Simulation of the precision limits of plastic scintillation detectors using optimal component selection

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Purpose: The purpose of this work was threefold: First, to determine which type of charge-coupled device (CCD) would provide the best dosimetric precision for plastic scintillation detectors (PSDs); second, to design a high-photon-efficiency PSD system by optimizing its signal-to-noise ratio (SNR) using off-the-shelf technology; and third, to establish the spatial, temporal, and dose precision limits of such a PSD system. The authors have attempted to design a dosimetric tool suitable for radiotherapy treatment modalities employing small fields or fast temporal modulation of the radiation fields, and to explore the current precision limits of PSD systems.

Methods: The authors used an SNR simulation model to design and calculate the dosimetric precision of a PSD employing a fiber taper to couple the optical fiber to the photodetector. The authors also used the SNR simulation model to evaluate the impact of the photodetector performance characteristics on the SNR and to establish the spatial, temporal, and dose precision limits.

Results: The authors found that a high-photon-efficiency PSD can provide a precision of 1% in 45 μ s of integration time for a dose rate of 400 cGy/min when a single image is taken, detect a dose of 1 cGy with a detector volume of 0.0007 mm³, and image over 15 000 detectors with a precision of 1% on a 30.7 $\times$ 30.7 mm² CCD imaging area.

Conclusions: These characteristics establish that PSDs theoretically constitute a suitable dosimetric tool for radiotherapy treatment modalities employing small fields or fast temporal modulation of the radiation fields. © 2010 American Association of Physicists in Medicine.

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I. INTRODUCTION

The need for quality assurance (QA) of complex radiotherapy treatment modalities, such as intensity-modulated radiotherapy, has resulted in three trends in the development of dosimetry systems: A reduction in the size of dosimeters in order to enable high spatial resolution and minimal perturbation of the beam fluence; the development of arrayed systems for the reconstruction of quasi-two-dimensional or three-dimensional dose patterns; and the design of dosimeters capable of providing real-time dose measurements. Collectively, these trends have motivated the design, development, and validation of minimally perturbing water-equivalent plastic scintillation detectors (PSDs).^{1–3} PSDs that utilize optical fibers to guide photons to the photodetector are referred to as “fiber-based,” while those that use in-air transport are referred to as “non-fiber-based.” The main components of

PSDs, both fiber-based and non-fiber-based, are a plastic scintillator, which can be millimeter-sized or in bulk, a liquid, or a scintillating fiber; an optional optical fiber to transport scintillation photons to the photodetector for fiber-based systems; and a photodetector.

In a previous study, we demonstrated that the signal-to-noise ratio (SNR) can be used to simulate the dosimetric precision of PSDs if the signal and noise parameters of the system components are known.⁴ The signal and noise parameters are provided by the manufacturer for most off-the-shelf optical and optoelectronic components that are used to construct PSDs. When designing a high-precision PSD, it is crucial to select an appropriate photodetector because photodetectors affect both the signal and noise through the quantum detection efficiency and the dark and readout noise components. Photomultiplier tubes (PMTs) have been used in quite

a number of PSD systems.^{5–9} Recently, however, charge-coupled devices (CCDs) have gained popularity. Today, three main types of CCDs are available: Standard CCDs (hereafter referred to simply as “CCDs”), electron-multiplying CCDs (EMCCDs), and intensified CCDs (ICCDs). EMCCDs use an analog stage to multiply electrons before they are digitized to render the electronic readout noise effectively negligible compared to the signal. This amplification stage theoretically allows EMCCDs to provide a better SNR at lower light levels than CCDs. However, the amplification stage itself introduces an additional source of noise, represented by the multiplication noise factor (n_f) because of the stochastic nature of the impact ionization amplification process. The typical value of n_f in EMCCDs is ~ 1.4 .¹⁰ In ICCDs, a microchannel plate is placed in front of the CCD chip to amplify the optical signal. The typical value of n_f in ICCDs is ~ 1.6 ,¹¹ which is slightly higher than for EMCCDs. Unlike EMCCDs, ICCDs do not need to be cooled when operated with a sufficiently high gain because the dark noise is not amplified.¹²

