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Dosimetric accuracy of planning and delivering small proton therapy fields

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Abstract

A detailed Monte Carlo model of a proton therapy treatment head was commissioned to simulate the delivery of small field proton treatments. Small fields are challenging in the planning and quality assurance process because of aperture scattering and dosimetric disequilibrium. Four patients with small fields used in all or parts of their treatment course were studied, including two stereotactic patients and two fractionated patients. For the two stereotactic patients the overall difference of the dose covering at least 95% of the gross tumor volume between the Monte Carlo calculations and the delivery was -0.2% and -1.6% , respectively. For the two fractionated patients the overall difference of the dose covering at least 95% of the clinical target volume was -3.0% and 1.0% , respectively. We have thus confirmed that our current planning and delivery process for small proton fields is accurate enough to treat small lesions in the patient. Furthermore, we studied the impact of field size corrections and identified limitations of the pencil beam algorithm for predicting hot and cold spots and range degradation in the target due to scattering in heterogeneities. For the four cases studied in this paper, these limitations appear to impact individual field calculations, but do not have a significant impact on the prescribed dose over multiple fields.

1. Introduction

The physical attributes of a proton beam make it one of the most attractive modalities for the treatment of cancer. The ability to conform radiation dose to the target through the precise delivery of the proton Bragg peak has resulted in improved local control and reduced local toxicity for a variety of treatment sites. Despite these advantages, sufficient care must be given during treatment planning to ensure that the dose deposition from the 'sharp-edged' proton beam is being deposited in the correct location. While modern technologies permit safe and accurate delivery of most proton treatments, uncertainty still remains regarding the accuracy of delivering small fields for spread-out Bragg peak (SOBP) proton therapy (Hong *et al* 1996, Titt *et al* 2008, Fontenot *et al* 2007, Daartz *et al* 2009). In the context of this paper, a small proton field is defined as a field in which the aperture is so small that there is little or no uniform dose region (Gottschalk 1999).

The dosimetric limitations of small proton fields have been broadly evaluated. Interestingly, some of the first treatments using protons were for small radiosurgery target volumes. As early as 1967, Larsson (1967) provided a brief explanation on the degradation of the Bragg peak in small proton fields. The author documented that 'when the diameter of the beam was reduced below 3 cm, the relative height of the Bragg peak decreased progressively' (Larsson 1967). The following year, Preston and Koehler (1968, unpublished) discussed the

dosimetric properties of small proton fields more extensively and provided theoretical approximations of the lateral and distal contributions of small fields with some experimental validation. The vanishing Bragg peak is a consequence of the loss of transverse equilibrium along the central axis of the beam (namely more protons are scattering out of the beam than in). As a result, the decrease in fluence begins to compensate the increase in proton stopping power and the Bragg peak begins to deteriorate. A second consequence of small fields is that the transverse profile of the beam becomes entirely penumbra. Depending on the depth of the target, the penumbra can also be significantly widened due to scattering in the patient (Gottschalk 1999). Finally, an equally palpable consequence of small fields is the influence of slit scattering or aperture scattering on the central axis dose. Gottschalk (1999) used measurements and Monte Carlo data to demonstrate the dosimetric contributions of slit scattering. These include an increase in entrance dose and the emergence of 'horns' or 'spikes' specifically at the field edge on shallow dose profiles due to small angle scattering originating at aperture (Gottschalk 1999, Titt *et al* 2008).

The dosimetric properties associated with small proton fields are a clinical challenge in terms of planning and delivery, i.e. planning systems need to be able to predict the correct dose distributions and clinical dosimetry has to predict the correct output factor. It suffices to say that attempts have been made to develop accurate dose algorithms that can accommodate small proton fields. In the past, broad beam dose algorithms were used in routine clinical practice. However, it was pointed out by Hong *et al* (1996) that broad beam dose algorithms do not fully account for dose perturbations due to heterogeneous regions in the patient. In addition, such broad beam models do not predict the loss of transverse equilibrium along the beam central axis associated with small proton beams. The authors presented a new pencil beam formalism that corrected for these deficiencies and improved the dosimetric accuracy of treatment plans using small proton fields. However, residual limitations in the new algorithm were acknowledged, including the lack of aperture scattering contributions and errors in the shadow of thick heterogeneities whose edge is parallel to the beam central axis, both of which have a greater impact on small fields compared to large fields. Currently, all treatment plans developed at our facility are based on an in-house implementation of a pencil beam algorithm that uses the physics model developed by Hong *et al* (1996).

