Received:
5 July 2016

Revised:
4 November 2016

Accepted:
8 November 2016

<https://doi.org/10.1259/bjr.20160596>

Cite this article as:

Pasquino M, Cutaia C, Radici L, Valzano S, Gino E, Cavedon C, et al. Dosimetric characterization and behaviour in small X-ray fields of a microchamber and a plastic scintillator detector. *Br J Radiol* 2017; **90**: 20160596.

FULL PAPER

Dosimetric characterization and behaviour in small X-ray fields of a microchamber and a plastic scintillator detector

¹MASSIMO PASQUINO, PhD, ¹CLAUDIA CUTAIA, PhD, ¹LORENZO RADICI, PhD, ¹SERENA VALZANO, PhD, ¹EVA GINO, PhD, ²CARLO CAVEDON, PhD and ¹MICHELE STASI, PhD

¹Medical Physics Department, AO Ordine Mauriziano di Torino, Turin, Italy

²Medical Physics Department, University Hospital, Verona, Italy

Address correspondence to: Dr Massimo Pasquino

E-mail: mspasquino@mauriziano.it

Objective: The aim of this work was to investigate the main dosimetric characteristics and the performance of an A26 Exradin ionization microchamber (A26 IC) and a W1 Exradin plastic scintillation detector (W1 PSD) in small photon beam dosimetry for treatment planning system commissioning and quality assurance programme.

Methods: Detector characterization measurements (short-term stability, dose linearity, angular dependence and energy dependence) were performed in water for field sizes up to $10 \times 10 \text{ cm}^2$. Polarity effect (P_{pol}) was examined for the A26 IC. The behaviour of the detectors in small field relative dosimetry [percentage depth dose, dose profiles often called the off-axis ratio and output factors (OFs)] was investigated for field sizes ranging from 1×1 to $3 \times 3 \text{ cm}^2$.

Results: Results were compared with those obtained with other detectors we already use for small photon beam dosimetry. A26 IC and W1 PSD showed a linear dose response. While the A26 IC showed no energy dependence, the W1 PSD showed energy dependence within

2%; no angular dependence was registered. P_{pol} values for A26 IC were below 0.9% (0.5% for field size $>2 \times 2 \text{ cm}^2$). A26 IC and W1 PSD depth-dose curves and lateral profiles agreed with those obtained with an EDGE diode. No differences were observed among the detectors in OF measurement for field sizes larger than $1 \times 1 \text{ cm}^2$, with average differences $<1\%$. For field sizes $<1 \times 1 \text{ cm}^2$, the effective volume of ionization chamber and non-water equivalence of EDGE diode become significant. A26 IC OF values were significantly lower than EDGE diode and W1 PSD values, with percentage differences of about -23 and -13% for the smallest field, respectively. W1 PSD OF values lay between ion chambers and diode values, with a maximum percentage difference of about -10% with respect to the EDGE diode, for a $6 \times 6 \text{ mm}^2$ field size.

Conclusion: The results of our investigation confirm that A26 IC and W1 PSD could play an important role in small field relative dosimetry.

Advances in knowledge: Dosimetric characteristics of Exradin A26 ionization microchamber and W1 plastic scintillation detector for small field dosimetry.

INTRODUCTION

The introduction of image guidance and new technology in radiation therapy, like intensity-modulated radiotherapy (IMRT), micromultileaf collimators and image-guided systems such as the CyberKnife® (Accuray, Inc., Sunnyvale, CA) and TomoTherapy® (TomoTherapy, Inc., Madison, WI), has improved the capability to treat small lesions, using small fields or non-standard fields made of small fields, in order to achieve the desired, precisely focused and highly conformal dose distributions. The commissioning of small-field system requires the acquisition of percentage depth-dose (PDD) curves and lateral profiles; patient-specific quality assurance is generally performed by delivering an IMRT plan to a phantom and thereafter comparing the dose distribution calculated by the treatment

planning system (TPS) with the one measured by ion chambers, gafchromic films and/or other detectors using gamma analysis.¹ However, the required dose accuracy within small photon fields is difficult to achieve, as lateral electronic equilibrium breaks down and traditional detectors have limitations to represent the unperturbed dose distribution in water as accurately as possible.^{2–8} As reported in literature,^{9,10} this is mostly due to dose averaging over the finite size of the active volume and the non-water equivalence of the materials that surround the active volume. With regard to volume-averaging effects, the commercial availability of detectors suitable for measurements of small fields down to the small field sizes mostly used in linear accelerators with multileaf collimator (MLC) reduces the importance of the effect. The same does not

apply to measurements in smaller fields like the ones produced by stereotactic radiosurgery systems working with cones or the Gamma Knife. Some ionization chambers were recently especially designed for small-field photon dosimetry and showed a high sensitivity, stability and very good linearity.^{3,6,11,12} Silicon detectors have a very high resolution; typically, unshielded *p*-type silicon diodes with very small active volume are used to perform small-field dosimetric characterization and output factor (OF) measurements.¹³ However, the mass density of the sensitive volume and surrounding medium strongly influences the small-field measurements with ionization microchambers and diodes.^{9,14}

