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Characterization of the Exradin W1 scintillator for use in radiotherapy

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Purpose: To evaluate the main characteristics of the Exradin W1 scintillator as a dosimeter and to estimate measurement uncertainties when used in radiotherapy. Methods: We studied the calibration procedure, energy and modality dependence, short-term repeatability, dose-response linearity, angular dependence, temperature dependence, time to reach thermal equilibrium, dose-rate dependence, water-equivalent depth of the effective measurement point, and long-term stability. An uncertainty budget was derived for relative and absolute dose measurements in photon and electron beams. Results: Exradin W1 showed a temperature dependence of \(-0.225\% ^\circ C^{-1}\). The loss of sensitivity with accumulated dose decreased with use. The sensitivity of Exradin W1 was energy independent for high-energy photon and electron beams. All remaining dependencies of Exradin W1 were around or below 0.5%, leading to an uncertainty budget of about 1%. When a dual channel electrometer with automatic trigger was not used, timing effects became significant, increasing uncertainties by one order of magnitude. Conclusions: The Exradin W1 response is energy independent for high energy x-rays and electron beams, and only one calibration coefficient is needed. A temperature correction factor should be applied to keep uncertainties around 2% for absolute dose measurements and around 1% for relative measurements in high-energy photon and electron beams. The Exradin W1 scintillator is an excellent alternative to detectors such as diodes for relative dose measurements. © 2015 American Association of Physicists in Medicine. [http://dx.doi.org/10.1118/1.4903757]

Key words: scintillator dosimeter, radiation therapy measurements, dosimeter characterization

1. INTRODUCTION

The use of scintillation light is one of the oldest techniques on record to detect ionizing radiation,\(^1\) but it was not until 2013 that a dosimeter based on a scintillator became commercially available. Scintillator materials may be organic or inorganic. Organic plastic scintillators are better than inorganic scintillators for radiotherapy purposes because they are almost water-equivalent and they can be manufactured in small dimensions. The reason why these plastic scintillators were not used for dosimetry purposes in radiotherapy until recently was that within the light fiber that guides the scintillation light to a photodiode or a photomultiplier, some Čerenkov light is also produced in the irradiated fiber. The amount of Čerenkov light depends on the length of the irradiated fiber, thus leading to a variable component that is not easy to quantify and eliminate.

One of the most widely used techniques to eliminate the Čerenkov light component is the spectral method.\(^2,3\) This method takes advantage of the fact that the scintillation light spectrum differs from that of Čerenkov light. The former has a maximum at a wavelength of 500–550 nm (green light) for many plastic scintillators while the latter decreases with the wavelength, and is mainly bluish. To reduce the Čerenkov light component from the scintillation light, some prototypes in the literature divide the light beam in two by means of two 45° reflective dichroic color filters (green and blue) placed at the end of the light fiber. The green filter selects the green light, where most of the scintillation light spectrum is located, and attenuates the blue light, which is mainly produced by the Čerenkov effect. The blue filter attenuates the scintillation light and allows the Čerenkov light pass through. Two photodiodes then convert these light beams into electrical currents that are integrated by a two-channel electrometer. Guillot et al.\(^3\) proposed a calibration method, based on the spectral method, that reduces the Čerenkov effect. The residual uncertainty of this calibration method is less than 0.7% for field sizes ranging from \(7 \times 7 \text{ cm}^2\) to \(40 \times 40 \text{ cm}^2\), an acceptable uncertainty for most dosimetry purposes in external beam radiotherapy.

Recently, a commercial plastic scintillator dosimeter became available. This commercial solution uses the spectral discrimination method to eliminate the Čerenkov light contamination and recommends the calibration method as proposed by Guillot et al.\(^3\)

Although much has been published on the development and characterization of prototypes of PSDs over the last decade,\(^4–8\) there is as yet no independent publication from a customer showing the characteristics of the aforementioned commercial solution for radiotherapy purposes. This paper aims to evaluate the main characteristics of Exradin W1 scintillator as a dosimeter and to estimate measurement uncertainties when used in radiotherapy.
2. MATERIALS AND METHODS

We tested an Exradin W1 Scintillator connected to a two-channel SuperMAX electrometer, both devices from Standard Imaging Inc., Middleton, WI. The scintillating fiber is based on polystyrene.

