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Monte Carlo calculations and measurements of absorbed dose per monitor unit for the treatment of uveal melanoma with proton therapy

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Abstract

The treatment of uveal melanoma with proton radiotherapy has provided excellent clinical outcomes. However, contemporary treatment planning systems use simplistic dose algorithms that limit the accuracy of relative dose distributions. Further, absolute predictions of absorbed dose per monitor unit are not yet available in these systems. The purpose of this study was to determine if Monte Carlo methods could predict dose per monitor unit (D/MU) value at the center of a proton spread-out Bragg peak (SOBP) to within 1% on measured values for a variety of treatment fields relevant to ocular proton therapy. The MCNPX Monte Carlo transport code, in combination with realistic models for the ocular beam delivery apparatus and a water phantom, was used to calculate dose distributions and D/MU values, which were verified by the measurements. Measured proton beam data included central-axis depth dose profiles, relative cross-field profiles and absolute D/MU measurements under several combinations of beam penetration ranges and range-modulation widths. The Monte Carlo method predicted D/MU values that agreed with measurement to within 1% and dose profiles that agreed with measurement to within 3% of peak dose or within 0.5 mm distance-to-agreement. Lastly, a demonstration of the clinical utility of this technique included calculations of dose distributions and D/MU values in a realistic model of the human eye. It is possible to predict D/MU values accurately for clinical relevant range-modulated proton beams for ocular therapy using the Monte Carlo method. It is thus feasible to use the Monte Carlo method as a routine absolute dose algorithm for ocular proton therapy.

1. Introduction

In 1974, Constable and Koehler proposed treating uveal melanoma using low-energy proton beams, and the theoretical advantages they predicted were subsequently confirmed in clinical studies at several proton centers throughout the world (Constable and Koehler 1974). The patient outcome data revealed >99.7% local control at five years after treatment for small-and medium-sized tumors, retention of the eye in most cases and preservation of useful vision (Egger *et al* 2003, Gragoudas *et al* 1978, 1977, 1980, 1982, Munzenrider 2001, Munzenrider *et al* 1988, 1989, Seddon *et al* 1987, 1986, Suit *et al* 1977). After three decades of experience, the techniques used for treating uveal melanoma with proton therapy are fairly well understood (Gragoudas and Marie Lane 2005, Munzenrider 2001), and recent research efforts have focused on improving the dosimetric accuracy of these treatments (cf Koch and Newhauser 2005, Newhauser *et al* 2002a).

Perhaps because of the low incidence of uveal melanoma, the development of treatment planning systems to design ocular proton treatments has been limited to a few reports (Goitein and Miller 1983, Koch *et al* 2006, Pfeiffer and Bendl 2001, Sheen 2003). Koch and Newhauser (2005) highlighted the need for dose calculations of increased accuracy. They reported that a widely used treatment planning system for ocular proton therapy predicted relative dose distributions that differed from measured values by up to 12% of the maximum dose or up to 30% of the local dose at shallow depths. In a proof-of-concept analysis, Newhauser *et al* (2005) found that accuracy could be improved by using a Monte Carlo simulation model. More recently, this model was used to study the relative dose perturbations caused by small implanted fiducial markers in the eye (Newhauser *et al* 2007b)

Another limitation of presently available treatment planning systems is that they do not predict the dose per monitor unit (D/MU) value. Accurate D/MU predictions to within 3% were reported by Kooy *et al* (2003) using an analytical algorithm developed for proton fields with a wide cross-sectional area and deep penetration range. Paganetti (2006) also reported on absolute D/MU predictions using the Monte Carlo method in large-diameter treatment fields. Newhauser *et al* (2007a) have commissioned a commercially available proton treatment planning system, for treatments using large fields, using Monte Carlo generated data. However, it is unclear that to what extent the Monte Carlo technique applies to ocular treatment beams with small cross-sectional areas. Herault *et al* (2005) reported on a Monte Carlo model to predict D/MU values of pristine Bragg peaks delivered from a low-energy (<70 MeV initial beam energy) ocular nozzle. Their predictions agreed to within 5% with measurements for circular field diameters between 5 mm and 34 mm. More recently, Herault *et al* (2007) extended their technique to predict D/MU values to within 2% for spread-out Bragg peaks (SOBPs) for collimator diameters between 7.5 and 34 mm. While Herault *et al* have successfully applied the MCNPX code (Pelowitz 2005) in low-energy ocular proton therapy research, the code has not been shown to predict absolute doses from a high-energy (>70 MeV initial beam energy) ocular proton nozzle. Higher energy nozzles present more of a challenge to Monte Carlo codes because the beam must be degraded first to a therapeutically useful energy, which emphasizes the accuracy of a wider range of proton stopping powers. Further, the simulation models and results reported thus far for ocular

