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Golden Beam Data for Proton Pencil Beam Scanning

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Abstract

Proton, as well as other ion, beams applied by electro-magnetic deflection in pencil-beam scanning (PBS) are minimally perturbed and thus can be quantified a-priori by their fundamental interactions in medium. This a-priori quantification permits an optimal reduction of characterizing measurements on a particular PBS delivery system. The combination of a-priori quantification and measurements will then suffice to fully describe the physical interactions necessary for treatment planning purposes. We consider, for proton beams, these interactions and derive a “Golden” beam data set. The Golden beam data set quantifies the pristine Bragg peak depth dose distribution in terms of primary, multiple Coulomb scatter, and secondary, nuclear scatter, components. The set reduces the required measurements on a PBS delivery system to the measurement of energy spread and initial phase space as a function of energy. The depth doses are described in absolute units of Gy(RBE).mm².Gp⁻¹, where Gp equals 10⁹ (giga) protons, thus providing a direct mapping from treatment planning parameters to integrated beam current. We used this Golden beam data on our PBS delivery systems and demonstrate that it yields absolute dosimetry well within clinical tolerance.

1. Introduction

Historically, proton beams have been clinically deployed with in-beam scattering and modulation delivery systems that create spread-out pristine Bragg peak (SOBP) depth-dose distributions with uniform lateral intensity. These delivery systems used mechanical means, with minimal electronic control, to create the SOBP distributions and thus allowed the early (circa 1960) use of proton beams in radiotherapy. The primary dosimetric characteristic of an SOBP field is its uniform dose profile over its distal to proximal energy (range) interval. Only recently have pencil-beam scanning (PBS) delivery systems been introduced to use the unperturbed proton beam with magnetic, energy and current control to create modulated fields to permit the delivery of arbitrary dose distributions in the patient. The clinical use of PBS fields for protons was first introduced at the Paul Scherrer Institute [1]. Its delivery system, however, used a combination of range-shifters and magnetic controls to create such fields and thus is not considered an “unperturbed” PBS delivery system.

Commissioning of SOBP delivery systems has been complicated by the variety in construction of such systems, with no accepted standard, and the perturbations from proton interactions with the materials of the delivery system. Thus, no standard approach is, in effect, possible and each system must be carefully analyzed for its effect on the desired

SOBP depth dose distribution. The commissioning of one such system is described by Engelsman *et al.* [2].

Fully electro-magnetic PBS delivery systems, in contrast, are by definition all equal. The future use of such PBS systems, in favor of SOBP delivery by scattering systems, is inevitable as PBS treatments will prove to outperform SOBP treatments in all clinical performance criteria. These include quality of dosimetry, efficiencies in planning and delivery, and the ability to effectively compete with photon radiotherapy techniques for image-guided and adaptive radiotherapy (see for example Schippers and Lomax [3], Kooy *et al.* [4] and Lomax [5]).

The accuracy required for PBS commissioning is atypical compared to conventional measurement practices in radiotherapy. The individual pristine Bragg peak depth dose distributions must be quantified accurately to achieve better than 1% dosimetric accuracy in standard geometries and to allow an accurate mapping between the treatment planning system derived pristine Bragg peak intensities (in units of Gp in our practice) and the measured current, often integrated, on the delivery system as observed in its reference ionization chamber. The purpose of this work is to simplify commissioning procedures and reduce the measurement burden associated with commissioning proton PBS systems by constructing a Golden data set of Monte Carlo generated pristine Bragg peaks that are validated with measurements.

We describe our process, based on GEANT4 Monte Carlo calculations and measurements, to determine the complement of parameters for commissioning any general PBS delivery system for treatment planning and delivery of PBS treatment fields. We distinguish between universal data, applicable to any PBS delivery system, and those that require measurements. GEANT4 has been used by others [6] [7] for proton dosimetry purposes. Our approach, however, aims to derive an equipment independent data set which, in turn, can be applied to any PBS delivery system.