At our facility an analytical model is used to model the output factors (i.e. cGy MU⁻¹) for proton treatments (Kooy *et al* 2003, Kooy *et al* 2005). This model uses standard measurements of a large set of ranges and modulations to predict the output factor, eliminating the need to measure the output factor for each treatment field (Kooy *et al* 2005, Engelsman *et al* 2009). However, it was made clear by the authors that such models have limitations for small field sizes (Kooy *et al* 2005, Engelsman *et al* 2009). Daartz *et al* (2009) compared measured and predicted outputs for a variety of field sizes and revealed large discrepancies for fields less than 5 cm in diameter. As a result, field-specific correction factors were incorporated into our output prediction (Daartz *et al* 2009). However, these correction factors are derived from measurements in a water phantom thereby ignoring scattering effects in patients resulting from tissue heterogeneities.

In this paper we set out to investigate the accuracy of our treatment planning process for proton treatments that use small fields. To do this we used a well-benchmarked Monte Carlo framework that is able to simulate proton therapy treatments in patients (Paganetti *et al* 2008) and systematically compared dose distributions predicted by our planning system and Monte Carlo simulations for a variety of treatments that use small fields. Prior to this study, the impact of small proton fields has been studied using measurements or Monte Carlo simulations in simple phantom geometry (Fontenot *et al* 2007, Titt *et al* 2008, Daartz *et al*

2009). To our knowledge, this is the first time that the dosimetric properties of small proton fields have been assessed in patients using Monte Carlo simulations.

2. Methods

2.1. Monte Carlo model of the treatment head

All simulations were performed using a detailed model of the Francis H. Burr proton treatment head that was developed by Paganetti *et al* (2004) and implemented in Geant4 version 9.0 (Agnostinelli *et al* 2003). All relevant components were modeled based on blueprints provided by the system manufacturer (Ion Beam Applications SA (IBA), Louvain-la-Neuve, Belgium). The physics settings were carefully selected based on work described elsewhere (Zacharatou Jarlskog and Paganetti 2008, Paganetti *et al* 2004). A diagram of the treatment head model is provided in figure 1(a). For stereotactic treatments that utilize very small proton fields an additional piece of hardware, known at our facility as the 'HiHat', abuts to the distal face of a standard aperture and is used to reduce the air gap between the aperture and the patient. A HiHat is an assembly of a brass aperture and a compensator, which are both smaller than standard hardware. A diagram of the HiHat is provided in figure 1(b). Accordingly, the HiHat was also modeled for simulations involving stereotactic treatments.

2.2. Dose to patients for small proton fields

Dose in the patient was simulated in the Monte Carlo system as described elsewhere (Paganetti *et al* 2008, Jiang and Paganetti 2004). The calculation is based on the actual CT grid. The CT data are imported from the planning system along with the necessary beam (e.g. range, modulation width, aperture, compensator, treatment head settings, etc) and patient (isocenter position, voxel size, couch rotation, etc) parameters. The CT numbers are converted to material compositions and mass densities for each tissue using the conversion scheme proposed by Schneider *et al* (2000). Normalization to the relationship between the CT number and the relative stopping power as used in the planning system was performed (Paganetti *et al* 2008). The dose calculation is retrospectively reconstructed on the planning grid once the simulation is completed for comparison with the planning system. Likewise, we tailored our Monte Carlo dose calculations to report dose to water using the methodology described by Paganetti (2009). We also compared dose to water versus dose to medium for small fields because the field size may have a secondary impact on the dose to water versus dose to tissue conversion, since the proton energy distributions is altered within the target when the field size is reduced. Due to the field size considered, a total of 1×10^8 source protons were simulated for each field to ensure reasonable statistical accuracy. For the field sizes considered the treatment head efficiency is only about 3%.

2.3. Validation of output factor simulations

To validate the output factors predicted by the Monte Carlo treatment head model, simulations were performed using standard circular fields and selected field conditions for which values were previously measured (Daartz *et al* 2009). Output factors from three different ranges with modulation widths of 2 cm were simulated, including ranges of 6.7, 10.2 and 17.7 cm. In total seven circular fields were considered for each range with the following diameters: 1.2, 2, 3, 4, 5, 7, 9 and 12 cm. The 1.2 cm diameter aperture is used in combination with the HiHat hardware. The scoring region in the simulations was set to a radius of 0.5 cm to ensure that charge particle equilibrium was established within the scoring volumes.

