

## Current status of scintillation dosimetry for megavoltage beams

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**Abstract.** Plastic scintillation dosimeters (PSD) have been a topic of a keen research interest for the past 20 years. Numerous PSD systems have been proposed, most times differentiating from the previous by a slight change in one or more components, such as the photodetector. However, a few major technological and engineering innovations have also been made. Over the past few years PSDs have been evaluated for small field dosimetry, in vivo dose measurements, building of arrays and much more. The present manuscript is intended to present the basic physics and properties of PSDs and its application over the past two decades.

### 1. Introduction

Detection of ionizing radiation using scintillating materials is one of the oldest techniques available to scientists. Key discoveries, such as x-rays by Roentgen, were made using a light-emitting screen. Various processes can lead to the emission of visible light following the interaction and excitation by ionizing radiation. The fast emitting component, usually in the range of  $10^{-9}$  s, is called fluorescence. It constitutes the signal of interest. Other components such as phosphorescence or delayed fluorescence also produce visible light but at a much longer time scale. Therefore, efficient scintillation detectors for radiation detection usually make use of the fast emitting component. Also, two types of scintillation detectors are used on a regular basis and produce visible light through very different physical processes: organic scintillators and inorganic crystals. The reader is referred to the seminal books by Birk [1] or Knoll [2] for detailed reviews.

The applications of scintillators to medicine are numerous ranging from x-ray detection screens, gamma camera and PET scanners [3]. However, one of the first application to radiation dosimetry was reported in a manuscript entitled "A new method of integral dose measurement with a plastic scintillator phantom" by Murai *et al* in 1963 [4]. In that paper, a plastic scintillator was chosen because is constituted a good approximation of soft tissues and the visible light emitted was shown to be proportional to the absorbed dose. However, one must wait for the publication of the two seminal papers by Beddar *et al* to see any significant interest for plastic scintillation detector as radiation dosimeters[5, 6].



The current manuscript presents a review of plastic scintillation dosimeters (PSDs), their challenges and their applications to photon and electron dosimetry.

## 2. Scintillation detector for radiation dosimetry

### 2.1. Basic physics properties

When ionizing radiation interacts with an organic scintillator, the energy deposited by the electrons generated in the medium excites orbital electrons in higher energy levels. This excitation process is very fast ( $10^{-15}$  s). Through internal conversion and vibrational relaxation, electrons decay to the  $S_{10}$  state within the  $10^{-12}$  s. It is from the decay of  $S_{10}$  state to the  $S_{0x}$  that visible fluorescence light is emitted. The whole process is very fast and takes about  $10^{-9}$  s. The gap between the  $S_1$  and  $S_0$  states is usually between 3 to 4 eV thus accounting for the blue-violet light usually seen from most scintillators. Furthermore, this band gap is large enough to ensure some temperature independence (room temperature translating to an energy of about 0.025 eV). Finally, due to the differences in excitation and decay processes, the energy emitted as visible light produces an emission spectrum that is of lower energy than the excitation one. This is called the Stokes shift [2]. Thus organic scintillators are mostly transparent to their own emission.

It is important to note that a competing de-excitation mechanism is possible in which decays from the  $S_1$  states to the triplet  $T_1$  states occur followed by further decays from the  $T_1$  to the  $S_0$  states. This also results in visible light and is called delayed fluorescence and phosphorescence. This process is much less likely and is discarded in most dosimetry applications.

