

A Superconducting Magnet for a Beam Delivery System for Carbon Ion Cancer Therapy

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Abstract—The therapy of cancer (hadrontherapy) with accelerated particle beams is an out coming method that allows an accurate and effective treatment of several kinds of tumors minimizing the irradiation of the surrounding tissues and avoiding intercepting vital organs. The deep tumors are usually treated with proton or ion beams by using a rotating gantry that allows the 360° irradiation around the patient. For carbon ion hadrontherapy, being the energy in the order of 400 MeV, huge gantries structure are required if resistive magnets are used. The size of a ion gantry can be of the order of 10 m in diameter and 100 tons in weight. An alternative solution has been studied, involving the use of a cryogenic free superconducting magnet. The magnet consists of ten 90° bent superconducting dipoles wound with an aluminum stabilized Cu/NbTi conductor and mechanically supported by an aluminum alloy structure. An optimization of the magnetic design is running by using genetic algorithms with the goal to obtain a field homogeneity within 0.2% in a region of $60 \times 200 \text{ mm}^2$ along the beam path. Furthermore, an accurate study of the winding technique is in progress. The magnet will be kept at 4.5 K by a cooling system based on two cryocoolers operating alternately in steady state or together during the cool down and the electrical transients. The current is supplied via a couple of HTCS current leads always connected. In the paper, a description of the design is given.

Index Terms—Aluminum stabilized conductor, cryogenic free, optimization, superconducting dipole magnets.

I. INTRODUCTION

THE CONSTRUCTION of an ion gantry puts serious technical difficulties if considering the conventional technology. The high magnetic rigidity of ion therapy beams (over 6 Tm), dictated by desirable penetration range of ions in the patient body (around 25 cm), requires carbon ions of 4.5 GeV energy. In order to fulfill this requirement the resulting gantry weight and dimensions exceed some practical limits, especially when considering a dedicated installation in a common hospital environment [1], [2]. This consideration suggested the idea of a more compact gantry equipped with a superconducting magnet.

II. DESIGN GUIDELINES

In order to avoid the negative impact on medical environment of devices involving cryogenics liquid (complex operations, use of large dewars), a superconducting bending dipole for

the gantry should be cooled with a fully automatic cryocooler. Cool-down, normal operation and warm-up may be performed in simple way by nonspecialized operators. Of course the impact of this choice on the dipole design is quite strong:

- I) The relatively low available cooling power at 4.5 K (3 W) imposes the design of a low loss coil, cooled indirectly through a medium. We chose to follow the design criteria of the classical indirect cooled thin solenoid for detectors in colliders, involving aluminum stabilized conductors.
- II) When the field is ramped up/down during a therapy cycle, we have dissipations inside the coil. On one side this dissipation shall be minimized by operating on the conductor layout; on the other side the thermal capacity of the winding shall be so high to limit the temperature increase within the critical temperature.
- III) As superconducting material we chose NbTi, because, in spite of the relatively low critical temperature (9.2 K at 0 T), it is the most known and reliable material involved in superconducting magnet technology.
- IV) Current leads are sources of dissipation. The maximum current was set to 1 kA. This is also the minimum current in order to limit the number of turns and then the coil inductance, which plays an important role in case of quench.
- V) The field uniformity required in the large space, in which the ion beam can move, requires a special winding layout. The use of several bent racetracks leads to a large number of geometrical parameters for the field optimization.
- VI) Since the winding is made of aluminum stabilized conductor, it is highly recommended to use aluminum alloy in the mechanical supporting structure. In this case we can have a containing structure with good thermal properties and the same thermal contraction as the winding.

The above mentioned points constituted the basic guidelines, around which the design was performed.

III. WINDING LAYOUT AND DESIGN CONSTRAINS

The goal of this work was to design a superconducting dipole able to generate a magnetic field of 4 T along a circular path, 1.6 m in radius and 90° angularly. The field uniformity in a region of rectangular cross section $200 \times 60 \text{ mm}^2$, describing the beam path and centered at $r = 1.6 \text{ m}$, should be better than 2‰. Such a magnet has to be composed by a certain number of bent racetrack coils, shaped as shown in Fig. 1(a).

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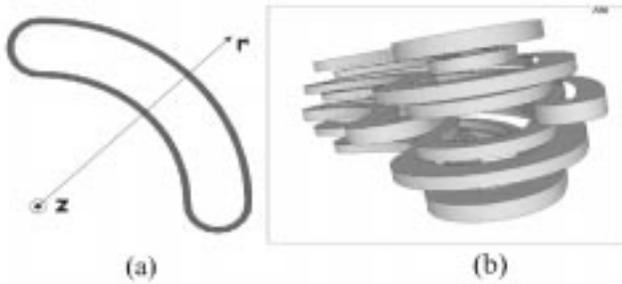


Fig. 1. (a) Shape of the bent racetrack coils composing the dipole. (b) Perspective 3D view of the superconducting dipole.

