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Exradin W1 plastic scintillation detector for *in vivo* skin dosimetry in passive scattering proton therapy

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Abstract

In vivo skin dosimetry is desirable in passive scattering proton therapy because of the possibility of high entrance dose with a small number of fields. However, suitable detectors are needed to determine skin dose in proton therapy. Plastic scintillation detectors (PSDs) are particularly well suited for applications in proton therapy because of their water equivalence, small size, and ease of use. We investigated the utility of the Exradin W1, a commercially available PSD, for *in vivo* skin dosimetry during passive scattering proton therapy. We evaluated the accuracy of the Exradin W1 in six patients undergoing proton therapy for prostate cancer, as part of an Institutional Review Board-approved protocol. Over 22 weeks, we compared *in vivo* PSD measurements with in-phantom ionization chamber measurements and doses from the treatment planning system, resulting in 96 *in vivo* measurements. Temperature and ionization quenching correction factors were applied on the basis of the dose response of the PSD in a phantom. The calibrated PSD exhibited an average 7.8% under-response ($\pm 1\%$ standard deviation) owing to ionization quenching. We observed 4% under-response at 37°C relative to the calibration-temperature response. After temperature and quenching corrections were applied, the overall PSD dose response was within $\pm 1\%$ of the expected dose for all patients. The dose differences between the PSD and ionization chamber measurements for all treatment fields were within $\pm 2\%$ (standard deviation 0.67%). The PSD was highly accurate for *in vivo* skin dosimetry in passively scattered proton beams and could be useful in verifying proton therapy delivery.

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Keywords

plastic scintillation detector; *in vivo* dosimetry; skin dose; proton therapy

Introduction

In vivo dosimetry (IVD) is an important component of a comprehensive quality assurance program as a final check at the point of radiotherapy delivery; IVD can measure target doses, prevent dose delivery errors, and enhance the standard of patient care. IVD has been widely used in conventional external beam radiotherapy [1] and brachytherapy [2]. In external beam radiotherapy, IVD is mostly used for entrance dose measurement because it is easy to place a detector on the skin of the patient. For photon beam radiotherapy, transmission measurements using an electronic portal imaging device (EPID) are also gaining interest for IVD [3]. However, in proton therapy such a technique is not possible because there is no exit dose. Because the entrance dose of modulated proton beams is much higher than that of photon beams, IVD is expected to be useful in proton therapy for monitoring the skin dose. IVD also opens up the possibility eventually to detect gross errors such as, malfunctions in the delivery system and use of wrong devices during the course of the treatment.

Many of the detectors used in IVD for external beam radiotherapy and brachytherapy have been investigated for proton therapy [4,5]. An ionization chamber, considered to be the gold standard for radiation dosimetry, cannot be used *in vivo* because it is fragile and requires a large applied voltage bias that may not be safe for placement on the patient. Most of the detectors commonly used for IVD in photon therapy have some features, such as dose, dose rate and linear energy transfer dependence, offline reading, neutron damage, and angular dependence, which make their use in proton therapy problematic.

Plastic scintillation detectors (PSDs) possess several characteristics that make them highly advantageous for IVD. PSDs can be utilized for real-time dosimetry, are resistant to radiation damage (allowing extended reuse), and are capable of high levels of precision and accuracy [1]. The response of PSDs has been found to be unaffected by angle of incidence, dose rate, and total dose [6,7]. PSDs are water-equivalent; as such, their presence does not perturb radiation fields, allowing measurements of treatment fields to be taken without compromising the treatment. The water equivalency of PSDs also means that no conversion factor is needed to convert the dose to the detector to the dose that would be received by an equivalent volume of tissue [6,7].

However, some challenges remain to be addressed before PSDs can be widely used for IVD in proton therapy. The scintillation light produced by PSDs exhibits both ionization quenching (a nonlinear response to high linear energy transfer radiation) and temperature dependence [8]. Both can be corrected for using empirically determined correction factors. PSD temperature response has been found to be linear [9,10]. Therefore, a correction factor based on the patient's skin temperature must be used. Ionization quenching can be corrected for by characterizing the PSD under-response compared with the response of an ionization chamber as a function of proton energy, as demonstrated in a previous study that

characterized a PSD for use as a beam entrance dosimeter in a passively scattered proton beam [8,11].

