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Fibre optic dosimetry in synchrotron microbeam radiation therapy

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Abstract
Synchrotron microbeam radiation therapy is a novel pre-clinical therapy method that uses high brilliance, spatially fractionated, low energy x-rays to deliver a very high dose rate. A conventional spatial fractionation for these beams is 50 μm microbeam width at 400 μm peak-to-peak separation. To perform dosimetry on these beams, a dosimeter with a high spatial resolution is required. We present a plastic scintillator fibre optic dosimeter that has been demonstrated to be able to resolve the microbeam structure. The advantages of such a dosimeter is the water equivalence, passive components exposed to the radiation field, relatively inexpensive components and simple fabrication. We use cylindrical BC-400 plastic scintillator optically coupled to a 1 mm diameter core plastic optical fibre. BC-620 reflective paint was coated on the end of the probe to maximise the capture of light. A silicon photomultiplier (SensL MiniSM) was used to measure the light output of the probe. A 50 μm resolution probe and 20 μm probe was tested at the Australian Synchrotron, on the Imaging and Medical Beam-Line. We were able to resolve microbeams with both resolution probes. The depth dose measurements matched that of a Pinpoint ionisation chamber except for an over-response at shallow depth.

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Fibre optic dosimetry in synchrotron microbeam radiation therapy

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Abstract. Synchrotron microbeam radiation therapy is a novel pre-clinical therapy method that uses high brilliance, spatially fractionated, low energy x-rays to deliver a very high dose rate. A conventional spatial fractionation for these beams is 50 μm microbeam width at 400 μm peak-to-peak separation. To perform dosimetry on these beams, a dosimeter with a high spatial resolution is required. We present a plastic scintillator fibre optic dosimeter that has been demonstrated to be able to resolve the microbeam structure. The advantages of such a dosimeter is the water equivalence, passive components exposed to the radiation field, relatively inexpensive components and simple fabrication. We use cylindrical BC-400 plastic scintillator optically coupled to a 1 mm diameter core plastic optical fibre. BC-620 reflective paint was coated on the end of the probe to maximise the capture of light. A silicon photomultiplier (SensL MiniSM) was used to measure the light output of the probe. A 50 μm resolution probe and 20 μm probe was tested at the Australian Synchrotron, on the Imaging and Medical Beam-Line. We were able to resolve microbeams with both resolution probes. The depth dose measurements matched that of a Pinpoint ionisation chamber except for an over-response at shallow depth.

1. Introduction

Synchrotron microbeam radiation therapy (MRT) is a novel pre-clinical therapy method that uses high brilliance, spatially fractionated, low energy synchrotron x-rays to deliver a very high dose rate within the microbeams. A conventional spatial fractionation for these beams is 50 μm microbeam width at 400 μm peak-to-peak separation (which is used on the Imaging and Medical Beam-Line at the Australian Synchrotron). To perform dosimetry on these beams, a dosimeter with a high spatial resolution and radiation tolerance is required. We present a plastic scintillator fibre optic dosimeter that has been demonstrated to be able to resolve the microbeam structure. The advantages of this dosimeter is its water equivalence, passive components exposed to the radiation field, relatively inexpensive components and simple fabrication.

Plastic scintillator dosimeters have been researched on for over two decades [1-3]. They have been applied to a range of medical radiation therapy fields, such as external beam therapy [4-6], brachytherapy [7, 8] and prostate cancer therapy [9]. The challenge to applying this type of dosimeter to MRT is the small sensitive volume required to properly resolve microbeams, which reduces the overall light signal that can be measured. We present the highest spatial resolution plastic scintillator fibre optic dosimeter found in the literature to date.
2. Materials and Methods
We use cylindrical BC-400 plastic scintillator optically coupled to a 1 mm diameter core Eska CK-40 plastic optical fibre (illustrated in figure 1). BC-620 reflective paint was coated on the end of the probe to maximise the capture of light. A silicon photomultiplier (SensL MiniSM) was used to measure the light output of the probe. The use of plastic components ensures that the probe is water equivalent for radiation interaction. BC-400 is energy independent over a large range of energies, linear, and temperature independent over 0-60 °C [1]. The maximum one dimensional spatial resolution of the probe is determined by the thickness of scintillator used. We present the results from two probes, with 50 μm and 20 μm spatial resolutions, with measurements of the intrinsic microbeam profile, and depth dose curves.

