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Monte Carlo calculations for absolute dosimetry to determine machine outputs for proton therapy fields

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Abstract

The prescribed dose in radiation therapy has to be converted into machine monitor units for patient treatment. This is done routinely for each spread-out Bragg peak (SOBP) field either by calibration measurements, by using analytical algorithms or by relying on empirical data. At the Northeast Proton Therapy Center, a monitor unit corresponds to a fixed amount of charge collected in a segmented transmission ionization chamber inside the treatment head. The goal of this work was to use a detailed Monte Carlo model of the treatment head to calculate the dose delivered to the patient as a function of ionization chamber reading, i.e. to yield absolute dose in patients in terms of machine monitor units. The results show excellent agreement with measurements. For 50 SOBP fields considered in this study, the mean absolute difference between the experimental and the calculated value is 1.5%, where ~50% of the fields agree within 1%. This is within the uncertainties of the data. The Monte Carlo method has advantages over analytical algorithms because it takes into account scattered and secondary radiation, does not rely on empirical parameters, and provides a tool to study the influence of parts of the treatment head on the ionization chamber reading. Compared to experimental methods the Monte Carlo method has the advantage of being able to verify the dose in the patient geometry.

1. Introduction

The Northeast Proton Therapy Center (NPTC) has two isocentric gantry treatment rooms and one room with three horizontal beam lines. The gantry treatment nozzles contain various components for beam monitoring and beam shaping (Jongen et al. 1996). The field shaping system for broad beam modulation consists of four parts. A double-scattering system ensures a flat lateral dose profile, a range modulator provides the range shift necessary to distribute the dose homogeneously in depth to create a spread-out Bragg peak (SOBP), a patient field specific aperture confines the dose in lateral direction, and a patient field specific compensator adjusts the distal edge of the dose distribution to correspond to the distal edge of the target volume.

For a patient treatment field, the proton fluence required to deliver the desired dose is specified in machine monitor units (*MU*). A *MU* corresponds to a fixed amount of charge collected in one of the segmented transmission ionization chambers (monitor chambers) close to the snout of the gantry nozzles (figure 1). At the NPTC, an output factor is defined as the dose delivered to a calibration point in the treatment field divided by the required *MU*. Strictly speaking this definition differs from the one used in conventional radiation therapy, where the output factor is defined as the ratio of the dose per monitor unit at a given field to the dose per monitor unit for a reference field. Therefore, the output factor is relative and unitless. In our case the output factor is absolute. Nevertheless, for historic reasons it is called output factor instead of just output.

Output factors for proton therapy fields depend on the required range in the patient and on the width of the SOBP (Kooy et al. 2003). Thus, the output factor has to be determined for each individual patient field prior to treatment. At the NPTC it is currently either measured or analytically predicted. The measurements involve the reading of the monitor chamber output charge and the measurement of absolute dose using an ionization chamber[#] at a calibration point in water, a standard procedure in proton therapy (ICRU 1998). The point is ideally chosen within the flat region of the SOBP and in the projected center of the patient aperture to minimize the fluence perturbations due to the aperture field edge and the range compensator profile. In clinical practice we rely on our treatment planning system to accurately predict the dose distribution including patient internal heterogeneities and scattering effects of the aperture and range-compensator (except for small apertures). This permits us to do field calibration without using the patient specific aperture and range-compensator (i.e. to do open field calibrations) as all presumed effects from these devices are modelled in the treatment planning system (Kooy et al. 2005). Kooy et al. (Kooy et al. 2003) developed an analytical method to reduce the time-consuming task of individual field calibration. The analytical method is in clinical use and the agreement with measured output factors is usually within 1.5%.

The analytical model needs empirical normalization factors because of the complexity of the treatment nozzles. These factors had to be determined as part of the commissioning and account for differences in the scattering system as a function of the field dependent scatterers and modulators. The amount of data needed to determine parameters for analytical prediction depends on the complexity of the delivery nozzle. The basic analytical method is limited to open field calibrations and requires additional information to account for scattering in the beam path by patient field specific aperture and compensator. This information can be provided by an accurate in-patient dose calculation such as described by Hong et al. (Hong et al. 1996) or by a Monte Carlo method such as described here.