proton therapy used a simple box phantom. Similar Monte Carlo simulations have not yet been performed using realistic patient geometry.

The main objective of this study was to determine if Monte Carlo methods could predict D/MU values at the center of an SOBP to within 1% of measured values for a variety of treatment fields relevant to ocular proton therapy. To accomplish this objective, we extended an existing Monte Carlo model (Newhauser *et al* 2005) to calculate pristine Bragg peaks and therapeutically useful SOBPs in terms of absolute dose in three-dimensional geometry.

2. Materials and methods

2.1. Measurements

Measurements were performed to provide information for completing the Monte Carlo model. These measurements included dose profiles and absolute dose values for validating Monte Carlo predictions under several combinations of beam penetration ranges and range-modulation widths.

2.1.1. Relative depth–dose profiles—Measurements were made at the Northeast Proton Therapy Center (NPTC, Boston, MA, USA) using a fixed horizontal beam line dedicated to ocular treatments. Relative depth–dose profiles were acquired using a Markus-type parallel-plate ionization chamber (PPIC) (model 23343; PTW, Freiburg, Germany) in a water phantom (model 160–1; Computerized Radiation Scanners Inc., Vero Beach, FL, USA). The front face of the phantom was located 5.6 cm downstream of a 24 mm diameter circular collimating aperture. A separate thimble ionization chamber (model T1; Exradin Inc., Lisle, IL, USA) was mounted in air in the periphery of the proton field near the front face of the water tank. The PPIC and thimble chamber were read out in synchrony; the PPIC response was divided by the thimble chamber response to minimize the effect of temporal fluctuations in the dose rate. A step size of 1 mm was used to measure each depth–dose curve, with a dwell time of 2.5 s at each depth.

Eight depth–dose curves were measured, including four unmodulated beams and four SOBPs (table 1). For a description of generating an SOBP or modulated proton treatment beam from unmodulated beam, please see Kostjuchenko *et al* (2001). The measurement depth interval began at a water-equivalent depth of 9 mm and extended beyond the end of the proton beam range. Measurement proximal to water-equivalent depth of 9 mm was not possible with our measurement system due to the thickness of the water phantom's front wall and the Markus chamber's front window.

2.1.2. Relative crossfield profiles—Radiographic films (Kodak X-Omat V2 Ready Pak; Carestream Health, Inc., Rochester, NY, USA) were secured between several slabs of polymethyl methacrylate ($C_5H_8O_2$, $\rho = 1.19 \text{ g cm}^{-3}$; commonly known as Lucite; GE Plastics, Inc., Pittsfield, MA, USA) at water-equivalent depths of 8, 14, 20 and 26 mm and irradiated using the SOBPs listed in table 1. The optical transmittance of the processed film was scanned at a resolution of 0.036 cm^2 (model VXR-16 DosimetryPro; Vidar Systems Corp., Herndon, VA, USA) and converted to gray-scale values. We used standard methods to calibrate the film's response as a function of absorbed dose. Specifically, a series of films

was irradiated at various doses; the net gray-scale values (i.e., the gray-scale values after background was subtracted) were correlated with the corresponding absorbed-dose values measured in an ionization chamber. The effects of changes in proton linear energy transfer with depth on the calibration were not explicitly taken into account because this study did not seek to compare films at different depths on an absolute dose scale. One-dimensional relative dose profiles were extracted from the two-dimensional dose images captured on film. The penumbral widths were extracted from the one-dimensional dose profiles and compared to values from the Monte Carlo simulations.

