Systematic evaluation of photodetector performance for plastic scintillation dosimetry

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Purpose: The authors’ objective was to systematically assess the performance of seven photodetectors used in plastic scintillation dosimetry. The authors also propose some guidelines for selecting an appropriate detector for a specific application.

Methods: The plastic scintillation detector (PSD) consisted of a 1-mm diameter, 10-mm long plastic scintillation fiber (BCF-60), which was optically coupled to a clear 10-m long optical fiber of the same diameter. A light-tight plastic sheath covered both fibers and the scintillator end was sealed. The clear fiber end was connected to one of the following photodetectors: two polychromatic cameras (one with an optical lens and one with a fiber optic taper replacing the lens), a monochromatic camera with an optical lens, a PIN photodiode, an avalanche photodiode (APD), or a photomultiplier tube (PMT). A commercially available W1 PSD was also included in the study, but it relied on its own fiber and scintillator. Each PSD was exposed to both low-energy beams (120, 180, and 220 kVp) from an orthovoltage unit and high-energy beams (6 and 23 MV) from a linear accelerator. Various dose rates were tested to identify the operating range and accuracy of each photodetector.

Results: For all photodetectors, the relative uncertainty was less than 5% for dose rates higher than 3 mGy/s. The cameras allowed multiple probes to be used simultaneously, but they are less sensitive to low-light signals. The PIN, APD, and PMT had higher sensitivity, making them more suitable for low dose rate and out-of-field dose monitoring. The relative uncertainty of the PMT was less than 1% at the lowest dose rate achieved (0.10 mGy/s), suggesting that it was optimal for use in live dosimetry.

Conclusions: For dose rates higher than 3 mGy/s, the PIN diode is the most effective photodetector in terms of performance/cost ratio. For lower dose rates, such as those seen in interventional radiology or high-gradient radiotherapy, PMTs are the optimal choice.

Key words: scintillator, dosimeter, PSD, photodetector

1. INTRODUCTION

The use of plastic scintillation detectors (PSDs) is becoming more common in all fields requiring measurement of radiation, including radiotherapy,1–3 brachytherapy,4–6 and, more recently, radiology.7–11 Among the numerous applications for PSDs, small field dosimetry,12–14 real-time in vivo measurements,15,16 and two-dimensional (2D) dosimetry17,18 have been evaluated. PSDs are known for their advantageous properties, such as near water-equivalence and small size, and their response is fast, reproducible, and dose rate-independent.19,20 However, to some extent, PSDs are energy- (for high-LET ionizing radiation)21 and temperature-dependent;22,23 thus, these parameters need to be controlled accordingly. Fiberoptic systems are also vulnerable to the “stem effect” from sources such as fluorescence, phosphorescence, and Cerenkov light.24–26 However, multiple approaches have been proposed to address this issue, such as using a second optical fiber to subtract contamination19 or discriminating light signals relative to their wavelength.17,27–29

PSDs come in various shapes and sizes and several photodetectors have been proposed in the past 20 years.3 A photomultiplier tube (PMT) was first suggested19,20 on account of its high sensitivity, but high voltage supply stability
dependence made it less convenient than a simple, if less sensitive, PIN photodiode.\textsuperscript{3,14,15} Polychromatic photodiodes were also introduced to subtract the stem effect signals.\textsuperscript{3} Charge-coupled device (CCD) and electron-multiplying CCD (EMCCD)\textsuperscript{30} cameras were then used to simultaneously read multiple fibers, allowing the development of PSD grids, although this task can now be performed with a PMT array.\textsuperscript{30} Although various PSD designs have been reported for diverse applications, to the best of our knowledge, no study has evaluated and compared the performance of these designs under the same conditions.

The purpose of the current study was to systematically assess the performance of seven photodetectors used in plastic scintillation dosimetry. We evaluated a polychromatic photodiode, an avalanche photodiode (APD), a PMT, and a commercially available W1 scintillator. Three camera systems were considered, one monochromatic and two polychromatic. One of the polychromatic cameras used a fiber optic taper instead of a glass lens to improve light collection efficiency.\textsuperscript{31} To cover all applications, whether in diagnostic radiology, orthovoltage therapy, or external radiotherapy, we explored energies ranging from 120 kVp to 23 MV. Photodetector response was then investigated by varying the dose rate to which the PSD was exposed. To consider all these detectors as candidates for real-time applications, their minimum readout frequency was set to 1 Hz. Real-time detector systems have the capability to determine the exact moment when deviations between planned and delivered doses occur,\textsuperscript{16} thus providing immediate feedback and no additional readout time. The PSD spectrum was also analyzed with a spectrometer for each of the studied energies, although this procedure required a longer integration time preventing evaluation of the spectrometer along with the other photodetectors. On the basis of our results, we propose guidelines for selecting an appropriate detector for a specific application.