CCDs now constitute the photodetector of choice for most PSD systems because of their low noise characteristics and their ability to simultaneously and cost-effectively image many dose points or detectors. CCDs have been used for beam characterization and IMRT QA in both fiber-based PSD systems^{2,13} and non-fiber-based PSD systems.^{14,15} EMCCDs have been used for intensity-modulated proton radiotherapy,¹⁶ proton beam characterization,¹⁷ and IMRT QA.³ To date, ICCDs have not been used in radiotherapy dosimetry applications, although they have been employed in nonradiotherapy dosimetry.¹⁸ Standalone photodiodes have also been used in a fiber-based PSD system developed for electron beam characterization.¹⁹

In this work, we used the SNR model previously developed⁴ to determine which type of CCD provides the best dosimetric precision for PSDs. Using this model, we designed a high-photon-efficiency PSD system by optimizing its SNR using off-the-shelf technology, and we further established the spatial, temporal, and dose precision limits of this system. We should note that the term “high-photon-efficiency” refers to the coupling efficiency of the entire optical train linking the scintillator to the photodetector.

II. MATERIALS AND METHODS

II.A. High-photon-efficiency PSD system design

The SNR of a fiber-based PSD can be calculated using Eq. (3.1) from Ref. 4

$$\text{SNR}_{\text{spot red,green or blue}} = \frac{\sqrt{N}\sqrt{n}\Phi_{p,\text{ave}}\eta_q T}{\sqrt{\Phi_{p,\text{ave}}\eta_q T + D_{\text{ave}}T + N_{r,\text{ave}}}}, \quad (1)$$

where the SNR is given in terms of CCD pixel values, i.e., analog-to-digital units (ADUs); n is the number of CCD pixels comprising the signal emitted by a specific fiber; N is the number of frames to be averaged; $\Phi_{p,\text{ave}}$ is the photon fluence (photons s^{-1}) per pixel; η_q is the quantum efficiency of the CCD; T is the integration time; D_{ave} is the variance of the

dark noise; and $N_{r,\text{ave}}$ is the standard deviation of the readout noise.

We assumed that the Čerenkov radiation produced inside the optical fiber was removed from the scintillation signal using a two-channel colorimetric subtraction as described by Frelin *et al.*²⁰ We should note that Eq. (1) does not explicitly take into account the dependence of various parameters, such as quantum detection efficiency, transmission losses, and coupling efficiencies, on the wavelength of the scintillation photons. We used spectral averages for each spectral band of interest in our model. These spectral bands correspond to the blue and green channels of a standard Bayer pattern filter deposited on the color CCD imaging chip. To design a high-photon-efficiency PSD, we systematically optimized the numerator component in Eq. (1) to maximize the signal and we minimized the noise terms. Both of these actions are required to maximize the SNR. A 1 mm³ scintillator volume was assumed in the design as it is suitable for small field dosimetry,²¹ which is one of the goals of our design. A 2 m long optical fiber was used to transport scintillation photons to the photodetector. This is done to allow placement of the photodetector on the treatment table, but away from the radiation beam to limit radiative contamination of the signal detected by the photodetector.²

The general PSD system design is based on the actions necessary to optimize a PSD system.⁴ The signal, which is equal to the product of the scintillator yield ($\Phi_{\text{sc int}}$) times the scintillator-to-fiber coupling efficiency (η_{fib}), the optical fiber attenuation (L), and the optical fiber-to-CCD coupling efficiency (η_{CCD}), is given by Eq. (4) of Ref. 4. The signal is maximized by optimizing the coupling efficiencies of the optical train ($\eta_{\text{fib}}, L, \eta_{\text{CCD}}$). The noise is minimized by an adequate selection of the photodetector. A fiber taper is used to couple the optical fiber to the photodetector in order to maximize the coupling efficiency. Previous research in the field of digital radiography has shown that the coupling efficiency to the CCD is dominant in the overall optical train gain and that a fiber taper provides a higher coupling efficiency from the scintillator screen to the CCD than an objective lens.^{22,23}

