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Monte Carlo study of the potential reduction in out-of-field dose using a patient-specific aperture in pencil beam scanning proton therapy

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1 **Monte Carlo study of the potential benefit of using a patient-**
2 **specific aperture in pencil beam scanning proton therapy**

3

4 Running head: PBS PT using a patient-specific aperture

5

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11

12 **Abstract**

13 This study aimed at identifying the potential benefits of using a patient specific aperture in proton
14 beam scanning. For this purpose an accurate Monte Carlo model of the pencil beam scanning (PBS)
15 proton therapy (PT) treatment head at Massachusetts General Hospital (MGH) was developed based
16 on an existing model of the passive double-scattering (DS) system. The Monte Carlo code specifies
17 the treatment head at MGH with sub-millimeter accuracy. The code was configured based on the
18 results of experimental measurements performed at MGH. This model was then used to compare out-
19 of-field doses in simulated double-scattering (DS) treatments and PBS treatments

20 For the conditions explored, the penumbra in PBS is wider than in DS, leading to higher absorbed
21 doses and equivalent doses adjacent to the primary field edge. For lateral distances greater than 10cm
22 from the field edge, the doses in PBS appear to be lower than those observed for DS.

23 We found that placing a patient-specific aperture at nozzle exit during PBS treatments can potentially
24 reduce doses lateral to the primary radiation field by over an order of magnitude. In conclusion, using
25 a patient-specific aperture has the potential to further improve the normal tissue sparing capabilities of
26 PBS.

27

28 Keywords: proton therapy, pencil beam scanning, Monte Carlo simulation, patient-specific aperture

29

30 **1. Introduction**

31 Pencil beam scanning (PBS) proton therapy (PT) has the potential to deliver highly conformal fields
32 with reduced dose external to the primary field compared to double scattering (DS) PT. In PBS,
33 magnets are used to scan the proton beam laterally and the energy of the proton beam is typically
34 altered without beam modifying devices to achieve conformality in three dimensions. In contrast to
35 DS, PBS has the ability to conform the dose to both the proximal and distal edges of the target
36 (Lomax *et al.*, 2004). PBS treatments can also be delivered without patient-specific hardware, which
37 is required to achieve conformality to the lateral and distal edges of the target in DS (Kooy *et al.*,
38 2010). Not using patient-specific hardware reduces the amount of material in the path of the beam,
39 which decreases the total number of proton interactions occurring in the treatment nozzle, leading to a
40 lower neutron dose from PBS (Schneider *et al.*, 2002). The definition of in-field used in this work is
41 the region traversed by primary protons (including penumbra), while out-of-field is defined as the
42 dose delivered by secondary particles in the region not traversed by primary particles.

43 There are two principle methods of beam scanning used in PT, which we define as spot scanning and
44 continuous scanning. Spot scanning involves delivering dose in a series of finite steps. After the
45 delivery of each spot, the beam is turned off while the elements which steer the beam are reconfigured
46 to deliver the next spot. This reconfiguration may require a change in the beam position, energy or
47 both depending on the individual facility and the chosen scanning methodology. The primary

48 difference between continuous scanning and spot scanning is that during continuous scanning, the
49 beam remains on while the position is altered. Scanning fields at MGH are delivered via a series of
50 two-dimensional layers of constant proton energy (and hence range). The distal layer is irradiated
51 first, then the energy is reduced and the subsequent layer irradiated. This process continues until the
52 entire field has been delivered. Other PT centers may use a different scanning method, depending on
53 the hardware properties of the given facility.

54 Previous work by Paganetti *et al* (2004) has demonstrated the ability to accurately simulate DS proton
55 therapy treatments using the Geant4 Monte Carlo toolkit (Agostinelli *et al.*, 2003). Modeling the DS
56 treatment nozzle with sub-millimeter accuracy allowed the simulation of dose distributions from
57 spread out Bragg peak (SOBP) fields with accuracy, in terms of range and modulation, on the order of
58 a millimeter. Previous studies have also employed Monte Carlo for examination of the dose delivered
59 by primary and secondary particles (Paganetti, 2002, Dowdell *et al.*, 2009, Perez-Andujar *et al.*, 2009,
60 Zacharatou Jarlskog *et al.*, 2008, Zacharatou Jarlskog and Paganetti, 2008b). There is also potential in
61 the use of Monte Carlo simulations for patient specific dose calculation (Paganetti *et al.*, 2008).

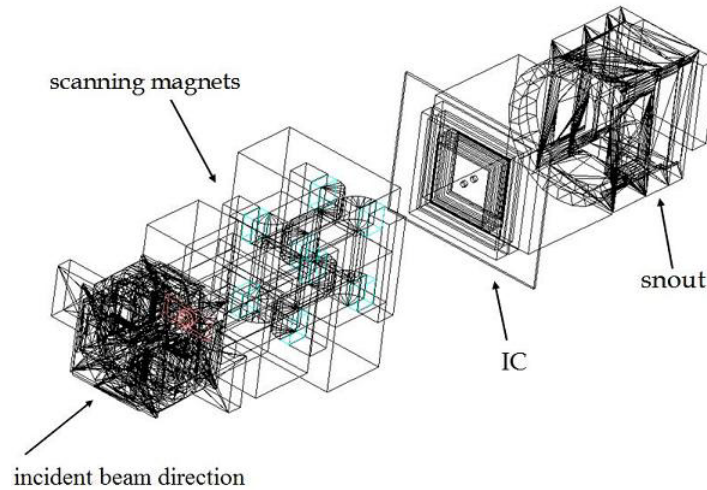
62 The aim of this work was to study the feasibility and potential benefit of incorporating a patient-
63 specific aperture at nozzle exit during delivery of PBS fields. To achieve this we have implemented a
64 model of the PBS treatment head at the Francis H Burr Proton Therapy Center, Massachusetts
65 General Hospital (MGH) using the Geant4 Monte Carlo toolkit. The model was used to simulate the
66 doses in PBS PT. The results of these simulations were compared to simulations of the DS treatment
67 head at MGH.

