A design methodology using signal-to-noise ratio for plastic scintillation detectors design and performance optimization

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Purpose: The design of novel plastic scintillation detectors (PSDs) is impeded by the lack of a suitable framework to simulate and predict their performance. The authors propose to use the signal-to-noise ratio (SNR) to model the performance of PSDs that use charge-coupled devices (CCDs) as photodetectors.

Methods: In PSDs using CCDs, the SNR is inversely related to the normalized standard deviation of the dose measurement. Thus, optimizing the SNR directly optimizes the system's precision. In this work, a model of SNR as a function of the system parameters is derived for optical fiber-based PSD systems. Furthermore, this proposed model is validated using experimental results. A formula for the efficiency of fiber coupling to CCDs is derived and used to simulate the performance of a PSD under varying magnifications.

Results: The proposed model is shown to simulate the experimental performance of an actual PSD to a suitable degree of accuracy under various conditions.

Conclusions: The SNR constitutes a useful tool to simulate the dosimetric precision of PSDs. Using the SNR model, recommendations for the design and optimization of PSDs are provided. Using the same framework, recommendations for non-fiber-based PSDs are also provided. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3231947](https://doi.org/10.1118/1.3231947)]

Key words: plastic scintillation detectors, signal-to-noise ratio (SNR), dosimeter array, dosimetry, dosimeter precision

I. INTRODUCTION

The amount of research on the use of plastic scintillation detectors (PSDs) in dosimetry applications has grown considerably in recent years. For instance, several researchers investigated the use of single- or multipoint PSDs in various clinical applications: Photon and electron beam characterization,^{1–4} quality assurance,^{5–7} small-field dosimetry,^{8–10} brachytherapy dosimetry,^{10–12} and proton beam dosimetry.¹³ PSDs possess a unique combination of properties including water equivalence over a broad energy range (from 0.2 to 25 MeV), a highly sensitive detecting medium which enables miniature active volumes ($<2 \text{ mm}^3$), dose rate- and energy-independent response, and real-time readout,^{1,2,4} all of which make them prime candidates for the construction of dense, water-equivalent two-dimensional (2D) or three-dimensional (3D) dosimeters.^{5,14–16}

Unlike liquid scintillator¹⁷ and plastic scintillator sheet^{15,16} detector systems, which do not use optical light

guides, PSDs are composed of a section of plastic scintillating fiber coupled to an optical fiber that transports the scintillation photons to a photodetector. The photodetector can be a photomultiplier tube (PMT), a photodiode, or a charge-coupled device (CCD). CCDs are fast becoming the photodetector of choice because of their ability to image many detectors simultaneously in a cost-effective manner. Within the past few years, several CCD-based PSD arrays, that is, using multiple (either 8 or 29) scintillating fibers as dose detection points (or “PSD arrays”) have been demonstrated.^{5,14} CCD-based planar scintillator sheet and liquid scintillator systems have also been demonstrated.^{15–17} Today, PSDs are usually designed using an empirical trial and error approach. This is likely to lead to suboptimal systems. We propose a systematic approach to PSD design using the concept of signal-to-noise ratio (SNR).

The SNR is a metric commonly used to characterize the global performance of optoelectronic systems but it has rarely been used to evaluate the performance of PSDs. To our

knowledge, Beddar *et al.*¹⁸ are the only ones who have used SNR (where the noise term includes the Čerenkov radiation) to model the performance of PSDs using PMTs as a photodetector.¹⁸ In CCDs, SNR is inversely related to the standard deviation of the image analog-to-digital units (ADUs) (Ref. 19) or counts (i.e., image pixel values) and, therefore, it characterizes the measurement precision of the dosimeter. In this work, we propose a numerical model that allows the SNR to be calculated as a function of the fundamental system parameters for fiber-based PSD arrays. We then validate the model using experimental data for an existing array.

