Stereotactical fields applied in proton spot scanning mode with range shifter and collimating aperture

C Bäumer, C Fuentes, M Janson, A Matic, B Timmermann and J Wulff

1 West German Proton Therapy Centre Essen, Hufelandstr. 55, Essen, Germany
2 University Hospital Essen, Hufelandstr. 55, Essen, Germany
3 West German Cancer Center (WTZ), Hufelandstr. 55, Essen, Germany
4 German Cancer Consortium (DKTK), Heidelberg, Germany
5 IBA PT, Chemin du Cyclotron 3, Lovain-La-Neuve, Belgium
6 RaySearch Laboratories, Sveavägen 44, Stockholm, Sweden
7 Department of Particle Therapy, University Hospital Essen, Hufelandstr. 55, Essen, Germany

E-mail: christian.baumer@uk-essen.de

Keywords: proton therapy, spot scanning, lateral dose fall-off, SRT, apertures

Abstract

Some clinical indications require small fields with sharp lateral dose gradients, which is technically challenging in proton beam therapy. This holds especially true for low-range fields applied with the spot scanning technique, where large beam profiles entering from the beam-line or the insertion of range shifting blocks lead to large lateral gradients. We regard the latter case and solve it by shifting the range shifting block far upstream in conjunction with a collimating aperture close to the patient. The experiments of the current work are based on a commercial proton therapy treatment head designed for several delivery modes. In a research environment of the spot-scanning delivery mode a range shifter is inserted downstream of the scanning magnets in a slot which is usually employed only in a scattering delivery mode. This configuration is motivated by equations assuming a simple model of proton transport. In the experiments lateral dose planes are acquired with a scintillation screen and radiochromic films. Dose distributions are calculated with the Monte Carlo dose engine of the RayStation treatment planning system. We demonstrate that proton fields with 80%–20% lateral dose fall-off values between 1.4 mm and 4.0 mm can be achieved for water equivalent depths between 0 cm and 10 cm. The simulated lateral dose profiles agree with the experimental dose profiles. The sharpening of the field edges is set off by a broadening of the proton spots towards the center of the fields. This limits the clinical application mainly to small fields for which the distal and proximal conformity is of minor importance.

1. Introduction

Stereotactical radiation therapy features typically small fields with high conformity and a large dose per fraction. The former requirement implies a rapid lateral dose fall-off if also the sparing of normal tissue is considered. Aiming for a sharp lateral dose fall-off, the creation of treatment fields with proton beams is demanding. The multiple Coulomb scattering of protons in a patient or in a technical phantom poses a lower physical limit of the lateral dose fall-off (Gottschalk et al 1993). In addition, there is a contribution of the treatment head, which is often called ‘nozzle’ in particle therapy, to the lateral dose fall-off. Proton nozzles, which adhere to the concept of passive beam delivery, use scattering foils for the lateral beam broadening and a downstream collimating aperture to conform the proton field laterally to the target volume. Using this technique, lateral dose fall-off values can be achieved which comply with the needs of stereotactical radiation therapy (see Wang et al (2014), Vernimmen et al (2016) and references therein).

Currently, however, a clear trend is observed towards proton field application with the pencil-beam scanning (PBS) delivery technique, especially for new installations of proton therapy centers. PBS employs a proton beam which is magnetically steered across the clinical target. The lateral profile of a PBS beamlet, called ‘spot’, can be
described by a Gaussian profile. In the frame of the current work the nozzle is operated in spot scanning mode (SSM), where the PBS spots are applied one by one with a beam off-time in between. The fluence modulation of spots in conjunction with energy switching at the level of the accelerator enables 3D-conformal dose distributions in patients. The lateral dose fall-off of PBS fields, which is determined by the underlying spot profiles, is inferior to the one delivered with passive delivery techniques and apertures, as, e.g. pointed out in Safai et al (2008).

Wang et al showed that the in-air-sigma of the Gaussian profile has to be below 4.3 mm in order to justify stereotactical treatments with protons as an alternative to stereotactical radiation therapy with photons (Wang et al 2014). This limit corresponds to a limit of about 6.4 mm for the 80%–20% lateral dose fall-off (Bäumer and Farr 2011). Many PBS delivery systems do not achieve this specification. This is particularly the case for target volumes at shallow depths. These require either low proton energies, which feature larger in-air-sigma values than energies for deep seated targets, or range shifting plastic blocks (RS) near the patient surface in order to reduce the proton energy below the minimum accelerator energy, which usually ranges between 60 MeV and 100 MeV (Moignier et al 2016). The latter option also increases the spot size.