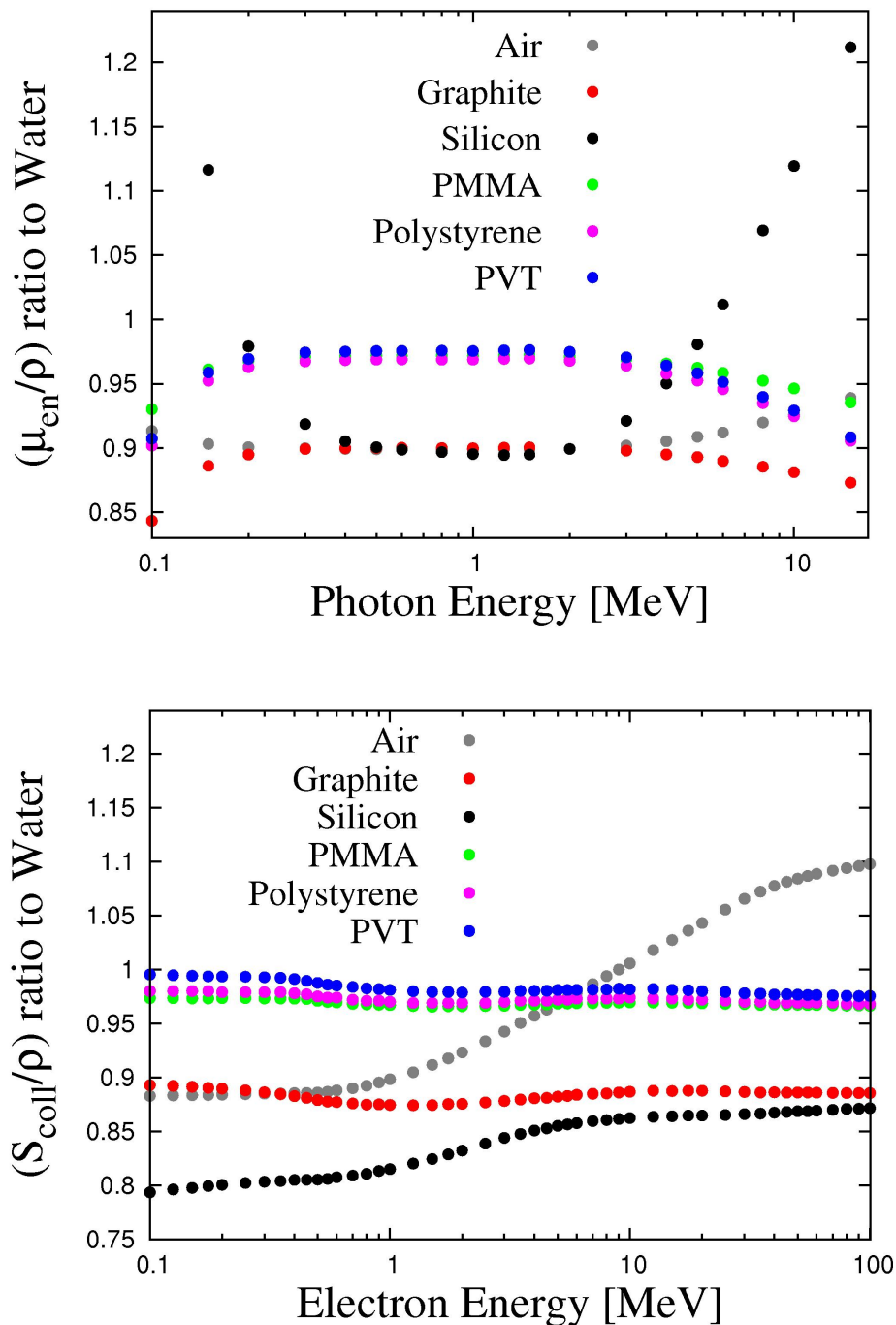
Plastic scintillators are composed of 97% of polyvinyltoluene (polystyrene in the case of plastic scintillating fibers) to which an organic fluor, the scintillating material, is added. These fluors are composed of aromatic chain with strong carbon double bonds. This is called a two component system. To obtain emission of the green or orange light, a third component called a wavelength shifter is added. In addition, scintillating fibers also possess a cladding that is usually made of PMMA, a material well known in radiation dosimetry.

Figure 1 shows some fundamental interaction properties of each of the materials described above, namely the mass energy absorption coefficients for photons and the stopping power for electrons. Both are normalized to water, the reference medium for radiation therapy dosimetry. Furthermore, relevant materials found in ion chambers and diodes are also shown. For materials composing scintillation detectors, it can be seen that for electrons, the stopping powers are almost independent of energy and within 2% of water over the full range covered by radiation therapy energies. Similarly, the mass energy absorption coefficients display a similar behavior for photons between 200 keV up to 3 MeV. This can be contrasted with silicon and even air. Thus from a basic interaction physics point of view, plastic scintillation detectors are water-equivalent and energy independent. This should translate to very little perturbation of the radiation fluence relative to the medium of reference (water); thus no correction factors should be needed for PSDs for most radiation therapy applications [7-9].