Owing to their water equivalence and good spatial resolution, plastic scintillation detectors (PSDs) could play an important role in small-field dosimetry. Several authors have studied PSDs in recent years:^{10,15–20} their linear response to absorbed dose, dose rate and energy independence are well known. Consequently, PSDs have just found large application in small-field dosimetry.^{21–24} The main drawback of plastic scintillators is represented by the generation of Čerenkov light in the optical fibre guide when the scintillator is exposed to a radiation field. Two main approaches have been suggested to subtract the Čerenkov light component: the method proposed by Beddar et al,^{15,16} which uses a background optical fibre not coupled with a scintillating fibre, and the two-fibre or spectral method, proposed by Fontbonne et al²⁵ and reformulated by Guillot et al,²⁶ which is based on the dependence of the intensity of the Čerenkov light on the length of the exposed fibre and measures the light signal by means of two different wavelength channels. Morin²² has proposed a modification of the measurement procedure proposed by Guillot et al²⁶ to determine the Čerenkov spectrum with the detector placed with the stem parallel to the beam axis. Considering that the Beddar et al's approach depends on fibre, coupling and photodetector equivalence, the Guillot et al method seems to be more appropriate, reducing the Čerenkov effect within 0.7%, which represents an acceptable uncertainty for the aims of radiation therapy dosimetry.

This study aimed to investigate the main dosimetric characteristics (short-term stability, energy, angular and polarity dependence, linearity) and the performance in small-field photon beam dosimetry of two relatively new small-field measurement tools: an A26 Exradin ionization microchamber (A26 IC) and a W1 Exradin plastic scintillation detector (W1 PSD). The results

are compared with those obtained with other detectors we already use for small photon beam dosimetry, contributing to complement data that can be found in the literature,^{24,27–29} in particular regarding the behaviour of these detectors in small-field beam dosimetry required for TPS commissioning (PDD, dose profile and OF measurements).

METHODS AND MATERIALS

We tested an A26 IC and a W1 PSD, connected to a dual-channel SuperMAX electrometer. Both detectors and electrometer are manufactured by Standard Imaging Inc., Middleton, WI. The small size of the sensitive volume and the good water equivalence for high-energy photon and electron beams make these detectors potentially suited for small-field dosimetry.

The main geometrical characteristics of the detectors used in the study are summarized in Table 1.

The A26 IC is composed of C552 Shonka air-equivalent plastic material; as indicated by the manufacturer, the characteristics of leakage, ion recombination, polarity effect (P_{pol}), initial recombination and stability make the detector a Task Group (TG) 51 reference-class chamber for small fields.

W1 PSD is produced with water-equivalent components (polystyrene with an acrylonitrile butadiene styrene plastic enclosure and a polyimide stem). As known, the major drawback of scintillator detectors is represented by the generation of Čerenkov light in the optical fibre guide when the fibre is exposed to a radiation field. To account for this effect, a Čerenkov correction factor (CLR) has been measured following both the procedures suggested by the manufacturer for the different orientations of the scintillating fibre relative to the radiation beam axis.^{22,30} Despite the difficulties of the four-step procedure recommended with the scintillator axis is orientated parallel to the beam axis (as mentioned by Underwood et al²⁴), by accurately controlling the positioning of the fibre, we found similar standard deviations, <1%, which confirm data reported by Carrasco et al²⁸, who performed the procedure with the detector perpendicular to the beam axis in plastic water. For convenience, we used the same approach for CLR calibration in the measurement sessions, with the detector perpendicular to the beam axis. OF measurements were performed with the detector in both orientations: parallel and perpendicular to the beam axis.

