

Scientific Paper

High resolution 2D plastic scintillator detectors for radiotherapy departments

Grzegorz Zbigniew WOŹNIAK^{1,2,ABCD,*}, Beata KOZŁOWSKA^{1,CDE}

¹*Institute of Physics, University of Silesia in Katowice, Poland*

²*Zakład Radioterapii, Katowickie Centrum Onkologii, Poland*

*Corresponding author: Grzegorz Woźniak; gwozniak@interia.pl

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Abstract

Introduction: Plastic scintillation detectors (PSD) have been developed for over four decades and are widely used in a variety of fields, but one can find relatively few reports of their clinical use compared to other dosimetric solutions.

Material and methods: The inexpensive detector setup made of a Saint-Gobain BC-400 plastic scintillator and commercially available on the market CMOS-based DSLR Pentax camera was investigated. Build PSD detectors were irradiated with 6, 10 and 15 MV flattening filtered (FF) and 6 and 10 MV flattening filter free (FFF) photon beams using a clinical linear accelerator. Data were processed using Matlab software to remove background and artefacts. A comparison of the spatial resolution parameters to the Gafchromic EBT3 films was performed.

Results: Average dose difference between TPS and PSD was 1.1%. The measured spatial resolution was 0.29 mm, and it differed from the film by 1.1%. MTF₅₀ for PSD was 0.57 mm higher than the Gafchromic film. Signal to dose fit function with an R-square equal to 0.999 was established. The standard deviation of mean pixels value for a series of measurements was below 0.1%, for variable dose rate dependence was below 0.6% and for different energies 1.1%.

It was demonstrated that such a setup allows a satisfactory signal-to-dose dependence and provides high spatial resolution at an affordable price compared to a 2D ion chamber or a diode detector array. Moreover, PSDs are reusable and provide a simple readout compared to Gafchromic films commonly used in radiotherapy departments.

Conclusions: Variable parameters of the camera allow to select signal values at the optimal level. The system presented excellent signal stability, high image resolution and a simple signal-to-dose relationship which encourages further work to investigate PSDs for use in radiation therapy departments.

Keywords: PSD; dosimetry; radiotherapy; plastic scintillation detectors; Gafchromic films.

Introduction

A radiation dosimeter must possess a number of characteristics starting from at least one physical property as a function of the dosimetric quantity being measured.¹ Plastic scintillation detectors (PSD) appear to be very promising regarding measurements of high energy beams in medical linear accelerators.^{2,3} Plastic scintillators can be formed in an assortment of shapes like sheets, tubes or fibers with dimensions ranging from 0.1 mm up to tenths of a centimeter. A number of publications show a wide range of applications and benefits of PSD, but clinical application and availability of commercial systems are low.^{4,5} One of the most important features of plastic scintillators is their electron density which is very close to water, 3.37 vs 3.35×10^{23} electrons/cm³. Hence, PSD can be treated as water equivalent. This makes the dose-to-medium/dose-to-water ratio close to unity, making the detector insensitive to changes

in the radiation energy spectrum. This, in turn, makes it the ideal detector for use in a radiotherapy department.⁶

On the other hand, an important issue in the use of PSD for dosimetry is the elimination of the unwanted light signal coming from the Cherenkov radiation produced by both the scintillator and the commonly used optical fibers. Cherenkov light can exceed the scintillation signal in some dosimeter geometries.^{2,3,7,8} At least five methods of reducing or eliminating the effect of Cherenkov radiation on the signal have been described in the literature.²⁻⁸⁻¹¹ Light guide-free solution proposed by Lambert et al.¹¹ looked promising for 2D and 3D applications, but their resolution was relatively low compared to the Gafchromic films. Previous works cited above focused on demonstrating the advantages of a particular solution, such as different ways of eliminating Cherenkov radiation, camera image correction or signal processing. In this study, the authors combined these methods and investigated the use of a thin scintillation sheet together with the commercially available

Complementary Metal-Oxide-Semiconductor (CMOS) camera and common computer software. The signal from the scintillator was measured as a single image by the camera without the use of optical fibers, so Cherenkov light is not expected to be generated in the air, as its refractive index is very close to 1. The use of an RGB colour camera can also work as a second method of Cherenkov radiation elimination by spectral differences or constant relation between scintillation and the Cherenkov spectrum.¹⁰ In addition, the use of different camera settings such as ISO, aperture or focal length allowed the sensitivity of the system to be adapted to the measurement conditions while maintaining high sensitivity. The single-shot approach allowed us to minimize the electronic shot noise.

With the development of radiotherapy, more sophisticated treatment techniques are being used to increase treatment conformality and allow the dose escalation to target and/or dose reduction to organs-at-risk (OAR). Accurate dose delivery is critical. The most commonly used techniques use irregular field shapes, often containing segments whose width is the width of a leaf and the length of a centimeter. Quality assurance of such treatment plans is challenging, especially with respect to spatial resolution. The highest spatial resolution of 0.05–0.5 mm/pixel, typically 0.3 mm/pixel after processing, is provided by film dosimetry, but due to its characteristics such as disposability, lack of online readout or the need for calibration of each batch, make it inconvenient on for day-to-day use. For this reason, despite the lower spatial resolution of several millimeters, dosimetry arrays with ionization chambers or solid-state detectors are more commonly used.

A novel, high-resolution, inexpensive detector setup made of a plastic scintillator sheet and a CMOS-based digital single lens reflex (DSLR) camera was chosen for this study. It would provide high spatial resolution and good dose dependence at an affordable price compared to a 2D ion chamber or diode detector array while still being reusable and providing easy readout compared to Gafchromic films.⁶ The following measurements were performed to assess held PSD: accuracy, precision, linearity, dose rate dependence, energy dependence, spatial resolution, physical size, readout convenience and convenience of use. Automated image processing according to accepted standards like de-noise, rotation correction, image scaling, vignetting and tilt was done using MATLAB (MathWorks, USA) scripts. Measurements were made to estimate the influence of the background in relation to measurement time, camera sensitivity, number of monitoring units, field size and the amount of Cherenkov radiation generated in the scintillation material. The system precision for repeatable measurements and response as a function of deposited dose and dose rate were then measured. The feasibility of constructing a useful high-resolution dosimeter using inexpensive and commonly available elements was investigated. In particular, the authors determined the extent to which this approach could be adopted for different dose registration by adjusting the camera settings.

Materials and Methods

The experimental system setup, whose main components were the PSD and the CMOS camera, was built using commercially available and inexpensive components. The experiments were performed in an oncology center on a medical linear accelerator. Initial measurements were taken to characterize the background signal being a source of artefacts, followed by a series of tests to determine the suitability of the system as an ionising radiation detector. Automated data processing procedures were established. The following chapter describes the system setup, data processing and measurements performed.

System setup and characteristics

Experiments were carried out using the Saint Gobain BC-400 general-purpose plastic scintillator shaped into 10 cm square sheets of 1 mm nominal thickness. The thickness measured with a micrometer was 0.97 with a variation of ± 0.01 mm, corresponding to the tool resolution. The detector's density was equal to 1.023 g/cm^3 , the maximum spectral emission at 423 nm (blue light), decay time 2.4 ns, light output of 65% of anthracene, refractive index of 1.58, and electron density 3.37×10^{23} per cm^3 . The readout system consisted of DSLR CMOS-based Pentax K-x camera with 12.4 megapixels (4288×2848 pixels) equipped with a 55–300 mm telephoto lens and an acrylic lens. The camera stores images with a signal resolution of 16 bits, which means that the maximum value of the recorded signal is $2^{16} = 65\,536$. All results obtained, especially for background measurements, should be related to this value as the maximum possible value. The PSDs were irradiated with 6, 10 and 15 MV flattening filtered (FF) and 6 and 10 MV flattening filter free (FFF) photon beams using the Elekta Versa HD linear accelerator (linac).