Monte Carlo simulations have not been used for output factor calculations in proton beams so far. However, they have been used extensively for photon and electron beams (Kapur et al. 1998, Liu et al. 2000, Haryanto et al. 2002, Ding 2003, Ding 2004, Ma et al. 2004, Turian et al. 2004, Popescu et al. 2005). These calculations, however, predict relative effects only. Calculations of absolute output require exact knowledge of the primary radiation field entering the treatment head and require the simulation of the monitor chamber reading as a function of delivered dose and particle fluence. For clinical use, accuracies of 2-3% are required. Obviously, this can only be achieved by modeling the machine head and the monitor chamber geometry in great detail. Our goal was the calculation of absolute machine outputs (not relative to a standard field) for proton therapy based on first principles without any empirical normalization factors.

2. Methods and Materials

The treatment heads at the NPTC were modeled using the Monte Carlo code GEANT4 (Agostinelli et al. 2003). Figure 1 shows the simulated geometry. Simulations are already being used for nozzle design, commissioning and quality assurance (Paganetti et al. 2004). The accuracy of the nozzle simulations in terms of beam range R and modulation width M of an SOBP was shown to be within 1 mm (R) and 2 mm (M) when compared with measurements (Paganetti et al. 2004). The accuracy was achieved by simulating more than 1000 individual parts in the treatment head. Our implementation takes into account explicit details of time dependencies in beam current and nozzle components to allow for four-dimensional (space and time) Monte Carlo simulations (Paganetti 2004).

[#]PTW Freiburg Germany, Model T23343

In each of the gantry nozzles, there are two redundant segmented ionization (monitor) chambers positioned downstream of the double-scattering and range modulator system and upstream of the final collimator and the patient specific collimator and aperture. Each chamber contains a small volume for absolute dosimetry. A monitor unit is defined as a fixed amount of charge collected in this volume, corrected for pressure and temperature effects. The chambers are open to the environment. Per definition, at NPTC, one *MU* corresponds to 3 nC collected. To be more specific, it corresponds to 300 counts, where a count is registered when the electronics has integrated 10 pC of charge. The output factor is defined as the dose in cGy to a point in the patient, or a point in the water phantom for pre-treatment calibration, delivered by one *MU*.

The collecting volume in a monitor chamber is modeled in the Monte Carlo geometry. It is a cylinder with a circular shaped base positioned with its center on the central beam axis. The radius of the active area, A_{IC} , of this cylinder is the radius of 0.95 cm plus half the gap (0.05 cm) to the next segmented part of the monitor chamber since protons passing through this half of the gap will produce charge being collected. The distance d_{IC} between the two parallel plates is 1 cm according to manufacturer's specification. The two cover plates, one Mylar and one Aluminum, have a thickness of 25 μm reducing the height of the dosimetry cylinder by $2 \times 25/2 \mu\text{m}$. All materials in the nozzle are implemented by exact material composition, e.g. taking into account particle interactions within the plates. Figure 1 shows the Monte Carlo simulated monitor chamber with a dedicated dosimetry volume for output factor measurements.

The current i_{IC} from the simulated plane parallel ionization chamber is ideally given as in (1) (e denotes the elementary charge ($=1.602 \times 10^{-19}$ C), ρ_{air} denotes the density of air ($=1.196 \times 10^{-3}$ g/cm³ at 22° C near sea level), W'_{air} denotes the mean energy to produce an ion pair, and $\dot{D}$ stands for the dose rate).

$$i_{IC} = \frac{e \cdot \rho_{air}}{W'_{air}} \cdot \dot{D} \cdot A_{IC} d_{IC} \quad (1)$$

