

Article

A Multi-Point Optical Fibre Sensor for Proton Therapy

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Abstract: As the technology to deliver precise and very high radiotherapeutic doses with narrow margins grows to better serve patients with complex radiotherapeutic needs, so does the need for sensors and sensor systems that can reliably deliver multi-point dose monitoring and dosimetry for enhanced safety and access. To address this need, we investigated a novel five-point scintillator system for simultaneously sampling points across a 74 MeV proton beam with a Hamamatsu 16-channel MPPC array. We studied the response across beam widths from 25 mm down to 5 mm in diameter and in multiple depths to observe beam penumbrae and output factors as well as depth-dose. We found through comparison to ionization chambers and radiochromic film that the array is capable of measurements accurate to within 8% in the centre of proton beams from 5 to 25 mm in diameter, and within 2% at 3.5 cm depth in water. The results from three trials are repeatable after calibration to within <1%. Overall, the five optical fibre sensor system shows promise as a fast, multipoint relative dosimetry system.

Keywords: optical fibre scintillator; multipoint; fibre array; proton radiotherapy; multipixel photon counter; photodetector



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1. Introduction

Radiotherapy continues to be a necessary step in 50% or more of cancer treatments worldwide, either alone or in combination with other treatments like surgery or chemotherapy. Across applications of radiotherapy, there are opportunities for multipoint, real-time or close to real-time sensing in both novel treatment modalities, due to increasing complexities, as well as in current and legacy treatments.

Radiotherapy systems employing or potentially benefiting from multipoint sensing range from particle radiotherapy [1–3] to brachytherapy [4–9] and teletherapy [10–13]. The simultaneous, real-time monitoring of multiple points inside and outside of an irradiated volume using fibre-based sensors is currently under investigation by several groups with some success, namely in the form of the commercially available Hyperscint™ scintillator dosimeter [14]—a system that utilises a spectrometer and employs a spectral separation method to chromatically remove induced Čerenkov and fluorescence from the total signal [15–17]. There are also many groups investigating proof-of-concept examples of fast, multipoint radiation dosimetry worldwide [18–22].

To date, though proton therapy is increasing in treatment sites and frequency [23], options available for multipoint sensing in protons, both pencil-beam scanned and passively scattered, are largely based on electronic sensors acting as both dosimeter and digitizer and therefore are located in the irradiation chamber [2,24–26], exposing sensitive elements to damaging radiation. A reuseable array where the sensing material is not subject to radiation damage and is (relatively) easily replaced at low cost, such as small volumes

of plastic optical fibre scintillators with long extensions leading to electronics outside of the irradiation chamber, would provide a cost-effective and long-lasting solution. Further, unlike the long processing time and one-time use of radiochromic film and the potential fading of signal over time and limited number of repeated readouts in optically stimulated luminescence (OSL) materials [27–29], after calibration, the analysis of scintillation signals obtained over an irradiation can be carried out in under 1 min using purpose-written Python code; there is no signal loss over time since the signal is digital and recorded in real time. These sensors can also be used repeatedly and, depending on the scintillator material used, experience little to no loss in signal at radiotherapeutic doses and dose rates [30]. The three-point Hyperscint™ system, which meets all the above criteria, presently consists of three scintillators and is capable of real-time dosimetry.

Many scintillation dosimeters are fabricated using plastic or silica optical fibres as transport fibres and organic or inorganic scintillators for radiation sensing [31–37]. When irradiated, scintillator atoms (inorganic scintillators) or molecules (organic scintillators) are excited, and as the product of the scintillator returning to a neutral state, light is emitted and radiation-induced luminescence (RIL) is detected. Optical transport fibres relay the scintillation light to photodetectors where the light is analyzed for intensity and/or energy, which is related to the amount of dose absorbed in the scintillator.

Optical fibre-based sensors made of organic materials like plastic are desirable for radiation detection and dosimetry due to their small size, simple and durable construction, and near-water equivalence. They are also flexible and waterproof, relatively insensitive to electromagnetic fields and temperature changes, and can deliver information very quickly—even in near-real time. Organic scintillators do have downsides such as quenching in regions of high linear energy transfer and induced Čerenkov light at proton energies over 320 MeV [38]; however, quenching correction is possible due to Birks' quenching correction formula, and since radiotherapeutic energy protons remain below the threshold of 320 MeV, plastic optical fibre scintillators remain viable dosimetric tools.

The scintillator employed in this study was the SCSF-3HF (1500) double-clad plastic scintillator fabricated by Kuraray (Chiyoda-ku, Tokyo, Japan), which was selected due to its sharp scintillation peak at 530 nm as well as its formulation for radiation hardness. A single sensor fabricated in the same manner as those used in this study was previously characterized using the TRIUMF 74 MeV proton therapy research centre (PTRC) beam, and shown to exhibit a linear dose and dose rate response as well as quenched but correctable (via Birks' formula) raw and spread-out Bragg peaks [37].