68

69 **2. Method**

70 *2.1. Monte Carlo modeling of a PBS nozzle*

71 2.1.1. *Nozzle Geometry*. The components of the PBS nozzle jointly developed by MGH and IBA (Ion
72 Beam Applications, Louvain La Neuve, Belgium) were modeled in the Geant4 code (version
73 4.9.0.p01)). The geometry as implemented in the Monte Carlo code is shown in figure 1.



74

75 **Figure 1: The pencil beam scanning treatment head implemented in the Monte Carlo code,**
76 **showing the incident beam direction, scanning magnets, ionization chambers (IC) and the snout.**

77 Protons are generated in the simulations at the entrance of the treatment head. Upon entering the
78 treatment head, the beam passes through the first and second scanning magnets, which are used to
79 scan the proton pencil beam horizontally and vertically respectively. The Monte Carlo code reads in
80 the prescribed lateral spot positions and proton energy at the treatment head exit via an input file
81 which is directly generated by the treatment planning system (TPS). These parameters are translated
82 into magnetic field settings using an automated script which converts the prescribed lateral position to
83 the required field strength in both scanning magnets based on the proton mass, the proton energy of
84 the current layer and the required deviation from the central axis. Using the file generated by the TPS
85 ensures that the beam moves throughout a single two-dimensional layer with the same scanning
86 pattern in the simulation and the clinical delivery. The three-dimensional dose distribution is delivered
87 in a series of two-dimensional layers of constant energy (and hence range). The layers are delivered
88 sequentially, commencing with the distal layer (highest proton energy) and concluding with the most

89 proximal (lowest proton energy) layer. The specification of the spot positions and the beam current
90 delivered in the input file also determines whether spot scanning or continuous scanning is delivered.

91 The magnetic fields generated by the scanning magnets are modeled as uniform fields inside the
92 magnet volume and the magnetic field strength set to zero external to the magnet volume. The field
93 centers are defined based on drawings provided by the manufacturer (Ion Beam Applications) and the
94 field lengths are defined by the effective lengths of the two scanning magnets. This same
95 methodology was also used for the specification of the magnetic fields in the study of Peterson *et al*
96 (2009).

97 As protons, or secondary particles, traverse the magnetic fields, their maximum step size in the Monte
98 Carlo was restricted to 2mm. Monte Carlo simulations typically model particle trajectories as a series
99 of straight lines and restricting the maximum step ensures more accurate modeling of the curved
100 trajectory of charged particles through the magnetic fields. Elsewhere in the treatment head, the
101 maximum step size was set to 100mm as high accuracy modeling of the trajectories was not important
102 in these areas of the treatment head and reducing the maximum step size would increase simulation
103 time. All particle interactions with different elements of the treatment head were still modeled, as the
104 particles trajectory and information (energy, momentum etc) are recalculated by default in Geant4
105 when crossing volume boundaries. It is thus unlikely that a particle travels up to the maximum step
106 size without changing its direction or energy.

107 After passing through the scanning magnets, the beam then passes through the ionization chambers,
108 which are used to monitor spot position and particle fluence. It is imperative to include the ionization
109 chambers in the model of a PBS nozzle as proton interactions with the chambers can result in wide-
110 angle scatter.

111 Finally, the beam passes through the snout, which can be used to hold a patient-specific aperture.
112 Each of the different snout sizes available in the clinic which were previously modeled in the Monte
113 Carlo for the DS system (Paganetti *et al.*, 2004) can also be included in the PBS simulation code. In

114 the design of the treatment head it was suspected that there may be clinical PBS cases in which
 115 patient-specific apertures or compensators are required (Kooy *et al.*, 2010). Examples of such cases
 116 would include treatments of tumors located in close proximity to critical structures (e.g. spinal
 117 column) where an aperture or range compensator could be used to sharpen the lateral or distal
 118 penumbrae respectively. The Monte Carlo code also has the capability of including these patient-
 119 specific devices.

120 *2.1.2. Definition of the proton phase space at treatment head entrance.* It has been shown previously
 121 that the phase space at nozzle entrance can be modeled using four parameters (energy, energy spread,
 122 geometrical sigma, and angular spread) and that these parameters can be treated independently for DS
 123 delivery simulations (Paganetti *et al.*, 2004). It was also shown that the beam delivery was insensitive
 124 to small variations in these parameters. However, PBS delivery can be expected to be more sensitive
 125 due to the lack of a scattering system. In this work, Twiss parameters are used in the Monte Carlo to
 126 govern the emission of protons at nozzle entrance. The Twiss parameters (α , β , γ) are calculated from
 127 a solution to the first order equation of motion. For an equation of the form

$$128 \quad \frac{d^2x}{ds^2} + k_x(s) = 0 \quad (1)$$

129 there exists a general solution of the form

$$130 \quad x(s) = \sqrt{\varepsilon\beta(s)} \cos(\psi(s) + \phi) \quad (2)$$

131 where ε and ϕ are arbitrary constants. Taking the derivative of equation 2 with respect to s yields

$$132 \quad x'(s) = \sqrt{\frac{\varepsilon}{\beta(s)}} \frac{\beta'(s)}{2} \cos(\psi(s) + \phi) - \sqrt{\varepsilon\beta(s)} \left(\sin(\psi(s) + \phi) \frac{1}{\beta(s)} \right) \quad (3)$$

$$= -\sqrt{\frac{\varepsilon}{\beta(s)}} (\alpha(s) \cos(\psi(s) + \phi) + \sin(\psi(s) + \phi))$$

