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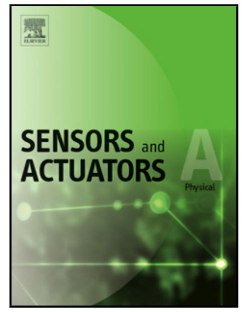

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Advanced Characterization of an Optical Fibre Sensor System based on an MPPC Detector for Measurement of X-ray Radiation in Clinical Linacs

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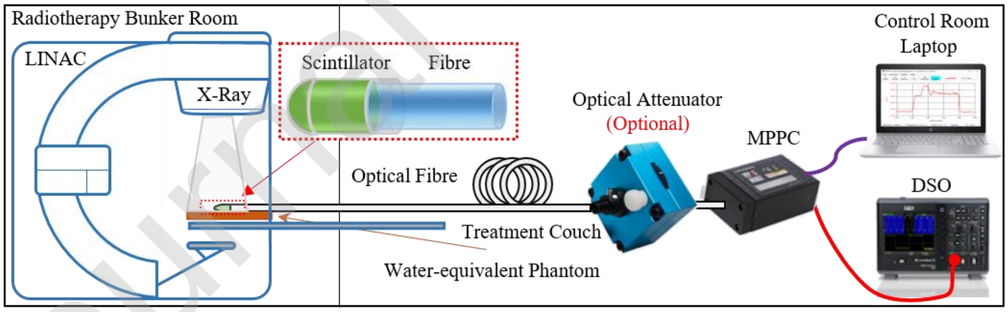
Graphical abstract



Highlights

- A reliable, accurate and in-vivo dosimetry system for measuring the radiation dose and profiling the X-ray beam during radiotherapy is reported.
- None of the currently available commercial dosimeters are capable of detecting radiation dose truly *in-vivo* during radiotherapy treatment.
- The system has been previously fully characterised using a novel in-house developed light emitting diode based system designed to fully replicate the light signal in the clinical setting.
- The use of this system has allowed a rigorous investigation of the characteristics of the multi pixel photon counting (MPPC) detector to establish the correct exposure conditions, which is essential for accurate dose measurement in the clinical setting.
- The use of the Optical Fibre Sensor (OFS) system is the first reported instance of ultra-high time resolution measurement (sub-microsecond) of the X-ray irradiation from a clinical based linac being used in the clinical setting.
- The system utilises both the digital count and analogue voltage outputs of the (MPPC) detector in a unique combination that allows the identification of a constant ratio (4.68) of the average incident photon number to the average output voltage exists which is universal across a wide range of OFS and can be used to determine the existence of correct exposure conditions.

Abstract:

A reliable, accurate and in-vivo dosimetry system for measuring the radiation dose and profiling the X-ray beam during radiotherapy is reported. Its dynamic range is investigated using an accurately controlled pulsed light emitting diode (LED) system. Highly resolved temporal analog and digital signals were captured from the analog and digital outputs of a multi-pixel photon counter (MPPC) detector when exposed to the LED system. The photon distribution of a low intensity pulsed LED light source was observed and is found to obey a Poisson distribution with changing light intensity. The average number of photons was obtained using the digital MPPC output signals which in turn allowed the appropriate intensity of the light source to be determined for the correct light exposure conditions for the detector. The average analog output voltage over a single 3 μs pulse is determined to indicate the intensity of the detected light. The MPPC detector output analog signal is limited to a narrow range (0.6 V to 1.4 V) to ensure adequate signal detection level (the lower limit) and prevention of entry into saturation (the upper limit) which also corresponds to a digital output signal range (in counts). An average photon number range of 3 to 7 for the digital output signal is established, which leads to the establishment of a unique and constant photon number to average output voltage ratio of 4.64 ± 0.10. Experimental results show that the establishment of this ratio is significant

as adherence to it ensures the correct exposure conditions of the MPPC and speeds up the measurement cycle in the clinical setting.

Keywords—Optical fibre sensors; Photodetectors; Radiation monitoring; Scintillators; Scintillation counters; Silicon photomultipliers; X-ray detectors.