Due to the favourable results obtained in the previous study, an array of these sensors was assessed for its accuracy and reliability in medical beam monitoring and assessment applications. The following study describes the sensors used, the phantom designed and built to house them, and the electronics used to digitize signal measurement setup and calibration procedures. Lateral and axial profiles were assessed and compared to radiochromic films and a T1 ionization chamber (respectively). This novel 5-point array takes advantage of an established, radiation hard plastic scintillator fibre with potential to expand up to 16 points sampled simultaneously. It is also the first study where a commercially available multipixel photon counter (MPPC model C13369-3050ED-04; Hamamatsu, Hamamatsu City, Shizuoka, Japan) is utilised for this channel-by-channel purpose.

2. Materials and Methods

2.1. Sensors

Five sensors were fabricated as described by Penner et al. (2023) [37] using 3 mm lengths of 1 mm diameter SCSF-3HF (1500) scintillating fibre (Kuraray) as the radiosensitive element, and 15 m fibre lengths of Eska Premier 1 mm Simplex Optical Fibre (Mitsubishi Chemical Corporation, Chiyoda-ku, Tokyo, Japan) for light transport, as shown in Figure 1. The exposed scintillator/fibre-coupled ends were coated with a liquid electrical tape (Gardner Bender, New Berlin, Wisconsin, United States) to eliminate ambient light contamination. The total outer diameter was 2.2 mm. The distal (non-sensing ends) of the sensors were

terminated with SMA 905 connectors (Thorlabs, Newton, New Jersey, United States), and each sensor was coupled via a Thorlabs fixed attenuator (FA26M) to a 15 cm PMMA extension permanently affixed to a faceplate, which was mounted to a light detector (MPPC). The sensing ends of the fibres were inserted into a custom-made phantom for irradiations.

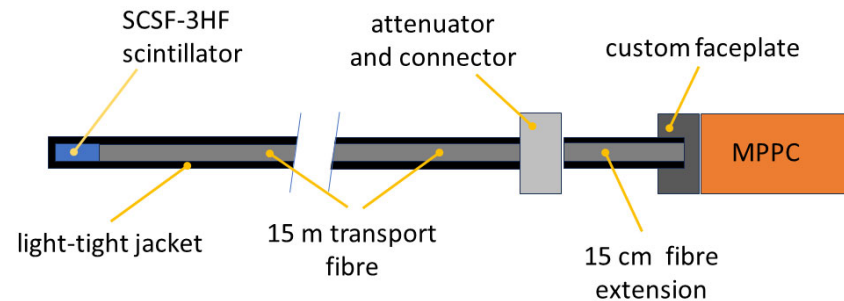


Figure 1. Schematic of one channel of the five-sensor system.

2.2. Phantom

A fibre holder acting as a phantom was modelled using Autodesk Inventor (2024) software and was 3D-printed on an Elegoo Mars Pro LCD 3D printer out of black resin cured to a density similar to that of PMMA (Anycubic plant-based UV resin). The resin cured to a density of 1.1 g/mL (PMMA has a density of 1.18 g/mL); thus, the materials of the fibres and of the applicator affected the beam in virtually the same manner. The phantom consists of a top (cover) piece and bottom (backing) piece with five channels to snugly fit the 2.2 mm jacketed fibres. Once printed and cured, the phantom was made homogeneous and light-protective to eliminate ambient light contamination. The phantom's five channels were each placed 4.69 mm apart centre to centre to capture the full range of the 25 mm collimator—the largest used when measuring different field sizes. The cover had a thickness of 3 mm, which is roughly equivalent to 3 mm of water, and could be added or removed as required by the measurements (Figure 2).

