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REDUCING STRAY RADIATION DOSE FOR A PEDIATRIC PATIENT RECEIVING PROTON CRANIOSPINAL IRRADIATION

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Abstract

The aim of this study was to quantify stray radiation dose from neutrons emanating from a proton treatment unit and to evaluate methods of reducing this dose for a pediatric patient undergoing craniospinal irradiation. The organ equivalent doses and effective dose from stray radiation were estimated for a 30.6-Gy treatment using Monte Carlo simulations of a passive scattering treatment unit and a patient-specific voxelized anatomy. The treatment plan was based on computed tomography images of a 10-yr-old male patient. The contribution to stray radiation was evaluated for the standard nozzle and for the same nozzle but with modest modifications to suppress stray radiation. The modifications included enhancing the local shielding between the patient and the primary external neutron source and increasing the distance between them. The effective dose from stray radiation emanating from the standard nozzle was 322 mSv; enhancements to the nozzle reduced the effective dose by as much as 43%. These results add to the body of evidence that modest enhancements to the treatment unit can reduce substantially the effective dose from stray radiation.

Keywords

shielding; proton craniospinal irradiation; stray radiation

I. INTRODUCTION

Proton therapy is an effective treatment modality for central nervous system tumors in pediatric patients. Proton beams have a finite range, which virtually eliminates the exit dose. This gives proton therapy a dosimetric advantage over photon radiotherapy in the sparing of nearby tissues and organs. Normal tissue sparing is paramount for children, who generally have longer expected survival times and are more sensitive to radiation. Furthermore, young children have smaller bodies than adults, and consequently, healthy organs are closer to the therapeutic radiation field. These factors increase the patient's lifetime risk of radiation carcinogenesis.

The therapeutic proton dose distribution within the patient has been thoroughly examined; exposures from stray radiation are poorly understood and controversial. Absorbed dose from stray neutrons is of particular concern because the relative biological effectiveness of neutrons for carcinogenesis is higher than that of photons. Stray neutrons are produced in both the treatment unit (or "external neutrons") and inside the patient (or "internal neutrons").1 The patient's exposure to external neutrons depends on the design of the proton treatment unit and the treatment technique.2 For passively scattered proton therapy (PSPT), the final, field-

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defining collimator is a predominant source of external neutrons.[3,4] Recently, Newhauser et al. estimated an effective dose from external neutrons of 98 mSv in an adult male patient who underwent a 36-Gy craniospinal irradiation (CSI) using PSPT (Ref. 5), and Taddei et al. estimated an effective dose from external neutrons of 344 mSv for a 10-yr-old boy who underwent a 30.6-Gy CSI with a 23.4-Gy boost PSPT treatment, both with PSPT, which corresponded to a 2.8% lifetime risk of second cancer mortality.[6] The results from these studies suggest that it is important to refine calculations of stray radiation doses and to find ways to reduce them.

Strategies were recently examined for reducing the effective dose from stray radiation for an adult patient undergoing PSPT for prostate cancer.[7] Modest modifications to the treatment unit (i.e., adding a proton collimator far upstream of the patient, changing the composition of the field-defining collimator from brass to tungsten alloy, and increasing the local shielding near the patient) reduced the effective dose from stray radiation emanating from the treatment unit from 320 to 108 mSv. These methods may also be effective for improving proton CSI, which is technologically more demanding than proton radiotherapy of the prostate.

The aim of this study was to evaluate two strategies to reduce stray radiation exposures for a pediatric patient undergoing CSI delivered using a PSPT unit:

1. modifying the field-defining collimator
2. adding an extra upstream collimator.

Monte Carlo simulations were performed to evaluate the performance of these modifications.

II. METHODS

II.A. Monte Carlo Model

In this study a proton radiotherapy treatment was simulated for a 10-yr-old male patient with a supratentorial primitive neuroectodermal tumor who underwent proton CSI using a PSPT treatment unit (PROBEAT; Hitachi America, Ltd., Tarrytown, New York) at our institution. The plan was created using a commercial treatment planning system (Eclipse Proton Planning; Varian Medical Systems, Inc., Palo Alto, California)[8] and wholebody kilovoltage computed tomography (CT) images. The treatment fields, patient, and nozzle specifications are described elsewhere[6] and are summarized here for the readers' convenience.