2.1.3. Absolute D/MU measurements—Using the phantom setup described in section 2.1.1, *D/MU* values on the central axis in each of the beams (table 1) were measured in accordance to an established proton dosimetry protocol described by ICRU Report 59 (1998). An electrometer (model 6512; Keithley Instruments, Inc., Cleveland, OH, USA) integrated the charge collected from the PPIC. For the unmodulated beams, the *D/MU* values were measured at the 9 mm water-equivalent depth, where dose gradients were the smallest. For each unmodulated beam, 10 measurements were made at 1 MU each, which corresponded to a dose of approximately 10–20 cGy at the Bragg peak. The *D/MU* value at the maximum peak of the Bragg curve was determined by dividing the *D/MU* value measured at the 9 mm water-equivalent depth by its corresponding per cent depth–dose value.

Similarly, the *D/MU* values of the four modulated beams were measured at the center of the SOBP, per standard practice at NPTC (Newhauser *et al* 2002b). The water-equivalent depths of the eight *D/MU* measurements are also given in table 1. For each modulated beam, doses between 84 and 117 cGy were delivered to the PPIC at the calibration depth. The average of 10 repeated dose measurements was divided by the average number of measured monitor units delivered to obtain the *D/MU* value of each beam considered here. Corrections for temperature and pressure variations were applied to the averaged measured dose and MU values from each of the unmodulated and modulated beams.

2.2. Monte Carlo model and simulations

The ocular nozzle was modeled using the general-purpose MCNPX code and featured a highly realistic geometric representation of the beam delivery apparatus, including a range modulator wheel, a variable range degrader for adjusting the beam range, transmission monitor chambers, brass collimators and phantoms (figure 1). The physics modeled in the transport code included energy straggling, multiple Coulomb scattering, elastic and inelastic scattering, and non-elastic nuclear reactions. Default model parameters were used for proton transport options, as described elsewhere (Newhauser *et al* 2005). Relative to this earlier work, we made several refinements to the model that were important for the absolute dosimetry predictions. Specifically, improvements were made to the geometry of the monitor chamber and brass collimator located immediately downstream of the fixed degrader. These improvements included modeling the correct number and thickness of aluminum foils of the monitor chambers and the correct thickness of the collimator immediately following the degrader. Owing mainly to the change in monitor chamber design, the Gaussian width of the beam's initial energy distribution required a small

adjustment from the previously reported value (Newhauser *et al* 2005) so that the width of a simulated pristine beam in the beam direction matched its corresponding measured curve. The procedure to determine the energy width of the initial source distribution was described previously (Koch and Newhauser 2005). Briefly, the initial step in configuring the Monte Carlo model of the nozzle was to adjust the initial energy distribution of the proton beam that entered the nozzle in the Monte Carlo simulations. By varying the width of this Gaussian distribution of proton energies in the Monte Carlo model, we matched the width of a simulated pristine Bragg peak at the highest therapeutic range (i.e., 4 cm in water) to the corresponding measurement. Then the initial energy distribution was held constant for the more extensive benchmarking tests at lower energies and for the SOBPs listed in table 1.

The Monte Carlo model used the same range-modulation functions that were used for the measurements. Specifically, we used the opening angles and thicknesses of each step in the range modulator wheel and included these values in the Monte Carlo input file. Previously, we created an SOBP by summing the weighted results of simulations of individual steps of the range modulator wheel, as also performed by others (cf Herault *et al* 2005, Koch and Newhauser 2005). In this work, the range modulator wheel was modeled as a fixed slab of polymethyl methacrylate with constant thickness. The range modulation effect achieved by rotating the range modulator wheel (thereby adjusting the polymethyl methacrylate thickness) was accomplished by modulating the starting location of the proton beam along the beam axis within the fixed slab. At each starting location, the thickness of the absorber in the proton path corresponded to the physical thickness of the actual range modulator wheel. The relative number of protons starting at each axial position, which corresponded to the relative opening angle of each step of the actual modulator wheel, was implemented in the MCNPX simulation using a feature that allows the source properties to be sampled from a user-specified probability density function. The Monte Carlo simulations tracked 5×10^8 histories using mesh-based weight windows (a variance reduction technique), yielding statistical uncertainties of 0.2% at the 68% level of confidence in the tallies. We took advantage of the radial symmetry of our model by using cylindrical tallies centered on the central beam axis (axis of symmetry) to further improve the simulation efficiency. The use of variance reduction techniques (e.g., mesh-based weight windows) reduced the overall simulation time by approximately half to achieve the same level of statistical uncertainty in the tally results. Proton energy deposition was tracked in the most upstream monitor chamber and in a cylindrical tally volume residing in the target, e.g. water phantom or eye model. The combination of the monitor chamber tally with the central axis tally in water from the same simulation yielded the D/MU value according to the relationship described below.

