DESIGN CHOICES OF THE MedAustron NOZZLES AND PROTON GANTRY BASED ON MODELING OF PARTICLE SCATTERING

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Abstract

MedAustron, the Austrian hadron therapy center is currently under construction. Irradiations will be performed using active scanning with a proton or carbon ion pencil beam which is subject to scattering in vacuum windows, beam monitors and air gap. For applications where sharp lateral beam penumbras are required in order to spare critical organs from unwanted dose, scattering should be minimal. A semi-empirical scattering model has been established to evaluate beam size growth at the patient due to upstream scattering. Major design choices for proton gantry and nozzle based on the scattering calculations are presented.

INTRODUCTION

In hadron therapy, a sharp lateral penumbra of the scanned pencil beam is important for reducing the dose to critical organs and healthy tissue in the vicinity of the tumor [1]. The MedAustron synchrotron will provide proton (60-250 MeV) and carbon ion (120-400 MeV/n) beams for medical use to two fixed beam-line irradiation rooms and one gantry (protons only). Degradation of the lateral penumbra due to scattering is mainly a low-energy proton issue. In order to estimate beam size growth at the patient due to upstream scattering, a semi-empirical scattering model [2] has been extended to take beam optics into account. Different design options for the MedAustron proton gantry and nozzles are evaluated with respect to beam sizes at the isocenter (IC).

METHOD

Scattering Power

The phase space ellipse of a Gaussian beam is uniquely defined by its angular variance, $\langle z'^2\rangle$, covariance, $\langle zz'\rangle$ and spatial variance, $\langle z^2\rangle$ (z and z' are the position and angular direction of a single particle in horizontal (x) or vertical phase space (y), $\langle\rangle$ denotes an average over the whole beam). Expressed in conventional Twiss parameters, α, β, γ, these three quantities are equivalent to:

$$\vec{Z} \equiv (\langle z'^2\rangle, \langle zz'\rangle, \langle z^2\rangle)^T = \frac{E_{z,1\sigma}}{\pi}(\gamma_z, -\alpha_z, \beta_z)^T \quad (1)$$

where $E_{z,1\sigma}$ is the horizontal or vertical 1σ emittance. The scattering power, $T(s)$, is defined as the rate of increase of the beam divergence due to scattering, with longitudinal coordinate s:

$$T(s) = \mathrm{d}\langle z'^2\rangle/\mathrm{d}s \quad [\mathrm{rad}^2/\mathrm{cm}] \quad (2)$$

The scattering power model used in this paper is the *differential Highland* model, introduced by Kanematsu [2], where $T(s)$ is a function of the beam energy and material at s, and the amount of material the beam has traversed up to s. It is applicable to heterogeneous geometries and thick scatterers.

Beam transport

Combining the three projection integrals [2] for the evolution of $\vec{Z}$ from $s = a$ to $s = b$ into a single equation gives:

$$\vec{Z}_b = M_{a\to b}\vec{Z}_a + \int_a^b M_{s'\to b} \times (T(s'), 0, 0)^T \mathrm{d}s' \quad (3)$$

where $M_{a\to b}$ is a 3×3 matrix, transporting $\vec{Z}$ from a to b (see below). Thus, $\vec{Z}_b$ is a sum of two terms: one depending on the phase space of the incoming beam ($\vec{Z}_a$), the other on the amount of scattering along the transfer line, $T(s)$.

In a drift space, the transport matrix is:

$$M_{a\to b}^{Drift} = \begin{pmatrix} 1 & 0 & 0 \\ b-a & 1 & 0 \\ (b-a)^2 & 2(b-a) & 1 \end{pmatrix} \quad (4)$$

while in a general transfer line, it must be calculated from the Twiss functions.

The Full Width at Half Maximum (FWHM), W_z, of a Gaussian beam is related to $\langle z^2\rangle$ as:

$$W_z = 2\sqrt{2\ln 2} \times \sqrt{\langle z^2\rangle} \approx 2.35\sqrt{\langle z^2\rangle} \quad (5)$$

The terms "beam width", or "beam size" refers in this document to the FWHM of the beam, which is assumed to be Gaussian in x and y. At the IC the unscattered beam width can be chosen in the range 4-10 mm.

DESIGN OPTIONS

Proton Gantry

The proton gantry is a 15.6 m long (beam path), rotating construction that allows for irradiation of the patient from any direction (see Fig. 1). To separate the rotating vacuum pipe of the proton gantry from the fixed, upstream, vacuum pipe, double vacuum windows (2×50 μm kapton) were originally foreseen at the coupling point (CP).

