BEAM CHARACTERISATION USING MEDIPIX3 AND EBT3 FILM AT THE CLATTERBRIDGE PROTON THERAPY BEAMLINE

J. S. L. Yap[†‡1], J. Resta-Lopez[1], R. Schnuerer[1], C. P. Welsch[1], Cockcroft Institute, Warrington, UK

N. J. S. Bal[†§2], M. Fransen, F. Linde, Nikhef, Amsterdam, Netherlands

J. L. Parsons, University of Liverpool, Merseyside, UK

A. Kacperek, Clatterbridge Cancer Centre, NHS Foundation Trust, Wirral, UK

[1]also at University of Liverpool, Merseyside, UK

[2]also at Amsterdam Scientific Instruments, Amsterdam, Netherlands

Abstract

The Clatterbridge Cancer Centre (CCC) in the UK is a particle therapy facility providing treatment for ocular cancers using a 60 MeV passively scattered proton therapy beam. A model of the beamline using the Monte Carlo Simulation toolkit Geant4 has been developed for accurate characterisation of the beam [1]. In order to validate the simulation, a study of the beam profiles along the delivery system is necessary. Beam profile measurements have been performed at multiple positions in the CCC beam line using both EBT3 Gafchromic film and Medipix3, a single quantum counting chip developed specifically for medical applications, typically used for x-ray detection. This is the first time its performance has been tested within a clinical, high proton flux environment. EBT3 is the current standard for conventional radiotherapy film dosimetry and was used to determine the dose and for correlation to fluence measured by Medipix3. The count rate linearity and doses recorded with Medipix3 were evaluated across the full range of available beam intensities, up to 3.12×10^{10} protons/s. The applicability of Medipix3 for absolute proton therapy dosimetry is discussed and measurements are compared against the performance of EBT3.

INTRODUCTION

The Clatterbridge Cancer Centre (CCC) in the UK is the world's first hospital based particle therapy facility and has been successfully treating patients for ocular cancers since 1989. Originally built and commissioned for fast neutron therapy trials, it was shortly after converted into a proton beamline and ocular treatment facility [2]. Given its operational historically, many of the treatment line components and quality assurance (QA) equipment were constructed in-house. Several modifications to the accelerator and transport line were necessary, and so, there are clear differences with the beam and system initially commissioned. As the facility handles a busy patient load and achieves high rates of treatment success, the beamline itself is constrained by clinical requirements; any beam diagnostics checks are performed within the treatment beamline, located much further downstream of the last steering dipole magnet. These associated uncertainties with beam parameters, quality and unique clinical conditions are not common by modern standards and as a result proved both a challenging and promising environment for our tests.

Essential to this study and other work carried out at the beamline, is an understanding of the behaviour and parameters of the beam being delivered. In this case, simulation studies are instrumental to characterise the beam and for comparison with experimental measurements. As such, a model of the CCC beamline [1] has been developed using the Monte Carlo simulation toolkit, Geant4 [3] and efforts are ongoing to verify and validate the model. Transverse beam profiles, beam divergence and lateral spread are all important beam observables and indicate the accuracy of certain model parameters. Therefore, measurements with EBT3 film and Medipix3 contribute to validation.

METHOD

The Medipix3 is a hybrid pixel detector and uses a 500 μm silicon sensor, the active area is 28x28 mm^2. At 60 MeV, every single proton that impinges on the sensor deposits charge along its path and passes through the sensor, the charge is collected in pixels. Anything from single pixel points to long tracks can be observed depending on the incident angle of the proton. At this energy, a proton beam perpendicular to the sensor will produce mostly one pixel events per proton, whereas a non-perpendicular beam to the sensor will produce long tracks where many pixels are triggered. With high energy deposition, such as from these 60 MeV protons, charge will be shared between pixels through the sensor. Both due to the high charge deposition and also simple geometric effects, the charge cloud generated by the protons does not have to be centred on pixels. This means that the detector will over count compared to the number of protons which actually impinge on the detector. This undesired effect is called 'charge sharing' [4]. No attempt to compensate for it has been made here due to the reduced count rate capabilities since this measurement was intended to measure high count rates. This was a 'worse case' scenario due to the relatively thick Si sensor and the charge sharing effect contributing to a guaranteed increase in count rate, both of these effects push the count rate up.

These measurements were performed using a Medipix3 based detector using a SPIDR readout system [5, 6] from Nikhef, with the experimental layout as in Fig. 1. The

---

† Both authors contributed equally to this work
‡ jacinta.yap@cockcroft.ac.uk
§ navritb@nikhef.nl

Medipix3 chip was bump bonded to 500 μm of silicon in fine pitch mode, biased to +100 V, the chip was configured to collect holes.