The intrinsic profile was measured by simply scanning the probe laterally across the field and sampling the detector response at 1000 Hz. The time scale is converted to position using the speed of the scanning motor. The depth dose curves were obtained by scanning the detector vertically through the field at each depth and integrating the response.

To reduce noise in the intrinsic microbeam profiles, a smoothing filter is run across the raw data. We use an unweighted moving average with a window size of 17 was used (further details on this method can be found in Archer et al 2017 [5]).

The experiments were performed at the Australian Synchrotron, on the Imaging and Medical BeamLine. The 50 μm probe was tested using Gammex Solid Water to provide attenuation and backscatter, while the 20 μm probe was tested in a water tank. To test the effect of the BC-620 paint on the light capture, the depth dose measurements with the 20 μm probe were repeated with and without the paint.

Figure 1. Left: Polished scintillator coupled to large core fibre (not to scale). Right: Microscope image of the probe tip, without the reflective paint. The scintillator is the dark, thin layer on the end. For maximum one-dimension resolution, the radiation beam must be incident perpendicular to the fibre axis. Reproduced from Archer et al [10].

3. Results and Discussion
We were able to resolve microbeams with both resolution probes (shown for the 50 μm and 20 μm probes in figure 2). The average microbeam widths measured with the 50 μm probe was (63 ± 2) μm, which compares favourably with a silicon strip detector of the same resolution [11, 12], which measured (62.4 ± 0.9) μm. The 20 μm probe measured the widths to be (52.1 ± 6.5) μm. The depth dose measurements with the 50 μm probe matched that of a Pinpoint ionisation chamber except for an over-response at lower depth. This result was similar to the 20 μm probe in water. The 20 μm probe was also scanned continuously through depth (Figure 4), which resulted in a similar over-response. Removing the BC-620 paint has no significant effect on the depth dose shape but reduced the total light output by 24% (seen in Figure 3).
Figure 2. (a) The intrinsic microbeam profile measured with the 50 μm resolution probe [13]. (b) The profile with the 20 μm probe [10]. Insets show the centre section of the microbeam scans.

Figure 3. Depth dose responses collected by the 50 μm probe scanning at (a) 10 mm/s and (b) 5 mm/s [13]. (c) 20 μm probe relative response with and without reflective paint. (d) Same probe normalised to 20 mm depth against ionization chamber [10]. Three BDAs are used with the 50 μm probe: 2.014 mm, 1.052 mm and 0.532 mm.

Figure 4. Continuous scan of the 20 μm probe in face-on mode, from 7 mm to 80 mm depth. Also shown is results from ionisation chamber. The 20 μm probe is calibrated to 20 mm depth against the ionisation chamber.

4. Conclusions
The results presented here demonstrates the ability for the fibre optic dosimeter probe to be able to effectively resolve microbeams. The depth dose results show a consistent discrepancy to the ionisation chamber results at low depths. The comparison with and without the BC-620 paint demonstrates that any radiation hardening effects cannot explain this discrepancy. The higher sensitive volume of the ionisation chamber makes a direct comparison of the results challenging.

The reduction in light output by removing the BC-620 paint has two reasons; a lower amount of light being reflected back into the fibre, and the higher Z material of the paint (TiO₂) having a higher
dose enhancement. As one of the main advantages of using a probe of this design is the water equivalence, not using the BC-620 paint is justified to remove any possible unwanted dose enhancements.

We anticipate that this dosimeter design can be refined to the point that it can be used in clinical quality assurance in the challenging field of MRT dosimetry. This will provide a relatively inexpensive, simple dosimetry tool to medical physicists responsible for MRT treatments.

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6. References