A sharper lateral dose fall-off of beam delivery systems, which are affected by a large emittance at nozzle entrance or large scattering contributions within the nozzle, can be realized by the combination of PBS and collimators (Engelsman et al 2013, Moteabbed et al 2016, Charlwood et al 2016, Yasui et al 2017, Ciocca et al 2019). In a previous work we demonstrated lateral dose fall-off values of 5.4 mm–9.0 mm for a commercial proton nozzle using collimating brass apertures right upstream of a RS (Bäumer et al 2018). The lateral gradients could be improved by about 20% if the aperture was mounted between RS and patient. This technique would in principle allow stereotactical treatments with protons for selected cases according to the reasoning of Wang et al (2014). Clearly, a sharper lateral dose fall-off is desirable to lower the normal-tissue complication probability. Pursuing a similar nozzle concept as Ciocca et al (2019), we demonstrate in the current work that lateral dose fall-off values well below the threshold of 6.4 mm are feasible with a proton treatment head operated in PBS delivery mode including RS. We show that this is possible with a single hardware change with respect to the commercial, certified nozzle: the upstream shift of the RS towards the scanning magnets. The necessary modification of the dosimetry control system is discussed. We also provide an analytical description motivating the novel nozzle concept (section 2). Measurements of lateral dose profiles are presented and compared with dose distributions simulated with a Monte Carlo dose engine in the treatment planning system RayStation (section 3.2).

2. Theory

The total spread $\sigma_{tot}$ of a PBS spot can be approximated by Kanematsu et al (2008)

$$\sigma_{tot}^2 = \sigma_{th}^2 + \sigma_{pt}^2$$

where the first term $\sigma_{th}$ on the right hand side of equation (1) describes the contribution from the treatment head to the width of a spot. Its square root is denoted as ‘in-air-sigma’ in the section above. The 80%–20% lateral dose fall-off, which is also referred to as lateral penumbra (LP) (ICRU 2007), is proportional to $\sigma_{tot}$ (Bäumer and Farr 2011). The second term $\sigma_{pt}$ in equation (1) concerns the scattering within a patient. Its influence can be minimized by choosing optimized beam portals in treatment planning. In the current work we deal with the optimization of the first term.

In the framework of Sabbas et al (1987) and Bäumer et al (2018) the in-air-sigma for a nozzle configuration at the edge of a field, in which the aperture is located downstream of the RS, can be approximated by:

$$\sigma_{th,edge}^{ap-ds} (z) = A_0^{RS} \left( \frac{z_{RS}}{z_{edge}} \right)^2 (z - z_{edge})^2$$

where $A_0^{RS}$ describes the angular variance after traversing the RS. The longitudinal distances $z_{edge}$, $z_{RS}$, and $z$ are measured from the virtual source position (see figure 1). Of course, equation (2) is a simplification, because there are contributions from the scattering in air and in the monitor chambers. As pointed out in Sabbas et al (1987) and Bäumer et al (2018) the virtual point source can be regarded as an isotropic point source which coincides with the source of a scanning beam. In the following theoretical discussion of this section we only deal with PBS configurations with an aperture downstream of all other beam shaping devices (figure 1 middle left/right). Here, $z_{edge}$ is the $z$ position of the downstream edge of the aperture. For the $ap-ds$ configuration, in which the aperture is mounted right behind the RS in downstream direction (see Bäumer et al (2018) and figure 1 middle left), the longitudinal position of the RS $z_{RS}$ is given by

$$z_{RS} = z_{edge} - B - \Delta z_{exp}$$

where $B$ is the physical thickness of the aperture block. $\Delta z_{exp}$ is the distance between the downstream edge of the RS ($= z_{edge} - B$ for $ap-ds$) and the $z$ coordinate of the effective scattering point, which is approximately in
the middle of a thick target if the residual range is well above the water-equivalent thickness (WET) of the RS (Gottschalk et al 1993, Hong et al 1996, Bäumer et al 2018). Regarding $z_{RS}$ in the ap-ds-stereo configuration, we refer to figure 1 (middle-right). The second factor in equation (2) ($z - z_{edge}$) is given by the sum of the air gap $G$ and the depth of the target volume in the patient. Assuming that $G$ has already a low value which is realizable in clinical practice (say 3 cm to 5 cm), the second factor in equation (2) cannot be minimized further.

In the current work we reduce the first factor $z_{RS}/z_{edge}$ of equation (2) to reduce the lateral dose fall-off (ap-ds-stereo in figure 1 middle-right). Using the mechanics of the IBA universal nozzle (see section 3.1) the RS can be mounted close to the scanning magnets (section 3.1) yielding a factor $z_{RS}/z_{edge}$ of 0.17. By comparison, this factor is 0.97 if the RS is mounted right upstream of the aperture as outlined in Bäumer et al (2018).

Above considerations concern the spot profile at the edge of the field. In the center of the field the angular variance transforms into a spatial variance as the protons propagate through the aperture opening into the patient or phantom:

$$\sigma_{Tot,center} \approx \sqrt{A_{RS}(z - z_{RS})}. \quad (4)$$

Considering a distance $z - z_{RS}$ on the order of 1.7 m the spots will be quite big at the detection plane (figure 1). Consequently, the effect of the fluence modulation of spots is limited. This could be tolerated, if the cross section of the beam port is small.