The computation time required for the simulation of 5×10^8 histories was approximately 546 computer processing unit (CPU) hours on a single 3.0 GHz processor. For practical reasons, simulations were carried out on 10 dual-processor nodes (i.e., 20 CPUs), cutting the processing time by a factor of approximately 20.

2.3. Monte Carlo prediction of D/MU value

The D/MU value is defined as the absorbed dose measured along the central axis at a calibration depth in water (D) divided by the number of monitor units (MU) required to deliver this dose. In our Monte Carlo simulations, the D/MU value was obtained from the result of an energy deposition tally in a water phantom (E_w^{MC}) divided by the result from an energy deposition tally in the active volume of gas inside the monitor chamber (E_g^{MC}), or

$$\frac{D}{MU} = \frac{E_w^{MC}}{E_g^{MC}} \cdot F \quad (1)$$

where F is a constant of proportionality that relates the results of the above energy deposition tallies in the simulation to the measured D/MU value. Therefore, F implicitly includes the various constants of proportionality for converting simulated energy deposition into MU and measured charge from the dosimeter to proton-absorbed dose. The value of F was determined by minimizing the root mean square (*rms*) of the difference between the measured and simulated D/MU values for all beams listed in table 1. Although it is possible to model the charge in the monitor chamber with Monte Carlo simulations (Paganetti 2006), the simpler approach used here was sufficient for the objectives of this study.

2.4. Eye model

To demonstrate the clinical utility of a Monte Carlo dose algorithm in ocular proton therapy, we implemented a realistic and customizable model of the human eye and used it to calculate dose distributions and D/MU values. The eye model was primarily based upon an earlier model described by Dobler and Bendl (2002), but we added anatomic features, such as the optic nerve, optic disk and macula. Arbitrary gaze angles and translations were easily applied to the model using the matrix transformation card available in MCNPX, along with a representative tumor shape, which has been demonstrated in the previous work that simulated relative doses (Koch and Newhauser 2005, Mourtada *et al* 2005). Our model was defined with the center of the eye at the origin of the coordinate system, with the gaze originating in the +z direction (figure 2).

3. Results

3.1. Comparison of measured and Monte Carlo profiles

To demonstrate the accuracy of the Monte Carlo simulations' dose predictions, absorbed dose profiles from simulations and measurements are compared below.

3.1.1. Depth–dose profiles—As mentioned above, an unmodulated proton beam with a 4 cm range was used to deduce the initial energy distribution of the Monte Carlo model (figure 3). When we used a Gaussian energy distribution with a standard deviation of 1.12 MeV for the initial proton source model, the simulated and measured Bragg peaks matched excellently. The cause of the initial energy distribution is rooted in the design of the proton beam transport system as the proton bunches are created, accelerated and steered toward the treatment room. The mean proton energy of 159 MeV was taken from the literature

(Newhauser *et al* 2002b). Given these parameters, the simulated and measured pristine peaks agreed to within 0.6% at depths proximal to the distal 90% dose level and to within 0.5 mm in the distal fall-off region. Figure 3 shows the measured and calculated pristine Bragg peaks, where the calculated curve was normalized to a maximum of 1 and the measured curve was fit to the calculated curve using points above the 90% dose level. This normalization procedure was used for all comparisons of relative depth dose profiles. The Gaussian energy width deduced from this procedure agreed well with the previously measured width of 1.15 MeV (Cascio *et al* 2004).

