

# A DRAMATICALLY REDUCED SIZE IN THE GANTRY DESIGN FOR THE PROTON-CARBON THERAPY\*

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## Abstract

Gantries in the proton/carbon cancer therapy machines represent the major cost and are of the largest size. This report explains a new way to the gantry design. The size and cost of the gantries are reduced and their use is simplified by using the fixed magnetic field. The "new" gantry is made of a very large momentum acceptance non-scaling Fixed Field Alternating Gradient (FFAG) quarter and half arc beam lines. The gantry is made of combined function magnets with a very strong focusing and small dispersion function. Additional magnets with a fast response are required to allow adjustments of the beam position for different energies at the beginning of the gantry. Additional strong focusing magnets following the gantry have also to be adjustable to provide required spot size and radial scanning above the patients. The fixed field combined function magnets could be made of small permanent magnets for the proton machine, or of the high temperature superconductors or superconductors for the carbon machine, reducing dramatically the size.

## INTRODUCTION

The cancer hadron therapy facilities exist today in a large number of medical facilities all around the world [1]. Many more are being commissioned or in process of being built. This progress is mostly due to multiple advantages of the hadron cancer therapy with respect to any other radiation methods [2]. A major challenge in present and future hadron therapy facilities is the beam delivery system. At least one of the patient delivery room is equipped with an isocentric gantry system. The gantry role is to deliver a precise ion dose to the patients with very good reliability and stability. The larger cancerous tumors require transverse position scanning at different beam energies and an angle variation around the patient provided by gantry rotation. Additional constraints are described in a presentation of the newest state of the art gantry for carbon hadron therapy facility in Heidelberg [3]. This report examines a new way for the gantry design by using a concept of the non-scaling FFAG [4] with very strong focusing with very small dispersion (few cm) obtained by small combined function magnets. The momentum acceptance is very large mostly due to small dispersion function (it could be  $\delta p/p \sim \pm 50\%$  or kinetic energy range 68-400 MeV/n). More details about this concept and solutions for hadron acceleration for medical facilities are described in the other report at this conference [5]. The weight of the gantries for the hadron

therapy ranges is in the order of 600 tons [3]. The non-scaling FFAG concept might be able to reduce overall weight due small magnets. The fixed field magnets have transversely linear variation of the magnetic field and they could be superconducting or high temperature superconductors.

## THE NON-SCALING FFAG

A revival of the scaling FFAG accelerators, a concept developed in fifties, is evident especially in Japan. The magnetic field varies transversely as  $B \sim (r/r_0)^k$  where the  $k$  value could be very large ( $\sim 1000$ ) and the negative bend is a third of the major bend. The non-scaling FFAG design has been studied and analyzed during the last ten years and building of a proof of principle machine is in progress. Interest for the non-scaling FFAG comes from a reduction of the aperture size with respect to the scaling one and a smaller value of the opposite bend angle. The non-scaling FFAG is made of fixed field combined function magnets with a linear transverse variation of magnetic field.

### The Basic Cell of the Gantry

An example a gantry basic cell with betatron functions is shown in Fig. 1.

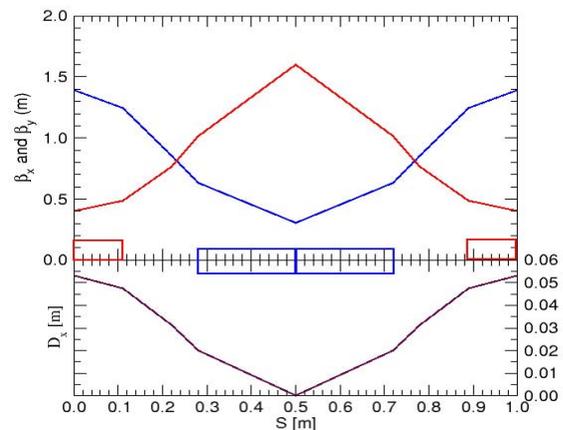


Figure 1: Betatron functions in one gantry cell.

The central magnet is a defocusing combined function magnet with minima of dispersion and horizontal  $\beta_x$  function at the middle. The minimum emittance lattice for the light sources requires very similar conditions.

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Dispersion function throughout FFAG lattice retains very small values. The large momentum acceptance and small momentum offsets are a consequence of the small dispersion:  $\Delta x = D_x \delta p/p$ . If the aperture is of the order of  $\Delta x \sim \pm 25$  mm and dispersion is  $D_x \sim 5$  cm, the momentum offset could be  $\delta p/p \sim \pm 50\%$ . 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 bending angles are presented in Fig. 2. The middle of the dipole is selected as a place for the input and output of the gantry.