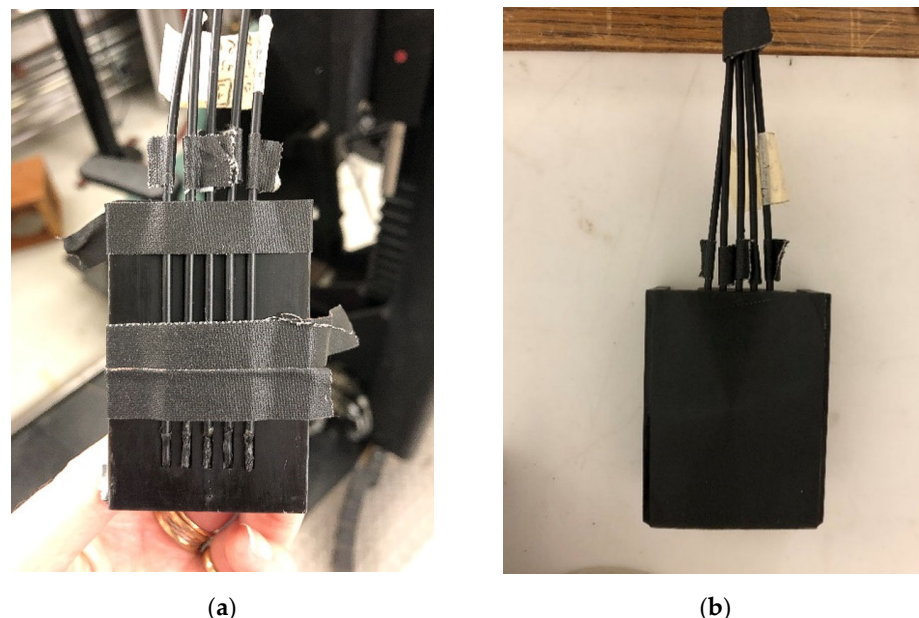


Figure 2. (a) Phantom backing piece for the sensor array with no cover and the five fibres inserted and taped in place. The tape at the top on the individual fibres served as a marker to ensure that the sensor remained at the correct depth in the phantom. (b) Phantom with 3 mm cover attached, viewed from the front with five sensors inserted.

2.3. Photodetector and Software

Once the sensors were assembled, they were characterized individually using the 16-channel MPPC. Coupling of the sensors to the MPPC was achieved by designing a custom black aluminium faceplate, on which 15 cm PMMA extensions were permanently fixed to each channel. Extensions withunjacketed fibre ends were inserted into a faceplate channel such that the end of the fibre was <0.5 mm from the centre of its designated MPPC (Figure 3a), and the connecting ends of the extensions were fitted with SMA 905 connectors. The multichannel MPPC module was operated at a threshold of 3.5 photon equivalent (p.e.), and 1000 μ s gate time, meaning 1000 data points were acquired per second.

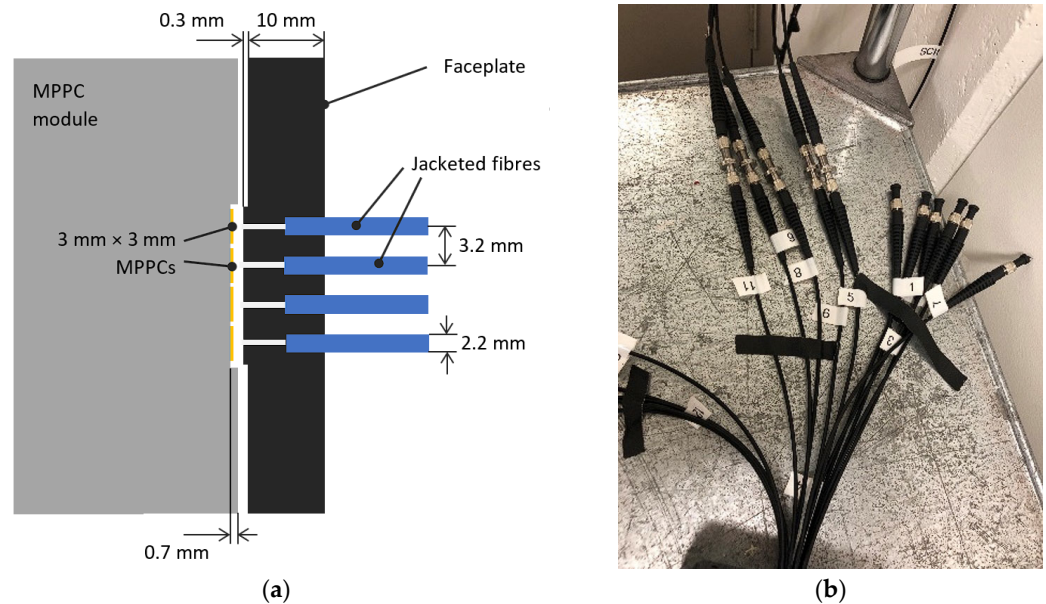


Figure 3. (a) Side schematic of MPPC (4×4) array with faceplate and 1 mm fibre ends. (b) Sensor extensions connected via attenuators to five sensors. Numbers on extensions note the channel number each extension is coupled to.

Each irradiation performed was immediately followed by a background recording, and every data point of the measured data was subtracted in Python by the background counterpart of the same channel. An averaging function was then used to smooth the data; in this case, every 300 points were averaged, and the resulting data were plotted as a function of time for each measurement.

Although every effort was made to fabricate identical sensors, much of the work was carried out by hand and inconsistencies could have arisen from the scintillator machining, coupling to the transport fibre, terminating the transport fibre, terminating and coupling the extension fibre, and the background and sensitivity of the individual MPPCs. Since the Hamamatsu multichannel MPPC array measured a dark noise of 1–4 kcps at 3.5 p.e., it was necessary to delegate each scintillator sensor permanently to its own MPPC (which corresponds to 1 of the 16 channels in the module) and include correction factors in the Python code to correct for any discrepancies between channels.

