Received:
23 May 2016

Revised:
27 July 2016

Accepted:
23 August 2016

<https://doi.org/10.1259/bjr.20160454>

Cite this article as:

Le Deroff C, Frelin-Labalme A-M, Ledoux X. Characterization of a scintillating fibre detector for small animal imaging and irradiation dosimetry. *Br J Radiol* 2017; **90**: 20160454.

SMALL ANIMAL IGRT SPECIAL FEATURE: FULL PAPER

Characterization of a scintillating fibre detector for small animal imaging and irradiation dosimetry

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Objective: Small animal image-guided irradiators have recently been developed to mimic the delivery techniques of clinical radiotherapy. A dosimeter adapted to millimetric beams of medium-energy X-rays is then required. This work presents the characterization of a dosimeter prototype for this particular application.

Methods: A scintillating optical fibre dosimeter (called DosiRat) has been implemented to perform real-time dose measurements with the dedicated small animal X-RAD® 225Cx (Precision X-Ray, Inc., North Branford, CT) irradiator. Its sensitivity, stem effect, stability, linearity and measurement precision were determined in large field conditions for three different beam qualities, consistent with small animal irradiation and imaging parameters.

Results: DosiRat demonstrates good sensitivity and stability; excellent air kerma and air kerma rate linearity; and a good repeatability for air kerma rates $>1\text{mGys}^{-1}$.

The stem effect was found to be negligible. DosiRat showed limited precision for low air kerma rate measurements ($<1\text{mGys}^{-1}$), typically for imaging protocols. A positive energy dependence was found that can be accounted for by calibrating the dosimeter at the needed beam qualities.

Conclusion: The dosimetric performances of DosiRat are very promising. Extensive studies of DosiRat energy dependence are still required. Further developments will allow to reduce the dosimeter size to ensure millimetric beams dosimetry and perform small animal *in vivo* dosimetry.

Advances in knowledge: Among existing point dosimeters, very few are dedicated to both medium-energy X-rays and millimetric beams. Our work demonstrated that scintillating fibre dosimeters are suitable and promising tools for real-time dose measurements in the small animal field of interest.

INTRODUCTION

Pre-clinical research in radiotherapy using small animals is crucial to improve or develop new cancer treatment approaches.¹ The actual irradiation techniques in clinical radiotherapy have become more and more sophisticated. Intensity-modulated radiotherapy, arc therapy or radio-surgery produce more conformal dose depositions with complex small beams and important dose gradients. In response to these sophisticated irradiation techniques, new dedicated small animal irradiators have been designed in order to reproduce more faithfully and transpose clinical treatments to small animal studies.^{2,3} These irradiators allow multiple beams delivery with millimetric collimation and medium-energy X-rays ($<250\text{keV}$). An on-board imaging system provides precise repositioning and improved targeting. This advanced technology involves high technical precision which requires adapted quality control, beam dosimetric characterization and treatment monitoring. This drives the need for an accurate and precise real-time dosimeter with high spatial resolution, for the kilovoltage energy range.

Kilovoltage dosimetry is a particular challenge. Since the photoelectric effect can be significant in high Z material ($Z > 10$), the dosimeter-sensitive material should be as close as possible to the tissue composition to avoid over- or underestimation of dose. Among existing dosimeters used in clinics, ionization chambers are well adapted for reference beam dosimetry in large field but suffer from too large volume for smaller beams. Other dosimeters such as diodes, thermoluminescent dosimeters and radiochromic films are not ideal for small animal dosimetry. None of them provides simultaneously high spatial resolution, adapted material composition and direct reading. Plastic scintillating optical fibres are good candidates for such measurements because they have a composition close to tissue and can be designed with small sensitive volume.

