Optimization of a multipoint plastic scintillator dosimeter for high dose rate brachytherapy

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Purpose: This study is devoted to optimize and characterize the response of a mPSD for applications to in vivo dosimetry in HDR brachytherapy.

Methods: An exhaustive analysis was carried out in order to obtain an optimized mPSD design that maximize the scintillation light collection produced by the interaction of ionizing photons. More than 20 prototypes of mPSD were built and tested in order to determine the appropriate order of scintillators relative to the photodetector (distal, center, or proximal), as well as their length as a function of the scintillation light emitted. The available detecting elements are the BCF-60, BCF-12 and BCF-10 scintillators (Saint Gobain Crystals, Hiram, OH, USA), separated from each other by segments of Eska GH-4001 clear optical fibers (Mitsubishi Rayon Co., Ltd., Tokyo, Japan). The contribution of each scintillator to the total spectrum was determined by irradiations in the low energy range. For the best mPSD design, a numerical optimization was done in order to select the optical components (dichroic mirrors, filters and photomultipliers tubes (PMTs)) that better match the light emission profile. Calculations were performed taking into account the measured scintillation spectrum and light yield, the manufacturer-reported transmission and attenuation of the optical components, and the experimentally characterized PMT noise. The optimized dosimetric system was used for HDR brachytherapy measurements. The system was independently controlled from the $^{192}$Ir source via LabView and read simultaneously using NI-DAQ board. Dose measurements in terms of distance to the source were carried out according to TG43 recommendations. The system performance was quantified in term of signal to noise ratio (SNR) and signal to background ratio (SBR).

Results: For best overall light-yield emission, it was determined that BCF-60 should be placed at the distal position, BCF-12 in the center and BCF-10 at proximal position with respect to the photodetector. This configuration allowed for optimized light transmission through the collecting fiber and avoided inter-scintillator excitation and self-absorption effects. The optimal scintillator length found was of 3,6, and 7 mm for BCF-10, BCF-12 and BCF-60, respectively. The optimized luminescence system allowed for signal deconvolution using a multispectral approach, extracting the dose to each element while taking into account Cerenkov stem effect. Differences between the mPSD measurements and TG-43 remain below 5% in the range of 0.5 to 6.5 cm from the source. In all measurement conditions, the system was able to properly differentiate the produced scintillation signal from the background one: SNR was found to be above 5 for all dose rates while the minimum SBR measured was 1.8 for a distance of 6.5 cm from the source.

Conclusion: Based on the spectral response at different conditions, a mPSD was constructed and optimized for HDR brachytherapy dosimetry. It is sensitive enough to allow multiple simultaneous measurements over a clinically useful range, up to 6.5 cm from the source. This study constitutes a baseline for future applications enabling real time dose measurements and source position reporting over a wide range of dose rate conditions.

Keywords: in vivo dosimetry, plastic scintillator, multi points plastic scintillation detector, HDR brachytherapy
I. INTRODUCTION

In brachytherapy, radioactive sources are placed a short distance from the target. The high dose gradients near brachytherapy sources (10% or more per millimeter for the first few centimeters from the source) provide a level of protection to healthy tissues surrounding the target. Despite the short distances involved in this modality, brachytherapy is not free from errors, which can be caused by humans (e.g., incorrect medical indication, source strength, patient identification, catheter, or applicator) or by failures in the treatment system (e.g., mechanical events). Even small errors in the source positioning can result in harmful consequences for patients. Systematic implementation of precise quality control and quality assurance protocols helps to improve treatment quality, and routine use of real-time verification systems and in vivo dosimetry are even more helpful in determining whether there are deviations from the prescribed dose during treatment delivery. Performing these tasks requires a precise and accurate detector whose presence does not perturb the particle fluence and the physics interactions. Tanderup et al.\textsuperscript{2} reviewed different detectors for potential use for in vivo brachytherapy dosimetry. The selection of the appropriate dosimetric system is a compromise between different requirements and constraints of the detector as well as the application sought.

Plastic scintillator detectors (PSDs) are a promising detector for obtaining accurate real-time radiotherapy dose measurements. Previous studies have demonstrated that PSDs can accurately measure dose in external beam radiotherapy and that they have high spatial resolution, linearity with dose, energy independence in the megavolt energy range, and water equivalence\textsuperscript{3–12}. In addition, some authors have found PSDs to be feasible for use in brachytherapy applications\textsuperscript{13–17}. Despite the aforementioned advantages, PSDs response is affected by the stem effect and temperature dependence. Temperature dependence was long considered to be negligible, but recent studies showed that, depending on the type of scintillator, changes on the order of 0.6% per degree Celsius should be expected\textsuperscript{9,18}. Moreover, a large fraction of the light collected by PSDs consists of Cerenkov radiation, which is especially undesirable in high dose rate (HDR) brachytherapy. Cerenkov radiation can cause large errors in dose reporting; therefore, numerous investigations have developed methods to remove Cerenkov radiation from the scintillation light collection procedure\textsuperscript{6,11,12,19–22}.