1. Introduction

Radiotherapy is a widely used modern-day oncology treatment which is used in isolation or in combination with other modalities. The linear accelerator (Linac) [1] is a widely used means of delivering radiation for cancer therapy to the point of treatment (the tumour), and is [1] aided by sophisticated computer technology order to maximize the radiation dose to the tumor and minimize potential damage to the surrounding healthy tissue. The planned dose is accurately delivered via an energetic X-Ray (or electron) beam being in the MeV range from the linac being controlled using a computerized control system. The energy of the beam is matched to the depth in the tissue at which the tumor exists. For example, in the case of breast cancer treatment, the tumor may be at a relatively deep location below the skin surface. Therefore several tissue layers located at different depths will be irradiated, and their received radiation dose varies with depth and different types of tissue and organic material, including fat, skin, bone and blood vessels [2].

To maximize the dose received by the tumor and minimize the dose received by the surrounding healthy tissue, the X-ray beams are delivered to the tumor from several directions. A reliable, accurate, *in-vivo* (indolence), and real-time dosimeter used for monitoring the real dose of the tumor or other critical tissue is therefore desired in the process of Quality Assurance (QA). The most common commercial electrical conductivity dosimeters in the market include the Farmer ionization chamber (IC) [3], semiconductor [4], diamond [5], and all are very accurate but limited for use *in-vivo* (internal use) by their characteristics. Optical fibre and scintillating material-based sensors [6, 7] have the potential for use in dosimetry applications for clinical purposes, due to their miniature size (their maximum diameter is typically less than 3 mm, less than half that of many the commercial dosimeters, e.g. ionization chambers [8]), easy and low cost of fabrication [9]. The group of electrical conductivity dosimeters are ionized by the radiation, releasing the free electrons or ion-pairs directly, and they are directed via an externally applied electric field and collected. In the case of the scintillator-based dosimeter, the incident X-ray is converted directly into visible light without the need for any external power source and being purely optical [10]. The signal in the optical fibre is immune to other external electromagnetic radiation sources [11]. A 20 m long optical fibre is sufficiently long to guide the emitted visible light to the readout system located outside of the treatment room in the control room of the clinic.

The origin of scintillating materials applied to ionizing radiation detection can be traced back to 1900s but have gained more intensive attention since the early 1950s and the development of X-Rays as a medical diagnostic tool [12]. The primary mechanisms of the scintillation interaction with radiation is through direct fluorescence but additional ‘interfering’ light emission can be generated through secondary energetic electron emission processes including the photoelectric effect, Compton scattering, and pair production depending on the energy of the incident X-ray beam and the effective atomic number (Z_{eff}) of the absorbing material [13]. These energetic electrons are capable of ionizing the inner orbital electrons or stimulating them to be excited. In this process the excited ions relax back to their original states, and additional visible light (including fluorescence and phosphorescence) is emitted [2].

Due to the relatively low conversion efficiency (< 20 %) [10] of scintillator material for converting X-ray radiation into visible light, it was necessary to use a high gain (~10^6) photodetector for detecting the visible light signal at the distal end of the plastic optical fibre (POF). The device of choice in this case is the multi pixel photon counter (MPPC) (C11208-01) by Hamamatsu. It comprises an array of high speed, high sensitivity avalanche photodetectors and has previously been identified by the authors of this article [14] as a highly suitable device for the challenging detection requirements of this measurement. The time resolved response of the analog signal output (voltage) and digital signal (photon counts) of the MPPC to a range of different dynamic response tests based on a pulsed light emitting diode (LED) source has been fully investigated by Chen *et. al* [14]. In the investigation described in this article, the performance of the MPPC has been further investigated, including identifying a unique relationship between the analog and digital output signals under the correct exposure conditions of the detector.

The results obtained using the improved LED characterization method of this investigation are used directly to facilitate accuracy of measurement in a series of tests undertaken in the real clinical environment for a wide range of Linac beam capture conditions. In this scenario, accurate results of the optical fibre scintillator-based dosimeter system used in the clinic have been efficiently obtained, producing significant cost saving as a result of reduced time spent in the 'live' linac beam in the clinic.