133 where

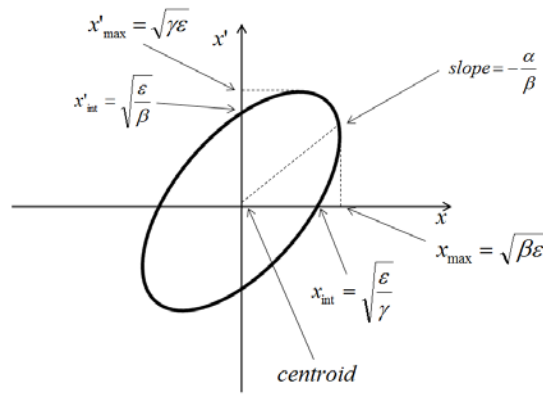
134
$$\psi(s) = \int_0^s \frac{d\tau}{\beta(\tau)}, \quad \alpha(s) \equiv -\frac{\beta'(s)}{2} \quad \text{and} \quad \gamma(s) \equiv \frac{1 + \alpha(s)^2}{\beta(s)} \quad (4)$$

135 The Twiss parameters govern the position and angular deviation of the protons generated in the Monte
 136 Carlo simulations. The emittance is given by

137
$$\varepsilon = \gamma(s)^2 + 2\alpha(s)xx' + \beta(s)x'^2 \quad (5)$$

138 and thus it is related to the area covered by the x, x' phase space ellipse shown in figure 2. Gaussian
 139 distributions ($\sigma = \sqrt{\varepsilon\beta}$) are assumed for the position and angular deviation.

140



141

142 **Figure 2: A two-dimensional ellipse of area $\pi\varepsilon$ based on the Twiss parameters α, β and γ .**

143 The Twiss parameters incorporated into the Monte Carlo code are functions of range, R (g/cm^2), and
 144 vary quadratically in a drift space. The variation in range arises from the increased lateral scattering
 145 as the proton energy (and momentum) is decreased. Additionally, lower energy beams have a wider
 146 momentum distribution, due to increased range straggling.

147 Initial values of the Twiss parameters were obtained from a transport calculation on the clinical
 148 beamline (Rohrer, 2007, Brown *et al.*, 1980). Manual optimization of width of the Gaussian
 149 distributions of position and angular distribution was performed such that the spot size obtained at
 150 isocenter in air in simulations of different proton beam energies matched the results obtained from

151 measurements at the clinical PBS system at MGH using a Wellhofer MatriXX detector (IBA
152 Dosimetry).

153 The Twiss parameters (α, β, γ) together with the emittance (ε) describe the beam trajectory at the
154 nozzle entrance. The initial energy of the proton is defined by the range required (specified in input
155 file from TPS) and the calculated initial energy spread distribution at nozzle entrance.

156 *2.1.3. Geant4 physics models.* The different physics models available in the Geant4 toolkit were
157 compared in a previous study by Zacharatou Jarlskog and Paganetti (Zacharatou Jarlskog and
158 Paganetti, 2008a) with the aim of finding the best models for use in Monte Carlo simulations of PT.
159 The results of this previous study dictated the physics models used in the PBS Monte Carlo code.

160 To model the electromagnetic interactions, the G4EmStandard (Agostinelli *et al.*, 2003) model was
161 used. This model governs the interactions of photons and all charged particles which have energy
162 greater than 1keV. The Bethe-Bloch equation is used for specifying the energy loss of hadrons of
163 energy greater than 2MeV. Inelastic hadronic interactions were modeled using a binary cascade
164 (G4BinaryCascadeFolger *et al.*, 2004) whilst the particle energy of is greater than 100MeV. Once the
165 energy of the particle falls below 100MeV, the precompound model (G4PreCompoundModel) is
166 invoked. Elastic interactions are governed by the UHElastic model (Ivanenko, 2006).

167 The DS code previously developed by Paganetti *et al* (Paganetti *et al.*, 2004) has been shown to
168 predict absolute doses in water within an accuracy of 1.5% compared to ionization chamber
169 measurements (Paganetti, 2006). Since we have used the same physics models for the Monte Carlo
170 simulations in this study, one could expect comparable agreement using the PBS code for absolute
171 doses. The comparisons performed in this study only considered relative doses, which may lead to
172 better agreement than 1.5%.

173 In terms of secondary radiation, extensive validation of the nuclear models used in the Monte Carlo
174 has been previously undertaken through comparisons with Faraday cup measurements (Zacharatou
175 Jarlskog and Paganetti, 2008a). Direct comparison of the Monte Carlo doses with ionization chamber

176 measurements external to the primary field in proton therapy demonstrated the suitability of the
177 chosen physics models for simulation of out-of-field doses in PT (Clasie *et al.*, 2009).

178 2.2. Calibration and validation of PBS Monte Carlo code

179 Depth dose curves were measured using a plane-parallel chamber with an entrance window of
180 diameter 84mm (PTW Freiburg GmbH). The depth dose curves were measured in a water tank for
181 proton pencil beams at five different ranges, 8.92g/cm², 12.64g/cm², 15.92g/cm², 21.1g/cm² and
182 25.15g/cm². The energy spread of the beam at nozzle entrance was determined by minimizing the χ^2
183 of the Monte Carlo results against experimental data for different energies over the range of
184 therapeutic energies deliverable at MGH. Combining the specification of the energy distribution at
185 nozzle entrance with the Twiss parameters allows the specification of all properties of the proton
186 beam required for Monte Carlo simulations.