The plastic scintillator was placed in the isocentre in a table plane. Using the table extension frame, PMMA build-up plates were placed on top of the PSD. An acrylic mirror was mounted directly under the detector. The acrylic mirror is safer to handle and has lower UV absorption than mineral glass. The camera was placed perpendicular to the beam axis about 1 meter from the mirror and shielded with steel blocks. The camera and the mirror were adjusted using the linac laser system, to point at the centre of the scintillator detector. The telephoto was set to cover twice the size of the detector. All light sources in the bunker were turned off or covered up. The source-to-surface distance (SSD) was set to 98 cm, and PMMA build-up was 2 cm. Unless otherwise stated, a $15 \times 15 \text{ cm}^2$ field was used in the experiment. The measurement setup is shown in **Figure 1**.

Since PSD measurements are relative, the number of monitor units (MU) was used as the dose in the experiments instead of the dose in gray (Gy). The linear accelerator photon energies were calibrated so that the delivery of 1 MU corresponded to a dose of 1 cGy for the field size $10 \times 10 \text{ cm}^2$ in water at a depth of 10 cm at the SSD equal to 90 cm.

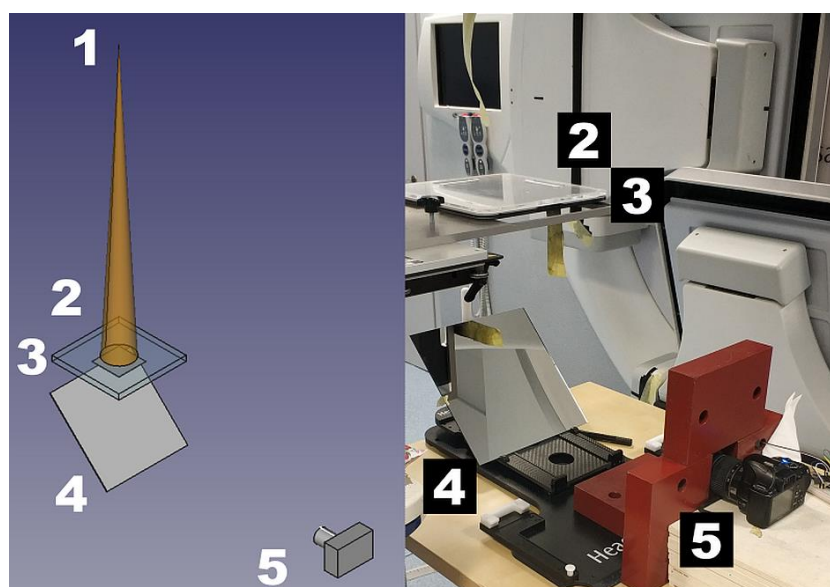


Figure 1. The system setup at the linac. Main elements on left schematic view: beam (1), build-up (2), PSD (3), mirror (4) and camera (5). The beam (1) is not visible in the right picture.

Data processing

Every single exposure was saved as a separate image stored in the Pentax Digital Camera Raw Image format (PEF), which is owned by the camera manufacturer Pentax. The images were batch converted using the Digital Camera Utility 5 software (Ricoh Imaging Company, Japan). The uncompressed 48-bit TIFF format was chosen to allow further processing without loss of image quality. Image post-processing was performed using the MATLAB environment. The images had different exposure times, so the background was proportionally corrected for the time. A selected region of interest (ROI) was copied across all measurements in one series. Batch processing was used to avoid human bias. The mean and the standard deviation of the ROI were then calculated.

The measured signal is stored by the camera in a single image in three colour channels. The BC-400 plastic scintillator used for the measurements has a maximum spectral emission at 423 nm, corresponding to blue light. Signal and background measurements were performed to select the optimal colour channel(s) for further analysis and to evaluate whether electronic noise is channel independent and whether the camera sensor registers the background signal uniformly over all channels.

Lens-derived correction was performed by the camera's built-in software. Stray radiation creates 'salt and pepper' artefacts, which were effectively reduced using a median filter.⁶ The multi-level filter script was developed to remove noise without causing distortion or blurring. The in-house automated script had a filter and corrected images for vignetting, tilt, and rotation as well as filtering and average background signal removal from electronic noise and stray radiation before further analysis.

Background and artefacts

Collected images contain background signal and artefacts. A background signal that comes from ambient light, unshielded light sources, stray radiation and electronic noise from a camera sensor is collectively named sensor-derived artefacts. Other artefacts can be divided into lens-derived and scintillator-derived.¹²

All light sources in the bunker were turned off or covered to prevent the collection of any light other than light that came from scintillations. The electronic noise related to the camera sensor and the exposure time was measured with exposure times between 10 and 90 seconds at ISO 100, 200 and 400. The measurements were carried out in the bunker with the same geometry as the following measurements in order to take into account any background signal or residual light in the bunker.

Stray radiation can directly affect the camera sensor and cause additional noise. The dependence of the stray radiation signal on the delivered dose was measured. Measurements were made with the scintillator detector removed from the setup, delivering 10, 25, 50, 100, 150, 200, 300, 400, 500, 750 and 1000 MU with 6 MV photon beam and a field size $15 \times 15 \text{ cm}^2$.

The linac output factor as a function of the field size has a phantom and a head scatter component. It can be expected that the scatter signal will depend on the field size used. Five different field sizes were tested: 3×3 , 5×5 , 10×10 , 15×15 and $20 \times 20 \text{ cm}^2$. The dose delivered was 100 MU for each test. The camera was set to ISO 100, 200 and 400, respectively. The aperture F was set at 4.5.

Lens-derived artefacts are known as distortion and vignetting. The camera's built-in firmware corrected distortion, but did not handle vignetting properly. Vignetting caused the signal value to decrease as the distance from the centre of the image

increased. To avoid this effect, the signal was measured from a homogeneous light field and the correction function:

$$\frac{f l^4}{(f l^2 + r^2)^2}$$

was fitted with the variable $f l$, where $f l$ is the focal length and r is the radius from the image centre.

Scintillator-derived artefacts cause image blur and can be described by the point spread function (PSF). The standard method for PSF evaluation is to calculate the modulation transfer function (MTF) at the edge of a sharp object. For films and PSD detectors, the penumbra of the linac is larger than the blur induced by the detector. Therefore, the MTF reflects both the size of the penumbra and the blur of the detector. Assuming that a constant penumbra difference in the MTF corresponds to the difference in spatial resolution between films and PSD, the MTFs were calculated at the edge of the $5 \times 5 \text{ cm}^2$ field in the 'Y' direction in the IEC 61217 coordinate system, since the penumbra created by the collimator jaws is smaller than the penumbra created by the leaves.