Table 1. Main characteristics of the investigated detectors

Detector's characteristic	Standard imaging A26	Standard imaging W1	Standard imaging A1SL	Sun nuclear EDGE
Detector type	Ionization chamber	Plastic scintillator	Ionization chamber	Diode
Inside diameter (mm)	3.3	1	4	0.8
Length (mm)	–	3	4.4	0.8
Volume (cm ³)	0.015	0.0024	0.053	1.9×10^{-5}
Polarization voltage (V)	+300		+300	

The work was mainly performed with a 6-MV photon beam provided by an Elekta® Synergy linear accelerator (Elekta, Stockholm, Sweden) equipped with a 160-leaf collimator (5 mm at isocentre). In order to check the W1 PSD energy dependence, 10- and 15-MV photon beams were also used.

Characterization measurements were performed in water in a three-dimensional scanner phantom (Sun Nuclear Corporation, Melbourne, FL) for field sizes up to $10 \times 10 \text{ cm}^2$. Reference conditions were 10-cm depth, 100-cm source-axis distance, $10 \times 10 \text{ cm}^2$ field size and 100 MU. The field sizes were defined by the MLC and orthogonal dynamic jaws. The A26 IC effective measurement point was set equal to 1 mm (0.6 times the internal radius towards the radiation source, according to the main dosimetry protocols); the W1 effective point of measurement was set equal to 0.8 mm from its surface according to Carrasco et al.²⁸ The effective point of measurement is derived by using a plane-parallel ion chamber as a reference detector: the 6-MV normalized PDD curves were matched in the build-up region and beyond the maximum dose. The verification of the effective point of measurements was performed for both the parallel and perpendicular orientations of the detectors. A26 IC measurements were performed by setting the polarization voltage to +300 V. W1 PSD measurements were performed with the detector axis perpendicular to the beam axis, except for the angular dependence, which was also investigated with the detector longitudinal axis parallel in relation to the beam axis.

Detector physical characterization

Short-term reproducibility of 10 consecutive 100-MU measurements was tested for 3×3 - and $10 \times 10 \text{ cm}^2$ fields; A26 IC measurements were performed before and after a 1-kGy warm-up pre-irradiation.

Dose linearity was evaluated for 3×3 - and $10 \times 10 \text{ cm}^2$ fields in the range 1–500 MU.

The A26 IC P_{pol} was calculated following the American Association of Physicists in Medicine protocol TG 51,³¹ varying the field sizes from 1×1 to $10 \times 10 \text{ cm}^2$. The results were compared

with those obtained with an Exradin A1SL ionization chamber that we currently use for small photon beam dosimetry at our institution.

The angular dependence at different gantry angles was investigated for W1 PSD only, given the cylindrical symmetry of the ionization chamber, irradiating the detector placed at the linear accelerator isocentre in air within a home-made cylindrical build-up cap. Measurements were performed every 30°

Figure 2. $1 \times 1 \text{ cm}^2$ (a), $2 \times 2 \text{ cm}^2$ (b) and $3 \times 3 \text{ cm}^2$ (c) field percentage depth doses measured with A26 Exradin ionization microchamber (A26 IC) (—), W1 Exradin plastic scintillation detector (W1 PSD) (×), A1SL IC (□) and EDGE diode (○).

Figure 1. Polarity effect (P_{pol}) at different field sizes for A26 Exradin ionization microchamber (A26 IC) (---) and A1SL IC (---).

from -180° to 180° under $10 \times 10\text{-cm}^2$ 6-MV beam and corrected for the linear accelerator output variation by means of a set of measurements performed with a calibrated Farmer-type ionization chamber.

A26 IC and W1 PSD energy dependence was investigated at 10-cm depth, 100-cm source-axis distance and $10 \times 10\text{-cm}^2$ field size for photon beam energies of 6, 10 and 15 MV. The detector readings were normalized to the absolute dose values measured with a calibrated Farmer-type ionization chamber in the same irradiation conditions and the percentage differences relative to the 6-MV energy measurements were calculated.

As far as W1 PSD temperature dependence is concerned, the results of Carrasco et al²⁸ can be considered.

Detector behaviour in small field relative dosimetry PDD curves and dose profiles [off-axis ratio (OAR)] were acquired for $1 \times 1\text{-}$, $2 \times 2\text{-}$ and $3 \times 3\text{-cm}^2$ field sizes. Given that it is not possible to use the W1 PSD connected to any three-dimensional scanning system, PDD and OAR measurements with W1 PSD were performed point-by-point using the scanning system to position the detector in the measurement position.

The results were compared with those obtained with the A1SL IC and an EDGE diode we already use for small field relative dosimetry. With regard to PDD curves, the agreement was evaluated in four comparison points at 2-, 5-, 10- and 20-cm depths. OARs were compared in terms of field size [full width half maximum (FWHM)] and penumbra values (defined as the distance between 80 and 20% dose points). OF measurements were performed at 100-cm source-axis distance and 10-cm water depth, for field sizes ranging from 0.6×0.6 to $10 \times 10\text{ cm}^2$; W1 PSD measurements were performed with the detector axis both parallel and perpendicular to beam axis.