3. Materials and methods

3.1. Delivery of proton fields

In the WPE facility, which is equipped with the ProteusPlus proton therapy system (IBA PT, Louvain-la-Neuve/Belgium), protons are accelerated by an isochronous cyclotron. In the subsequent energy-selection system (ESS) the energy can be degraded down to 100 MeV. In the experiments the proton beam was guided to the fixed horizontal treatment head, which is located in one of the four treatment rooms. The treatment head, which is called ‘universal nozzle’ and sketched in figure 1, can deliver fields in single scattering, double scattering, SSM, or uniform scanning mode. It features a pair of quadrupole magnets (not shown in figure 1) and a pair of dipole magnets which steer the pencil beam to the desired lateral position. In SSM discrete spots are applied with in-air-sigma values in the isocenter plane ranging from 3.3 mm (227 MeV) to 7.9 mm (100 MeV) (Bäumer et al 2018).

The nozzle comprises segmented monitor chambers at the entrance (‘IC1’) and close to the exit (‘IC2/3’) (Courtois et al 2014) (at a distance of 116.5 cm from isocenter). The spatial profiles of the ionization within IC1 are used to control the centricity of the beam downstream of the last bending magnet. As this feature is identical
in the clinical operation and in the current research project and IC1 has no effect on the positioning and dosimetry of PBS spots, we omit IC1 in figure 1 and in the following discussions. The IC2/3 contains two layers with strip electrodes (width =6.3 mm, pitch =6.5 mm) which provide information about the position and the size of individual spots (Courtois et al 2014). In SSM the spots are applied energy layer by energy layer. The beam energy is changed in the ESS accompanied by a change of the current setpoints of the magnets of the beam line. In the control software of the dosimetry system the centering of the individual energy layers is checked and adjusted with the application of a so-called tuning spot.

The universal nozzle has also a range modulator wheel (‘RMW’ in figure 1) to create spread-out Bragg peaks (SOBPs) in the scattering modes and in the uniform scanning mode. Scattering foils (fixed scatterer ‘FS’ in figure 1) and a contoured scatterer can be inserted into the beam path for the scattering modes. A snout is mounted at the exit of the universal nozzle providing holders for collimating apertures and range compensators or RS. The snout can be translated linearly to minimize the air gap between these beam-shaping devices and the patient surface. The collimating brass apertures have been designed to conform the proton fields laterally to the target volume in the scattering modes and uniform scanning delivery modes. The snout holder contains either two aperture blocks in subsequent slots with a total physical thickness of \( B = 6.5 \text{ cm} \) or a single aperture in the downstream slot with \( B = 3.3 \text{ cm} \) thickness. The apertures and a RS, which is a homogeneous plastic slab, can be mounted on the snout as beam-shaping devices in SSM.

The universal nozzle used in the current research project was not commissioned for single and double scattering mode. Thus, a RS with a physical thickness of \( D = 5.43 \text{ cm} \) could be mounted in an open slot of the contoured scatterer located between the scanning magnets and the IC2/3 (figure 1). The WET of the RS was measured with the Zebra multi-layer ionization chamber yielding \( 6.27 \text{ g cm}^{-2} \). In order to apply the proton fields, the treatment control system was operated in a service mode, in which the interlocks derived from IC2/3 were disabled. In all other regards the field delivery was performed in the same way as clinical irradiations. The disabling of the IC2/3 based interlocks was necessary, because amongst other things the spot widths were larger than in the nominal operation of the nozzle. The larger spots might affect the signal quality of the IC2/3 readout channels and the spot positioning accuracy. These possible consequences were evaluated retrospectively using the data acquired during the cyclic checks of the monitor chambers.

### 3.2. Field design and dose calculation

A beam model was established within a research installation of the commercial treatment planning system RayStation (version 7). The beam model was derived from our clinical model (Bäumer et al 2017, Bäumer et al 2018) applying a single modification: A new RS with a WET of \( 6.27 \text{ g cm}^{-2} \) and with a distance of its center of 171 cm to the isocenter plane was added to the Snout180, which holds one or two 3.3 cm thick apertures with an inner diameter of 16 cm (see section 3.1). Apertures were produced with a milling head of 1 cm diameter. The virtual source axis distance for the lateral directions were 230.3 cm and 193.5 cm reflecting the longitudinal separation of the scanning dipole magnets of about 35 cm. For analytical considerations (see section 2) the average of both virtual source axis distances of both (212 cm) can be used to approximate a singular virtual source located 212 cm of the isocenter. Distributions of absorbed dose were simulated with the Monte Carlo dose engine of RayStation (Saini et al 2017, Bäumer et al 2018, Maes et al 2019) using a virtual water phantom with a 1 mm dose grid.