This fiber is surrounded by an acrylonitrile butadiene styrene (ABS) plastic enclosure and a polymide stem. The sensitive volume of the Exradin W1 scintillator is approximately 1 mm diameter by 3 mm long (0.0024 cm³), and the optical output is guided to a photodiode by a clear optical fiber. The electrometer was set in triggered charge collection mode for channel 1 (green light, mainly scintillation light). We used the automatic start and stop trigger thresholds preconfigured as a default in the Supermax electrometer (start = 0.4 pA and stop = 0.2 pA). Channel 2 collected the signal from blue light mainly produced by Čerenkov radiation. The electrometer software allows automatic correction of the Čerenkov effect. For this work, however, we recorded raw charges for both channels and we manually applied the correction to check how it was implemented. We investigated the calibration procedure and the dosimetric characteristics of Exradin W1 for energies and modalities used for external beam radiotherapy. These characteristics were energy and modality dependence, short-term repeatability, dose–response linearity, angular dependence, temperature dependence, time to reach thermal equilibrium, dose-rate dependence, the water-equivalent depth of the effective measurement point, and long-term stability.

We derived A-type uncertainties from the standard deviation of series of measurements. Each measurement was repeated at least three times. As most tests were relative to a reference value, many uncertainties such as those present in absolute dose calibration cancelled out. To account for B-type uncertainties, we assumed a uniform probability density function and the associated standard deviation was the interval amplitude divided by (12)½ according to international recommendations.⁹ We always included two sources of B-type uncertainties: first, all electrometer uncertainties stated in its calibration certificate and, second, the beam/measuring system stability. The latter was estimated from dose measurements in reference conditions before and after performing the test. In specific tests such as the study of temperature dependence, we accounted for additional B-type uncertainties, like the uncertainty in the temperature measurement. In this specific case, we performed two temperature measurements, one before and one after irradiating the scintillator. Additionally, we included the temperature uncertainty measurement from the thermometer calibration certificate. Standard uncertainty propagation was applied.

2.A. Calibration

We calibrated the Exradin W1 against a traceable ionometric dosimetry system for a 6 MV x-ray beam (Clinac 2100C/D, Varian Medical Systems, Palo Alto, CA). We performed dose measurements in reference conditions [field size 10 × 10 cm², source–surface distance (SSD) = 100 cm, z = 10 cm] following the IAEA TRS-398 Code of Practice¹⁰ using a NE2571 ionization chamber (Thermo Electron Co., Waltham, MA) and an Inovision 35040 electrometer (Fluke Biomedical, Cleveland, OH) traceable to a secondary laboratory.

We performed the calibration in a 30 × 30 × 30 cm³ slab Plastic Water™ phantom (CIRS Inc., Norfolk, WA), placing the scintillator inside the calibration plate provided with the Exradin W1, as shown in Fig. 1, with 10 cm of Plastic Water on top. We set the SSD to 100 cm. The scintillator was therefore located at 10 cm depth. The dosimetric equivalence of Plastic Water to water for megavoltage photon beams was verified in a previous study¹¹ and it is well documented in the literature.¹²

We followed the calibration procedure recommended by the manufacturer and based on the paper published by Guillot et al.³ This procedure consists of four steps aiming to determine two parameters: the Čerenkov Light Ratio (CLR) and the “Gain” value (that in a sense is the “calibration coefficient”). For all four steps, measurements are carried out for a fixed number of monitor units (MU). The first three steps consist of the following measurements to be performed with the scintillator: first, a charge measurement with the cable in its calibration certificate and, second, the beam/measuring system stability. The latter was estimated from dose measurements in reference conditions before and after performing the test. In specific tests such as the study of temperature dependence, we accounted for additional B-type uncertainties, like the uncertainty in the temperature measurement. In this specific case, we performed two temperature measurements, one before and one after irradiating the scintillator. Additionally, we included the temperature uncertainty measurement from the thermometer calibration certificate. Standard uncertainty propagation was applied.