RESULTS

Detector physical characterization

With regard to short-term reproducibility, repeated measurements with A26 IC showed a monotonical increase with a 2% maximum variation, which confirms literature data concerning microchamber leakage currents.³² However, after a 1-kGy warm-up, A26 IC readings showed a 0.2 and 0.1% relative standard deviation for $3 \times 3\text{-}$ and $10 \times 10\text{-cm}^2$ fields, respectively. W1 PSD measurements showed a relative standard deviation of 0.1% for both $3 \times 3\text{-}$ and $10 \times 10\text{-cm}^2$ fields.

A26 IC and W1 PSD dose response was linear to within 0.5% ($R^2 = 1$) both for $3 \times 3\text{-}$ and $10 \times 10\text{-cm}^2$ fields.

In Figure 1, P_{pol} of the A26 IC and the corresponding values for the A1SL IC, as reference, are reported. As illustrated, the P_{pol} was within 0.5% for field size $>2 \times 2\text{ cm}^2$; for field size $<2 \times 2\text{ cm}^2$, the effect increased, still remaining within measurement standard deviation.

Table 2. Percentage depth dose (PDD) values measured at different depth points with A26 Exradin ionization microchamber (A26 IC) and percentage differences of W1 Exradin plastic scintillation detector (W1 PSD), A1SL IC and EDGE diode measurements

Depth (cm)	EDGE diode PDD value (%)			W1 PSD Δ (%)			A1SL IC Δ (%)			A26 IC Δ (%)		
	$1 \times 1\text{ cm}^2$	$2 \times 2\text{ cm}^2$	$3 \times 3\text{ cm}^2$	$1 \times 1\text{ cm}^2$	$2 \times 2\text{ cm}^2$	$3 \times 3\text{ cm}^2$	$1 \times 1\text{ cm}^2$	$2 \times 2\text{ cm}^2$	$3 \times 3\text{ cm}^2$	$1 \times 1\text{ cm}^2$	$2 \times 2\text{ cm}^2$	$3 \times 3\text{ cm}^2$
2	97.1	98.8	98.4	0.5	-0.7	0.3	0.9	-0.3	0.2	-0.3	-0.7	0.1
5	79.1	81.6	82.8	4.2	-0.4	0.7	3.0	0.1	-0.1	0.9	0.0	-0.4
10	56.8	58.8	60.1	3.9	0	1.7	3.9	1.5	0.3	0.9	0.3	0.7
20	30.1	31.2	32.3	5.3	3.2	0.9	4.7	2.9	1.5	0.7	-0.3	2.5

Table 3. Full width half maximum (FWHM) and penumbra values measured with A26 Exradin ionization microchamber (A26 IC), W1 Exradin plastic scintillation detector (W1 PSD), A1SL IC and EDGE diode

Dose profiles' characteristic	1 × 1 cm ²				2 × 2 cm ²				3 × 3 cm ²			
	A26	W1	A1SL	EDGE	A26	W1	A1SL	EDGE	A26	W1	A1SL	EDGE
FWHM (mm)	9.9	9.7	10.1	9.8	18.7	18.6	18.9	18.6	28.0	27.8	27.9	27.6
Penumbra (mm)	4.2	4.0	4.4	4.4	4.8	4.5	5.0	4.3	4.9	4.4	5.2	4.4

As expected, measurements of A26 IC and W1 PSD energy dependence confirmed the energy independence of the microchamber, while W1 PSD showed a slight dependence, within 1.5%, with increased response at greater energies, predominantly owing to a change in the CLR from 6 to 15 MV.

Angular dependence showed a $\pm 0.5\%$ maximum reading variation. However, comparing W1 PSD response with that of a calibrated Farmer-type ionization chamber, no statistically significant differences were observed; therefore, the reading variations have to be ascribed to linear accelerator output angular dependence. No dependence on the detector orientation with respect to the beam axis was registered.

Detector behaviour in small field relative dosimetry PDDs for the 1 × 1-, 2 × 2- and 3 × 3-cm² field size are shown in Figure 2. In Table 2, PDD values measured at 2-, 5-, 10- and 20-cm depth points with the EDGE diode are reported and compared with those measured with W1 PSD, A1SL IC and A26 IC, in terms of percentage differences. As reported, A26 IC measurements agreed with those obtained with the EDGE diode; conversely, W1 PSD and A1SL IC showed a growing disagreement with decreasing field sizes, with maximum values equal to about 5% at 20-cm depth for the 1 × 1-cm² field size.