In the frame of the validation of the beam model the following quasi-monoenergetic fields were applied for energies 100 MeV, 130 MeV, 150 MeV, 180 MeV, 200 MeV:

- An almost square field with an aperture opening of \( 3 \text{ cm} \times 3 \text{ cm} \) and aperture thickness of \( 6.5 \text{ cm} \). Due to the longitudinal separation of the scanning magnets described above the projected shape of the aperture opening on the detection plane is not exactly square. The target design started with a rectangular contour. Due to the 1 cm diameter of the milling head the corners of the apertures are rounded off. The applied SSM fields extend laterally from \( -17.5 \text{ mm} \) to \(+17.5 \text{ mm} \) with a spot spacing of \( 2.5 \text{ mm} \).
- Single, central axis spots with an open aperture (16 cm diameter).

Furthermore, a SOBP has been designed representing a possible clinical application. The field was optimized to cover a cylindrical target volume with a range of \( 2.5 \text{ cm} \), a modulation width of \( 1 \text{ cm} \) and an absorbed dose of \( 1 \text{ Gy} \). Spot weight optimization was performed after a circular aperture (diameter \( 1 \text{ cm} \), \( B = 3.3 \text{ cm} \) thickness, 1.3 mm aperture margin (Wang et al 2015)) had been created in the proton beam design module. The lateral hexagonal spot grid had a spacing of 2 mm. The field comprised eight energy layers. This field serves as an example of the irradiation of a very shallow target, e.g. an eye tumor.
3.3. Measurements

All measurements of the current work refer to the configuration \textit{ap-ds-stereo} (figure 1 middle right). The air gaps ranged between 5 cm and 15 cm. Two-dimensional profiles perpendicular to the central beam axis were acquired with a Lynx2D detector (IBA, Schwarzenbruck/Germany) (Russo et al 2017) and with EBT3 GAFchromic film (Ashland, Bridgewater/USA) (Devic 2011) scanned with an Epson flatbed scanner 10 000 XL with 72 dpi resolution. The pixel values of the red color channel were converted into doses, which proved to be a practical and accurate procedure up to dose readings of about 8 Gy (Devic 2011). This procedure was successfully used in conjunction with PBS-with-apertures (Maes et al 2019). In a previous work we have shown that the Lynx2D detector combined with a polystryrene plate phantom (SP34, (IBA, Schwarzenbruck/Germany) provides a good approximation of the relative dose at shallow depths (Bäumer et al 2018). A similar experimental approach was pursued in van Luijk et al (2001). The data processing has been carried out in OmniPro-IMRT (IBA, Schwarzenbruck/Germany) as described in Bäumer et al (2018). The dose of the quasi-mono energetic layers and the single spots was about 0.1 Gy for the Lynx2D measurements and 0.4 Gy for the film measurements.

4. Results

4.1. Delivery of PBS spots

Figure 2 shows the example of the tuning in the frame of the delivery of the highest energy layer of the SOPB (R2.5M1, see section 3.2). The lateral full-width-at-half-maximum is 46 mm. Although the spot profile extends over a major part of the IC2/3, the spot is fully covered providing a sufficient number of sample points for the Gaussian fit. The example shows that the monitoring and dosimetry system of the nozzle can detect deviations of the lateral spot position from the nominal one and correct for it. The retrospective analysis of all applied spots showed that the maximum deviation on IC2/3 was 0.2 mm.

4.2. Lateral profiles of broad fields

Figure 3 visualizes exemplary one-dimensional dose profiles. The sharp lateral dose fall-off is clearly visible. There is a low dose tail, which gets stronger with increasing with proton energy. It is almost negligible for 100 MeV. At 200 MeV it is on a relative dose level of percent extending over a lateral distance $>1$ cm. The high dose region is not uniform. At 100 MeV there is a dose enhancement (‘horns’ (van Luijk et al 2001)) towards the edges of the field. The edge enhancement reverses with increasing energy turning into a rounding off of the corners.

Table 1 provides an overview of the extracted values of the lateral dose fall-off. At 5 mm depth the lateral dose fall-off is on the order of 2 mm. There is no distinct dependence on the proton beam energy. The difference between the lateral dose fall-off of measured and simulated lateral profiles is typically on the order of 0.2 mm (average deviation 0.1 mm/2%). A marked difference is observed for the film measurements at 0.1 mm depth. Here, the Monte Carlo simulation overestimates the lateral dose fall-off by about 0.5 mm. Table 1 also shows that the Monte Carlo simulations predict LP between 3 mm and 5 mm at depths between 7.5 cm and 12.5 cm.

Figure 4 visualizes the lateral dose-fall off as a function of the air gap. The air gaps of the clinical fields in our proton therapy center are typically between 3 cm and 7 cm. The measured and simulated data points at 5 cm and
7 cm indicate that there is only a slight increase of the lateral dose fall-off with air gap in the range of clinical air gaps.

