LARGE MOMENTUM ACCEPTANCE SUPERCONDUCTING NS-FFAG GANTRY FOR CARBON CANCER THERAPY*

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Abstract
Carbon cancer radiation therapy has clear advantages with respect to the other radiation therapy treatments. Cost of the ion cancer facility is dominated by the delivery systems. New superconducting carbon ion gantry designs, based on the Non-Scaling FFAG (NS-FFAG) principle, are presented. The magnet size and weight is dramatically smaller with respect to existing gantries. 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 of the top-notch gantry in the Heidelberg ion therapy facility.

INTRODUCTION

A number of ion cancer therapy facilities are rapidly rising due to a clear advantage to any other radiation treatment of the cancerous tumors. This is due to a very localized energy deposition of ions at the "Bragg peak-as suggested by R. Wilson 1946 [1]. The ion cancer facility starts with an ion source, accelerator, and the delivery system (major cost). The Coulomb scattering and struggling define the final beam spot size–penumbra at the tumor. The best results in the cancer treatment are obtained by use of the spot size "pencil" beams with the transverse and longitudinal scanning technique. Usually the treatment time is of the order of couple of minutes. In proton cancer therapy the minimum beam volume size-voxel at the tumor (for the maximum distance of 27 cm in the patient) is ~715 mm^3, while for carbon ions it is ~14 mm^3. The transverse intrinsic spot for 206 MeV protons is ~11 mm, while for carbon ions is 2.9 mm. Carbon ions have advantage in the patient treatment, as they are more precise. Normal angle of ion arrival to the patient reduces the skin radiation. The Source to Axis Distance (SAD) parameter in the cancer treatment shows an optimum with SAD=∞ (beam arrives with normal angle of incidence to the patient skin).

A recommendation from the Workshop on Ion Therapy [1] in Bethesda MD (organized by Department of Energy (DOE), National Cancer Institute (NCI), National Institute of Health (NIH), and Department of Health & Human Services DHHS) was that the carbon or other ion therapy facility should include rotatable gantry, possibly made of superconducting magnets (to be able to reduce the size and cost) and with a large momentum acceptance.

This is a report on a large momentum acceptance superconducting NS-FFAG gantries. They have many advantages like: reduced size and weight, large momentum acceptance (200-400 MeV or 100-200 MeV/u etc.), fixed magnetic field, with the angle of incidence always normal to the patient (SAD=∞), and easy to operate, as the magnetic field is fixed during the treatment. We present two examples: the isocentric gantry and an eccentric gantry proposed, by Marco Pullia (CNAO-Italy). Both gantries have a large momentum acceptance – for one fixed magnetic field setting they can operate covering a large kinetic energy range (either between 200-400 MeV/u or 100-200 MeV/u). The normal angle of incidence is obtained by use of two scanning dual plane scanning magnets.

NS-FFAG PRINCIPLE

In recent years there has been a clear revival of the previous concept of the FFAG's, developed mostly in the 1950's. A concept of the NS-FFAG is based on strong focusing principle with a small dispersion function. The relationship: $\Delta x = D\ \delta p/p$ shows that the beam offset is small even for the large $\delta p/p$. Major differences between the NS-FFAG's with respect to the scaling FFAG's are variations of the tunes, time of flight, dispersion and amplitude functions with a momentum as shown in a gantry cell in Figure 1.

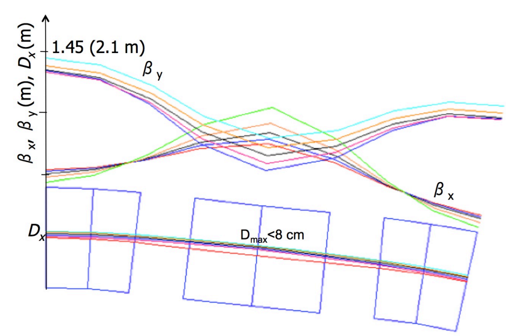

Figure 1: Betatron and dispersion functions in a gantry cell for the treatment energies.

BEAM DELIVERY - ION GANTRIES

The delivery system is the most expensive part of the ion cancer therapy facilities. Figure 2 shows the top-notch proton gantry at PSI in Switzerland. The scanning magnets are placed usually in front of the large 90°-bending magnet above the patient. This is to allow for the SAD=∞. To avoid the large magnet above the patient (for the proton gantries its weights is ~60 tons) a scanning system is placed above the patient. This solution (submitted to the Patent Office) not only removes the necessity of the large magnet but also always provides normal angle of incidence to the patient skin. A proton gantry design is shown in Figure 3.

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Figure 2: The proton top-notch proton therapy facility at PSI, with the large bending magnet at the end.