Dosimeters based on plastic scintillators have been widely studied for clinical megavoltage dosimetry.^{4,5} Several studies showed that in the entire radiotherapy energy range, they are water equivalent; have excellent reproducibility and response

linearity with dose and dose rate; and offer high spatial resolution because of their small size. Therefore, they have been popular for small beam dosimetry.⁵ However, it is known that in addition to scintillating light emitted by the scintillator, a stem effect is produced in the fibre material and contributes to the detected signal.^{6–8} When using megavoltage beams, the major component of this stem effect is Čerenkov radiation. Multiple studies and solutions have been proposed in literature to remove the Čerenkov light influence from the final signal.⁹ Small dependence to temperature has also been reported with a small decrease of the plastic scintillator light output with increasing temperature.^{10–12} This temperature dependence is an important consideration for a detector dedicated to *in vivo* dosimetry. Nevertheless, it becomes negligible in a temperature-controlled environment.

Interest has grown for the use of plastic scintillators in kilovoltage energy range, in particular for dosimetry in radiology and brachytherapy. The different prototypes described in literature present dosimeters with linear, reproducible dose measurements.^{13–15} However, other studies also found an energy dependence of the dosimeter response, expressed as a decrease of sensitivity with decreasing energy.^{16–19} In addition, in the kilovoltage energy range, the stem effect is mainly due to the fluorescence of the optical fibre and its contribution has been reported to be small.⁸ Its small influence on the final signal allows to simplify the dosimeter readout system.

The purpose of the present study was to demonstrate the ability of a plastic scintillator to satisfy the dosimetry requirements for small animal irradiation and imaging. We characterized the performances of a homemade real-time plastic scintillating fibre dosimeter. The dosimeter sensitivity and stem effect were evaluated, then we focused on the dosimeter response stability

and its response with dose and dose rate, and finally on the precision achievable with our prototype.

METHODS AND MATERIALS

Dosimeter prototype

The prototype dosimeter, called DosiRat, is composed of a scintillating probe, represented in Figure 1a and a photo-detection and electrometer part, displayed in Figure 1b. The scintillation probe is composed of a scintillating fibre coupled to an optical fibre which end is inserted in a straight tip (ST) connector. The scintillating fibre is a polystyrene-based BCF-12 fibre (Saint-Gobain Crystals, Nemours, France), the emission peak of which is located at 435 nm and is of 1-mm diameter thick and 15 mm long. This length was chosen purposely for our study; it cannot provide an optimal spatial resolution but maximizes the output signal. This scintillating fibre is coupled to a polymethyl-methacrylate (PMMA) optical fibre of the same diameter, thanks to cyanoacrylate glue, the refraction index of which is close to one of polystyrene and PMMA. It is therefore a good candidate for an optimized optical coupling.²⁰ The fibre alignment was achieved under the microscope using micrometer-operated three-axes stages (Thorlabs, Inc., Newton, NJ) and by maximizing the transmitted power of a red laser (Visual Fault Locator, 630 nm; Westover Scientific, Inc., Bothell, WA) through both the scintillating and the optical fibre. Prior to coupling, all the optical interfaces were polished with sheets of decreasing grain size. The scintillator tip was covered with titanium dioxide to optimize light reflection. The other end of the optical fibre was inserted into a ST connector. The fibres were covered by a polyethylene jacket and by black acrylic paint for the stripped part, to shield the scintillating probe from the external light. This shielding allows to measure scintillation light exclusively, and the measurements can therefore be performed

Figure 1. (a) Scintillating probe consisting of a BCF-12 scintillating fibre coupled to a polymethyl-methacrylate clear optical fibre inserted in a straight tip (ST) connector. (b) DosiRat dosimetric system including the scintillating probe, the photodetection and electrometer parts. CAMEL, CARte Acquisition Multivoies ELEctromètre; TiO₂, titanium dioxide; Si, silicon.