W'_{air} for protons was measured for dry air as to be 34.2 ± 0.5 J/C for a mean proton energy of 180 MeV using a water calorimeter for ionization chamber calibration (Siebers et al. 1995). We use this number for protons and neutrons. Further we use 33.97 ± 0.5 J/C for electrons and photons (Boutillon and Perroche-Roux 1987), 34.2 ± 0.5 J/C for deuterons, 34.2 ± 0.5 J/C for tritons, 34.5 ± 0.5 J/C for ³He (Kanai et al. 1993), and 34.5 ± 0.5 J/C for ⁴He. For charged particles heavier than electrons we account for the small energy dependence of the W_{air} -values by use of equation (2) where A and z_{eff} are the atomic number and the effective charge of each specific projectile (ICRU 1979), respectively:

$$W_{air}/W'_{air} = Z_{eff}^{0.1} \left(1 + 0.4058 \cdot \left[E/Az_{eff}^{1.3333} \right]^{-0.4194} \right) \quad (2)$$

The efficiency ε_{IC} of the monitor chamber, which includes recombination effects, is $99.97 \pm 0.01\%$, according to the manufacturer. Clinically it is ensured that the efficiency is insensitive to small voltage variations and that it does not depend on the dose rate. The W_{air} -values depend slightly ($\approx 1\%$ effect) on the air humidity resulting in a correction factor h_{air} . In the clinical beam line the monitor chamber is open and the chamber reading depends on the air temperature and pressure. However, in the Monte Carlo simulation we set the correction factors t_{air} and p_{air} to unity, since it is assumed that *MU* determinations are always made at 22° C at standard atmospheric pressure. With the dose rate parameterized as the stopping power in air (averaged over the energy spectrum for the particle flux), $(dE/dx)_{air}$, times the particle flux Φ through the dosimetry volume, the chamber current may be expressed as

$$i_{IC} = \frac{e \cdot \rho_{air} \cdot A_{IC} \cdot d_{IC} \cdot \varepsilon_{IC} \cdot p_{air} \cdot t_{air}}{W_{air} \cdot h_{air}} \times \left(\frac{1}{\rho_{air}} \left(\frac{dE}{dx} \right)_{air} \Phi \right) = \frac{e \cdot \varepsilon_{IC}}{W_{air}} \times \left(\frac{dE}{dx} \right)_{air} \cdot \Phi \cdot A_{IC} \cdot d_{IC} \quad (3)$$

Event though the dosimetry volume is only a small part in a segmented ionization chamber, i.e. the electric field lines are perfectly parallel, the relationship given in (3) has some serious simplifications. Particles do not always enter the dosimetry volume through the cylinder base and travel the exact distance of d_{IC} , i.e. 1 cm. To be more accurate one has to consider that particles may enter the dosimetry volume through the cylinder mantle. These may be particles scattered at devices close to the dosimetry volume. In addition, the path length may vary depending on the angle with respect to the central beam axis. Although the average total path length may differ only slightly from d_{IC} , low energy particles (scattering or nuclear interaction products) tend to have a larger angle to the beam axis, which increases their path length through the chamber while they also have a higher stopping power. Thus, the corrected equation, using the path length, p , of each particle per step, ξ , integrated over the particle flux is

$$i_{IC} = \frac{e \cdot \varepsilon_{IC}}{W_{air}} \times \iint \left(\frac{dE}{dx} \right)_{air} p \cdot d\xi \cdot A_{IC} \cdot \frac{d\Phi}{dE} \cdot dE \quad (4)$$

In the Monte Carlo simulations, we track primary protons as well as secondary photons, electrons, protons, deuterons, tritons, ^3He , ^4He , and neutrons. The physics setup is described elsewhere (Paganetti et al. 2004). All particles are tracked through the entire nozzle. Secondary particles, except for electrons and photons, are ‘created’ without limitations (i.e. thresholds). The production threshold for photons is set to ensure that particles are ‘created’ only if the projected CSDA-range is bigger than 1 mm in air. For electrons in air a production threshold in energy of 1 keV was chosen. Otherwise the energy is deposited locally. Using smaller cut-offs did not change the results for i_{IC} . Note that the cut-off does only influence generation, not tracking. Each particle is tracked down to a kinetic energy of 0.