187 The lateral deviation of the proton beam from the central axis is directly related to the strength of the
188 magnetic field in the scanning magnets. The script used to convert lateral position to scanning magnet
189 field strength in the Monte Carlo code was verified to ensure that the lateral spot position observed in
190 simulations matched the position prescribed by the input file within 1mm at five different ranges,
191 8.92g/cm², 12.64g/cm², 15.92g/cm², 21.1g/cm² and 25.15g/cm².

192 The simulation of a complex two-dimensional test pattern was compared to measurements using a
193 Beam Imaging System (BIS) (Ion Beam Applications Dosimetry) at isocenter in air. The BIS uses a
194 30cm x 30cm scintillator and a CCD camera to capture the image. The energy deposition in the Monte
195 Carlo simulation was obtained at isocenter in air using a 30cm x 30cm x 0.2cm volume of water with
196 the front face placed at isocenter and the maximum step size restricted to 0.02mm. The lateral voxel
197 size used in both the Monte Carlo simulation and experimental measurement was 0.73mm x 0.73mm.
198 Monte Carlo simulations and experimental measurement were compared via gamma analysis (Low
199 *et al.*, 1998). The scanning pattern used in this comparison validates all aspects of the Monte Carlo

200 simulations (other than range), by combining position, spot size and dose delivery checks into a single
 201 field.

202 Table 1 shows the different parameters considered in the PBS Monte Carlo code and the method of
 203 calibration and/or validation.

204 **Table 1: Methods of calibration and validation of the different aspects of the PBS Monte Carlo**
 205 **code used in this study. (MC = Monte Carlo, BIS = Beam Imaging System)**

Parameter	Configured and validated by
Proton range	Depth dose curves in water tank
Energy spread at nozzle entrance	Minimizing χ^2 of MC results against experimental measurement in water tank
Scanning magnet field strength	Measurement of lateral spot positions for different ranges using the BIS Complex 2D irradiation
Spot size	Calibration of Twiss parameters (α, β, γ) Complex 2D irradiation

206

207

208 *2.3. Monte Carlo simulations of clinical prostate field*

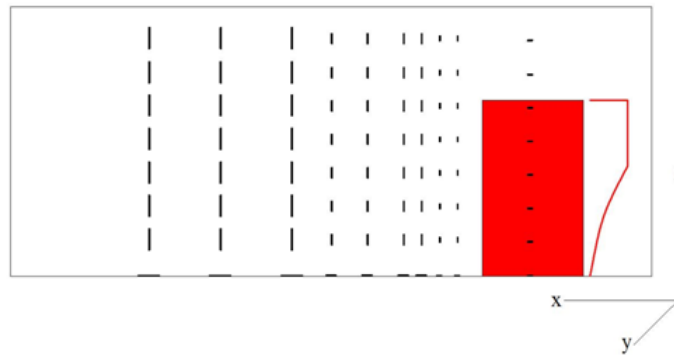
209 A prostate field was chosen because of the typically large range used in these treatments. The lateral
 210 field size is defined in DS as the area contained within the projection of the patient-specific aperture
 211 upon the phantom. The PBS field was generated to match the lateral dimensions of the DS field. The
 212 PBS field had a range (R_{90}) of 22.8g/cm^2 , a modulation (M_{90-90}) of 10.4g/cm^2 and a lateral field size of
 213 45.2cm^2 .

214 All simulations of the clinical prostate field were performed in two separate parts. In the first step, the
215 primary protons (and secondary particles) were transported through the treatment head (either DS or
216 PBS), resulting in the generation of a phase space file at nozzle exit which contained the energy,
217 position and direction of all particles which exited the treatment head. The properties phase space files
218 were then used to define the initial conditions of the second step which transported the particles from
219 nozzle exit into the phantom. The maximum step size was restricted to 0.2mm for all particles in the
220 phantom. Using the phase space allowed the field to be simulated upon the phantom multiple times
221 whilst requiring calculation of the particle transport through the treatment head only once, thereby
222 making the simulation process more time efficient. Each full simulation (treatment head and phantom)
223 took approximately 5 days to complete. This long simulation time was required to achieve acceptable
224 statistics in the scoring volumes distal to and at large lateral distances from the primary field.

225 In addition to conventional PBS, the aperture used in the DS simulations was placed at nozzle exit for
226 a series of PBS simulations. The primary fields were incident upon a Lucite phantom of size 90cm x
227 26cm x 37.76cm in all the simulations for all delivery techniques.

228 In addition to scoring volumes along the central axis, distances from the lateral field edge of 2.5cm,
229 5cm, 7.5cm, 10cm, 15cm, 20cm, 30cm, 40cm and 50cm were considered at depths of 4.72cm,
230 9.44cm, 14.16cm, 18.88cm, 23.6cm, 28.32cm and 33.04cm in Lucite. To increase statistics, larger
231 volumes were used out-of-field where the dose gradient is not as steep. The volume sizes varied based
232 on the lateral distance (x) from the field edge. The sizes used were $3.8 \times 3.8 \times 1.1 \text{ mm}^3$ ($x \leq 5\text{cm}$), 7.6
233 $\times 7.6 \times 1.1 \text{ mm}^3$ ($5\text{cm} < x \leq 20\text{cm}$) and $15 \times 15 \times 1.1 \text{ mm}^3$ ($x > 20\text{cm}$) (see figure 3). A similar
234 methodology was adopted in the previous work of Clasié *et al* (2009).