Most studies that have characterized PSDs have been conducted using an optical fiber.
connected to a single point of measurement as the sensitive volume. Multiple scintillation sensors attached to a single optical chain have been used, but their application is limited to measurements made within 3 cm of an HDR brachytherapy source\textsuperscript{22–24}. New multipoint PSDs (mPSDs) could assess the dose at multiple points simultaneously, thereby improving treatment quality and accuracy. The multi-hyperspectral filtering method proposed by Archambault et al.\textsuperscript{22} led to the conception of an mPSD in which each scintillator has an independent signal. Such an arrangement would allow simultaneous determination of the absorbed dose at different locations in a volume\textsuperscript{23–25}.

The purpose of this study was to evaluate the performance of an mPSD in terms of sensitivity and accuracy, resulting in a thorough optimization of the optical chain, with applications in in vivo HDR brachytherapy in mind. To achieve this goal, 3 steps were followed. First, the base system was experimentally characterized. Second, the mPSDs system was numerically optimized to maximize the mPSDs performance. Finally, it was characterized the optimized system.

### II. MATERIALS AND METHODS

#### II.A. Experimental characterization of the base system

**II.A.1. Experimental set-up**

The optical chain in the proposed system has components that (1) generate scintillation light in the mPSD, (2) detect the scintillation light, and (3) analyze the signal.

The scintillating fibers used in this study were the plastic scintillators BCF-10, BCF-12,
and BCF-60 from Saint Gobain Crystals (Hiram, OH, USA). Figure 1 shows the design schematic of a typical mPSD. The scintillators were separated from each other by 1 cm of clear optical fiber (Eska GH-4001, Mitsubishi Rayon Co., Ltd., Tokyo, Japan). The same type of fiber was also used to conduct the scintillating light to the photodetector surface. (Here, clear refers to a fiber in which no scintillation light is produced.) Figure 1 shows an mPSD with 2 scintillators, but both 2-point and 3-point mPSDs were explored in this study. All optical interfaces (scintillators and clear optical fibers) were polished using a SpecPro automated optical fiber polisher (Krell Technologies, Neptune City, NJ, USA) with successive grain sizes of 30 µm, 9 µm, 3 µm, and 0.3 µm. The detectors were constructed using a previously described coupling technique. It was verified that the polishing and coupling techniques were reproducible within margins of less than 5%. Each 1-mm-diameter detector prototype was made light-tight using a black polyether block amide jacket from Vention Medical (Salem, NH, USA). The scintillating tip was sealed with a mixture of epoxy and black acrylic paint.

The scintillation light signal was guided to the photodetector surface by a clear optical fiber, which was attached to the photodetector with a subminiature version A connector (11040A, Thorslab, Newton, NJ, USA). Two types of photodetector were used: (1) an Ocean Optics QE65Pro spectrometer (Dunedin, FL, USA) and (2) a set of photomultiplier tubes (PMTs) coupled to dichroic mirrors and filters from Hamamatsu (Bridgewater, NJ, USA). When the PMTs were used as the photodetector, the signal was read by a data acquisition board (NI USB-6289 M Series Multifunction I/O Device, National Instruments, Austin, TX, USA), and LabVIEW software version 15.0f2 (National Instruments) was used during the signal analysis stage.

II.A.2. Irradiation measurements

This study was designed to cover energy ranges of clinical interest. Several measurements were carried out in the low-energy range using an Xstrahl 200 X-ray therapy system (Xstrahl Ltd., Camberley, UK) to avoid Cerenkov light emission. Continuous beam irradiations with a tube current of 10 mA were performed according to the specifications shown in Table I. Three beam energies were used during these irradiations.