2. Methods

2.1. Depth Doses

We use GEANT4 Monte Carlo Version 4.8.1p01 [8][6] for generating pencil-beam depth-dose distributions and cross calibrate these results against measurable, composite, observables such as pristine and SOBP depth doses, absolute point doses and broad-field planar measurements. We use the standard electromagnetic model and the HElastic and binary cascade nuclear interaction models. Protons, neutrons, electrons, and photons are tracked with maximum step size of 0.2 mm and default cut value of 0.05 mm while particles below this threshold or other types of particle deposit their energy locally. We generate a full complement of pristine Bragg peaks in the clinical range of 20 to 350 mm (in water), all with an energy spread $\Delta E/E = 0$, and at peak-to-peak intervals of 5 mm. The pristine peaks are generated for a zero emittance beam in a $300 \times 300 \times 0.2 \text{ mm}^3$ slab geometry. The pristine peaks have units of $\text{Gy(RBE).mm}^2.\text{Gp}^{-1}$ in correspondence with Pedroni *et al.* [9].

We normalize the absolute units of the generated peaks against depth dose measurements with a Markus plane-parallel ionization chamber[‡] for single energy (i.e. pristine peak) broad-fields ($300 \times 300 \text{ mm}^2$). The Markus chamber is cross calibrated in passive-scattering fields according to IAEA TRS-398 dosimetry protocol [10]. The use of broad fields, again, best approximates the full capture of scattered and secondary contributions and also permits the use of a finite-size ionization chamber. The measurements thus verify and establish the

[‡]Model N23343, PTW Freiburg Germany

exact ratio of Gy to Gp. The measurements are done at three depths in water: 1 cm, $R_{80}/2$ and a point intermediate to the other two. Three depths are used to quantify the uncertainty due to the small, but nonzero energy spread of the beam in the measurement. Because the energy spread has negligible effect on the absolute dose at shallow depths, and only becomes significant near the Bragg peak, the uncertainty is given by the variation in the normalization between the three depths. Further measurements are done with reduced slit settings in the energy selection system to verify the normalization is indeed independent of the energy spread. The normalization factor, $C_D(R_{80})$, is applied to the generated peaks to yield the set of experimentally verified pristine peaks of $\Delta E/E = 0$.

We quantify the energy spread of a particular PBS delivery system by measuring individual, but limited number of, pristine peaks with the Bragg peak chamber[§]. The Bragg peak chamber is a wide parallel-plate ionization chamber that is designed to intercept all the protons in a pencil beam aligned with the center of the chamber. Measurements are at shallow depths and in the Bragg peak. The Bragg peak chamber area does not suffice to capture all energy deposited by the beam and we correct these measurements with a depth- and energy-dependent correction for the non-primary contribution derived from Monte Carlo. These measurements yield the total depth-dose of the delivery system, $\Delta E/E \neq 0$, Bragg peaks. These are fitted against a convolution of the $\Delta E/E = 0$ pristine peaks as a function of the energy spread as in Kimstrand *et al.* [11]. We subsequently fit the results against a 3d-order polynomial to describe $\Delta E/E(R_{80})$. This function yields the energy spread to convolve the 0-width peaks to produce a complete set for the observed, delivery-system specific, Bragg peaks. This last step is the only facility-dependent calibration. No correction for inverse square is necessary as an individual pencil-beam does not diverge.

2.2. Lateral Dose Profile

We assume the lateral dose profile at any R_{80} and depth is described by a superposition of two Gaussian distributions, one for the primary dose and the other for the “halo” contribution which comes from wide, single scattered protons and secondary particles. The relative contribution of the primary Gaussian as a function of depth is given by the parameter $\alpha(d, R_{80})$ and where the primary and halo Gaussian widths are $\sigma_1(d, R_{80})$ and $\sigma_2(d, R_{80})$, respectively. Both widths are convolved with the initial in-air spread σ_0 of the

pencil-beam itself giving $\sigma_1(d, R_{80}) = \sqrt{\sigma_M^2(d, R_{80}) + \sigma_0^2(R_{80})}$ and

$\sigma_2(d, R_{80}) = \sqrt{\sigma_H^2(d, R_{80}) + \sigma_0^2(R_{80})}$. The energy and depth dependence of the primary spread due to multiple Coulomb scattering σ_M is described, for example, by Hong *et al.* [12]. Generally, the in-air spread can depend on depth, *i.e.* $\sigma_0(d, R_{80})$, and this can be adopted without any change to the method.