Using the proton source parameters described above, we simulated the unmodulated and modulated beams listed in table 1. These calculated beams were compared to measurements, revealing agreement at a level similar to that for pristine peaks. For example, figure 4 shows the level of agreement for a modulated beam with a 30 mm range and a 15 mm SOBP width. The maximum difference between the simulated and measured relative depth-dose profiles, proximal to the distal 90% point, was 1.9%. At depths greater than the distal 90% point in the high-gradient region, the average and maximum per cent differences were –2.1% and –5.4%, respectively. The maximum per cent difference occurred at the distal 50% dose point (i.e., the location of the inflection point in the distal fall-off); however, the distance to agreement was still within 0.4 mm of the measured value.

3.1.2. Cross-field dose profiles—As a part of the extensive benchmarking procedures, the simulated and measured relative cross-field dose profiles were compared to assess the accuracy of modeling lateral scattering in the Monte Carlo simulations and to verify the geometric expansion of the simulated beam. The cross-field profiles from Monte Carlo simulation matched well with measured profiles, as shown in figure 5, and validated the scattering models in the Monte Carlo model. The lateral 80%-to-20% penumbral widths typically agreed within 0.1 mm. The simulated geometric magnification of the beam was verified by comparing with measurements of the 50%-to-50% cross-field widths at several depths, which also revealed agreement that was typically within 0.1 mm. The slight asymmetry of the measured profile was an artifact of beam steering, which was verified with a separate measurement of the beam spot location at the entrance of the nozzle.

3.2. Comparison of absolute D/MU values

The Monte Carlo method accurately predicted D/MU values for all of the unmodulated and modulated beams considered (table 1), and these values are listed in table 2 along with the corresponding per cent differences. The largest difference was less than 1%, indicating that the Monte Carlo simulations accurately predicted the D/MU values, and the *rms* of the per cent differences listed was only 0.56%.

Although this work considered only circular fields 24 mm in diameter, the MCNPX code has been used to predict the output of pristine peaks and SOBPs for various field sizes in an ocular nozzle (Herault *et al* 2005, 2007). Similarly, we expect that the methods described in this work could be successfully extended to predict D/MU values for other field sizes common to ocular proton therapy, although this extension of the methods will have to be carefully confirmed using additional measured data.

3.3. Absolute dose profiles

Figure 6 shows the central-axis depth–dose profile of the most energetic proton beam exiting the ocular nozzle in terms of absolute dose, instead of relative dose as figure 3 shows. The maximum dose difference between these profiles, from the surface to the depth of the distal 90% dose level, was 0.2 cGy MU^{-1} or 0.9% of the peak dose. Similarly, the maximum dose difference and its corresponding distance-to-agreement in the distal fall-off region were 0.6 cGy MU^{-1} and 0.2 mm, respectively. Accurate simulation of this pristine beam is of central importance because dosimetric differences tend to accumulate when the pristine beams are combined in the formation of an SOBP.

Representative measured and simulated SOBPs are shown in figure 7, also revealing similar good agreement. The maximum dose difference from the surface to the distal 90% dose level was -0.2 cGy MU^{-1} or -1.8% of the measured beam output at the depth of calibration. In the distal fall-off region, the maximum dose difference and its corresponding distance-to-agreement were -0.8 cGy MU^{-1} and 0.4 mm, respectively.

3.4. Monte Carlo predictions in an eye model

In addition to performing tests using a box-shaped water phantom, we also evaluated the impact of an irregularly shaped external surface (e.g., the curvature of the eye) on absolute dose predictions. This approach allowed us to demonstrate the predictive model under more clinically realistic conditions. As expected, the curved surface of the eye model was echoed in the two-dimensional plots of the isodose contours. Figure 8 shows two representative SOBPs incident upon the eye model shown in figure 2.