Fig. 1. Calibration plate with the scintillator in the maximum fiber configuration (left) and minimum fiber configuration (right).
another measurement in the “minimum fiber configuration” but under a 10 × 10 cm² field. We repeated each measurement five times and we recorded the readings for both channels. The fourth and last step is a dose measurement using an ionization chamber at 10 cm depth under IAEA reference conditions. We measured the absorbed dose before and after measuring with the scintillator, and the dose we used for the calibration (Dose₁₀) was the average of the two measurements.

According to the manufacturer’s specifications, if we use sub-indices “1” and “2” for channels 1 and 2, respectively, sub-indices “max” and “min” for the maximum and minimum fiber configurations, sub-indices “10” and “40” for the 10 × 10 and 40 × 40 cm² fields, and R for the collected charge, then

\[ CLR = \frac{(R_{1\text{, max}} - R_{1\text{, min}})}{(R_{2\text{, max}} - R_{2\text{, min}})} \]  
\[ \text{Gain} = \frac{\text{Dose}_{10}}{(R_{1\text{, min}} - R_{2\text{, min}}) \times \text{CLR}}. \]

Once these parameters are determined, the dose for any situation can be calculated from the charges collected by the two channel electrometer (R₁ and R₂)

\[ \text{Dose} = \text{Gain} \times (R₁ - R₂ \times \text{CLR}). \]

As stated above, we compared the correction for Čerenkov radiation, that is implemented in the electrometer software, against our manual calculation.

2.B. Energy and modality dependence

We evaluated the energy and radiation modality dependence of the detector response for high energy photon and electron beams delivered by a linear accelerator Clinac 2100C/D. Table I specifies the beam quality indices for each energy and modality. To verify the energy and/or modality dependence, we compared the dose measured by the scintillator to that measured by an ionometric system in reference conditions. All chambers used in this work had calibration coefficients traceable to secondary laboratories for the beam qualities.

Although Plastic Water is water-equivalent for high-energy photon and electron beams within 0.5% in the absorbed dose, we performed dose measurements in water with a PTW MP3 Waterphantom (PTW, Freiburg, Germany) to eliminate the uncertainty related to this slight non-water equivalence. For high-energy photon beams, we used a PTW 30013 Waterproof Farmer Chamber, while for electron beams we measured the dose with a NACP02 (IBA dosimetry, Schwarzenbruck, Germany) plane-parallel chamber. For consistency reasons, the value of Dose₁₀ in Eq. (2) was replaced by the dose measured for 6 MV photons in water because all values in this section referred to water.

We investigated whether the CLR value could depend on the beam modality, the beam energy or the amount of optical fiber lying close to the phantom. For electron beams, we were unable to directly determine the CLR value using the calibration plate because it is designed for a 40 × 40 cm² field that cannot be set in a Varian Clinac 2100 Linac (the maximum electron applicator is 25 × 25 cm²). We explored three methods to obtain CLRs. First, we placed the scintillator in the calibration plate at the reference depth under a 25 × 25 cm² electron field, and we increased the source–surface distance up to SSD = 140 cm. At this distance, we achieved a field size that exceeded the phantom. Second, we set SSD = 100 cm and we followed the standard CLR determination procedure. At this distance, some parts of the fiber did not lie within the beam at the maximum fiber configuration and, consequently, this setup was not expected to yield accurate results. However, it was included to clarify whether the precise value of the CLR could be conditioned by the increased SSD to 140 cm. Third, we placed the scintillator in minimum fiber configuration and we obtained the CLR value by comparing the output under the 6 × 6 cm² applicator to that under the 25 × 25 cm² applicator. The sub-indices “min” and “max” in formula (1) in this third case corresponded to the 6 × 6 cm² and 25 × 25 cm² applicators, respectively. We obtained CLRs for the most extreme electron energies, 6 and 20 MeV, using the first method. We applied the second and third methods for 6 MeV electron beams only.