The dose at a distance r from the central axis of a pencil-beam with range R_{80} is [9]

$$\delta(d, r) \propto D_{\infty}(d, R_{80}, \Delta E/E) [\alpha(d, R_{80})G(r, \sigma_1(d, R_{80})) + (1 - \alpha(d, R_{80}))G(r, \sigma_2(d, R_{80}))] \quad (1)$$

where $G = (2\pi\sigma^2)^{-1} \exp(-r^2/2\sigma^2)$ and $D_{\infty}(d, R, \Delta E/E)$ is the depth dose of pristine proton beam infinite broad width and infinite source-to-axis distance. We determine the parameters α and σ_H in relation to the measured depth dose D_{∞} .

We adapt a radial geometry of beam delivery and measure the dose to the center of the scanning pattern. Pristine pencil-beams are delivered in concentric circles centered on the

[§]Model TN34070, PTW Freiburg Germany

measurement point. The increasing circumference of the circle pattern with r amplifies the small secondary contribution that decreases with r . The pattern, furthermore, creates a flat dose distribution in the center which reduces uncertainties due to interpolation between pixels.

We define circles of radii r_j , where $r_{j+1} - r_j \approx \sigma_O$, and where

$$n_j = \max\left(1, \text{floor}\left(0.5 + \frac{2\pi r_j}{\sigma_O}\right)\right) \quad (2)$$

pencil-beams are placed in circle j . The dose at depth d from a circle j packed with n_j identical pencil-beams and with identical flux is $D_j \propto n_j \delta(r_j, d)$ which, when normalized by D_0 , is

$$D_j = \frac{n_j(\alpha G(r_j, \sigma_M) + (1-\alpha)G(r_j, \sigma_H))}{\alpha/2\pi\sigma_M^2 + (1-\alpha)/2\pi\sigma_H^2} \quad (3)$$

The packing of circles results in quantized number of pencil-beams and D_j is not a continuous function. We fit equation 3 to the measured data to extract the $\alpha(d)$ and $\sigma_{M,H}(d)$ parameters.

3. Results

3.1. The Golden Bragg Peaks

A complete set of pristine Bragg peaks is generated by Monte Carlo in absolute units of Gy(RBE).mm².Gp⁻¹. We use the generated Bragg peaks to calculate the expected absolute dose in the center of a broad field and cross-calibrate Gy per Gp at three normalization depths by measuring dose with a Markus chamber and charge per spot with a Faraday cup. These measurements yield a small correction factor, $C_D(R_{80})$, shown in figure 1. The $C_D(R_{80})$ is consistent between the three depths within $\pm 1\%$ so the points are not sensitive to the finite, but nonzero energy spread of the real beam and thus can be used to normalize the Monte Carlo results. Further experimental verification of this assumption is made by closing the energy selection slits from 40 mm to 5 mm, reducing the $\Delta E/E$, and there is no difference in the measured dose per gigaproton at the normalization depths.

The factor $C_D(R_{80})$ is applied to each pristine Bragg peak and results in the set of peaks shown in figure 1. The set of Golden pristine Bragg peaks, (GPPS), with $\Delta E/E = 0$, is universal and accurate within measurement tolerances characteristic of the Markus chamber (2.2%) and the Faraday Cup (3%). The total, but largely systematic, error in this measurement is thus $\approx 4\%$. We address the final correction for this error in section 3.3.

3.2. Equipment Commissioning

3.2.1. Energy Spread Calibration—We claim that the energy spread $\Delta E/E$ is the only parameter that describes the pristine peak depth doses for a specific PBS delivery system. In principle, one could use models of the accelerator and, if applicable, the energy selection system to determine this spread [13]. We determine $\Delta E/E$ by measuring proton pencil-beams with a PTW Bragg peak chamber at a limited set of energies. The PTW Bragg peak chamber (BPC) is too “small,” however, and requires a correction for dose deposited outside the 81.6 mm diameter chamber. This correction has also been observed in Grevillot *et al.* [7].

Monte Carlo simulations are used to determine this correction, which can be as large as 7.0% at 225 MeV. For these simulations, a beam with zero $\Delta E/E$ and zero position and angular spread is incident on a water phantom that incorporates a model of the Bragg peak chamber. The correction, D_{BPC} in absolute units [Gy(RBE).mm².Gp⁻¹], is a function of depth d in units [g.cm⁻²] and range R_{80} in units [g.cm⁻²]. The result is parameterized as follows

$$D_{BPC}(d, R_{80}) = 110 \times \max(0, p_{35}t^{35} + p_4t^4 + p_3t^3 + p_2t^2 + p_0), \quad (4)$$

where,

$$t = d/R_{80},$$

$$p_{35} = \min(0, 1.585 \times 10^{-5} R_{80}^2 - 0.004559 R_{80} + 0.04214),$$

$$p_4 = 0.002133 R_{80}^2 - 0.04721 R_{80} + 0.195,$$

$$p_3 = -0.003363 R_{80}^2 + 0.0668 R_{80} - 0.3058,$$

$$p_2 = 0.001297 R_{80}^2 - 0.0174 R_{80} + 0.08441, \text{ and,}$$

$$p_0 = (0.4087 R_{80}^3 - 21.25 R_{80}^2 + 326 R_{80} + 215.6) \times 10^{-6}.$$