Detector's characteristics

Radiation detectors are characterised by a number of parameters. The following tests were carried out to investigate the system: precision, dose rate dependence, dose response, camera aperture and ISO dependence, energy dependence and spatial resolution. Since all experiments were performed by one person with one set, the precision is evaluated for a series of repeated measurements. Most of the tests were performed with 6 MV FF beams. In general, the measured signal depends on the delivered dose, geometry, energy, dose rate, irradiation time, aperture, ISO, field size area, background signal and artefacts, and camera and optics settings. The measurement result is relative and must be multiplied by a calibration factor to convert the relative dose to the absolute dose. Ideally, for a measured setup and with both geometry and focal length fixed, the dose should not depend on any parameters other than the signal and a camera setting, i.e. ISO and aperture, and should be in the form of:

$$D = f(g(S)) \cdot P_{ISO} \cdot P_{f l} \cdot P_F \cdot D_{cf} \quad \text{Eq. 1}$$

where S is the sensor signal corrected for background and de-noised, function g is the image correction function for the lens- and sensor-derived artefacts, function f is the signal-to-dose conversion, P_{ISO} is the correction factor for different ISO settings, $P_{f l}$ is the correction factor for different focal lengths, P_F is the correction factor for different apertures F and D_{cf} is the signal-to-dose coefficient. In this paper, we have investigated the effects of these factors except $P_{f l}$ on the dose and, if a relationship exists, described it.

System repeatability defines the stability of the result obtained in a series of measurements. In the test, the PSD was irradiated 10 times in a row under the same conditions. A $15 \times 15 \text{ cm}^2$ beam was used to irradiate the entire detector uniformly. For signal calculation, a $5 \times 5 \text{ cm}^2$ square in the central part of the scintillation was selected as a region of interest (ROI). For each

image acquired, the mean signal and the standard deviation within the ROI were calculated. The system was tested at ISO 200 and 400, respectively, and F 4.5.

The detector system should be dose-rate independent. This is essential when verifying plans with a variable dose rate. The linac in service mode has limited capability to adjust the dose rate continuously, but it can do so in steps. In order to cover as wide a range of dose rates as possible, five measurements were performed for 6 MV FF and 10 MV FFF beams, each delivering 100 MU. 6 MV FF has the lowest and 10 MV FFF have the highest dose rate on the linac. 6 MV FF beams with a dose rate of 30 to 500 MU/min and 10 MV FFF beams with 140 to 2100 MU/min were used.

One of the most important characteristics of a radiation detector is the dependence of one of its parameters on the administered radiation dose. A series of measurements were performed to assess the dependence of the system response on the dose administered.

The signal was measured in a ROI selected in the central part of the scintillator as a square with a side equal to half of the side of the scintillator, i.e. $5 \times 5 \text{ cm}^2$. The mean signal and the standard deviation of the pixel values were determined for this area. The dose administered varied from 10 to 1000 MU at the full dose rate (500 MU/min). A $15 \times 15 \text{ cm}^2$ beam was used to uniformly cover the plastic scintillator.

Using the LAB fit Curve Fitting Software (Universidade Federal de Campina Grande, Brazil), more than 200 functions with 2, 3 and 4 parameters were tested to fit the collected data, including fitting functions for the Gafchromic films published in the literature. Curves with more parameters should have a better fit, but they require more calibration points. Therefore, it is important to select a calibration curve with as few parameters as possible. It was assumed that a 3-parameter curve with a constant parameter for background compensation should be used.

The camera aperture, denoted as F , is responsible for the amount of light entering the camera and the depth of the field. By changing F it is possible to adjust the amount of light collected and to ensure that measurements are made under optimum conditions for the CMOS sensor, in particular to avoid saturation of the sensor. A series of images were taken for F ranging from 5.6 to 22 to investigate the dependence of the aperture on the signal.

Unlike the Gafchromic film, the electronic sensor of the DSLR camera has variable sensitivity via ISO settings. Increasing the ISO will increase the sensitivity of the chip, but it will also increase the current in the sensor and generate heat which can increase the noise and degrade the stability of the readout system. Measurements were taken at ISO 100, 200, 400 and 800 to assess the effect on the stability and electronic background of the system.

Image series were acquired for the following photon energies: 6, 10 and 15 MV FF beams and 6 and 10 MV FFF beams, respectively, delivering 100 MU for each energy. This corresponded to a slightly different dose due to the MU/cGy calibration, so differences in responses were expected. The reference dose was calculated in the previously commissioned Monaco (Elekta AB, Sweden) treatment planning system (TPS) using the Monte Carlo algorithm with a standard deviation of 0.5%. The collected signal was recalculated to the relative dose. D_{cf} calibration factor, which maps the relative dose to the absolute dose, was determined by numerical optimisation. It allowed the signal to be converted to dose in cGy.

High spatial resolution in dosimetry is required for quality assurance of sophisticated radiotherapy treatments such as Stereotactic Radiation Therapy (SRS). A series of measurements for different sizes of treatment fields were performed using the developed PSD system. Results were analyzed using the developed scripts written in MATLAB. The collimator of the linear accelerator was set to produce square fields with edge lengths ranging from 0.5 to 7 cm. The results were compared both with the set field size and with the results of the dosimetric Gafchromic EBT3 films and processed in FilmQA Pro (Ashland, USA). The Gafchromic films were irradiated in the same geometry as the plastic scintillators and processed according to the "one-scan protocol". Films were scanned at the maximum resolution accepted by the FilmQA Pro software, i.e. 600 dpi.

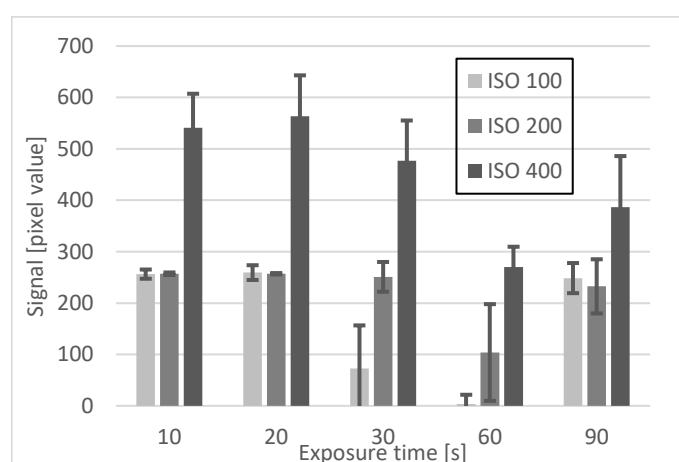


Figure 2. Signal as mean pixel value for different exposure times at different ISO settings. The results were filtered using the median and Wiener filters. Error bars represent the standard deviations of the individual measurements.

Results

Color channel selection

The camera CMOS sensor captures the signal in three colours: red, green and blue. 99% of the signal captured by the camera sensor in the scintillator area was collected in the blue channel. The remaining two channels showed no statistically significant differences in signal from the scintillator and outside the scintillator area. To minimise noise, only the blue component of the acquired images was used for further data processing.

Background and artefacts

The electronic noise measurements showed that the background signal was independent of the exposure time. It was 0.3% of the maximum signal recorded by the sensor at ISO 100 and 200, and 0.7% at ISO 400. The mean signal as a function of exposure time in the range from 10 s to 90 s for ISO 100, 200 and 400, respectively, is presented in **Figure 2**.

ISO 100 is an extended mode for Pentax K-x camera, while the first regular mode is ISO 200. It can explain similar results for both of these settings. The noise reduction observed for 60 s is difficult to explain. The authors attribute this effect to the signal processing of the camera's firmware.

The dependence of the collected stray radiation signal on the amount of the radiated dose is shown in **Figure 3**. The measurements were performed with a 6 MV beam, $15 \times 15 \text{ cm}^2$ field size in the same geometry as other tests but with the plastic scintillator removed.