II. METHODS AND MATERIALS

II.A. SNR for CCD-based PSD arrays

Figure 1 illustrates the basic components comprising a CCD-based PSD array. Sections of plastic scintillating fibers are glued to the ends of optical fibers (ideally water-equivalent plastic fibers) which are used as a waveguide to transport photons to the photodetector. The emission cone of optical fibers is large, thus a significant fraction of the light emitted by the fiber is not captured by the objective lens. Since the scintillation photons carry the dose information, it is crucial in scintillation dosimetry to maximize the photon throughput to the photodetector. The fraction of the light incident on the objective lens can be increased by placing the optical fibers closer to the lens, but this also decreases the CCD's field of view and thus the number of fibers that can be imaged. There are thus trade-offs between the number of fibers that can be imaged and the efficiency with which they can be imaged.

In image processing, the SNR is defined as the ratio of the mean to the standard deviation of a pixel value.¹⁹ Therefore, the inverse of the SNR directly provides the normalized standard deviation of the signal as follows:

$$\sigma_s = \text{SNR}^{-1}. \quad (1)$$

However, the image SNR must be clearly distinguished from the optical spot SNR. While in imaging applications the SNR of an image pixel is the point of interest, in PSD arrays it is not. Rather, the precision of the dose measurement is linked to the SNR of the collection of pixels that receives the optical signal emitted by a given fiber. Usually, the optical signal of a single fiber (the “spot”) falls on several tens of adjacent pixels. The SNRs of each individual pixel must be combined to determine the “spot SNR,” which is the metric of interest in scintillation dosimetry. To obtain the spot SNR, the signal and noise power for each pixel must be summed over n pixels composing the spot, yielding

$$\text{SNR}_{\text{spot}} = \sqrt{n} \text{SNR}_{\text{ave}}, \quad (2)$$

where SNR_{ave} is the average SNR of the pixels within the spot. Signal and noise terms are needed before SNR can be calculated. The signal term is given by the optical photon fluence detected by the CCD per unit time per pixel. The noise term is the sum of the various noise components that affect this signal. For CCDs, three main sources of noise are

FIG. 1. Schematic diagram of a CCD-based PSD array.

usually considered: (1) Shot, or photon, noise, (2) dark noise, and (3) readout noise.¹⁹ Note that readout noise includes all the CCD electronic noises such as truncation noise and processing noise.¹⁹ The variance of the noise sources present in the system can be added in quadrature to calculate the total variance of noise since all noise terms are uncorrelated. Transient noise in the image, which is produced by high-energy particles incident on the CCD pixels and, in turn, generates spurious charge carriers, can also be present in the image if the CCD is located inside the radiation area (i.e., the treatment room used for radiotherapy). Transient noise is usually removed by using a temporal or spatial filtering algorithm.²⁰ Combining N images to form a new median image improves the SNR by a factor of $\sqrt{N}$.¹⁹ The SNR for a spot for a particular spectral band (red, green, or blue) is calculated by summing over the n red, green, or blue pixels of the spot as follows:

$$\begin{aligned} \text{SNR}_{\text{spot red, green, or blue}} &= \sqrt{N} \frac{\sum_{i=1}^n \Phi_p(i) \eta_q T}{\sqrt{\sum_{i=1}^n \Phi_p(i) \eta_q T + \sum_{i=1}^n D(i) T + \sum_{i=1}^n N_r^2(i)}}, \end{aligned} \quad (3a)$$

where $\Phi_p(i)$ is the photon fluence (photons $\text{pixel}^{-1} \text{s}^{-1}$) on pixel i , η_q is the quantum efficiency T is the integration time for one image (in seconds), $D(i)$ is the variance of the dark noise (electrons) on pixel i , and $N_r^2(i)$ is the variance of the readout noise (electrons) on pixel i . In general, the imaged fiber subtends a finite number of pixels (n). N represents the number of images or frames that are averaged. When using average values within a spot for the photon fluence and noise terms, Eq. (3a) reduces to

$$\begin{aligned} \text{SNR}_{\text{spot red, green, or blue}} &= \sqrt{N} \sqrt{n} \frac{\Phi_{p,\text{ave}} \eta_q T}{\sqrt{\Phi_{p,\text{ave}} \eta_q T + D_{\text{ave}} T + N_{r,\text{ave}}^2}}. \end{aligned} \quad (3b)$$

The photon fluence term Φ_p in Eq. (3) can be rewritten to account for the PSD system construction as follows:

$$\Phi_{p,\text{ave}} = S \Phi_{\text{sci}} \eta_{\text{fib}} L \eta_{\text{obj}} A. \quad (4)$$