Figure 2 shows the correction factor for two pristine peaks. It is, if nothing else, this unmeasurable correction factor that makes commissioning of a PBS delivery system by measurement-only difficult.

The pencil beam incident on the phantom in the measurement has small $\Delta E/E \sim 1\%$ (1σ) and small lateral width ($\sigma_0 \approx 5$ mm). The dose correction is, without modification, added to the measured data to obtain the total depth-dose distribution $D_{\infty}(d, R_{80}, \Delta E/E)$ even though the simulation is done with an infinitesimally narrow width beam. We call this the narrow-beam approximation. This approximation is valid when the primary beam width, σ_1 , is much smaller than the radius of the chamber so that the dose deposited outside the chamber is

completely described by the halo with width σ_2 . Furthermore, because $\sigma_2 = \sqrt{\sigma_H^2 + \sigma_0^2}$, we also need σ_0 to be much smaller than σ_H to use this approximation. To estimate the error due to the narrow-beam approximation in our configuration, we ran the Monte Carlo again with $\sigma_0 = 5$ mm and the change in the total depth-dose distribution is less than 0.5%. In fact, one can make the narrow-beam assumption for dose deposited outside the chamber with $\sigma_0 = 10$ mm and accuracy better than 1% relative to the total dose.

A database of facility-specific $D_{\infty}(d, R_{80}, \Delta E/E)$ is constructed by smearing the zero energy width GPPS by a Gaussian function of width $\Delta E/E$. At our institution, the Gaussian parameters come from a polynomial fit to the limited set of measurements with the Bragg peak chamber corrected using equation 4. Figure 3 (left) shows the necessary steps to achieve the fit for a single pristine peak, while figure 3 (right) shows the derived $\Delta E/E$ as a function of energy. Figure 3 (right) also shows an independent measurement of the energy spread using an NaI scintillator in spectroscopic mode [13] and the agreement is much better than 0.1 MeV. The derived $\Delta E/E$ are fitted to a polynomial and applied in the convolution of each GPPS pristine peak to yield the pristine peak data set for our PBS delivery system. We verify the MGH specific pristine peaks in a broad-field SOBP in section 3.3.

This process may appear circular at first glance, since we start this section with the measurement of non-zero $\Delta E/E$ Bragg peaks, fit the energy spread, then calculate non-zero $\Delta E/E$ Bragg peaks at all energies. This process, however, reduces the required number of measurements because $\Delta E/E$ is generally a slowly varying function of energy. The

parameterization of $\Delta E/E$ together with the GPPS from figure 1 allows us to interpolate between measured Bragg peaks. If the energy spread is already known, the D_{∞} database can be constructed in one step, by convolving the $\Delta E/E$ together with the GPPS.

3.2.2. Delivery System Calibration—The PBS delivery system must be calibrated in terms of the required range in patient, as controlled by the energy selection system of the accelerator, and in terms of the absolute number of protons, as controlled by a reference ionization chamber. In our treatment planning system, we use the range calibration to generate a lookup-table of beamline energy against range in patient. The latter is expressed both in terms of R_{80} as well as R_{90} as the latter is used in standard communication about range in patient. Figure 5 (left) shows the difference between requested and observed range in patient.

The reference ionization chamber response, MU, is calibrated against the number of protons using a Faraday cup. The ionization in the air of the reference ionization chamber depends on the energy of protons passing through the chamber and the response MU is proportional to the stopping power in air. This proportionality only requires, in principle, to establish the relation $MU(R_{80})$ at one point. This proportionality to the stopping power is confirmed by our measurements as shown figure 5 at multiple such points.