Finally, the physical density (1.032-1.06 g/cm<sup>3</sup>), electron density ( $3.238\text{--}3.272 \times 10^{23}$  e-/g) and elemental composition by weight forming the effective  $Z$  (7.7-8.5% H and 91.5-92.3% C) of the various components of a scintillator are all close to water.

### 2.2. Experimental validation

The various properties described above are all entrenched in a solid-state physics background. Yet, experimental validation under clinically realistic conditions is a critical step for adoption of PSDs. Beddar *et al* were the first to fully explore experimentally PSD properties for high-energy beam dosimetry. Namely in a series of two manuscripts, they have demonstrated the independence of PSDs to beam energy and dose rate, linearity of the light to dose, temperature independence, high spatial resolution and water-equivalence [5, 6]. Later on Archambault *et al* have shown that for small, point-like detectors PSDs also show no angular dependence if the Cerenkov effect is corrected for (to be discussed in details below) [10].



**Figure 1:** Mass energy absorption coefficients (upper panel) and electron mass stopping power (bottom panel) as a function of energy. Both are normalized to water and values were taken from the NIST website ([www.NIST.org](http://www.NIST.org)).

Validation of the same properties to clinical electron beams was detailed extensively by independent groups [11-13]. In particular Lacroix *et al* demonstrated that because PSDs measured directly the dose deposited by electrons without any correction factors, they can be used to extract ion chamber correction parameters [13].

While this is not the subject of this review, it is interesting to note that the measurements of absolute deposited dose for clinical proton beams with PSDs has not been thoroughly investigated at this time. Issue related to quenching effect are still being investigated in order to determine appropriate correction methods to correct for it [14].

### 2.3. Component of a PSD and design optimization

A PSD is made of three major building blocks: the scintillating probe (presented above), a light guide and a photodetector. This is called the optical chain. At all time, the linearity between the dose and light output depends not only of the scintillation properties but also of all component in the optical chain. Furthermore, each stage added in the path of the optical photons lead to a decrease of the light collection efficiency. First, the visible light produced in the scintillator must travel (through internal reflection) toward one of the exit face of the scintillator and be injected in the light guide (e.g. a clear optical fiber) (collection efficiency of about 5%). Interface between various components, e.g. scintillator to optical guide) is also a source a loss and the coupling efficiency is generally around 75-85% [15]. Light guide have intrinsic transport capability. Optical fibers, in particular the water-equivalent and flexible plastic types, are often used because of their enhanced light transport properties. Light attenuation in an optical fiber guide is generally less than 20% over a few meters. It is interesting to note that while most fiber-based PSDs use a cylindrical geometry, Yoo *et al* have shown that square fiber could be used as well [16]. The output of a light guide must be capture by a photodetector. Depending if the coupling to the photodetector is direct, passing through filters or a lens (or a combination thereof), the coupling efficiency can be as low as 5% and as high as 90%. Finally, the photodetector itself possesses an intrinsic efficiency (quantum efficiency), which can vary from 20% to 90%. The overall efficiency over the complete optical chain is thus only of a few percent. Therefore, optimization of each component is critical [15, 17, 18].

Various photodetectors have been used over the years. In the 1990s, Beddar *et al* pioneered the use of a photomultiplier tube, while a similar design but using a photodiode was first validated by Létourneau *et al* [19]. Because of its simplicity, this configuration has been adopted by multiple groups ever since [12, 20, 21]. The use of segmented RGB (red, green, blue) diode has been introduced in the early 2000s by Fontbonne *et al*, to implement a new Cerenkov subtraction feature [22]. RGB CCD camera has also been shown to be highly efficient for single point PSD and PSD arrays. Furthermore, high sensitivity electron multiply CCD was used by Archambault *et al* for fast (ms) in vivo dosimetry [1, 23]. Finally the used of PMT array was recently demonstrated by Liu *et al* [2, 24].

