# INNOVATIVE SUPERCONDUCTING NON SCALING FIXED FIELD ALTERNATING GRADIENT ISOCENTRIC GANTRY FOR CARBON CANCER THERAPY

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*Abstract*

Numbers of proton/carbon cancer therapy facilities in recent years is rising fast due to a clear advantage with respect to the other radiation therapy treatments. Cost of the ion cancer therapy is dominated by the delivery systems. An update on a design of the carbon and proton isocentric gantries is presented; using the non-scaling alternating gradient fixed field magnets (NS-FFAG). The magnet size and weight is smaller with respect to other magnets used today in cancer therapy treatment. The weight of the transport elements of the carbon isocentric gantry is estimated to be 1.5 tons to be compared to the 130 tons weight of the top-notch Heidelberg facility gantry. For the carbon transport elements estimated weight is 1500 kg.

## INTRODUCTION

This is a proposal to use direct wound iron free superconducting magnets in medical application - cancer therapy facilities, specifically for the isocentric gantries. A motivation comes from a reduction of the gantry size, weight, cost, simplicity in operation and reduction of electrical power consumption. At first, we describe essentials of the ion-cancer therapy method, showing its advantage over other radiation methods used. The beam delivery systems, like isocentric gantries, as well as other major components of an ion therapy facility follow. We then describe a new concept of the Non Scaling Fixed Field Alternating Gradient (NS-FFAG) accelerators. The NS-FFAG concept applied in isocentric gantries is the next section. Here we describe the magnet properties for isocentric gantries as well as the beam transport characteristics. The superconducting magnet design is shown in the last section ending with a summary.

## ION CANCER THERAPY

Cancer is the second most frequent cause of death; in Europe one of three will confront some form of cancer during their lifetime. Forty five percent of the cancer cases can be treated mainly by surgery and/or radiation therapy. By success rate ion therapy is the second only to surgery. Proton and carbon cancer therapy are expanding very fast with a number of patients growing exponentially. The hadron-ion therapy has advantage with respect to other radiation methods primarily because the radiation can be deposited directly (the Bragg peak) where the tumor is located and damage is not done to the skin. Also, because of the Bragg peak, the radiation is localized patient inducing unnecessary damages to the cells. The cancer treatment requires very often radiation treatment. In the ion cancer therapy energy of the beam must be a variable quantity; as a location of the Bragg peak needs to be exactly at the tumor depth. The transverse scanning-preferably spot scanning with respiration adjustments with 2D conformal treatment is another necessity. The scanning range should cover an area of at least ±10 cm an energy scanning ±20 %. The intensity modulation of up to ~$10^3$ in current is preferable.

### S.A.D. Definition

The maximum dose to the patient surface relative to the dose in the Spread Out Bragg Peak (SOBP) increases as the effective source-to-axis distance (SAD) decreases. Large SAD's could be achieved in a fixed horizontal beam but not in a gantry beam lines. A smaller gantry with a physical outer diameter of less than 2 meters may have important cost implications. Such a gantry would require magnetic optics to ensure that the effective source-to-axis distance is large enough to provide adequate skin sparing.

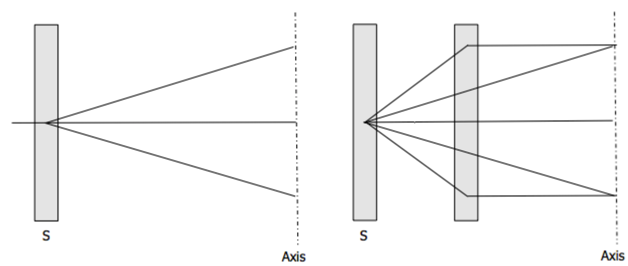


Figure 1: The Source-to-axis (SAD) distance: from a single point at the middle of the scanning magnet. A possibility for an infinite SAD with two scanning magnets.