2.C. Short-term repeatability

We obtained short-term repeatability by irradiating the Epxradin W1 dosimeter 40 times in reference conditions for 6 and 15 MV x-ray beams and evaluating the standard deviation. We performed three tests. For the first and third tests, we delivered 100 MU (~0.75 Gy) in each irradiation. For the second test, we gave only 20 MU (~0.15 Gy) to see whether the dose value affected repeatability. The intention of the third test was to show the effect of the recording method (manual versus automatic trigger in channel 1).

2.D. Dose-response linearity

We determined the linearity of the response with the dose in reference conditions for 6 and 15 MV x-ray beams for doses between 0.15 and 6 Gy. We performed this test both for the manual and for the automatic triggered collection modes. As any potential drift in the beam monitoring system would mask the results in this test, we used the NE2571 ionization chamber to monitor the beam output, and hence to correct for such an effect. We placed this chamber in a Plastic Water slab located beneath the scintillator with a 1 cm offset with respect to the beam axis.

Table I. Quality indices of the various modalities and energies studied for the energy dependence test. These indices are TPR₂₀,₁₀ for high-energy x-rays, R₅₀ (g cm⁻²) for high-energy electrons.

<table>
<thead>
<tr>
<th>Modality</th>
<th>Nominal energy</th>
<th>TPR₂₀,₁₀</th>
<th>R₅₀ (g cm⁻²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>High-energy x-rays</td>
<td>6 MV</td>
<td>0.685</td>
<td>—</td>
</tr>
<tr>
<td></td>
<td>15 MV</td>
<td>0.761</td>
<td>—</td>
</tr>
<tr>
<td>Electrons</td>
<td>6 MeV</td>
<td>—</td>
<td>2.460</td>
</tr>
<tr>
<td></td>
<td>9 MeV</td>
<td>—</td>
<td>3.686</td>
</tr>
<tr>
<td></td>
<td>12 MeV</td>
<td>—</td>
<td>5.138</td>
</tr>
<tr>
<td></td>
<td>16 MeV</td>
<td>—</td>
<td>6.733</td>
</tr>
<tr>
<td></td>
<td>20 MeV</td>
<td>—</td>
<td>8.493</td>
</tr>
</tbody>
</table>
2.E. Angular dependence

We placed the scintillator at the Linac isocenter in air, within a 1.5 cm radius home-made cylindrical buildup cap made of bolus (Superflab, Mick Radiation Nuclear Instruments, Mount Vernon). This buildup cap has a central hole to insert the scintillator. From any angle, the beam always traversed 1.5 cm of a water-equivalent material before reaching the scintillator. We irradiated the scintillator with this buildup cap every 30° under a 10×10 cm² 6 MV x-ray beam.

2.F. Temperature dependence and time to reach thermal equilibrium

We used a water phantom equipped with a thermostat (Fig. 2) to study the influence of temperature on the scintillator signal. The scintillator was tapped on a thin slab of PMMA that was in contact with the water. We measured the temperature with a thermistor provided with an immersion probe that was attached to the PMMA slab beside the scintillator. The temperature was increased from 22°C to 38.5°C. Each temperature was maintained for approximately 10 min in order to reach thermal equilibrium between the water and the scintillator. A water pump stirred the water to keep its temperature homogeneous. We performed dose measurements at SSD = 100 cm for a 10×10 cm² x-ray beam at seven temperatures ranging from 22°C to 38.5°C. We recorded the temperature before and after each measurement and included differences in temperature in the uncertainty budget. We did this test for 6 and 15 MV x-ray beams.