Here, Φ_p is the photon fluence detected by the CCD per pixel (photons $\text{pixel}^{-1} \text{s}^{-1}$). S is an apodization term that is the inverse of the area of the spot on the CCD imaging area and that converts photons to a photon fluence. It is approximately equal to the inverse of the total number of pixels on the CCD chip times the pixel surface area (given by $[1/An_{\text{tot}}]$). Φ_{sci} is

the photon light yield (photons s⁻¹) isotropically produced inside the scintillating fiber or scintillating plastic when exposed to a certain dose rate, η_{fib} is the coupling efficiency from the scintillating fiber to the optical fiber, L is the transmission of the optical fiber, η_{obj} is the coupling efficiency from the fiber output through the objective lens to the CCD plane, and A is the pixel area. Thus, as shown in Eq. (3b), the SNR can be calculated as a function of the optical photon fluence produced in the scintillator, the coupling efficiency from the scintillator to the optical fiber, the optical fiber loss, the objective lens coupling efficiency, and the noise terms.

II.B. Objective lens coupling efficiency

One point of much interest is the coupling efficiency that can be obtained using objective lenses. A similar coupling problem has been encountered in digital radiography systems. In their study on lens coupling efficiency for such systems, Yu and Boone²¹ derived coupling formulas based on the assumption that a phosphorescent radiology screen is an isotropic point source or a Lambertian extended source. However, optical fibers emit light within a divergent but finite cone. The most accurate way to solve the coupling problem in optical fiber systems would be to model and simulate the optical system using an optical design software such as CODEV.²² However, it is not necessary to perform this time-consuming optical modeling task to obtain a first-order approximation that can serve as a design guideline. To solve the problem analytically, the following two assumptions must be made: (1) The fiber is a point source and (2) the distribution of light is uniform angularly within a cone of a certain half-angle obtained from the numerical aperture of the optical fiber. The problem is then reduced to a form in which geometrical optics calculations can be performed. The expression for coupling efficiency of optical fibers therefore can be derived following the guidelines outlined in Ref. 21,

$$\eta_{\text{obj}} = \frac{m^2}{4F^2(m+1)^2 \tan^2(\theta)}, \quad (5)$$

where (m) is the magnification, defined as the image size divided by the object size, and θ is the optical fiber emission cone half-angle. The coupling efficiency (η_{obj}) is proportional to the magnification (m) squared and inversely proportional to the square of the F -number (F). The SNR obtained in the image produced by the CCD is what ultimately determines the dosimeter's performance. Equation (4) can be rewritten to include coupling efficiency explicitly for PSD arrays,

$$\Phi_{p,\text{ave}} = S \Phi_{\text{sci}} \eta_{\text{fib}} L \frac{m^2}{4F^2(m+1)^2 \tan^2(\theta)}, \quad (6)$$

where η_{obj} is replaced with Eq. (5) and S is replaced with $(1/An_{\text{tot}})$. Note that the units of Φ_p are photons pixel⁻¹ s⁻¹. The full expression for the SNR for a single pixel for one frame then becomes

$$\text{SNR}_{\text{pixel}} \approx \frac{S \Phi_{\text{sci}} \eta_{\text{fib}} L \frac{m^2}{4F^2(m+1)^2 \tan^2(\theta)} \eta_q T}{\sqrt{S \Phi_{\text{sci}} \eta_{\text{fib}} L \frac{m^2}{4F^2(m+1)^2 \tan^2(\theta)} \eta_q T + DT + N_r^2}}. \quad (7)$$

Equation (7) is given in terms of electrons and provides the SNR as a function of the fundamental parameters for PSD arrays. In the shot noise limit, as the shot noise becomes much more important than dark and readout noises (this happens at high photon fluences, i.e., when many thousands of photons are detected by each pixel), Eq. (7) reduces to

$$\text{SNR}_{\text{pixel}} \approx \sqrt{S \Phi_{\text{sci}} \eta_{\text{fib}} L \frac{m^2}{4F^2(m+1)^2 \tan^2(\theta)} \eta_q T}. \quad (8)$$

The dose calculation is performed using the image pixel values given in ADU. A certain number of electrons must be detected by the CCD to produce 1 analog digital unit (ADU). This is termed the CCD conversion factor g . Equation (7) can be converted to a numerically exploitable form. The apodization factor S is replaced by $1/An_{\text{tot}}$.