2. Experimental Set-up

2.1 LED system based in laboratory

Fig. 1 illustrates the system consisting of the LED light source driven by an arbitrary waveform generator (AWG) and the readout system, including a 20 m long plastic optical fibre (POF, provide manufacturer name and fibre type as previously deployed by Chen et al [14]), an optical attenuator and MPPC detector, which is connected to a digital storage oscilloscope (DSO) and a laptop. In this work, the LED was driven using an AWG (also referred to as the signal generator (SG), Teledyne LeCroy WaveStation 2012) which replaced the in-house developed field programmable gate array (FPGA) based system previously deployed by Chen et al [14]. In comparison with the SG, the FPGA based system is flexible and good for operation outside the laboratory e.g. in-field calibration. However, it is limited in its minimum pulse duration of 5 µs. However, the duty cycle of the pulse signal from a SG must be larger than 0.1 %, e.g. if the signal produced from the SG has a 4 µs pulse width, then the period must be less than 4 ms and could be set to any period that less than 4 ms but more than 4 µs. In this investigation, the SG was preferred to the FPGA system. Firstly, the response of the MPPC to different pulse periods was fully investigated in [14] and hence the duty cycle limitation of the SG was not a factor in this investigation. Secondly, the pulse train of X-ray beam delivered from a medical linear accelerator (linac) is typically an ideal rectangular signal with a pulse width less than 3 ± 0.2 µs and the in-house developed FPGA system was not capable of producing an ideal rectangular pulse with a pulse width less than 3 µs. However, the rectangular pulse produced from the SG was close to ideal with a short rise time (several tens of ns). In this system, an in-line optical fibre-based attenuator was applied to reduce the incident light to the MPPC to a suitable level, and thereafter its setting remained fixed for the entire set of measurements in this investigation. The SG was set to produce a pulse train with a 3 µs pulse width and 1 ms period. But the amplitude of the pulse was changed using 9 discrete output voltages, which was also difficult to achieve accurately using the FPGA system. In this case, the intensity of the LED was accurately varied with the voltages in the range 2.8 V to 7.6 V in steps of 0.6 V.

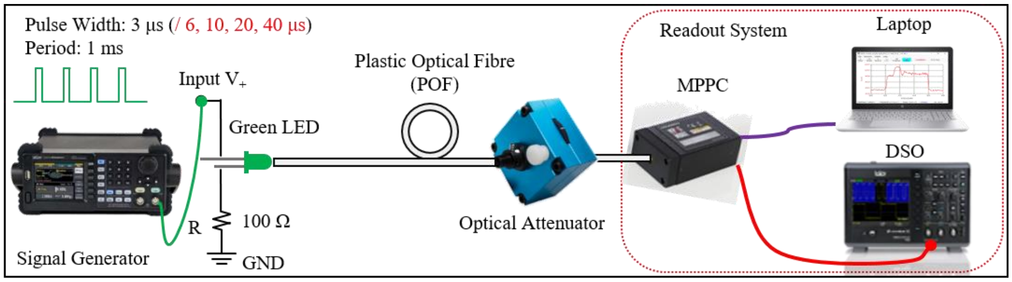


Fig. 1 The schematic diagram of the LED light source driver circuit with an AWG

2.2 Dosimetry system based in Galway Clinic, Ireland

Fig. 2 illustrates the schematic diagram of the dosimetry system to detect the X-ray beam delivered from the linac (Elekta VersaHD). The sensor with scintillating material was attached on the top of a water-equivalent phantom located in a centralized beam position 100 cm below the X-ray beam delivery window.

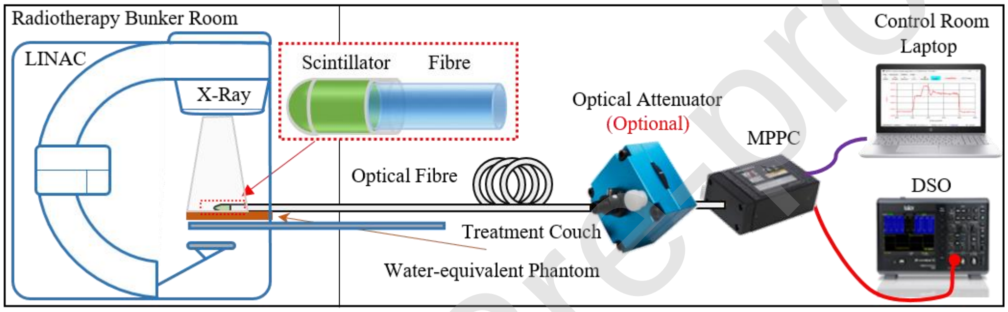


Fig. 2 Schematic illustration of the dosimetry system set-up in Clinical environment and an inset of the structure of the sensor

