INNOVATIVE DESIGN OF THE ISOCENTRIC PROTON/CARBON ION GANTRIES*

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Abstract

We describe an update on the design of proton/carbon gantries using the principle of the Non-Scaling Fixed Field Alternating Gradient (NS-FFAG) accelerators. The novel design consists of the NS-FFAG cells with the final focus and scanning system above the patient. The cells are made of small superconducting combined function dipoles. The beam is transported through the gantry transport system under a fixed magnetic field covering the whole required energy range. The weight of transport elements is dramatically reduced to a total less than one ton to be compared to the 135 tons of the present Heidelberg new facility. This is due to small magnet size resulting from the extremely strong focusing and small dispersion function (150-400 MeV/u for fully stripped carbon ions). The cost reduction, ease on operation due to the fixed magnetic field, and simplified treatment, make this design very competitive to the common used designs. We also present preliminary magnet design for the fully stripped carbon application.

INTRODUCTION

A major challenge and cost in present and future hadron therapy facilities is the beam delivery system. In almost all facilities, at least one of the patient delivery rooms is equipped with an isocentric gantry system. An isocentric gantry system is becoming a necessity for each facility; it is especially important for treating hard-to-reach tumors especially around the spine (Chordomas and low grade Chondrosarcomas, unresectable sacral chordomas). Hadron therapy treatment requires different incident angles to avoid damage to sensitive areas (such as the spine) by radiation. The spot scanning technique is another necessity. It allows the accumulated dose within the restricted tumor area to be accurate within 1-2%. Presently, most of the scanning systems use a "parallel" beam technique, with a very large bending magnet above the patient. We assume that an area of ±10 cm or 20x20 cm in both planes represents the largest transverse ion positions. The ion dose has to be delivered with very good reliability and stability. Larger cancerous tumors require transverse position scanning at different beam energies and an angle variation around the patient provided by gantry rotation. The newest state-of-the-art gantry for hadron therapy, made of standard warm magnets, is in the facility in Heidelberg [1].

The non-scaling FFAG (NS-FFAG) concept should dramatically reduce overall weight [2] of the carbon gantry made of warm magnets. The fixed field magnets have transversely linear variation of the magnetic field and they could be made using the superconducting wire without any iron. The reduction of weight comes from the small magnets. The very large bending magnet at the end of the gantry is replaced with a scanning system at 3.9 and 2.5 meters above the patient. Two proton isocentric proton gantries are presented. The first gantry is made of small warm combined function magnets (permanent magnets are possible) with a height difference of ~7.8 meters. The second one is a smaller size isocentric gantries and it is made of superconducting magnets with heights 4.8 meters. A comparison between the "parallel beam scanning" in contrast with scanning with an angle the order of 25-32 mrad is discussed in details. An easier operation and use of the gantry is expected, as the magnetic field remains fixed. During the whole treatment over the required energy range no adjustment is necessary. The proton gantry operates within 79-250 MeV range and this corresponds to a momentum range of $\delta p/p=-30\%$ up to 30%.

THE NON-SCALING FFAG

The non-scaling FFAG is made of fixed field combined function magnets with a linear transverse variation of magnetic field [3]. An example a gantry basic cell with orbits through the gantry is shown in Figure 1.

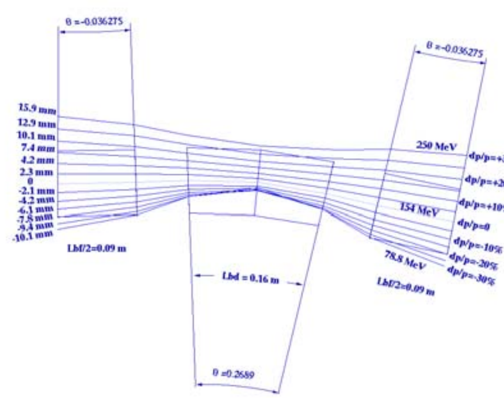

Figure 1: Orbits in one gantry cell for treatment energies.

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The central magnet is a defocusing combined function magnet with a length of 16 cm. The minimum of dispersion and horizontal β_x function is at the middle of it. Dispersion function throughout the FFAG lattice retains very small values. The large momentum acceptance and small orbit offsets are a consequence of the small dispersion. The gantry, made of the non-scaling FFAG cells, accepts and propagates different energies ions with very small variation of the orbit. The combined function magnets of the basic cell with betatron functions are presented in Figure 2. The middle of the dipole is selected as a place for the input and output of the gantry.

