008 shielded photon diode (SD)

The p-type silicon 60 008 shielded photon diode [Physikalisch-Technische Werkstatten (PTW), Freiburg, Germany] has a sensitive volume of 2.5×10^{-6} cm^3 and a diameter of 1.12 mm. Being made out of silicon ($Z = 14$) the sensitive volume of this detector over-respond to low energy scattered photons due to increase of photoelectric effect interactions with the sensitive material. In order to compensate this effect, the silicon chip is embedded in a metallic plate (made of tungsten) in order to partially block the low energy scattered photons.[17]

II.A.4. SFD stereotactic unshielded diode (UD)

The SFD stereotactic UD (Scanditronix AB, Uppsala, Sweden) is a p-type silicon diode having a 1.7×10^{-5} cm^3 sensitive volume of 0.6 mm of diameter. This diode is not shielded which causes an over-response to low energy scattered photons in large fields.

II.A.5. Gafchromic EBT2

The Gafchromic EBT2 (ISP, NJ) is a radiochromic film having active components shaped as needlelike particles of 1–2 μm in diameter and 15–20 μm in length. However, the resolution of the dose distribution is driven by the resolution of the scanner used. In this study, the scanning has been performed with an EPSON EXPRESSION 10000XL having a resolution of 150 dpi, resulting in a 0.17 mm pixel size. For each $S_{c,p}$ measurement, 21 pixels in the center of the field are averaged corresponding to a 0.85 mm diameter detector measurement (5 pixels of diameter). For dose profile measurements, each point of measurement corresponds to one pixel. Gafchromic EBT2 films are principally made of carbon (42.37%), hydrogen (40.85%), and oxygen (16.59) for a Z_{eff} of 6.84. Scanner homogeneity response was characterized using a preirradiated film with doses ranging from 0 to 3 Gy. The scanning was performed taking RGB scans in transmission mode and keeping only the information coming from the red channel (16 bits). Films were calibrated using 15 different fields of known dose distributed on 3 Gafchromic films of the same batch as those taken for measurement to interpolate a calibration curve. Films were kept in a light tight envelop after the irradiation and were all scanned after the same autodeveloping time of 24 h.

II.A.6. PSD crosshair array

The PSD crosshair array (PSDCA) is composed of 49 unjacketed PSDs forming two perpendicular lines of 32.5 mm of 25 PSDs (Fig. 2) each line having 1.3 mm detector spacing. All detectors are inserted in a 1.25 cm thick plastic water slab (CIRS, Norfolk, VA) and are coupled to the RGB CCD camera with optical fibers. Each detector is recovered

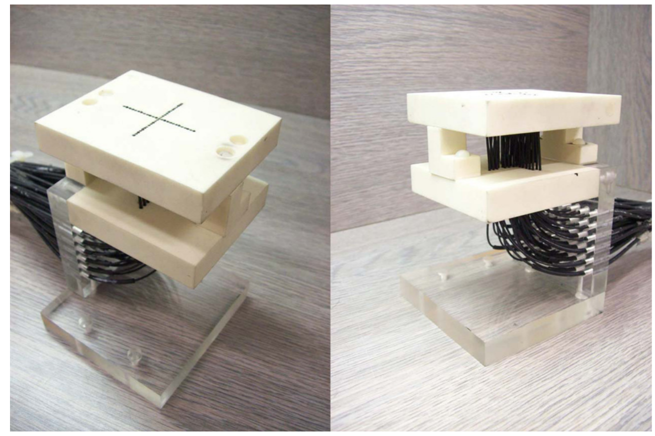

FIG. 2. Photo of the PSD crosshair array prototype containing 49 PSDs. The upper half is made in plastic water slab and the bottom half in acrylic.