2.4. Patient population and treatment planning

We selected four patients with small lesions that were treated with small proton fields and are representative for this patient population. Two of the patients (patients 1 and 2) were stereotactic patients who had common intracranial lesions treated with small field sizes. Patient 1 was treated for a benign neoplasm in the pituitary gland. Patient 2 was treated for a solitary brain metastasis. The remaining two patients received fractionated radiotherapy. Patient 3 was treated for cancer in the clivus and patient 4 was treated for ependymoma. In the case of patient 3, all of the ten fields qualified as small fields. Patient 4 had a rather large contoured CTV, but the treatment also included a small field boost to a residual section of the GTV. Details of the field characteristics are provided in table 1. Note that some of the patients received part of their treatment with photons, but this additional dose was not considered in the analysis.

All of the treatment plans were created using the treatment planning system XiO (XiO, Computerized Medical Systems Inc.) in combination with an in-house pencil beam algorithm. Typically, small fields full coverage cannot be achieved in the target volume during the initial planning process. Comparison of measured and calculated relative output as a function of field size showed overestimation of calculated dose for small fields (Daartz *et al* 2009). As a result, any computed treatment beam is renormalized in the treatment planning system to undo the effect of charge particle disequilibrium on the dose at the point of interest ensuring full coverage in the target volume. In general, this renormalization is done using an isodose line circumscribing the target volume. This normalization is accounted for during the analytical calculation of the number of monitor units (MUs) for our stereotactic treatments. A field size correction factor is applied during the MU calculation for stereotactic patients. The field size effect on MU is typically not considered for our fractionated patients.

3. Results

3.1. Comparison of measured and calculated output factors for reference fields

Figures 2(a)–(c) provide a comparison of calculated and measured output factors using several beam diameters and beam ranges of 6.7, 10.6 and 17.7 cm. The modulation width of each field was set to 2 cm. For the 6.7 cm range the largest discrepancy between the calculated and measured data was 2.1% using the smallest field diameter of 1.2 cm. For the 10.6 and 17.7 cm ranges the largest discrepancy was 1.6% and 1.2%, respectively using the 2 cm field diameter. These differences are well within our clinically defined tolerance of 3% (Engelsman *et al* 2009). The results demonstrate the accuracy of the Monte Carlo model and serve as validation for the patient related studies.

3.2. Comparison of Monte Carlo and treatment planning dose calculations

Figures 3–6 show dose-volume histograms (DVHs) for patients 1–4. Patients 1 and 2 were single fraction stereotactic patients. Figure 3 presents the DVHs for patient 1 and demonstrates that the treatment planning system only slightly overestimates the dose to the gross tumor volume (GTV) compared to the Monte Carlo. For the entire treatment, the dose covering at least 95% of the GTV volume was 0.2% higher for the treatment planning system ($D_{\text{TPS},95}$) compared to Monte Carlo ($D_{\text{MC},95}$). In addition, dose to organs-at-risk (OARs) was slightly overestimated by the pencil beam algorithm with the optic chiasm having the largest discrepancy where the mean dose difference between the Monte Carlo and the treatment planning was 2.1%. Figure 4 presents the DVHs for the stereotactic patient with a brain metastasis. Again, the treatment planning system overestimated the dose to the GTV where $D_{\text{TPS},95}$ was 1.6% higher than $D_{\text{MC},95}$. The target volume was located far away from critical structures; hence no OAR doses need to be considered.