2.4. Irradiations

All measurements were carried out at the TRIUMF Proton Therapy Research Centre (PTRC), previously a clinical proton therapy facility for the treatment of ocular melanomas [39]. The beam used in all described tests was an extracted 74 MeV proton beam from the TRIUMF 500 MeV cyclotron at beamline 2C [40,41]. All measurements were taken in triplicate, and each one was followed by a dark measurement for background subtraction. All scintillator measurements were planned for a duration of 15,000 monitor counts (MCs), an arbitrary unit.

For each measurement configuration, the MC to dose calibration factor was measured using an Exradin T1 chamber (Standard Imaging Inc. Middleton, Wisconsin, United States). The actual number of delivered MCs varies due to the speed of the shutter which closes on the beam when the set number of MCs is reached. For this reason, each trial received a slightly different number of monitor counts. All background-subtracted signals were summed and then divided by the total number of actual MCs delivered in order to compare the different trials with the same beam energies.

2.5. Ionization Chamber and Radiochromic Film

A T1 ionization chamber (Standard Imaging, Middleton WI, USA) was used for dose calibration and depth-dose comparison, and EBT-XD films by GAFCHROMIC™ (Ashland Specialty Ingredients, Bridgewater, NJ, USA) were irradiated at all depths and collimator sizes for comparison and validation of the scintillator data.

Radiochromic film has many qualities, making it an excellent tool for radiotherapy measurements: it is not sensitive to visible light; it has very high spatial resolution (<1 mm, which may be higher depending on the resolution of the scan) making it well-suited for high-dose-gradient fields, such as those present in the measurements taken here [42]; and it is self-developing, easy to handle, and water-equivalent [43]. Although it may seem that radiochromic film would be a superior tool, it is impractical for everyday use because it is expensive and non-reusable, and developing time typically is >24 h. Radiochromic film can also be sensitive to air temperature, humidity, and ultraviolet light. Furthermore, the film must be kept in the same orientation when irradiated and when scanned due to the active component being oriented a particular way [44]; it can also exhibit a non-uniform dose response [43,45]. The EBT-XD film used here has a dose range of 0.4 cGy to 40 Gy.

The light absorption of the films, which corresponds to absorbed dose, was determined by scanning the films using a Xerox2 (Altalink C8170, Calgary, AB, Canada 600 dpi) scanner. The green channel pixel values were extracted from the image and inverted using Fiji (ImageJ2) computing software. The green channel was used because it has the largest dynamic range.

The resulting image data were analysed using Python code (version 3.9). The average background pixel value of the film was found by sampling a square 25×25 pixel area from each of the four corners, and the average pixel value of the area with the lowest standard deviation in pixel value was used as the background value. All pixels were then subtracted by this background. To convert the pixel value to dose in Gy, the film's non-linear response to dose also needed to be accounted for. Calibration data were obtained by irradiating 6 films with dose ranging between 0.38 and 14.37 Gy, corresponding to pixel values ranging between 15 and 113. The beam amplitude pixel values were fit in Python to the applied dose in Gy using the four-parameter Rodbard (logistics) function shown in Equation (1), which represents the non-linear behaviour of the film when irradiated [46,47]:

$$D = d + \frac{a - d}{1 + \left(\frac{x}{c}\right)^b} \quad (1)$$

where D is the dose in Gy; x is the pixel value; and constants b , c , and d are found using the Python function `curve_fit` from the module `scipy.optimize` with a fixed to 0 because the data are background-subtracted. The Python fit automatically generates the constants by using non-linear least squares to fit the function to the data. Response to radiation may differ between different film batches, even for the same type of film; therefore, the values of best fit must be found for each film batch. With the constants of best fit obtained (see Table 1), the pixel dose values of the film scans were converted to doses in Gy.

To compare the scintillator outputs to the radiochromic films, the films were analyzed as rectangular regions of interest of length 3 mm and width 1 mm at each pixel across the beam spot, producing a continuous profile while taking into account the shape of the scintillators.

Table 1. Parameters of best fit for the four-parameter Rodbard function, fit to the calibration data, and the maximum pixel value at which data points were capped.