2. MATERIALS AND METHODS

2.A. Dose rate protocols

2.A.1. Orthovoltage

Low-energy beam measurements were obtained using an Xstrahl 200 x-ray therapy system (Xstrahl Ltd., Camberley, UK). This unit presents nine operating energies, ranging from 40 to 220 kVp. We used three beam qualities that have been commissioned for clinical use, described in Table I. The x-ray tube used a tungsten target with a 30° anode angle. There was an inherent filtration of the beam consisting of 0.8 ± 0.1 mm of beryllium, and additional filtration was added for each beam quality. The tube current used was 10 mA and the beam was continuous. Nominal dose rate was measured using a Farmer type Exradin A-12 ionization chamber (Standard Imaging, Middleton, WI; calibrated by the National Research Council, Ottawa, Canada), located at the opening of the standard conic applicator, which had a 10-cm diameter and a 20-cm source-to-axis distance.

Owing to the ionization chamber diameter, nominal dose rate was measured at a source-to-detector distance (SDD) of 20.7 cm. The x-ray tube assembly, which was mounted on a ceiling rail, was then pulled away from the ionization chamber. For each beam quality, successive dose rate measurements were then obtained by increasing the SDD up to 433 cm, which allowed dose rates ranging from 55 mGy/s down to 0.075 mGy/s. We preferred this method over depth dose measurement, which alters primary beam quality and increases scattering. Measurements of dose rate relative to SDD follow the radial inverse square law, but absolute dose rate measurements are required to account for the field flattening, attenuation in air, and treatment room backscattering at high SDD.

2.A.2. Megavoltage

A Varian Clinac iX (Varian Medical Systems, Palo Alto, CA) was used to produce high-energy beams. Measurements were obtained in a 10 × 10 cm\textsuperscript{2} field in a plastic water phantom. Dose rate was measured with the ionization chamber located inside a 2-cm thick slab at 1.5-cm depth for a 6 MV beam or 3.5-cm depth for a 23 MV beam. The phantom extended 15 cm beyond the chamber to consider the backscattered contribution. The gantry was rotated by 90° and the phantom was set on its side and on a cart moving along the beam axis. Dose rate was measured for a SDD of 62 cm, up to 360 cm. The accelerator monitor unit rate was also varied to widen the dose rate range, which was from 265 mGy/s at minimum SDD down to 1.3 mGy/s at maximum SDD. The accelerator beam is pulsed at a maximum rate of 360 Hz, delivering about 0.28 mGy per 4 μs pulse at the isocenter.

2.A.3. Signal acquisition and analysis

The set of photodetectors was composed of three cameras, two passive photodetectors, and two active photodetectors. The camera images were acquired and processed through a homemade analysis software program. With the aim of considering the cameras as real-time photodetectors, integration time was set to 1 s and 50 frames were acquired back to back for each measurement sequence. Every other photodetector was connected to an electrometer (SuperMax; Standard Imaging), controlled by a MATLAB (Mathworks, Natick, MA) code, and reading rate and accumulated charge were set to 10 times/s for 50 s.

The signal acquisition routine was independent of the beam source and detector and was defined as follows. Prior to any exposure, the background signal was collected for at least 30 s, and then the mean value and standard deviation were recorded. The beam was turned on, but the scintillator signal was not.

| Peak energy (kVp) | Added filtration (mm) | HVL 1 (mm) | Dose rate (mGy/s) |
|-------------------|-----------------------|------------|-------------------|
| 120               | 0.5Al + 0.10Cu        | 5.0Al      | 32.9              |
| 180               | 1.5Al + 0.15Cu        | 0.5Cu      | 55.1              |
| 220               | 1.0Al + 0.25Cu + 0.45Sn | 2.0Cu    | 33.3              |
considered until both the beam and the detector outputs were stable. This took less than 15 s for the linear accelerator, but up to 30 s for the orthovoltage beam. This stabilization time was induced by either the slow response circuit of the electrometer low range (if used) or the kilovoltage peak (kVp) and milliamperage second (mAs) rise of the orthovoltage unit. Once stabilization was met, the acquisition began, lasting for 50 s. The background contribution was then subtracted from the exposure value to get the signal measurement mean value ($\mu_{\text{sig}}$) and standard deviation ($\sigma_{\text{sig}}$). For high-energy beams (6 or 23 MV), a correction needed to be applied to remove the Cerenkov contamination, as described in Sec. 2.A.4.