Patients 3 and 4 were fractionated patients. Figure 5 presents the DVHs for patient 3 with cancer within the clivus. In this case the target volume contoured by the physician was rather small (i.e. 3 cm³), resulting in ten small fields planned to cover the clinical target volume (CTV) and the GTV boosts. As seen in the figure, the treatment planning system overestimated the dose to the CTV compared to Monte Carlo calculation. Considerable discrepancies were also seen in the optic chiasm. $D_{\text{TPS},95}$ to the CTV for this patient was 3.0% higher compared to $D_{\text{MC},95}$. Finally, figure 6 presents the DVHs for patient 4, which used a small residual GTV boost (i.e. field 7) as part of the treatment for an ependymoma. The overall differences between $D_{\text{TPS},95}$ and $D_{\text{MC},95}$ were small, where $D_{\text{TPS},95}$ was 1% larger than $D_{\text{MC},95}$ for the field sizes considered. For this patient larger discrepancies were seen for the small GTV boost field, where $D_{\text{TPS},95}$ was 3% larger than $D_{\text{MC},95}$. A detailed comparison of the relevant dose values for each field calculated using the treatment planning system and Monte Carlo is presented in table 2. Overall, it seems that the treatment planning system systematically overestimates the dose to the GTV compared to Monte Carlo for most of the fields. For individual fields, maximum differences up to 8.6% were seen between $D_{\text{TPS},95}$ and $D_{\text{MC},95}$. However, the differences between $D_{\text{TPS},95}$ and $D_{\text{MC},95}$ for full stereotactic or fractionated treatments were all below 3.0%. Treatment plans for the stereotactic fields were closer to the Monte Carlo-predicted dose distributions. This can be attributed to the use of a field size correction factor used in the output model for all small stereotactic fields (Daartz *et al* 2009).

Table 3 shows the dose differences between the dose to water (D_w) and dose to medium (D_m) at different dose levels in the GTV for each treatment plan considered in this study. For all but one treatment, the difference between $D_{w,95}$ and $D_{m,95}$ was less than 1%. Patient 3 was the only exception where the difference between $D_{w,95}$ and $D_{m,95}$ was 3.4%. The GTV in patient 3 surrounds the clivus, which is a bony surface in the posterior cranial fossa. Large differences (~10% or higher) between D_w and D_m in bony anatomy have been previously reported (Paganetti 2009). In soft tissue the differences are much less, between 0.1 and 3% (Paganetti 2009). The greatest differences between D_w and D_m are expected at the end of the proton range, i.e. low proton energies (Paganetti 2009). This is the case for patient 3 where bony anatomy is within the GTV.

4. Discussion

We have demonstrated that a treatment planning system using a pencil beam algorithm generally overestimates the dose delivered to the target volume for small fields compared to Monte Carlo calculations. The loss of fluence along the central axis associated with small proton fields becomes difficult to reproduce analytically in complex heterogeneous volumes. Despite being better suited to predict dose distributions of small proton fields compared to older broad beam algorithms, the accuracy of the pencil beam algorithm remains in question when thick heterogeneities are present within the proton field (Petti 1996, Hong *et al* 1996). These algorithms typically have difficulties predicting the correct location of dose shadows produced by upstream heterogeneities resulting in the misrepresentation of hot and cold spots within the target volume (Petti 1996). Of course, this situation becomes more detrimental as the target volume is reduced and shadow effects have a greater impact on the overall dose coverage in the target.

To demonstrate this effect figures 7(a) and (b) present a comparison of dose contours from our Monte Carlo and the pencil beam calculations for patient 1, field 2. The dose was normalized to the dose of 90% of the GTV volume. This normalization was done to provide a clear demonstration of the dose shadow effects from heterogeneities located upstream of the tumor. The statistically significant dose shadows not predicted by the pencil beam algorithm stand out in the Monte Carlo dose contours. In this case, the dose shadows are due

to the bony structures located upstream of the GTV. One result of this limitation is a missing cold spot in the planned GTV, as shown in the figures, resulting in a less uniform tumor coverage than originally predicted. Figures 8(a) and (b) compare depth dose distributions and dose profiles through the GTV calculated with Monte Carlo and the pencil beam algorithm along the designated scoring axes shown in figure 8(c). As seen in the figures, the pencil beam algorithm is able to predict the deterioration of the high dose plateau region of the SOBP. However, the uniformity predicted within the GTV is noticeably unrealistic where the variation between the Monte Carlo and the pencil beam algorithm is almost 5%. Note that the statistical fluctuation in the Monte Carlo calculation was less than 1% for all data points. Furthermore, the degradation of the distal gradient of the Bragg peak due to heterogeneities is illustrated in figure 8(a). The pencil beam algorithm in contrast does not account for this range degradation.

Ultimately, the limitations of the pencil beam algorithm may have the most significant impact on the MU calculation. Again, to ensure 100% dose coverage to the target during planning an isodose line from the original plan that circumscribes the target is renormalized to 100%. This normalization is accounted for in the calculation of the MUs by a normalization factor. Consequently, the MU calculation is dependent on the isodose line used for normalization. As clearly shown in figures 7(a) and (b), the isodose lines predicted by the pencil beam algorithm do not account for residual effects of heterogeneities upstream of the target. The hot and cold spots in the target volume, and the range degradation could lead to an under-prediction or over-prediction of the MUs needed to deliver the prescribed dose. It is important to note that the physical limitations of the pencil beam algorithm appear to cancel out with multiple fields.