4.3. Lateral profiles of individual central spots

Figure 5 shows exemplary one-dimensional dose profiles of central-axis PBS spots acquired in the transverse plane 5 cm upstream of isocenter. The figure shows also the corresponding simulated dose profiles. The spot widths are on the same scale as the field widths considered in the current work. The profile of the 100 MeV spot does not fit into the opening of the 16 cm diameter aperture ring. Qualitatively, the simulated profiles match the measured ones. For the 130 MeV field the deviation between the two profiles deviates with increasing distance from the field center.
In table 2 the full-width-at-half-maximum values of the spots are listed. The spot widths are about a factor of seven larger than the widths of the spots without RS in the beam path (last column of table 2).

4.4. The impact of the aperture thickness on the lateral dose fall-off
Figure 6 shows a simulation study about the effect of the aperture thickness on the lateral dose profile. At 100 MeV the reduction from two aperture blocks (6.5 cm thickness) to a single aperture block (3.3 cm thickness) leads to a reduction of the edge enhancement. In the example of the 180 MeV field the rounding-off of the corners is slightly more pronounced when using a single aperture block instead of two.

5. Discussion
The successful delivery of the radiation fields and the evaluation of the irradiation log files demonstrate that a spot scanning mode is operational with a scatterer upstream of the IC2/3 (see section 4.1). The criterion for the feasibility of such a SSM is that the dosimetry including the control of the number of applied monitor units (MUs), the check and the tuning of spot position and the check of the spot size can be performed for individual spots. The field delivery was facilitated by the use of an existing slot in the commercial treatment head, which is clinically used only in the double scattering mode, in a research environment. The requirement of the acquisition of a substantial part of a spot on IC2/3 within the dosimetry control software would restrict the maximum field.
size. Because the IBA universal nozzle is designed for field sizes up to 30 cm × 40 cm, this is, however, not a concern for small stereotactical fields.

Regarding the scanning of large spots over a large lateral area, the nozzle design of the current work resembles the nozzle design of the uniform scanning mode (figure 1 right). The similarity of the nozzle design correlates with the sharp lateral penumbrae, which are a feature of the US mode, too (Nichiporov et al 2012, Rana et al 2013, Bäumer et al 2017). However, in uniform scanning mode the pencil beam is broadened upstream of the scanning magnets and the scanning path follows a regular pattern with continuous beam-on phase during each individual energy layer. Furthermore, the energy layer switching is carried out within the nozzle. The advantage of the uniform scanning mode over the special SSM of the current work is the conformal shaping of the dose distribution at the distal side with a range compensator. Further studies will have to evaluate how this would affect the overall conformity in a multi-field treatment plan. For stereotactical applications with small tumor volumes we expect a small impact.

In order to develop a clinical version with a RS in second-scatterer position two functions of the nozzle would have to be modified. First, the increased spot sizes would have to be considered in the control software of the cyclic nozzle checks. This also includes finding the fastest and most robust numerical procedure to determine the position and the width of spots. Secondly, the MUs would have to be recalibrated, because the RS will alter the beam quality incident on the IC2/3. Especially, low-energetic secondary protons generated in the RS will lead to an over-proportional large signal in the monitor chambers due to their high stopping power. A similar argument holds for the longer path within the monitor chambers of scattered particles. While a clinical commissioning is still pending, the modified treatment head might already be used for small-animal experiments (van Luijk et al 2001).

The analytical considerations outlined in section 2 correctly predict a sharp lateral dose fall-off (section 4.2) if the distance between a RS and a downstream collimating aperture is large. At the same time the spots in the

### Table 2.

| Energy (MeV) | Measurement FWHM (mm) | Simulation FWHM (mm) | Mean σ (mm) | FWHM Meas. (mm) | Simulation FWHM (mm) | Mean σ (mm) | σ diff. (%) | σ diff. (mm) | w/o RS (mm) |
|-------------|------------------------|----------------------|-------------|-----------------|----------------------|-------------|-------------|-------------|-------------|
| 100         | 147.5                  | 148.2                | 147.9       | 160.8           | 161.6                | 161.2       | 7.9         |             |             |
| 130         | 94.4                   | 94.8                 | 94.6        | 100.8           | 100.3                | 100.6       | 3.5         | 8.8         | 6.4         |
| 150         | 78.7                   | 78.9                 | 78.8        | 80.9            | 81.3                 | 81.1        | 1.5         | 4.4         | 5.4         |
| 180         | 64.1                   | 63.9                 | 64.0        | 65.7            | 66.3                 | 66.0        | 2.7         | 5.4         | 4.3         |
| 200         | 57.0                   | 57.1                 | 57.1        | 59.2            | 59.7                 | 59.5        | 2.6         | 5.8         | 3.8         |