To evaluate the time to reach thermal equilibrium, we placed a Plastic Water phantom at room temperature (25°C) beside the thermal water phantom. We chose a high temperature of the water phantom (38°C) in order to obtain an upper limit on the time it takes to reach thermal equilibrium for Exradin W1. We kept the distance from the radiation source to the anterior surfaces of both phantoms at 100 cm. Therefore, their placement under the radiation beam required only a simple longitudinal movement of the treatment table. We first attached the scintillator to the thermal water phantom. We then moved it onto the Plastic Water phantom and switched on a chronometer. Then, the room was quickly vacated and we irradiated the scintillator several times until its response was stable. In a second round, we moved the scintillator back to the thermal bath and performed another set of measurements, also keeping track of the timing. Finally, we moved it back to the Plastic Water phantom and performed a new set of measurements, again recording the time. Summarizing, we recorded two sets of measurements while cooling the scintillator and one while heating it up. We defined the time to reach thermal equilibrium as the point of time at which the dose measurement no longer changed in more than the short-term repeatability.

2.G. Dose rate dependence

This section consisted of two tests.

First, we checked the scintillator response on the six dose rates given by the Linac, ranging from 100 to 600 MU/min every 100 MU/min. This test will be referred from now on as “Repetition Rate” test. It was performed in reference conditions for 6 and 15 MV photons. For this test, we used a monitor ionization chamber as described in Sec. 2.D to correct for beam dose rate variations. However, as Varian accelerators do not change the dose per pulse but the number of pulses per unit time, this test did not truly check the dose-rate dependence.

Second, we performed measurements changing the source-detector distance (SDD) as the change in the SDD clearly changes the dose rate. We performed measurements at six SDDs ranging from 80 to 130 cm every 10 cm. Any deviation with respect to the Inverse Square Law (ISL) can be attributed...
to dose-rate dependence. On continuation, we refer to this test as the “ISL” test.

We performed the ISL test by placing the scintillator at the Linac isocenter in air with a buildup cap under a narrow beam (5 × 5 cm²). We investigated two buildup caps. First, we used the 1.5 cm radii cap made of bolus from Sec. 2.E for 6 MV x-ray beams only. This cap was not used for 15 MV as it did not guarantee electron equilibrium for this energy. Second, we tried a 9.5 mm radius aluminium bronze alloy cap (5 cm water equivalent thickness) for both 6 and 15 MV photons. As the uncertainty in the positioning of the detector under a small field size at 80 cm has a direct impact on the result of this test, we performed two independent determinations and we derived a positioning-related B-type uncertainty. This was included in the reported uncertainty.

2.H. Water-equivalent depth of the effective measurement point for x-ray beams

We placed the scintillator on the calibration plate and we mounted it onto a 12 cm slab phantom of Plastic Water to provide backscatter. The SDD was set at 100 cm to the calibration plate. Then, we measured TMRs for a 10×10 cm² field for 6 and 15 MV photon beams. We compared these TMRs with TMRs measured using the NACP chamber to estimate the water-equivalent depth of the effective measurement point of Exradin W1. Note that this geometry, with the scintillator axis perpendicular to the radiation beam, is not valid for field sizes less than 1.5×1.5 cm².

2.I. Long-term stability

We derived long-term stability from the change in the calibration coefficient during the present study. The total accumulated dose was 127 kGy. We applied the temperature correction factor obtained in Sec. 2.F.

2.J. Uncertainty budget

We defined an uncertainty budget including all dependencies except those we can correct (temperature effects and loss of sensitivity with accumulated dose). This overall uncertainty would therefore be the quadratic sum of these dependencies. We considered two scenarios: first, relative measurements considering the results of the previous tests and including the 0.7% maximum difference found by Guillot et al. to account for different field sizes; and second, reference dose measurements. For this second scenario, we included the uncertainties in dose calibration for the various modalities and energies as detailed in TRS-398.

3. RESULTS AND DISCUSSION

3.A. Calibration

We verified that the electrometer software correction for Čerenkov radiation was correctly implemented. The results of formulas (1) and (2) compared to the results from the Table III. Percentage differences between the dose measured using the scintillator calibrated at 6 MV x-rays and the dose measured using the ionization chamber. The uncertainty is expressed with \( k = 1.0 \).