The D/MU values for each beam were calculated in the eye and are listed in table 2. Upon examination of these results, and correcting for inverse square effects, we observed a decrease in the D/MU values of the unmodulated beams of approximately 2% in the eye model compared to the flat-faced phantom results. The third column of table 2 shows the D/MU values for the unmodulated beam taken from table 2 and corrected for inverse square effects due to the proximal surface of the eye phantom being located 0.377 cm closer to the beam source compared to the water phantom. These results suggest a systematic difference in the D/MU value when calibrating the beam in a flat-faced phantom compared to patient-specific anatomy. This comparison was not made for the modulated beams since the effective source position of a modulated beam is less defined.

4. Discussion

We found that the Monte Carlo simulation results agreed well with measurements; the absorbed dose profiles agreed to within $\pm 3\%$ of the peak dose or within $\pm 0.5 \text{ mm}$ at every measured point in the depth–dose profiles. Further, a comparison of simulated and measured D/MU values at the depth of calibration revealed a maximum difference of -0.8% . The findings of this work are important because the predictions are the most accurate yet reported for ocular proton therapy and well within clinical requirements on accuracy.

Our findings are similar to those from Kooy *et al* (2003), who predicted D/MU values to within 3% using an analytical method, and Paganetti (2006), who used Monte Carlo

methods for 50 proton fields and predicted D/MU values that had a mean absolute deviation of 1.5% and maximum deviation >4%.

The results of the present study demonstrate the ability of the Monte Carlo transport code, MCNPX, to correctly simulate the physics of proton transport through a wider energy range than previously shown in ocular nozzles. More specifically, the Harvard ocular nozzle design (considered in this study) differs greatly from the nozzle simulated by Herault *et al.* The latter accepts a 65 MeV proton beam into a thin tantalum foil to create a uniformly flat beam, which is then immediately passed through a range modulator. The former accepts a 159 MeV proton beam that is range modulated before being degraded 70 MeV or less. Despite the dramatic differences in the energies and nozzle designs, the MCNPX predictions faithfully reproduced measured dose distributions in water phantoms.

Monte Carlo calculations are computationally expensive, which has inhibited adaptation of the technique to routine clinical use. However, access to computational resources continues to increase as it has within our laboratory. On a LINUX cluster of 512 dual-processor nodes presently available to our laboratory (i.e. 1024 CPUs), a simulation similar to those presented above would be complete after approximately 0.5 h. Therefore, it appears practical to use the Monte Carlo method as a routine dose algorithm with currently available computing technology.

With a single simulation, the three-dimensional relative dose distribution and the D/MU value for a patient-specific treatment plan can be obtained. This technique could simplify the clinical workflow and save valuable beam time. Given the accuracy and realism demonstrated in this report, as well as the accuracy achieved by others (Herault *et al* 2007), it is conceivable that the D/MU predictions could replace many routine measurements. However, additional development and verification will be necessary before such practice can be implemented.

5. Conclusion

The results of this study demonstrated that it is possible to predict D/MU values accurately for clinically relevant range-modulated proton beams for ocular therapy using the Monte Carlo method. The differences between the Monte Carlo predictions and measurements of D/MU values were less than 1%. The Monte Carlo method predicted the measured central-axis relative depth–dose profiles of SOBPs to within $\pm 3\%$, or ± 0.5 mm. We also demonstrated the feasibility of predicting absolute dose distributions in customizable patient geometry within practical calculation times. These findings, which reinforce those from previous studies, suggest it is feasible to use the Monte Carlo method as a routine absolute dose algorithm for ocular proton therapy.

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Figure 1.

Schematic drawing of the Monte Carlo model geometry representing the ocular nozzle used in this study. A 159 MeV proton beam enters the nozzle from the left. In the case of a range-modulated beam, the proton beam will start within the slab of polymethyl methacrylate located at (A) to simulate range modulation as described in the text. Next in the modeled nozzle is a thick slab of polycarbonate resin thermoplastic (B) that acts to adjust the beam range and laterally scatter the beam, producing a therapeutically useful proton field with a maximum energy of approximately 70 MeV. The monitor chambers (C) monitor the output of the nozzle as the beam exits through the final collimating aperture (D) and stops in the water phantom (E).

Figure 2.