Figure 2. X-RAD® 225Cx irradiator: (a) shielded cabinet and remote control panel; (b) inside of the cabinet.

in the X-RAD® cabinet with the light on. This probe is connected to the measurement part (located outside the X-ray shielded cabinet) by an approximately 2-m long optical fibre and a ST connection. This part includes six measurement channels connecting the probe to a silicon PIN photodiode (S1223, Hamamatsu Photonics, Hamamatsu City, Japan), with about 0.2 A/W photosensitivity at 435 nm. No bias voltage was applied to the photodiodes. As well as for the fibres, this measurement box is shielded from external light with an aluminium covering. The charge produced by the photodiodes is collected by a numerical electrometer CARte Acquisition Multivoies ELectromètre card included in Fast Acquisition SysTEm for nuclear Research acquisition system,²¹ initially designed for nuclear physics experiments. This numerical electrometer simplifies the acquisition circuit and gives a compact detection system. Since the CARte Acquisition Multivoies ELectromètre card has 32 channels, several photodiodes can be used simultaneously, especially as they are low cost and easy to use.

For the present study, only one channel of DosiRat was used, but six channels are currently available. The electrometer charge readout frequency was set to 10 Hz.

Small animal irradiator

Our dosimeter was evaluated with the X-RAD® 225Cx (Precision X-ray Inc., North Branford, CT) image-guided irradiator. This system is composed of a 225-kVp X-ray tube (MXR225/22, COMET) mounted on a C-arm gantry in front of an amorphous silicon flat-panel imaging detector, all inside a shielded cabinet (Figure 2).

The tube inherent filtration is made of a 0.8-mm beryllium foil, and either a 2-mm aluminium foil or a 0.3-mm copper foil can be placed as additional filter. The beam size can be modified with manually placed fixed collimators. Typical irradiations are performed using the maximum energy and current (225 kV, 13 mA) with the copper filter while image acquisitions are performed with lower energy (40–120 kVp) and current, with the aluminium filter.^{3,22}

DosiRat performances were therefore evaluated with three different beam qualities, described in Table 1, representative of pre-clinical applications (treatment and imaging). The corresponding

half-value layers (HVLs) were measured according to the American Association of Physicists in Medicine (AAPM) Task Group 61 (TG-61) protocol recommendations,²³ using aluminium sheets. Unless otherwise stated, the tube current used for each beam quality is the one reported in Table 1, in agreement with the defined protocols used in routine.

Reference dosimetry

The three tested beam qualities were calibrated in air kerma following the protocols for low and medium energy, described in the AAPM TG-61 report.²³ The reference conditions were 10 × 10-cm field size, free in air measurement at a source-to-detector distance (SDD) of 30.7 cm.

Two different calibrated ionization chambers were used, according to the different beam qualities (Table 2). Each chamber has a traceable calibration to national standards of the National Metrology Institute of Germany, Physikalisch-Technische Bundesanstalt, Braunschweig.

The signals from the ionization chambers were acquired using a PTW (PTW-Freiburg, Freiburg, Germany) UNIDOS® E electrometer with bias voltages of +400 V for the PTW 30013 and +300 V for the PTW 23342 chamber. Corrections for ambient

Table 1. Specifications on the beam qualities used for the dosimeters exposures, including half-value layers (HVLs)

Beam quality	225 kVp 0.3 mm Cu	100 kVp 2 mm Al	40 kVp 2 mm Al
Tube voltage (kVp)	225	100	40
Tube current (mA)	13	0.7	7
Added filtration (mm)	0.3 Cu	2 Al	2 Al
Measured HVL (mm Al)	11.5 ± 0.2	2.3 ± 0.4	1.2 ± 0.1
Application	Irradiation	Imaging	Imaging

Al, aluminium; Cu, copper.

Table 2. Ionization chambers used as reference dosimeters depending on beam quality

Beam quality	225 kVp 0.3 mm Cu	100 kVp 2 mm Al	40 kVp 2 mm Al
Ionization chamber	PTW 30013 cylindrical air cavity chamber	PTW 23342 parallel plate air cavity chamber	

Al, aluminium; Cu, copper.

temperature and pressure (k_{TP}) were applied to the chamber readings. Ion recombination and polarity corrections (k_{ion} , k_{pol}) were measured and applied for each beam quality using AAPM TG-61 protocol formulas. Beam quality corrections of the calibration factor (k_Q) were obtained by measuring the beam HVL and interpolating the closest specified correction factors given by the calibration certificates.