$\text{SNR}_{\text{pixel}}$

$$\approx \sqrt{\frac{1}{g} \frac{\Phi_{\text{sci}} \eta_{\text{fib}} L \frac{m^2}{n_{\text{tot}} 4F^2(m+1)^2 \tan^2(\theta)} \eta_q T}{\sqrt{\Phi_{\text{sci}} \eta_{\text{fib}} L \frac{m^2}{n_{\text{tot}} 4F^2(m+1)^2 \tan^2(\theta)} \eta_q T + DT + N_r^2}}}. \quad (9)$$

In Eq. (9) note the dependence of the SNR on g . The CCD conversion factor is normally chosen by the manufacturer so that the bit depth spans the whole well depth. The well depth is the number of electrons that a pixel can contain. Using a small g will result in a higher SNR, albeit at the expense of a reduced CCD dynamic range. The derivation presented

Fig. 2. Experimental (symbols) and model (full lines) results SNR for the blue and green channels as a function of the integration time.

above assumes that the Čerenkov radiation is entirely subtracted from the optical signal. This can be achieved using a variety of methods.^{2–5} Most of these methods rely on a spectral analysis of the broadband optical scintillation and Čerenkov signal output from the fiber using two spectral passbands to eliminate the Čerenkov radiation, a gain coefficient being used to convert the luminosity to dose.³ Since the inverse of the SNR provides the normalized standard deviation of the signal, the normalized instrument's precision can be calculated if the calibration method is known. For spectral analysis methods to eliminate the Čerenkov radiation, the normalized variance on the dose will be equal to the sum of the normalized variances or the inverse of the SNR for each spectral channel used (blue and green in this case) to calculate the dose as shown below

$$\sigma_s \approx \sqrt{\left(\frac{1}{\text{SNR}_{\text{blue}}}\right)^2 + \left(\frac{1}{\text{SNR}_{\text{green}}}\right)^2}. \quad (10)$$

II.C. Model validation

We validated the SNR model using a PSD array employing a color interlined CCD (Alta U-2000C; Apogee Instruments, Inc., Roseville, CA) and plastic fiber scintillators. The design details and experimental beam characterization results using this system were previously published.⁵ The specifications of the Apogee CCD are available from the manufacturer.²³

We used small scintillating fibers ($1 \times 3 \text{ mm}^3$) with a peak emission wavelength of 435 nm (BCF-12, Saint-Gobain Crystals and Detectors, Inc., Newbury, OH). The optical fiber was a plastic optical fiber (ESKA, Mitsubishi, Inc., Tokyo, Japan) providing a loss of 0.2 dB/m in the wavelengths of interest. The objective lens had a focal length of 8 mm and an F -number of 1.4 (Megapixel NT58-000, Edmund Industrial Optics, Barrington, NJ). The optical photon production rate inside the scintillator is estimated to be 8000 photons/MeV of energy deposited (per BCF-12 technical specifications). Because this model does not take explicitly into account spectral effects, calculations were made based on spectral averages for each spectral passband of interest. The spectral average was estimated from the passband of the Bayer filters on the CCD. Since the elimination of unwanted Čerenkov radiation relies on a spectral filtering technique,³ two of the RGB channels (green and blue) of the color CCD were used to calculate the dose as provided by Eq. (10). The SNR achieved by the green spot was lower than that achieved by the blue spot despite the fact that there are twice as many green pixels as blue pixels on the CCD chip Bayer pattern because the scintillator used had an emission peak in the blue part of the spectrum (435 nm), which was quite distant from the green (550 nm).

The first step in the validation was to calibrate the calculated pixel value or ADU in each channel against the measured pixel value or ADU. This calibration step was necessary to account for the following unknowns of the PSD system: Objective lens attenuation, uncertainty in the value of the CCD conversion factor provided by the manufacturer,

the presence of masked rows of blocked pixels alongside the active pixels in interline transfer CCDs, the presence of a microlens on top of the active pixels, which affects the optical coupling in an unknown fashion; and the spectral passbands of the objective lens (which are not specified by the manufacturer).