In Table 3 the absolute values of FWHM and penumbra values measured with the different detectors are reported. OAR dose profiles of 1 × 1-, 2 × 2- and 3 × 3-cm² fields are illustrated in Figure 3.

Finally, Table 4 shows OF values measured with A26 IC, W1PSD, A1SL IC and EDGE diode.

DISCUSSION

In modern radiation therapy, when modalities like IMRT and stereotactic radiosurgery are chosen, small field sizes are often used to treat target volumes, sparing normal tissues.^{33–36} Consequently, it becomes fundamental to determine the dose in small photon beams with high accuracy, both in TPS commissioning and patient-specific quality assurance. However, the lack of charged particle equilibrium, the finite size of the detectors and the non-water equivalence of the materials limit the accuracy of small-field dosimetry. Several authors have investigated this issue: new formalisms for small and non-standard field dosimetry have been proposed,³⁷ Monte Carlo simulations have been used to provide correction factors for several detectors in non-standard fields^{38,39} and the behaviour of traditional or new detectors in small photon beams has been investigated.^{3,6,11}

However, the selection of suitable detectors remains a subject of debate. Microchambers especially designed for small-field dosimetry have been more recently commercialized, but the low density of their sensitive volume still represents an issue. Diode detectors and plastic scintillators seem to be a better choice, being less influenced by volume averaging compared with ionization chambers. Plastic

Figure 3. $1 \times 1\text{-cm}^2$ (a), $2 \times 2\text{-cm}^2$ (b) and $3 \times 3\text{-cm}^2$ (c) field off-axis ratio dose profiles measured with A26 Exradin ionization microchamber (A26 IC) (---), W1 Exradin plastic scintillation detector (W1 PSD) (---), A1SL IC (---) and EDGE diode (---).

scintillators also offer better water equivalency, in principle.^{17,28,40,41} However, diode detectors have a high density in comparison with water.

In this work, we have investigated the dosimetric characteristics of two relatively new detectors: an ionization microchamber and a plastic scintillator that have not yet been extensively studied.^{27,28,42} We did not take into account detector energy and temperature dependence, which would allow achieving uncertainties lower than those found.

The A26 IC showed reproducibility within 0.3%, minimal dependence on P_{pol} vs field size, minimal energy dependence and dose linearity.

The W1 plastic scintillator showed reproducibility and dose linearity, energy and angular independence, confirming recent literature data.²⁸

Following recommendations from the American Association of Physicists in Medicine TG 135,⁴³ the diode from this study (EDGE) was considered as the reference detector. Considering PDD measurements, both W1 PSD and A26 IC showed a better agreement with the EDGE diode (maximum variation within 1%) than the A1SL IC, with significant volume-averaging effects at a $1 \times 1\text{-cm}^2$ field (maximum variation $>10\%$). With regard to dose profile analysis, W1 PSD and A26 IC FWHM and penumbra values show a good agreement with the EDGE diode values (maximum variation within 0.5 mm).

Finally, with regard to OF measurements, no differences were observed in detector response for field sizes $>1 \times 1\text{ cm}^2$, with average differences $<1\%$. For field size $<1 \times 1\text{ cm}^2$, the averaging effect due to the large sensitive volume of the ionization chambers starts to play a non-negligible role, leading to a dose underestimation. In fact, A26 IC OF values were comparable with those of A1SL IC and significantly lower than those of EDGE diode and W1 PSD values, with percentage differences of about -29% and -14% for the smallest field, respectively.

W1 PSD OF values lay between ion chambers and diode values, with a maximum percentage difference of about -17% with respect to the EDGE diode for the $6 \times 6\text{-mm}^2$ field size. The fact that there were no significant differences observed for different orientations of the W1 PSD (parallel or perpendicular to the beam axis) allows for the length/volume effect to be discarded as the cause for the observed discrepancy in this comparison. The discrepancies could be explained to be due to the presence of non-water-equivalent materials in the active and non-active parts of the Edge detector, which, within the small field, becomes important, more than the differences in detector Z, as demonstrated by Scott et al,¹⁴ causing an overreading of the detector. In case of small beams, the fraction of photons scattered from parts of the detector is higher compared with the case of large beams, possibly resulting in an overestimation of the OF. Considering a 7% overresponse of the EDGE detector, as reported by Bassinet et al⁴⁴ for the $6 \times 6\text{-mm}^2$ field of a Clinac 2100 equipped with additional micro-MLC, the differences between W1 PSD and the corrected EDGE diode are reduced to roughly 4 and 10% for $8 \times 8\text{-}$ and $6 \times 6\text{-mm}^2$ field sizes, respectively.