In this investigation, five different sensors were tested. Three of them were fabricated using the same scintillator material (cerium doped yttrium aluminum garnet, YAG:Ce) but in different physical forms, namely YAG powder, YAG ceramic and a YAG doped silica fibre. The two remaining sensors were fabricated using a commercially available organic scintillating fibre (BCF-12™: Saint-Gobain Crystals [15]) and the other was bare fibre with no scintillator material. These were labelled as YAG-P, YAG-C, YAG-F, BCF and Bare respectively (shown in Fig. 3 without YAG-F). The optical attenuator is optional for some measurements. The X-ray beam delivered from the linac beam delivered a 100 MU (monitor unit) dose at a 300 MU/min dose rate. The energy was set to 6 MV or 15 MV. The source to sensor distance was fixed at 100 cm. The field size was 10 × 10 cm^2. Under these standard conditions, 100 MU is equivalent to a dose of 1 Gy at the maximum dose depth (dmax). In this investigation the sensor was located on the surface of the phantom and hence the dose is always less than 1 Gy. However, this may be used as a guide value in establishing the correct exposure conditions for the MPPC detector.

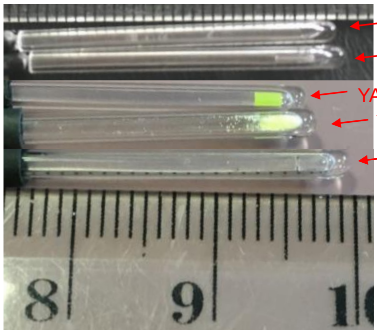


Fig. 3 A photograph of the sensors with different scintillator materials

3. Results and discussion

DSO (Teledyne LeCroy, WaveAce 1012) is only capable of capturing a single or very limited number of pulses of a pulse train due to its short recording duration, particularly when used in the highest time resolution e.g. for a resolution of 1 ns the total available recording window is 41 µs. The analog signal from the DSO is the electronically converted signal (volts) from the APD array representing the total photons accumulated by the MPPC detector. The sampling rate of the DSO means that the analog signal output is sampled every 10 ns. The shorter sampling interval provides more highly temporal resolved results. A sampling interval of 10 ns was used in this investigation. The proprietary software from Hamamatsu is designed to count the number of photons over the duration that the array is exposed to the light signal (gate time). There are two selectable options that allows the user to set up the conditions for acquiring data, namely gate time and threshold level. The gate time is analogous to the sample interval, and it can be set in discrete values in the range 1 ms to 100 ms. With this limited time resolution, it is not capable of profiling the pulse pattern (individual linac pulses are 1-3 µs in duration), but it can record over an extended time, period being more than 1 minute, which is sufficient to cover a whole linac treatment delivery cycle, this being typically no more than a few 10s seconds in duration. The threshold level represents the number of photons that the APD array requires to register a count (also referred to as the photon number). It is selectable only in a limited range of discrete values from 0.5 P.E. (Single Photon Equivalent) to 7.5 P.E. in a step of 1 P.E., e.g. in the case of the 1.5 P.E. value, the software only registers a count when 2 photons are detected simultaneously by the MPPC, if 1 photon or more than 2 photons are detected by the MPPC, the event is ignored by the software and no count is registered. While the software is clearly sensitive to the preset threshold value (it can be set too high or too low) the analogue voltage output is not subject to any such restriction, but its value must maintained in the range 0 to 3 Volts to avoid saturation effects of the detector [14].

The distribution of the photon number of a very low intensity level laser obeys a Poisson distribution [16]. In this experiment, it is shown that this rule is also obeyed when the LED light source was used in a very low but constant intensity (pulse amplitude) condition in order to avoid analog signal output oscillation. Under the correct exposure conditions the voltage output signal is characterized by highly oscillatory spikes and dips during the illumination period [14]. Therefore, an average voltage over the pulse period has been used to indicate the intensity of the detected LED light intensity. However, it is much more complicated to ascertain an average number to indicate the intensity of the detected LED light source using the count values provided by the proprietary software, as it requires at least three sets of count values using different P.E. values. Only in that case, may the Poisson distribution fitting be used to determine the average photon number ($\bar{n}$). It is a process that requires multiple repeated testing which is not an option in the clinical setting.

3.1 Analog output voltage results

Fig. 4 shows the 9 discrete analog signals captured using the variable pulse amplitudes of the LED drive circuit and their average voltages are shown using a dashed magenta line. It is clear that the average voltage increases when the pulse driver voltage amplitude (received light intensity) increases. For each frame in Fig. 4, the red solid line represents the LED driver signal and the green line the detected analog voltage signal. There is a very short delay time (circa 110 ns) between the driver signal and detected signal, which is caused by the time of flight in the 20 m length of transmission POF with a core refractive index of 1.49 [17]. Furthermore, in previous laboratory tests using a similar set-up but with a shorter length of transmitting fibre (2 m) no such delay was evident [14]. The average voltages of Fig. 4 can therefore be used to reflect the intensity of the detected light source.