Patients receiving passively scattered proton beam therapy have an increased risk of radiation dermatitis compared with patients treated with other external radiation modalities. This effect is caused by the limited number of beams used in proton therapy and the lack of skin sparing [12]. PSDs have been studied for proton dosimetry in the form of scintillator screens for two-dimensional measurements, but they have never been applied to real-time *in vivo* skin dosimetry [13,14]. In the present study, we investigated the utility of the Exradin W1, a commercially available PSD, for *in vivo* skin dosimetry during passively scattered proton beam therapy. We compared in-phantom and *in vivo* dose measurements to determine temperature and quenching correction factors. We found that we could successfully verify the dose delivered at the entrance of a proton beam by using a PSD *in vivo*. For full characterization of the Exradin W1 scintillator in radiotherapy we direct you to read the publications by Beierholm *et al.* and Carrasco *et al.* [15,16].

Methods

A. Detectors

The Exradin W1 scintillator (Standard Imaging Inc., Middleton, WI, USA) was used for the present study. The detector consists of a scintillating fiber 1.0 mm in diameter by 3.0 mm in length (Figure 1). The detector was connected to a two-channel SuperMAX Electrometer (Standard Imaging Inc.). For calibration, the electrometer's two channels were used to detect the separated blue and green components through the dual-channel photodiode using the chromatic removal technique [17-19], which removes the light contribution from Cerenkov radiation [20]. The chromatic removal technique takes advantage of the fact the light signal measured from the detector will have two independent components, one that is composed of scintillation light and proportional to dose and another that is composed of Cerenkov radiation [17]. To extract these components a linear set of equations is used, each irradiated with different condition. In this work, a ^{60}Co unit was used to calibrate the W1 and extract the two factors C_s and C_{ck} to calculate dose. C_s is the scintillation gain factor of the PSD, and C_{ck} is the Cerenkov light contribution reduction factor. The calibration irradiation conditions were done in water at depth 0.5 cm for 2 min irradiation time. The field sizes were 5 cm by 5 cm, 10 cm by 10 cm and 15 cm by 15 cm. The PSD dose *in vivo* was then calculated using the following equation:

$$D_{in\ vivo} = C_t C_q C_s [M_1 - C_{ck} M_2]$$

where C_t is the temperature correction factor, C_q is the quenching correction factor, M_1 and M_2 are the two channel outputs. The methods by which C_t and C_q were determined are outlined in the temperature response and quenching calibration sections below.

To compare absolute doses in the proton beam, we used a calibrated plane-parallel ionization chamber (PTW 34045; PTW, Freiburg, Germany) with an active volume of 0.02 cm³ to determine the dose, following the IAEA TRS 398 protocol [21]. A 0.9-mm buildup cap was

used for the plane-parallel ionization chamber because the active scintillator volume housing of the Exradin W1 PSD has a 2.8-mm-diameter jacket, which results in 0.9 mm of buildup material surrounding the scintillator. All ionization chamber readings were corrected for ambient pressure and temperature. The ionization chamber was operated at +300 V and the charge was read using a Scanditronix/Wellhofer electrometer (Scanditronix/Wellhofer North America, Bartell, TN, USA). Background subtraction was used for both the SuperMax and the Scanditronix/Wellhofer electrometers to determine the amount of light (Supermax) or charge (Scanditronix/Wellhofer) generated by the proton beam only. To verify the accuracy of the dose calculated by the Eclipse treatment planning system (Varian Medical Systems, Inc., Palo Alto, CA, USA), we compared the skin dose from the treatment planning system with that measured using the plane-parallel ionization chamber. The treatment planning system dose was defined at the central axis of the treatment field using a point that is 0.9 mm below the skin surface.

B. Temperature response

To assess temperature response, we irradiated the PSD with a ^{60}Co beam in a temperature-controlled water tank while all other conditions remained constant. The two temperature points used were 20°C and 37°C. The temperature dependence of PSDs is linear and can be corrected for [9]; the correction factor we used was simply the ratio of the PSD dose response at skin temperature to the temperature at which the PSD was calibrated.