For the complete treatment plan, the prescribed dose to 100% of the volumes of the brain and spinal cord was 30.6 Gy, and the prescribed dose to the boost clinical target volume (a subvolume of the brain) was an additional 23.4 Gy. The boost field for this patient contributed only 5% to the effective dose from stray radiation and, for simplicity, was not considered in the present analysis. Therefore, the simulated treatment included four CSI fields: an inferior posterior-anterior (IPA) spinal field, a superior posterior-anterior (SPA) spinal field, a right posterior oblique (RPO) cranial field, and a left posterior oblique (LPO) cranial field. The spinal and cranial fields had proton beam energies of 140 and 180 MeV at the entrance of the treatment head, penetration ranges in the patient of approximately 9- and 10-cm water equivalent thickness, spread-out Bragg peak widths of 7 and 16 cm, and nominal air gaps of 10 and 2 cm between the distal component of the treatment unit and the proximal surface of the patient, respectively. Beam modifiers included a range-modulator wheel, scattering foil, range shifter, collimator block, and range compensator.[8] The largest of the three available snouts was used for all treatment fields (field size up to 25×25 cm^2). The thickness of the field-defining collimator was 4 cm for the spinal beams and 6 cm for the cranial beams.

The Monte Carlo simulations were performed using the Monte Carlo N-Particle eXtended (MCNPX) code (version 2.6b) (Ref. 9) with parallel computing methods. The MCNPX model of the treatment unit included a realistic proton source, the beam-modifying devices, the structural and housing components, and various static collimators.[8,10] A voxelized phantom represented the patient and was based on CT images for the 10-yr-old male patient, which included the entire body except the feet. The Hounsfield unit, or CT number, in each pixel was converted to a mass density and a biological material composition in the corresponding voxel.[11] Material compositions, variance-reduction information, selected physics options, and other model specifications were provided in detail elsewhere[6] and are available from the authors upon request.

II.B. Treatment Unit Modifications

In the first modification, the thickness of the field-defining collimator was increased from 4 to 8 cm for the spinal fields and from 6 to 8 cm for the cranial fields. Additionally, the composition of each field-defining collimator was changed from brass (density = 8.38 g cm^{-3}) to tungsten (density = 19.29 g cm^{-3}). This increased the shielding in the nozzle. In the second modification, a pair of jaws made of tungsten alloy were introduced inside the nozzle 186 cm upstream from the isocenter. The orthogonal jaws collimated the proton field to a rectangular shape. The lateral dimensions of the jaws were based on the maximum and minimum lateral extents of the field-defining collimator, reduced (i.e., demagnified) to take into account the proximity to the virtual source at 270 cm from the isocenter and expanded by 20% to minimize the edge-scatter effects[12] on the dose distribution of the therapeutic beam.

II.C. Dosimetric Calculations

The effective dose from external neutrons E was calculated as

$$E=\sum_{T}(w_T\, H_T)\,, \tag{1}$$

where

w_T = weighting factor for each organ or tissue T

H_T = organ equivalent dose from stray radiation.[13]

The w_T values were taken from the International Commission on Radiological Protection (ICRP) Publication 60 (Ref. 14). H_T was calculated for external neutrons for each field as the product of the radiation weighting factor w_R and the mean absorbed dose for each organ or tissue D_T or

$$H_T=w_R\, D_T\,. \tag{2}$$

The w_R values were calculated using the neutron spectral fluence incident upon the voxelized phantom according to the relationship between w_R and neutron energy given in ICRP Publication 92 (Ref. 13). Because w_R varied only slightly (<4%) for similar nozzle configurations,[6] for simplicity a mean value of w_R was estimated for each beam of the standard nozzle and applied to the modified nozzles. D_T was the mass-weighted average of the absorbed dose of all voxels within the organ or tissue. An exception was made for the skin and remainder organs and tissues, for which, for each field, D_T was approximated as the average absorbed dose for all voxels within the phantom. Although the effective dose from internal neutrons was calculated in a previous work,[6] in this study internal neutrons were not considered because their intensity could not be reduced by improvements to the nozzle.