The mPSD response was then assessed for HDR brachytherapy. A Flexitron HDR after-
Table I  Main irradiation parameters when using a Xstrahl 200 x-ray therapy system.

| Energy (kV) | HVL1 (mm) | Added filtration | Field size (cm) |
|-------------|-----------|------------------|-----------------|
| 120         | 5.0 Al    | 0.5 Al+0.10 Cu   | 10*             |
| 180         | 0.5 Cu    | 1.5 Al+0.15 Cu   | 10*             |
| 220         | 2.0 Cu    | 1.0 Al+0.25 Cu+0.45 Sn | 10*         |

*HVL = Half-Value Layer
*diameter to circular shape

loader (Elekta, Stockholm, Sweden) was used with an $^{192}$Ir source. The cylindrical $^{192}$Ir source pellet was 0.6 mm in diameter and 3.5 mm in length and was housed inside a stainless steel capsule 0.86 mm in diameter and 4.6 mm in length. The source activity was 7.8 Ci and the air kerma strength was 32226 cGy·cm$^2$·h$^{-1}$.

II.A.3. mPSD design optimization

The purpose of this step was to identify the best configuration of scintillators in an mPSD. Interchangeable radiosensitive tips containing the scintillators were coupled to a clear optical fiber, which conveyed the light signal to the photodetector. We built and tested more than 20 prototype 2-point and 3-point mPSDs using various combinations of scintillator positions and lengths to determine (1) the optimal position for each scintillator within the fiber and (2) the optimal length of each scintillator.

To evaluate the effect of scintillator position on the recorded signal, the lengths of the scintillators inside the radiosensitive tips were kept identical, but the positions of the scintillators inside the fiber were altered. Each scintillator was placed in a distal, central, or proximal position relative to the photodetector surface. To determine the optimal scintillator length in terms of the scintillation light produced and spatial resolution, each scintillator was varied in length from 3 to 14 mm. The hyperspectral approach$^{22}$ dictated a balanced signal contribution from each scintillator to the overall light collection, while the spatial resolution of the dose required as small as possible detection elements. For each possible combination of scintillator number, position, and length, the total emission spectrum was
obtained from simultaneous irradiation of all the scintillators in the fiber. Then, in order to obtain their individual contributions to the total emission spectrum, each scintillator was individually irradiated. Lead blocks were used to shield the neighboring scintillators from the incoming radiation. This analysis was performed under a fixed irradiation condition. An energy of 120 kV was selected to avoid Cerenkov light production. The spectrometer was cooled to -20 °C, and the integration time was 40 s. Background signals acquired prior to exposures were subtracted from the scintillation light.

II.A.4. Cerenkov radiation removal

The stem effect, which can have 2 sources, is always present in PSDs. The first source is direct excitation of the polymer chain, or fluorescence, from the plastic optical fiber guides, and the second is Cerenkov light production. Therriault-Proulx et al.\textsuperscript{29} evaluated the nature of the stem effect light produced in an optical fiber for clinical energy ranges. They found that the fluorescence component produced in the optical fiber was present over the entire range of clinically relevant irradiation energies (from kV to MV), even where Cerenkov light emission is possible; however, the light produced by fluorescence is orders of magnitude less intense than the Cerenkov light\textsuperscript{29,30}. The production and removal of Cerenkov light in PSDs are widely discussed topics\textsuperscript{30}. The intensity of Cerenkov light emitted in the optical fiber guide is up to 2 orders of magnitude lower per millimeter than the intensity of the scintillation light produced by the scintillator. However, the optical fiber guides within the radiation field are usually much longer than the scintillation probes several centimeters versus a few millimeters at most for the scintillator\textsuperscript{30}.

Whether Cerenkov light requires removal in brachytherapy depends on the radioactive source used and the measurement geometry. Low dose rate brachytherapy sources like $^{125}$I or $^{103}$Pd do not have an energy high enough to produce Cerenkov radiation in a plastic optical fiber. The context is completely different for the $^{192}$Ir sources used in HDR brachytherapy. Therriault-Proulx et al.\textsuperscript{14,15,19} showed that when an HDR brachytherapy source is within 10 mm of the optical fiber and more than 25 mm from the scintillator, Cerenkov removal is necessary. Their study in a water phantom demonstrated that an error on the order of 25% could be expected if stem effect removal is not performed. They also showed that in conditions where the source is placed close to the scintillator and far from the optical fiber,
the stem effect is negligible. The further the source moves away from the scintillator and closer to the optical fiber, the more important the stem effect becomes. In our study, multiple probes were read by a single clear collecting optical fiber; thus, it was used a single removal method, the hyperspectral filtering technique proposed by Archambault et al.\textsuperscript{22}

\[
m = Rx
\]