The combined standard uncertainty on air kerma measurements, for a given beam quality and tube current, was computed according to the AAPM TG-61 protocol recommendations. It includes the uncertainty on the certificate calibration factor (1% for the PTW 30013 and 2.5% for the PTW 23342), the effect of beam quality difference between calibration and measurement (2%), the statistical uncertainty in charge measurement (determined on ten successive measurements) and the uncertainties on correction factors (k_{TP} , k_{ion} , k_{pol}), which do not exceed 0.5%.

DosiRat characterization

Experimental setup

All measurements were performed at isocentre with a SDD of 30.7 cm and a 10×10 -cm square field collimator. The DosiRat probe and the parallel plate chamber were placed on 1.5-cm-thick expanded polystyrene foam lying in a dedicated rat bed (Equipement vétérinaire Minerve, Esternay, France). The cylindrical chamber was held in air by a dedicated support. The dosimeters were positioned thanks to a cone beam CT acquisition. The irradiations were performed in the axial plan of symmetry of both DosiRat probe and the cylindrical chamber (with the beam perpendicular to the dosimeter's axis). The parallel plate chamber was irradiated with the beam perpendicular to its surface.

Unless otherwise stated, the exposure time of DosiRat and the ionization chambers was 40 s, and the charge was integrated over 30 s. The 10 first seconds of exposure were omitted to allow beam stabilization in kilovoltage and milliamperage.

The measurement of statistical uncertainties with DosiRat was determined by the standard deviation of 10 successive readings for a given kilovoltage and tube current combination (see the Precision vs beam quality and integration time section). The temperature inside the X-RAD® cabinet was stable during measurements ($21 \pm 2^\circ\text{C}$). Thus, the uncertainty generated by the small temperature dependence of the BCF-12 light output ($-0.263 \pm 0.028\%/^\circ\text{C}$ reported by Lee et al¹²) was not taken into account. It should not exceed 1% in our

conditions. However, if DosiRat is used in a non-controlled temperature environment, the uncertainty on dose measurement could be significant if the temperature is different from the reference one, used for DosiRat dose calibration. In that case, the temperature should be reported and a correction factor should be included for each dose measurement. DosiRat could also be calibrated for a given set of temperatures.

Sensitivity and stem effect

DosiRat response (pC) was measured for each beam quality of Table 1. The corresponding air kerma (Gy) measured with the ionization chambers allowed to express DosiRat sensitivity in pC Gy^{-1} .

The intensity of the stem effect generated in DosiRat probe is expected to be small compared with the scintillation signal. Since it is proportional to the length of the irradiated fibre, the stem effect can be assessed and maximized by irradiating a very long optical fibre. Its intensity can then be expressed per centimetre of irradiated optical fibre. The stem effect intensity in a normal irradiation condition can then be derived in proportion to the length of exposed optical fibre.

Therefore, a 194-cm-long optical fibre of PMMA was prepared (cut, polished and shielded from light) with the same protocol as the scintillator probe. As shown in Figure 3, the fibre was rolled up to be contained in a 10×10 -cm irradiation field at a SDD of 30.7 cm, on 20-cm-thick expanded polystyrene foam, to approach free in-air measurement conditions. The fibre was irradiated with the beam qualities shown in Table 1, applying maximum tube currents to maximize the stem effect signal (13, 30 and 45 mA for 225, 100 and 40 kVp, respectively). Similarly, air kerma

Figure 3. Experimental setup for the stem effect evaluation. A long clear optical fibre was contained in a 10×10 -cm field size. Irradiation was performed on expanded polystyrene foam to simulate an exposure in free air.