In separate simulations for therapeutic protons and external neutrons, the absorbed dose in each voxel per source particle D_v/sp (in Gy sp^{-1}) was computed. For the external neutron simulations, all proton trajectories were terminated immediately upstream of the patient by a proton-stopping plane (imp:h = 0). For the therapeutic proton simulations (mode n h; imp:h > 0; imp:n = 0), the mass-averaged absorbed dose per source particle in the target volume $D_{T=target}$/sp was calculated for each treatment beam. This value was then used to normalize D_v/sp from external neutrons separately for each nozzle configuration and each treatment field, resulting in E (in mSv Gy^{-1}), H_T (in mSv Gy^{-1}), and D_T (in mGy Gy^{-1}). The E and H_T values from individual treatment fields were combined to yield the corresponding quantities for the entire course of 30.6 Gy. The methods for calculating these quantities are described in detail elsewhere.[6,7]

External neutron simulations were performed using the following nozzle configurations: the standard nozzle, the standard nozzle with the modified field-defining collimator, the standard nozzle with the additional jaws, and the standard nozzle with both the modified collimator and the additional jaws. Therapeutic proton simulations were performed for the standard nozzle and for the standard nozzle with the jaws; a comparison confirmed that the therapeutic dose was not perturbed by the nozzle modifications.

To achieve acceptable statistical uncertainties in D_T, 1×10^9 and 5×10^8 source particle histories were tracked for the spinal field and cranial field simulations, respectively. Geometry splitting at the surface of the voxelized phantom reduced the variance in the values of D_v. Statistical uncertainties were based on the coefficients of variation reported by the MCNPX code. The uncertainties in the mean w_R values were disregarded because the variance was minimal; the uncertainties in the w_T values were assumed to be zero because these values were set by definition rather than measurement. Statistical uncertainties were reported at the 68% confidence interval.

III. RESULTS

The values of effective dose from external neutrons E for each nozzle configuration are listed in Table I. As shown, E for the standard nozzle was 322.0 ± 0.6 mSv. E was reduced by 104.9 ± 0.8 mSv (33%) by changing the field-defining collimator material, 58.2 mSv ± 0.8 (18%) by adding jaws, and 138.7 ± 0.8 mSv (43%) by making both modifications.

Table II summarizes the exposure to external neutrons for the standard nozzle and the nozzle with modified collimator and jaws. The results are broken down by organ and by treatment field. Reductions in H_T ranged from 33 to 59% for the spinal fields and from 10 to 26% for the cranial fields. The modifications reduced E by 51% (from 236.5 to 115.6 mSv) for the spinal fields and by 21% (from 171.0 to 135.4 mSv) for the cranial fields.

The total computing time for all simulations was 15.3 cpu·years using parallel processing on 2.6-GHz, 64-bit processors (AMD Opteron; Advanced Micro Devices, Inc., Sunnyvale, California).

IV. DISCUSSION

This study has demonstrated that improved shielding of a proton treatment unit can substantially reduce the effective dose from stray radiation emanating from the treatment apparatus. The clinical implication is that the reduction in the effective dose will reduce the lifetime risk of radiation carcinogenesis. For the standard nozzle, the effective dose from external neutrons E was 322 mSv. By replacing the 4- or 6-cm brass field-defining collimator with an 8-cm tungsten field-defining collimator, E was reduced by 33%. By adding a pair of jaws made of tungsten-alloy far upstream of the patient, E was reduced by 18%. By making

both modifications, E was reduced by 43%, and percentage reductions in the organ equivalent dose from external neutrons H_T ranged from 33 to 59% for the spinal fields and from 10 to 26% for the cranial fields. The increased performance of the modifications for the spinal fields versus that of the cranial fields may be attributed to several variables, including the proton beam energy, field size, field shape, and proximity of radiosensitive organs (see also Ref. 2).