of black acrylic paint to eliminate the crosstalk that would occur, if scintillating light was allowed to penetrate into the surrounding PSDs. To perform measurements, the prototype is immerged in water. The first step of the calibration process of the PSDCA is similar to the calibration of a single PSD.[12–15] It consists of two measurements with different optical fibers length irradiated, in order to obtain a "Cerenkov light ratio" that is used to remove the Cerenkov light from the scintillation light for every detector (see Guillot *et al.*[15] for more details and comparison of calibration methods). The second step is to calibrate accurately each PSD with a known dose using a reference field. For this study, a 10×10 cm^2 reference field has been used. To obtain the dose distribution along both axes of the crosshair array, an Exradin A12 farmer chamber (Standard imaging, Middleton, WI) is used to perform the measurement prior to the PSDCA calibration. The same reference field is then used to calibrate the PSDCA in a single irradiation, by extracting a coefficient factor relating the light output from each PSD to the specific dose it receives.

II.B. Measurements

Measurements were taken on a 6 MV Siemens Mevatron linac. To investigate the behavior of each detector for small field dosimetry, $S_{c,p}$ have been measured with each of the five detectors for circular collimators (cones) used in stereotactic radiosurgery with diameters ranging from 4 to 40 mm, the detector being positioned at the isocenter at a 1.8 cm depth. The $S_{c,p}$ is defined as the ratio of dose measured for a given cone diameter over the dose of a reference field size. Two different reference field sizes were used to normalize the measurements: a 10×10 cm^2 field and a 35 mm diameter cone. Three to five $S_{c,p}$ measurements per field were done with each detector. For dose profiles, ten measurements per point were taken for the IC, the PSD, the SD, and the UD. Five different EBT2 films measurements were averaged to obtain each dose profile per field. Each field used to perform $S_{c,p}$ and dose profile measurements were set to 200 monitor units for all detectors. Longitudinal profiles were taken at a

5 cm depth at the isocenter for the same five dosimeters. Each profile measurement where centered using the middle of the full width half maximum and the central dose have been normalized to 1.000. The microionization chamber and the EBT2 films used for profiles and $S_{c,p}$ measurements were positioned perpendicular to the beam axis, while the two diodes and the PSDCA (as shown in Fig. 2) were placed parallel to the beam axis. Measurements were achieved in a motorized water tank (Blue Phantom, IBA Dosimetry America, Bartlett, TN) having a reproducibility of ± 0.1 mm. To increase the number of points per dose profiles taken with the PSDCA, two translations of 0.4 mm have been done for the 4 mm and 10 diameter cone. To cover the entire 40 mm diameter cone dose profile, a 24.7 mm translation was applied to the PSDCA. When a gamma evaluation is made on a reference curve having a higher number of measurement points per distance than the other curve, smaller gamma values will be found if the reference curve is noisy.[18] Because the dose profiles have been done to compare the dose measured by the PSDCA with the UD and the Gafchromic EBT2 films, smoothing those two curves gives a more accurate gamma comparison with the PSDCA measurement. Thus, to perform the gamma evaluation, Gafchromic EBT2 and UD dose profiles have been smoothed using a Savitzky–Golay smoothing filter,[19,20] to get rid of small fluctuations (of the order of 1%) observed in the plateau of the 40 mm cone. The center of radiation was found for $S_{c,p}$ and dose profiles doing one inline and one crossline scan prior to each measurement for every detector.

An SRS treatment has been planned using the Xknife planning system and delivered on a phantom simulating a patient head (see Fig. 3). It must be pointed out that the treatment planning system has been commissioned using the shielded diode to measure the $S_{c,p}$. This implies that any treatment plan will be affected by this imperfect measurement (mainly due to the nonwater equivalence of the SD detector). Prior to the planning session, three computed tomography (CT) scans of the phantoms have been performed; each of them including a different dosimeter (either the IC, the UD, or the PSD) inserted approximately 8 cm deep. Each detector was enclosed in a water equivalent adaptor made of polyflex (Austenal, Inc., Chicago, IL) in order to remove air gaps between the detector and the phantom. The similar treatment plan, calculated on their respective CT images, was delivered to all three detectors and consisted in four noncoplanar arcs. 15 Gy was prescribed at the isocenter, where the effective measurement point of each detector was precisely placed. Arcs have been distributed in order to decrease the presence of high gradients regions at the isocenter and its close neighborhood to reduce the impact of a small position error between the dosimeter and the radiation beam (that could be caused by factors such as sag effect of the gantry). Two measurements per dosimeter where taken using cones of 5 and 35 mm diameter. Regarding the positioning of the detector when performing the QA of the SRS plan, radiation isocenter was found by doing three scans always perpendicular to the beam axis in the three dimensions. Those scans were done by moving the head phantom with the submillimetric adjustments of the stereotactic couch mount adapter. This procedure was also repeated before each measurement of a SRS plan and for the three dosimeters. The center was always found using the middle of the full width half maximum for each scan that was done with the Xknife system having a ± 0.1 mm precision.