The energy deposited is sampled for each particle step that takes place in the dosimetry volume. A step refers to a computational step of the particle track, which is determined by a pre-defined maximum step length and the distance between two interactions. The monitor chamber reading is determined within the Monte Carlo simulation by using (4) replacing flux by fluence and thus simulating the collected charge rather than the current. Note that A_{IC} is not used explicitly in the Monte Carlo sampling because it is implicit when considering only tracks within the dosimetry volume. The collected charge is proportional to the particle fluence, i.e. the simulated number of histories. The use of energy deposition and W_{air} -value instead of applying directly the energy lost by secondary electrons causes the results to be independent of the step size (if reasonably small; maximum step sizes of 0.25 mm and 0.01 mm were used in air and in the foils, respectively) and independent of the production threshold in range for secondary particles (if significantly smaller than the chamber size).

The GEANT4 versions prior to 8.0 had shortcomings with respect to electron tracking, i.e. the code permitted electrons to reach geometric boundaries in large steps, and underestimated lateral displacement near interfaces (Poon et al. 2005). Such multiple scattering effects depend on the chamber wall thickness. Because of our small step sizes and because the thin walls (25 μm) of the ionization chamber cause only a small field disturbance within an extended air volume, possible step size artifacts due to distortions in electron fluence and angular distributions do not play a role.

The path length of a particle traveling through the dosimetry volume differs slightly from the pure distance between the two foils due to the foil thickness and the beam’s angular spread. In fact, the median pathlength is about 5 μm longer than the air gap between the two foils, which,

in turn, depends slightly on R and M (figure 2). The more material the beam passes through, the bigger is the angular spread of the beam at a given energy.

The energy deposited in the monitor chamber per particle step is illustrated in figure 3. The distributions depend on the particle energy distributions and the particle fluences. These in turn depend on the combination of scattering and absorbing material applied to produce the prescribed SOBP. The maximum scales roughly with the beam energy, i.e. with the required range. The two groups of distributions correspond to two different scatterers in the double-scattering system. Each nozzle at the NPTC has three different contoured scatterers. Note that the energy deposited per step shown in figure 3 depends on the step size and may thus depend on the chosen maximum step size without altering the total energy deposited.

To relate the monitor chamber reading to the dose at calibration point, D_C , the dose in water, given in MeV per considered mass, is simulated. We consider a volume in the water tank that corresponds to the size of the calibration chamber used for measurements. The volume of the calibration chamber is 0.055 cm^3 according to the manufacturer[#]. The output factor ψ can thus be calculated using equation (5) (with $e_{ic}^{\text{collected}}$ being the collected charge in units of 3 nC). The output factor has a unit and is truly absolute and not relative to a reference field. Thus, the presented Monte Carlo framework offers absolute dosimetry.

$$\Psi = \frac{D_c}{e_{ic}^{\text{collected}}} \left[\text{cGy} / \text{MU} \right] \quad (5)$$

An experimental calibration in practice yields the absolute dose D_C directly. To simulate D_C , the calibration chamber response was not modeled explicitly. Our goal is the clinical use of Monte Carlo simulations for output factor determination for a given position in the SOBP within the patient. Thus, D_C should be given as accurate as possible (independent of the response of a measuring device) by considering the energy deposited directly as predicted by the Monte Carlo. Simulation of the ionization chamber used for dose measurements in water is a separate issue and could introduce uncertainties (Palmans and Verhaegen 1998, Poon et al. 2005).