Parameter	<i>a</i>	<i>b</i>	<i>c</i>	<i>d</i>	Maximum Pixel Value
Value of best fit	0	0.79	7.249	182.057	113

2.6. Measurements

2.6.1. Correction Factors

For the MPPC channel assessment, each MPPC channel was coupled in turn to the same fibre, and a uniform amount of light was relayed to each of the 16 channels over multiple trials. The light source was an HL-2000-LL (Ocean Insight, Orlando, FL, USA) broad-spectrum tungsten halogen light, and one end of an optical fibre was fixed at a distance where the light coupled into the fibre would not saturate the MPPCs. The other end of the fibre was coupled to each channel through multiple thresholds and gate times. In this manner, we identified the five MPPC channels with the highest repeatability and selected those as the ones to be used in this study. For each of the five channels selected, a fibre sensor was assigned permanently.

Once all five sensors were assigned to their channels, they were irradiated in a 74 MeV, 6 nA, uniform 5 cm × 5 cm beam to assess the inherent differences between the sensors and transport fibre channels when exposed to the same dose; this was repeated three times. For each trial, a set of correction factors was found and their average applied in subsequent signal analysis.

2.6.2. Field Size Response

To assess the capability of the array to accurately relay signal intensities with decreasing field size, 6 circular brass collimators with decreasing diameters of 25, 20, 15, 10, 7 and 5 mm were, in turn, fitted to the end of the proton nozzle, as shown in Figure 4. The array was placed downstream in the isocentre (roughly aligned with the beam centre using vertical and horizontal lasers) with the central scintillator in the approximate centre of the beam horizontally, and all scintillator volumes at the approximate centre vertically. Three measurements were made for each collimator.

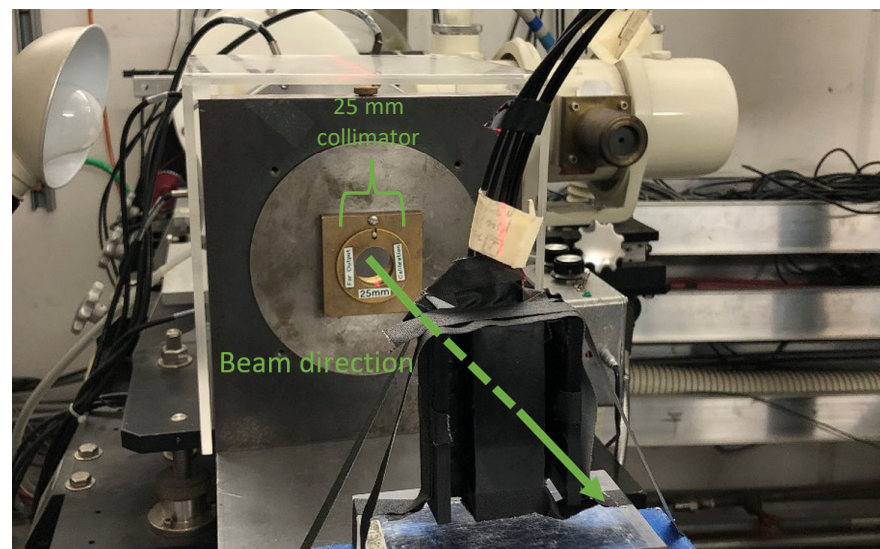


Figure 4. Field size response setup with five sensors in the phantom and 25 mm collimator.

2.6.3. Depth–Dose

To measure the relative dose deposition with depth in solid water, the array was placed directly behind increasing depths of solid water slabs (in 0.5 cm increments up to 3.5 cm), as shown in Figure 5. The beam was collimated to a 15 mm diameter and the

phantom cover was removed. The centres of the scintillators were roughly aligned using lasers and three measurements were made at each depth.

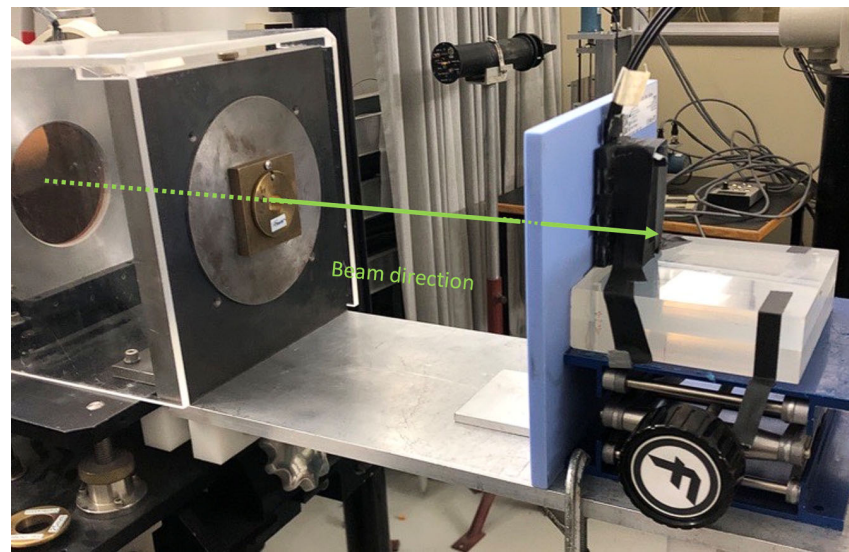


Figure 5. Setup for depth-dose measurements showing 1 cm of solid water directly in front of the phantom without the cover and holding the five sensors.