<table>
<thead>
<tr>
<th>Modality</th>
<th>Nominal energy</th>
<th>Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>High-energy x-rays</td>
<td>6 MV</td>
<td>—</td>
</tr>
<tr>
<td></td>
<td>15 MV</td>
<td>−0.2 ± 0.6</td>
</tr>
<tr>
<td>Electrons</td>
<td>6 MeV</td>
<td>−0.5 ± 0.7</td>
</tr>
<tr>
<td></td>
<td>9 MeV</td>
<td>−1.2 ± 0.6</td>
</tr>
<tr>
<td></td>
<td>12 MeV</td>
<td>0.5 ± 0.6</td>
</tr>
<tr>
<td></td>
<td>16 MeV</td>
<td>−0.4 ± 0.6</td>
</tr>
<tr>
<td></td>
<td>20 MeV</td>
<td>−0.2 ± 0.6</td>
</tr>
</tbody>
</table>

electrometer software were negligible (<0.06%) and due to electrometer resolution.

3.B. Energy and modality dependence

Table III compares the dose measurements made by the scintillator with those made by the ionization chamber for different modalities and energies. The third column shows the percentage dose difference obtained using the calibration coefficient calculated for 6 MV x-rays without any further correction. Differences were negligible taking into account the uncertainties with a coverage factor \( k = 2 \). We can affirm that Exradin W1 shows an energy-independent response for megavoltage photon and electron beams. As the response of the scintillator was energy independent in this range, for all remaining sections except Sec. 2.H, we merged the data for 6 MV and for 15 MV x-rays and we analyzed all data together. Any energy dependence is therefore also incorporated within the uncertainty budget.

Regarding CLR dependencies, for photon beams, changing the length of the fiber exposed to scatter radiation had a strong impact on raw readings for both channels. However, this impact was not so strong on the CLR value, yielding a standard deviation of only 0.5%. For all electron beams, CLRs obtained with the different methods and the different energies agreed within 0.9%. We used the average value of 15 independent determinations (mostly for photon beams) and took the 1.5% standard deviation from all CLR values into account in its uncertainty.

As Table III shows, we obtained clinically acceptable differences (around or less than 1%) for high-energy x-rays and electron beams.

3.C. Short-term repeatability

Table IV shows the results when we performed an automatic triggered collection on channel 1. However, if the collected charges were manually triggered, the result of this test increased up to 0.40% ± 0.17% and 2.22% ± 0.17% for 0.75 and 0.15 Gy, respectively. Due to the low currents, timing effects are of paramount importance. We strongly recommend a dual electrometer with automatic triggered collection in order to avoid the uncertainty related to poor repeatability that becomes unacceptable for dosimetry purposes at low doses.
Table IV. Main characteristics of Exradin W1. The uncertainty column includes all sources of uncertainty described in Sec. 2 for each test, with the exception of temperature dependence and loss of sensitivity with accumulated dose tests. For these tests, the statistical uncertainty in the slope of the least-square fit is shown. We calculated the root-mean-square value (RMS) as $\text{RMS} = \sqrt{\frac{1}{n} \sum_{i=1}^{n} x_i^2}$, where “$x_i$” are percentage deviations from a reference value and $i = 1$ to n are all measurements within the corresponding test.