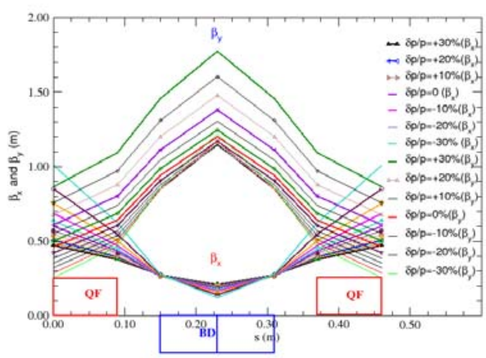

Figure 2: Betatron functions β_x and β_y in the basic cell.

The offsets at the end of the cell are obtained from the Polymorphic Tracking Code (PTC) [4] in a kinetic energy range between in momentum $\delta p/p < \pm 30$ % or in kinetic energy range 78.8-250 MeV/n.

Gantry designs

The first example is an isocentric proton gantry made of small warm magnets is shown in Figure 3.

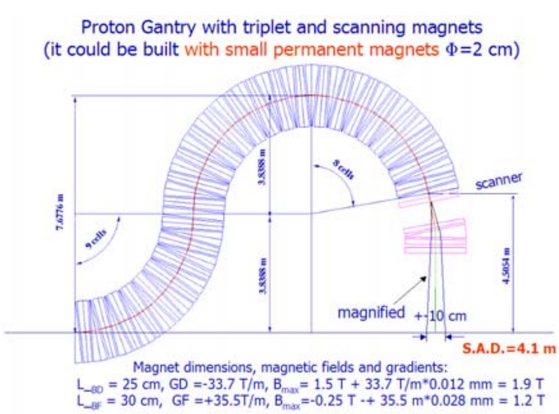

Figure 3: Proton permanent magnet NS-FFAG gantry.

A reduction in the gantry size came from raising the maximum of the magnetic field in the magnets from 1.9 T to 4.35 T and by reducing a number of cells and the magnet size to 18 and 16 cm. The maximum height is ~4.8 m. Beam amplitudes within ellipses with maximum values for the horizontal axis of x_{max}=3 mm and y_{max}=3 mm are used as initial conditions for particle tracking through the gantry for the kinetic energy range between 79-250 MeV.

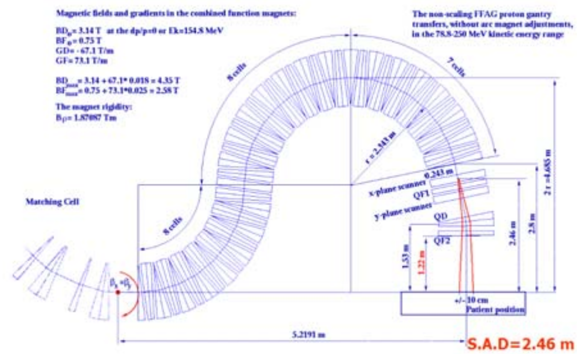

Figure 4: The superconducting proton gantry.

At the largest kinetic energies protons show at the end of the gantry offsets within –11 and + 8 mm. This is easy to adjust by the scanning magnet to the required position at the patient. The triplet magnets could adjust the spot size of the beam at the patient within a large range. The smallest values of the betatron functions are $\beta_x = \beta_y$ =0.45 m to provide the beam size of σ_{OPT}=0.23 mm ($\sigma^2_{OPT}=(\beta_{twiss}\ \varepsilon_N)/(6\ \pi\ \beta\gamma)$ where β_{twiss} is the amplitude function, ε_N is the normalized emittance, while $\beta\gamma$ are the relativistic factors). The beam size could also be enlarged by the quadrupole adjustments if it is required by the treatment.