The performance of the photodetectors was compared in terms of signal-to-noise ratio (SNR), which enabled a systematic and analytical exploration of PSD optimization. The SNR can be defined as $\mu_{\text{sig}}/\sigma_{\text{sig}}$ or by the inverse ratio, the coefficient of variation (CV), also known as the relative standard deviation. Each photodetector system was calibrated at the specific dose rate of 30 mGy/s on both treatment units. The deviation from this calibration curve was then analyzed to compare the responses of each photodetector in the setting of different dose rates.

2.A.4. Cerenkov correction

When exposed to radiation, scintillating material emits light proportional to the dose received. However, this signal is contaminated by the Cerenkov light produced in the exposed optical fiber. The Cerenkov effect occurs when charged particles travel faster than the speed of light in a dielectric medium. The threshold energy depends on the medium’s refractive index and the charged particle. For electrons traveling in a polymethyl methacrylate (PMMA) optical fiber, this threshold is 178 keV. Two common methods have been proposed to remove the Cerenkov contamination. Beddar et al. suggested the use of two parallel and adjacent optical fibers, with only one coupled to a scintillating material. Assuming the Cerenkov light is the only factor responsible for the signal difference, and subtracting the Cerenkov light should result in the scintillator signal. However, this method requires two identical photodetectors, a multichannel photodetector or one photodetector collecting each fiber signal successively. This method is limited when steep dose gradients cross over the fiber. It also depends on fiber, coupling, and photodetector equivalence. In the current study, the two-fiber method was used for the PMT, APD, monochromatic camera, and the spectrometer systems.

The second approach, known as the spectral method, was proposed by Fontbonne et al. and explicitly formulated by Frelin et al. It relies on the spectrum analysis of the Cerenkov light, the intensity of which depends directly on the length of the exposed fiber. Guillot et al. reformulated this method by expressing the calibration factors independently. Any measurement $M$ made with a polychromatic photodetector can then be corrected for its Cerenkov contamination using the spectral method and the following equation:

$$M = a \cdot (T_G - T_B \cdot \text{CLR}),$$

where $T_G$ and $T_B$ are the light signals measured by the green and blue color channels, CLR is the Cerenkov light ratio, and $a$ is a gain factor. In the current study, we used this method for the W1 scintillator, PIN photodiode, and the polychromatic and taper cameras.

2.B. PSD

2.B.1. Design

The PSD was made of a polystyrene scintillating fiber, BCF-60 (Saint-Gobain Crystals, Paris, France) which emitted at a peak of 530 nm. The scintillator was 10 mm long and had a 1-mm diameter. It was coupled to a PMMA clear optical fiber (Eska GH-4001, Mitsubishi International Corporation, NY) of the same diameter, which was 10 m long. The use of PMMA fiber is known to minimize fluorescent light generation. Each interface had been polished with an automated optical fiber polisher (SpecPro, Krell Technologies, NJ) with grain size down to 0.3 µm. Both fibers were covered by a light-tight polyethylene jacket and the scintillating tip was sealed with PTFE tape and a mixture of epoxy and black acrylic paint. The other end was uncovered when used with a camera or inserted into a SMA connector (11040A, Thorslab, Newton, NJ) to plug into the other photodetectors studied. To remove the Cerenkov effect at high energies, we prepared a second fiber of the same length and diameter, but without the scintillator component.

2.B.2. Calibration protocol

At low energies, the PSD was held on a stand at the same location as the reference ionization chamber in air, as described in Fig. 1(a). At high energies, the PSD and the clear fiber (if needed) were sandwiched between a plastic water slab and a 1-cm thick superflab bolus (Mick Radio-nuclear Instruments, Mount Vernon, NY) to preserve fiber integrity and medium homogeneity. The fiber was located at the same depth as the ionization chamber and handled similarly, as in Fig. 1(b).

![Fig. 1.](image-url)
2.C. Spectrum analysis

The scintillator spectrum was acquired for each energy beam using a QE65 Pro spectrometer (Ocean Optics, Dunedin, FL) cooled to −20°C. The scintillating and clear fibers were located at nominal dose rate locations: SDD of 20.7 cm for the orthovoltage unit and SDD of 1 m for the linear accelerator. The spectrometer integration time was set to 50 s and the signals of both fibers were collected successively, with and without exposure, to remove the background noise and the stem effect.