3. Results and Discussion

The Monte Carlo method must simulate the delivered treatment field – in terms of produced range R , modulation M and lateral profile - with high accuracy to provide reliable output factors. The agreement between measured and calculated R and M was previously tested and is typically within 1-2 of mm (Paganetti et al. 2004). The NPTC treatment nozzles each have nine different rotating, circular, range-modulator tracks. The optimal track for a particular patient field is chosen according to the required range modulation width and field size. Combinations of range-modulator tracks and contoured scatterers are grouped together and define treatment options (Paganetti et al. 2004). The settings of beam energy, double-scattering system and variable collimators vary within one option. Output factors can be conveniently analyzed as a function of $r = (R-M)/M$ (Kooy et al. 2003). Figure 4 shows measured and calculated output factors for such options. The data were taken for large apertures where aperture and range compensator effects can be neglected. The points within one option do not lie on a continuously increasing curve which can be explained by a range dependent shift in virtual source position, which introduces up to 7% variation in output within an option for a given value of r (Kooy et al. 2005).

The Monte Carlo simulations predict the output factor with great accuracy. For the 50 fields considered, the mean absolute difference between the experimental and the calculated value is 1.5% where ~50% of the fields agree within 1%. Figure 5 shows a histogram with the

deviations from the experimental values. The statistical uncertainty of the simulated data is about 1.5% for the number of histories tracked (10 million per field). The biggest discrepancies occur for large values of r . Presumably this is because for small values of M it is difficult to measure exactly in the flat dose region of the SOBP. On average, the calculated values exceed the experimental values by 1.1%. This could be a systematic shift due to a small difference between the actual radius of the ionization volume compared to the design value of 0.95 cm. A change of 0.05 mm already yields a 1% difference in ψ . The difference between experimental values and calculated values is certainly within the statistical uncertainties of the Monte Carlo simulation and the experimental uncertainties.

For routine clinical use output factors have to be known with accuracy in the order of 2-3%. To achieve the accuracy illustrated in figure 4 and figure 5 with Monte Carlo simulations, the respective beam delivery nozzle has to be modeled in great detail. Components that are not part of the beam shaping system, e.g. frames holding foils and housing of ionization chambers, need to be modeled because they may account for scattered radiation. Further, nuclear interactions can play a significant role and secondary particles have to be tracked. Nuclear interaction products contribute significantly to ionization chamber response in particular in high energy beams (> 100 MeV) (Laitano et al. 1996). The monitor chamber detects the energy loss that corresponds to the entrance region of a SOBP where the influence of nuclear interactions on dose is in general more pronounced than in the peak region (Paganetti 2002).

Figure 6 shows the contribution of secondary radiation to the monitor chamber reading for two fields ($R=22$ cm, $M=4$ cm; $R=10$ cm, $M=4$ cm). The main contributions are from primary protons, secondary protons (from nuclear interactions) and electrons. All other directly ionizing particles' contributions are negligible. Note that for electrons a energy cut-off of 1 keV was used. Thus, the contribution of secondary electrons with a smaller range is included in the primary particle. The contribution of different particle types to the output factor is shown in table 1 for four fields. The variations with field are generally small and mainly depend on the initial beam energy, R , M , and the settings of the double-scattering system. For the four fields, the relative contribution of primary protons goes up with decreasing M , which is an indication that the most significant scattering takes place in the range modulator wheel. Table 2 shows the 0.5-2% contribution of secondary or scattered radiation to the output factor from the scanning magnets and variable collimators alone. Contributions from other devices were negligible with the exceptions of the beam modulating devices and ionization chambers.

The data shown so far did not include patient specific apertures. The output factor does not depend on the aperture as long as its size allows neglecting edge scattering for the dose at isocenter. If the aperture opening is bigger than ~ 6 cm², as it is for almost all fields we use clinically, neglecting the aperture is justified if the calibration point projects near the center of the aperture opening. Fields used for stereotactic radiosurgery usually have small M and small aperture opening. Such fields are particularly challenging for output factor determination. Analytical models have difficulties because of significant aperture scattering. Measurements are difficult because the reference point in the target is positioned within a very small area of homogeneous dose distribution. Very small thimble chambers are therefore needed. Table 3 shows an example of the influence of aperture scattering by comparing output factors for a small and a large aperture opening. In the case of the radiosurgery field, the Monte Carlo simulation predicts a reduction in output factor by 17.7% if the dose at isocenter is integrated over a radius of 2 mm around the central axis. This agrees very well with the measurement.