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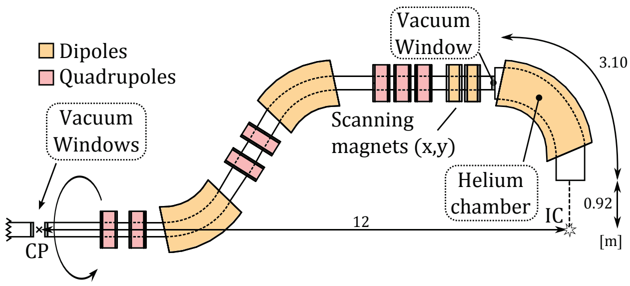

Figure 1: Schematics of one proton gantry option: double CP vacuum windows and helium in last bending dipole. Without helium in the dipole, the last vacuum window would be located just before the nozzle.

There is no dedicated vacuum chamber in the 90° bending dipole at the end of the gantry. Instead, the magnet yoke serves as vacuum chamber in order to increase the aperture and maximize the scanning area. In the event of a vacuum leakage, reparations of the 70 ton dipole could be time-consuming. By filling the entire dipole with helium (which has a lower scattering power than air), it would be robust against minor leakages.

The beam growth due to scattering in gantry vacuum windows and helium in the dipole is evaluated later in this paper.

Scanning Nozzles

Fig. 2 shows schematic models of the considered nozzle options, containing beam intensity, position and profile monitors (1.1 mm water equivalent thickness (WET) [3]). Ridge filters to widen the Bragg peak can be inserted when irradiating with carbon beams. The overall length from the last vacuum window to the IC is 92 cm. It ends with a 0.5 mm thick Plexiglas window, preventing the patient from accidentally reaching into the nozzle. Two methods for minimizing the beam growth in the nozzle are considered:

1. Minimizing the nozzle-to-isocenter air gap by moving the monitors as close as possible to the patient (Nozzle 1)
2. Inserting a helium-filled bellow between the vacuum window and the monitors to reduce scattering in air (Nozzle 2)

Additionally, the use of a movable Plexiglass range shifter (RS) to reduce the beam range below the limit set by the minimum extraction energy is evaluated (Nozzle 3). The RS thickness (3 cm) roughly corresponds to the minimum proton and carbon ion beam penetration depths, which allows for irradiation of superficial tumors. In situations where the penetration depth of the lowest possible extraction energy is too great, a range shifter (RS) can be used. The impact on beam size of a movable range shifter attached to the nozzle (Nozzle 3) is evaluated. For safety reasons, the gap between nozzle (or RS) and patient should be at least 10 cm.

With movable monitors, the deflection angle of the scanned beam (φ, Fig. 2) necessitates correlating the measured beam position to the longitudinal position of the monitors. This increases the complexity of beam verification during scanning and certification of the beam delivery system. In the proton gantry, though, the optics is such that $\varphi \approx 0$ even when irradiating at the edges of the field.

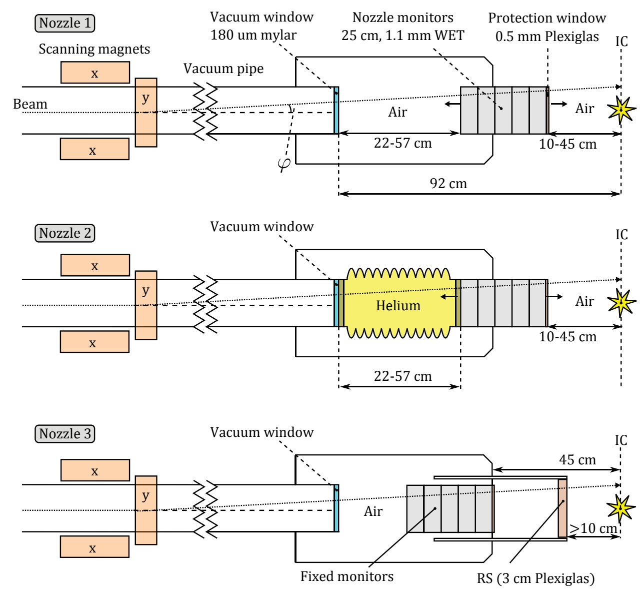

Figure 2: Different nozzle options. 1: Movable monitors. 2: Movable monitors and helium chamber after vacuum window. 3: Fixed monitors, movable (and removable) range shifter.

SIMULATION RESULTS

Proton Gantry

Eq. 3 has been applied to a 60 MeV proton beam, using the optics of the MedAustron proton gantry with/without CP vacuum windows and dipole helium, neglecting scattering in air and nozzle. The unscattered FWHM at IC is 4 mm in both planes. Beam sizes when taking scattering into acccount are summarized in Table 1.

Without vacuum windows at the CP, or helium in the dipole, the vacuum window just before the nozzle is the only scattering element. This results in beam sizes in both planes of 5.9 mm. Due to the different optics in the horizontal and vertical plane, double vacuum windows at the CP cause a large beam growth in the horizontal plane, while the vertical beam size is unaffected. Filling the dipole with helium would result in beam sizes of 15-20 mm in x and y. Thus, in order to produce symmetric and small beams at the IC - even for low energy proton beams - neither vacuum windows at the CP, nor helium in the 90° dipole will be used.