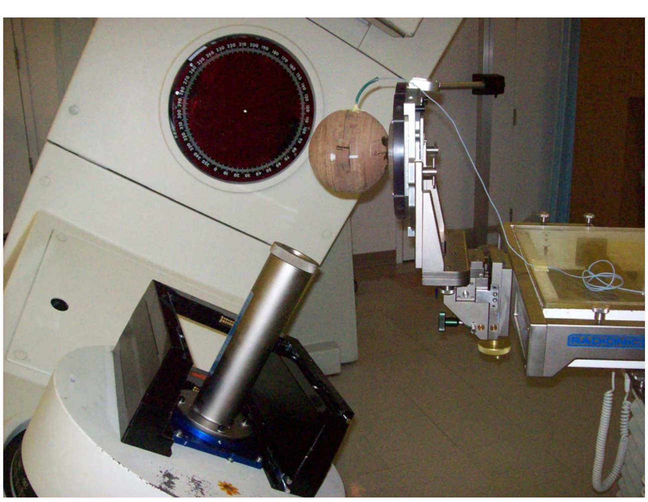

FIG. 3. Setup for the measurements of the complete treatment plan of stereotactic radiosurgery on the phantom containing a dosimeter (here the UD).

III. RESULTS AND DISCUSSION

Output factors measured with all detectors at different field sizes are shown in Fig. 4. In Fig. 4(a), each $S_{c,p}$ have been normalized to a 10×10 cm^2 field and in Fig. 4(b), the $S_{c,p}$ have been normalized to the 35 mm diameter cone. As mentioned previously, detector design can have an important impact on detector measurements, and it has to be taken into account while interpreting curves in Fig. 4(a). The IC shows an increasing under-response compared to the other detectors as the field size is reduced due to the width of the cross-section of the sensitive volume faced by the beam, causing dose averaging in small fields. The photon SD has been optimized to provide measurements in agreement with an ion chamber in large field conditions by embedding the sensitive volume in a metallic plate. However, when a large field is compared to a small field, the mean energy of the beam is higher in the small field.[21] This is due to smaller contribution of the low energy scattered photons and the loss of CPE resulting in a dose deposited in the detector that comes more from higher energy electrons compared to large fields. Another effect arises from the diode nonwater equivalent material. Because the lateral range is shorter in silicon and in the shielding material (tungsten) than in water, the lateral CPE remains for smaller field size than in a water equivalent detector. This effect implies an over-response compared with detector made of water equivalent material[5] therefore explaining the over-response obtained with the SD in apertures smaller than 20 mm of diameter compared to the other dosimeters investigated. Francescon *et al.*[22] found correction factors to apply to the $S_{c,p}$ obtained with the 60 008 shielded diode for small collimator sizes ranging from 5 to 10 mm of