When using a fiber taper in fiber-based PSDs, the limiting factor to the overall system gain is the scintillator-optical fiber coupling efficiency η_{fib} . We should note that η_{fib} is limited by the finite acceptance cone of the optical fiber and the isotropic nature of the scintillation emission. The scintillator-optical fiber coupling efficiency is 3.4% for typical round single-clad optical fibers and 5.6% for round multicladd optical fibers.²⁴ This limit on the overall system gain could be decreased with the use of a lens to couple the scintillator and optical fiber,²⁵ which would have the unwanted consequence of decreasing the water equivalence of the PSD as no water-equivalent plastic coupling lenses are commercially available. The general design is as follows:

- A 1 mm³ blue scintillating optical fiber (BCF-12, Bicon, Inc., Hiram, OH) is used.
- The scintillating fiber is glued to a plastic optical fiber (ESKA, Mitsubishi, Inc., Tokyo, Japan) with a diameter

TABLE I. Technical specifications for the Apogee Alta U3041 CCD, the Andor iXon^{EM}+860 EMCCD, and the Andor iStar 720 Gen III ICCD.

Specification	Alta U3041 CCD	iXon ^{EM} +860 EMCCD	iStar 720 Gen III ICCD
Pixel size	15 μm	24 μm	26 μm
Number of pixels	2048 $\times$ 2048	128 $\times$ 128	1024 $\times$ 256
Imaging area	30.7 $\times$ 30.7 mm^2	3.1 $\times$ 3.1 mm^2	25 $\times$ 6.7 mm^2
CCD quantum efficiency (η_q at 435 nm)	92%	75%	15%
Readout noise (N_r^2)	8 e ⁻ RMS at 0.7 MHz	48 e ⁻ RMS at 10 MHz; 18 e ⁻ RMS at 1 MHz	8 e ⁻ RMS at 0.031 MHz
Dark current (D , typical)	2 e ⁻ /pixel/s (-20°C)	0.002 e ⁻ /pixel/s (-85°C)	...
Gain	1	1–1000	1000
Relative cost	+	++	+++

of 1 mm and a length of 2 m, which allows the CCD to be placed distant from the radiation beam.

- The plastic optical fiber is glued to a low (1:1) or high (4:1) magnification ratio fiber taper that is bonded to the CCD imaging chip. The high magnification ratio fiber taper (4:1) is used to calculate the maximum number of detectors that can be imaged (see Sec. III).

II.B. Photodetector comparison

The SNR of CCDs and EMCCDs can be calculated by dividing the signal by the shot, dark, and readout noise variances, using the following equation:

$$\text{SNR} = \frac{\Phi_p \eta_q T}{\sqrt{n_f^2 (\Phi_p \eta_q T + DT) + \frac{N_r^2}{G^2}}}, \quad (2)$$

where Φ_p is the photon fluence (photons $\text{pixel}^{-1} \text{s}^{-1}$) on a pixel; η_q is the quantum efficiency; T is the integration time for one image in seconds; n_f is the multiplication noise factor; D is the variance of the dark noise (electrons); N_r^2 is the variance of the readout noise (electrons); and G is the electron multiplication gain. For CCDs, $n_f=1$ and $G=1$. For EMCCDs, $n_f \cong 1.4$ and $G=1-1000$. From Eq. (2), it is evident that the following characteristics are desirable for the photodetector: (1) High quantum efficiency and (2) low dark and readout noises. Back-illuminated photodetectors possess a much higher quantum efficiency than conventional front-

illuminated devices ($>80\%$); therefore, they are a logical choice for scintillation dosimetry.