\[
\begin{pmatrix}
  d_{p1,1} & d_{p1,2} & \cdots & d_{p1,N+1} \\
  d_{p2,1} & d_{p2,2} & \cdots & d_{p2,N+1} \\
  \vdots & \vdots & \ddots & \vdots \\
  d_{pN,1} & d_{pN,2} & \cdots & d_{pN,N+1}
\end{pmatrix}
= 
\begin{pmatrix}
  X_{p1,L_1} & X_{p1,L_2} & \cdots & X_{p1,L_{N+1}} \\
  X_{p2,L_1} & X_{p2,L_2} & \cdots & X_{p2,L_{N+1}} \\
  \vdots & \vdots & \ddots & \vdots \\
  X_{pN,L_1} & X_{pN,L_2} & \cdots & X_{pN,L_{N+1}}
\end{pmatrix}
\times 
\begin{pmatrix}
  m_{L_1,1} & m_{L_1,2} & \cdots & m_{L_1,N+1} \\
  m_{L_2,1} & m_{L_2,2} & \cdots & m_{L_2,N+1} \\
  \vdots & \vdots & \ddots & \vdots \\
  m_{L_{N+1},1} & m_{L_{N+1},2} & \cdots & m_{L_{N+1},N+1}
\end{pmatrix}
\]

\[
D = X \times M
\]

Dose calculation in mPSDs is based on the assumption that the recorded signal results from the linear superposition of spectra; no self-absorption interactions among the scintillators composing the mPSD are considered. The idea behind this formalism is that once the light emission of each component at different wavebands is known, the total signal recorded can be decoupled, and the signal fraction contributed by each scintillator can be determined.

In equation (1), \(m\) is a vector of \(L\) elements, \(R\) is the response matrix of dimensions, and \(x\) is a vector of \(N\) elements representing the photon flux (the number of photons emitted for a given emission source, either scintillating elements or any other source of light). \(L\) represents different wavelength filters or channels. The number of measurement channels \(L\) should be equal to \(N + 1\). The additional channel is included to take into account the stem effect, which should be removed from the measured signal.\textsuperscript{14}

The dose \(d_{i,k}\) received by the scintillator during irradiation is directly proportional to the number of scintillation photons in the absence of losses (quenching); for this reason, \(d_{i,k} = a_i x_{i,k}\), \(a_i\) being a proportionality constant and \(x_{i,k}\) the photon fluence in the scintillating material \(i\) during the measurement \(k\). However, knowing the dose at a specific point requires a previous calibration to determine the calibration factor \(X\) for each scintillation point as well as each measurement channel. For such a calibration the dose (e.g., \(d_{p1,p2,\ldots,pN}\)) should be known at each point. We calculated these dose values by using the TG-43 formalism.\textsuperscript{31}

To account for the finite size of each scintillator, TG-43 dose values were integrated over each scintillators sensitive volume. Therefore, equation 2 is the general mathematical
equation for determination of the calibration factor. Once the calibration factor $X$ is known, the dose $D$ at each point can be determined using equation 3, where $M$ represents the raw data acquired during measurements. The capital $M$ is used to highlight that this is a new set of measurements whose goal is to determine the absorbed dose, not the calibration factor, which is already known at this stage.

II.B. Optimization of the light collection system

In the second step, we sought the appropriate optical components for the light detection system used during measurements. Having a complete spectral characterization led us to perform a numerical analysis to determine the optical chain that would allow optimal scintillation light collection.

In our experimental set-up, scintillation light was read by an assembly of PMTs, which were coupled to a set of dichroic mirrors and filters that deconvolved the collected light into spectral bands. PMTs were chosen as the photodetectors because they have a high signal-to-noise ratio (SNR) and readout speed that overcomes many of the sensitivity issues of charge-coupled device-based systems. Generally, PMTs more accurately measure low light signals and have a faster response, making them more suitable for the demands of in vivo dosimetry applications. Henceforth, an assembly composed of a dichroic mirror, filter, and PMT will be referred to as a channel. From an optimization perspective, the signal produced in each channel was calculated, taking into account the measured scintillation spectrum and light yield, the manufacturer-reported transmission and attenuation of the optical components, and the experimentally characterized PMT noise. The experimental spectral characterization obtained for the mPSD constituted the main input. That spectral information was then used to construct the optical system and simulate its response when interacting with a radiation beam. No particle transport was performed. A large set of possible component combinations (brute force) was explored to find the configuration that provided the best SNR. For the calculations, we used the characteristics of filters and dichroic mirrors from Hamamatsu series A10033 and A10034, respectively. For the PMTs, the models used corresponded to Hamamatsu series H10722. The numerical optimization took into account the fact that the number of channels depended on the number of scintillator points $N$ composing the mPSD and equaled $N + 1$. This procedure allowed us to optimize light
transmission and to minimize the contribution of elements generating spurious light (as will be shown in Figure 7).