The Medipix3 detector was placed at three locations throughout the treatment beamline and irradiated under varying beam conditions (Table 1). Sections of EBT3 GAFchromic^{TM} film [7] were also positioned in front of the detector (Fig.1) and irradiated simultaneously in order to directly compare performance.
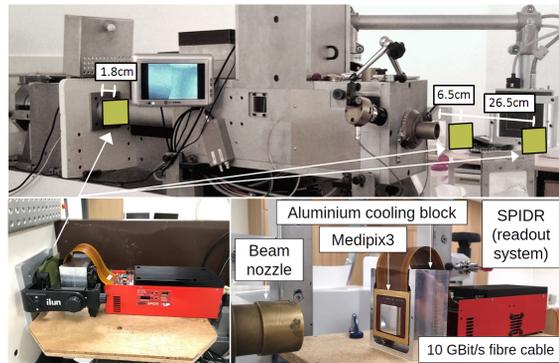
Figure 1: The film is held 3.5 cm directly upstream of the detector face. Film irradiations were performed just after the modulation box (integration zone) and at two distances after the final collimator.

EBT3 film is a standard radiochromic film dosimeter commonly used for quality assurance in radiation therapy. For patient specific or machine verification, it is essential to be able to perform checks with high accuracy and reliability, prior to delivering a course of treatment. Film dosimetry allows a visual representation of the beam as well as an analysis of the spatial characteristics and 3D dose distribution. EBT3 GAFchromic^{TM} film is made of a 28 μm layer of lucite, enclosed by 125 μm of polyester substrate on each side. Exposure to ionising radiation results in polymerisation of free radicals within the active layer, inducing the film to darken [8]. EBT3 self develops and the dark colouring or optical density (OD) is proportional to the extent of irradiation, increasing with absorbed dose. For hadron therapy however, the use of EBT3 film is limited due to quenching effects and saturation at points of high doses, such as the Bragg Peak [9].

RESULTS

Following complete development, the irradiated EBT3 pieces were scanned using an EPSON 750 scanner, saved as 48-bit images with no colour corrections at 150 dpi (dots per inch) and analysed using the image processing software ImageJ [10]. Using the software, a region of interest (ROI) is chosen, generating a plot of the grey values per pixel against distance. This gives a simple indication of the beam profile however due to saturation effects, it is necessary to evaluate these values against calibrated film measurements. Calibration measurements were performed separately using practised methods which are described in [9, 11–13],

Table 1: Run Data of Beam Conditions

Run #	Beam current (nA)	Time (secs)	Distance from nozzle (cm)
2	0.012	97.2	9.5
3	0.052	99.8	9.5
4	0.35	49.8	9.5
5	0.69	44.9	9.5
6	0.27	32.6	9.5
7	0.27	29.5	30.0
9	1.35	68.9	9.5
10	1.97	66.4	9.5
14	2.2	103.0	Integration zone
15	2.1	285.3	Integration zone

providing calibration curves. A curve fit to the red channel allows for correlation between grey pixel values and OD, to dose. Given these known quantities, it is then possible to convert the grey values from our irradiated experimental films to dose and plotted against position, determines the beam profile (blue plots in Fig. 2 and Fig. 3). It is also noted that the selection of the ROI is important and can affect the shape of the profile significantly. For our case, the ROI was specified such that there was total horizontal beam coverage and a sufficient height to generate a smooth profile.
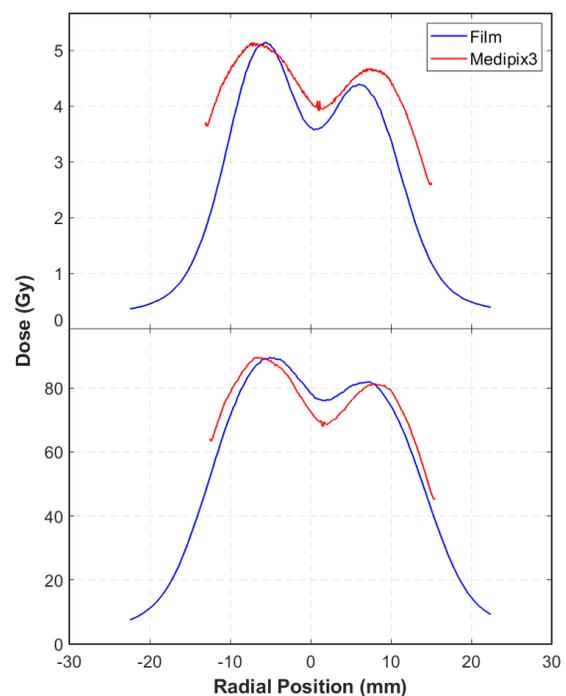
Figure 2: Beam profiles for runs 14 (top) and 15 (bottom).