For ICCDs, Eq. (2) must be modified as follows:

$$\text{SNR} = \frac{\Phi_p \eta_q T}{\sqrt{n_f^2 (\Phi_p \eta_q T) + \frac{DT}{G^2} + \frac{N_r^2}{G^2}}}. \quad (3)$$

For ICCDs, $n_f \cong 1.6$ and $G \cong 1000$.

Note that Eqs. (2) and (3) are well established in such fields as astronomy and photometry and have been experimentally validated. For examples, see Refs. 11 and 26. Table I shows the technical specifications for the high quantum efficiency back-illuminated CCD, the EMCCD, and the ICCD selected for the high-photon-efficiency PSD design. Both the CCD and EMCCD are back-illuminated CCD and EMCCD models to maximize the quantum efficiency. The cameras that were selected correspond to the criteria described in Ref. 4 as well as the criteria outlined in Tables II and III. In addition, they were found to be among the top commercial models of their type in terms of quantum efficiency (η_q), dark noise (D), and readout noise N_r^2 . We must emphasize that these CCDs were chosen for illustration purposes only; thus, our selection does not constitute a recommendation or an endorsement of these products. We could equally have selected photodetectors produced by different manufacturers. Note that no attempt was made to optimize the design in terms of cost.

TABLE II. Signal parameters.

Parameter	Optimal selection	Value
Scintillation yield ($\Phi_{\text{sc int}}$)	Select a blue (435 nm) or green (550 nm) scintillator. Match the numerical apertures of the scintillator and the optical fiber; use multiclad fibers; finely polish all interface surfaces; use an index matching glue.	8000 photons/MeV
Scintillator-to-fiber coupling efficiency (η_{fib})	Select a high-quality optical fiber; minimize fiber length by placing PSD on the treatment table.	5.6%
Optical fiber attenuation (L)	Maximize by using a fiber taper to couple the optical fiber to the CCD.	0.2 dB/m
Optical fiber to CCD coupling efficiency (η_{CCD})	Use a back-illuminated CCD for maximum sensitivity.	85% (for 1:1 taper)
CCD quantum efficiency (η_q)	Application-dependent.	$\sim 80\%$
CCD integration time (T)		N/A

TABLE III. Noise parameters.

Parameter	Optimal selection	Value
See Table II	All parameters that affect the signal also affect the shot noise. Maximizing the signal also maximizes the shot noise.	See Table II
Dark noise (D)	Dark noise can be rendered negligible by strongly cooling the CCD.	$0.1 \text{ e}^-/\text{pixel/s}$
Readout noise (N_r^2)	Select a low readout noise CCD; select a slow readout speed.	$7\text{--}50 \text{ e}^- \text{ RMS}$

We used Eqs. (2) and (3) to calculate the SNR performance of the Alta U3041 CCD (Apogee Instruments, Inc., Roseville, CA), the iXon^{EM}+860 EMCCD (Andor Technology, Belfast, Northern Ireland), and the iStar 720 Gen III ICCD (Andor Technology). SNR calculations were performed using an optical photon fluence ranging from 0.1 to 100 000 photons per pixel on the CCD imaging area. Because the pixel areas vary between these devices (i.e., $225 \mu\text{m}^2$ for the Alta U3041 CCD, $576 \mu\text{m}^2$ for the iXon EMCCD, and $676 \mu\text{m}^2$ for the iStar ICCD), we normalized the photon fluence to the pixel size of the Alta U3041 CCD. We varied the electronic gain of the EMCCD from 1 to 1000. To determine the effect of the EMCCD's readout speed on its performance we simulated two readout speeds and associated readout noises ($18 \text{ e}^- \text{ RMS}$ at 1 MHz and $48 \text{ e}^- \text{ RMS}$ at 10 MHz). The SNR provided by an "ideal" purely shot noise-limited photodetector was used for comparison, which was quantified in terms of ADU. The integration time in all simulations was set equal to an integration time of 1 s.