Another known limitation of the pencil beam algorithm is the inability to account for aperture scattering (Hong *et al* 1996). This limitation is demonstrated in the depth dose distributions provided in figure 8(a). The heightened dose near the entrance region shown in the Monte Carlo distribution is due to scattering from the brass aperture located directly above the patient. For this stereotactic patient the HiHat apparatus was abutted to the end of the snout, which almost completely eliminated the air gap between the treatment head and the patient. Scattering from the small aperture together with the reduced air gap between the aperture and the patient increases the dose in the entrance region. Therefore, the dose to the skin and other superficial organs may be slightly higher than what is predicted by the pencil beam algorithm. Note, that our pencil beam model includes scattering effects from the range compensator. Nonetheless, the influence of compensator scattering on MU calculations is negligible (Fontenot *et al* 2007).

Finally, this comparison also indicated that larger discrepancies between the Monte Carlo and the pencil beam algorithm occur in fractionated patients compared to stereotactic patients. For fractionated patients a field size correction factor is not applied during the MU calculation. For most fractionated patients a correction may not be necessary since a majority of the fields are not considered small, as is the case for patient 4. However, for some patients several fields are small and neglecting the field size effect on the output factor may lead to under-dosage in the target. This was demonstrated in patient 3 where all ten of the treatment fields were small enough to impact the output factor. Table 4 compares the doses to the GTV of patient 3 with and without a field size correction applied to the MU. The differences between the treatment planning system and the Monte Carlo calculation for the entire plan are reduced from 3% to 0.9%. The residual difference for the corrected treatment plan also improved for each individual field. However, similar to the stereotactic fields, differences between the planning system and Monte Carlo larger than 3% still arise due to scattering and the choice of the normalization contour.

5. Conclusion

An investigation of the dosimetric uncertainties associated with small proton fields in patients using Monte Carlo has hitherto not been done. To this end, we tested the accuracy of both stereotactic and fractionated small field proton treatments using four patients treated at our facility. Based on the results, we are confident that our current planning and delivery process for small proton fields provides the level of accuracy needed to treat small lesions in the patient; the overall dose differences between the treatment planning system and Monte Carlo calculations was at most 3%. The largest differences were seen for non-stereotactic patients suggesting that field size corrections might be warranted. We also presented dosimetric limitations of the pencil beam algorithm for predicting hot and cold spots and range degradation in the target due to scattering in heterogeneities. These deficiencies may impact the accuracy of single proton field calculations; however the residual consequence of such shortcomings over multiple fields appears insignificant.

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References

- Agnostinelli S, et al. GEANT4—a simulation toolkit. *Nucl. Instrum. Methods Phys. Res. A*. 2003; 506:250–303.
- Daartz J, Engelsman M, Paganetti H, Bussire M. Field size dependence of the output factor in passively scattered proton therapy: influence of range, modulation, air gap, and machine settings. *Med. Phys.* 2009; 36:3205–10. [PubMed: 19673219]
- Engelsman M, Lu H-M, Herrup D, Bussire M, Kooy HM. Commissioning a passive scattering proton therapy nozzle for accurate SOBP delivery. *Med. Phys.* 2009; 36:2172–80. [PubMed: 19610306]
- Fontenot JD, Newhauser WD, Bloch C, White RA, Titt U, Starkschall G. Determination of output factors for small proton therapy fields. *Med. Phys.* 2007; 34:489–98. [PubMed: 17388166]
- Gottschalk, B. Physical parameters of small proton beams. *PTCOG XXXI Conf. Proc.*; Bloomington, IN, USA. 1999.
- Hong L, et al. A pencil beam algorithm for proton dose calculations. *Phys. Med. Biol.* 1996; 41:1305–30. [PubMed: 8858722]
- Jiang H, Paganetti H. Adaptation of GEANT4 Monte Carlo dose calculations based on CT data. *Med. Phys.* 2004; 31:2811–8. [PubMed: 15543788]
- Kooy HM, et al. The prediction of output factors for spread-out proton Bragg peak fields in clinical practice. *Phys. Med. Biol.* 2005; 50:5847–56. [PubMed: 16333159]
- Kooy HM, Schaefer M, Rosenthal S, Bortfeld T. Monitor unit calculations for range-modulated spread-out Bragg peak fields. *Phys. Med. Biol.* 2003; 48:2797–808. [PubMed: 14516102]
- Larsson B. Radiological properties of beams of high-energy proton. *Radiat. Res. Suppl.* 1967; 7:304–11. [PubMed: 4964576]
- Paganetti H. Dose to water versus dose to medium in proton therapy. *Phys. Med. Biol.* 2009; 54:4399–421. [PubMed: 19550004]
- Paganetti H, Jiang H, Lee S-Y, Kooy H. Accurate Monte Carlo for nozzle design, commissioning, and quality assurance in proton therapy. *Med. Phys.* 2004; 31:2107–18. [PubMed: 15305464]
- Paganetti H, Jiang H, Parodi K, Slopsema R, Engelsman M. Clinical implementation of full Monte Carlo dose calculation in proton beam therapy. *Phys. Med. Biol.* 2008; 53:4825–53. [PubMed: 18701772]