From the results collected, we observed that for doses below 500 MU, the signal did not depend on the amount of dose delivered, and the mean value was equal to 259. An increase in the signal was observed for doses equal to and above 500 MU. For these results, a two-parameter (a , b) linear function $ax+b$ was fitted with $a = 1.2476$ and $b = 167$.

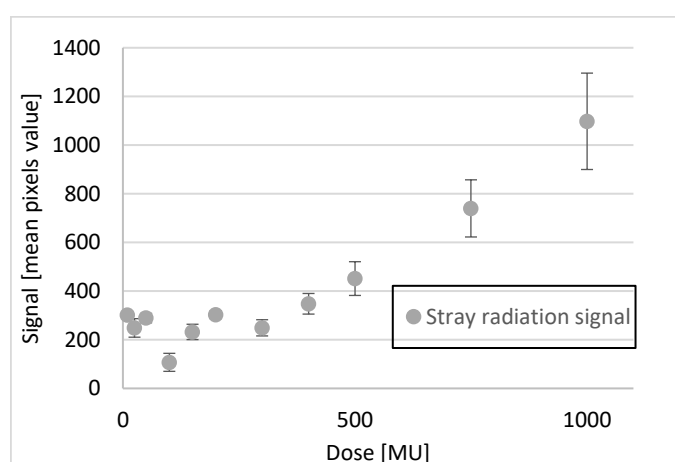


Figure 3. Measurement of stray radiation as a function of dose. Signal as mean pixel value vs the delivered dose for the field $15 \times 15 \text{ cm}^2$. Each measured value contains the electronic noise and stray radiation signal. The CMOS camera ISO was set to 100. Error bars represent the standard deviations of the individual measurements.

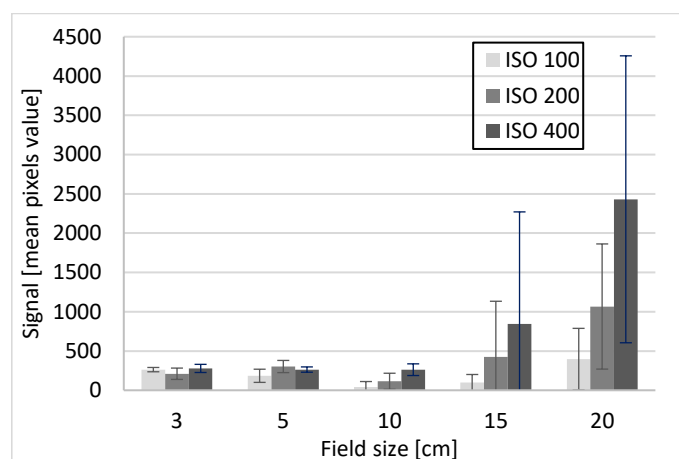


Figure 4. Measurement of stray radiation in relation to field size. Signal as mean pixels value vs different square field sizes (side of the square field). Results presented for three different ISO values. The standard deviation of each signal value is imposed on bar graph.

The amount of stray radiation also depends on the field size. The signal dependence for 100 MU for different field sizes using five different field sizes for ISO 100, 200 and 400 is shown in **Figure 4**. Fields ranging in size from $3 \times 3 \text{ cm}^2$ to $20 \times 20 \text{ cm}^2$ were investigated. The increase in signal from scattered radiation was observed for field sizes above $10 \times 10 \text{ cm}^2$. A constant response was observed for smaller fields.

The MATLAB scripting language was used to find the optimal focal length for vignetting correction. The same optimisation methods were used for tilt correction. Collected images were corrected for vignetting before further processing.

The MTF for a dosimetry film and PSD was calculated for the same geometry for the $5 \times 5 \text{ cm}^2$ field. The results are presented in **Table 1**. The distance corresponding to spatial frequency where MTF is 50% of maximum value (MTF_{50}) shows that the blur caused by the PSD is 0.57 mm higher and the minimum size of distinguishable detail is 0.1 mm greater for the PSD. The profile for the $5 \times 5 \text{ cm}^2$ field in the Y direction is shown in **Figure 5**.

System Repeatability

Ten measurements in the series for ISO 200 and 400 were taken to assess the system response stability. The results obtained are presented in **Figure 6**. For clarity, the signal in the graphs is shown as a percentage of the mean value for the series of measurements.

The mean signal value for ISO 400 was 33144, and for ISO 200 was 21096. The standard deviation of the mean for ISO 400 and ISO 200 was 0.43% and 0.13%, respectively. The average standard deviation for single measurements was 0.93% for ISO 400 and 1.41% for ISO 200.

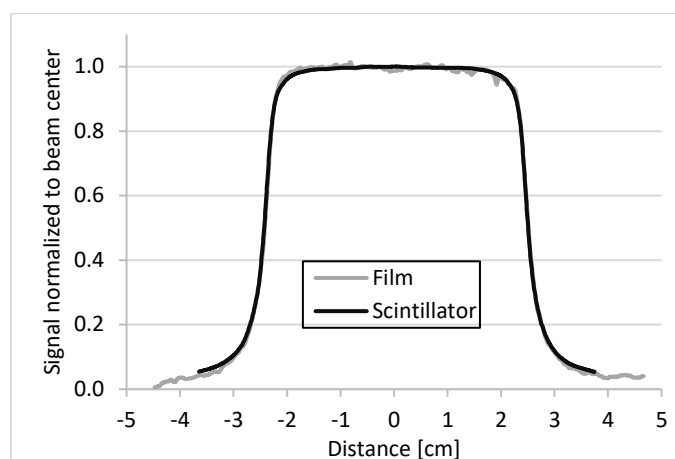


Figure 5. PSD profile to a Gafchromic film profile for $5 \times 5 \text{ cm}^2$ beam. Results have been normalised to the centre value for both PSD and Gafchromic films.

Table 1. MTF [mm] for a film and PSD

$\text{MTF}_{50} \text{ film}$	$\text{MTF}_{50} \text{ PSD}$	$\text{MTF}_{10} \text{ film}$	$\text{MTF}_{10} \text{ PSD}$
3.20	3.77	1.26	1.36

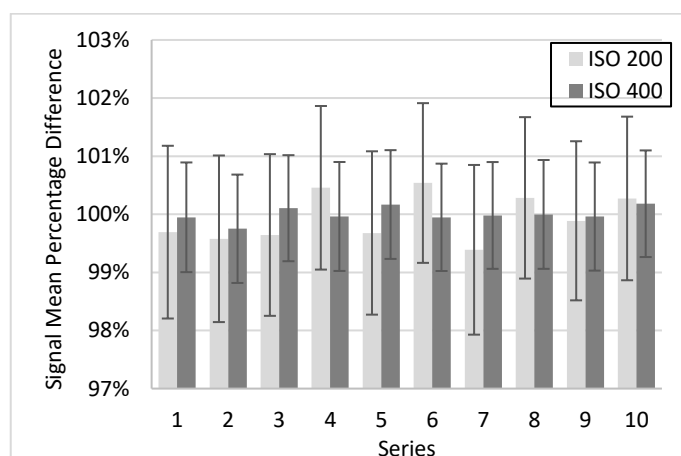


Figure 6. Repeatability measurements at ISO 200 and 400 as percentage differences to average results from all series. The standard deviation (SD) of each signal value is imposed on bar graph. The average SD for ISO 200 and 400 is 1.41% and 0.93% respectively. The lower SD for ISO 400 is due to the higher signal, while the noise is comparable for both setups.