**Figure 6.** Simulated lateral profiles with single aperture (3.3 cm thickness) and double aperture (6.5 cm thickness) for 100 MeV and 180 MeV fields using the ap-ds-sito configuration with the range shifter 171 cm upstream of the isocenter. The air gap was 5 cm and the water equivalent depth of the measurement plane 5 mm. The relative dose profiles are scaled to have a maximum of 105%/95% (100 MeV/180 MeV).
middle of the field get large (section 4.3) as predicted by the theory. This result might be regarded as counterintuitive, because in alternative nozzle concepts the RS is mounted close to the patient. In the current work RS and aperture act as a ‘filter’: the sharp cut employs the strong correlation between angle and position of the protons at the aperture edge. Quantitatively, however, there are limitations of the analytical equations of section 2. According to equation (2) the lateral penumbra should be reduced by a factor of about five compared to the aperture downstream configuration (figure 1 middle left). The reduction obtained from the simulations and from the measurements at 150 MeV is, however, a factor of less than three (see table 3 in Bäuer et al. 2018).

Furthermore, the theory of section 2 is not able to predict the dependence of the lateral dose fall-off on the proton energy $E_p$. According to equation (2) the lateral dose fall-off scales with the square-root of the angular variance which in turn scales with the scattering power. The scattering power scales approximately with $1/E_p$ (Gottschalk 2010). Thus, a decrease of the lateral dose fall-off is expected with increasing proton energy $E_p$. Table 1 shows that this dependence was not observed in the experiments. For instance, the lateral dose fall-off values are larger for the 200 MeV field than for the 100 MeV field. We conclude that the lateral gradients can only be explained if edge scatter effects are included as, e.g. described in Kimstrand et al. (2008) and van Luijk et al. (2001). This holds especially for the ‘horns’ (or ‘ears’) and the rounded corners at the edges of the high-dose region (van Luijk et al. 2001).

Ciocca et al. (2019) describes a treatment head design similar to the one of the current work adhering to the principle of a RS far upstream of the aperture. The resulting sharp lateral dose gradients qualitatively agree with the finding of the current work. The lateral dose fall-off values are a bit smaller than the ones of the current work. The thinner RS used in Ciocca et al. (2019) (2.8 g cm$^{-2}$) is a possible explanation. In contrast to the current work, Ciocca et al. (2019) reports about a clinically commissioned treatment room. Our work extends the work Ciocca et al. (2019) by

- providing a theoretical motivation for the chosen arrangement of beam shaping components in the nozzle
- extending the energy range beyond the specification for ocular treatments
- demonstrating the feasibility of a sharp lateral penumbra for a rather thick RS with a WET of 6.3 g cm$^{-2}$ (2.8 g cm$^{-2}$ in Ciocca et al. (2019))

The RS of the current work degrades a proton beam from 89 MeV to 0 MeV. Considering that the minimum accelerator energy is 60 MeV to 70 MeV in new installations of proton therapy centers, even sharper values of the lateral penumbra are expected than reported in this work. In the current work we applied beam energies up to 200 MeV. A beam energy of 200 MeV in conjunction with a RS would be clinically applied if the modulations width exceeds 12 cm, which is unlikely considering the small target volumes considered in this work. So the measurements and simulations at high beam energies only serve to validate the beam model and the dose engine, but are not representative of clinical cases.

The current work continues the discussion if the aperture-downstream configuration (figure 1 middle left) is superior to the aperture-upstream configuration (figure 1 left) in terms of lateral dose fall-off (Winterhalter et al. 2018). Ciocca et al. (2019) and the current work have shown the superiority of the aperture downstream configuration at the field edge. However, there are two restrictions. First, the spot width towards the field center increases thereby decreasing the benefits of thefluence modulation. Secondly, an extreme reduction of the lateral penumbra is only possible if the RS can be mounted close to the virtual source.

Owing to the simple beam delivery method, the beam modelling is simple, too. As Monte Carlo dose engines are now available in commercial treatment planning systems, an accurate estimation of the dose distribution is possible. The large distance between RS and calculated dose volume puts high demands on the implementation of the multiple Coloumb scattering in the dose engine (see figure 5). In this regard and considering also the extreme heterogeneity at the aperture edge, the fields of the current project may serve as experimental benchmark data for Monte Carlo dose simulations (Gottschalk et al. 1993, Gottschalk 2018). This may partly explain the differences between measurements and simulations (table 1 in section 4.2). The observed differences of the measured and simulated spot profiles (table 2 in section 4.3) could be explained by the fact that the beam model has been created using spot profiles measured over a longitudinal distance of 40 cm around isocenter, but is now used in a geometry where the transport starts 171 cm from isocenter. This quite long extrapolation increases the uncertainties of the beam modelling process. We also point out that the sharp gradients observed in the current work require a fine dose grid. Ideally, the granularity of the dose grid should be below 1 mm, which is the case for the ocular module in RayStation under development.