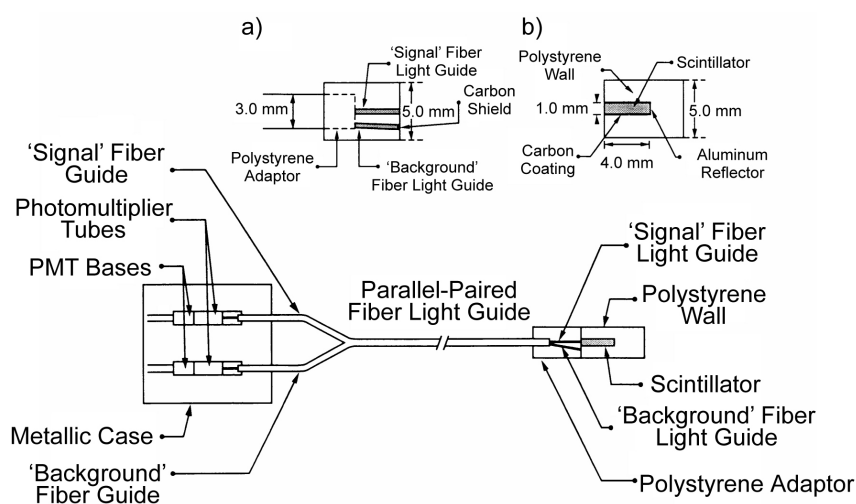
Overall the choice of plastic scintillator versus plastic scintillating fibers, cylindrical versus square fibers and of a particular photodetector should be dictated based on the specific application sought. In particular parameters such as light production (related to dose and dose rate), light collection, readout frequency and number of PSDs to read simultaneously (size of an array of PSDs) are all factors that need to be considered in the optimization of optical dosimeters, Lacroix *et al* have set-out a general framework enabling a quantitative evaluation [3, 25, 26]

### 3. The Cerenkov "problem"

Cerenkov is produced in a dielectric medium when electrons travel faster than light. This happen because the speed of light in the medium changes as  $c/n$ , where  $n$  is the index of refraction. For most plastics used in PSDs, the index  $n$  is between 1.48-1.60. This translates to electrons traveling faster than light in the medium for electron energy above 145-180 keV. Ongoing electrons polarize the medium with orbital electrons slightly perturbed; which results in dipole oscillations, then the relaxation process leads to the emission of visible light. Because that light travel slower than the exciting electrons, constructive interference occurs and the wave front will travel preferentially in the direction of the electrons [27]. Cerenkov light spectrum is independent of the medium and peaks in the violet region. All other wavelengths in the visible spectrum are also present but the intensity varies as  $\lambda^{-3}$ .

Cerenkov light emission per mm of medium is factor of 20 to 100 lower than a scintillator. However the clear optical light guides can be meters long, with many cm exposed to ionizing radiation [28, 29]. If exposure was constant (always the same amount of fiber), Cerenkov would also be a constant and easily taken into account during the light to dose calibration. However, in radiation therapy field sizes, beam orientations and points of measurement all lead to variation in the amount of medium exposed. Therefore, the unwanted Cerenkov must be eliminated. In 2000 [19, 30], a detailed study of six types of dosimeters in small field photon beams concluded the following: *“If the problem with the relatively high cable signal can be solved, the plastic scintillator offers high resolution, near water equivalence and small directional dependence”*.

The very first method proposed to eliminate this stem effect was the two fibers approach (figure 2). One clear fiber is used to collect the light emitted by the scintillator while a reference fiber is placed side-by-side but without the scintillator [5, 28]. Assuming that both fibers see the same electron energy fluence in all irradiation condition, subtracting the signal of the reference fiber from the PSD results only in the scintillation light. This design has been replicated by multiple authors, with small variations, through out the years [12, 19, 31, 32]. This method worked perfectly for in-fields measurements in situation where no dose gradient is seen differentially but the two fibers, however, suffered from spatial limitation when both fibers see a different dose gradient.



**Figure 2:** Diagram of the first prototype plastic scintillation detector, illustrating the two-fiber technique developed by Beddar *et al* from ref (5,6).