Table 3. Air kerma rates in mGy s^{-1} delivered by the irradiator for different combinations of tube voltages (kVp) and tube currents (mA)

mA	kVp		
	225	100	40
0.1	0.33	0.19	0.03
0.3	1.00	0.53	0.09
0.5	2.00	0.88	0.15
0.7	2.67	1.25	0.22
1	3.67	1.79	0.32
5	11.3	8.98	1.58
7	18.7	12.6	2.21
10	–	17.9	3.16
13	48.3	–	–
20	>tube max. power	35.9	6.31
30	>tube max. power	53.7	9.44

max., maximum.

Uncertainties on air kerma rates do not exceed 2.3% for the 225-kVp beams and 3.5% for the 100- and 40-kVp beams.

– : no measurement performed for these kVp-mA combinations.

measurements were performed with the ionization chambers, and the stem effect signal was expressed in pC Gy^{-1} per centimetre of irradiated optical fibre. To express the stem effect generated when DosiRat probe was exposed to a 10×10 -cm square field, its intensity was reduced in proportion to the length of exposed optical fibre (5.75 cm in our case). This stem effect was then compared with the scintillation probe signal for the beam qualities shown in Table 1.

Short-term stability

DosiRat response stability over the course of 10 days was evaluated. Daily measurements of DosiRat response along with air kerma measurements were performed for the 225-kV, 13-mA beam. The average response and relative standard deviation were then computed for both dosimeters.

Linearity

The linearity of DosiRat response with air kerma was evaluated for each beam quality shown in Table 1 by increasing the

integration time from 1 to 120 s while keeping the tube potential and current constant. As previously, the first 10 seconds of irradiation were omitted to reach beam stabilization. DosiRat charge was then plotted as a function of the air kerma measured by the ionization chambers. A linear fit was applied, and the coefficient of determination R^2 was used to quantify the linearity of the dosimeter.

Linearity with air kerma rate was evaluated as well. The different rates were obtained by holding the exposure time constant and varying the tube current from 0.1 to 30 mA (13 mA at 225 kVp due to the maximal tube power of 3 kW), delivering air kerma rates ranging from 0.03 to 53.7 mGy s^{-1} . Table 3 displays the air kerma rates in mGy s^{-1} achieved by the different beam qualities and tube current. A linear fit was applied to DosiRat measurements plotted vs air kerma at different air kerma rates, and the coefficient of determination R^2 was used again to quantify the dosimeter's linearity.

Precision vs beam quality and integration time

The measurement precision of DosiRat was tested for different beam parameters (kVp, mA) and for different charge integration times. For that purpose, measurements were repeated 10 times, and the average reading ($\bar{x}$) and standard deviation (σ_x) were evaluated for each tested air kerma rate. The coefficient of variation expressed as $C_v(\%) = \sigma_x / \bar{x} \times 100$ was then calculated and plotted as a function of the air kerma rate, for either 30 or 2 s of charge integration time.

RESULTS AND DISCUSSION

Sensitivity and stem effect

Table 4 shows DosiRat sensitivity in air for a 10×10 -cm field size, for each beam quality. The corresponding stem effect (for 5.75 cm exposed optical fibre) is also reported, as well as its relative contribution to the total light signal. The stem effect contribution was found to be $<0.13\%$ and is therefore negligible. It must be noticed that this contribution is directly proportional to the irradiated fibre length and will be smaller for smaller irradiation fields. As reported by Therriault-Proulx et al,⁸ the main source of stem effect at those energies is fibre fluorescence. Indeed, Čerenkov radiation is generated in PMMA by secondary electrons $>178 \text{ keV}$. Such electron energy cannot be reached by Compton diffusion for photons $<225 \text{ keV}$,²⁴ and photoelectric absorption represents $<0.2\%$ of the total absorption in 1 mm of PMMA.²⁵ These results confirm that at pre-clinical energies, the stem effect is exclusively composed of optical fibre fluorescence and does not need to be corrected.