3. Results and Discussion

3.1. Correction Factors

Results for three trials in which all five sensors were irradiated with the same dose yielded very uniform results in each individual sensor (Figure 6). Repeatability in the three trials performed was over 99% in all sensors (Table 2). After applying correction factors calculated from the average of the three trials, the results were the same across the sensors to within <1% (Figure 6).

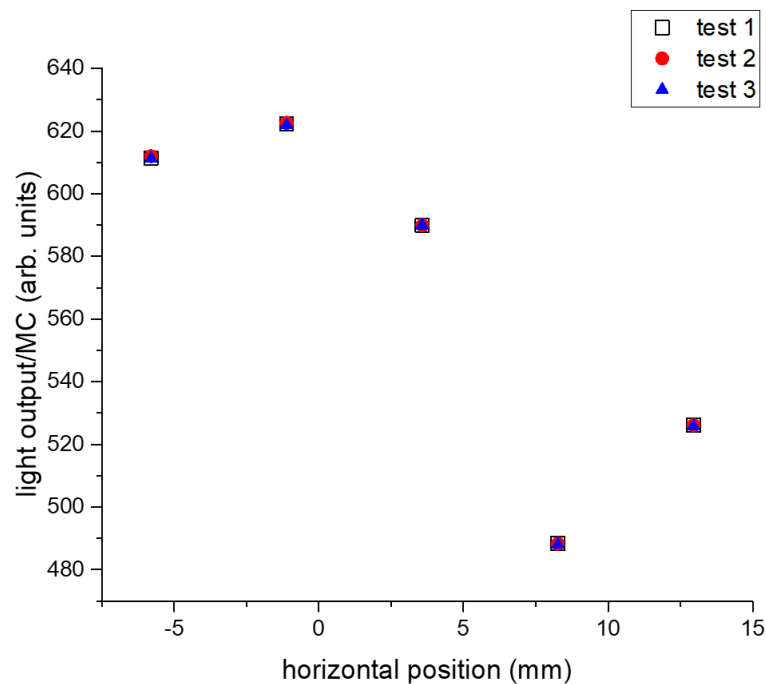


Figure 6. Results for 15,000 MC irradiations of a 74 MeV proton beam with a beam current of 6 nA. Results were background-subtracted and divided by the exact MC delivered but are otherwise uncorrected and shown for each of three individual trials for the five scintillators in the array.

Table 2. Background-subtracted total signal, mean, standard deviation and calculated correction factor for the three trials conducted. An extremely uniform response was observed for all sensors.

Test	Fibre 1	Fibre 2	Fibre 3	Fibre 4	Fibre 5
Test 1	9.460×10^6	9.628×10^6	9.128×10^6	7.557×10^6	8.141×10^6
Test 2	9.501×10^6	9.666×10^6	9.159×10^6	7.581×10^6	8.168×10^6
Test 3	9.546×10^6	9.712×10^6	9.213×10^6	7.622×10^6	8.212×10^6
Mean	9.502×10^6	9.669×10^6	9.167×10^6	7.587×10^6	8.174×10^6
Std. Dev	3.512×10^4	3.444×10^4	3.506×10^4	2.685×10^4	2.910×10^4
Correction factor	1.018	1.000	1.055	1.274	1.183

3.2. Horizontal Profile and Field Size

Horizontal profiles with five collimator sizes measured with the fibre array and compared to the EBT-XD film are shown in Figure 7. Both the film and scintillators were normalized to 1 at their highest point. A very good agreement can be observed between the film and the fibre array data.