- Petti P. Evaluation of a pencil-beam dose calculation technique for charged particle radiotherapy. *Int. J. Radiat. Oncol. Biol. Phys.* 1996; 35:1049–57. [PubMed: 8751415]
- Preston WM, Koehler AM. The effects of scattering on small proton beams. Harvard Internal Report. 1968 unpublished.
- Schneider W, Bortfeld T, Schlegel W. Correlation between CT numbers and tissue parameters needed for Monte Carlo simulations of clinical dose distributions. *Phys. Med. Biol.* 2000; 45:459–78. [PubMed: 10701515]
- Titt U, Zheng Y, Vassiliev ON, Newhauser WD. Monte Carlo investigation of collimator scatter of proton-therapy beams produced using the passive scattering method. *Phys. Med. Biol.* 2008; 53:487–504. [PubMed: 18185001]
- Zacharatou Jarlskog C, Paganetti H. Physics settings for using the Geant4 toolkit in proton therapy. *IEEE Trans. Nucl. Sci.* 2008; 55:1018–25.

Figure 1.

(a) Wired-frame rendering of the entire treatment head modeled in Geant4. (b) A zoomed-in rendering of the HiHat arrangement abutted to the snout. The green lines outline the miniature aperture with the aperture opening shown in white and the red lines outline the miniature compensator.

Figure 2.

Comparison of measured and calculated output factors as a function of the beam diameter for ranges of (a) 6.7 cm, (b) 10.6 cm and (c) 17.7 cm. All output factors are normalized to the value for a 12 cm diameter aperture. The relative error in the Monte Carlo data is less than 1%.

Figure 3.
Comparison of DVHs calculated with the treatment planning system and Monte Carlo for patient 1. The DVHs considered were for the GTV and several organs-at-risk.

Figure 4. Comparison of DVHs calculated with the treatment planning system and Monte Carlo for patient 2. Due to the location of the lesion no OAR needed to be considered.

Figure 5. Comparison of DVHs calculated with the treatment planning system and Monte Carlo for patient 3. The DVHs considered were for the GTV, CTV and several organs-at-risk.

Figure 6.

Comparison of DVHs calculated with the treatment planning system and Monte Carlo for patient 4. The DVHs considered were for the GTV, CTV and several organs-at-risk.

Figure 7.

A comparison of different dose contours from the (a) Monte Carlo and (b) pencil beam calculations for patient 1, field 2. The dose was normalized to the dose to 90% of GTV volume. The beam is directed left to right.

Figure 8.

Comparison of (a) depth dose distributions along the beam central axis and (b) transverse dose profiles at a depth equal to the center of the target volume. These figures illustrate the non-uniformity in the GTV predicted by Monte Carlo but not by the treatment planning system. The scoring axes for these distributions are shown as dotted white lines in (c). The statistical fluctuation in the Monte Carlo calculation was less than 1% for all data points. In order to improve clarity, magnified insets of the depth dose distribution and dose profile are provided.

Table 1

Field characteristics for the four patients considered.