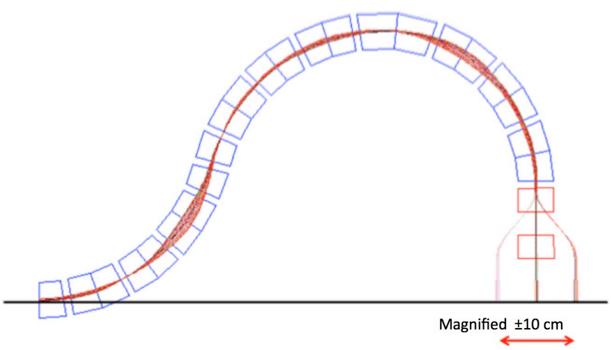

Figure 3: Our proton gantry design eliminates need for the large dipole and provides always 90° angle to the skin.

The carbon gantry operating at the Heidelberg ion cancer facility is of a large size as at the time when it was built and designed, relied on the existing safe warm magnet technology. The maximum kinetic energy of the carbon ions in Heidelberg is 430 MeV/u with a corresponding $B\rho$=6.624 Tm. For the magnetic field of 1.6 T produced by the warm magnet the radius of a curvature is R=4.1 m. To better perceive the size of this gantry in figure 4 a visitor is shown. The NS-FFAG superconducting gantry design is shown together with the Heidelberg gantry in Figure 6. A large number of small superconducting combined function magnets will reduce the size of the gantry. The number of magnets is 31, including the two scanning and the triplet magnets for defining the beam size. The superconducting magnets are made by superconducting wire wound directly onto the beam pipe, without using any iron. This reduces the weight. They have a small aperture r~3 cm. The dipole field in the defocusing magnet is B_{yD}= 4.55 T with gradient of G_d= −90 T/m, while the focusing magnet has opposite bending field of B_{yF}=-0.385 T with a gradient of G_F=150 T/m. The length of the focusing magnet is 20 cm while the defocusing one is 26 cm long. The bending angle of the defocusing magnet is θ_D=0.233 rad while the focusing combined function magnet has an angle of the opposite sign of θ_F=−0.152 rad. The magnetic field in the defocusing combined function magnet is equal to: $B_{DTOT} = B_D + (G_D * x)$, while for the focusing magnet it is: $B_{FTOT} = B_F + (G_F * x)$. The maximum magnetic field in the defocusing magnet is equal to B_{Dmax}=4.55 T at the lowest energy and B_{Dmax}=5.63 T at the highest energy (this is a combination of the bending field plus gradient multiplied by the orbit offsets for energies other than the central energy). The maximum field in the focusing combined function magnet is, for the lowest energy, B_{Fmax}=2.34 T, while for the highest energy it is B_{Fmax}=-3.4 T.

Figure 4: The Heidelberg gantry with a visitor. The large 45° magnet is shown above.

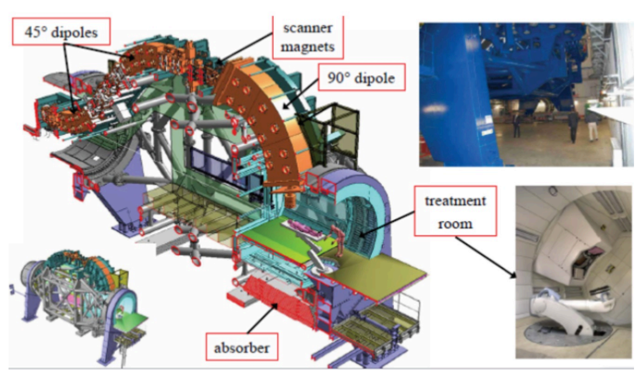

Figure 5: The whole structure of the Heidelberg carbon gantry with detail of the treatment room on the right side.

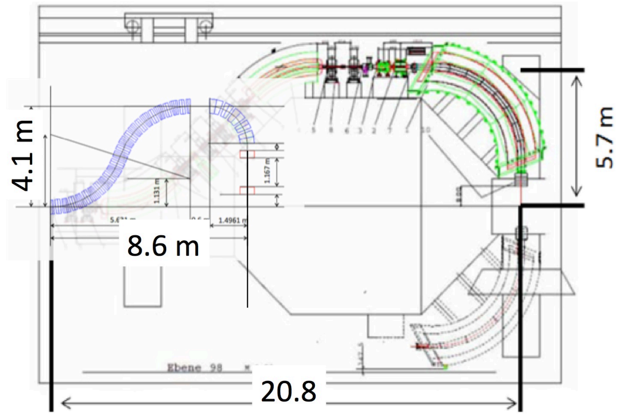

Figure 6: The carbon ion gantry with superconducting magnets is shown together with the Heidelberg gantry.