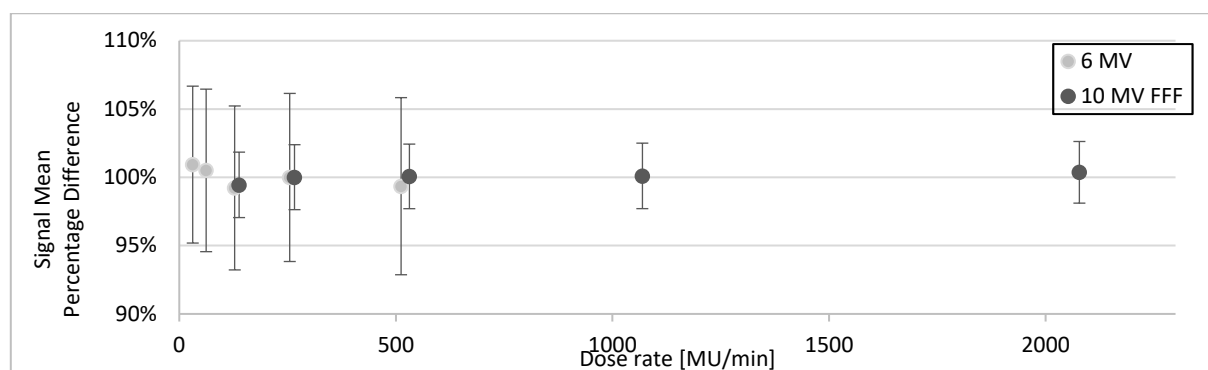


Figure 7. Relative signal difference versus the dose rate for 6 MV FF and 10 MV FFF beams. Error bars represent standard deviations for the single measurements.

Dose Rate dependence

The measured 6 MV FF and 10 MV FFF beams were calibrated to deliver 1 cGy/MU at 10 cm depth at a source-surface-distance (SSD) of 90 cm for the $10 \times 10 \text{ cm}^2$ field. The experimental setup had a different geometry: the plastic scintillator was covered with 2 cm PMMA plates, and the SSD was 98 cm. As a result, different absolute doses were delivered for the same amount of MU. The experiments were performed over several and the beams were measured in different setups. 6 MV FF were delivered with a smaller mirror-camera distance, $5 \times 5 \text{ cm}^2$ field size and had about 78% of the maximum sensor signal, while 10 MV FFF were delivered with $15 \times 15 \text{ cm}^2$ field size and had 48% of the maximum sensor signal. Therefore, to compare the signals for different energies, the data were normalised to the average signal for each energy. The results are shown in **Figure 7**.

Dose dependence

The dose-response function f of the detector introduced in **Equation 1** was investigated by delivering different doses to the scintillator. Sample images are shown in **Figure 8**.

For ISO 100 and 400, the doses were delivered in the same setup, while for ISO 200 in a different one. This allowed a selection of curves to be verified against different geometries. Acquired data were normalised to the signal at 100 MU for each curve (**Figure 9**).

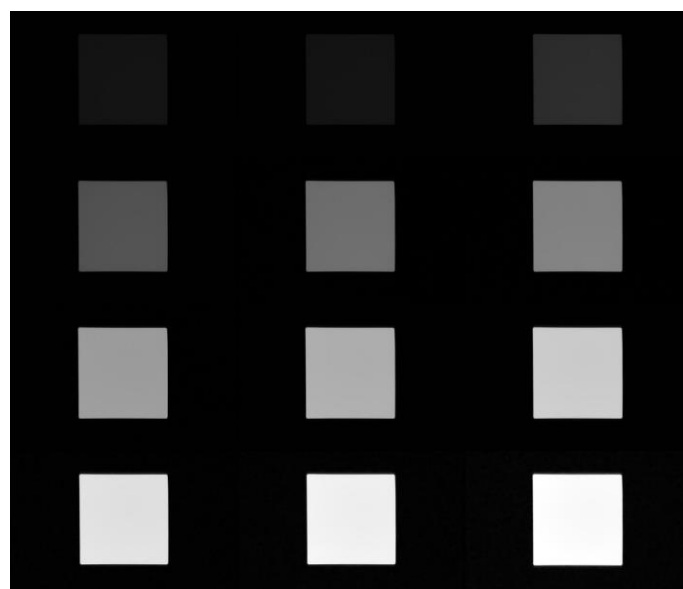


Figure 8. The collage of the images from doses delivered 12 times with values from 10 MU to 1000 MU for ISO 100. Starting from the upper left corner, first in the rows and then in the columns, each picture represents a single image for 10, 25, 50, 75, 100, 150, 200, 300, 400, 500, 750 and 1000 monitor units, respectively.

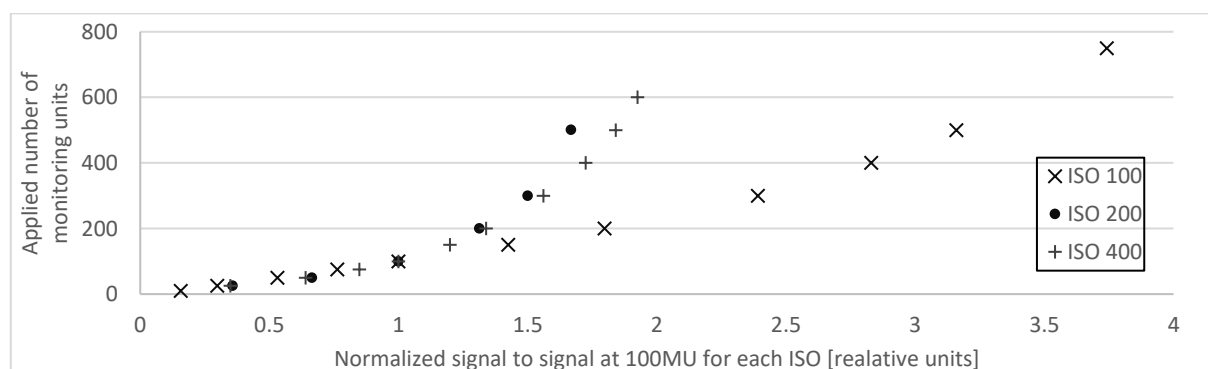


Figure 9. Doses versus normalised signals for three different ISO values. The delivered signal was normalised to unity at 100 MU for each ISO.

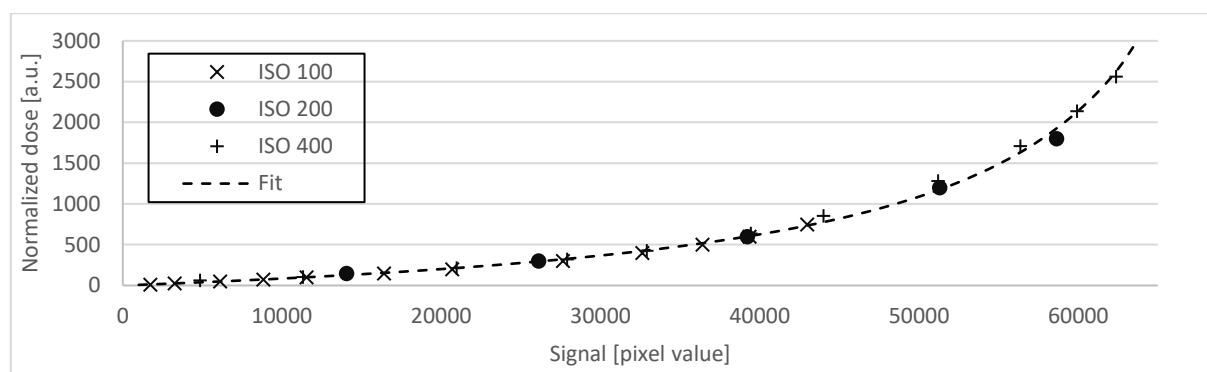


Figure 10. The normalised signal versus dose (arbitrary units).