<table>
<thead>
<tr>
<th>Test</th>
<th>Result</th>
<th>Uncertainty</th>
</tr>
</thead>
<tbody>
<tr>
<td>Short-term repeatability (at 0.75 Gy)</td>
<td>$\sigma = 0.10%$</td>
<td>0.07%</td>
</tr>
<tr>
<td>Short-term repeatability (at 0.15 Gy)</td>
<td>$\sigma = 0.25%$</td>
<td>0.05%</td>
</tr>
<tr>
<td>Dose–response linearity</td>
<td>RMS = 0.61%</td>
<td>0.20%</td>
</tr>
<tr>
<td>Angular dependence</td>
<td>RMS = 0.21%</td>
<td>0.07%</td>
</tr>
<tr>
<td>Temperature dependence</td>
<td>$-0.225% , ^{\circ}\text{C}^{-1}$</td>
<td>0.008% $^{\circ}\text{C}^{-1}$</td>
</tr>
<tr>
<td>Time to reach thermal equilibrium$^a$</td>
<td>1 min 40 s 16 s</td>
<td>16 s</td>
</tr>
<tr>
<td>Repetition rate dependence</td>
<td>RMS = 0.53%</td>
<td>0.06%</td>
</tr>
<tr>
<td>Deviation from ISL</td>
<td>RMS = 0.38%</td>
<td>0.26%</td>
</tr>
<tr>
<td>Loss of sensitivity with accumulated dose</td>
<td>$-0.28% , \text{kGy}^{-1}$ (0–15 kGy)</td>
<td>0.06% kGy$^{-1}$</td>
</tr>
<tr>
<td></td>
<td>$-0.032% , \text{kGy}^{-1}$ (15–127 kGy)</td>
<td>0.018% kGy$^{-1}$</td>
</tr>
</tbody>
</table>

$^a$For the conditions defined in Sec. 2.F.

The short-term repeatability obtained in this work was better than that reported by Lacroix et al.$^{13}$ for an array of 29 plastic scintillator detectors embedded in a water-equivalent plastic sheet (0.8%), worse than that reported by Beddar et al.$^{14}$ for a Bicron Co. BC-400 PSD (0.1%), and much lower than that reported by Andersen et al.$^{15}$ for a completely different kind of dosimeter like the Al$_2$O$_3$:C luminescence dosimetry system (1.3%).

3.D. Dose–response linearity

For most situations, the deviation from linearity was well below 0.5%. However, there was a maximum deviation of $–1.9\%$ for a very low dose (0.075 Gy), leading to the RMS value shown in Table IV (0.61%). The result of this test with manual triggered collection yielded worse results, reaching values as high as $–10\%$ for this very low dose. These effects are also due to the timing effects. They can be avoided by increasing the dose, and with an automatic trigger. An unexpected finding, we observed when performing this test was that when environmental relative humidity levels were high (around 75%), leakage currents in the electrometer became significant when measuring small doses. This value is consistent with the operating range given by the manufacturer$^{16}$ (relative humidity between 20% and 80%).

3.E. Angular dependence

The maximum angular dependence was 0.34% ± 0.07%, only slightly larger than the RMS shown in Table IV. In this section, the RMS value refers to percentage differences with respect to the value measured at 0°. The build-up cap used for this test was home-made, adding an uncertainty of unknown magnitude due to little flaws in its design. Consequently, the result of this test must be taken as an upper limit. However, with the excellent result reported in Table IV, we can state that Exradin W1 shows an almost isotropic response around its symmetry axis.

3.F. Temperature dependence and time to reach thermal equilibrium

Temperature dependence is shown in Fig. 3 and summarized in Table IV. It is linear and almost of the same magnitude as that observed for diodes$^{17}$ but it has an opposite sign. The decrease in the light output was close to that obtained for a BCF-60 PSD (0.32% $^{\circ}\text{C}^{-1}$) by Wooton and Beddar.$^{18}$ For absolute dose measurements, it should be included as a correction factor. Otherwise, the uncertainty would increase significantly if the temperature was different from that at calibration, i.e., in a potential application for in-vivo dose measurements. The results of the present study confirm significant temperature dependence, thus disagreeing with claims that PSDs response is temperature independent$^{19,20}$ and with the manufacturer’s technical sheet.$^{16}$

Figure 4 shows that after 40 s (the time to quickly leave the treatment room and continue the irradiation), the reading differed 1% from the thermal equilibrium value, after 1 min,
less than 0.3%; and at only 1 min 40 s, the detector reached thermal equilibrium, defined as the time the detector response no longer changed in more than 0.1% (short term repeatability), as detailed in Table IV. The temporal behavior of the Exradin W1 response was the same when heating as when cooling. This issue should not therefore be a concern in clinical routine as the time to vacate the room after placing the detector will always exceed the time to reach thermal equilibrium.