III.C. Performance of high-photon-efficiency PSD system

We determined the performance limits of the high-photon-efficiency PSD system designed in Sec. II A by calculating (1) the maximum spatial resolution; (2) the minimum integration time; (3) the minimum dose; and (4) the maximum number of dose points that can be resolved assuming fixed precisions of 0.1%, 1%, and 2% and a dose rate of 400 cGy/min. Precision was defined as one standard deviation from the mean, and a precision worse than 2% was assumed to be unacceptable for clinical measurements. A low magnification ratio fiber taper (1:1) was used to couple the transport optical fiber to the CCD imaging chip in order to obtain the maximum optical gain to calculate the maximum spatial resolution, minimum integration time, and minimum dose. A high magnification ratio fiber taper (4:1) was used to calculate the maximum number of dose points that could be imaged assuming fixed precisions of 0.1%, 1%, and 2% and a dose rate of 400 cGy/min.

III. RESULTS

III.A. High-photon-efficiency PSD system design

III.A.1. Maximizing the signal

Table II provides the optimal parameter selections to maximize the signal. Additional information on the listed

parameters can be found in Ref. 4. The choice of scintillator wavelength was made to maximize the scintillation yield. Alternatively, were a CCD to possess a low quantum efficiency at the peak of the scintillator scintillation yield, the scintillator wavelength could be selected to match the CCD peak.

III.A.2. Minimizing the noise

Table III provides the optimal parameter selections to minimize the noise. Note that using a slow readout speed CCD as suggested to reduce the readout noise will also lead to long CCD deadtime.

III.B. Photodetector comparison

Figure 1 shows the SNR as a function of the photon fluence per pixel for a shot noise-limited photodetector, the CCD, the EMCCD at two gain levels (1 and 1000), and the ICCD. The EMCCD at high gain (1000) provides a better SNR than the CCD for photon fluences up to around 60 photons $\text{s}^{-1} \text{ pixel}^{-1}$. However, the CCD provides a better SNR for photon fluences greater than 60 photons $\text{s}^{-1} \text{ pixel}^{-1}$. Since the dosimetric precision is related to the SNR,⁴ it is possible to conclude that CCDs will provide better dosimetric precision once the photon fluence exceeds 60 photons $\text{s}^{-1} \text{ pixel}^{-1}$.

For all photon fluences, the EMCCD at low gain or at gain equal to unity performs notably worse than the CCD.

FIG. 1. SNR as a function of photon fluence per pixel for a shot noise-limited photodetector, a CCD, an ICCD, and an EMCCD at two gain levels (1 and 1000).

FIG. 2. Number of images that need to be averaged as a function of the integration time per image in order to attain a 0.1%, 1%, and 2% precision for a CCD and an EMCCD at high gain (1000).

There is thus little rationale to use an EMCCD at low gain settings. The reason for the relative poor performance of EMCCDs is the multiplication noise factor n_f , which increases the absolute value of shot noise, and is the dominant source of noise at high photon fluences. We also found that, for all photon fluences, the ICCD performs slightly worse than the EMCCD at high gain. Again, the reason for this performance difference is that the multiplication noise factor n_f is higher for the ICCD than for the EMCCD.

III.C. Performance of high-photon-efficiency PSD system

III.C.1. Minimum integration time

Figure 2 shows the number of images that need to be averaged as a function of the integration time per image to attain precisions of 0.1%, 1%, and 2% for the CCD and the EMCCD. The ICCD was not simulated because it performs slightly worse than the EMCCD at high gain, and is therefore of little interest in scintillation dosimetry. The curve representing the performance of the EMCCD at high gain can be used as a surrogate for the performance of the ICCD. Figure 2 can be understood as follows: If 100 images are taken per measurement, then the shortest possible integration time of 1 μ s per image will be achieved by using an EMCCD at high gain. Since 100 images need to be taken per measurement, the total integration time is $100 \times 1 \mu\text{s} = 0.1 \text{ ms}$, ignoring CCD dead time. Note that this is less than the linac pulse repetition rate on most clinical linear accelerators, thus opening the door to real-time, per pulse characterization of the radiation beam. The maximum precision that can be achieved with an integration time per image of 1 μ s is 2%. Switching from the EMCCD to the CCD for the same precision and number of images slightly increases the integration time per image from 1 to 1.7 μ s. However, if only one image is taken per measurement, then the integration time per image assuming a precision of 2% would be 0.1 ms with an EMCCD and 0.019 ms with a CCD. The need for precisions

FIG. 3. Number of images that need to be averaged as a function of the dose in order to attain a 0.1%, 1%, and 2% precision using a CCD an EMCCD at high gain (1000).

better than 2% would increase the integration time per image if the number of images averaged were kept constant.