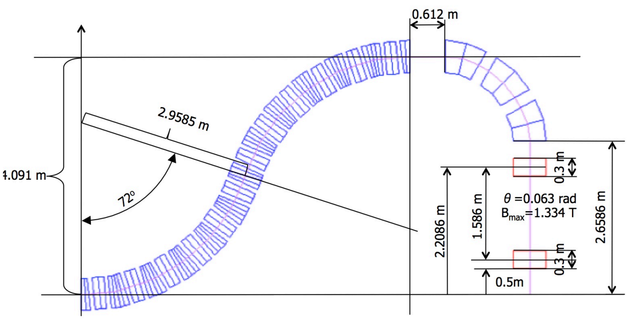

Figure 7: The carbon ion superconducting fixed field isocentric gantry layout. The scanning magnets are 30 cm long with a maximum required magnetic field of 1.35 T for the kinetic energy of 400 MeV/u (to be able to bend the beam ±10 cm at the patient (a distance between the two magnets is 1.566 m).

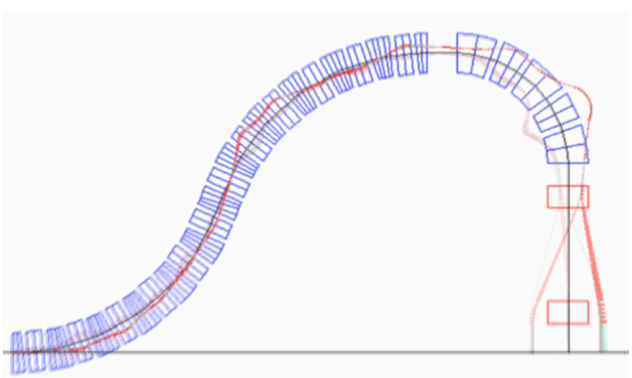

Figure 8: The fixed filed superconducting carbon isocentric gantry with beams of 202, 270, and 400 MeV/u. With the magnified (x10) scanning ±10 cm positions.

Two examples of the superconducting carbon ion gantries: the isocentric gantry (the same as in Fig. 6) and the eccentric one are shown in Figures 7&8, and Figures 9&10, respectively.

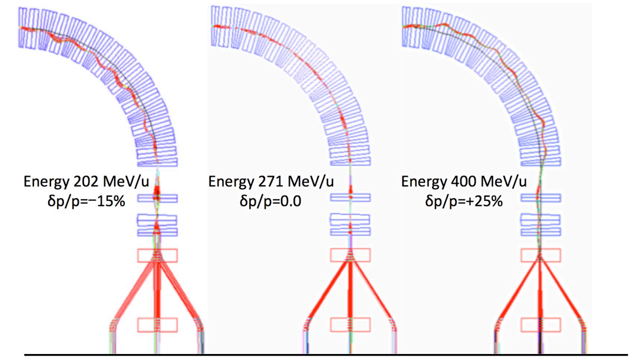

Figure 9: The 90°-superconducting gantry shown with carbon ions passing with three energies 202, 271, and 400 MeV/u under the fixed magnetic field, with scanning magnets. Positions of ±10 cm at the patient are magnified.

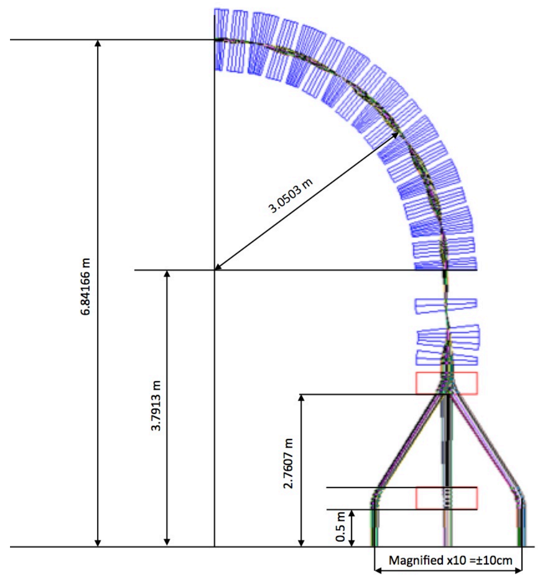

Figure 10: The layout of the 90° superconducting fixed field carbon gantry designed to fit to the CNAO eccentric gantry.

SUMMARY

The NS-FFAG superconducting gantries for carbon ion design with isocentric and eccentric case are presented. The gantry design fulfills a requirement of the very large momentum acceptance with a fixed magnetic field. Major advantages of the proposal are reduced size, weight, and easier operation as the magnetic field is fixed during the treatment time. New scanning system with SAD=∞ is presented. It reduces the size of the gantry, as it does not need the very large magnet at the end. The triplet magnets, placed above the scanning system, are used to define the beam size at the patient.

REFERENCES

[1] Robert Wilson, Radiology **47**, 487 (1946).
[2] Summary Report, Workshop on Ion Therapy, Bethesda, January 11-13, 2013.

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