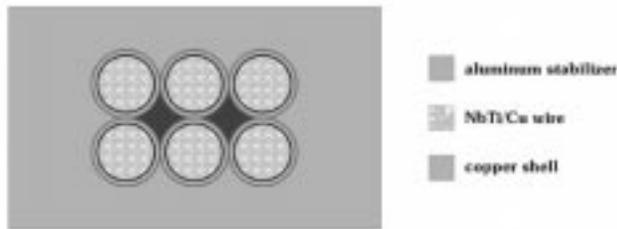


Fig. 2. Layout of the conductor to be used in the winding.

A preliminary analysis allowed us to define the overall shape of the superconducting magnet, shown in Fig. 1(b): the coil on the equatorial plane was split into two independent coils, in order not to occlude the path to the ion beam, and the total number of bent racetrack coils was fixed to 10, symmetric with respect to the equatorial plane. Subsequently, an exhaustive optimization analysis was carried out, in order to define completely the magnetic design of the superconducting dipole. The starting point of that analysis was the identification of all the constraints that tie the design; then the final optimization, performed using Finite Element analyses driven by a genetic algorithm, led to a best design which fulfills all the project requirements.

The magnetic optimization analysis was performed in two steps. The first step consists in 2-D axis-symmetric analysis performed on a section sufficiently far from the coil heads. The aim of that analysis is to maximize the field uniformity in the region $200 \times 60 \text{ mm}^2$, keeping into account all the design requirements listed before. The second step, (3D Layout still on run), consist in shaping the dipole heads in such a way to optimize the tracking of the ion beam along its path.

Design Constraints

Central Field: The nominal central field should be kept as near as possible to 4 T.

Operative Current: The maximum operative current should be kept within 1000 A.

Peak Field in the Winding: The peak field in the winding must be $B_{peak} \leq 5.5 \text{ T}$.

Operative Current Density: The existing or under construction magnets all have current densities of the same order of magnitude, $(0.3 \div 0.8) \cdot 10^8 \text{ A/m}^2$. To be conservative, we decided to choose a current density at the higher limit of that range, but anyway inside it: $(0.6 \div 0.7) \cdot 10^8 \text{ A/m}^2$.

TABLE I
CONDUCTOR CHARACTERISTICS

Conductor overall dimensions	$5.0 \times 3.0 \text{ mm}^2$
Strand Diameter	0.8 mm
Filament Diameter	$<10 \mu\text{m}$
Cable Dimension	$2.4 \times 1.6 \text{ mm}^2$
Number of Strands	6
Strand twist pitch	5 cm
Cu/Sc ratio	1.5
Critical Current at 4.2K and 5T	3400A
Critical Current at 4.5K and 5.5T	2712A
Sharing Temperatures at 1000A	6.0K

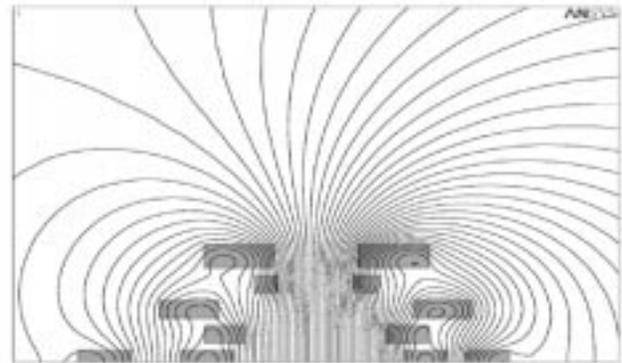


Fig. 3. Flux lines of the best 2-D configuration (0.85% in uniformity).

Conductor: Having fixed the operating current, $I = 1000 \text{ A}$, and the current density $J = (0.6 \div 0.7) \cdot 10^8 \text{ A/m}^2$, the same in all the coils, it comes immediately the conductor cross section: $A = I/J = (14417) \text{ mm}^2$. Choosing an appropriate aspect ratio for the conductor itself, about 1.5, we finally defined the conductor cross section as $5 \times 3 \text{ mm}^2$. The conductor to be used to wind the bending dipole is then shown in Fig. 2. It is composed of a Rutherford cable obtained by cabling 6 multi-filamentary NbTi/Cu wires 0.8 mm in diameter. The Rutherford is then embedded in a pure aluminum stabilizer through a co-extrusion or conform process.

Total Inductance: We should keep as low as possible the induced voltage during magnetic field variations and the hot spot temperature. We found that a reasonable higher limit for the total inductance is 25 H.