The calibration images were taken of a 1 mm diameter $\times$ 3 mm length scintillator exposed to a dose rate of 400 cGy/min at the isocenter of a $10 \times 10 \text{ cm}^2$ field using a 6 MV beam of a Varian Clinac iX (Varian Medical Systems, Inc., Palo Alto, CA) operating at d_{max} , with an integration time of 10 s. The model was calibrated to provide the number of ADU measured inside the 80% of the full width at half maximum (FWHM) of a spot in the blue and green images, which means that the numerical signal calculated by the model was multiplied by a constant to reproduce the measured number of ADUs in the spot for the specified integration time. The spot magnification used was 0.11. After the calibration, we compared the model's results to the experimental results. Measurements were performed on a ^{60}Co teletherapy machine (TheraSphere; MDS Nordion, Ottawa, Canada) at the isocenter of a $10 \times 10 \text{ cm}^2$ field at d_{max} . The scintillator was 1 mm in diameter and 3 mm in length. The dose rate was 200 cGy/min. Ten images of 10 s each were acquired, and a temporal median filtering algorithm was used to eliminate the radiative noise. The median ADU and standard deviation for each blue and green spot for a radius equal to 80% FWHM of the spot was then determined. The median spot value divided by the standard deviation provided the SNR. The SNR was simulated using the calibrated model for the same experimental conditions. Note that the CCD was turned on and cooled to -20°C at least 30 min before use, thus eliminating any warm-up effects.

III. RESULTS

The simulation and experiment results are plotted in Fig. 2. As shown, the values predicted by the model closely match the experimental results. The error bars were estimated by calculating the standard deviations of the median for subsamples of the ten measurements to estimate the variability of the standard deviation and propagating the error in the ratio of the standard deviation and median.

Next, the SNR model is validated against the published experimental results for the PSD array modeled in Ref. 5. Figure 3 in Ref. 5 indicates a 0.8% precision for an integration time of 15 s in a $10 \times 10 \text{ cm}^2$ field and 10% precision for measurements in the field penumbra for 400 MU/min. The SNR model indicates 0.9% and 9.2% precision for the same conditions. The match is satisfactory as the experimental measurement precision is an estimate calculated from a series of ten samples in Ref. 5 and is thus subject to uncertainty.

III.A. System performance versus number of fibers

The model can be used to determine the system performance with varying magnification to determine how the optical coupling efficiency affects the SNR and the maximum

FIG. 3. Precision as a function of the integration time for three magnifications (0.1, 0.3, and 0.5).

number of detectors that can be imaged. We plotted the dose measurement precision as a function of the integration time for each magnification. Magnification was varied from 0.1, 0.3, and 0.5 (corresponding to spot sizes of 100, 300, and 500 μm , respectively, on the CCD), which means that a maximum of about 7744, 860, and 289 contiguous spots could be imaged on the CCD's imaging area, respectively. These numbers are based on the assumption that no space on the CCD's imaging area is wasted and ignore optical aberrations and vignetting, which affect spot size and coupling efficiency away from the system's optical axis.

The SNR is plotted as a function of integration time in Fig. 3. An integration time of 10 s was chosen as the upper limit for the CCD integration time. A 10 s upper bound seemed reasonable to obtain a measurement in clinical situations, although CCDs can integrate for hours. Figure 3 illustrates the trade-off that exists between the number of spots imaged and the precision achievable in PSD arrays. Table I lists the integration times needed to obtain $\pm 1\%$ precision as a function of the number of dose points imaged.

IV. DISCUSSION

A theoretical framework using the SNR for PSD design and performance evaluation was presented. Using Eq. (9), we formulated several recommendations to maximize the SNR for an arbitrary fiber-based PSD array (Table II). Recommendations for systems that do not use an optical fiber to

guide photons to the photodetector such as scintillator sheet and liquid scintillator systems ("non-fiber-based PSDs") are also given. For such non-fiber-based PSD systems, Eq. (5) needs to be modified to account for the fact that the emission for scintillator sheets or its liquid scintillators cannot be described by a cone because it is isotropic or Lambertian. Except for this modification, the theoretical framework developed in this work for PSD arrays can be applied to non-fiber-based PSDs. Similarly, the denominator in Eq. (9) can be modified to accommodate other types of photodetectors such as *pin* diodes or electron multiplying CCDs.