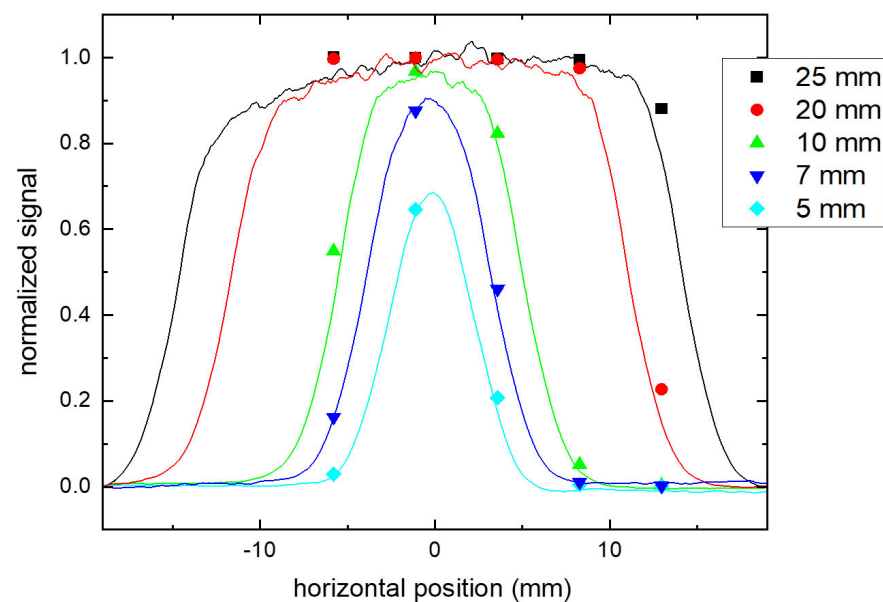


Figure 7. Horizontal profile of a proton beam collimated with different collimator sizes. The symbols are the fibre results averaged over three irradiations and compared to EBT-XD film results represented by lines. Both the film and scintillator array were normalized to 1 at their highest point with the largest collimator.

Overall, the measured profiles, as suggested by the five fibres, agree well with the film measurements at the five fibre positions; percent differences were low, ranging from 8% at best to 20% at worst in the penumbral region, probably due to the extended size of the fibres being compared to rectangles of 1 mm by 3 mm. This is comparable to 2021 field size results seen for the HyperscintTM multipoint system, which saw up to 32% discrepancy between scintillator results and the treatment planning system's calculation in the penumbral region for MeV photons [14]. The results are superior to those observed in a 3D multilayer OSL foil eye study performed by Sadel et al. in 2021, though the 3D nature of the 2021 study significantly increased its complexity [27].

The results of the fibre and film field size measurements are summarized in Figure 8, both normalized to the data of the largest field size. Again, a very good agreement can be observed between the fibre and the film data.

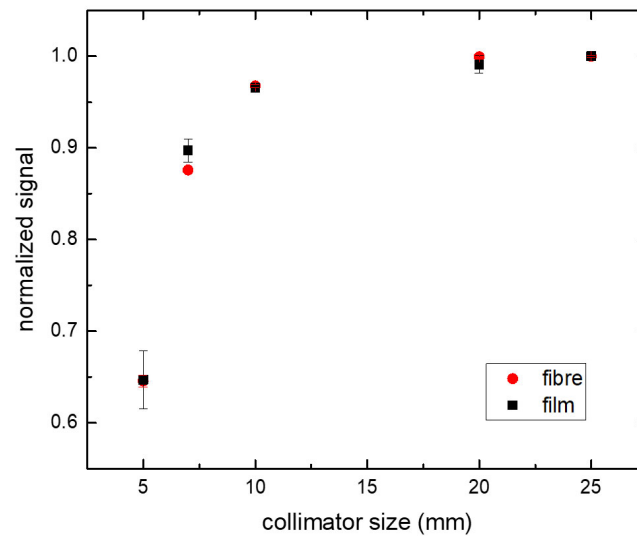


Figure 8. Field size results normalized to 1 at the highest point (25 mm) with varying collimator size and measured with the central fibre at horizontal position -1.1 , averaged over three irradiations; EBT-XD film results were averaged from -1.6 to -0.4 mm. Error bars represent the uncertainty in the outputs by adding, in quadrature, the standard deviation in the 25×25 background pixel range and the standard deviation in the 30×30 beam centre range.

3.3. Depth–Dose Response

Film and scintillator signal strengths over solid water depths of 0–3.5 cm were normalized to 1 at 0 cm depth and compared; some of the horizontal profiles are plotted in Figure 9. In shallow depths, the film and scintillator measurements were well aligned; though this was the case for field size comparisons, due to the 3D volume of the scintillator, the fibre data show slightly wider signals overall. The height of the film and scintillator results at depths greater than 1 cm do not agree as well, and these discrepancies are discussed along with depth–dose.

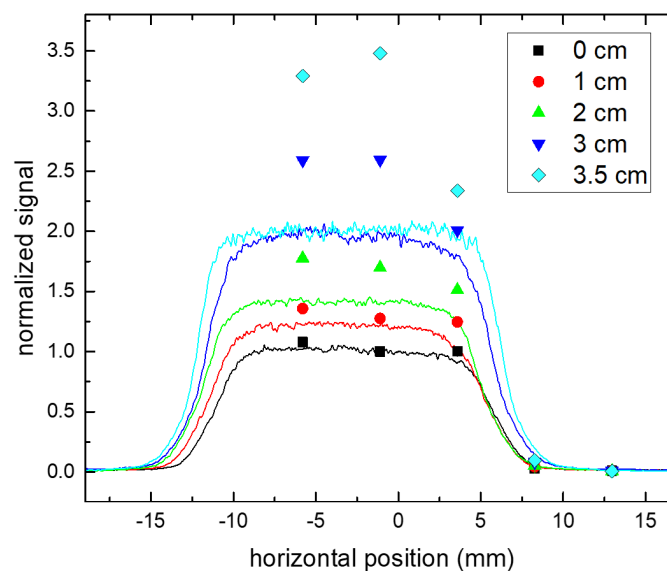


Figure 9. Normalized beam profiles compared to the five scintillators at different solid water depths with a 6 nA beam current and a 15 mm collimator size. The symbols are the scintillator results averaged over three irradiations and compared to EBT-XD film results (which are represented by lines).