Table 1: IC Beam Sizes in the Horizontal and Vertical Plane for a 60 MeV Proton Beam Passing Through the Gantry

He in dipole	Vac. Wind. at CP	FWHM at IC [mm]	
		Hor.	Vert.
-	-	5.9	5.9
-	√	13.7	6.0
√	-	15.0	18.8
√	√	19.9	19.6

Isocenter Beam Sizes

In air: Horizontal and vertical FWHM at the isocenter, in air, have been calculated for the three nozzle options for all beam energies. Results are presented in Fig. 3. With Nozzle 1, beam sizes of 9-11 mm can be achieved for low-energy protons, depending on air gap. Adding the helium chamber reduces the beam sizes by at most 2 mm. At higher energies, the scattering effect decreases and the FWHM at the IC approaches 4 mm.

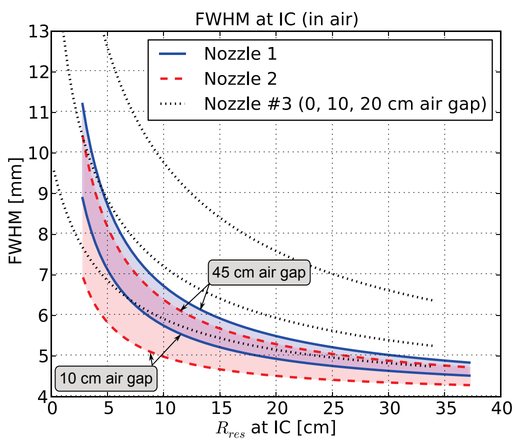

Figure 3: FWHM (horizontal and vertical) in air vs. residual beam range in water, R_{res}, at the IC. The three dotted lines indicate IC beam sizes for Nozzle 3, assuming 0, 10 and 20 cm air gap between RS and IC.

Fig. 3 also shows beam sizes for Nozzle 3 for RS-IC air gaps of 0, 10 and 20 cm (No air gap would be equivalent to attaching the range shifter/pre-absorber directly on the patient's skin). The primary use of the RS is to reduce the penetration depth below 3 cm, but, compared to Nozzle 1, it could also be used to reduce beam sizes of targets at up to 6-7 cm depth, if the RS-IC air gap is small enough.

In water: By placing a water phantom at the IC, the beam sizes at the Bragg peak (in water) can be evaluated. The amount of water in front of the IC is matched to the energy of the incoming beam such that the Bragg peak is located exactly at the IC, where the beam is focused.

At low energies, the beam size is approximately equal to the beam size in air, and scattering in the nozzle and air gap is dominant. With increasing energy, scattering in the nozzle decreases and the Bragg peak beam sizes feature a minimum at a depth of 5-8 cm. At high energies, scattering

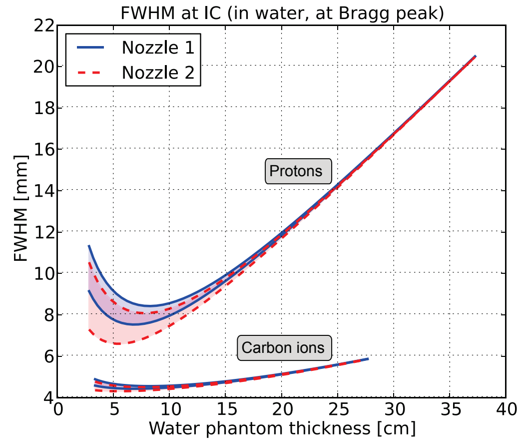

Figure 4: FWHM at IC, in water (at Bragg peak). Upper bands: 60-250 MeV protons, lower bands: 120-400 MeV/n carbon ions.

in the water phantom is dominant. The gain in optimizing the nozzle and reducing the air gap would be less than 1 mm for Bragg peaks deeper than 10 cm.

CONCLUSIONS

In order to produce symmetric beam sizes smaller than 13 mm FWHM at the isocenter of the proton gantry at low energies, a windowless solution is required at the proton gantry coupling point, as well as keeping the vacuum in the 90° bending dipole.

For superficial targets and minimum air gap, the FWHM of the beam could be reduced from 11 mm to 7 mm with movable monitors and a helium chamber that reduces the total amount of air in the beam path. However, a similar reduction can be achieved by attaching a range shifter directly on the patient.

In the two fixed beam line irradiation rooms, the scanned beam will not be parallel ($\varphi > 0$), and potential benefits of movable monitors must there be weighed against the increased complexity of manufacturing, beam position verification and certification. One possibility is to install non-movable monitors in these fixed, dual-particle, beam lines and use Nozzle 2 only in the proton gantry. The facility would then still have the capability to treat patients with indications of narrow, low-energy proton beams without patient-specific hardware.

REFERENCES

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