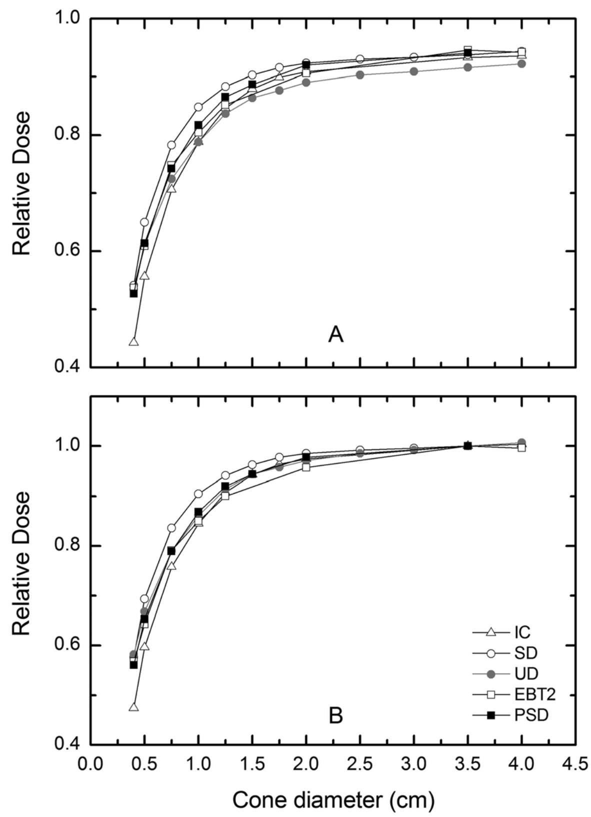

FIG. 4. Output factors of radiosurgery cones measured at 1.8 cm depth. Measurements are made with a PSD, a IC, a SD, an UD, and Gafchromic EBT2 (a) normalized to a 10×10 cm^2 field, (b) normalized to the 35 mm diameter cone. Error bars are smaller than symbols.

diameter of a CyberKnife radiosurgery system. This factor is to compensate the higher proportion of scatter near the active region due to material having higher atomic number than water in its neighborhood. The correction factors obtained in the Francescon *et al.*[22] paper were applied to the SD measurements (there is no correction factor applied on the SD data shown in Fig. 4) even if they were measured on a CyberKnife system and using a different PTW 60 008 shielded diode to have a first approximation of a corrected $S_{c,p}$. When the corrected $S_{c,p}$ were compared to the $S_{c,p}$ measurements taken with the PSD, the agreement was better than 1% for the 5, 7.5, and 10 mm cone. The UD has been built especially for small field measurements. Since this diode is not shielded, it over-responds in large fields. When the $S_{c,p}$ measurements of the SRS cones are normalized to a 10×10 cm^2 field, the latter bias the results, leading to an undermeasurement of the $S_{c,p}$. However, it can be overcome by normalizing the $S_{c,p}$ to a smaller field such as the 35 mm diameter cone as shown in Fig. 4(b). The presence of high Z material near the sensitive zone in the UD is considerably smaller compared to the SD since there is no shielding plate and the silicon chip size is smaller. However, this can still cause a small over response for that detector.

Both diodes measure the dose deposited in silicon and not in water. The paper from Eklund and Ahnesjo[23] proposes a correction method to take consideration of this issue. Nevertheless, it is required to know correction factors (scatter factors) specific to the measurements conditions for each beam. This approach can become more complex in clinical situation, for QA of SRS, IMRT, or SRT plan where multiple different correction factors have to be applied and can be seen as a disadvantage compared to other dosimeters that does not need correction such as the PSD.

Regarding Gafchromic EBT2 film and PSD, they have no disadvantage *a priori* for SRS measurements because they are water equivalent, have a good spatial resolution and no angular and energy dependence. However, the PSD have a major advantage compared to the Gafchromic films which is its capability of doing real-time measurements. When the PSD is compared with the Gafchromic measurements, $S_{c,p}$ have a mean difference of 1.3%. A similar comparison can be made between the PSD and the UD when measurements are normalized to a 35 mm cone (in order to take account for the energy dependence of the diode) to obtain a mean discrepancy of 1%. If only large fields are considered, the PSD can be compared to the SD, the IC and Gafchromics, all four detectors are within a 1.3% margin for the 35 mm cone. Finally, the response of PSD has been evaluated with Monte Carlo by Wang and Beddar[24] and has shown variations within 0.4% for field sizes ranging from 0.5×0.5 cm^2 to 10×10 cm^2.