De Boer *et al* proposed another approach called optical filtering that takes advantage of the fact that the Cerenkov light is peaked in the violet region and that any scintillators emitting in a wavelength band slightly shifted will generate a signal that is much larger than the stem effect [33]. Thus an appropriate optical filter is selected, which removes wavelengths shorter than the scintillator emissions but let the higher wavelengths pass. This method was shown to remove about 50% of the stem signals over no removal [33]. Further optimization of this technique by Clift *et al* was able to reduce the stem effect by a factor of 2.4 [34]. The main advantage of this approach is to have a single probe and thus bypassing the limitation of the two-fibers method. However, because the Cerenkov light is present at all visible wavelengths, it is not as efficient.

A third approach, the temporal method, was proposed by Clift *et al* [35]. While plastic scintillators usually have very fast rise time (a few ns) and short decay constant (10-20 ns), this group uses a scintillator with a long decay constant (264 ns) and the fact that under pulsed beam, the stem effect shows a shorter decay time than the latter scintillator. Thus by timing the read-out of the scintillator relative to the beam pulse, essentially waiting for the stem effect to decayed-out after the beam pulse

and measuring the scintillation light afterward. They were able to show that 99.9% of the stem effect was removed and doses as low as 0.2 Gy/min could be measured [35]. This method takes advantage of using a single probe, like the optical filtering approach. On the other hand, it also needs complex timing set-up using information about beam pulse from the accelerator. This further makes clinical implementation difficult and possibly accelerator model dependent.

In 2002, Fontbonne *et al* introduced what would become a leading method for stem removal: the chromatic technique [22]. This solution capitalizes on the differences between the optical spectra of plastic scintillation light and Cerenkov light. By measuring the light output of a PSD under two different wavelength filters, it is possible to decouple each phenomenon (scintillation and Cerenkov) and isolate the portion of the signal that is only proportional to the dose delivered to the sensitive volume of the detector. This method has been shown to be very efficient while keeping the detection apparatus simple: filtering can be made by using external filters or even better on-chip RGB filter [10, 11, 22]. Archambault *et al* were the first to demonstrate in a systematic study that this method is superior to the approaches described above, in particular in the case of dose gradients (small fields, brachytherapy, beam penumbra) [10]. Recently, Guillot *et al* have shown the calibration procedure for this technique is critical to ensure optimal subtraction in all measurement conditions [36].

Finally, Lambert *et al* proposed in 2006 a different approach for getting rid of the stem effect by using a hollow light-guide instead of a plastic optical fiber. Because no Cerenkov light is produced in such light-guide, there is no problem of Cerenkov [20]. Uses of this technique for small field applications have shown promises [12]. On the limitation side [37, 38], the hollow core fiber are rigid compared to plastic fiber which limits its use for in vivo applications and light transport is less efficient, showing attenuation between 5 to 10 times higher than plastic core fiber.

## 4. Modern Application

### 4.1. Small Field Dosimetry

Accurate measurement of small radiation fields is a well-known challenge for many dosimeters [1]. Lack of lateral charge-particle equilibrium [5, 6, 41], dose-averaging within the sensitive volume of dosimeters [2, 42] and differences between the composition of detectors and their surrounding media [5, 6, 43] all cause perturbations of the radiation field. Thus, discrepancies are seen between the measured dose and the actual dose that would be deposited in the medium in the absence of a detector [10, 43-45]. Semiconductor diode detectors are often used as a reference dosimeter for small field measurements due, in part, to their extremely small sensitive volume [11-13, 46]. Even with diodes, Monte Carlo simulations may nevertheless be required to accurately determine the dose delivered by small fields [13, 43, 46, 47]. Combining high precision measurement with detailed Monte Carlo simulation is not a trivial task and cannot be performed routinely in most clinics.

The use of small fields is becoming ubiquitous in radiation therapy and may be a source of errors if measurements are not conducted properly. Discrepancies have been observed between treatment planning systems and measurement in Intensity modulated Radiation Therapy (IMRT) that can be attributed to the presence of small fields [14, 41, 42, 48]. Techniques such as radiosurgery and stereotactic body radiation therapy (SBRT) rely heavily on small fields. Furthermore, specialized equipment such as the Cyberknife and Gamma-knife are only relying on small fields for delivering their treatments. To address such non-standard irradiation schemes a work-group has been established to better define how reference measurement should be made in non-standard conditions [15, 49].