A calibration in water does not necessarily accurately reflect scattering inside the patient. The significance of this effect depends on the complexity of the radiation field and the patient geometry. Measurements or analytical calculations cannot overcome this problem. It can only be solved by importing the patient's CT into the Monte Carlo geometry. Based on a detailed

simulation of treatment head and patient geometry we are currently also performing output factor calculations for reference points in the patient. This is done by importing the patient CT information (Jiang and Paganetti 2004), then calculating the dose distribution in the patient (to define D_C), and then relating the dose to the simultaneous simulation of the chamber reading (using (4) and (5)). Thus, the Monte Carlo simulation method predicts absolute delivered dose to the patient without using any empirical normalization factors.

This study shows the feasibility of absolute proton dosimetry using Monte Carlo simulations by modeling the treatment head geometry and the monitor chamber reading. However, to reduce statistical uncertainties to a level acceptable for clinical use ($\sim 2\%$), most primary particles (protons) and secondary particles (protons, electrons, neutrons, and photons) need to be tracked through a complex geometry with small step sizes (depending on the size of the monitor chamber). Consequently, Monte Carlo calculations can become quite time consuming. This is in particular true in our case because of the small size of the dosimetry volume. Calculating output factors therefore takes about 10 hours on ten CPUs (2GHz) running LINUX. Because the limiting factor for statistical accuracy is the small size of the monitor chamber volume, we did additional simulations where we increased the size of the dosimetry volume to 3 cm, compared to the real value of 1 cm. However, the pathlength distribution turned out to be affected by this, changing the output factor by about 1%. A practical solution is to assume several, parallel volumes, each with a radius of 1 cm. This speeds up the simulations by about a factor of 10, i.e. to ~ 1 hr per field and can be applied for the majority of fields. However, one has to be careful not to use this approximation in cases where small upstream apertures may cause an inhomogeneity of particle fluence across the radiation field, e.g. if the variable collimators are moved significantly into the field.

4. Conclusion

Absolute machine outputs for SOBP proton fields have been simulated for the first time using a full Monte Carlo method. Accurate modeling the proton beam delivery nozzle, including areas used for dosimetry, enables the use of Monte Carlo simulations for absolute dosimetry. Output factors for proton therapy treatments can thus be calculated. For the 50 fields considered in this study, the mean absolute deviation from the experimental value was just 1.5%, where $\sim 50\%$ of the fields agreed within 1%. There could be a 1.1% systematic shift due to a difference in actual, but unknown, ionization chamber construction details. Monte Carlo predicted output factors can be part of an extended quality assurance program. The influence of specific devices in the treatment head on the output factor reading can be studied. Furthermore, the Monte Carlo method can be used to estimate output factor in daily clinical routine to verify or replace some of the time-consuming experimental verifications. Monte Carlo simulations may be the method of choice for treatment fields where, due to extreme scattering conditions (e.g., small apertures), analytical models may have difficulties and experiments are tedious. In addition, absolute doses as a function of machine monitor units can be calculated for a CT based patient geometry. Because of the successful calculation of absolute doses in a complex geometry, the present work also serves as a validation of the GEANT4 package for use in proton radiation therapy.

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Figure 1.

Left: Gantry nozzle as simulated in GEANT4. The various components are described in more detail elsewhere (Paganetti et al. 2004). Right (upper): Magnified view of the two redundant ionization (monitor) chambers near the nozzle exit (arrow). Right (lower): Beam's eye view of the monitor chamber. The simulated cylindrical dosimetry volume (1 cm radius) used for output factor measurements is shown in grey.