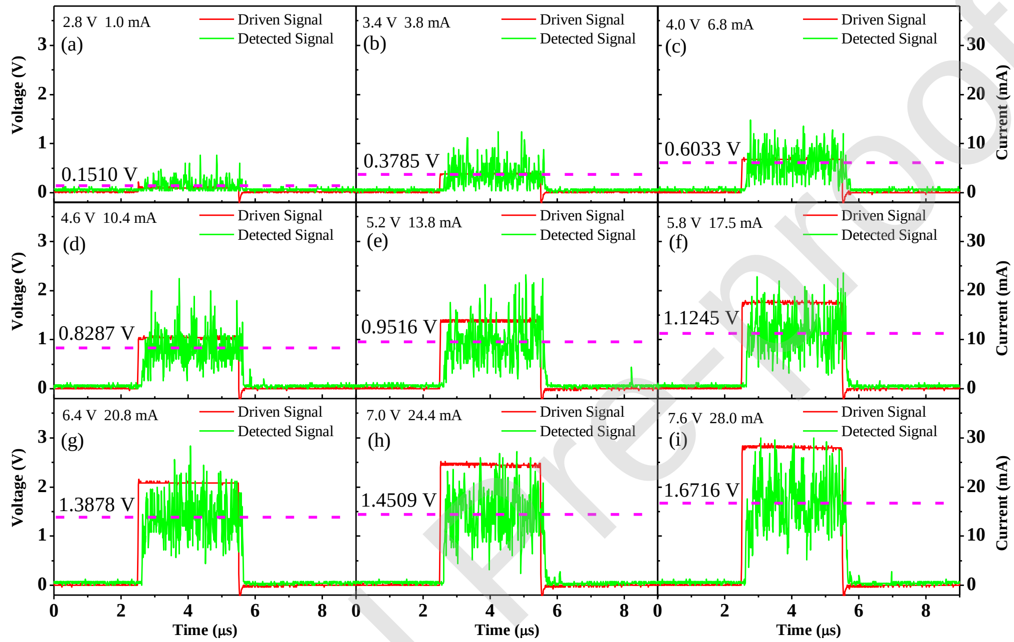


Fig. 4 9 sets of detected signals (in green) through the MPPC and driven signals (in red) of the LED obtained from the DSO

3.2 Digital results

Fig. 5 shows the 9 sets of the photon distribution with the Poisson fitting applied and each frame corresponds to the counts accumulated when the LED light source was driven using the 9 discrete applied voltages. For a given Frame of Fig. 5 (the frames correspond to the LED driver pulse values of Fig. 4) the red circle represents the average count value over the entire range of P.E. values as captured by the software and the blue cross symbols represent the probability of the existence of a certain photon number obtained using the Poisson distribution. This number is used to indicate the intensity of the detected light source and is called the average photon number ($\bar{n}$). As the proprietary software of the MPPC only has 8 discrete P.E. values of the threshold level, in each frame, only 8 discrete points are depicted.