3.G. Dose rate dependence

The repetition rate test resulted in values of up to 0.5% with the exception of the lower repetition rate, 100 MU/min. In this case, we obtained a difference of −1% and a higher uncertainty of 0.8%. Table IV shows the RMS value of the test. In this case, the RMS value referred to percentage differences with respect to the dose measured at a repetition rate of 300 MU/min. The results for Exradin W1 are different from those reported by Beddar et al.14 for a BC-400 PSD (dose-rate independence within ±0.1%) between 80 and 400 MU/min at a Varian Clinac 2100C Linac.

When investigating the dose rate dependence by verifying the ISL with the metal cup, we obtained acceptable differences (<0.6%) and the RMS of all differences was low (0.38% ± 0.26%), as shown in Table IV (for this test, RMS values referred to percentage differences with respect to the dose measured at a SDD = 100 cm). The values we obtained with the metal cap agreed with those obtained with the bolus cap (0.36% ± 0.14%).

3.H. Water-equivalent depth of the effective measurement point for x-ray beams

Figure 5 shows TMR curves obtained with the NACP02 plane-parallel chamber measurements for 6 and 15 MV x-ray beams, and those measured with the scintillator. To achieve the best match between the scintillator and the ionization chamber for both energies, the scintillator TMRs have been shifted 0.8 mm in depth. We could therefore say that the effective measurement point is 0.8 mm from its surface. The diameter of the scintillator housing is 2.8 mm, so the effective measurement point is 0.6 mm upstream respect to the Exradin W1 axis center. The uncertainty bars have not been included in Fig. 5 for the sake of clarity, but are around 0.2%.

3.I. Long-term stability

Figure 6 shows the loss of sensitivity with accumulated dose. We can see two clearly different regions, one for accumulated doses less than 15 kGy and another for accumulated doses over 15 kGy. The slopes of these linear fits are shown in Table IV. The scintillator loss of sensitivity with accumulated dose behavior is similar to that of diodes, although the rational might be completely different. The value reported by Beddar et al.20 for a BC-400 PSD was 2.8% for 10^4 Gy, coinciding exactly with the value reported in Table IV (−0.28% ± 0.06% kGy⁻¹), corresponding to the first fit in Fig. 6.

3.J. Uncertainty budget

In the first scenario (relative dose measurements), the combination of all uncertainties for high-energy photon beams yielded a value of 0.92%, provided that temperature correction was performed and that calibration was recent. For
high-energy electron beams, the overall uncertainty would be between 1.0% and 1.7%. Table V shows the composite uncertainty for the different modalities. The last column corresponds to the second scenario (reference dose measurements). It was obtained by including the uncertainty estimates given by TRS-398 for dose measurements in reference conditions for these modalities.

### Table V. Uncertainty budget \((k = 1)\) for different beam qualities. Dose measurement uncertainties are the quadratic sum of the relative measurement estimates and the estimated relative standard uncertainty of \(D_{eq}\) as given by TRS-398 in reference conditions \((1.5\% \text{ for high-energy photon beams and } 2.1\% \text{ for electron beams})\).

<table>
<thead>
<tr>
<th>Modality</th>
<th>Relative measurement (%)</th>
<th>Dose measurement (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>High-energy x-rays</td>
<td>0.9</td>
<td>1.8</td>
</tr>
<tr>
<td>High-energy electrons</td>
<td>1.2</td>
<td>2.4</td>
</tr>
</tbody>
</table>

4. CONCLUSION

The ExradinW1 scintillator presents excellent dosimetric characteristics, making it a good alternative to other detectors, such as diodes, for relative dose measurements. It is especially useful for small field sizes due to its small dimensions and because the Čerenkov effect is minimized. The main advantage of the ExradinW1 scintillator over diodes is its water equivalence that makes its response independent from energy for high energy x-rays and electron beams. A temperature correction factor should be applied to keep uncertainties around 2% for dose measurements.

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\(^{16}\) Standard Imaging Exradin W1, Reference 92739, DOC #80643-00.