II.C. Performance of the optimized light collection system in HDR brachytherapy

With the optimized system, we next evaluated the performance of a 3-point mPSD in HDR brachytherapy. The HDR brachytherapy unit was remotely controlled and able to move the source to a desired position in a water tank by means of a 30-cm needle set from Best Medical International (Springfield, VA, USA). The mPSDs dimensions allowed it to be inserted into an additional catheter for use during real-time dose verification.

To be consistent with the TG-43 formalism\textsuperscript{31}, measurements were performed with the source and detector covered by at least 20 cm of water isotropically. To ensure the accuracy and reproducibility of the source-to-detector distance, all the catheters were inserted in a custom-made poly(methyl methacrylate) phantom (Figure 2a), which was in turn placed inside a $40 \times 40 \times 40$-cm$^3$ water tank. As shown in Figure 2, the phantom was composed of 2 catheters with insertion templates of $12 \times 12$ cm$^2$, separated by 20 cm. This phantom allowed for source-to-detector parallel displacement. Figure 2b shows the experimental set-up used during measurements. The source dwelled at 30 positions ranging from 0.5 to 7.0 cm away from the mPSD. Following the axis convention shown in figure 2a, the source dwell positions were chosen to ensure source displacement in the $x$ and $y$ directions but with the scintillator volumes center placed at the same $z$ location.

The dosimetric system was initially calibrated under the same conditions used to perform the measurements, following the TG-43\textsuperscript{31} and hyperspectral\textsuperscript{22} formalisms. The luminescence dosimetry system was controlled using LabVIEW software. A gain input voltage of 1 V was assigned to each PMT, producing a channel output of $2 \times 10^6$. The gain linearity with voltage was assessed by producing variations in the input voltage from 0.5 to 1.1 V. For each measurement channel, 70000 samples per second were acquired. Dose values were recorded in real time by the mPSD. All measurements were repeated at least 5 times, and the measurement set-up was completely unmounted between each measurement. Statistical variations in the readings were determined by taking 60 readings at each depth at intervals of 1 s.
Fig. 2  (a) Schematic of the poly(methyl methacrylate) (PMMA) phantom constructed for HDR brachytherapy measurements with an mPSD. The catheter positioning allowed source displacement parallel to the mPSD. (b) Experimental set-up for HDR brachytherapy measurements. (1) PMMA phantom, (2) mPSD, (3) $^{192}\text{Ir}$ source, (4) 30-cm catheters, (5) Flexitron HDR afterloader unit, (6) $40 \times 40 \times 40$-cm$^3$ water tank, (7) solid-water slabs.

Fig. 3  Typical signal pulse used for signal-to-noise ratio and signal-to-background ratio calculations. The indicated values are: $\mu_s$, mean signal value; $\mu_b$, mean background value; $\sigma_s$, signal standard deviation; $\sigma_b$, background standard deviation and the difference between $\mu_s$ and $\mu_b$.

To quantify the sensitivity of the dosimetry system, it was evaluated the SNR and signal-to-background ratio (SBR) associated with each scintillator during HDR brachytherapy measurements. Figure 3 is a representation of a typical signal pulse, showing the magnitudes that were used in SNR and SBR determination: $\mu_s$, the mean signal; $\mu_b$, the mean
background signal; $\sigma_s$ the signal standard deviation; and $\sigma_b$ the background standard deviation. 

$$SNR = \frac{\mu_s}{\sigma_s}$$ (4)

$$SBR = \frac{\mu_s}{\mu_b}$$ (5)

The SNR is a commonly used metric for characterizing the global performance of optoelectronic systems. In case of PSD performance assessment, the noise term includes Cerenkov radiation. A few SNR studies using PMTs as the photodetector have been performed\textsuperscript{32-34}. The scintillator SNR as function of dose rate was obtained using equation 4 where the numerator represents the mean signal for a determined irradiation in a fixed time, and the denominator is the standard deviation in a homogeneous background region. SBR was determined according to equation 5 and is the proportion between the mean signal value and the mean background value for a fixed irradiation time.

III. RESULTS AND DISCUSSION

III.A. mPSD signal proportion and optimal configuration

The initial investigations were performed with a 2-point mPSD configuration. Each scintillator spectrum was measured independently. Figure 4 graphs the individual spectra with intensities normalized to a 1-mm scintillator length. As shown in the figure, the scintillation intensity was strongest in the BCF-10 scintillator, whereas the BCF-60 scintillator had the weakest scintillation intensity.