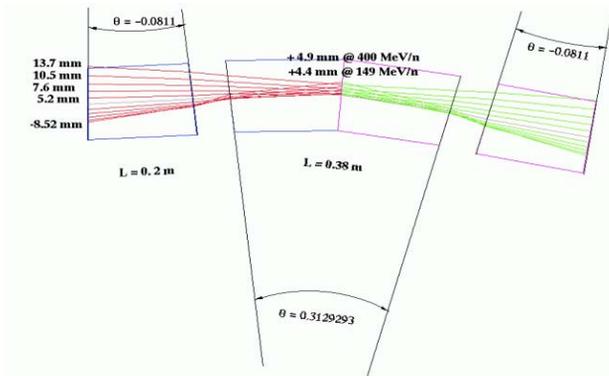


Figure 2: The basic cell of the gantry made of non-scaling FFAG combined function magnets.

The offsets at the end of the cell are obtained from the Polymorphic Tracking Code (PTC) [5] in a kinetic energy range between in momentum  $\delta p/p < \pm 30\%$  or in kinetic energy range 131.6-400 MeV/n. The dispersion function and the slope are set to zero at the beginning of the gantry in the middle of the required momentum range. The orbit offsets are smallest at the middle of the major bending magnets.

### Gantry Design

This design has been submitted for patent approval.

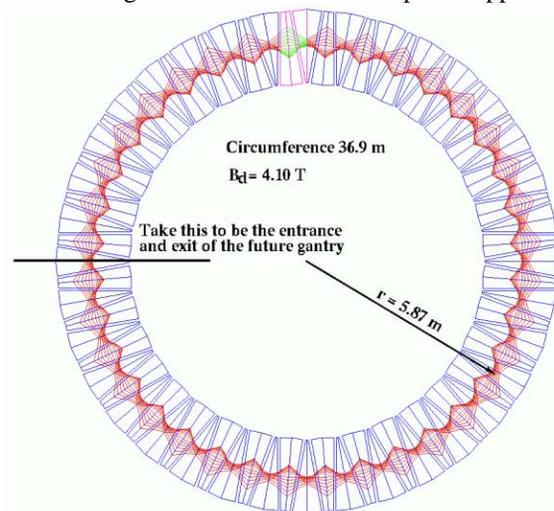


Figure 3: Non-scaling FFAG ring to be used for the gantry design. Orbit offsets are magnified 25 times.

A design of the gantry requires a ring with zero dispersion and slopes at the middle of the major combined function dipoles as presented in Fig. 3. Stable orbits for carbon ions are found within  $\delta p/p = \pm 30\%$ . Symmetric values of the largest orbit offset, at the focusing quadrupoles, throughout the energy range are obtained by optimization of the magnet gradients. Symmetry of the ring is broken at the middle of the major bending magnet (this is marked in Fig. 3). The beam line –the gantry takes an orbit swing towards the patient after a specific number of cells. This depends on the required geometry and available space of the patient gantry treatment room, and on a required distance between the end of the gantry and the patient. The radial scanning and orbit correction is provided at the end of the gantry. A construction of the gantry follows the ring solution and  $1/4$  of the ring is used in the first example for the beginning of the gantry, as presented in Fig. 4.

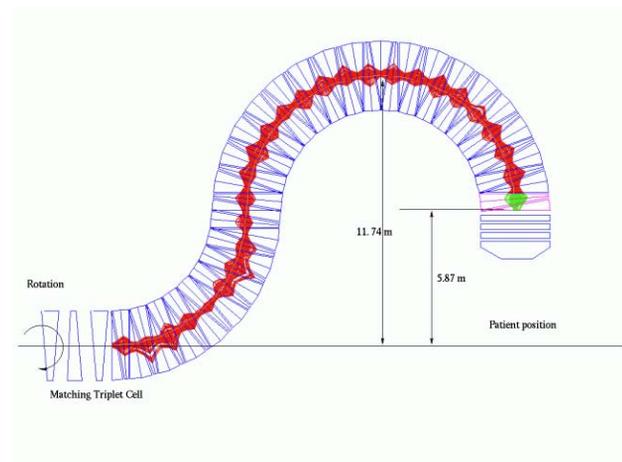


Figure 4: A gantry made of  $1/4$  and  $1/2$  combination of the ring presented in previous figure. Orbits are obtained by tracking with the PTC. Orbit offsets are magnified 25 times. The betatron functions at the central momentum are shown in Fig. 5.

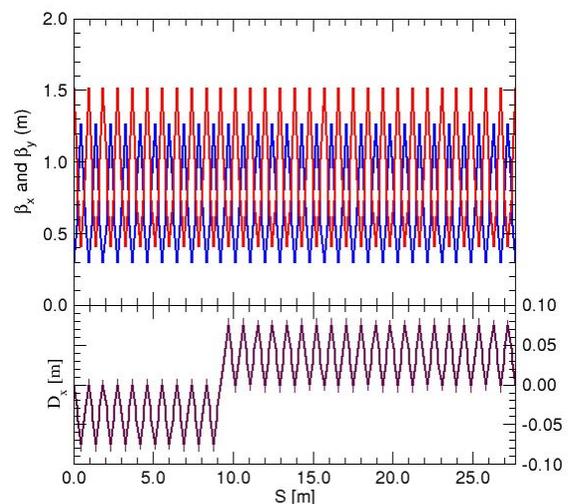


Figure 5. Betatron functions in the gantry.