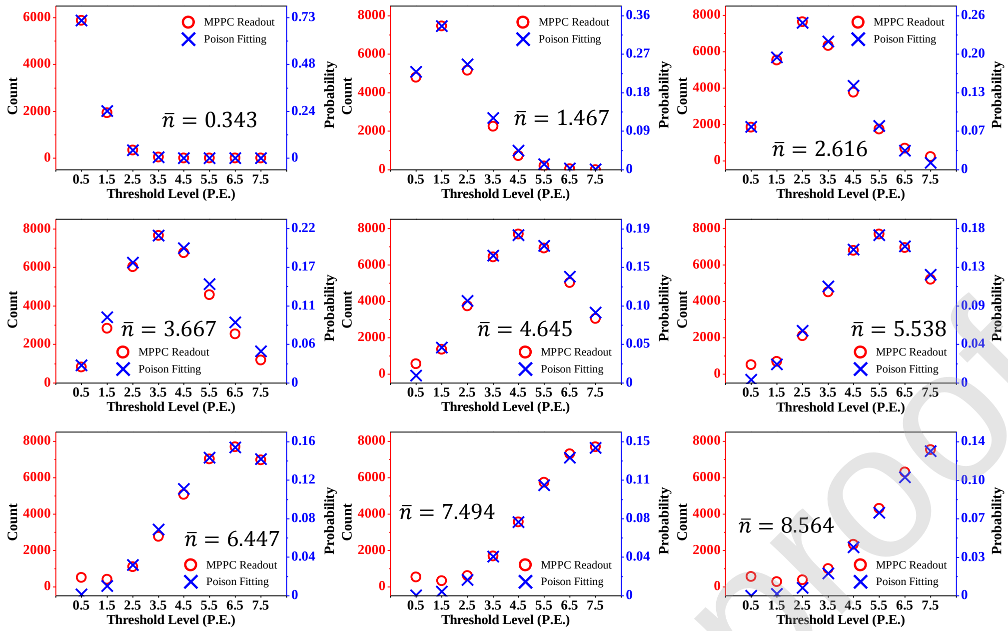


Fig. 5 9 sets of the photon counts (in red circle) from the MPPC software and their Poison Fitting (in blue cross) against different P.E. level

3.3 A comparison of the analog and digital signals obtained using the MPPC detector to the signal obtained using a spectrometer

To avoid saturation occurring in the MPPC detector, the intensity of the incident LED light was reduced to an extremely low level using an optical attenuator. A highly sensitive fluorescence spectrometer (Ocean Optics QE65000) was used to measure the received light at the end of the 20 m transmission POF when the LED source was driven using the SG with the same 9 pulse amplitude settings as described in the previous section. However, in this case, the attenuator was adjusted to be fully opened. In that case, the intensity of the LED light source was bright enough to be detected by the spectrometer.

Fig. 6 shows the comparison of the normalized intensity to a median level. The median level is used as a reference value to form the normalized intensity value (the vertical axis of Fig. 6). The normalized value is derived as the average value in Fig. 4(e) for the DSO and 5(e) for the digital output. The average voltages (red circles), average photon number (blue triangles) and the output signal from the spectrometer (green square) for the 9 discrete pulse amplitude settings are thus shown together in Fig. 6. In each case the value of the result obtained using the DSO, software (average photon number) and voltage (from the DSO) were normalized to their respective maximum values over the entire P.E. range. From Fig. 6, it is clear that the normalized intensity of these three different results exhibit very good agreement from the 3rd voltage level to the 7th level. However, in the case of the first two voltage amplitude settings, the results obtained from the software are much closer to the results captured using the spectrometer. In the case of the last two voltage levels (the higher values), the results captured using the DSO exhibit better agreement with the results from the spectrometer than the corresponding results from the software.

The deviation of the DSO at the lower end amplitude values is caused by the background noise and the relatively weak MPPC output voltage signal at these driver pulse values. It is clear in Fig. 4 (a) and (b) especially, that the low background noise is not zero and manifests as a few spikes (in the green line) even during the LED light off-phase and it cannot be removed or compensated for in a ‘single-shot’ time capture event. Due to the limitation of the availability of the number of threshold levels (the maximum is 7.5 P.E.) imposed by the MPPC proprietary software, the maximum count value of the detected light signal with higher intensity is shifted to the higher P.E. level (Fig. 5), such as 7.5 P.E. or higher. However, the software is not capable of acquiring the data for such conditions. Therefore, the acquired data is no longer optimal for the Poisson fitting to allow calculation of an accurate average photon number ($\bar{n}$).

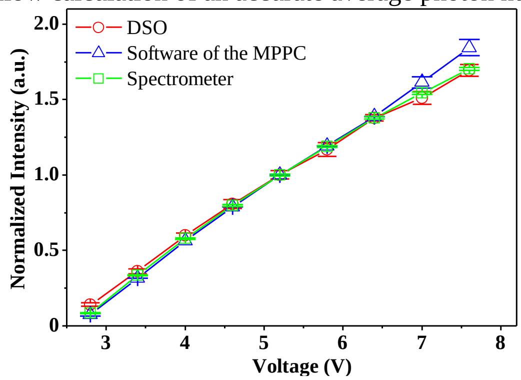


Fig. 6 The normalized intensity of the average voltage (DSO), average photon number (Software of the MPPC) and count value (Spectrometer) to the median level (the fifth value) against the 9 discrete LED driver pulse amplitude values

3.4 The relationship between the analog signal and digital signal

Since both the average voltage and the average photon number are used to show the intensity of the same light source, the average voltage and the average photon number ($\bar{n}$) should ideally share a constant relationship across the full range of light exposure conditions. The results of the previous sections indicate that this cannot be the case over the entire range, but there may exist a sub-range over which this is true. This section deals with a comparison of the analog signal output of the MPPC (in volts) and the count values obtained using the output of the MPPC software (average photon number) in order that the existence of the range referred to above may be identified.