Table 4. DosiRat sensitivity and stem effect in pC Gy^{-1} for different beam qualities

Beam quality	225 kVp 0.3 mm Cu	100 kVp 2 mm Al	40 kVp 2 mm Al
Sensitivity (pC/Gy)	202.6 ± 4.7	126.4 ± 7.8	90.6 ± 5.7
Stem effect (pC/Gy)	0.19 ± 0.01	0.12 ± 0.01	0.12 ± 0.01
Stem effect contribution	0.10%	0.10%	0.13%

Al, aluminium; Cu, copper.

The stem effect proportion in DosiRat signal is also reported.

Table 5. DosiRat relative sensitivity for the different beam qualities

Beam quality	225 kVp 0.3 mm Cu	100 kVp 2 mm Al	40 kVp 2 mm Al
Relative sensitivity to 225 kVp 0.3 mm Cu	100%	62%	45%
Beam mean energy $\bar{E}$ (keV)	85.4	47.0	27.3
$CF(\bar{E})$	0.85	0.54	0.39
Relative CF to 225 kVp 0.3 mm Cu	100%	64%	46%

Al, aluminium; Cu, copper; CF, correction factor.

Beams mean energies computed by SpekCalc²⁷ were used to calculate a CF for each beam quality. The CF is defined as the mass energy-absorption ratio between polystyrene and air.

The results also highlight that the sensitivity is not constant with beam quality. The sensitivities normalized to the one obtained at 225 kVp are reported in Table 5. They show a factor >2 between 225-kV beam and 40-kV beam. As a consequence, calibration factors (pC Gy^{-1}) should be determined for each specific beam quality used for measurements.

This energy dependence of plastic scintillator has first been expressed by Williamson et al¹⁶ who showed a reduction of the scintillation light output with decreasing energy for photons <133 keV. Other authors declare an energy dependence of plastic scintillator detectors sensitivity in the radiologic energy range (25–150 kVp).^{13,14,16–19} As reported by Williamson et al,¹⁶ Lessard et al,¹⁸ and Peralta and Rêgo,¹⁹ it is partly related to energy absorption differences between the reference dose medium and the scintillator medium (air and polystyrene respectively in our case). As shown in Figure 4, the polystyrene-to-air mass energy-absorption coefficients ratio shows significant variations <100 keV. Lessard et al¹⁸ proposed to account for this medium difference with CFs. These are expressed as the kerma ratio between the two media that can be expressed as their mass energy-absorption coefficient ratio, given the negligible range of secondary electrons at those energies. This ratio can be written as

$$CF = \frac{\left[\frac{\mu_{\text{en}}}{\rho}(\bar{E}) \right]_{\text{polystyrene}}}{\left[\frac{\mu_{\text{en}}}{\rho}(\bar{E}) \right]_{\text{air}}} \quad (1)$$

where $\mu_{\text{en}}/\rho(\bar{E})$ is the mass energy-absorption coefficient for the spectrum mean photon energy $\bar{E}$.

Equation (1) and data from Figure 4 were used to calculate CFs for our different beam qualities. The beams' mean energies were obtained with SpekCalc software²⁷ using the parameters of Table 1 as inputs. Indeed, SpekCalc calculates the X-ray spectrum from a tungsten anode X-ray tube. For a given set of tube potential, anode angle and filtration, SpekCalc computes, in particular, the mean beam energy averaged over the fluence spectrum. The resulting CFs are summarized in Table 5 with the relative sensitivities and clearly show that the medium difference is the major part of energy dependence. Nevertheless, these calculations are very approximate and

do not take into account accurate spectrum considerations. Indeed, accurate CF should be computed over the all-energy spectrum of each different beam, as carried out by Williamson et al,¹⁶ Lessard et al,¹⁸ and Peralta and Rêgo.¹⁹ Corrections factors do not explain entirely the energy dependence and remaining energy dependence has been observed after medium correction by Williamson et al¹⁶ and Peralta and Rêgo.¹⁹ Besides, Frelin-Labalme et al²⁸ also measured a non-linearity of response <100 keV for different plastic scintillators. It is explained as scintillation quenching, which results in a decrease of scintillator light output for high ionization density, due to the interactions of low-energy secondary electrons in the scintillator medium.