To estimate the equilibrium temperature of a PSD attached to skin, we measured the skin temperature of a volunteer using a metal thermal probe encased in a polyethylene jacket to simulate the conditions of the scintillator volume as closely as possible (Taylor 9867FDA Digital Thermocouple; Taylor precision products, Las Cruces, NM, USA). The thermometer had an accuracy of $\pm 1^\circ\text{C}$. The thermometer was placed on the volunteer's skin laterally on the abdomen and temperature readings were recorded after 2 minutes. The thermometer was then reset and the readings were conducted four more times. A single correction factor based on the mean of these measurements was considered to account sufficiently for temperature response, because variations of a few degrees in either direction produced a change of less than 1% in the detector response.

C. Quenching calibration

To account for ionization quenching, the PSD and the plane-parallel ionization chamber were simultaneously irradiated in a phantom with a single open-field (aperture size $9.8\text{ cm} \times 9.8\text{ cm}$) proton beam with a range of 26.9 cm (225 MeV) and a spread out Bragg peak (SOBP) width of 8 cm (typical field parameters for the patients enrolled in the study). The phantom used was a plastic water block (CIRS, Norfolk, VA, USA) with an ionization chamber slot. The PSD was attached in front of the ionization chamber and to its side outside of the active volume surface as demonstrated in figure 2 [11]. The snout position was set at 30 cm. It is important to note that in a previous study by Wootton *et al* comparing a PSD with a plane-parallel ionization chamber, a snout position of 45 cm was used, whereas we used 30 cm, as required for patient treatment fields [11]. Thus, the air gap was 15 cm larger in the previous study than in the present study. In proton therapy, a smaller air gap between the beam nozzle and patient surface is preferred because it reduces the beam's

lateral penumbra¹⁹, which helps to spare normal tissue and critical organs. Because of this difference in the size of the air gap, we conducted further measurements comparing snout positions. The same single open-field beam was delivered at different snout positions (30, 37, and 45 cm) as well as at one of the patient's treatment fields with the appropriate apertures and compensators.

To determine whether the aperture and compensators affected the PSD dose response, we compared the PSD response with that of the ionization chamber measurements for patient-specific treatment fields in the phantom using the custom apertures and compensators for each patient. The average ratio of PSD-to-ionization chamber measurements from the open-field beam was used to generate a quenching correction factor for the *in vivo* measurements. Because the patient-specific treatment field aperture and compensators were not found to have a substantial effect on the PSD response, the final quenching correction factor (C_q) was calculated from the single open-field proton beam. Table 1 lists some field parameters for each patient in this study.

D. Stability measurements

Weekly baseline measurements were obtained for 22 weeks to assess the stability of the PSD response over time and thereby its ability to accurately measure patient skin dose. To determine the baseline measurements, we used a reproducible setup in a ⁶⁰Co unit with the Exradin W1 PSD calibration slab on top of a gantry attachment. The baseline dose was measured with a 10 cm × 10 cm field and an 8-mm buildup slab. The acquisition time was 2 minutes. The source decay was accounted for in the comparisons.

E. In vivo skin dose measurements

Six patients undergoing passively scattered proton beam therapy for prostate cancer were selected for our study of *in vivo* skin dosimetry. We chose prostate cancer treatment plans because the skin is easily accessible with little discomfort to the patient. Written informed consent was obtained from all patients, and the study protocol was approved by the Institutional Review Board of The University of Texas MD Anderson Cancer Center. The PSD was fixed to the patient's skin with medical tape at the central axis of the two laterally opposed treatment fields using the isocenter setup lasers. Due to manual positioning we expect a positioning accuracy of 1-2 mm. Dose measurements were obtained once per week for each patient for the duration of treatment, resulting in a total of 96 *in vivo* measurements. The dose measured *in vivo* by the PSD was compared with the expected dose, which had been measured at the surface of a phantom with a calibrated plane-parallel ionization chamber according to patient treatment plan parameters, including the patient-specific treatment field apertures and compensators discussed above. The average difference over all measurements and per-patient differences ± standard deviations were computed.