As shown in Fig. 2, the CCD provides a shorter integration time than the EMCCD in most cases, except for a precision of 2% when 100 and 50 images are averaged. In these cases, the EMCCD at high gain slightly outperforms the CCD, which is consistent with our finding that the EMCCD outperforms the CCD at low photon fluences.

III.C.2. Minimum dose

Figure 3 shows the number of images that need to be averaged as a function of the absorbed dose to attain precisions of 0.1%, 1%, and 2% for the CCD and the EMCCD at high gain. The integration time per image is 1 s. Figure 3 can be understood as follows: If 100 images are taken per measurement, then the smallest possible dose of 7 μ Gy will be measured by using an EMCCD at high gain. Since 100 images need to be taken, the total integration time is $100 \times 1 \text{ s} = 100 \text{ s}$, ignoring CCD dead time. The maximum precision that can be achieved when measuring 7 μ Gy is 2%. Switching from the EMCCD to the CCD for the same precision and number of images averaged slightly increases the dose that can be measured from 7 to 10 μ Gy. If only one image is taken per measurement, then the minimum dose that can be measured with a precision of 2% is 0.08 cGy with the EMCCD and 0.013 cGy with the CCD. The need for precisions better than 2% would increase the minimum dose that could be measured if the number of images averaged were kept constant.

III.C.3. Minimum detector volume

We calculated the minimum dose that can be measured in Fig. 3 for a 1 mm³ scintillator. Because the number of scintillation photons produced per unit of absorbed dose is proportional to the scintillator volume, based on Fig. 3, the EMCCD at high gain can measure a dose of 7 μ Gy with a precision of 2% when 100 images are averaged. Therefore,

to measure a dose of 1 cGy under the same conditions, the scintillator volume can be reduced to 0.0007 mm^3 while keeping the same precision. If only one image were to be taken under the same conditions, then the scintillator volume would be 0.08 mm^3 .

III.C.4. Maximum number of detectors

Using the same high-photon-efficiency PSD design described above (see Sec. II A) with a high magnification ratio fiber taper (4:1), we calculated that the EMCCD-based PSD can image 153 detectors on its $3.1 \times 3.1 \text{ mm}^2$ imaging area, while the CCD-based PSD can image 15 079 detectors on its $30.7 \times 30.7 \text{ mm}^2$ imaging area, assuming a fill factor of 100%. This is equivalent to having one dose detector every 3 mm over a $40 \times 40 \text{ cm}^2$ radiation field.

IV. DISCUSSION

We found that the CCD performed better than the EMCCD and ICCD in most clinically relevant measurement situations for a high-photon-efficiency PSD system. ICCDs seem to be ill-suited for radiotherapy scintillation dosimetry applications for the following reasons: (1) They provide a slightly worse SNR performance compared to EMCCDs used at high gain for the same quantum efficiency because ICCDs have a higher multiplication noise factor than EMCCDs, which increases the absolute value of shot noise; (2) they are significantly more expensive than CCDs and EMCCDs; and (3) their strengths, such as their picosecond gating capability and limited cooling requirements, are neither useful to scintillation dosimetry nor advantageous over those of competing technologies. Figure 1 shows that a CCD will outperform an EMCCD assuming that the photon fluence per pixel is sufficient. This is the case in a high-photon-efficiency PSD system, where the optical train gain is optimized. The presence of the electron multiplication stage in EMCCDs and the associated multiplication noise factor (n_f) decrease the maximum SNR achievable compared to a CCD.