Beam Spot Scanning

The beam scanning should be of the order of 100 Hz. The 20 cm long scanning magnet for the proton therapy has a maximum magnetic field of B_{max}=0.38T, at the highest kinetic energy of E_k=250 MeV. A cartoon showing the "parallel" beam scanning with respect to scanning above the patient is shown in Figure 7. The protons or carbon ions penetrate through the tissue with energy deposited at the Bragg peak. During passage of ions through the body due to multiple Coulomb scattering a small spread of the beam occurs. As previously shown [5,6] the spread is a function of the energy and density of the tissue. The straggling makes longitudinally a small tail behind the Bragg peak. The transverse beam size is [5] defined as $\sigma_T^2=\sigma_{MCS}^2+\sigma_{OPT}^2$, where σ_T is the total beam size, σ_{MCS} presents the effect of the multiple Coulomb scattering, and σ_{OPT} is the beam size at the patient skin defined by the beam optics and emittance. The spot scanning with "parallel" beam reduces the radiation at the layers upstream of the tumor if the beam emittance and beam optics creates $\sigma_{OPT} \ll \sigma_{MCS}$. If the beam size σ_{MCS} due to multiple Coulomb scattering is of a similar size as the beam arriving to the patient σ_{OPT} this reduction is not accomplished. If there is a possibility of making the beam size at the skin of the patient with a condition $\sigma_{OPT} \ll \sigma_{MCS}$ a reduction of the radiation could be accomplished with the spot scanning few meters above the patient. We will analyze in more details an example of proton treatment at the energy of 200 MeV. The beam size at the Bragg peak has previously been compiled [5] as σ_{MCS} =6.5 mm. The beam size of the incoming beam is σ_{OPT} =0.23 mm by the beam optics and with the normalized emittance of ε_N=0.5 π μm. A distance of 2.8 m between the scanner and the patient is selected from the third example of the isocentric proton gantry presented above. Compilation results for the transverse beam sizes of the 200 MeV proton beam [5] are presented in Table 1.

Table 1: Transverse beam sizes

β_{twiss}(m)	σ_{OPT}(mm)	σ_{MS}(mm)	σ_T(mm)	$\theta_{spr.}$(mrad)
0.45	0.23	6.5	6.504	12.06
1	0.348	6.5	6.509	11.84
10	1.101	6.5	6.59	10.73
100	3.483	6.5	7.37	7.475

The largest angle of the beam at the patient is 32 mrad when the sweeping magnet is 2.8 m above the patient. An analysis of this beam position and the step size selection is shown in Figure 9. The accumulated dose could be within 1% of the required one if the scanning steps are correctly selected as had been previously discussed in details by S. Peggs [5]. The next step should overlay the half of the previous σ_T at the tumor.

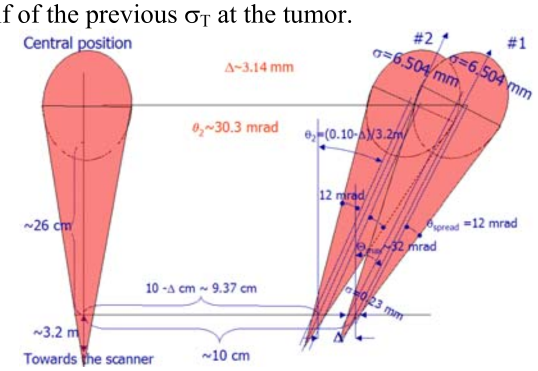

Figure 5: Step size selection during an "angle" spot scan.

This analysis shows that a careful selection of the beam size at the entrance and of the step size can reduce the effect of radiation as well as in the case of the "parallel" spot scanning. More details about this study will be soon presented elsewhere.

The carbon gantries where previously presented [2][7] and [8]. If the present proton design is scaled by the new value of the $B\rho$=4.882 Tm at the central kinetic energy of 253.1 MeV/n, for the energy range of 131.6-400 MeV/n.

The Size of the Gantry's Magnets

There are 23 non-scaling FFAG cells in the smallest proton gantry design (each 0.46 meters long). The same number of major bends-combined function magnets is required. Magnet dimensions, field *B*, gradients *G*, and maximum aperture A_p, are summarized in Table 2.

Table 2: Magnet properties

	L(m)	B(T)	G(T/m)	A_p(m)	B_{max}(T)
BD	0.18	3.1442	-67.099	+/–.018	4.35
BF	0.16	0.7541	73.101	+/-.025	2.58

A row of cells made of short (16-18 cm) combined function magnets could be made with continuous windings. If it is required, an additional layer of correction coils could be added. These do not have excessive field requirements for superconducting magnets and could be built as the coil dominated magnetic field operating at lower temperatures (2-4K). The configuration has a simple inner quadrupole surrounded by a thicker outer dipole coil and a very thin dipole coil (active shield) at a much larger radius. Other details of the design had been presented earlier [8].

SUMMARY

A few small non-scaling FFAG isocentric proton gantry designs are presented. The proton gantry with the highest point from the patient plane at 7.8 meters could be made of permanent magnets. Reduction of gantry size could come by using small superconducting magnets, which are already available. The highest point above the patient could be reduced to few meters while the longitudinal gantry length could be 5 meters. A discussion about spot scanning replacing the large bending magnets above the patient with the scanning magnets ~3m above the patient was initiated. Dramatic reduction in weight from 130 tons to less than one ton and simplified operation with the fixed magnetic field throughout the whole treatment, makes this design very interesting for future proton/carbon therapy facilities.

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