Dose profile measurements of 4, 10, and 40 mm diameter cones depict the capacity of a detector to measure dose gradients (Figs. 5–7). Dose profiles of the 4 and 10 mm aperture clearly show the averaging effect of the IC in gradient dose distribution. In order to assess dose profiles measurements obtained by the PSD with dose profiles from other detectors, gamma evaluation[25] has been conducted. Because Gafchromic EBT2 and the UD have small sensitive area, they are considered as a good reference detector for dose profiles comparison with the PSD crosshair array. Hence, gamma values have been calculated using acceptance criteria of 0.3 mm and 2%, and the results are presented in Table I. When compared to Gafchromic EBT2, the PSD crosshair array has 98.3% of its detectors passing the acceptance criteria over the combined three radiosurgery aperture profiles. The detectors that failed have a 1.07 and 1.16 gamma value that represents a

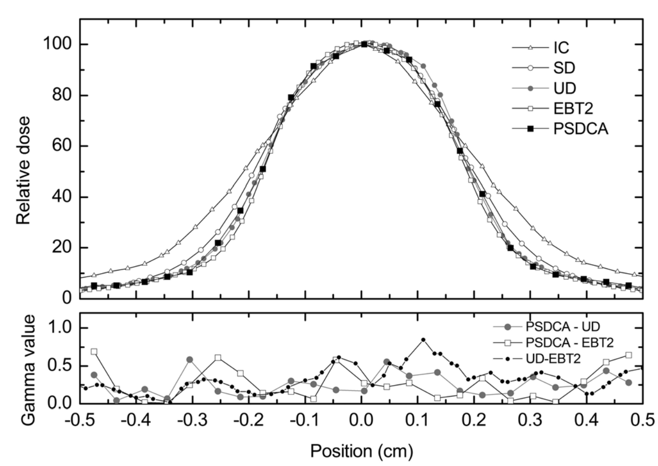

FIG. 5. Dose profiles of 4 mm diameter radiosurgery cone measured at 5 cm depth and normalized to the dose measured at the center of the field for each detector. Error bars are smaller than symbols.

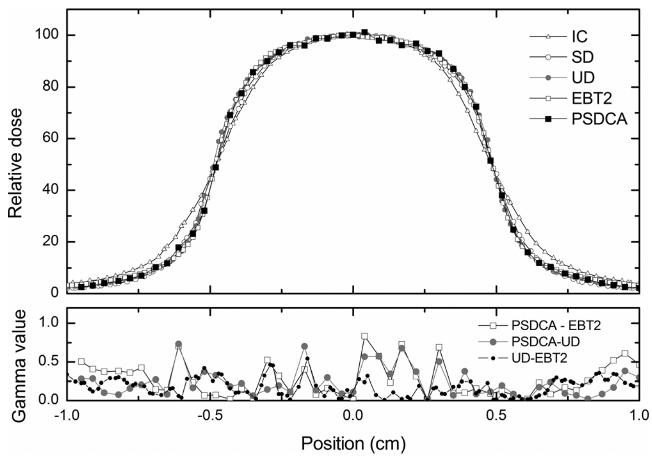

FIG. 6. Dose profiles of 10 mm diameter radiosurgery cones measured at 5 cm depth and normalized to the dose measured at the center of the field for each detector. Error bars are smaller than symbols.

2.14% and 2.32% offset, if the difference is considered to be completely dosimetric. These two points of measurements are located on either side of the bottom region of the penumbra of the 40 mm circular collimator dose profile. All PSD crosshair array detectors pass the gamma evaluation when they are compared to the UD. The mean gamma value of 4 and 10 mm apertures compared with the Gafchromic films, and the UD is around 0.25, which can be interpreted as a 0.5% mean dose difference if the positioning component is considered negligible. Similarly, a 0.39 average gamma value is obtained when comparing the 40 mm cone dose profiles and represents a 0.78% mean dose difference.