A depth–dose comparison was performed between the scintillator at a horizontal position of -1.1 mm (indicated by the results to be the most centrally located sensor), the EBT-XD film, and the T1 ionization chamber (Standard Imaging, Middleton, WI, USA), see Figure 10. All three data sets were normalized to 1 at a solid water depth of zero (no solid water), and all show characteristic Bragg peaks. The different types of sensors begin to diverge at 0.5 cm depth, but the T1 and scintillator reconverge near the Bragg peak—around 3.5 cm depth—without any correction for quenching. The divergence seen is likely due to the effects of differing density and volume between the two sensors. The different materials (gas in the T1 ionization sensor vs. plastic in the scintillator) in combination with the different sensitive volumes—the volume in the ionization chamber being 0.056 cc and that of the scintillator being 0.0024 cc—results in the effective measurement points being relatively upstream (closer to the proton nozzle) for the air-filled T1 (as compared to the sensor’s geometric centre) and relatively downstream for the denser but smaller scintillator.

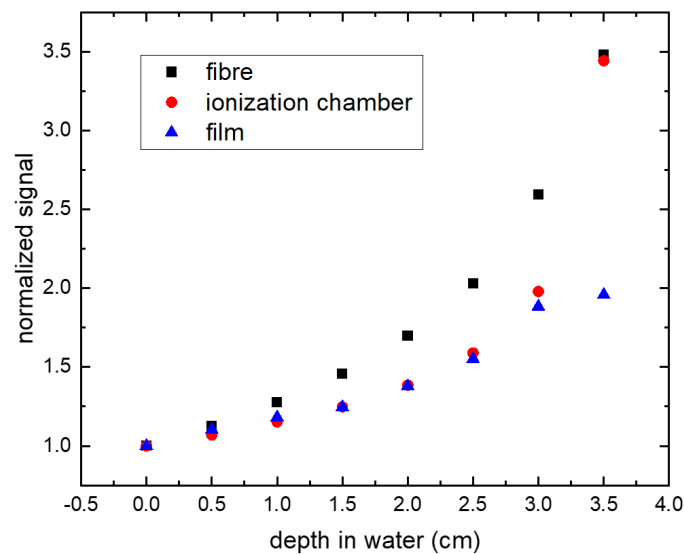


Figure 10. Normalized depth–dose comparison between the scintillator at a horizontal position of -1.1 mm, T1 ionization chamber, and EBT-XD film for a 15 mm collimator and a beam current of 10 nA.

The ultimate result of these variations is that the scintillator measurements appear higher at a shallower depth as compared to the T1 measurements at that same depth, as shown in Figure 10. With correction for the volume and density discrepancies, it is likely the two sensors would have more closely matched responses and the scintillator would display characteristic quenching in the Bragg peak region. A depth comparison with finer increments (mm rather than cm) would be very useful to this end. The film is noticeably saturated and shows an underresponse at 3.5 cm depth, though up to that point, it exhibits the same response as the T1 ionization chamber [43,45,48–50].

4. Conclusions

Here, we presented a multipoint optical fibre array that is able to measure proton beam profiles accurately (within 8%) and within one minute of irradiation. The assembly consists of a multi-channel MPPC, a custom faceplate and extensions to couple the fibre channels to the detector, five scintillator/fibre sensors, and a custom-designed phantom. The array shows promise in the field of multipoint relative proton dosimetry in its ability to relay relative dose and spatial information. The small size of the probes and the sensitivity of the photodetectors allow for a relatively high spatial resolution with a repeatability of the corrected sensor signal of >99% over the three trials for each measurement. The five-point 1D array of scintillators is effective for depth and horizontal relative dose information on par with lateral and axial data reported in other systems in early development [14,27]. Our system can be easily scaled up to 16 sensors because our MPPC detector has that many

channels. Spatial resolution can be improved in the future by choosing smaller scintillator sizes and optical fibre diameters; smaller scintillators would also reduce scintillator over-response. Finally, this multipoint array has potential use in fast particle discrimination in mixed fields resulting from proton irradiation, as described by Niedermeyer et al. [51], where multiple different sensors could be employed at once.

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