## NS-FFAG CONCEPT

The author had come to a concept of NS-FFAG by using the relationship between the orbit offsets and dispersion function: $\Delta x=D\bullet\Delta p/p$, where $\Delta p/p$ is the difference in momentum. To reduce the orbit offsets to ±4 cm range, for momentum range of $\Delta p/p$ ~ ± 50 % the dispersion function $D_x$ has to be of the order of $D_x$~ 4 cm, as 4/0.5= 8 cm. During the fifties a concept of scaling FFAG was developed theoretically [1-3] as well as by building electron models of the proton accelerators. The fixed magnetic field has non-linear $B=B_o(r/r_o)^k$ transverse dependence. The transverse orbits in the magnets follow the momentum change as $p_o=eB_or_o$, $p=eB_rr$, $p=p_o(r/r_o)^{k+1}$.

There are many very advantageous properties of the scaling FFAG's: like zero chromaticity for all energies, orbits are parallel to each other for each energy, both tunes are fixed for all energies, very large available momentum with very large momentum acceptance. Unfortunately they require relatively large apertures. The author's first attempt to design a smaller aperture FFAG with large momentum range acceptance was by using the synchrotron light source lattices adjusting magnet lengths to obtain very small dispersion function $<H>$. Disadvantages of the NS-FFAG's are tunes, chromaticities, and path length variation with momentum. Advantages are very small aperture - small magnets. A comparison between the NS-FFAG and scaling FFAG is shown in Fig. 1.

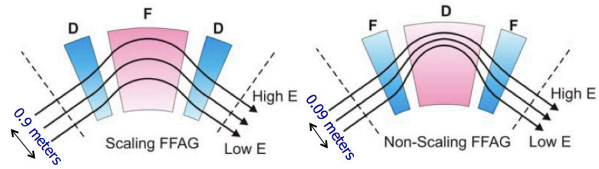


Figure 1: A comparison between scaling FFAG and NS-FFAG's. The aperture is at least an order of magnitude smaller but orbits are not parallel to each other.

## NS-FFAG ISOCENTRIC GANTRY

A major motive for choosing the NS-FFAG gantry solution for isocentric gantry application is dramatically smaller size of the magnets with respect to the existing designs and operating gantries in the world. This becomes more clear in the carbon cancer therapy where the bending of the 400 MeV/u beam becomes very difficult with the regular warm magnets ($B$~1.6 T) as the $B\rho$=6.347 Tm is to large making the radius of curvature of the order of 4 m.

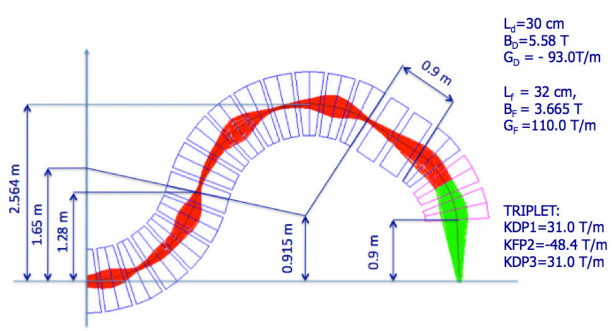


Figure 2: Superconducting NS-FFAG isocentric gantry for carbon cancer therapy with the height less than the smallest operating proton isocentric gantry.

We will present three latest different NS-FFAG designs in carbon ion application: the first one is with adjustable field for energies required with a goal of reducing the size of the radius to be less than the smallest operating proton isocentric gantry as shown in Fig. 2.

The maximum dipole field in the 30 cm long defocusing combined function magnets is $B_{dmax}$=5.58 T with a gradient of $G_d$=-93 T/m. The maximum dipole field in the 32 cm long focusing combined function magnet is $B_{fmax}$=3.7 T with a gradient of $G_f$=110 T/m. For very small apertures with a radius of ~5-7 mm gradients of 110 T/m require field at that radius of 0.8 T. Orbit offsets in the defocusing combined function magnet are always smaller then at the focusing one.

Next example of the NS-FFAG isocentric gantry is with a larger height as it accepts momentum range $\Delta p/p$=±25 % or in the kinetic energy 161 MeV/u < $E_k$ < 400 MeV/u.