The intrinsic lateral spread σ_0 as a function of energy of the proton pencil-beam emerging from the beam-line is also characterized. This parameter is independent of the others and characteristic of the beam-line and accelerator. The beam-line is characterized by the series of magnets in-between the accelerator and delivery system. The accelerator, currently, is either a cyclotron or synchrotron. The second will have better initial emittance, compared to a cyclotron, as there is no need to degrade the beam at the exit of the synchrotron. Modeling the beam-line optics, with the initial emittance as input, will allow a calculation of the pencil-beam spread at isocenter.

3.3. Absolute Dose Verification

At this point, all the elements to deliver broad-field depth dose distributions, in absolute dose, have been calibrated. These are: (1) absolute pristine peak depth doses in units of $\text{Gy(RBE).mm}^2.\text{Gp}^{-1}$, (2) absolute calibration of ionization chamber response in terms of MU per Gp, and (3) calibrated range in water versus equipment energy setting. We use these elements to construct, in our treatment planning system (Astroid), SOBP's as these are sensitive to the exact shape of the pristine peaks that comprise the SOBP. The delivered dose (figure 4) agrees within $\pm 2\%$ and even exhibits the ripples in the computed SOBP distribution.

3.4. Secondary Contributions

We measure the dose in the center of concentric circles of increasing radii and increasing number of pencil-beams centered on the circles (equation 3). The result for one such measurement is shown in figure 6. The measurements are tedious and we limited our measurements to $R_{90} = 150, 200$ and 250 mm. Our aim is to achieve a compact description of the secondary contribution whose major effect on dosimetry is to leak energy away from the pristine peak dose along its axis. The leaked energy is small, on the order of 5%, and only requires a phenomenological quantification (see Soukup *et al.* [15]). We therefore extract the parameter α for these three energies, only, and analyze its behavior against measurement depth normalized by range, i.e. $t = d/R$ where t and R are in units of [cm of water]. We fit α to a 2^d -order polynomial

$$\alpha(R) = \min(1, a_0(R) + a_1(R) \cdot t + a_2(R) \cdot t^2) \quad (5)$$

where $a_i(R)$ is further, to the point of incredulity but in an effort for optimal data reduction, fitted to a 2^d -order polynomial of R

$$a_i(R) = b_{0,i} + b_{1,i} \cdot R + b_{2,i} \cdot R^2 \quad (6)$$

This results in the fit values in table 1. Figure 7 shows the fitted results for $\alpha(R, t = d/R)$ and illustrates the ability of the derived form to decompose a pristine peak depth dose into its primary and secondary components.

Finally, the Gaussian spread of the secondary contribution, is observed to be nearly invariant of depth and is described as a function of R in units of [cm of water] by

$$\sigma_H(R) = 6.5 - 3.4 \cdot R + 0.078 \cdot R^2 \text{ mm} \quad (7)$$

The overall validity of the data analysis, in spite of its possible overextension, is verified against central axis dose build-up of which one field is shown in figure 8.

4. Discussion

We derived a universal set of pristine Bragg peaks of $\Delta E/E = 0$ energy spread. These peaks have units of $\text{Gy(RBE)} \cdot \text{mm}^2 \cdot \text{Gp}^{-1}$. These peaks can be used to create arbitrary absolute dose-distributions (those in figure 4, for example) and the required intensities of the peaks that comprise these distributions is directly characterized in the number of protons (Gp) required to deliver those distributions. We independently calibrated our reference ionization chamber in terms of MU per Gp as a function of proton energy. Note that there are no free parameters in this calibration. That is, treatment planning specifies charge (Gp) which is converted, linearly, to MU (figure 5). In fact, we are able to deliver absolute dose distributions within 1% using only the above-stated information.

PBS dosimetry is different and simpler compared to IMRT dosimetry. In IMRT, dose in patient is mapped, non-linearly, to multi-leaf positions and motions. In PBS, in contrast, dose in patient is specified in charge which becomes the direct input to the PBS delivery system (after conversion with the stopping power in air) and which, again, is directly observed in the reference ionization chamber. The observed MU can be directly mapped to delivered charge which, in turn, can be directly re-input into the treatment planning system to compute the delivered dose in medium. The dosimetry of PBS is therefore considerably simpler, and safer because of its direct back and forth mapping of dose and charge, compared to both SOBP field dosimetry and IMRT field dosimetry.