The current work did not characterize the out-of-field dose as performed in Ciocca et al. (2019). Because the snout of IBA universal nozzle was designed to minimize leakage in the double scattering mode and uniform scanning mode (figure 1 right), which use more scatterers than SSM with RS, the dose level outside the collimated volume is assumed to be clinically acceptable. According to Ciocca et al secondary neutrons are mainly produced in the brass aperture (Ciocca et al 2019). The wider spots of the pbs-ds-stereo configuration increase the area of
brass which intercepts protons compared to the ap-us and ap-ds configurations. Thus, we expect a larger secondary neutron dose in the pbs-ds-stereo configuration. The experimental validation of the out-of-field dose should be subject of future studies.

Although the title of the current work refers to stereotactical applications, we did not discuss some important aspects of stereotactical radiation therapy systems, e.g. geometrical accuracy. The aim of the current project was to overcome the most important obstacle of using protons in stereotactical applications, which is the lateral dose-fall off (Wang et al 2014). This was motivated by the clinical quest to expand the use of protons in stereotactical radiotherapy (Vernimmen 2016). Considering also the simulation study of Wang et al (2014), the lateral dose fall-off up to water equivalent depths up to 12.5 cm is sharp enough to facilitate treatment plans with better quality than corresponding photon techniques. Furthermore, the results of the current work are not limited to the use of fixed apertures. We expect similar benefits for dynamic collimation systems with moving trimmer blades, which are described, e.g. in Moignier et al (2016). As an outlook we note that, besides small intracranial lesions, the results of the current project and those of Ciocca et al (2019) are also expected to be beneficial for stereotactical body radiotherapy: As the field is composed of large spots, it is assumed that the fields have favorable properties in regions affected by respiratory motion as shown in Grassberger et al (2013).

6. Conclusions

It was shown as a proof-of-principle that proton fields with 80%–20% lateral dose fall-offs between 1.4 mm and 4.0 mm can be realized in spot-scanning mode for water equivalent depths between 0 cm and 10 cm. This was achieved by a single, mechanically simple hardware modification of an existing commercial proton therapy nozzle. In this test environment lateral dose planes were measured. A theoretical motivation based on analytical equations was provided indicating the benefit of a range shifting block far upstream of a collimating aperture for divergent fields. The sharpening of the field edge is inherently connected to an enlargement of proton spots in the center. Thus, the experiments and simulations focused on small fields. It was argued that Monte Carlo simulations are necessary to estimate the dose distribution. We have shown that the accuracy of the Monte Carlo dose engine of a commercial treatment planning system is sufficient. The envisaged applications are stereotactical irradiations and small-animal experiments.

Acknowledgments

The authors would like to thank Manuel Beck for design and manufacturing of the range shifter in the downstream configuration. Many thanks to the WPE physics team and the local IBA team Essen. We acknowledge Ajvar Kern for his support regarding the film calibration.

Conflict of interest statement

Christian Bäumer, Beate Timmermann and Jörg Wulff declare no conflicts of interest. Martin Janson works for RaySearch Laboratories. Carolina Lina Fuentes and Andrija Matic work for IBA PT.