A number of curves can be fitted to acquired data.¹³⁻¹⁸ Using the fitting software, more than 400 curves of 2 and 3 parameters were tested for the best fit. Of all the curves tested, the best results were obtained for the following curves:

2 parameters: $\frac{x}{(a + \frac{x}{b})}$ Eq. 2

$a(1 - e^{bx})$ Eq. 3

3 parameters: $\frac{x}{(a + \frac{x}{b})} + c$ Eq. 4

$a(1 - e^{bx}) + c$ Eq. 5

The R^2 fit coefficients are shown in **Table 2**. Of the curves listed in **Table 2**, the curve in **Equation 2** was selected because it had only 2 parameters, and the average R^2 fit did not differ by more than 0.004 from the 3-parameter curves.

The calculated parameters for the selected curve (**Equation 2**) for ISO 100, 200 and 400 with 95% confidence bounds are presented in **Table 3**.

Having determined which curve best described the dose-to-signal relationship for different ISOs, an attempt was made to unify this curve so that the differences between readings for different ISOs would be reduced to a single parameter – the P_{ISO} characteristic for each ISO. This parameter is related to the different sensitivity of the sensor at different ISO settings. For ISO 100, the P_{ISO} parameter was assumed to be equal to 1. For the curve described in **Equation 2**, the following parameters were fitted to all data with 95% confidence bounds: $a = 139.8$ (135.6, 143.9), $b = 556.2$ (535.9, 576.4). R-square: 0.999.

ISO dependence

Measurements for ISO values equal 100, 200, 400 and 800 were performed to obtain signal-ISO dependence. As the camera ISO settings are a closed set, instead of fitting a function, the data for each ISO value are presented in **Table 4**. The resulting signal was compared with the signal for ISO 100. The dose was then calculated for each ISO setting using the signal-dose equation and compared to the ISO dose at ISO 100. It is possible to fit a curve describing the relationship between these settings and the signal, but since these parameters have strictly defined values, it seems preferable to use a look-up table to reduce the standard deviation.

Table 2. The best R^2 for 2- and 3-parameters fit functions for signal to dose dependence. Fitting was performed separately for different ISOs. a , b and c are fitting parameters, while x is a pixel value.

	Fit Function	R^2		
		ISO 100	ISO 200	ISO 400
2 parameters fit	$\frac{x}{(a + \frac{x}{b})}$	0.9994	0.9929	0.9982
	$a(1 - e^{bx})$	0.9887	0.9984	0.9889
3 parameters fit	$\frac{x}{(a + \frac{x}{b})} + c$	0.9995	0.9944	0.9977
	$a(1 - e^{bx}) + c$	0.9986	0.9983	0.9919

Table 3. Fitting parameters a and b for the selected fit functions for different ISO values.

	ISO 400 (95% confidence band)	ISO 200 (95% confidence band)	ISO 100 (95% confidence band)
a	749.5 (711.5, 787.4)	137.9 (127.8, 148)	151.2 (145.2, 157.3)
b	104.9 (98.8, 111.9)	566.8 (516, 617.6)	469.7 (436.5, 502.9)

Table 4. Camera ISO, the measured signal, the signal converted to the dose [cGy] and calculated ISO correction factor P_{ISO} .

ISO	Measured Signal [pixels value]	Calculated dose response [cGy]	P_{ISO}
100	11488	95	0.990
200	20914	202	2.117
400	33113	410	4.301
800	44451	742	7.853

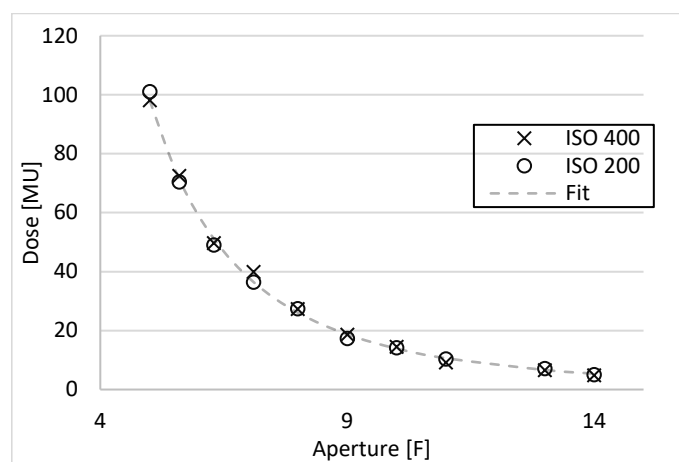


Figure 11. Dose vs the camera aperture F curve.

Fit using $\frac{x}{(a+\frac{x}{b})}$ function.

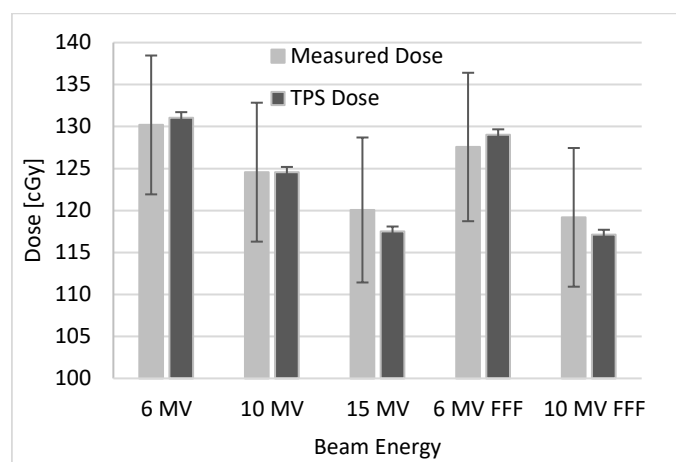


Figure 12. Measured signal versus dose from TPS for different photon energies. The signal is converted to dose and rescaled to the absolute value. The standard deviation is plotted on the bar graph.

Table 5. The measured signal as the mean pixel value, the signal recalculated to monitor units using the proposed signal-to-dose formula, the dose calculated from measurements, the calculated dose (TPS) for different beam energies.

Beam	Measured signal [pixels value]	Measured signal converted to MU	Measured dose [cGy]	TPS dose [cGy]	Difference between TPS and measured dose (%)
6 MV	32881	99.8	130.2	131.1	0.7%
10 MV	32062	95.5	124.6	124.6	0.0%
15 MV	31384	92.1	120.1	117.5	-2.2%
6 MV FFF	32503	97.8	127.6	129.0	1.1%
10 MV FFF	31249	91.4	119.2	117.1	-1.8%

Camera aperture

The amount of light captured by a lens is theoretically proportional to the area of the aperture, which is proportional to the reciprocal square of the F . Measurements were made for ISO 200 and ISO 400. The signal was converted into the dose using the signal-dose function.

Similarly to the signal-dose response, more than 400 curves were tested, and the best fit was obtained with $R^2 = 0.9987$ for the $x/(a+x/b)$ function. The calculated coefficients with 95% confidence bounds were: $a = 9162$ (7724, 1060) and $b = -2.819$ (-2.909, -2.729). The results with the fitted function are shown in Figure 11.

Energy dependence

Measurements were made for all available photon beam energies. The results are shown in Figure 12. The signal was converted to the relative dose and then rescaled by the signal to dose coefficient D_{cf} in order to obtain the absolute dose and be able to compare it with TPS results.