This finding is similar to previous results for reducing E for PSPT for prostate cancer. Taddei et al.[7] reported that modest modifications to a PSPT treatment unit reduced by 66% the effective dose from stray radiation originating in the nozzle. Thus, this study shows that for a second treatment site, the stray radiation emanating from a PSPT treatment unit was reduced substantially by making modest modifications to the treatment unit, approaching what may be achieved through alternative nozzle designs (e.g., an ideal nozzle that implements a spot-scanning technique).

In a previous study,6 the contribution of internal neutrons to effective dose was estimated to be 74 mSv for all four CSI treatment fields. Therefore, the effective dose from both internal and external stray radiation was 396 mSv for the standard nozzle and 257 mSv for the nozzle with a tungsten field-defining collimator and additional jaws. The National Council on Radiation Protection and Measurements (NCRP) has recommended a risk coefficient of 8.1%/Sv of effective dose for low-dose-rate, exposure-induced death for males between the ages of 0 to 19 yr (Ref. 15). The corresponding predicted excess attributable lifetime risk of second cancer fatality for the boy in this study was 3.2% for the standard nozzle and 2.1% for the nozzle with both modifications. That is, the modifications may result in about 11 fewer deaths from stray radiation per 1000 male pediatric patients treated in this way. For pediatric females, the same reduction in effective dose would correspond to about 16 fewer deaths per 1000 treated. However, the recommendations of the NCRP were based on data from a healthy population. The relative biological effectiveness of neutrons for radiogenic cancer mortality may be higher for a cancer population than for the general population. Therefore, the true risk (and risk reduction) for pediatric patients undergoing proton CSI may be even larger, which underscores the importance of improving proton beam delivery techniques to reduce patient exposures to stray radiation.

In conclusion, using Monte Carlo simulations, we found that it is possible to reduce the effective dose from stray radiation emanating from a PSPT treatment unit for a pediatric patient undergoing proton CSI by 43%.

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TABLE I

Effective Dose from External Neutrons E and Standard Deviation in E, σ_E, for the Standard Nozzle and Modified Nozzles

Nozzle	E (mSv)	σ_E (mSv)
Standard	322.0	0.6
Modified collimator	217.1	0.5
Additional jaws	263.7	0.6
Modified collimator and jaws	183.2	0.5

TABLE II

Values of Equivalent Dose H_T, Tissue Weighting Factor w_T, and Effective Dose E, from External Neutrons for the Standard Nozzle and the Nozzle with a Modified Field-Defining Collimator and Additional Jaws

		H_T (mSv)							
		Standard Nozzle				Modified Nozzle			
Organ or Tissue	w_T	SPA	IPA	LPO	RPO	SPA	IPA	LPO	RPO
Gonads	0.20	28.9	54.1	13.3	12.9	18.6	27.0	10.7	10.1
Red bone marrow	0.12	49.5	66.1	22.5	21.3	28.8	32.9	19.6	17.4
Colon	0.12	80.6	166.0	27.3	23.3	41.5	76.5	24.3	19.3
Lungs	0.12	266.5	177.3	67.0	64.2	135.0	79.0	50.9	47.3
Stomach	0.12	135.8	177.8	46.2	31.6	62.7	81.3	41.7	23.4
Bladder	0.05	40.0	88.3	13.9	14.4	26.9	42.6	11.5	11.9
Breasts	0.05	157.1	129.2	64.2	62.1	70.6	53.4	57.6	51.0
Liver	0.05	148.1	187.1	32.3	45.9	70.0	86.5	25.3	38.5
Esophagus	0.05	248.5	137.1	104.0	87.9	132.0	61.5	77.5	64.8
Thyroid	0.05	207.9	69.8	149.9	131.9	111.4	30.4	109.4	98.1
Skin	0.01	107.9	108.6	59.4	50.3	57.9	53.5	49.5	40.4
Bone surface	0.01	105.4	82.5	108.2	84.0	55.5	41.1	90.8	67.6
Remainder	0.05	107.9	108.6	59.4	50.3	57.9	53.5	49.5	40.4
E (mSv)		117.3	119.2	45.1	40.4	60.5	55.1	36.5	31.2