ORCID iDs

J Wulff https://orcid.org/0000-0001-8260-3523

References

Bäumer C and Farr J B 2011 Lateral dose profile characterization in scanning particle therapy Med. Phys. 38 2904–13
Bäumer C et al 2017 Comprehensive clinical commissioning and validation of the RayStation treatment planning system for proton therapy with active scanning and passive treatment techniques Phys. Med. 43 15–24
Bäumer C, Janson M, Timmermann B and Wulff J 2018 Collimated proton pencil-beam scanning for superficial targets: impact of the order of range shifter and aperture Phys. Med. Biol. 63 085020
Charlwood F C, Atkinson J H and Mackay R I 2016 A Monte Carlo study on the collimation of pencil beam scanning proton therapy beams Med. Phys. 43 1462–72
Ciocca M et al 2019 Design and commissioning of the non-dedicated scanning proton beamline for ocular treatment at the synchrotron-based CNAO facility Med. Phys. 46 1852–62
Courtois C, Beissonnat G, Brusasco C, Colin J, Cassoul D, Fontbonne J, Marchand B, Mertens T, De Neuter S and Peronnel J 2014 Characterization and performances of a monitoring ionization chamber dedicated to IBA-universal irradiation head for pencil beam scanning Nucl. Instrum. Methods Phys. Res. A 736 112–7
Devic S 2011 Radiochromic film dosimetry: past, present, and future Phys. Med. 27 122–34
Engelsman M, Schwarz M and Dong L 2013 Physics controversies in proton therapy Semin. Radiat. Oncol. 23 88–96
Gottschalk B 2010 On the scattering power of radiotherapy protons Med. Phys. 37 352–67
Gottschalk B 2018 Deterministic proton dose calculation from first principles Phys. Med. Biol. 63 135016
Gottschalk B, Koehler A, Schneider R, Sisterson J and Wagner M 1993 Multiple coulomb scattering of 160 MeV protons *Nucl. Instrum. Methods Phys. Res. B* 74 467–90
Grassberger C, Dowdell S, Lomax A, Sharp G, Shackelford J, Choi N, Willers H and Paganiatti H 2013 Motion interplay as a function of patient parameters and spot size in spot scanning proton therapy for lung cancer *Int. J. Radiat. Oncol. Biol. Phys.* 86 380–6
Hong L, Goitein M, Bucciolini M, Comiskey R, Gottschalk B, Rosenthal S, Serago C and Urie M 1996 A pencil beam algorithm for proton dose calculations *Phys. Med. Biol.* 41 1305
ICRU 2007 ICRU REPORT 78: prescribing, recording, and reporting proton-beam therapy *J. ICRU* 7
Kanematsu N, Yonai S, Ishizaki A and Torikoshi M 2008 Computational modeling of beam–customization devices for heavy-charged-particle radiotherapy *Phys. Med. Biol.* 53 3113
Kimstrand P, Traneus E, Ahnesjö A and Tilly N 2008 Parametrization and application of scatter kernels for modelling scanned proton beam collimator scatter dose *Phys. Med. Biol.* 53 3405
Maes D, Begmi R, Taddel P, Bloch C, Bowen S, Nevitt A, Leuro E, Wong T, Rosenfeld A and Saini J 2019 Parametric characterization of penumbra reduction for aperture-collimated pencil beam scanning (PBS) proton therapy *Biomed. Phys. Eng. Express* 5 035002
Meignier A, Gelover E, Wang D, Smith B, Flynn R, Kirk M, Lin L, Solberg T, Lin A and Hyer D 2016 Theoretical benefits of dynamic collimation in pencil beam scanning proton therapy for brain tumors: dosimetric and radiobiological metrics *Int. J. Radiat. Oncol. Biol. Phys.* 95 171–80 (particle therapy special edition)
Moteabbed M, Yock T I, Depauw N, Madden T M, Kooy H M and Paganiatti H 2016 Impact of spot size and beam-shaping devices on the treatment plan quality for pencil beam scanning proton therapy *Int. J. Radiat. Oncol. Biol. Phys.* 95 190–8 (particle therapy special edition)
Nichiporov D, Hsi W and Farr J 2012 Beam characteristics in two different proton uniform scanning systems: a side-by-side comparison *Med. Phys.* 39 2559–68
Rana S, Zeidan O, Ramirez E, Rains M, Gao J and Zheng Y 2013 Measurements of lateral penumbra for uniform scanning proton beams under various beam delivery conditions and comparison to the xio treatment planning system *Med. Phys.* 40 091708
Russo S, Mirandola A, Molinelli S, Castiglia E, Vai A, Magro G, Mairani A, Boi D, Donetti M and Ciocca M 2017 Characterization of a commercial scintillation detector for 2-D dosimetry in scanned proton and carbon ion beams *Phys. Med.* 34 48–54
Sabbas A M, Jette D, Rozenfeld M, Pagnamenta A and Land L H 1987 Collimated electron beams and their associated penumbra widths *Phys. Med. Biol.* 14 996–1006
Safai S, Borlind T and Engelsman M 2008 Comparison between the lateral penumbra of a collimated double-scattered beam and uncollimated scanning beam in proton radiotherapy *Phys. Med. Biol.* 53 1729
Saini J, Maes D, Egan A, Bowen S R, James S S, Janson M, Wong T and Bloch C 2017 Dosimetric evaluation of a commercial proton spot scanning Monte Carlo dose algorithm: comparisons against measurements and simulations *Phys. Med. Biol.* 62 7659
van Luijk P, van Luijk P, van ’t Veld A A, Zelle H D and Schippers J M 2001 Collimator scatter and 2D dosimetry in small proton beams *Phys. Med. Biol.* 46 653
Vernimmen F 2016 Intracranial stereotactic radiation therapy with charged particle beams: an opportunity to regain the momentum *Int. J. Radiat. Oncol. Biol. Phys.* 95 52–5
Wang D, Dirksen B, Hyer D E, Buatti J M, Sheybani A, Dinges E, Felderman N, Ten Napel M, Rayborth J E and Flynn R T 2014 Impact of spot size on plan quality of spot scanning proton radiosurgery for peripheral brain lesions *Med. Phys.* 41 121703
Wang D, Smith B R, Gelover E, Flynn R T and Hyer D E 2015 A method to select aperture margin in collimated spot scanning proton therapy *Phys. Med. Biol.* 60 N109–19
Winterhalter C, Lomax A, Oxley D, Weber D C and Safai S 2018 A study of lateral fall-off (penumbra) optimisation for pencil beam scanning (PBS) proton therapy *Phys. Med. Biol.* 63 025022
Yasui K, Toshihito T, Otomichi C, Hayashi K, Tanaka K, Asai K, Shimomura A, Muranatsu R and Hayashi N 2017 Evaluation of dosimetric advantages of using patient-specific aperture system with intensity-modulated proton therapy for the shallow depth tumor *J. Appl. Clin. Med. Phys.* 19 132–7