The calculated signal to cGy conversion factor for the given measurement geometry was 1.304. The system was calibrated using a 6 MV beam, so the expected value of the D_{cf} should be close to the value of the dose delivered by the 6 MV beam. The results presented in Table 5 show that this is the case, and the TPS dose value was 131.1 cGy for 100 MU and the difference was about 0.5%. The average difference between TPS and measured dose for all energies was 1.1%. The differences between the measured doses and the doses from TPS were on

average -0.4%, ranging from -2.2% to 0.7%, with the maximum difference being for the 15 MV beam.

Spatial resolution

The collected images from each scintillator detector were denoised, corrected for vignetting and tilted. The Gafchromic films were irradiated with the same setup. The collected results were compared with the TPS calculations. The dose plane for the same geometry as the experimental data was exported. The results are presented in Table 6.

The field size result for the scintillator detector was calculated using two methods. In the first method, the physical size of the scintillator was measured and compared with the pixel size to obtain the pixel size in the X and Y directions. The second method was to calculate the linear regression between the measured profile width and the results obtained from the films.

In the first method, the average difference between the film and the scintillator was 0.27 mm. The average relative difference was 0.91%. The relative difference increases for smaller fields.

For the second method, the calculated fitted line has the following parameters: for direction X, $a = 0.0541$, $b = 0.1111$ and for direction Y, $a = 0.0517$, $b = 0.0343$. The parameters a correspond to the pixel size in a given direction, while b corresponds to the image blur related to the point spread function. In the second method, the average difference between the film and the scintillator was 0.28 mm, and the average relative difference was 0.94%.

Table 6. Comparison of field sizes for TPS, films and PSDs [mm] (two methods).

Set field size		TPS		Film		Scintillator (first method)		Scintillator (second method)	
Field size X	Field size X	Field size Y	Field size X	Field size X	Field size X	Field size Y	Field size X	Field size Y	Field size Y
5	-	-	5.70	7.70	5.61	7.82	5.73	7.86	
10	10.86	10.31	10.90	11.80	10.82	11.91	10.96	11.96	
15	15.56	15.12	16.00	16.90	15.91	16.84	16.06	16.90	
20	20.49	20.12	20.70	22.10	20.91	21.64	21.07	21.70	
30	30.48	30.18	30.70	31.60	30.92	31.97	31.10	32.05	
40	40.56	40.20	40.60	42.00	40.78	41.71	40.99	41.81	
50	50.62	50.18	50.10	51.00	50.84	51.61	51.07	51.72	
60	60.59	60.21	60.90	61.60	60.53	61.65	60.77	61.78	
70	70.64	70.19	71.00	71.80	70.56	71.44	70.83	71.59	

As shown in **Table 6**, both methods give similar results. It can be noticed that the field size results for the scintillator detector relative to the films were closer than the field size results for the model in TPS. The average difference between the Gafchromic films and PSD was 0.9%.

Discussion

Background and artefacts were thoroughly analysed in the investigated system. Measurement results for different exposure times showed that the measurement system was well isolated from light. Overall, the residual light in the bunker was negligible and the electronics of the camera used contributed a small, constant noise to the signal in the order of 0.3 to 0.7% of the maximum signal. As expected, the higher the ISO setting, the higher the signal background was measured. Measurements have shown that as the sensitivity changes, electronic noise increases proportionally. At ISO values below ISO 800, the background noise was at an acceptable level of around 1% of the maximum signal.

The conclusion was as follows. The signal-to-noise ratio for long exposures is not affected since the background signal is independent of the exposure time. The lowest possible ISO setting should be used.

Figure 3 shows the relationship between the amount of MU irradiated and the stray radiation signal collected. There was no significant difference between the signal below 500 MU and the background. For doses equal to and greater than 500 MU there was a clear correlation between the administrated dose and the signal value. Above 500 MU, the linear fit coefficient R^2 between dose and signal was 0.997. There was also an increase in the standard deviation, indicating an increase in the collected noise.

This experiment showed that for doses below 400 MU, the stray radiation signal was mainly background noise for the studied setup. For doses above 400 MU, an increase in noise of 1.2% of the maximum signal for every 1000 MU given can be expected.

In **Figure 4**, a rapid increase of the stray radiation noise was observed for fields larger than $10 \times 10 \text{ cm}^2$. Increasing the area of field size, enhanced scatter radiation resulting in a higher background was observed. For fields smaller than $10 \times 10 \text{ cm}^2$,

the observed background was constant. In the tested system, the scintillator size was $10 \times 10 \text{ cm}^2$. Therefore, for such a detection system, it can be concluded that the noise does not depend on the field size for fields smaller in size than the PSD.

As the experimental results show a strong dependence between the field size and the signal (**Figure 4**), the system limits of the presented setup are revealed. For the field size larger than $20 \times 20 \text{ cm}^2$ and ISO 400, the signal noise and its standard deviation were 3.7% and 2.7% of the maximum signal, respectively. The system developed in this study was equipped with a $10 \times 10 \text{ cm}^2$ scintillator detector. Future measurements in this configuration are not planned for fields larger than $10 \times 10 \text{ cm}^2$.

The repeatability of the results obtained in a series of measurements (10 times in a row, **Figure 6**) was presented. The normalised mean value ranged between 99.39-100.58% with a standard deviation of 0.32%. The standard deviation of a single measurement was, on average, 1.18%. The obtained result indicated that the standard deviation for the single measurement was higher than the standard deviation of the mean for the series of measurements, indicating the high repeatability of the system. The relatively high standard deviation for the single value with the low standard deviation of the mean value for the series of measurements may be due to the characteristics of the CMOS matrices used in the experiment. Most likely, this is due to the fact that each pixel of the CMOS matrix has its own analogue-to-digital converter and a voltage converter making the measurement on such a matrix comparable to that of several detectors with similar characteristics. Each of these provides a repeatable, accurate signal, but small differences in construction add up to a larger standard deviation of the average. Deviation in the signal for a single measurement can also be caused by variations in the thickness of the plastic film.

For PSD irradiated at different dose rates, the pixel value normalised to the average of the measurement series for each energy ranged between 99.45-100.37% with a standard deviation of 0.23%. Splitting the results for individual energies, 6 MV FF and 10 MV FFF ranged between 99.22-100.93% and 99.86-100.09% with standard deviations 0.34% and 0.10%, respectively. The mean standard deviations for the single measurements were 0.42% and 2.46% for 6 MV FF and 10 MV FFF, respectively. The lower SD for 6MV FF is due to the use

of a $5 \times 5 \text{ cm}^2$ field compared to a $15 \times 15 \text{ cm}^2$ field for 10 MV FFF, resulting in lower stray radiation. The higher signal for 6MV FF compared to the maximum sensor value also results in a lower SD for a single measurement.

The signal versus dose response described by a curve in **Equation 2** with $R^2 = 0.9968$ is shown in **Figure 10**. This relationship is non-linear. For the signal below 50% of the maximum signal for the CMOS sensor, the system has a higher dose resolution.

The characteristics of the CMOS sensor used in the experiment are similar to those of the Gafchromic films regarding a saturation above a certain amount of collected signal. However, unlike with the film, the amount of light reaching the CMOS sensor can be adjusted by changing the aperture and sensitivity of the sensor. This is very convenient and allows one to reduce or increase the collected signal so that the resulting measurement value has an optimal signal-to-noise ratio and does not saturate the detector.