The Bragg peak chamber response is almost independent of the particulars of the PBS beamline for proton pencil-beam spreads ($1\sigma_0$) as large as ≈ 10 mm. The cross-calibration of the Monte Carlo generated pristine peaks with the Bragg peak chamber can be done accurately after correcting for the chamber response. We argue, of course, that the GPPS data set is universal and can be applied to any PBS delivery system without the need of repeating or correcting Bragg peak chamber measurements.

The range calibration between range-in-patient and energy of the energy selection interface to the accelerator and beamline is, of course, specific to each vendor if not to each individual installation.

The secondary interactions distribute energy over a much larger volume compared to the primary interactions and the combined lateral distribution is modeled by a superposition of two Gaussian distributions. The primary objective is to account for the quantity of energy that must be subtracted from the pristine peak (figure 7). The secondary objective is to redistribute this energy over the larger volume characterized by the spread (equation 7) of the secondary energy. Both these objectives, however, depend on the particulars of a dose algorithm implementation. In our pencil-beam algorithm, for example, we first compute the primary proton transport at high resolution, with mathematical pencil-beams of zero emittance and an area of $2 \times 2 \text{ mm}^2$ and aggregate the primary transport into the physical, with an area of $9\pi\sigma_0^2$ pencil-beam spots. The secondary transport is only considered at the physical pencil-beam spot level.

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References

1. Pedroni E, Bacher R, Blattmann H, Böhringer T, Coray A, Lomax A, Lin S, Munkel G, Scheib S, Schneider U, Tourovsky A. The 200 MeV proton therapy project at the Paul Scherrer Institute: Conceptual design and practical realization. *Medical Physics*. 1995; 22(1):37–53. [PubMed: 7715569]
2. Engelsman M, Lu H-M, Herrup D, Bussiere MR, Kooy HM. Commissioning a passive-scattering proton therapy nozzle for accurate SOBP delivery. *Medical Physics*. 2009; 36(6):2172–2180. [PubMed: 19610306]
3. Schippers JM, Lomax A. Emerging technologies in proton therapy. *Acta Oncologica*. 2011; 50(6): 838–850. [PubMed: 21767183]
4. Kooy HM, et al. A case study in proton pencil-beam scanning delivery. *Int J Radiat Oncol Biol Phys*. 2010; 76(2):624–630. [PubMed: 20117294]
5. Lomax A. Intensity modulation methods for proton radiotherapy. *Phys Med Biol*. 1999; 44:185–205. [PubMed: 10071883]
6. Paganetti H, Jiang H, Lee S-Y, Kooy HM. Accurate Monte Carlo simulations for nozzle design, commissioning and quality assurance for a proton radiation therapy facility. *Medical Physics*. 2004; 31(7):2107–2118. [PubMed: 15305464]
7. Grevillot L, Frisson T, Zahra N, Bertrand D, Stichelbaut F, Freud N, Sarrut D. Optimization of GEANT4 settings for Proton Pencil Beam Scanning simulations using GATE. *Nuclear Instruments and Methods in Physics Research B*. 2010; 268(20):3295–3305.
8. Agostinelli S, et al. Geant4 - a simulation toolkit. *Nuclear Instruments and Methods A*. 2003; 506(3):250–303.
9. Pedroni E, Scheib S, Böhringer T, Coray A, Grossmann M, Lin S, Lomax A. Experimental characterization and physical modelling of the dose distribution of scanned proton pencil beams. *Physics in Medicine and Biology*. 2005; 50(3):541–561. [PubMed: 15773729]
10. Technical report series no. 398. International Atomic Energy Agency; 2000. Absorbed Dose Determination in External Beam Radiotherapy: An international Code of Practice for Dosimetry Based Standards of Absorbed Dose to Water.
11. Kimstrand P, Traneus E, Ahnesjö A, Grusell E, Glimelius B, Tilly N. A beam source model for scanned proton beams. *Physics in Medicine and Biology*. 2007; 52(11):3151–3168. [PubMed: 17505095]
12. Hong L, Goitein M, Bucciolini M, Comiskey R, Gottschalk B, Rosenthal S, Serago C, Urie M. A pencil beam algorithm for proton dose calculations. *Physics in Medicine and Biology*. 1996; 41(8): 1305–1330. [PubMed: 8858722]
13. Cascio EW, Sisterson JA, Gottschalk B, Sarkar S. Measurements of the energy spectrum of degraded proton beams at NPTC. *Radiation Effects Data Workshop, IEEE*. 2004:151–155.