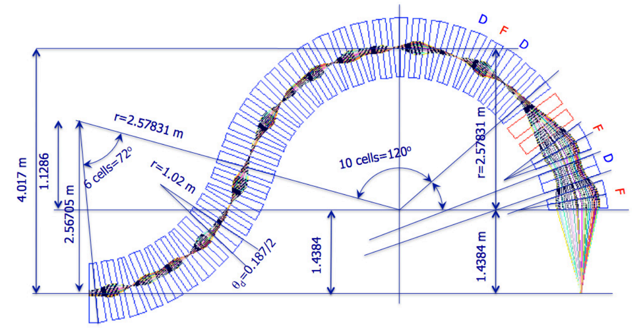


Figure 3: Superconducting NS-FFAG isocentric gantry for carbon cancer therapy with $\Delta p/p$=±25%.

The magnet lengths for the large momentum acceptance isocentric gantry are 19 cm. The maximum dipole field in the focusing combined function magnet is $B_{fmax}$=0.625 T with a maximum gradient of $G_f$=200 T/m, with maximum orbit offsets of 9.5 mm. The maximum field in the focusing combined function magnet is $B_f$= 0.625+203*0.0095 = 2.55 T.

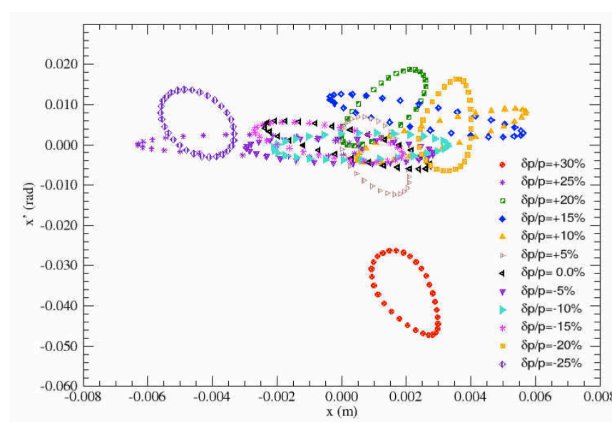


Figure 4: Orbit offsets ±6 mm at the end of the superconducting NS-FFAG isocentric for carbon cancer therapy in the x and x' phase space.

The maximum dipole field in the defocusing combined function magnet is $B_{dmax}$=4.97 T with a maximum gradient in the focusing combined function magnet is $G_d$=-158 T/m, with maximum orbit offsets of 9.5 mm. The maximum field in the focusing combined function magnet is $B_d$=4.97+-(157)*(-0.0072)=6.1 T. The gradients and dipole fields in magnets in this example are equal for each magnet. This produces small offsets at the end of the gantry as shown in fig. 4.

The third example of the superconducting isocentric gantry for the carbon cancer therapy is capable of accepting the momentum range $\Delta p/p$ =±20 % or in the kinetic energy 196MeV/u<$E_k$<400 MeV/u. The advantage of this design is that all energies pass through the magnets with a fixed magnetic field and finish without any offsets with a focal point at the center of the scanning magnet as shown in fig. 5.

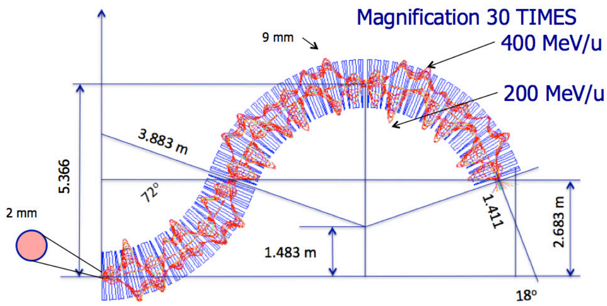


Figure 5: Superconducting NS-FFAG isocentric gantry for the carbon cancer therapy with a focal point at the middle of the scanning magnet.