In order to establish that the sub-range is suitable for all operating conditions of the light source, more experiments have been conducted. The SG was set to produce 4 more sets of driver signals using the same 9 discrete amplitudes (2.8 V to 7.6 V) and same pulse periods (1 ms) as described in the previous sections, but the 3 μs pulse width was replaced with the values 6, 10, 20 and 40 μs.

Fig. 7 shows the ratio of the average photon number ($\bar{n}$) over the average voltages obtained for the 5 LED pulse widths against the 9 different LED brightness (pulse amplitude) settings. The ratio of the LED driven using the SG with an amplitude from 4.6 V to 6.4 V are quite consistent and close to a unique ratio value of 4.68. For the case of the lower brightness LED, the higher normalized average voltage values in Fig. 6 obtained using the DSO compared with the corresponding normalized average digital output reduces the ratio; whilst for the higher brightness LED, the higher average photon number ($\bar{n}$) makes the ratio larger.

According to the results of Fig. 7, when the readout system of the MPPC device is connected to both a laptop computer (digital signal) and DSO (analogue output voltage), the intensity of the incident light source should be limited, such that the average output analogue voltage over a single pulse should be maintained within a range of 0.6 V to 1.4 V and the digital count output should correspond to the range 3 to 7 average photon numbers. Otherwise, the results of analog signal and digital signal outputs are not capable of accurately recording the intensity of the detected light source, i.e. some form of distortion may be incurred. In that case, it is sufficient that only the average analog voltage needs to be recorded to ensure that limits for signal quality are not exceeded. Also it is a relatively easy and quick process to obtain the ratio of the average voltage compared to the average photon number ($\bar{n}$). Hence, the need for repeated clinical measurements is removed by monitoring the digital (count) output voltage in each case of recording the linac output. This can be achieved 'on-line' and simultaneous with monitoring the analogue output.

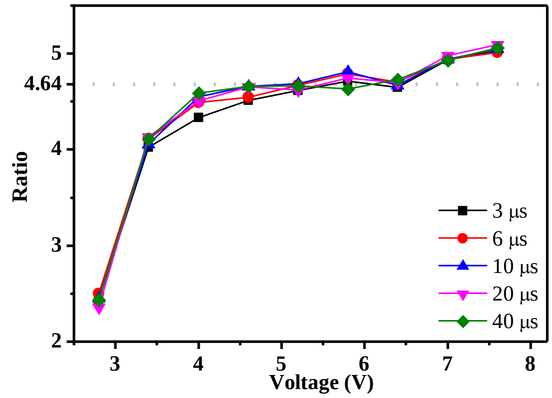


Fig. 7 The ratio of the average photon number ($\bar{n}$) over the average voltage

3.5 Clinical based results

The readout system described in this article has been applied to the dosimetry system for clinical purposes to measure the dose and temporal profile the single pulse of the X-ray beam delivered from a linac.

Fig. 8 shows the ratio of the average photon number over the average voltage (as used in the previous section) for a wide range of different sensor system configurations when irradiated under different X-ray beams delivered from the Elekta Versa linac. Table 1 provides the indexes of the X-labels in Fig. 8. The first item refers to the scintillator of the sensor, being YAG fibre, YAG powder, YAG ceramic, Bare fibre sensor or BCF organic scintillating fibre. The second term refers to whether the attenuator was applied or not. The third item refers to the energy of the X-ray beam, i.e. 6 MV or 15 MV. The last term refers to the dose rate of the X-ray beam. All sensors in this work have been exposed to the X-ray beam with a dose of 100 MU delivered at a rate of 300 MU/min.

Table 1 The rule of the X-labels in Fig. 8

Scintillator	Attenuator	Energy (MV)	Dose Rate (MU/min)
YAG-F/ YAG-P/ YAG-C/ Bare/ BCF	with/without	6/15	300

Fig. 8 shows good agreement with the LED based results of Fig. 7. The constant ratio is about a value of 4.91 ± 0.19, which is slightly higher than the value of 4.64 ± 0.10 of Fig. 7. As the sensors with a scintillator using YAG fibre and bare fibre sensor are previously known to have a low yield efficiency [11], the emitted light detected by the MPPC through an optical attenuator is too weak and hence lowers the ratio below the average value. When the same sensor with a YAG fibre without an attenuator was used the light output is known to be much higher in this case, and the more intense emitted light results in a higher ratio value than the average level (4.91 ± 0.19) in this case.