Table II summarizes the results obtained after doing a typical stereotactic radiosurgery treatment on the PSD, the UD, and the IC. The $D_{measured}/D_{prescribed}$ ratio shows the dose measured with each detector over the dose prescribed in the treatment planning system, for 5 and 35 mm cones. For the 5 mm cone, when the IC measurements are compared with the PSD, the IC is under responding by 9.1%, which is consistent with the discrepancy observed in $S_{c,p}$ measurements and can be explained by volume averaging of the IC. To compare each detector with the treatment planning system, the latter have to be corrected since it has been commissioned using the SD and this detector is known to over-respond in small field due to its composition. $S_{c,p}$ ratio of the investigated detector (X) and the SD have to be applied to correct the bias induced by the latter dosimeter. After applying this correction, each detector measurement can be compared with the prescribed dose; however, any discrepancies of the measured dose of the radiosurgery treatment due to $S_{c,p}$ will be corrected too in the $D_{measured} \cdot S^{X}_{c,p}/D_{prescribed} \cdot S^{SD}_{c,p}$ ratio. In so doing, the latter ratio for the IC has a 1% discrepancy with the planned dose since the averaging effect of the IC in the 5 mm cone is compensated in the $S_{c,p}$ ratio. For the PSD, the measured dose of the treatment plan is in agreement with the treatment planning system within 1.4% for the 5 mm cone and 0.7% for the 35 mm cone. Comparing the $D_{measured} \cdot S^{X}_{c,p}/D_{prescribed} \cdot S^{SD}_{c,p}$ ratios of the IC and the PSD reveals a 0.4% and 1.0% difference for the 5 and 35 mm cone, respectively. The UD is known to have angular dependence.[8] When its signal is normalized to an irradiation coming from the opposite side of the stem, a 3% response decrease is observed[8] when the detector is rotated by 90°. The impact of the angular dependence is reflected in the measurement of the whole stereotactic radiosurgery treatment using arcs since, even when corrected by

TABLE I. Means, standard deviations of gamma values[a] and percentage of point passing the gamma evaluation, obtained from dose profiles comparison of the PSD crosshair array, the EBT2 and the UD.

		Mean	σ[b]	% passing
4 mm cone	PSDCA–EBT2	0.263	0.217	100.0
	PSDCA–UD	0.252	0.150	100.0
	UD–EBT2	0.230	0.168	100.0
10 mm cone	PSDCA–EBT2	0.273	0.222	100.0
	PSDCA–UD	0.234	0.186	100.0
	UD–EBT2	0.191	0.105	100.0
40 mm cone	PSDCA–EBT2	0.387	0.300	95.8[c]
	PSDCA–UD	0.397	0.255	100.0
	UD–EBT2	0.360	0.203	99.7[c]

[a]Acceptance values are 0.3 mm and 2%.
[b]σ = standard deviation.
[c]Implies that two detector failed the gamma criteria for the PSDCA–EBT2 gamma evaluation and that one detector failed the gamma criteria for the UD–EBT2 gamma evaluation.

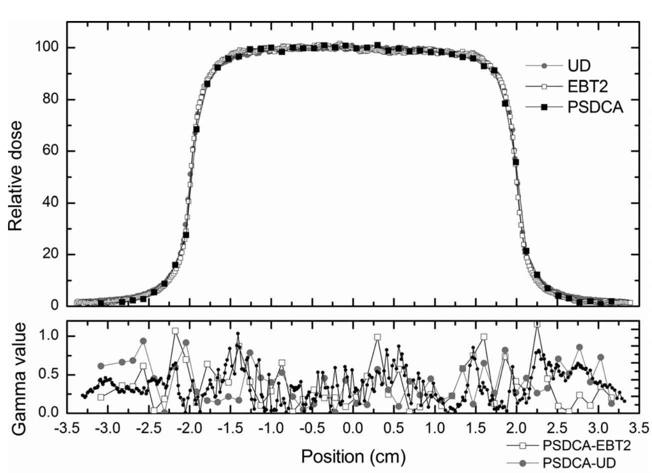

FIG. 7. Dose profiles of 40 mm diameter radiosurgery cones measured at 5 cm depth and normalized to the dose measured at the center of the field for each detector. Error bars are smaller than symbols.