### Particle Tracking in the Gantry

The non-scaling FFAG accelerator rings, previously reported [7], have momentum range of  $\delta p/p = \pm 50\%$ . The additional constraints of zero dispersion  $D_x=0$  and  $D_x'=0$  for  $\delta p=0$  make the available range smaller for the same size. The additional constraints required larger value of the negative bend angle. The magnetic field of the gantry's magnet is fixed for the whole required energy range for the patient treatment simplifying the operation. The initial conditions (shown in Fig. 5) during particle tracking through the gantry, for both horizontal and vertical phase space  $x, x'$  and  $y, y'$ , assumed both amplitudes to be  $x_{max}=3$  mm and  $y_{max}=3$  mm. Tracking results showed that for the negative part of the momentum range at  $\delta p=-30\%$  particle offsets at the end of the gantry were too large. The energy range of the presented gantry is  $-25\% < \delta p/p < 30\%$ .

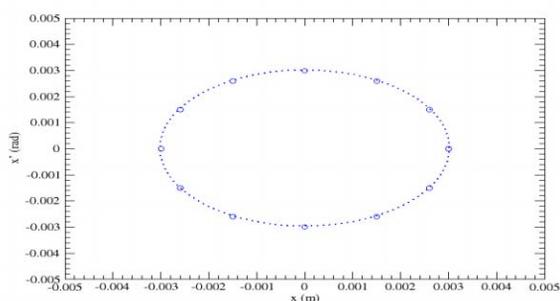


Figure 6: The initial conditions for  $x$  and  $x'$  phase space.

### Matching the Gantry to the Accelerator

An additional independent triplet cell is added at the entrance of the gantry to provide matching at the central momentum from the accelerator.

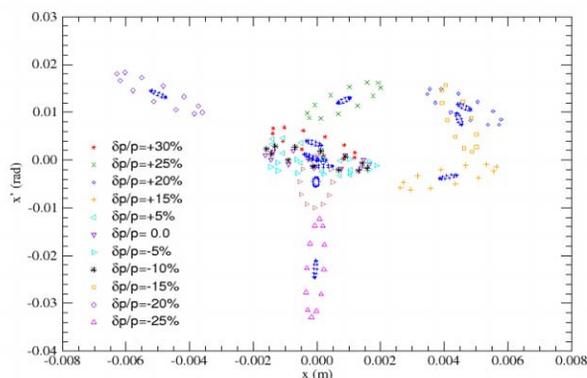


Figure 7: Particle tracking for different momenta.

At the point of entrance to the gantry's rotational part, the betatron functions are equal and the beam is round. In addition the slopes of the betatron functions and dispersion are equal to zero. The zero dispersion is obtained by adjusting the bending angles of the triplet's combined function elements. Two quadrupoles rotate together with the gantry and are on the axis of rotation.

They provide matching to the gantry's input betatron functions for each angle of the gantry.

### The Size of the Gantry's Magnets

There are about 24 to 30 non-scaling FFAG cells (0.92 meters long) used in the gantry design. 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 1.

Table 1: Magnet properties

	L(m)	B(T)	G(T/m)	$A_p$ (m)	$B_{max}$ (T)
BD	0.38	3.2-4.1	52.6	$\pm .008$	4.5
BF	0.40	1.98	54	$\pm .015$	2.8

The 40 cm long magnets of this size 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). A configuration has a simple inner quadrupole surrounded by a thicker outer dipole coil and a very thin dipole coil (active shield) at much larger radius. The cryostat has an  $OD=170$  mm, and around the whole magnet is a thin warm iron shell to take care of the field not caught by shield coil. The estimated weight of the magnets is in the range 45-75 kg/m or average  $\sim 50$  kg/m. The  $\sim 30$  meter long gantry beam line is  $\sim 1500$  kg. This does not include the weight of the support system, but it is to be compared to the 135 tons of the "transport components" [3]. The magnets could be built as super ferric magnets with a high temperature superconductor (HTS) racetrack coil distribution. They operate at temperatures of 35-45 K commercially available cryosystems without need for liquid helium.

### SUMMARY

The non-scaling FFAG gantry, with the fixed magnetic field for the whole energy range of the carbon/proton cancer treatment is presented. Small magnet sizes should reduce the cost and weight and ease the gantry operation. The transverse and final focus scanning system is assumed to be at end of the gantry transport above the patient. This work has been submitted for the patent approval.

### REFERENCES

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