Our results showed a good sensitivity of DosiRat and highlighted its energy dependence. Although medium CF can correct it partially, further measurements and calculations are currently under progress with DosiRat to correlate the dose-meter response to deposited energy and to assess its "intrinsic" sensitivity. This will allow to differentiate sensitivity variation due to mass energy-absorption coefficient variations from scintillation quenching.

Figure 4. Mass energy-absorption coefficient (μ_{en}/ρ) ratios between polystyrene and air as a function of energy. Data taken from the National Institute of Standards and Technology.²⁶

Figure 5. Daily responses of DosiRat over 10 days when exposed to the 225-kV, 13-mA beam.

Short-term stability

Figure 5 shows the daily responses of DosiRat measured over a period of 10 days. The responses are normalized to the first measurement of this study (day 1). The standard deviation of the responses is 0.6% and all the measurements fell within $\pm 1.2\%$ of the average value as displayed by the dashed lines in Figure 5. The evolution of air kerma measurement was also analyzed separately and exhibits a standard deviation of 0.3%. Since the ionization chamber present excellent repeatability (standard deviation of 0.03%), this 0.3% standard deviation accounts for the irradiator beam daily variations. The fluctuations in DosiRat responses are then partially due to those daily beam variations. The remaining fluctuations are attributable to DosiRat system variations such as slight displacements in optical coupling at the optical interfaces (in the ST connector and between the fibre and the photodiode).

Figure 6. DosiRat response as a function of air kerma at 225 kVp, 13 mA with the copper filter.

These results highlight the capacity of DosiRat for stable and therefore reliable measurements over a short-term period. Indeed, the cumulated dose of 16.6 Gy delivered in 10 days does not affect the system response. However, plastic scintillators exposed to very high levels of dose are known to have a decrease in sensitivity, caused by radiation damage. Beddar et al²⁹ reported 2.8% decrease in the output signal of a polystyrene-based plastic scintillator after delivering 10^4 Gy. Likewise, Carrasco et al³⁰ reported the same decrease for 10^4 Gy of cumulated dose in the Exradin W1 scintillating fibre detector (polystyrene based) and also observed up to 8% decrease in sensitivity after delivering 1.25×10^5 Gy.

Overall, given the range of deposited dose during animal irradiations compared with the important cumulated dose needed to damage the scintillator, DosiRat can be used for reliable dose measurements without too frequent recalibration. However, the cumulated dose in a plastic scintillating fibre dosimeter should be monitored to prevent the effect of radiation damage. This allows to define a periodical recalibration or even the scintillator replacement in order to maintain constant accurate measurements.

Linearity

Plots of DosiRat response vs air kerma are displayed in Figure 6 for the irradiation beam quality and in Figure 7 for the imaging beam qualities. A trend line was fitted to each plot. Whatever the beam quality, the response was observed to increase linearly with the delivered air kerma, and the fit coefficient of determination was found to be at least 0.998.

Similarly, Figure 8 shows DosiRat response (in $\text{pC } 30 \text{ s}^{-1}$) as a function of the air kerma rate, for the three beam qualities. A linear fit was applied to each data curve, yielding a coefficient of determination of at least 0.997.

These results demonstrate that DosiRat response is perfectly linear with respect to the air kerma and to the air kerma rate, whatever the beam quality, for kerma ranging from 5 mGy to

Figure 7. DosiRat response as a function of air kerma at 100 kVp, 0.7 mA; and 40 kVp, 7 mA with the aluminium filter.

Figure 8. DosiRat response integrated over 30 s as a function of air kerma rate, for three different beam qualities. Al, aluminium; Cu, copper.

6 Gy and kerma rates ranging from 0.3 to 53.7 mGy s⁻¹, which correspond to exposure ranges of interest in small animal irradiation and imaging.

Precision vs beam quality and integration time

The C_V over 10 exposures is plotted as a function of the air kerma rate, for 30 and 2-s charge integration time (Figures 9 and 10, respectively).