In all the tested combinations, the scintillators individual spectra evidenced no self-absorption or cross-excitation effects. The main differences observed related to the position occupied by each scintillator inside the fiber. Table II shows the measured signal proportion for different 2-point mPSD configurations. For each configuration, the proportion of the total signal coming from each independent scintillator was calculated by determining the area under the curve. The signal proportion was more balanced when a BCF-60 scintillator was placed in the distal position and coupled to a BCF-10 scintillator. At 530 nm, the BCF-60 signal represented 37% of the total signal when coupled to a BCF-10 scintillator but only
Fig. 4 Individual scintillator spectra normalized to a 1-mm-long scintillator.

Table II Scintillator signal proportion (in %) for each combination of scintillators and positions inside the optical fiber.

| Distal  | BCF-10 | BCF-12 | BCF-60 |
|---------|--------|--------|--------|
| BCF-10  | -      | 8.6/91.4 | 17.6/82.4 |
| BCF-12  | 18.8/81.2 | -      | 7.2/93.8  |
| BCF-60  | 37.4/62.6 | 8.8/91.2 | -      |

9% when coupled to a BCF-12 scintillator. Therefore, this combination of scintillators and positions is recommended for a 2-point mPSD.

The signal analysis demonstrated that the lower-wavelength scintillator should always be placed closer to the photodetector and the less-energetic scintillator in the distal position. Because of the Stokes shift, the absorption spectrum always has a lower wavelength range than the emission spectrum. If the aforementioned configuration is not used, inter-scintillator excitation and self-absorption effects take place, and as a consequence, the light transmission through the collecting fiber is not optimal. To exemplify this effect, 3-point mPSDs with 2 different configurations of scintillator positions inside the fiber were constructed. Their spectral distributions are shown in Figure 5. Figure 5a shows the spectra with the BCF-10 placed at the distal position, the BCF-60 in the center, and the BCF-12 in the proximal position. Almost all of the light produced by the BCF-10 scintillator was absorbed by the
neighboring scintillators, whose photon intensities were higher than they were in the optimal configuration (Figure 5b). Hence, in the subsequent experiments, we used mPSDs in which the scintillators were placed inside the optical fiber in increasing order of energy from distal to proximal positions.

In the case of a 3-point mPSD composed of 3 mm of BCF-60 at the distal position, BCF-12 in the center, and BCF-10 at the proximal position, we observed no self-absorption or cross-excitation effects, but the scintillators independent signals were not balanced at all. As shown previously, the scintillation process is more efficient in BCF-10 than in the other scintillators, accounting for almost 80% of the total signal. The intensities of BCF-12 and BCF-60 were closer to one another, with 17% and 10% of the total signal, respectively.

### III.B. Optimal scintillator length

The amount of measured scintillation light depends on the scintillator size, the coupling method, the fiber core size, and the numerical aperture. To determine the optimal length of each scintillator, 9 different 3-point mPSD prototypes were constructed. Figure 6 shows the contributions of individual scintillator signals for each of the 3-point mPSDs and specifies the length of each scintillator in millimeters. As indicated by the shadowed region in Figure 6, detector configurations P6 to P9 provided the required balanced signals for optimal hyperspectral deconvolution. P9 was selected as the optimal detector because it also minimized
Fig. 6 Fraction of the total scintillation light produced by each scintillator as a function of its length. The shaded region indicates the combination of sensor length that result in balanced signals.

There were variations in sensor length. In the P9 mPSD, the BCF-10 scintillator was 3 mm long, the BCF-12 scintillator was 6 mm long, and the BCF-60 scintillator was 7 mm long.

III.C. Optimized light collection system

Following the determination of the optimal length, the numerical optimization allowed us to determine the best combination of components to be used for the measurements of the light collection system. Figure 7 shows a schematic of the appropriate arrangement of the components of the light collection system obtained from these calculations.

This assembly filtered the total emission spectrum from the mPSD to produce a filtered spectrum entering each PMT. Figure 8 shows each channel’s filtered spectrum and the total emission spectrum of the scintillation light generated by the P9 mPSD. Each measurement channel produced a voltage value that, when converted into dose, was proportional to the scintillation light generated by each scintillator without the Cerenkov radiation. A signal was generated each second, so no signal integration in time was required.

The study done by Therriault-Proulx et al. in HDR brachytherapy, uses the same scintillating elements described in this work (BCF-10, BCF-12, and BCF-60; Saint-Gobain
Fig. 7  Schematic of the light collection system obtained from calculations. All the components used were from Hamamatsu. D indicates dichroic mirrors from series A10034, F indicates filters from series A10033, and P210 and P020 refer to PMTs H10722 210 and 020, respectively. CH indicates the channel number. FN refers to a filter that transmits 100% of the incoming light to the photodetector in the wavelength range of 300-500 nm.