The gantry is made of superconducting combined function magnets but with adjusted gradients though it. The maximum dipole field in the 17 cm long focusing combined function magnets is $B_{fmax}$=0.403 T with a gradient of $G_f$=212 T/m. The maximum dipole field in the 16 cm long defocusing combined function magnet is $B_{fmax}$=4.76 T with a gradient of $G_d$=-174 T/m. The betatron function $\beta_x$ and the dispersion $D_x$ are shown in Fig. 6.

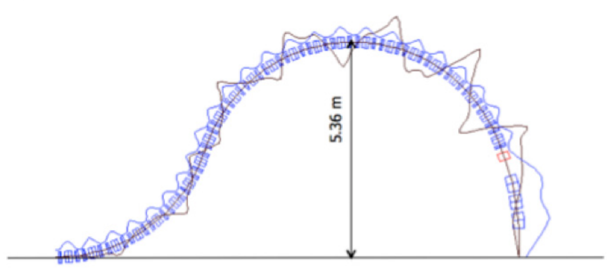


Figure 6: Betatron function $\beta_x$ and the dispersion function.

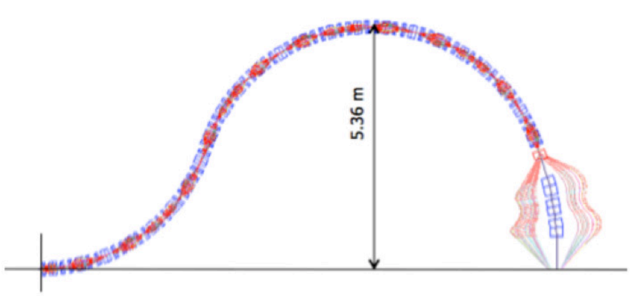


Figure 7: Tracked particles with the scanning magnet turned on. Orbits are magnified.

Three modules make a complete transport line of the gantry - each bending 72°. Each module is made of eight triplet magnets DFD with drifts between them. The gradients of four focusing and defocusing combined function magnets are variables: $D_1F_1D_2F_2D_3F_3D_4F_4D_5$ .

The scanning system is made of two scanning magnets in opposite directions in the transverse plane. A solution with the beam size control by the triplet magnets downstream of the scanners is shown in Fig. 7.

## MAGNET DESIGN

Superconducting combined function magnets designs are with the superconducting cable direct wound to the beam pipe directly defining the magnetic field. This makes them dramatically lighter as iron is not necessary. The Advanced Magnet Lab (AML) had already designed and built similar combined function magnets [4] as shown in Fig. 8.

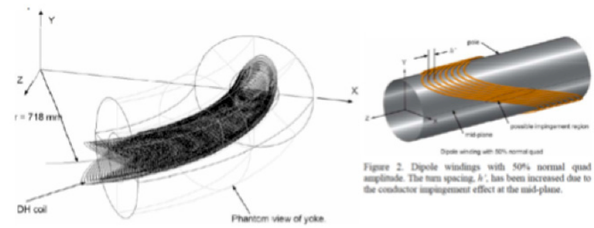

Figure 8: AML superconducting combined function magnet.

At Brookhaven National Laboratory, Brett Parker did a preliminary design of the combined function magnet as shown in Fig. 9.

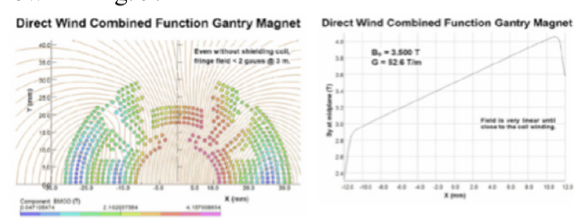


Figure 9: BNL preliminary design of the combined function magnet. The fringe magnetic field at 3m is less than 3 G.

## SUMMARY

An update on the superconducting NS-FFAG isocentric gantry for the carbon cancer therapy is presented. A reduction in cost and simplified operation in future carbon cancer therapy facilities can be achieved. The cost and weight are reduced for two orders of magnitude and simplification in operation is expected.

## REFERENCES

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