CONCLUSION

In conclusion, the results of our investigation and in particular the comparison of the behaviour with detectors that have been available longer, like the A1SL IC and the EDGE diode, confirm

Table 4. Output factor values measured with the different detectors

Detector	Field size (cm ²)									
	0.6 × 0.6	0.8 × 0.8	1 × 1	2 × 2	3 × 3	4 × 4	5 × 5	6 × 6	8 × 8	10 × 10
A26 IC	0.382	0.538	0.635	0.801	0.848	0.884	0.906	0.934	0.974	1
W1 PSD _p	0.442	0.579	0.668	0.811	0.853	0.888	0.910	0.935	0.972	1
W1 PSD _n	0.444	0.581	0.670	0.807	0.851	0.879	0.908	0.933	0.970	1
A1SL IC	0.381	0.525	0.632	0.800	0.846	0.880	0.907	0.931	0.970	1
EDGE diode	0.536	0.640	0.702	0.804	0.843	0.875	0.902	0.925	0.966	1

A26 IC, A26 Exradin ionization microchamber; W1 PSD, W1 Exradin plastic scintillation detector. W1 PSD measurements were performed with detector axis parallel (p) and normal (n) to beam axis.

that A26 IC and W1 PSD could play an important role in small-field relative dosimetry.

Our measurements confirm that the A26 IC could be considered as a reference-class ionization chamber for small fields, as indicated by the manufacturer; however, the small dimensions and

the water equivalence of W1 PSD makes this detector the best solution for small field relative dosimetry.

ACKNOWLEDGMENTS

The authors would like to thank Tecnologie Avanzate and Exradin for supporting the study.

REFERENCES

- Ezzell GA, Burmeister JW, Dogan N, LoSasso TJ, Mechalakos JG, Mihailidis D, et al. IMRT commissioning: multiple institution planning and dosimetry comparisons, a report from AAPM Task Group 119. *Med Phys* 2009; **36**: 5359–73. doi: <https://doi.org/10.1118/1.3238104>
- Sibata CH, Mota HC, Beddar AS, Higgins PD, Shin KH. Influence of detector size in photon beam profile measurements *Phys Med Biol* 1991; **36**: 621–31. doi: <https://doi.org/10.1088/0031-9155/36/5/005>
- Martens C, De Vagter C, De Neve W. The value of the PinPoint ion chamber for characterization of small fields segments use in intensity modulated radiotherapy. *Phys Med Biol* 2000; **45**: 2519–30. doi: <https://doi.org/10.1088/0031-9155/45/9/306>
- Laub WU, Wong T. The volume effect of detectors in the dosimetry of small fields used in IMRT. *Med Phys* 2003; **30**: 341–7. doi: <https://doi.org/10.1118/1.1544678>
- Leybovich LB, Sethi A, Dogan N. Comparison of ionization chambers of various volumes for IMRT absolute dose verification. *Med Phys* 2003; **30**: 119–23. doi: <https://doi.org/10.1118/1.1536161>
- Stasi M, Baiotto B, Barboni G, Scielzo G. The behavior of several microionization chambers in small intensity modulated radiotherapy fields. *Med Phys* 2004; **31**: 2792–5. doi: <https://doi.org/10.1118/1.1788911>
- Pappas E, Maris TG, Papadakis A, Zacharopoulou F, Damilakis J, Papanikolaou N, et al. Experimental determination of the effect of detector size on profile measurements in narrow photon beams. *Med Phys* 2006; **33**: 3700–10. doi: <https://doi.org/10.1118/1.2349691>
- Das IJ, Ding GX, Ahnesjo A. Small fields: nonequilibrium radiation dosimetry. *Med Phys* 2008; **35**: 206–15. doi: <https://doi.org/10.1118/1.2815356>
- Underwood TS, Winter HC, Fenwick JD. Mass-density compensation can improve the performance of a range of different detectors under non-equilibrium conditions. *Phys Med Biol* 2013; **58**: 8295–310. doi: <https://doi.org/10.1088/0031-9155/58/23/8295>
- Papaconstadopoulos P, Tessier F, Seuntjens J. On the correction, perturbation and modification of small field detectors in relative dosimetry. *Phys Med Biol* 2014; **59**: 5937–52. doi: <https://doi.org/10.1088/0031-9155/59/19/5937>
- McKerracher C, Thwaites DI. Assessment of a new small-field detectors against standard-field detectors for practical stereotactic beam data acquisition. *Phys Med Biol* 1999; **44**: 2143–60. doi: <https://doi.org/10.1088/0031-9155/44/9/303>
- Capote R, Sanchez-Doblado F, Leal A, Lagares JJ, Arrans R, Hartmann GH. An EGSnrc Monte Carlo study of the micro-ionization chamber for reference dosimetry of narrow irregular IMRT beamlets. *Med Phys* 2004; **31**: 2416–22. doi: <https://doi.org/10.1118/1.1767691>
- Cranmer-Sargison G, Weston S, Sidhu NP, Thwaites DI. Experimental small field 6 MV output ratio analysis for various diode detector and accelerator combinations. *Radiother Oncol* 2011; **100**: 429–35. doi: <https://doi.org/10.1016/j.radonc.2011.09.002>
- Scott AJ, Kumar S, Nahum AE, Fenwick JD. Characterizing the influence of detector density on dosimeter response in non-equilibrium small photon fields. *Phys Med Biol* 2012; **57**: 4461–76. doi: <https://doi.org/10.1088/0031-9155/57/14/4461>
- Beddar AS, Mackie TR, Attix FH. Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: I. Physical characteristics and theoretical considerations. *Phys Med Biol* 1992; **37**: 1883–900.
- Beddar AS, Mackie TR, Attix FH. Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: II. Properties and measurements. *Phys Med Biol* 1992; **37**: 1901–13.
- Wang LL, Klein D, Beddar AS. Monte Carlo study of the energy and angular dependence of the response of plastic scintillation detectors in photon beams. *Med Phys* 2010; **37**: 5279–86.
- Wang LL, Beddar AS. Study of the response of plastic scintillation detectors in small-field 6 MV photon beams by Monte Carlo simulations. *Med Phys* 2011; **38**: 1596–9.
- Gagnon JC, Theriault D, Guillot M, Archambault L, Beddar AS, Gingras L, et al. Dosimetric performance and array assessment of plastic scintillation detectors for stereotactic radiosurgery quality assurance.