Fig. 8  Emission and filtered spectra produced by the P9 mPSD prototype used for absorbed dose determination. The filtered spectra for each channel were obtained by applying optical filtration to the light entering each PMT.

Crystals, Hiram, OH), but with different positions inside the optical assembly. In that work it was suggested to use the BCF-10 in the central position while BCF-12 at proximal position with respect to the photodetector surface. Based on the signal analysis performed in this study, we propose an optimized mPSD design that maximize the scintillation light collection, resulting in the configuration shown in Figure 7, which invert the BCF-10 and BCF-12 positions. Furthermore, the mPSD system studied by Therriault-Proulx et al. 25
is limited to HDR brachytherapy measurements within 3 cm from the source. The system here proposed is able to accurately perform dose measurements beyond 3 cm with a high collection efficiency as described in the following sections.

III.D. mPSD brachytherapy implementation

Dose distributions in terms of distance to an HDR brachytherapy source were obtained, with the P9 mPSD calibrated at 1.5 cm from the $^{192}$Ir source. This calibration distance represented a compromise between measurement uncertainties and positioning errors. Andersen et al. demonstrated that positioning uncertainty dominates in measurements close to the source, whereas measurement uncertainty dominates at long distances. In order to ensure that we had enough data in the response recording, a source dwell time of 60 s was used at each dwell position.

Figure 9 shows dose rate readings for each scintillator, and Table III details the standard deviations for each distance to the brachytherapy source. For all 3 scintillators, the standard deviations were generally no greater than 5% of the mean dose reading, although this value, as expected, increased with distance from the source. At a distance of 6.5 cm, the standard deviation exceeded 10% for all scintillators. At that distance, the source radiation does not produce enough scintillation in the mPSD, so the recorded signal can be considered to be background. Nonetheless, the absolute standard deviation was small relative to the mean dose. TG-43 dose values were used as a reference; the last column in Table III presents the
differences between the measured dose and the TG-43 dose values at each distance to the brachytherapy source. In general, the measured mPSD dose and the TG-43 dose agreed well at short distances to the source, but the difference grew as the source moved away from the mPSD.

III.E. Evaluation of scintillation signal and system sensitivity

Figure 10a shows the SNR in terms of dose rate for each scintillator in detector P9. According to the Rose criteria, proper recognition (detection) of an object strongly depends on SNR, only becoming possible when SNR exceeds 5; detection performance degrades as SNR approaches zero. Thus, an SNR of 5 was the minimum sensitivity considered in this study. The BCF-10 and BCF-12 scintillators produced an SNR greater than 5 at all distances to the source. At dose rates below 22 mGy/s, the SNR produced by the BCF-60 scintillator fell below 5. These data suggest that, with regard to SNR, the dosimetric system characterized in this study is sensitive enough to measure dose rates above 22 mGy/s.

SNR analysis of PSD responses for various photodetectors was conducted by Boivin et al. in the energy range of clinical interest. SNR values in a range of 100 to 1000 for dose rates between 0.1-30 mGy/s were reported, being of around 2 orders of magnitude greater than the results obtained in the low dose rate range of this study. The differences are easily explained by the differences in design. First, the study done by Boivin et al. was conducted with a single point configuration detector that only have a single coupling interface, while the one reported in this study have 5 coupling interfaces intrinsically decreasing the overall light collection in mPSD. Secondly, the PSD size was of 10 mm in Boivin et al. vs. a maximal size of 7 mm. Thirdly and most important, the system here used is subject to optical filtration of the light produced; Boivin et al. used a system with no optical filtration, recording the signal of a PSD directly on a PMT module.

To evaluate how well the dosimetric system differentiated a signal pulse from the background signal, SBR values were calculated at each dose rate. It is important to mention that in order to determine the background signal; several signal acquisitions were performed without irradiation. From this analysis, it was obtained a $\mu_b \pm 0.16 \%$. To properly differentiate signal from background, a minimum SBR of 2 is required. Figure 10b shows the SBR results obtained for each scintillator in the mPSD. The SBR for the BCF-10 scintillator
TABLE III  Standard deviation (SD) of 3-point mPSD measurements and deviation of the mean measured dose from the values predicted by TG43.