Of the points listed in Table II, points 1, 2, and 3 are the easiest to modify and are the most likely to have a large impact on the SNR. Note that when shot noise is dominant, increasing the integration time (T) or averaging multiple frames (N) will increase the SNR by a function of $T^{1/2}$ or $N^{1/2}$, respectively. When the system is not dominated by shot noise, Eq. (9) indicates that the SNR will increase more rapidly when the integration time is increased than when multiple frames are averaged (for the same total integration time).

The geometrical optics model used in this work is limited by the fact that we used the thin lens approximation to model the objective lens and we ignored lens vignetting. Vignetting will result in a nonuniform light distribution on the CCD's imaging area and will lower the light fluence received away from the optical axis. Thus, the model presented in this work tends to overestimate the number of detectors that can be imaged satisfactorily in the case when vignetting is important. Note that vignetting is highly dependent on the details of the lens construction and can be minimized through proper lens design.²⁴ Using a high-performance lens will allow the vignetting problem to be curtailed. It should be understood that the model developed in this work provides a first-order approximation for optimizing the system parameters of a PSD array. Simulations of the dosimetric precision as a function of the number of dose points imaged could be used to verify the adequacy of a given PSD array design for intensity-modulated radiotherapy quality assurance applications, for instance, where sampling a wide surface area to subcentimeter spatial resolution is desired and many hundreds of fibers must be imaged.

V. CONCLUSION

The SNR is one of the most important parameters to optimize when designing scintillation dosimetry systems. A SNR calculation model can be used to perform the following tasks: (1) Simulate the dosimetric precision of scintillation dosimetry systems and (2) plan experiments and plan measurements to optimize the integration time, number of images, field of view, and image magnification to extract the optimal SNR for a specific measurement situation.

The SNR of scintillation dosimeters can be calculated using relatively simple modeling and geometrical optics calculations, without having to actually build the instrument. This constitutes a significant advantage for the design of novel PSDs. Extending the SNR model to scintillator sheet or liq-

TABLE I. Integration time needed to obtain $\pm 1\%$ precision and the number of dose points imaged during that time.

No. of dose points	Integration time needed to obtain $\pm 1\%$ precision (s)
7744	8.1
860	0.3
289	0.1

TABLE II. Recommendations and associated actions to maximize the SNR in PSD arrays and nonfiber PSDs.

	Recommendations for PSD arrays	Recommendations for nonfiber PSDs
1	Minimize the optical fiber emission cone <i>Collimate the fiber output</i>	<i>Select a forward peaked scintillator screen</i>
2	Maximize the magnification <i>Change the object distance</i>	Same
3	Maximize the objective lens coupling efficiency <i>Minimize the objective lens F-number</i>	Same
4	Maximize the scintillator-optical fiber coupling efficiency <i>Properly polish the interfaces and match the numerical apertures of the scintillator and optical fibers</i>	NA ^a
5	Minimize the optical fiber attenuation <i>Select a high-quality plastic optical fiber and match the scintillator emission peak to the fiber attenuation minima</i>	NA ^a
6	Maximize the photodetector quantum efficiency <i>Select a backilluminated CCD and match the scintillator emission peak to the CCD quantum efficiency maxima</i>	Same
7	Increase the integration time	Same
8	Average multiple images	Same
9	Use a scintillator having a high scintillation yield possible <i>Select a short wavelength scintillator</i>	Same
10	Minimize dark noise <i>Cool the CCD down to a suitable level</i>	Same
11	Minimize readout noise <i>Select a low readout speed/Select a low readout noise CCD model</i>	Same

^aNA: Not applicable.

uid scintillator systems or to systems that use alternate photodetectors is straightforward provided that the scintillation yield and the photodetector noise terms are known.

The use of the SNR as a system performance evaluation tool also enables a systematic and analytical exploration of PSD system optimization. The SNR model developed in this work could be used by system designers to perform cost-benefit analyses for PSDs. Briefly, the cost of optimizing a certain parameter could be weighed against the benefit in terms of SNR and dosimeter precision and the most cost-efficient design could then be selected. This has the potential to accelerate the development of novel PSD systems for various clinical applications.

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