235



236

237 **Figure 3: Diagram of the scoring volume positions simulated. The black rectangles show the**
 238 **position, orientation and relative size of the detector volumes. The size of the scoring volumes**
 239 **used varied based on the distance from the field edge (x) and were $1.1 \times 3.8 \times 3.8 \text{ mm}^3$ ($x \leq 5\text{cm}$),**
 240 **$1.1 \times 7.6 \times 7.6 \text{ mm}^3$ ($5\text{cm} < x \leq 20\text{cm}$) and $1.1 \times 15 \times 15 \text{ mm}^3$ ($x > 20\text{cm}$). The detectors at the**
 241 **phantom entrance and along the central axis were rotated by 90° so the majority of particles**
 242 **passed through the largest face of the scoring volumes. The colored area shows the section of the**
 243 **phantom irradiated by the primary field and a depth-dose curve demonstrates the modulation**
 244 **width of the SOBP.**

245 The absorbed dose was obtained in each of the scoring volumes and separated based on particle type.
 246 This allowed application of particle specific weighting factors for low dose radiations. The absorbed
 247 dose due to protons (D_p), neutrons (D_n) and photons (D_γ) was tallied separately. Neutron energy
 248 spectra were also collected in 1MeV bins during the Monte Carlo simulations which allowed
 249 calculation of the average radiation weighting factor, w_R , based on the ICRP definition (ICRP 2003).
 250 The average neutron weighting factors were then used to convert the absorbed dose (D) to equivalent
 251 dose (H) using equation 6. Photons are given a factor of 1, whilst protons are assigned a factor of 2 to
 252 account for secondary particles which deposit dose locally such as δ -electrons and charged nuclear
 253 fragments (ICRP, 2003).

254

$$H = 2D_p + w_R D_n + D_\gamma \quad (6)$$

255

256 **3. Results**

257 *3.1. Monte Carlo calibration and validation*

258 *3.1.1. Depth dose characteristics.* Depth dose curves were measured using a plane-parallel chamber
259 with an entrance window of diameter 84mm (PTW Freiburg GmbH) in a water tank for proton pencil
260 beams of range, 8.92g/cm², 12.64g/cm², 15.92g/cm², 21.1g/cm² and 25.15g/cm² (see figure 4). The
261 dose was scored in the Monte Carlo using cylindrical voxels of diameter 84mm and thickness 0.2mm
262 to give high depth resolution and to match the lateral dimensions of the plane-parallel chamber. The
263 results shown in figure 4 are the mean of 10 independent simulations which each cycled through their
264 respective phase space files 5 times (total ~34 million particles transported into the phantom in each
265 simulation). The uncertainty in the depth-dose curves was defined as the standard deviation of the 10
266 simulations and is less than 1% at all depths for all simulations. Emphasis was placed on taking
267 measurements close to the Bragg peak and on the distal edge for range verification and calibration of
268 the initial energy spread at nozzle entrance. The initial energy spread (ΔE) in terms of proton energy
269 (E) was determined by comparing experimental and simulated pristine Bragg curves and is given in
270 equation 7.

271
$$\frac{\Delta E}{E}(\%) = 4.7 \times 10^{-5} E^2 - 0.021E + 2.60 \quad (7)$$

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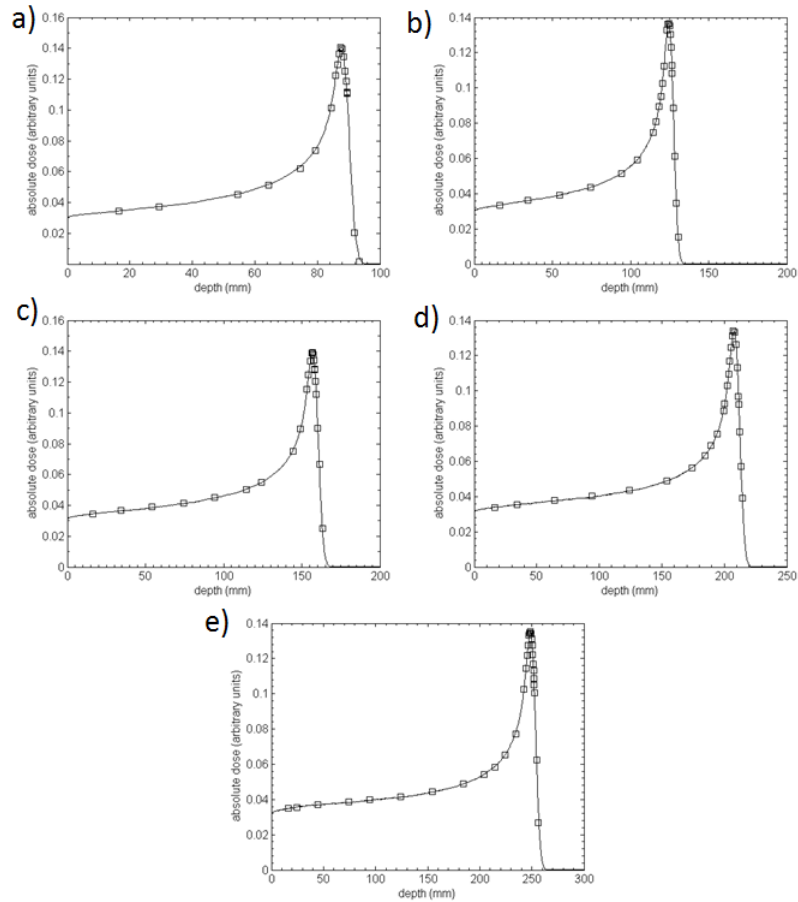
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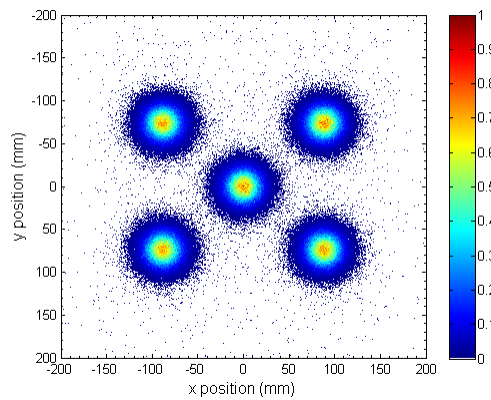
280 **Figure 4: Depth dose curves for (a) 8.92g/cm^2 , (b) 12.64g/cm^2 , (c) 15.92g/cm^2 , (d) 21.1g/cm^2 and**
 281 **(e) 25.15g/cm^2 . The solid line is the Monte Carlo data and the squares represent experimental**
 282 **data points. The uncertainty in the Monte Carlo is $<1\%$ at all points.**