2.D. Cameras

2.D.1. Polychromatic camera

We used an Alta U2020CL polychromatic CCD camera (Apogee Imaging System, Roseville, CA). The array size was 1600 × 1200 pixels of 7.4 µm. The camera was cooled down to −18°C to maximize SNR. This camera had a 50-mm focal length lens (JML Optical Industries, Rochester, NY) and was set on an optical plate located in the treatment room maze. A ridged stand held the PSD at a fixed distance. The camera was shielded by 3-mm lead sheets at low energies and 50-mm lead bricks at high energies. Black blankets covered the setup and light was turned off during exposure to minimize ambient light contamination on the CCD. To consider the camera as a real-time photodetector, we set the integration time to 1 s and acquired 50 frames back to back for each measurement, with and without exposure. Acquisitions were started when the radiation beam reached a stable dose rate. Median images were then processed for each sequence with the background removed, and the spot median gray level and standard deviation were extracted.

2.D.2. Monochromatic camera

The monochromatic CCD camera (U-4020ML; Apogee Imaging System) was slightly different from the polychromatic camera. Optical lens, pixel size, and cooling temperature were unchanged, but the array size was larger (2048 × 2048) and the nominal peak quantum efficiency was higher (55%, compared with 37% for the polychromatic camera) when there was no color distinction. At high energies, a clear fiber was positioned next to the scintillating fiber in front of the camera and the two spots were imaged and analyzed simultaneously. The stem signal was then subtracted from the scintillator signal. Otherwise, the acquisition protocol was the same as with the polychromatic camera.

2.D.3. Taper camera

The third camera (U4020CL; Apogee Imaging System) was a polychromatic version of the monochromatic camera described above. The main difference was that a custom-made fiberoptic taper was coupled to the CCD instead of the optical lens. This component comprised bundles of thousands of optical fibers welded together and stretched into a conic shape, acting as a 3:1 reducing lens. These bundles had a diameter of about 450 µm and are shown in Fig. 2. Individual optical fibers had a minimal diameter of 5 µm; because this is less than the pixel size, the fibers are indistinguishable in the picture. The PSD was directly coupled to the taper surface, which increased the amount of light collected up to the CCD (compared with a lens-based system) and minimized light scattering. The resulting image appeared as a sharp spot, whereas the other cameras produced gaussian-shaped spots, as shown in Fig. 3. However, the CCD cooling circuit was disconnected when the taper was mounted to avoid strains and condensation. Without temperature control, the camera needed to be operated until the CCD temperature reached a stable temperature of 30°C, thus increasing the background noise relative to the lens system. The acquisition and signal processing procedures were unchanged from those used with the polychromatic camera.

2.E. Passive photodetectors

2.E.1. PIN photodiode

A SMA connector was used to connect the optical fiber to the surface of a polychromatic PIN photodiode (Sensor-ICs True Color Sensor; MAzet GmbH, Jena, Germany). The photodetector had 19 sensors collecting light in three colors (red, green, and blue), and each color readout was generated using a triax output. The green channel photosensitivity reached 0.30 A/W at the scintillator wavelength of 530 nm. The photodiode was set in a light-tight aluminum box located outside of the treatment room, so that no additional shielding was necessary. The charge produced was read by the dual-channel electrometer, monitored in real time by a computer. Signal was acquired for 50 s, and the charge was read.
10 times/s. Because the green channel sensitivity closely matched the scintillator peak used in our study, the green channel was used to measure the scintillation signal under a low-energy beam. However, the blue channel was required when the spectral method was applied.

2.E.2. W1

The Exradin W1 scintillator (Standard Imaging) is a commercially available PSD intended for small field dosimetry. It has recently been characterized and compared with a similar house-made system. The W1 relies on a two-channel photodiode enclosed in a shielded case providing triax outputs for both green and blue signals. Its clear fiber is made of PMMA with a polyethylene jacket similar to the system described above. However, the fiber was only 3 m long, so the photodiode remained in the treatment room during the exposure. The scintillating fiber was also shorter (3 mm long and 1 mm in diameter). The scintillating material itself was unknown but the signal output read by the different channels suggested that it was a blue-emitting scintillator. Thus, it was different from the BCF-60 fiber used with all the other photodetectors. The manufacturer described the scintillating fiber as polystyrene with an ABS plastic enclosure and a polyanime stem. Signal acquisition and treatment procedures were the same as described for the PIN photodiode.