Similarly, the images generated with the Medipix3 detector could also be evaluated using ImageJ for direct comparison. Minimal image processing was performed for the Medipix3 data, outlying pixels (noisy and dead) are interpolated from neighbouring pixels, the images are summed and
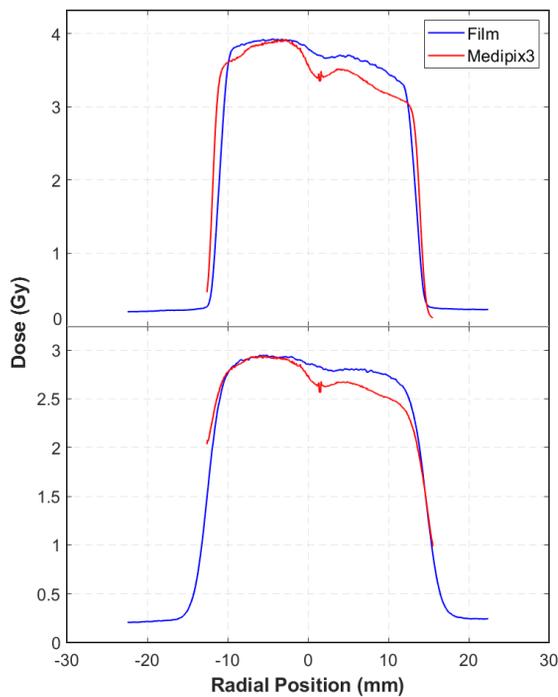
Figure 3: Runs 6 (top) and 7 (bottom), after the nozzle.
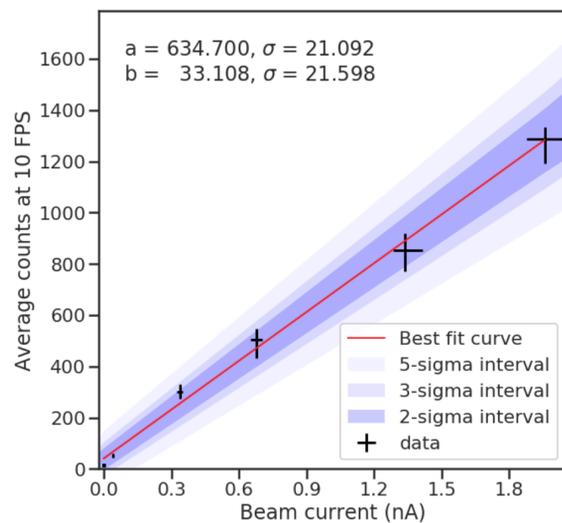
Figure 4: Count rate linearity over all active pixels recorded at 10 FPS (frames per second) for 6 beam currents between $0.012^{+13\%}_{-9\%}$ and $1.97^{+4\%}_{-7\%}$ nA.

then a Gaussian blur of sigma equal to 1 pixel is applied in order to reduce pixel-to-pixel gain variations. The grey pixel values correspond to hits and for simplicity, direct conversion to dose was done by scaling the pixel values to the film irradiated at the same experimental location. This preserves the grey value linearity but also correlates the magnitude of hits to a determined quantity, dose. We present the beam profiles obtained with Medipix3 (red plot in Fig. 2 and Fig. 3) at each of the concurrent film and detector locations. There is observable agreement between the distributions and any variances can be related to the image analysis uncertainties as well as fundamental differences between the detection processes. Most of the profiles do not extend completely across as high doses are detected right through to the edges of the sensor.

Figure 4 shows that the detector appears to have a linear response across the entire tested range of beam currents from 0.012 to 1.97 nA. There is relatively large uncertainty of the average count rate from the electrometer measuring the beam current, the residuals of the data are approximately 10% including this uncertainty. This uncertainty is dominated by the very infrequent, manual readings and secondly, the variation in beam current are in the order of 10%. This is why almost all of the uncertainty in the count rate likely is explained by the aforementioned reasons. Further measurements with relatively better beam current control would significantly reduce the count rate uncertainty and thus show the count rate linearity of the Medipix3 detector more reliably. The ultra low beam currents from runs 2 and 3 would not be possible to measure with other commonly used instruments with the precision of single protons, this detector therefore enables semi-destructive beam current measure-

ments from single events to 10^{10} protons per second with the possible temporal resolution down to 0.5 ms.