In addition to several types of diodes, the current arsenal used for measuring small radiation includes small-volume ion chambers, films (radiographic and radiochromic), diamond detectors, liquid ion chambers and thermoluminescent dosimeter (TLD). The use of plastic scintillation detectors (PSDs) for small field dose measurement has been suggested nearly 15 years ago [16, 19, 50]. PSDs are ideal for such measurement because they are water-equivalent and can be designed with small sensitive volume (i.e. PSDs can easily have dimensions smaller than a millimeter).

In recent years, PSDs have been increasingly used in radiosurgery and other small field techniques [12, 45, 51]. High precision/high accuracy measurements of MLC-shaped small fields [45] have shown output factors up to 15% higher than those measured with miniature ion chamber. This difference can be explained by the small size and water equivalence of PSDs. Thus proving that PSDs can be used to investigate possible discrepancies between TPS and measurements. To fully characterize the dose distribution produced by a given radiation beam, several profile and depth dose measurements are required. Such data acquisition can be simplified by the use of a detector array. However, given the high spatial resolution required for the characterization of a small field, most commercial arrays are inadequate for this task. Ion chamber arrays have individual detector widths of several millimeter and diode arrays cannot be packed much closer than 5 mm because of the surrounding electronics. PSDs do not suffer from such limitations, thus making PSD arrays an ideal tool for small-field characterization [16, 51].

Furthermore, because of their intrinsic water-equivalence, PSD arrays can be fully three-dimensional without the risk that detector element upstream affect the dose received by the detectors downstream. Films can also be used for 2D dose distribution measurements. However they are offline, integrating dosimeters while PSDs and other can be read in real-time [23], which provide useful information when delivering rotational treatments such as VMAT [52]. Another method for fully characterizing the dose distribution of a small radiation field could be to use the novel tomodosimetry concept [53]. Using tomodosimetry, it is possible to measure a 2D dose distribution with an arbitrarily high spatial resolution using only a small number of plastic scintillating fibers. With commercially available PSD and a clear demonstration of the potential of closely packed arrays [51] and tomodosimeters [53], it is likely that PSDs will become a reference for small field dosimetry.

## 5. In vivo dosimetry with PSDs

### 5.1. External beam radiotherapy

In addition to the dosimetric advantages PSDs have when compared to other detectors, they also can be easily incorporated with patient treatment applicators, inserted into Foley catheters or intraluminal interstitial needles as well as flexible catheters. The first study highlighting the use of PSDs for in vivo dosimetry and demonstrating their real-time monitoring capability of the dose rate delivery was demonstrated for external beam radiotherapy treatments by Archambault *et al* [23]. In that study, PSD systems designed into a modular detector patches, consisting of 5 closely packed sensors, were retrofitted to different rectal balloons and rectal inserts. Miniature PSD probes made out of green-emitting multi-clad scintillating fibers (0.5 mm diameter by 2mm length) were optically coupled to water-equivalent optical fibers were used to sample the dose reading using an electron multiplying (EM) CCD camera.

Doses of 200 cGy and 2cGy for all the PSDs were measured at a 0.15 s sampling rate and were accurate within 0.4% and 2.3% respectively. For an eight-beam intensity-modulated radiotherapy plan delivered to a prostate phantom with a rectal balloon inserted within it resulted in a very good agreement between the doses measured by the mounted PSDs and the treatment planning system. Overall, for all beams, the PSDs agreed with the treatment plan to within 0.5%. The main findings of the above study were: 1) a PSD multi-probe array could be read in real time with a CCD-based acquisition apparatus, 2) PSDs with high-spatial resolution (0.4 mm<sup>3</sup>) could be mounted on rectal balloons and used reliably for in vivo dose measurements, 3) the precision of the PSDs was 3% for measurements performed every 150 ms (1.5 cGy) and improved to 0.4% for a total dose of 200 cGy and 4) PSDs are capable of measuring dose with a high degree of accuracy.