It is clear from the results of Fig. 8 that it is necessary to consider the whole structure of the system to obtain a better output, not just the scintillator. It is therefore necessary that the entire system is configured to achieve the appropriate incident intensity level into the MPPC device, i.e. that the correct exposure conditions are ensured. This must also be geared to the known likely exposure conditions arising from different linac settings, but this can be relatively easily monitored using the analogue output voltage signal of the MPPC.

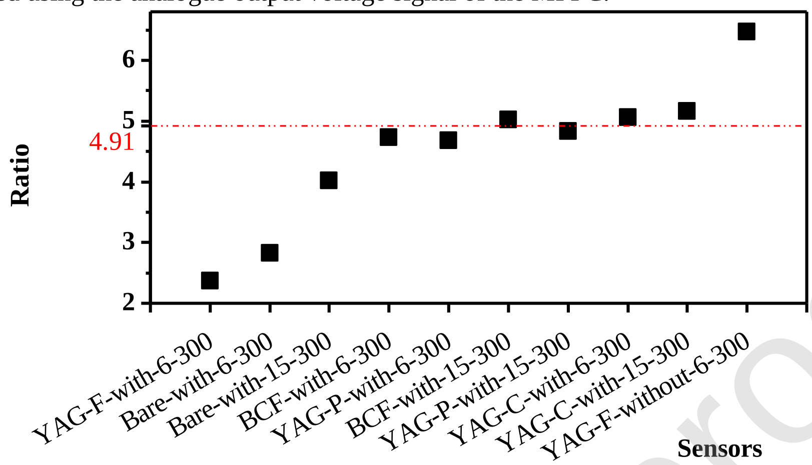


Fig. 8 The ratio of the average photon number ($\bar{n}$) over the average voltage in clinical based results

4. Conclusions

In conclusion, the MPPC is an excellent device in terms of its sensitivity and dynamic (time) response when used to measure the radiation dose and hence temporal profiling of the individual pulse of the X-ray beam as a part of forming an effective and accurate real time dosimetry system in the clinical environment. However, the dynamic range of the incident light to the MPPC device is strictly limited, not only to avoid saturation, but to meet the overall signal quality requirement that the ranges of 0.6 V to 1.4 V (the average voltage) and 3 to 7 (the average photon number) should be adhered to. The results of the investigation described in this article have shown that when these conditions are met, a constant ratio of 4.64 ± 0.10 of the average photon number to the average voltage exists. In that case, the data quality can be assured by simply monitoring the analogue output voltage of the MPPC during testing and ensuring that the average voltage value has stayed within range, which is relatively easily and quickly acquired.

It has also been shown that the dynamic range of the MPPC measured using the LED system also applies to the clinical dosimetry system. However, it is necessary that the response of the whole sensing system needs to be considered and not only the sensitivity of the scintillator as the sensing component in order to ensure that the MPPC is subject to the correct exposure conditions. It is also important that a prior knowledge of the linac operating characteristics are gained to ensure the latter condition is met.

As the ratio of the average photon number over the average voltage from the LED system and dosimetry system are not identical, further experiments are required to investigate why this may be so. Apart from the reasons mentioned above, it is likely that the photon-distribution of the light entering into the MPPC may lead to the difference between the LED-based system and the light source emitted by the linac source. As the photon-distribution of the low-level intensity LED light source follows a Poisson distribution, much better than the visible light emitted from the irradiated scintillation material, which is much closer to a sub-Poisson distribution.

Author statement

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Yong Sheng Ong, Software
Shuilin Chen, Resources
Sinead O'Keeffe, Funding acquisition, Supervision
Sean Gillespie, Resources, Investigation
Peter Woulfe, Resources, Investigation
Benxue Jiang, Funding acquisition, Resources
Ian Grout, Supervision
Elfed Lewis, Conceptualization, Investigation, Formal analysis, Writing – review & editing, Supervision

Declaration of Interest
None

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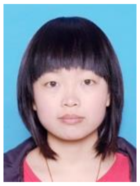
Author biography



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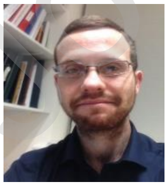

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