283

284 The results of the depth dose curves demonstrate good agreement between the measurement and
 285 simulation data.

286 *3.1.2. Scanning magnet field strength.* The magnetic field calibration was performed for beams of
 287 range 8.92g/cm^2 , 12.64g/cm^2 , 15.92g/cm^2 , 21.1g/cm^2 and 25.15g/cm^2 . These ranges correspond to
 288 nominal energies of 95.69MeV , 112.53MeV , 138.08MeV , 156.67MeV and 174.74MeV at nozzle
 289 entrance for the MGH PBS system, respectively. Five distinct spots were irradiated for each of the
 290 energies at different lateral positions (Figure 5). The results shown in Figure 5 are averages of 10

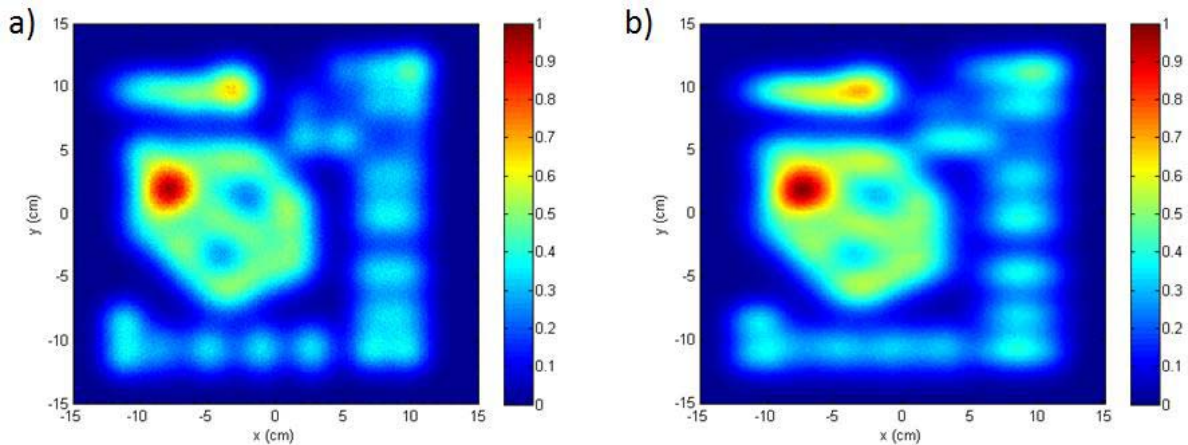
291 independent simulations, each containing ~34 million primary protons. The results shown in Figure 5
292 are a superposition of the 5 spots for the 5 different energies considered (i.e. total of 25 spots). The
293 lateral spot positions in the Monte Carlo matched those measured using the BIS within 1mm for each
294 considered energy, demonstrating that the conversion of lateral position to magnetic field strength
295 varies correctly with proton energy (and hence range).



296

297 **Figure 5: Calibration of magnetic fields to control lateral position of the proton beam in the**
298 **Monte Carlo simulations. The units on the color scale are relative to the maximum observed**
299 **value. The data in the figure contains the 5 considered ranges (8.92g/cm², 12.64g/cm²,**
300 **15.92g/cm², 21.1g/cm² and 25.15g/cm²).**

301 *3.1.3. Complex two-dimensional irradiation.* Figure 6 shows the results of the Monte Carlo simulation
302 (a) and experimental delivery (b) of the test pattern.



303

304 **Figure 6: Simulated (a) and measured (b) complex 2 dimensional irradiation containing areas of**
 305 **variable dose, continuous scanning and spot scanning.**

306 The Monte Carlo simulation plot is an average of 10 independent simulations, each containing ~34
 307 million primary protons. The uncertainty in the Monte Carlo data was less than 1% at all points
 308 considered. The dose was normalized to the maximum dose observed in both the simulated and
 309 experimental data. Relative doses were used rather than absolute as the BIS is not capable of
 310 measuring absolute dose. The simulated and measured results were compared via gamma analysis
 311 (Low *et al.*, 1998), with 100% of the points passing the 2mm/2% criteria.