### 5.2. Brachytherapy

The first PSD system built with the intention to be used for <sup>192</sup>Ir high dose rate (HDR) brachytherapy was developed by Arnfield *et al* in 1996 [54]. A plastic scintillator (BC408S, 1 mm diameter and 3

mm long) was attached to a 10 m silica fiber used a light guide and read by a photomultiplier tube. This system was never used for in vivo dosimetry however.

Also designed for HDR brachytherapy, a system has been described by Lambert *et al* (20) in 2006 and used clinically in a patient study in 2011[55]. The system called BrachyFOD is similar in design to the original of Beddar *et al* [5]. It uses a BC400 scintillator (0.5 mm diameter by 4 mm length) coupled to PMMA optical fiber guide (0.48 mm). The OPTIDOS Reader (PTW, Freiburg, Germany) composed of a photomultiplier and electrometer was used. 24 high-dose rate brachytherapy patients were enrolled in the study and dose to the urethra monitored. The PSD system underwent change after the first 14 patients in order to improve the positioning accuracy. A maximum dose departure of 9% from the expected values obtained by the treatment planning system was seen in the remaining 10 patients. This is inline with other brachytherapy in vivo studies using OSL [56] and Mosfet dosimeters [57, 58]. The authors of this study did not report seeing any temperature dependence for their in vivo measurement performed inside the urethra nor experiencing any stem effect when using these PSDs.

Another real-time in vivo PSDs for  $^{192}\text{Ir}$  HDR brachytherapy was developed by Therriault-Proulx *et al* [59] and validated in a phantom study [60]. The importance of removing stem effects for in vivo dosimetry using PSD in HDR brachytherapy treatments was assessed and was found to be required in order to be able to measure doses accurately at this energy range [59]. Therefore, a PSD system capable of stem removal was built using a tricolor (red-blue-green) photodiode connected to a dual-channel electrometer. The PSD system consisted of a 1.0 mm diameter x 3.0 mm long green scintillating fiber (BCF-60) coupled to a light-shielded 1 mm diameter 7 m long polymethylmethacrylate optical fiber (Eska Premier GH-4001). A clinically realistic Ir-192 HDR brachytherapy treatment plan was simulated by loading computed tomography images of an arbitrarily chosen patient with prostate cancer into the Oncentra version 3.2 Nucletron Treatment Planning System (TPS). The target volume (prostate) and surrounding organs at risk (i.e., bladder, urethra, and rectum) were contoured, and catheters were reconstructed in the software to reproduce the experimental setup. Simulations of treatment delivery were performed in a customized water phantom, which consisted of eleven catheters that were utilized as channels for dose delivery from the Ir-192 radioactive source, while two others were used to mimic dosimetry at the rectum wall and in the urethra using the above described PSD. The measured dose and dose rate data were compared to the expected values from the planning system. In-phantom dosimetry measurements of the treatment plan led to a ratio to the expected dose of  $1.003 \pm 0.004$  with the PSD at different positions in the urethra and  $1.043 \pm 0.003$  with the PSD inserted in the rectum. The verification of the dose delivered to the urethra within each catheter and at specific dwell positions led to average measured to expected ratios of  $1.015 \pm 0.019$  and  $1.014 \pm 0.020$ , respectively. These values at the rectum wall were  $1.059 \pm 0.045$  within each catheter and  $1.025 \pm 0.028$  for specific dwell positions [60]. The study also demonstrated that the above-described system has the ability to detect positioning errors of the source and also allows for temporal verification of the treatment delivery.

### 5.3. Superficial therapy, radiology and interventional radiology

For low energy ionizing radiation, PSDs are not expected to retain their water-equivalence (see Fig 1, bottom panel). Furthermore, effect such as quenching can become important, displaying non-linearity between the deposited energy and light production [61, 62]. Jones and Hintenlang demonstrated experimentally that PSDs show a strong dependence to tube voltage between 40 and 120 kVp but for a given voltage they show linearity between light output and air kerma and a reproducibility of about 6% [63]. Similar findings were made by Hyer *et al* in the context of dose measurements for CT scanner [64]. More recently, Lessard *et al* performed a systematic studies combining PSD measurements, cavity theory and Monte Carlo simulations to elucidate the energy dependence and possible quenching effect in this energy range [65]. A simple correction approach, using calculated x-ray tube spectra, resulted in absolute dose measurements within 2% of the expected values [65]. It is therefore quite likely that we will see more work in this area in the coming years.