EMCCDs and ICCDs were designed for photon-starved applications, such as low-light level microscopy, which is not the situation when scintillation dosimetry is performed with a PSD system using a fiber taper for coupling. The situation might arise, however, when an objective lens is used to couple the optical fiber to the CCD. Objective lenses provide coupling efficiencies that are 100 or 1000 times smaller than a fiber taper, depending on the magnification and lens f -number.^{4,22,23} From Fig. 3, it can be seen that the regime where an EMCCD outperforms a CCD using our system design is pushed from the measurement of doses of around $1 \text{ } \mu\text{Gy}$ to 0.01–0.1 cGy when an objective lens is used for coupling. In addition, if a PSD system were to use scintillator volumes of less than 1 mm^3 , such as might be the case to allow insertion of the scintillator and optical fiber into catheters for *in vivo* measurements, then the point where EMCCDs outperform CCDs would also be affected. Since there is a direct relation between the scintillator volume and the number of photons produced, we can calculate that using a 0.25 mm diameter and 1 mm long scintillator generates 20

times fewer photons than when using a 1 mm^3 scintillator. In this case, EMCCDs will outperform CCDs when measuring doses of 0.2–2 cGy. These points highlight the paramount importance of the question of the optical train gain in scintillation dosimetry. The optimization of the optical system is one of the most important tasks of the PSD system designer.

One issue that arises with using EMCCDs in scintillation dosimetry is gain aging: The gain tends to decrease as a function of time in a nonlinear fashion. Although EMCCDs are conditioned at the factory to eliminate the initial, rapid gain variation, the gain may still vary by a non-negligible amount (10%) over a day of continuous operation.^{26,27}

We must point out that the integration time calculated in Fig. 2 ignores the actual frame rates that can be achieved by the electronic circuits of the detectors. While the iXon EMCCD can operate at high speed (i.e., it can operate at 500 frames/s with 2 ms of integration time), the Alta U3041 CCD cannot operate at such high speed (i.e., it can only operate at 0.14 frames/s with 33 ms of integration time). Thus, frame averaging with the Alta U3041 CCD cannot be performed without incurring an unacceptably long dead time. This dramatic decrease in the frame rate from the EMCCD to the CCD is due partly to the fact that, for photodetectors, the frame rate decreases as the number of pixels increases. Because the Alta U3041 CCD has ~ 256 times the number of pixels of the iXon EMCCD ($4\,194\,304$ vs $16\,384$), the CCD has a considerably slower frame rate than the EMCCD.

Scintillation dosimetry is not an imaging application; therefore, image quality does not constitute a performance metric *per se*. The number of pixels provided by current photodetectors is needlessly large for scintillation dosimetry applications. Ideally, custom-built CCDs would be designed with a smaller number of larger pixels. Each optical spot in the CCD imaging area needs to be sampled only by a few pixels to allow the colorimetric Čerenkov subtraction to be performed. Frame rates for CCDs could be dramatically increased by using such a custom-built photodetector for PSD systems.

V. CONCLUSION

This work has presented the design of a high-photon-efficiency PSD constructed for the measurement of small fields or fast temporal modulation of the radiation fields and highlights the importance of optimizing the optical train coupling efficiency in order to achieve the best possible dosimetric precision in PSD systems.

Our results show that CCDs will provide a superior performance compared to EMCCDs for the majority of dosimetry tasks in radiotherapy when using a high-photon-efficiency PSD system. The main reason for this is the presence of the multiplication noise factor (n_f) in EMCCDs, which constitutes a significant source of additional noise compared to CCDs when used at moderate photon fluxes.

The performance that can theoretically be attained with a high-photon-efficiency PSD is extremely high. Real-time, high-precision measurements of very small doses

(<1 cGy) or high spatial resolution measurements (<1 mm³) are achievable, and therefore PSDs can be used to characterize the beams generated by advanced radio-therapy treatment modalities.

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