2.F. Active photodetectors

2.F.1. APD

An APD C4777-01 (Hamamatsu Photonics, Hamamatsu, Japan) was used to substantially increase the amount of signal collected. Electron-hole pairs created by incoming photons were accelerated by a bias voltage, triggering an avalanche process through multiple impact ionizations. The fiber was connected with a SMA connector inside a light-tight cover and coupled directly to the photosensitive surface, which had an active area of $\phi$ 3 mm. The APD gain was 50, with a 50 MΩ feedback resistance, and its photosensitivity reached 0.15 A/W at 530 nm. A two-stage thermoelectric cooler maintained the APD chip temperature at a constant 0°C. A triax output was connected to the electrometer and the signal was acquired and processed as for the PIN photodiode. However, because no polychromatic information was available, the clear fiber irradiation was performed successively to remove the steam effect under high-energy beams.

2.F.2. PMT

We used an H10721-20 PMT module (Hamamatsu Photonics) with a multialkali photocathode. A light-tight SMA adapter (E5776-51, Hamamatsu Photonics) aligned the fiber to the photosensitive surface of $\phi$ 8 mm. The PMT had an adjustable gain of $5 \times 10^3$ to $5 \times 10^6$, although we selected the minimum gain to avoid saturation of the electrometer, connected through triax. The PMT photosensitivity reached 0.075 A/W at 530 nm and 375 A/W when the minimum gain voltage is applied. The PMT module was operated at ambient temperature. Signal was acquired and processed as for the PIN photodiode, and irradiation of a clear fiber was also required at high energies, as described above.

3. RESULTS

3.A. Absolute dose rate measurements

The ionization chamber measured the absolute dose rate as the SDD was increased. Because the exposure follows
the inverse square law, the dose rate increased linearly with $SDD^{-2}$. Measurements showed that the dose rate difference remained below ±3% for high-energy beams. However, attenuation in air caused the low-energy beam dose rate to decrease relative to the inverse square law. At $SDD = 350$ cm, the difference was 7% at 120 kVp, 4% at 180 kVp, and 2% at 220 kVp. This illustrated the need for a reference absolute dose reading at every measurement point.

3.B. PSD measurements

3.B.1. Energy dependence

The clear fiber was exposed so that we could observe the fluorescence and Cerenkov contamination. At low energies, the spectrometer detected no significant signal above the background noise, indicating that the fluorescence signal can be considered negligible in our study. At high energy, the Cerenkov signal was visible and this contamination relative to the scintillator signal amounted to up to 9% for a 6-MV beam and 13% for a 23-MV beam when 5 cm of the clear fiber was exposed (isocenter, $10 \times 10$ field).

The scintillator response is known to be energy-dependent under a low-energy beam.\textsuperscript{21} This was expressed by a rise in the number of counts per mGy when increasing energy. Relative to a 120 kVp beam quality, the response increase of 180 and 220 kVp beam qualities was 31% and 83%. For megavoltage 6- and 23-MV beams, the increase reached 125% and 129%. Figure 4 shows these spectra normalized to the maximum value when the clear fiber signal and background are removed. The superimposition of spectra reveals that the scintillator spectrum shape was energy-independent (at least above 120 kVp) and did not suffer from wavelength shifting. These results indicate that there is no need to apply correction for spectrum variation as long as the Cerenkov contribution is removed. Both correction methods (two-fiber and spectral) performed equally well for the type of measurements obtained in our study.

3.B.2. SNR

Figure 5 presents the SNR curves for a 120-kVp and a 220-kVp orthovoltage unit beam and for a 6- and a 23-MV linear accelerator beam. These four graphs use common axes and each set of points is related to a power law fit to compare the SNRs of the photodetectors. The low-energy graphs [Figs. 5(a) and 5(b)] show how the amplified photodetectors (APD and PMT) outperformed the passive photodetectors.
(PIN and W1), especially at low dose rates, whereas both the polychromatic and monochromatic lens cameras offered a weak SNR. However, the taper camera reached a SNR similar to that of the PIN system, even though the background level of the taper camera was higher than that of the other cameras. The SNR increase of the PMT low-energy measurements around 0.2 mGy/s relative to higher dose rates measurements up to 1.5 mGy/s is explained by the electrometer scale range. Below 0.500 nA, the low-range scale is selected, which added a fourth significant digit, thus increasing the SNR. The low-energy SNR curves were generally similar between the three energies tested. However, the SNR increased slightly between 120 and 220 kVp owing to the increased response of the scintillator, as indicated by the spectrometer readings.