## 6. 2D scintillation dosimetry

The water-equivalence property is of primary importance for applications beyond single-point dosimetry: it implies that the presence of a scintillating fiber in a water-equivalent phantom does not alter the original dose distribution in the phantom (i.e. the dose distribution in the absence of the detector).

This property makes scintillating fibers ideal to use in an array of detectors. The first PSD array encountered in the literature was made by Flühs *et al* for ophthalmic plaque dose measurements [31]. It uses a 1D array composed of 16 PSDs and uses the two-fiber method for handling the stem effect. Furthermore, this 1D array was mounted on a vertically translating stage also capable of translation. This gave the capability to map the dose in 3D space. Numerous investigators have developed experimental devices in both linear (1D) and planar (2D) configuration [16, 24, 51, 53, 66-71]. Among these, the highest density of detector was achieved by Gagnon *et al.* and Yoo *et al.*, who achieve near 1 mm detector spacing [16, 51]. Their detectors were composed of two, crossed linear array of small length (< 5cm) and were developed for small field dosimetry. The highest number of detector in a single array was achieved by Guillot *et al*, with almost 800 scintillating fibers placed inside a 26×26 cm<sup>2</sup> detection plane [69, 71]. The resolution of this array was 10×10 mm<sup>2</sup> (except the two central profiles at 5 mm spacing) and was developed for IMRT and VMAT plan verification. As of today, no 3D experimental configuration of scintillating fiber arrays has been reported in the literature, although such arrangement would definitively be possible because of the water-equivalence of these detectors. However Kirov *et al* used liquid scintillator to characterize the dose distribution around brachytherapy sources [72] and Poenisch *et al* for high-energy photon beams [73].

Other investigators have opted for scintillator sheets instead of fibers to perform 2D dosimetry [74-76]. The scintillator sheets used ranged from 15×15 cm<sup>2</sup> to 25×25 cm<sup>2</sup> and were imaged with the combination of a CCD camera and a 45 degrees mirror. This methodology has the advantage of measuring 2D dose distribution in real-time with high resolution and without the discrete (i.e. incomplete) sampling performed by 2D arrays of scintillating fibers. 3D dosimetry has also been reported as possible using a vertical translation of the scintillator sheet [75].

Investigators have also looked for application of scintillating fibers in longer length (i.e. more than a few millimeters, as used in scintillating fiber arrays previously described) [68, 77]. Goulet *et al* developed a transmission detectors composed of 60 parallel, 27 cm long scintillating fibers for the on-line monitoring of MLC-driven radiotherapy treatment [68]. Collecting the scintillation light from both sides of each fiber and taking advantage of the important optical attenuation inside scintillating fibers, they were able to extract both the position and the fluence related parameters for each field under monitoring. These two independent parameters could then be evaluated and compared in real-time to their expected value (either theoretically or experimentally deduced). Using a similar fiber configuration, these same authors have developed a high resolution 2D dosimeter using tomographic principles [77]. Using a rotating array of 50 parallel fibers inside a water-equivalent phantom, they were able to measure the dose inside a 10 cm radius circle of detection with high spatial resolution (1×1 mm<sup>2</sup>). This kind of methodology has the advantage of limiting the number of scintillating fibers needed to perform high-resolution dosimetry, and should find interesting applications in 3D dosimetry.

## 7. Conclusion

PSDs have gain in technological innovation and engineering refinement over the years. The major issue of the stem effect (Cerenkov and fluorescence) has been addressed and example clinical use shown in this documents. PSDs are poised to become reference dosimeters for small field dosimetry. Early results shows that PSDs can be used without need of correction for fields of 4-5 mm, such as encountered on the Cyberknife (Accuray, California) [78, 79]. New major developments are also expected based on the recent conference presentations demonstrating the possibility of having multiple scintillation probes on a single light guide [80, 81]. Finally, a point-like PSD system is now available commercially, enabling widespread clinical adoption [82].

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