- Med Phys* 2012; **39**: 429–36. doi: <https://doi.org/10.1118/1.3666765>
20. Francescon P, Beddar S, Satariano N, Das JJ. Variation of kQ_{clin} , Q_{msr} , f^{clin} , f^{msr} for the small-field dosimetric parameters percentage depth dose, tissue-maximum ratio, and off-axis ratio. *Med Phys* 2014; **41**: 101708–14.
  21. Ralston A, Liu P, Warrenner K, McKenzie D, Suchowerska N. Small field diode correction factors derived using an air core fibre optic scintillation dosimeter and EBT2 film. *Phys Med Biol* 2012; **7**: 2587–602. doi: <https://doi.org/10.1088/0031-9155/57/9/2587>
  22. Morin J, Beliveau-Nadeau D, Chung E, Seuntjens J, Theriault D, Archambault L, et al. A comparative study of small field total scatter factors and dose profiles using plastic scintillation detectors and other stereotactic dosimeters: the case of the CyberKnife. *Med Phys* 2013; **40**: 011719. doi: <https://doi.org/10.1118/1.4772190>
  23. Warrenner K, Hug B, Liu P, Ralston A, Ebert MA, McKenzie DR, et al. Small field in-air output factors: the role of miniphantom design and dosimeter type. *Med Phys* 2014; **41**: 021723. doi: <https://doi.org/10.1118/1.4861710>
  24. Underwood TS, Rowland BC, Ferrand R, Vieilleveigne L. Application of the Exradin W1 scintillator to determine Ediode 60017 and microdiamond 60019 correction factors for relative dosimetry within small MV and FFF fields. *Phys Med Biol* 2015; **60**: 6669–83. doi: <https://doi.org/10.1088/0031-9155/60/17/6669>
  25. Fontbonne M, Iltis G, Ban G, Battala A, Vernhes C, Tillier J, et al. Scintillating fiber dosimeter for radiation therapy accelerator. *IEEE Trans Nucl Sci* 2002; **49**: 2223–7. doi: <https://doi.org/10.1109/TNS.2002.803680>
  26. Guillot M, Gingras L, Archambault L, Beddar AS, Beaulieu L. Spectral method for the correction of the Čerenkov light effect in plastic scintillation detectors: a comparison study of calibration procedures and validation in Čerenkov light-dominated situations. *Med Phys* 2011; **38**: 2140–50.
  27. Beierholm AR, Behrens CF, Andersen CE. Dosimetric characterization of the Exradin W1 plastic scintillator detector through comparison with an in-house developed scintillator system. *Radiat Meas* 2014; **69**: 50–6. doi: <https://doi.org/10.1016/j.radmeas.2014.08.005>
  28. Carrasco P, Jornet N, Jordi O, Lizondo M, Latorre-Musoll A, Eudaldo T, et al. Characterization of the Exradin W1 scintillator for use in radiotherapy. *Med Phys* 2015; **42**: 297–304.
  29. Tanny S, Sperling N, Parsai E. Correction factor measurements for multiple detectors used in small field dosimetry on the Varian edge radiosurgery system. *Med Phys* 2015; **42**: 5370–6.
  30. Standard Imaging. Calibration of the Exradin W1 scintillator for small field measurements technical note 4792-01, 2014. Available from: www.standardimaging.com/uploads/tech_notes/4792_01_Exradin_W1_Scintillator_Calibration_Tech_Note.pdf