Cross-sectional view in the xz -plane of the eye model used in Monte Carlo simulations. All dimensions of the eye model were customizable to patient-specific anatomy. Selected anatomic features of the eye, which may also be transformed to arbitrary gaze angles to represent the eye's treatment position, are labeled. Since no tumor was the target of the proton beams considered in this work, none appears here.

Figure 3.

The relative dose (D) as a function of depth (d) in water from measurements with an ionization chamber (open circles) and the Monte Carlo simulation (solid line). The figure shows the Bragg curve for the most penetrating beam available for the ocular nozzle.

Figure 4.

The relative dose (D) as a function of depth in water (d) for the modulated beam, with a nominal range of 3 cm and a modulation width of 1.5 cm, as measured with an ionization chamber (open circles) and predicted by the Monte Carlo simulation (solid line). The two curves have been normalized to their interpolated values at the depth of 2.3 cm, which was the calibration depth for measurement and simulation.

Figure 5. Cross-field profile from measurements on film (open circles) and Monte Carlo simulations (solid line). Error bars on the film data points indicate the standard deviation from three repeat measurements of the same beam. For clarity of presentation, the measured points between $-1 < x < 1$ are plotted at half-resolution.

Figure 6. Absolute depth-dose profile in water from ionization chamber measurement (open circles) and Monte Carlo simulation (solid line). This profile shows the most penetrating beam available from the ocular nozzle, with a depth at the distal 90% dose level of approximately 4 cm in water.

Figure 7.

Absolute depth-dose profile in water from ionization chamber measurement (open circles) and Monte Carlo simulation (solid line). This profile shows an SOBP with a depth to the distal 90% dose level equal to 2.9 cm and modulation width between the proximal and distal 90% dose levels equal to 1.4 cm.

Figure 8.

Dose distribution of an SOBPs incident upon a model of the human eye from Monte Carlo simulation in the treatment machine's coordinate system. Absolute values of the isodose contours are indicated. The two SOBPs are shown (a) a half-modulated beam that has a range of 20 mm and a modulation width of 10 mm and (b) a fully-modulated beam that has a range of 30 mm, i.e., the SOBPs encompasses the entire penetration range of the beam.

Table 1

The D/MU values of eight beams were measured and calculated. Each combination of penetration range and modulation width in water was given a unique label so the combinations could be quickly referenced later in the text and other tables. The far right column lists the water-equivalent depths of the D/MU calibration measurements, which matched the depths in water at which the D/MU values were calculated by Monte Carlo simulations.

Beam label	Nominal range (mm)	Nominal modulation width (mm)	Calibration depth (mm)
1500	15	–	9.0
2000	20	–	9.0
3000	30	–	9.0
4000	40	–	9.0
2011	20	11	15.0
2020	20	20	9.3
3015	30	15	23.0
3030	30	30	20.0

Table 2

The D/MU values from measurements with an ionization chamber (IC) and Monte Carlo simulations (MC) are shown in columns two and three. The column four lists the per cent differences between the corresponding values from each of the eight beams considered in this work. The per cent differences for all beams were within $\pm 1\%$. Columns five and six show the D/MU values from Monte Carlo simulations in the eye model (Eye) listed with the corresponding corrected D/MU values in from the flat-faced water phantom (WP), respectively. The column on the far right lists the per cent difference between the 'Eye' and 'WP' columns.

Beam label	IC (cGy MU ⁻¹)	MC (cGy MU ⁻¹)	(MC-IC)/MC (%)	Eye (cGy MU ⁻¹)	WP (cGy MU ⁻¹)	(Eye-WP)/Eye(%)
1500	14.5	14.6	0.4%	14.4	14.7	-2.1%
2000	16.4	16.5	0.5%	16.3	16.6	-1.8%
3000	19.7	19.8	0.5%	19.6	20.0	-2.0%
4000	22.7	22.8	0.4%	22.4	22.9	-2.2%
2011	11.7	11.6	-0.8%	11.5	–	–
2020	8.4	8.4	0.2%	8.35	–	–
3015	13.1	13.0	-0.7%	12.9	–	–
3030	9.9	9.9	-0.6%	9.75	–	–