The high-energy graphs [Figs. 5(c) and 5(d)] also showed similar results between the two energies tested. At dose rates higher than about 20 mGy/s, the PIN photodiode SNR caught up with and exceeded that of the PMT. The SNRs of the W1 and the cameras were similar, but the W1 SNR increased at 23 MV as the dose rate increased. The APD was unable to measure dose rates above 100 mGy/s at 6 MV and above 20 mGy/s at 23 MV as light input reached the saturation level. The APD was also affected by heating and needed to be powered up for an hour before the measurements were obtained.

3.B.3. Linearity

To characterize the photodetector responses, we performed a calibration for each energy beam at the specific dose rate of 30 mGy/s for the orthovoltage unit and the linear accelerator. This calibration point was then used to compare each photodetector measured dose rate with the ionization chamber absolute measurement. Figure 6(a) shows the photodetector linearity under a 120-kVp beam. Deviation curves are shown in Fig. 6(b) to describe the relative difference between the photodetector and the ionization chamber readings. These results are consistent with the SNR data and define the operating range of each photodetector relative to the expected dose rate. Once again, the PMT results stood out, suggesting that the PMT has potential for use in very low dose rate applications.

Although the dose rate was higher, the results were similar at high energy beam while increasing the SDD, but also by varying the monitor unit rate. This indicated that there was no specific dependence to the linear accelerator pulse rate.

4. DISCUSSION

4.A. Photodetector performance evaluation

The performances of every evaluated photodetector make them suitable candidates for the making of a PSD. However, the features and qualities of each one make some of them more appropriate than others for specific applications. Evaluation of the photodetectors in a systematic approach offered an opportunity to compare the performance of detectors on the same footing. A common index such as SNR or CV can then be useful to identify the operational dose rate range. Table II describes, for each energy range, the minimum dose rate at which the CV of a photodetector remains below 1% or 5%, which is equivalent to a SNR of 100 or 20. At dose rates used in superficial therapy (see Table I) and in-field radiotherapy, it appears that every photodetector we tested was sensitive enough to be considered as a suitable PSD component. However, many applications involve lower dose rates. In radiotherapy, quality controls require depth dose and profile measurements. In vivo measurements are also considered, such as in-field and out-of-field skin dose measurements, like fetal and pacemaker dose monitoring. In high dose rate brachytherapy, seed insertion can also be monitored by a sensitive PSD, although its use in low energy low dose rate brachytherapy has not yet been investigated. For information purposes, the air kerma rate in air for a high dose rate 41 000 U 125I source is about 114 mGy/s at 1 cm, whereas the air kerma rate in air for a low dose rate 1.27 U 125I seed at the same distance is approximately 3.5 μGy/s. Nuclear medicine may also offer some interesting applications for PSDs, such as syringe dose indicator or skin dose monitoring for out-of-screen organs at risk. The field of radiology may involve substantial cumulative dose and a PSD could be useful for skin dose measurement, especially in CT and long-lasting fluoroscopic procedures such as in interventional radiology.

The photodetectors in Table II are grouped as described in Sec. 2 (i.e., as cameras or passive or active photodetectors). The cameras were less sensitive to the light produced by the scintillator, especially the polychromatic camera. The focal length, optical fiber numerical aperture, and the lenses themselves all contributed to light loss. Moreover, the color pixels did not equally participate in the spot formation because the blue and red channels had a lower response than the green.
channel to the green light of the BCF-60 scintillating material. Grayscale pixels allowed the monochromatic camera to read dose rates two times smaller than on the other cameras for the same CV. However, the Cerenkov correction required a second clear fiber for this camera, which was not a major technical drawback when working with cameras but might be a hurdle for small-field and high-gradient measurements. The taper camera had high collection efficiency, which allowed it to measure dose rates four times lower than those measured by the lens cameras, even though the noise level was higher and no cooling was available. It is reasonable to assume that a cooled taper camera could reach a higher SNR similar to the active photodetectors.