The results for 30-s charge integration time show highly repeatable measurements for air kerma rates >8 mGy s⁻¹ as evidence by $C_V < 0.8\%$. The best C_V (0.2%) is reached when using the irradiation beam quality (225 kVp, copper filter and air kerma rate >18 mGy s⁻¹). For medium air kerma rates (1–8 mGy s⁻¹), the C_V is about 6%. For low air kerma rates,

Figure 9. Coefficient of variation (C_V) for 10 successive measurements with a charge integration time of 30 s, as a function of the air kerma rate. The black solid line represents a fitting function inversely proportional to the air kerma rate. Al, aluminium; Cu, copper.

Figure 10. Coefficient of variation (C_V) for 10 successive measurements with a charge integration time of 2 s, as a function of the air kerma rate. The black solid line represents a fitting function inversely proportional to the air kerma rate.

<1 mGy s⁻¹, the fluctuations between measurements are more important and the C_V reaches 11%. The C_V evaluated with 2-s charge integration time, compared with 30 s integration time, is slightly increased by 3.5% maximum, except for the lowest kerma rate which is degraded by 5%. Therefore, the precision is independent of the beam quality and only depends on the kerma rate (inversely proportional as shown by the black fitting curves in Figures 9 and 10). The precision also depends on the integration time, but satisfying precision was obtained with 2-s integration time for kerma rates >8 mGy s⁻¹.

These results show that DosiRat has to be used carefully for small animal imaging. For low dose imaging protocol, measurement uncertainties could reach 11% for low kerma rates or about 6% for medium kerma rates. DosiRat is, however, highly suitable for precise measurement with high dose rates, in particular for irradiation protocols (225 kVp, copper filter and tube current >4 mA). In those conditions, precise measurements can be made by either short (about 2 s) or long (>30 s) exposure times. These conclusions are in agreement with Boivin et al.³¹ recommendations concerning the dose rate ranges of use for one-dimensional plastic scintillation detector with a PIN diode photodetector.

CONCLUSION

Small animal image-guided irradiators have recently been developed in the field of pre-clinical research in radiotherapy. Scaled to animal size, they can deliver millimetric beams and mimic the delivery techniques of clinical radiotherapy. The high technical precision of such irradiators requires thorough dosimetric characterization in order to ensure accurate and precise dose delivery.

A scintillating optical fibre dosimeter has been implemented to perform real-time dose measurements with the dedicated small animal X-RAD® 225Cx irradiator. The dosimeter prototype

characterization was performed for three different beam qualities representative of small animal irradiation and imaging conditions. It demonstrated good sensitivity and stability, excellent air kerma and air kerma rates linearity and a good signal repeatability for air kerma rates $>1 \text{ mGy s}^{-1}$. The stem effect, mainly due to fluorescence in the optical fibre was found to be negligible for the studied conditions. One of the system drawbacks concerns DosiRat-limited precision for low air kerma rate measurements, typical for imaging protocols ($<1 \text{ mGy s}^{-1}$). The dosimeter is also limited to measurements with dose rates $>0.3 \text{ mGy s}^{-1}$, its minimum of detectability. DosiRat energy dependence constitutes another drawback that can be accounted for by calibrating the dosimeter at the beam qualities needed for dose measurements. Some uncertainties will still remain because of beam hardening and scattering effects. Thus, extensive studies of DosiRat energy dependence are required and currently under

progress to get a full description of the dosimetric system, for any irradiation condition that could modify the primary energy spectrum.

Overall, the dosimetric performances of DosiRat are very promising. Further developments are currently ongoing to make DosiRat a suitable dosimeter for several dosimetric applications. The reduction of the scintillator size will allow to satisfy the high spatial resolution needed for the millimetric beams delivered by small animal image-guided irradiators. In-phantom measurements will allow the validation of DosiRat as a useful device for quality assurance as well as for small animal treatment plan verification. With a proper characterization of its angular dependence, this system could also be used for small animal *in vivo* dosimetry to control if the delivered dose to the target is the prescribed one.

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