TABLE II. Results obtained after performing a complete stereotactic radiosurgery treatment composed of four noncoplanar arcs and a prescription dose of 15 Gy at the isocenter where one of the above detectors was placed.

	5 mm cone		35 mm cone	
	$\frac{D_{measured}}{D_{prescribed}}$	$\frac{D_{measured}S^{X}_{c,p}}{D_{prescribed}S^{SD}_{c,p}}$	$\frac{D_{measured}}{D_{prescribed}}$	$\frac{D_{measured}S^{X}_{c,p}}{D_{prescribed}S^{SD}_{c,p}}$
IC	0.847	0.990	0.992	0.997
PSD	0.932	0.986	1.010	1.007
UD	0.913	0.970	0.955	0.978

$S_{c,p}$ ratios, the measured dose by the UD is lower than the prescribed dose by 3.0% for the 5 mm cone and 2.2% for the 35 mm cone.

The results obtained measuring output factor, dose profile, and a complete stereotactic radiosurgery quality assurance show that the IC does volume averaging in small cone sizes of the order of 9.1% in the 5 mm aperture. This implies that this detector is not suited for small field dosimetry. The UD shows good results compared to the EBT2 films and the PSD when measuring $S_{c,p}$ and dose profiles when proper measures are taken in order to consider the energy dependence, such as renormalization of $S_{c,p}$ to a smaller field size like the one used in this paper (35 mm cone). However, when performing a SRS QA, the UD has an angular dependence that affects the measurement in the order of 2.5% for the plan measured in this study. Furthermore, this factor can vary according to the arcs angles used in the measured plan. The PSD results are in agreement with the UD and the EBT2 films in dose profiles and $S_{c,p}$ measurements. The SRS QA results obtained with the PSD are within 1.4% and 0.7% of the planned dose for the 5 mm cone and the 35 mm cone, respectively. These results demonstrate that the PSD is an excellent dosimeter for application in SRS treatment planning system commissioning and SRS QA. Furthermore, dose profile measurements obtained using the PSDCA expose the great potential of PSDs for the construction of 2D arrays having specific resolution for SRS field sizes. A gamma evaluation having 0.3 mm and 2% criteria has been passed within more than 95% of the measurement points despite the presence of 49 PSD in a 1.63 cm radius, implying that the dosimeters do not perturb the dose distribution. The results obtained by Klein *et al.*[11] suggest that a better resolution for a 2D array could be achieved using PSD having 0.5 mm of diameter.

Even if the measurements done in this study where taken in SRS fields, the discussion pertaining to $S_{c,p}$ and profiles also applies to other treatment modalities using very small fields and steep dose gradients such as fractionated stereotactic radiotherapy, intensity modulated radiotherapy, and intensity modulated arc therapy. In small field dosimetry, the specific characteristics of the detectors can induce important discrepancies in dose measurements. In this regard, we recommend to cross-check measurements with different detectors including at least Gafchromic films or PSDs.

IV. CONCLUSION

The PSD has been compared to various dosimeters and has shown a good agreement with detectors considered reliable in different irradiation conditions encountered in stereotactic radiosurgery. In addition to the good characteristics of the PSD such as water equivalence, having a high spatial resolution, no angular, and energy dependence, its main advantage is its real-time nature. The PSD also measures accurately steep dose gradients, which is an asset for profile measurements of radiosurgery fields. The presence of the 49 detectors in a 1.63 cm radius have shown no impact on the dosimetry of profiles obtained by the crosshair array as the measurements are consistent with Gafchromic EBT2 and UD. Therefore, small size fully water equivalent plastic scintillation detectors can be considered particularly well suited for small field dosimetry.

ACKNOWLEDGMENTS

This work was supported by a grant from the Natural Sciences and Engineering Research Council (NSERC) (Discovery Grant No. 262105).

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