The PIN photodiode and the W1 systems offered similar dose rates at specific CVs, even though the W1 scintillating fiber was 3 mm long, and the custom fiber was 10 mm long. It is possible that the commercial product (i.e., W1) had an improved optical chain (e.g., polishing, coupling), thus improving the light collection. Although the W1 was able to measure lower dose rates than the PIN photodiode at low energies, the PIN system performed better for measuring low dose rates at high energies. This discrepancy may be explained by different photodiodes, scintillating materials, or radiation noise at high energies because the W1 had to remain in the treatment room. The main issue with passive photodetectors is their extremely low output signal, in the range of picoamperes or less. A very sensitive electrometer is required to obtain these measurements. However, because these instruments are not powered, their signal is not affected by the electric noise. Interestingly, the taper camera achieved a performance closer to that of the passive detectors than did the other cameras.

The APD offers great potential because it generates a current readout about 2000 times higher than that of the PIN system. However, the avalanche process requires high voltage within the APD module, which is responsible for heating. Despite the manufacturer claim that a temperature control circuit keeps the APD chip temperature constant and ensuring stable measurements, it was obvious that the APD module was affected by the heat. Thus, it was necessary to power up the APD for an hour prior its use to avoid background drifting. The APD also failed to read high dose rates because of the signal saturation.

The PMT generated a current on the order of 30000 times higher than the PIN photodiode. For the data presented, a minimum gain setting was selected so that the signal remained within the electrometer reading range. An additional gain of 1000 would be possible at maximum input gain, sufficient to allow data acquisition through a more compact and affordable instrument than an electrometer. In our experiments, we found no temperature issue with the selected PMT and it featured a superior SNR, even at the lowest measured dose rate of 0.075 mGy/s on the orthovoltage unit. It should also be noted that active photodetectors amplify both the signal and the related noise, whereas noise in low-output passive photodetectors is related to the electrometer circuits. Thus, as dose rate increases, the SNRs of passive photodetectors are similar to and may even exceed those of active photodetectors, as shown in Fig. 5.

In order to perform coherent measurements and for its convenience, a single electrometer was used to read current from the ionization chamber, the passive and the active photodetectors. However, this instrument had its own limitations, such as a fairly low saturation level (500 nA) and a low reading rate (10 Hz). A dedicated acquisition system would be more appropriate for a specific detector, especially with the active photodetectors that are not used at their full potential. For example, a higher frequency would allow dose-per-pulse measurements, redefining real-time acquisition.30,39

### 4.B. Photodetector selection guidelines

Section 4.B is intended to offer some guidelines for choosing an appropriate detector for a specific application. One should first ask these three questions: (1) Will the detector be used in-field or out-of-field; in other words, what is the expected dose rate range? (2) Is the detector intended to measure low- or high-energy beams? (For high-energy beams, it is necessary to remove the Cerenkov contamination.) (3) Is a single fiber enough or should many fibers be read simultaneously? The photodetectors evaluated in the current

|                  | 120 kVp | 180 kVp | 220 kVp | 6 MV  | 23 MV |
|------------------|---------|---------|---------|-------|-------|
|                   | 1%      | 5%      | 1%      | 5%    | 1%    | 5%    |
| Cameras           |         |         |         |       |       |       |
| Polychromatic     | 15      | 2.9     | 12      | 2.5   | 8.6   | 1.6   | 12    | <2.0  | 13    | <2.0  |
| Monochromatic     | 8.8     | 1.5     | 7.2     | 1.2   | 4.9   | 0.8   | 6.2   | <2.0  | 7.9   | <2.0  |
| Taper             | 4.0     | 0.69    | 3.3     | 0.46  | 2.4   | 0.39  | 3.9   | <2.0  | 3.9   | <2.0  |
| Passive photodetectors |       |         |         |       |       |       |
| PIN               | 5.0     | 1.0     | 3.8     | 0.78  | 2.9   | 0.58  | <2.0  | <2.0  | 2.4   | <2.0  |
| W1                | 3.6     | 0.66    | 2.6     | 0.44  | 1.7   | 0.27  | 5.1   | <2.0  | 5.1   | <2.0  |
| Active photodetectors |       |         |         |       |       |       |
| APD               | 1.1     | 0.10    | 0.63    | 0.10  | 0.59  | 0.10  | 3.8   | <2.0  | 4.2   | <2.0  |
| PMT               | <0.10   | <0.10   | <0.10   | <0.10 | <0.10 | <0.10 | <2.0  | <2.0  | <2.0  | <2.0  |
study are classified according to these three parameters, and selection guidelines are summarized in Table III.