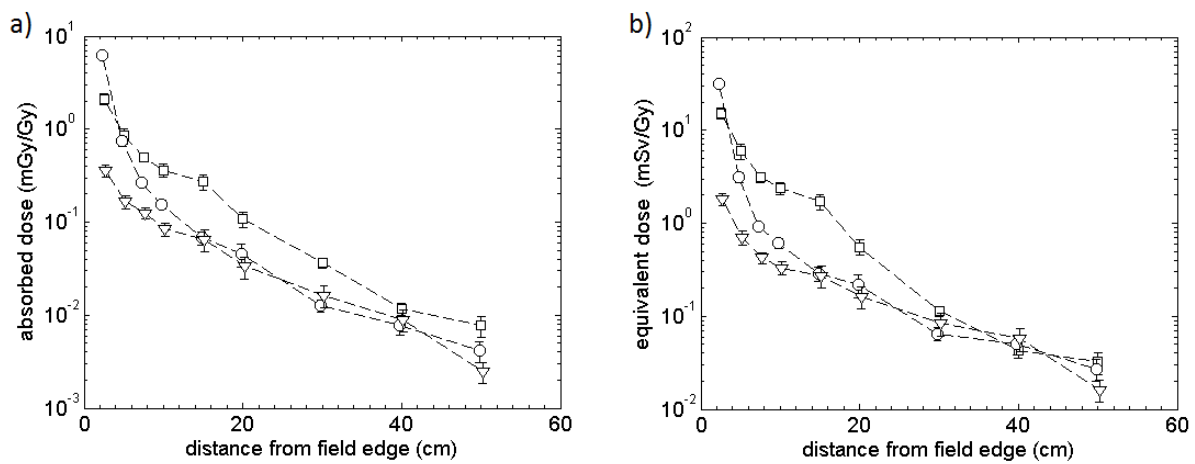
312

313 3.2. The impact of apertures in beam scanning on the example of a prostate treatment field

314 3.2.1. Doses at depths proximal to the SOBPs. At a depth of 4.72cm in Lucite, the absorbed dose at
 315 2.5cm out-of-field was found to be approximately 3 times higher for PBS than DS. The reason lies in
 316 the relatively large spot size used in PBS at the MGH (~12mm at isocenter). At a depth of 9.44cm, DS
 317 shows close to 5 times higher doses than PBS. The latter is caused by secondary doses created in the
 318 aperture when using DS.

319 Including an aperture at nozzle exit reduces the penumbral width by preventing wide-angle scatter
 320 from reaching the phantom (or patient). At larger lateral distances from the field edge, the difference

321 in the doses between the considered delivery techniques increases. The absorbed dose from PBS with
 322 an aperture is an order of magnitude lower than for the other techniques at 2.5cm from the field edge
 323 at a depth of 4.72cm in water (see figure 7). For lateral distances less than 10cm from the field edge in
 324 PBS, primary protons dominate the total equivalent dose. At greater lateral distances, the contribution
 325 of scattered primary protons is not significant and the absorbed dose from PBS becomes less than DS.
 326 As the absorbed dose at lateral distances from the field edge greater than 20cm is dominated by
 327 internally produced secondary particles, the benefit of using an aperture is somewhat diminished in
 328 such regions.

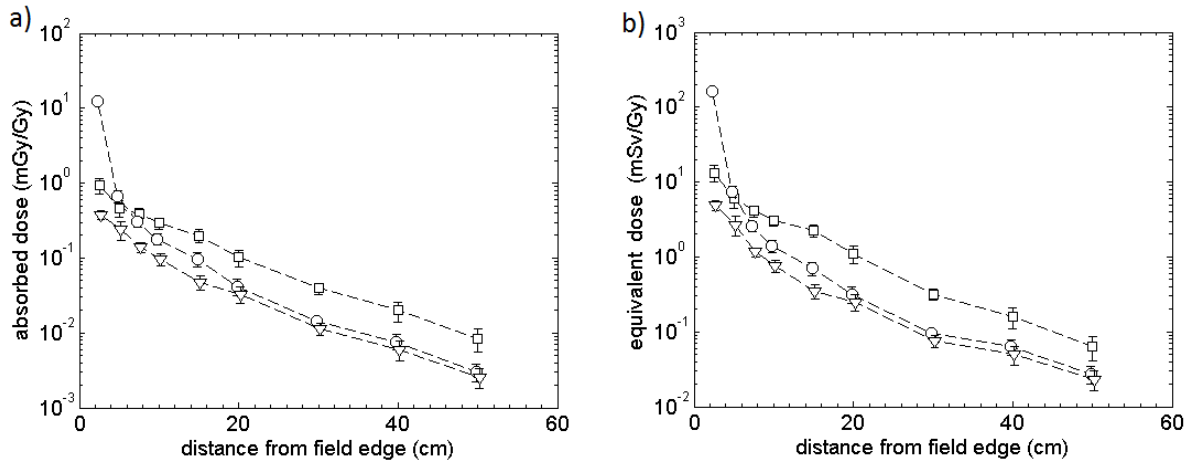


329

330 **Figure 7: Simulated absorbed dose (a) and equivalent dose (b) at different lateral distances from**
 331 **the field edge at entrance depth of 4.72cm. The doses are relative to the absorbed dose delivered**
 332 **in the SOBP. The data shown is for the double scattering (squares), pencil beam scanning**
 333 **(circles) and pencil beam scanning with an aperture (triangles). The error bars represent two**
 334 **standard deviations.**

335

336 *3.2.2. Doses at depths corresponding to the SOBP.* Figure 8 shows the absorbed dose and equivalent
 337 dose at a depth of 18.88cm. The absorbed dose close to the field edge is again clearly higher in PBS
 338 than the other considered delivery techniques. The absorbed dose from DS is higher than PBS at all
 339 lateral distances greater than 10cm from the field edge.