Results

A. Temperature response

The PSD showed an under-response of 4% when used at 37°C relative to its response at the calibration temperature of 20°C. The average skin temperature of the volunteer across 3

measurements was $32.4 \pm 1^\circ\text{C}$, resulting in a temperature correction factor (C_t) of 1.027. This value was applied to all measurements across all patients. Naturally, skin temperature varied among patients. However, with the above correction factor, a 2°C change in skin temperature would result in only a 0.4% change in dose. This shows that temperature variations among patients can be expected to have only minor effects on dose discrepancies, likely less than 0.5%.

B. Ionization quenching correction

We found that the average under-response of the PSD attributable to ionization quenching was $7.8\% \pm 1.0\%$, similar to the results published by Wootton *et al* [11]. The under-response for each field is shown in Figure 3. One field showed an under-response of 9.4%; however, as we show below, the average ionization quenching correction factor yielded acceptable results when we compared the PSD dose with the dose measured by the plane-parallel ionization chamber. The single open-field proton beam exhibited a similar under-response, $7.8\% \pm 0.5\%$, suggesting that the patient-specific treatment field apertures and compensators had little to no effect on the dose response of the PSD. Because it would be more clinically practical to use the quenching correction factor of the single open-field proton beam, we applied the quenching correction factor (C_q) of 1.078.

The patient treatment field under-response was greater at larger air gaps (10% under-response). However, the open-field beam showed little difference at different snout positions, falling within the overall uncertainty of 1%. We concluded that when the PSD was used at the minimum air gap distance, the quenching correction factor used here remained valid.

C. Stability measurements

The ^{60}Co calibration baseline results shown in Figure 4 illustrate that the PSD provided a stable and accurate dose response over time. The dose was normalized to the average value over 22 weeks. At least four measurements were conducted each week.

D. In vivo measurements

Skin doses were measured for at least eight fractions of treatment for each patient. The average dose difference between the PSD and the plane-parallel ionization chamber across all patient measurements was $0.27\% \pm 0.67\%$. The average dose difference between the plane-parallel ionization chamber and the calculated treatment planning dose was $0.56\% \pm 0.92\%$. Figure 5 shows the differences in each patient's per-field doses. Each measurement performed *in vivo* was plotted for each day in Figure 6. All measurements fell within $\pm 2\%$ of the dose calculated by the treatment planning system.

Discussion

The present study demonstrates that PSDs can be used for accurate and practical *in vivo* skin dosimetry in proton therapy. Measuring the quenching correction factor by comparing PSD and ionization chamber doses for a single open-field proton beam simplified the setup and minimized the time needed to calibrate the PSD and verify the dose delivered. The proton beam, in particular, passively scattered proton beam, was not perturbed by the presence of

the PSD because PSDs are small and water-equivalent. With parallel opposed field arrangement with adequate distal and proximal margins added for every field, the effect of additional 0.28 cm of water-equivalent material in the path of the beam was found to be negligibly small (on average $< 0.1\%$) on the target coverage. Typical distal and proximal margins for prostate patients are around 1.2 cm and 0.9 cm respectively (The distal margin is 3.5% of the range of the beam + 3 mm, and the proximal range is 3.5% of the (range-SOBP width) + 3mm), which are much larger than the 0.28 cm thickness of the PSD. In evaluating the effect of differences in air gap/snout placement on the response of the PSD, we have shown that the quenching correction factor was similar to that used in the Wootton *et al* study when the PSD was used with small air gaps [11]. Using PSDs for general IVD in proton dosimetry (for example, measuring rectal wall dose) is a more difficult problem because the conditions would vary more from beam to beam and patient to patient and would be harder to determine (e.g. range uncertainty would be a problem).