  31. Almond PR, Biggs PJ, Coursey BM, Hanson WF, Huq MS, Nath R, et al. AAPM's TG-51 protocol for clinical reference dosimetry of high-energy photon and electron beams. *Med Phys* 1999; **26**: 1847–70.
  32. McEwen MR. Measurement of ionization chamber absorbed dose kQ factors in megavoltage photon beams. *Med Phys* 2010; **37**: 2179–93.
  33. Das JJ, Downes MB, Corn BW, Curran BJ, Wasik MW, Andrews DW. Characteristics of a dedicated linear accelerator-based stereotactic radiosurgery-radiotherapy unit. *Radiation Oncol* 1996; **38**: 61–8.
  34. Adler JR, Chang SD, Murphy MJ, Doty J, Geis P, Hancock SL. The CyberKnife: a frameless robotic surgery system for radiosurgery. *Stereotact Funct Neurosurg* 1997; **69**: 124–8.
  35. Ruchala KJ, Olivera GH, Schloesser EA, Mackie TR. Megavoltage CT on a tomotherapy system. *Phys Med Biol* 1999; **44**: 2597–621. doi: <https://doi.org/10.1088/0031-9155/44/10/316>
  36. Ezzell GA, Galvin JM, Low D, Palta JR, Rosen I, Sharpe MB, et al. Guidance document on delivery, treatment planning, and clinical implementation of IMRT: report of the IMRT subcommittee of the AAPM radiation therapy committee. *Med Phys* 2003; **30**: 2089–115.
  37. Alfonso R, Andreo P, Capote R, Saiful Huq M, Kilby W, Kjäll P, et al. A new formalism for reference dosimetry of small and non-standard fields. *Med Phys* 2008; **35**: 5179–86. doi: <https://doi.org/10.1118/1.3005481>
  38. Francescon P, Cora S, Cavedon C. Total scatter factors of small beams: a multidetector and Monte Carlo study. *Med Phys* 2008; **35**: 504–13.
  39. Francescon P, Cora S, Satariano N. Calculation of kQ_{clin} , Q_{msr} , f^{clin} , f^{msr} for several small detectors and for two linear accelerators using Monte Carlo simulations. *Med Phys* 2011; **38**: 6513–27. doi: <https://doi.org/10.1118/1.3660770>
  40. Lacroix F, Beaulieu L, Archambault L, Beddar AS. Simulation of the precision limits of plastic scintillation detectors using optimal component selection. *Med Phys* 2010; **37**: 412–18. doi: <https://doi.org/10.1118/1.3276734>
  41. Lacroix F, Guillot M, McEwen M, Cojocar C, Gingras L, Beddar AS, et al. Extraction of depth-dependent perturbation factors for parallel-plate chambers in electron beams using a plastic scintillation detector. *Med Phys* 2010; **37**: 4331–42.
  42. Mathew D, Tanny S, Parsai E, Sperling N. Are small field detector correction factors strongly dependent on machine-specific characteristics. *Med Phys* 2015; **42**: 3191.
  43. Dieterich S, Cavedon C, Chuang CF, Cohen AB, Garrett JA, Lee CL, et al. Report of AAPM TG 135: quality assurance for robotic radiosurgery. *Med Phys* 2011; **38**: 2914–36.
  44. Bassinet C, Huet C, Derreumaux S, Brunet G, Chéa M, Baumann M, et al. Small fields output factors measurements and correction factors determination for several detectors for a CyberKnife® and linear accelerators equipped with micro MLC and circular cones. *Med Phys* 2013; **40**: 071725. doi: <https://doi.org/10.1118/1.4811139>