Figure 2. Path length distributions of particles crossing a dosimetry volume in one of the segmented ionization chambers. The two curves correspond to $R=10$ cm, $M=4$ cm (solid) and $R=22$ cm, $M=4$ cm (dashed).

Figure 3.

Energy deposited $(dE/dx)_{air}$ per step length (maximum 0.25 mm) divided by W_{air} for all particles traveling through a dosimetry volume for nine different SOBPs. The solid lines represent $(R,M) = (22 \text{ cm}; 8 \text{ cm})$, $(20 \text{ cm}; 12 \text{ cm})$, $(15.5 \text{ cm}; 12 \text{ cm})$, and $(15.5 \text{ cm}; 7.5 \text{ cm})$. The dashed lines represent $(R,M) = (12 \text{ cm}; 9.5 \text{ cm})$, $(12 \text{ cm}; 6 \text{ cm})$, $(12 \text{ cm}; 2.5 \text{ cm})$, $(8.5 \text{ cm}; 7.5 \text{ cm})$, and $(8.5 \text{ cm}; 3.5 \text{ cm})$.

Figure 4.

Measured (open circles) and Monte Carlo simulated (crosses) output factors, ψ , as a function of range minus modulation width, $(R - M)$, divided by the modulation width, M . The circles represent an area of $\pm 3\%$ around the measured value. All data are for large apertures (negligible aperture scattering).

Figure 5.
Deviation of Monte Carlo simulated output factors from experimental values.

Figure 6.

Energy deposited $(dE/dx)_{air}$ per step length (maximum 0.25 mm) divided by W_{air} for particles traveling through the dosimetry volume for two different SOBPs (upper: $R=10$ cm, $M=4$ cm; lower: $R=22$ cm, $M=4$ cm). Shown is the total energy deposited (solid) as well as the contributions of primary protons (short-dashed), secondary protons (dotted), and electrons (long-dashed).

Table 1

Contribution of different particle types (electron production cut-off 1 keV in air) to the output factor. (Column headers are (R,M) pairs in cm).

	(R=7.5, M=7)	(R=10, M=4)	(R=15.5, M=15)	(R=22, M=4)
Beam energy at nozzle entrance	153.02 MeV	169.23 MeV	180.99 MeV	215.45 MeV
Primary Protons	70.5 %	71.7%	69.8 %	71.3 %
Secondary Protons	1.1 %	1.3 %	1.3 %	1.6 %
Electrons	27.5 %	27.0 %	27.2 %	26.8 %
Helium Ions	0.9 %	0.1 %	1.6 %	0.6 %
All other particles	< 0.1 %	< 0.1%	< 0.1 %	< 0.1 %

Contribution of different particle types (electron production cut-off 1 keV in air) to the output factor for particles that are generated or scattered at sources other than range modulating material, double-scattering system, and ionization chambers. These are the scanning magnets and the variable collimators. (Column headers are (R,M) pairs in cm).

	R=7.5 M=7	R=10 M=4	R=15.5 M=15	R=22 M=4
From scanning magnets				
Primary Protons	0.1 %	0.3 %	0.2 %	0.3 %
Secondary Protons	<0.1 %	<0.1 %	<0.1 %	<0.1 %
Electrons	<0.1 %	0.1 %	0.1 %	0.1 %
From variable collimators				
Primary Protons	0.1 %	0.3 %	0.2 %	0.5 %
Secondary Protons	0.2 %	0.3 %	0.3 %	0.6 %
Electrons	0.1 %	0.2 %	0.3 %	0.6 %

Table 3
Measured, ψ_{EXP} , and simulated, ψ_{SIM} , output factors (in cGy/MU) for a breast cancer treatment field and for a radiosurgery field.

Treatment	R [cm]	M [cm]	aperture opening [cm ²]	ψ_{SIM} w/o aperture	ψ_{SIM} with aperture	ψ_{EXP} with aperture
Breast	12.6	10.6	45.68	0.931	0.934	0.915
Radiosurgery	12.84	1.87	1.57	1.64	1.35	1.37