If a single-fiber setup is considered, the PMT is the first choice at low energies because it is sensitive enough to measure the lowest dose rates. However, one PMT alone is limited by the Cerenkov contamination at high energy. The two-channel photodetectors, with one connected to a scintillator-free fiber, are more appropriate at high energies to remove this contamination. Two PMTs could be used, but the PIN photodiode SNR is nearly the same as that of the PMT at high dose rates, offering a very cost-effective solution for light readout. Alternatively, the two-PMT setup could be connected to a single fiber through an optical chain, splitting the scintillator light into different spectra windows so that the spectral method could be used. Passive photodetectors are more useful for standard in-field measures, while PMTs are sensitive enough for out-of-field readings.

In a multiple-point system, the cameras are more convenient instruments, than the other photodetectors, because many fibers, up to hundreds, can be gathered toward the CCD. This leads to interesting applications, such as linear arrays, 2D matrices, or even 3D scintillation dosimetry. However, the sensitivity of the cameras is limited compared with single-point photodetectors. The dose rate range could be extended by increasing either the length of the scintillator or the integration time, but that will be at the cost of spatial or time resolution. According to the SNR model developed by Lacroix et al., the SNR is directly proportional to the photon fluence, which is itself proportional to the scintillator volume, meaning that doubling the length of the scintillator volume should double the SNR.

More sensitive cameras than those we evaluated are now available, such as intensified CCD and EMCCD cameras. The benefit of these instruments appears only at low dose rates. The lower multiplication noise factor of EMCCDs compared with intensified CCDs makes the EMCCDs more suited for radiotherapy scintillation dosimetry than the intensified CCDs, but this multiplication noise factor still constitutes a major source of additional noise in EMCCDs compared with CCDs. The taper system provides a higher SNR than that of a conventional lens camera and allows a more compact setup. Because the taper cooling circuit was disabled in our study, the full potential of the taper camera could not be revealed. At low dose rates (<1 mGy/s), the cameras described in the current study were unable to provide a sufficient SNR to allow accurate real-time measurement. However, a combination of a taper and an EMCCD camera could offer an interesting way to maximize camera light collection efficiency, because the potential of both of these cameras has already been shown for real-time dosimetry.

An other possible design is a combination of a multichannel PMT and optical filters, as the one developed by Liu et al., using a multianode PMT. A PMT array system would allow simultaneous reading as a CCD camera, but with the speed and sensitivity of a PMT.

It should be noted that the focus of the current study was on the photodetectors, but the PSD is composed of two other components that can also be selected. Manipulation of the length, shape, and composition of the scintillator or the use of multiple scintillators offers countless combinations. The light guide length, material, and optical chain should also be optimized to increase the SNR of the system.

### Table III. Guidelines for selecting an appropriate photodetector component for a plastic scintillation detector.

| Source | Low energy | Medium dose rate | High energy |
|--------|------------|-----------------|-------------|
|        | Low dose rate | Medium dose rate | High dose rate |
|        | (<1 mGy/s)     | (1–10 mGy/s)     | (>10 mGy/s) |
| Single point | PMT          | PMT             | PMT
|          | PIN/W1     | 2 × PMT         | 2 × PMT |
| Multiple points | Multiple PMTs | Taper/monochromatic | Taper/monochromatic |
|          | Monochromatic | Polychromatic    | Polychromatic |

*These configurations require a scintillator-free fiber to remove the Cerenkov contamination.

5. CONCLUSION

The purpose of the current study was to systematically assess the performance of seven photodetectors used in scintillation dosimetry. A spectrometer was also involved to measure the scintillator light spectrum. The scintillators were exposed to both low- and high-energy beams. To characterize the linearity and SNR of each photodetector, we investigated multiple dose rates by varying the SDD along the beam axis. The active photodetectors presented the highest SNRs at low dose rates and low energies. However, the passive photodiodes had similar SNRs at higher dose rates and energies. The cameras allowed multiple probes to be used simultaneously, but they were less sensitive to low-light signals. However, coupling an optical fiber taper to a CCD camera improved the SNR up to the level of the passive photodetectors.

The current study also provides some guidelines for selecting an appropriate detector according to a specific application. The criteria were principally based on the expected dose rate, energy, and the number of simultaneous points of measurement. The PSD design should reflect the expected operating conditions, because no photodetector is appropriate for every setup. The data presented here are intended to ease and clarify such decisions.
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(Author’s note: The electronic mail is not relevant to the content and should be ignored.)