The detector signal from the PSD also increases with increasing ISO value (**Figure 9**). From the characteristics of the signal/dose curve described by **Equation 2**, it can be seen that the relative increase in signal for large signal values above half of the sensor's maximum value yields larger changes in dose. This means that the system is less sensitive to relative dose changes at high signal levels.

It has been shown that collected signal could be multiplied with higher ISO up to 7.9 times comparing ISO 100 to ISO 800 or decreased by up to 20 times comparing aperture F 5.6 to F 14. The measured dependencies of the camera ISO and aperture settings are shown in **Table 4** and **Figure 11**. It was concluded that it was possible to set up the measurement in such a way that the measured signal was in the highest dynamic range of the signal-dose curve and had a satisfactory signal-to-noise ratio.

Irradiation of PSDs with a fixed number of units within a fixed geometry but with different photon energies (MU) resulted in different doses. This was due to the different shapes of the percentage depth dose (PDD) curves for the different energies and the linac MU/cGy equal to 1 for the calibration at SSD 90 and the depth 10 cm, which differs from the measured geometry - SSD 98 and the depth 2 cm. The measurement results presented in **Figure 12** show that the relationship of the doses for different energies was maintained. Higher energies delivered the lower doses at 2 cm of depth. In all measurements, the TPS data were treated as the reference data. The differences between the measured doses and the doses from TPS were, on average -0.4%,

ranging from -2.2 % to 0.7 %, with a standard deviation of 1.5%. This means that the measurement result was within one standard deviation of the TPS value. The difference of about 2% at higher energies can be attributed to the measurements taken in a build-up region, the influence of the PSD thickness resulting in the dose averaging in the PDD and the inaccuracy of the TPS calculation region, which can be up to 4% at the depth of the dose maximum.¹⁹

A radiotherapy beam is not intended to be an imaging beam. The comparison of MTF_{50} and MTF_{10} for the Gafchromic film and the PSD showed that the films had 0.6 mm (18%) higher resolution than PSD and 0.1 mm (9%) higher minimum distinguishable object size. It can be concluded that the greater thickness of the plastic detector compared to a film was the main reason for the higher blurring of the beam image for PSD. Additional contributions to the blur were the inaccuracy of the optics, image filtering, imperfections in the scintillator surface and variations in scintillator thickness.

Conclusion

The excellent repeatability of PSD detectors with the normalised mean pixel value ranging from 99.39 to 100.39% with a standard deviation of 0.32%, small differences of 0.28 mm between measured field size compared to films, small differences in MTF_{50} and MTF_{10} equal to 0.6 mm (18%) and 0.1 mm (9%) compared to films, adjustable camera sensitivity using ISO and aperture settings and the simple signal-to-dose relationship were demonstrated in the presented studies. This indicated the possibility of using PSD detectors for a relatively fast and inexpensive check of the quality of the delivered radiation beam dose and shape, encouraging further work to investigate PSD for routine use in a radiation therapy department. Future development will focus on the construction of a phantom for quick and standardised measurements and software to streamline image processing. Using a thinner detector would reduce signal intensity but increase system resolution. This can be compensated for by using a camera with a higher sensitivity or lenses with a lower aperture number.

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References

1. Podgorsak EB. Radiation Oncology Physics, A Handbook for Teachers and Students: IAEA; 2005.
2. Beddar AS, Mackie TR, Attix FH. Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: I. Physical characteristics and theoretical consideration. *Physics in Medicine & Biology*. 1992;37(10):1883-1900. <https://doi.org/10.1088/0031-9155/37/10/006>
3. Beddar AS, Mackie TR, Attix FH. Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: II. Properties and measurements. *Physics in Medicine & Biology*. 1992;37(10):1901-1913. <https://doi.org/10.1088/0031-9155/37/10/007>
4. Goulet M, Rilling M, Gingras L, Beddar S, Beaulieu L, Archambault L. Novel, full 3D scintillation dosimetry using a static plenoptic camera. *Medical Physics*. 2014;41(8):082101. <https://doi.org/10.1118/1.4884036>
5. Carrasco P, Jornet N, Jordi O, et al. Characterization of the Exradin W1 scintillator for use in radiotherapy. *Medical Physics*. 2015;42(1):297-304. <https://doi.org/10.1118/1.4903757>
6. Beddar AS, Beaulieu L. Scintillation Dosimetry (1st ed.): CRC Press; 2016. <https://doi.org/10.1201/9781315372655>
7. Law SH, Suchowerska N, McKenzie DR, Fleming SC, Lin T. Transmission of Cerenkov radiation in optical fibers. *Optics Letters*. 2007;32(10):1205-1207. <https://doi.org/10.1364/OL.32.001205>
8. de Boer FS, Beddar SA, J R. Optical filtering and spectral measurements of radiation induced light in plastic scintillation dosimetry. *Physics in Medicine & Biology*. 1993;38(7):945-958. <https://doi.org/10.1088/0031-9155/38/7/005>
9. Clift MA, Johnston PN, Webb DV. A temporal method of avoiding the Cerenkov radiation generated in organic scintillator dosimeters by pulsed mega-voltage electron and photon beams. *Physics in Medicine & Biology*. 2002;47(8):1421-1433. <https://doi.org/10.1088/0031-9155/47/8/313>
10. Frelin AM, Fontbonne JM, Ban G, et al. Spectral discrimination of Cerenkov radiation in scintillating dosimeters. *Medical Physics*. 2005;32(9):3000-3006. <https://doi.org/10.1118/1.2008487>
11. Lambert J, Yin Y, McKenzie DR, Law S, Suchowerska N. Cerenkov-free scintillation dosimetry in external beam radiotherapy with an air core light guide. *Physics in Medicine & Biology*. 2008;53(11):3071-3080. <https://doi.org/10.1088/0031-9155/53/11/021>
12. Robertson D, Hui C, Archambault L, Mohan R, Beddar S. Optical artefact characterization and correction in volumetric scintillation dosimetry. *Physics in Medicine & Biology*. 2013;59(1):23-42. <https://doi.org/10.1088/0031-9155/59/1/23>
13. Brown TAD, Hogstrom KR, Alvarez D, II MKL, Ham K, Dugas JP. Dose-response curve of EBT, EBT2, and EBT3 radiochromic films to synchrotron-produced monochromatic x-ray beams. *Medical Physics*. 2012;39:7412-7417. <https://doi.org/10.1118/1.4767770>
14. Ashland. Gafchromic EBT2 self-developing film for radiotherapy dosimetry film. http://www.filmqapro.com/Documents/GafChromic_EBT-2_20101007.pdf
15. Devic S. Radiochromic film dosimetry: past, present, and future. *Physica Medica*. 2010;27(3):122-134. <https://doi.org/10.1016/j.ejmp.2010.10.001>
16. Devic S, Tomic N, Lewis D. Reference radiochromic film dosimetry: Review of technical aspects. *Physica Medica*. 2016;32(4):541-556. <https://doi.org/10.1016/j.ejmp.2016.02.008>
17. Micke A, Lewis DF, Yu X. Multichannel film dosimetry with nonuniformity correction. *Medical Physics*. 2011;38(8):2523-2534. <https://doi.org/10.1118/1.3576105>
18. Alfonso R, Andreo P, Capote R, et al. A new formalism for reference dosimetry of small and nonstandard fields. *Medical Physics*. 2008;35(11):5179-5186. <https://doi.org/10.1118/1.3005481>
19. Puyssleynr AD. Absorbed Dose in the Build-up Region in Modern Megavoltage Photon Radiotherapy: PhD Thesis; 2016. <http://hdl.handle.net/1854/LU-7099984>