| Distance to source (cm) | Dose per reading (mGy) | SD (mGy) (%) | Deviation from TG43 (mGy) (%) |
|------------------------|------------------------|--------------|-------------------------------|
| BCF-10                 |                        |              |                               |
| 0.5                    | 1117.9                 | 0.7          | 7.7                           | 3.0 33.2 |
| 0.7                    | 594.4                  | 0.7          | 4.2                           | 1.1 6.5 |
| 1.0                    | 332.0                  | 1.2          | 4.1                           | 1.1 3.6 |
| 1.1                    | 261.6                  | 1.6          | 4.2                           | 3.7 9.7 |
| 1.5                    | 138.7                  | 2.8          | 3.9                           | 3.8 5.3 |
| 2.0                    | 80.9                   | 4.2          | 3.4                           | 5.5 4.5 |
| 2.5                    | 51.3                   | 4.5          | 2.3                           | 7.5 3.9 |
| 3.2                    | 31.0                   | 4.1          | 1.3                           | 7.3 2.3 |
| 6.5                    | 6.0                    | 12.8         | 0.8                           | 21.3 1.3 |
| BCF-12                 |                        |              |                               |
| 0.5                    | 2146.6                 | 0.5          | 10.3                          | 1.1 23.2 |
| 0.7                    | 1082.6                 | 0.7          | 7.2                           | 0.8 9.1 |
| 1.0                    | 633.6                  | 0.8          | 5.3                           | 2.5 15.6 |
| 1.1                    | 500.9                  | 0.8          | 3.9                           | 2.2 11.1 |
| 1.5                    | 255.4                  | 2.0          | 5.2                           | 2.0 5.1 |
| 2.0                    | 144.8                  | 4.2          | 6.1                           | 2.2 3.2 |
| 2.5                    | 94.6                   | 3.1          | 2.9                           | 2.0 1.9 |
| 3.2                    | 56.6                   | 3.2          | 1.8                           | 2.3 1.3 |
| 6.5                    | 13.3                   | 11.2         | 1.5                           | 12.4 1.6 |
| BCF-60                 |                        |              |                               |
| 0.5                    | 2881.6                 | 0.4          | 11.3                          | 2.3 66.9 |
| 0.7                    | 1293.6                 | 0.7          | 8.5                           | 0.6 7.2 |
| 1.0                    | 778.5                  | 1.2          | 9.4                           | 3.4 26.7 |
| 1.1                    | 614.3                  | 1.2          | 7.1                           | 4.3 26.3 |
| 1.5                    | 302.6                  | 1.7          | 5.0                           | 0.8 2.5 |
| 2.0                    | 175.2                  | 6.5          | 11.4                          | 1.0 1.8 |
| 2.5                    | 119.2                  | 5.0          | 5.9                           | 2.4 2.9 |
| 3.2                    | 68.1                   | 6.8          | 4.6                           | 1.7 1.2 |
Fig. 10 Signal-to-noise ratio (SNR) and (b) signal-to-background ratio (SBR) as function of dose rate for BCF-10, BCF-12, and BCF-60 scintillators in the P9 mPSD prototype. Red shaded areas at the bottoms of the graphs indicate the cutoff values for each parameter.

fell below the SBR cutoff value for dose rate values of around 6 mGy/s, with an SBR of 1.8. SBR is directly proportional to the photon fluence from scintillation, which is in turn proportional to the scintillator volume. According to the previously determined optimal mPSD design, the length of the BCF-10 scintillator was only 3 mm. The dose rate range could be extended by increasing the length of the scintillator, but at the cost of spatial or temporal resolution. As the background signal was almost constant in the explored dose rate range, all the scintillators evidenced almost perfectly linear behavior.

IV. CONCLUSIONS

In this study, we optimized an mPSD system that can be used clinically in HDR brachytherapy. We found that the scintillation light emission per millimeter of scintillator was more efficient in BCF-10 than in BCF-12 or BCF-60 scintillators. Furthermore, we experimentally determined the appropriate position of each scintillator inside the fiber: the scintillating element with the lower wavelength should be placed closer to the photodetector, whereas the scintillator with the longer wavelength (less energy) should be placed distally. In a 2-point mPSD, the most balanced signal was obtained with BCF-10 placed proximally and BCF-60 placed distally. We also evaluated a 3-point mPSD consisting of
BCF-10, BCF-12, and BCF-60 scintillators. The best prototype used 3 mm of BCF-10, 6 mm of BCF-12, and 7 mm of BCF-60. Those dimensions were determined not only on the basis of light emission balance, but also with the aim of improving the detectors spatial resolution. Finally, an optimal light collection system was evaluated in HDR brachytherapy simulations. The evaluated mPSD produced minimal deviations in dose rate readings, and analysis of SNR and SBR showed that the detector provided accurate real-time dose measurements.

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