Overall Dimensions: The overall dimensions of the bending dipole make up a further constraint. In fact one of the main reasons to choose a superconducting dipole is the possibility to reduce considerably the overall dimension of the rotating gantry. Secondly, the radiation losses depend linearly on the external surface and the conduction losses depend linearly on the mass, and then on the volume. We decided to keep within 1 m^2 the area of the minimum rectangle containing in section all the coils composing the dipole.

Aspect Ratio of the Single Coil: Due to the particular technique proposed by Ansaldo Superconduttori to wind the bent racetrack coils, a particular attention should be paid on the race-track coil aspect ratio L_r/L_z . In order to reduce possible risks

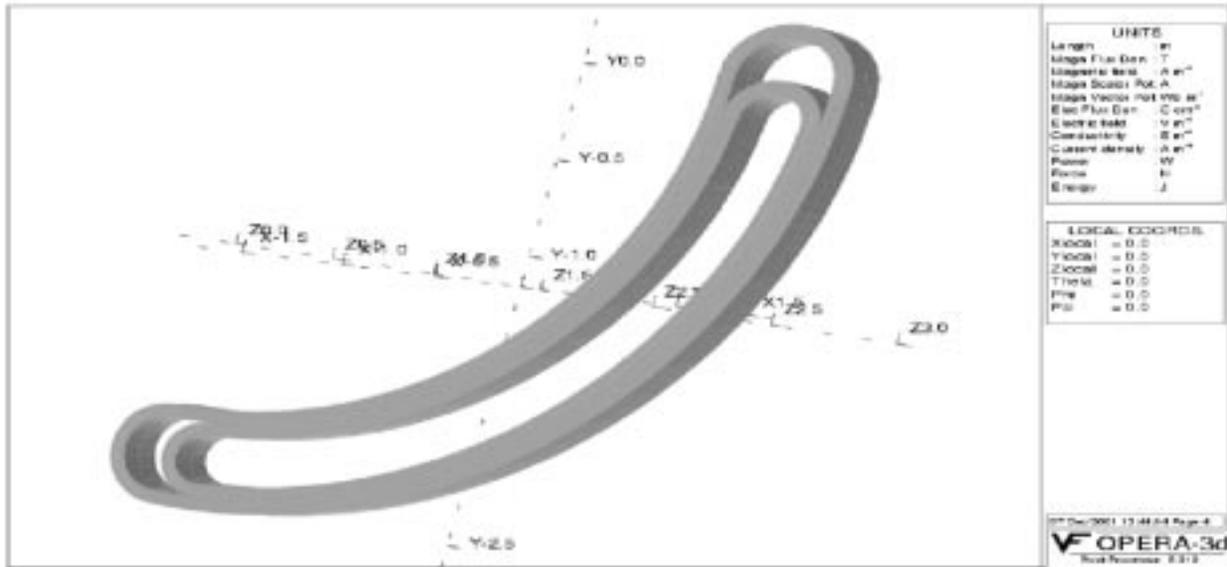


Fig. 4. A single coil with split heads.

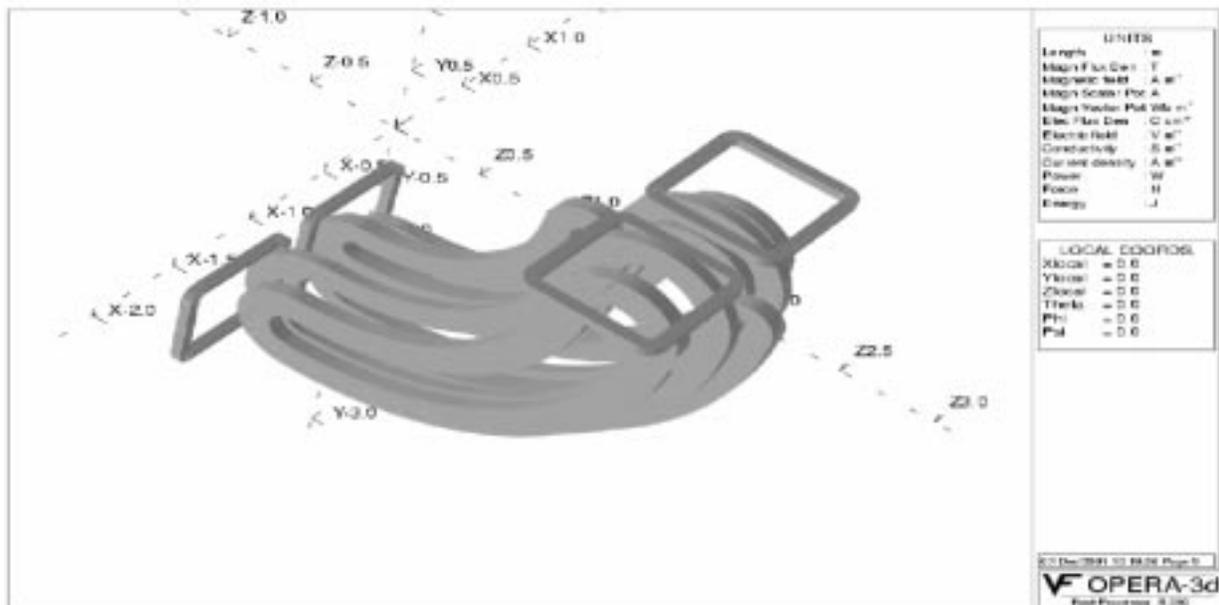


Fig. 5. Dipole model containing four racetrack compensating coils.