340

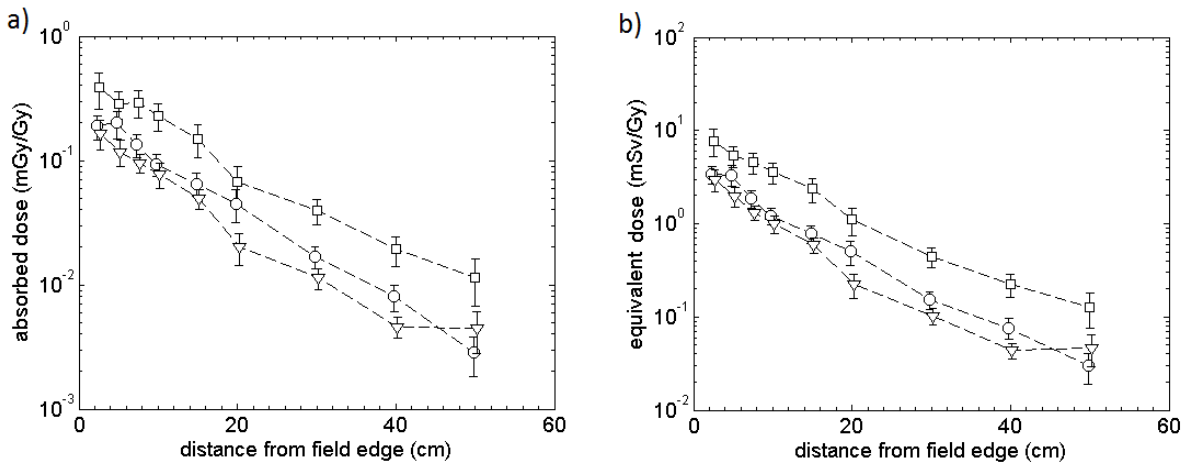
341 **Figure 8: Simulated absorbed dose (a) and equivalent dose (b) for the double scattering**
 342 **(squares), pencil beam scanning (circles) and pencil beam scanning with an aperture (triangles)**
 343 **at a depth of 18.88cm. The doses are relative to the dose delivered in the SOBP. The error bars**
 344 **represent two standard deviations.**

345 Using an aperture significantly reduces the absorbed dose from PBS for lateral distances up to 20cm
 346 from the field edge at depths within in the SOBP. At 2.5cm from the field edge, the absorbed dose is
 347 reduced by more than an order of magnitude when the aperture is used. For larger distances, the
 348 benefit of an aperture is not as pronounced due to the higher contribution of internally produced
 349 secondary and scattered particles to the total dose.

350 The equivalent dose curves agree within the uncertainty limits for the PBS and PBS with an aperture
 351 data for lateral distances greater than 20cm. The higher equivalent dose values observed for the DS
 352 data highlights the significant contribution of neutrons generated in the treatment head (in particular
 353 the aperture) for this delivery technique. The equivalent dose from DS is significantly higher than all
 354 other techniques for all lateral distances greater than 5cm from the field edge. Note that when using an
 355 aperture in PBS only a small portion of the field is restricted by the aperture while in DS typically the
 356 majority of the incident proton therapy field will be blocked by the aperture.

357 *3.2.3. Doses at depths distal to the SOBP.* The contribution of primary protons to the total absorbed
 358 dose and equivalent dose distal to the SOBP is zero due to their finite range. The absorbed dose and

359 equivalent dose values obtained distal to the SOBP (see figure 9) is entirely due to secondary
 360 particles. The higher doses observed distal to the SOBP in DS are due to the higher number of
 361 secondary particles generated in the treatment head compared to the other delivery techniques. As the
 362 depth in the phantom increases, the relative contribution from internally generated secondary particles
 363 to the total dose increases. The increased neutron fluence from DS contributes to a higher neutron
 364 absorbed dose and equivalent dose compared to the other considered techniques. The increased
 365 number of secondary particles, especially neutrons, incident upon the phantom in DS leads to an
 366 increased production of secondary particles, the effect of which can still be observed distal to the
 367 SOBP.



368
 369 **Figure 9: Simulated absorbed dose (a) and equivalent dose (b) for the double scattering**
 370 **(squares), pencil beam scanning (circles) and scanning with an aperture (triangles) at a depth of**
 371 **28.32cm. The dose values are relative to the dose delivered in the SOBP. The error bars**
 372 **represent two standard deviations.**

373

374 4. Discussion and Conclusion

375 Placing a patient-specific aperture at nozzle exit reduces the out-of-field doses from PBS. Proton
 376 interactions occurring in the brass of the aperture lead to a higher neutron fluence compared to PBS
 377 with no aperture. The neutron component of the total dose out-of-field increases when using an

378 aperture in PBS, however the total doses are still reduced. The reduction in the dose from primary
379 protons, due to the large angle scatter not reaching the phantom leads to a reduction in the total
380 absorbed dose and equivalent dose when an aperture is placed at nozzle exit. The number of proton
381 interactions in the aperture is much lower in PBS than in DS. In DS, a large proportion of the primary
382 beam interacts with the aperture, leading to a high neutron and secondary particle fluence. In PBS, a
383 comparatively small amount of interactions occur in the aperture, as the majority of the primary
384 protons pass through the aperture without undergoing an interaction. The aperture only interacts with
385 particles which have been scattered through large angles.

386 The benefit of incorporating an aperture in PBS is diminished distal to the SOBP. Some benefit can
387 still be observed distal to the SOBP, demonstrated by the reduced absorbed doses and equivalent
388 doses in this region compared to PBS with no aperture. The reduction in the penumbral width reduces
389 the proton fluence and dose lateral to the primary field in PBS. This reduction in proton fluence leads
390 to a decrease in secondary particle production, the effect of which can be seen in the lower out-of-
391 field doses throughout the phantom compared to PBS with no aperture.

392 One of the reasons for moving to PBS in preference to DS is the removal of the dependency upon
393 patient-specific hardware, which adds to the cost of operating a clinical proton facility. However, one
394 of the issues with scanning is the wider penumbra when the pencil beam width is large and/or the
395 proton energy is small, which is demonstrated in the simulation results presented here. The use of a
396 patient-specific aperture may be required to decrease the penumbral width for certain clinical cases,
397 but the other advantages of PBS compared to DS will still be largely maintained. The results of the
398 simulations in this work show that using an aperture can potentially reduce the absorbed dose and
399 equivalent dose lateral to the primary field in PBS by an order of magnitude.

400

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