14. Cascio EW, Gottschalk B. A Simplified Vacuumless Faraday Cup for the Experimental Beamline at the Francis H. Burr Proton Therapy Center. Radiation Effects Data Workshop, IEEE. 2009:161–165.
15. Soukup M, Fippel M, Alber M. A pencil beam algorithm for intensity modulated proton therapy derived from Monte Carlo simulations. Physics in Medicine and Biology. 2005; 50(21):5089–5104. [PubMed: 16237243]

Figure 1.

Left: Normalization correction to the Monte Carlo generated pristine peaks. The normalization correction, defined as the ratio of measurement over monte carlo, is measured at three depths as indicated in the legend. The “optimal” position is an intermediate depth between 1 cm and $R_{80}/2$. The correction factor as a function of R_{80} is parameterized by a 4-th order polynomial and applied to the Monte Carlo pristine peaks. Right: Monte Carlo generated pristine peaks normalized against measurement.

Figure 2.

The response of the PTW Bragg peak chamber must be corrected for the (small) fraction of dose deposited outside the detection volume. The grey curve, and its parametrization as a dashed curve, shows the missing dose as a function of depth.

Figure 3.

Left: The Golden data set is convolved with a Gaussian function, with R_{80} and $\Delta E/E$ as parameters, and fitted to the measured data (green) corrected with the PTW Bragg chamber correction factor (blue +). The $\Delta E/E = 0$ peak is shown in red. In this case $R_{80} = 25.2 \text{ g/cm}^2$ and $\Delta E/E = 1.04 \text{ MeV}$. Right: The derived $\Delta E/E$ as a function of energy. The dashed lines indicate an uncertainty band of $\pm 0.1 \text{ MeV}$ which is our requirement to achieve a 1% accuracy in measured dose distributions (see figure 4 (right)). The point (★) is an independent measurement (in NaI) of the energy spread [13].

Figure 4.

Left: Absolute dose and depth dose verification for high, medium and low energy SOBPs (parallel-plane ionization chamber measurement ($\diamond$), Astroid (in-house) treatment planning calculation (solid line) and T1 ionization chamber measurement ($\square$)). The bands indicate the $\pm 2\%$ bounds. Right: The MGH requirement for $\Delta E/E$ accuracy requires a distal fall-off accuracy better than 1%. The largest effect of $\Delta E/E$ on an SOBPs is in the distal region at low Range. The MGH requirement requires ± 0.1 MeV to achieve $\pm 1\%$ accuracy in the dose in the distal region of the plateau.

Figure 5.

Left: The proton Range delivered by the machine is calibrated with an accuracy of ± 0.06 g/cm². Right: The ionization chamber, which responds in MU, is calibrated in units of MU versus giga-protons (Gp) as a function of energy. Measurements are done with a Faraday Cup [14]. The black line is dE/dx in air scaled to match the measurements.

Figure 6.

Left: Dose in the center of concentric circles of radii r normalized to the dose at $r = 0$ for a pencil-beam with a range of $R_{90} = 200$ mm. The symbols indicate the measurements at various depth, while the dashed lines are the result of fitting equation 3. Of note, the distribution reaches a maximum at the radius $r = \sigma_O$. Our pencil-beam spread at this energy is 9 mm. Right: The values for α are plotted as a function of the range-scaled depth t/R and fitted to a 2nd-order polynomial (eq. 5). The polynomial coefficients, $[a_0(R), a_1(R), a_2(R)]$, are subsequently fitted as a function of R (eq. 6).

Figure 7.

(Left: Example decomposition of the pristine peak depth dose into the primary and secondary components. Right: The fraction of the depth dose (eq. 5) that is delivered from secondary protons as a function of range R and range-scaled depth $t = d/R$. Note the general feature where most of the secondary energy is dissipated in the middle of the pristine peak and which rises to as much as 10% at the highest energy.

Figure 8.

Dose build-up (measured and predicted (line)) on central axis at various depths for a pristine peak of range $R = 200$ mm (water) and range $R = 250$ mm (water) versus square field of increasing size s .

Table 1

Fit parameters to obtain the coefficients a_i to compute the depth-dependent halo fraction $\alpha(R, t)$ (equations 6 and 5).

f	a_0	a_1	a_2
b_0	1.002	2.128e-03	-2.549e-03
b_1	-5.900e-04	-2.044e-02	2.125e-02
b_2	0	3.178e-04	-3.788e-04