CONCLUSION AND OUTLOOK

Figure 4 shows that the Medipix3 detector appears to have sufficient count rate linearity and sensitivity for beam characterisation and quality assurance. Once cross-checked with another detector for accurate verification of the beam current, the impact of the 'charge sharing' effect on absolute dosimetry could be quantified. It is also anticipated that the Medipix3 chip configuration could be optimised significantly for high flux protons (>60 MeV) from the default X-ray (4-30 keV) settings. For future tests, it would be relevant to use the 'charge summing mode' on Medipix3 for furthering detector characterisation and possible mitigation of the 'charge sharing' effect. This has been shown to work with relatively low energy X-rays [4], however the energy deposition from these 60 MeV protons is significantly higher, and so should be tested. Further analysis will determine the energy deposited in the sensor and obtain definitive conversion factors for direct comparison with the dose distributions obtained by the film.

For this first test in a clinical proton beam environment, beam profile measurements comparable to EBT3 film were achieved with the detector, suggesting the potential of Medipix3 for quality assurance applications. Results obtained provide a clear representation of the actual spatial distribution and spread of the beam as necessary for beam characterisation. For the Clatterbridge eye proton therapy beamline, these beam profiles contribute meaningfully to benchmarking and complete validation of the simulation model.

ACKNOWLEDGEMENTS

REFERENCES

[1] Clatterbridge simulation model. `http://www.hep.ucl.ac.uk/pbt/wiki/Clatterbridge`.

[2] A. Kacperek. Proton therapy of eye tumours in the UK: A review of treatment at Clatterbridge. *Applied Radiation and Isotopes*, 67(3):378–386, 2009.

[3] S. Agostinelli et al. GEANT4 - A simulation toolkit. *Nuclear Instruments and Methods in Physics Research, Section A: Accelerators, Spectrometers, Detectors and Associated Equipment*, 506(3):250–303, 2003.

[4] E. N. Gimenez, R. Ballabriga, M. Campbell, I. Horswell, X. Llopart, J. Marchal, K. J. S. Sawhney, N. Tartoni, and D. Turecek. Study of charge-sharing in MEDIPIX3 using a micro-focused synchrotron beam. *Journal of Instrumentation*, 6(01):C01031–C01031, Jan 2011.

[5] J. Visser, M. van Beuzekom, Henk Boterenbrood, B. van der Heijden, J.I. Muñoz, S. Kulis, B. Munneke, and F. Schreuder. SPIDR: a read-out system for medipix3 & timepix3. *Journal of Instrumentation*, 10(12):C12028–C12028, Dec 2015.

[6] B. van der Heijden, J. Visser, M. van Beuzekom, H. Boterenbrood, S. Kulis, B. Munneke, and F. Schreuder. SPIDR, a general-purpose readout system for pixel ASICs. *Journal of Instrumentation*, 12(02):C02040–C02040, Feb 2017.

[7] Ashland inc. gafchromic™ radiology. `www.gafchromic.com/`

[8] M. Cristina Battaglia, D. Schardt, J. M. Espino, M. Isabel Gallardo, M. Cortés-Giraldo, J. Quesada Molina, A. Lallena, H. Miras, and D. Guirado. Dosimetric response of radiochromic films to protons of low energies in the bragg peak region. *Physical Review Accelerators and Beams*, 19, 06 2016.

[9] A. Mirandola R. Castriconi, M Ciocca, C. Sini, M. Schwarz S. Broggi, F. Fracchiolla, M. Martišíková, G. Aricò, G. Mettivier, and P. Russo. Dose–response of EBT3 radiochromic films to proton and carbon ion clinical beams. *Physics in Medicine and Biology*, 62(2):377–393, 2016.

[10] Imagej. image processing and analysis in java. `www.imagej.nih.gov/ij/`

[11] M. Vadrucci, G. Esposito, C. Ronsivalle, R. Cherubini, F. Marracino, R. M. Montereali, L. Picardi, M. Piccinini, M. Pimpinella, M. A. Vincenti, and C. De Angelis. Calibration of GafChromic EBT3 for absorbed dose measurements in 5 MeV proton beam and 60Co γ-rays. *Medical Physics*, 42(8):4678–4684, 2015.

[12] S. Devic, J. Seuntjens, G. Hegyi, E. B. Podgorsak, C. G. Soares, A. S. Kirov, I. Ali, J. F. Williamson, and A. Elizondo. Dosimetric properties of improved GafChromic films for seven different digitizers. *Medical Physics*, 31(9):2392–2401, 2004.

[13] J. Sorriaux, A. Kacperek, S. Rossomme, J. A. Lee, D. Bertrand, S. Vynckier, and E. Sterpin. Evaluation of Gafchromic EBT3 films characteristics in therapy photon, electron and proton beams. *Physica Medica*, 29(6):599–606, Nov. 2013.