Patient	Field	Site	Range (cm)	Modulation width (cm)	Target dose (Gy)	Target volume (cm ³)
1	1-1	Pituitary gland, craniopharyngeal duct	15.6	1.5	5.63	0.90 (GTV)
	1-2		11.4	2.7	5.54	
	1-3		11.9	2.8	5.43	
2	2-1	Brain, lt. temporal lobe	7.1	1.5	7.40	0.46 (GTV)
	2-2		7.1	1.5	7.39	
	2-3		11.3	1.5	7.48	
3	3-1	Clivus	15.2	4.0	21.9	3.00 (GTV)
	3-2		10.6	4.1	8.00	
	3-3		11.3	4.0	6.00	
	3-4		14.5	3.3	12.0	
	3-5		10.8	4.3	6.00	
	3-6		11.5	4.2	8.00	
	3-7 (boost)		9.7	3.2	3.06	
	3-8 (boost)		10.5	3.2	3.07	
	3-9 (boost)		9.7	3.2	4.08	
	3-10 (boost)		10.5	3.2	4.09	
4	4-1	Ependymoma	11.2	7.9	16.0	107 (GTV)
	4-2		12.6	7.5	16.7	
	4-3		12.6	7.8	16.8	
	4-4		11.2	7.9	1.79	
	4-5		12.2	7.8	1.80	
	4-6 (boost)		9.3	4.0	1.79	2.6 (GTV residual)

Table 2

Differences between the dose calculated with Monte Carlo and the treatment planning system, $(D_{MC} - D_{TPS})/D_{MC} \times 100\%$, for different volume levels (95%, 50%, 10%) and the mean dose in the GTV. All values are given in percent.

Patient/field/volume	Volume level (%)			Mean dose
	95	50	10	
1/1/GTV	2.1	1.1	-4.5	1.2
1/2/GTV	-3.9	-3.8	-2.8	-2.6
1/3/GTV	-2.4	-2.0	-1.0	-1.1
1/TOT	-0.2	-0.8	-0.1	-0.3
2/1/GTV	-5.3	-4.1	-2.4	-3.0
2/2/GTV	-1.4	-0.1	-1.8	-0.5
2/3/GTV	3.9	3.8	4.4	3.3
2/TOT	-1.6	-1.3	-3.4	-1.4
3/1/CTV	-3.8	-2.5	-1.8	-1.9
3/2/CTV	-4.5	-2.1	-0.2	-1.5
3/3/CTV	2.6	3.7	5.0	4.6
3/4/CTV	-2.6	-3.0	-1.8	-2.0
3/5/CTV	-5.5	-2.6	-0.5	-1.8
3/6/CTV	-4.2	-2.7	-0.6	-1.6
3/7/CTV	-6.3	-3.8	-1.7	-2.0
3/8/CTV	-6.2	-2.5	-0.7	-1.0
3/9/CTV	-6.3	-5.4	-2.5	-3.4
3/10/CTV	-8.6	-3.4	-1.0	-1.5
3/TOT	-3.0	-2.9	-2.4	-2.0
4/1/CTV	-2.3	0.2	1.0	0.1
4/2/CTV	-1.4	1.0	1.5	0.6
4/3/CTV	-1.2	0.7	1.7	0.6
4/4/CTV	-	-4.0	-2.8	-3.6
4/5/CTV	-	-3.4	-1.7	-3.4
4/6/GTV	-3.1	-5.2	-4.5	-3.2
4/TOT/CTV	-1.0	0.5	0.9	0.2

Table 3

Difference between D_w and D_m , $(D_w - D_m/D_w)$, in the GTV for different volume levels (95%, 50%, 10%) and the mean dose.

Patient	Volume level (%)			Mean dose
	95	50	10	
1	-0.2	-0.3	-1.0	-0.4
2	-0.6	-0.7	-0.6	-0.7
3	3.4	3.5	3.1	2.7
4	-0.6	-1.0	-0.7	-0.5

Table 4

Differences between D_{95} of the CTV in patient 3 calculated with Monte Carlo and the treatment planning system, $(D_{MC,95} - D_{TPS,95}/D_{MC,95})$, when the MU is both corrected and not corrected for the field size effect.

Field/volume	$(D_{MC} - D_{TPS}/D_{MC})$ (%)	
	Uncorrected	Corrected
1/CTV	-3.8	-0.4
2/CTV	-4.5	-3.2
3/CTV	2.6	3.7
4/CTV	-2.6	-0.1
5/CTV	-5.5	-4.1
6/CTV	-4.2	-2.9
7/CTV	-6.3	-3.7
8/CTV	-6.2	-6.2
9/CTV	-6.3	-2.4
10/CTV	-8.6	-3.5
TOT	-3.0	-0.9