The availability of a commercial system enables any clinic to obtain a PSD and utilize its advantages for IVD. Wootton *et al* previously demonstrated the feasibility of using PSDs for entrance proton beam dosimetry and here we have confirmed and demonstrated the accuracy of PSDs for *in vivo* skin dosimetry using a commercial system. Although our protocol was limited to patients with prostate cancer, IVD with PSDs could also be very useful for cancer at other sites, such as the lung and breast [23], where radiation dermatitis can be a limiting factor for proton therapy [24]. The PSD can be used to monitor the skin dose delivered to these other sites in a similar fashion to the methods used in the present study. While the number of patients was small, the purpose of the study was to demonstrate the utility of the Exradin W1 detector for IVD. Thus, the total number of *in vivo* measurements (n=96) demonstrated agreement between expected and measured dose to within $\pm 2\%$. In the future, it would be interesting to investigate inter-patient variability by increasing patient sample size, and other cancer sites.

Our study was also limited to only passively scattered proton beams. It has been shown that in spot-scanning proton beams, the skin dose is 12% lower than that received in treatment with passively scattered proton beams [25]. This could mean that skin toxicity is not as critical for spot-scanning proton beams. The Exradin W1 PSD can still be used in spot scanning proton beams, although a quenching correction factor would need to be calculated for each set of proton beam spot energies. Furthermore, detector placement plays a role in PSD accuracy, and further investigation is needed to determine the dose effects at positions away from the central axis.

Conclusion

We have shown that the Exradin W1 PSD is a strong candidate for *in vivo* skin dosimetry in passively scattered proton beam therapy. PSDs are water-equivalent and very small, permitting them to produce accurate measurements that do not perturb the delivered dose. The quenching and temperature corrections are easy to calculate, as shown in the present study and in a previous study by Wootton *et al*. [11]. The overall PSD dose response was within $\pm 1\%$ for all patients after we applied temperature and quenching corrections. The PSD exhibited a stable response over the course of the 22 weekly measurements.

Furthermore, PSDs were used for IVD without interfering with the clinical workflow, demonstrating that their implementation is practical and useful.

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Highlights

- *In vivo* skin dosimetry method for proton therapy.
- Used the commercially available Exradin W1 plastic scintillation detector.
- *In vivo* measurements were compared to in-phantom ion chamber measurements.
- Method did not alter patient treatment plans and treatment workflow.

Figure 1.
The Exradin W1 scintillator and SuperMAX Electrometer.

Figure 2. A sketch of the beams-eye view of the in-phantom setup

The ion chamber (dashed line) is placed in its slot inside the water-equivalent phantom. The center of the ion chamber is placed at the central axis. The Exradin W1 scintillator was placed in front of the ionization chamber and to its side outside of the active volume surface.

Figure 3. Percentage of plastic scintillation detector under-response that was attributable to ionization quenching in the phantom for each treatment field

Error bars represent the standard deviation of at least three repeated measurements.

Figure 4. ^{60}Co calibration results

A: Distribution of normalized baseline measurements. The distribution of the measurements was centered very close to zero and fell within a range of $\pm 2\%$. **B:** Weekly ^{60}Co baseline measurements of plastic scintillation detector dose response normalized to the average dose delivered across 22 weeks and corrected for source decay. All measurements fell within $\pm 1\%$. Error bars represent the standard deviation of repeated measurements.

Figure 5. Dose differences by field for each patient

Dose difference between the plastic scintillation detector and plane-parallel ionization chamber for each field (left and right lateral fields) by patient.

Figure 6.
Dose differences between the plastic scintillation detector and plane-parallel ionization chamber for all patient skin dose measurements acquired.

Table 1

Field parameters for each patient.

Patient	Field	Range in water (cm)	Field SSD (cm)	SOBP width (cm)	Skin dose per fraction (cGy)
1	1	24	250.6	8	67.3
	2	24.3	251.3	9	68.4
	3	23.3	252.6	9	68.4
	4	23.8	252.4	9	68.4
3	5	24.5	251.7	10	70.1
	6	24.9	251.4	9	68.2
4	7	23.9	251.6	7	63.9
	8	23.9	252.3	7	64.1
5	9	22.3	252.7	8	66.4
	10	22	253.1	8	66.5
6	11	22.8	252.7	8	66.5
	12	23.7	252.1	9	68.1