during the winding operations, it was decided to keep the aspect ratio below $2.5 \div 2.6$.

Stray Field: The stray field is the last constraint we found in this preliminary design, and it has to be taken inside the limits imposed by Italian law for hospital environment. Even if we are aware of the problem, we decided to postpone its solution; actually several possibilities are under study, as compensating coils or a cryostat made by iron, but none of them was implemented in the optimization algorithm.

IV. 2-D OPTIMIZATION

The finite element code ANSYS [3] used for the 2-D calculations has been linked to a genetic algorithm (GA) [4]. For

each set of parameters given from the GA (called individual), ANSYS calculates the field uniformity in a rectangular region $200 \times 60 \text{ mm}^2$, assigns a fitness to the individual and passes it to the GA.

An axis-symmetric finite element model is used to perform magneto-static analyses. The parameters taken up as design independent variables are: i) the number of turns in the z direction for every coil (5 variables, assuming the same value for the two coils on the equatorial plane); ii) the aspect ratio of every coil (6 variables); iii) the aperture, i.e., the radial distance between the two circular arcs of every coil (6 variables); iv) the distance between the two circular arcs of the equatorial coils (2 variables); v) the axial distance between two adjacent coils, supposed to be equal between all the coils (1 variable).

TABLE II
HEAT LOSSES

Losses	P_1 (W)	P_2 (W)
Radiation	<0.1	5
Conduction (tie rods and wires)	0.6	10
Current leads 1–1000 A	1.5	100
Current leads 1–0	1.5	60
TOTAL	2.2	115

TABLE III
AC LOSSES

Losses	P (W)	Q (J)
Hysteretic losses in filaments	4.6	374
Coupling losses	0.75	62
Eddy currents	4.5	180
Total	9.85	616

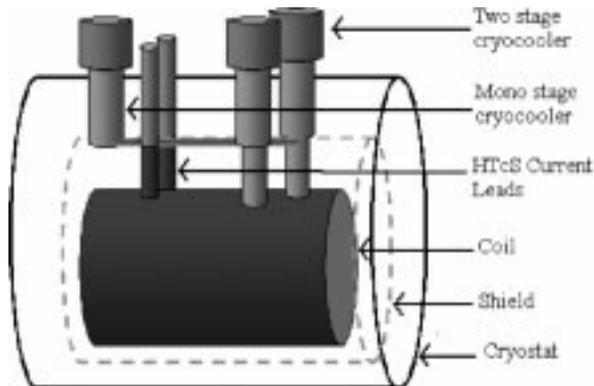


Fig. 6. Cryogenic layout.

The optimization procedure driven by the genetic algorithm produced many solutions, fulfilling all the design requirements. Among them, the best one has a uniformity of 0.85%, with $B_{peak} = 5.1$ T, and $L = 24.5$ H. The central field is 4.000 T, by imposing a current $I = 993$ A. In Fig. 3 the flux lines are shown.

V. 3-D LAYOUT

In the 3-D design our attention is now focused on the beam optics more than on the field uniformity. Our main goal is to obtain an uniform beam size at the patient location during a scan of the 200×60 mm² region. The 3-D design is still on run and is performed linking the electromagnetic finite element code OPERA 3D [5] with the same GA used with ANSYS. Two different approaches are followed. The first one is normally used in dipole design and consists in the head splitting (Fig. 4) (2 parameters for each coils, i.e., 12 parameters), the second one involve the use of compensating coils (Fig. 5) (5 parameters for each racetrack, i.e., 20 parameters). The optimization procedure is carried out as follows: 1) the GA gives the set of parameters; 2) OPERA evaluates the correction angles of the dipole in order to have a 90 degrees bent beam. 3) Starting from the scanning focus three test beam of the dimensions 10×10 mm² are tracked at three angles such to cover the $+100, -100$ mm region. 4) Then the beam distribution at the patient location is obtained and a parameter taking into account the beam distortion at the three locations is calculated. This is the parameter passed to the GA for the optimization.

VI. WINDING TECHNIQUES

Our coils have an aspect that makes difficult the winding operations. In general a racetrack coil, because of its two parallel

straight legs, imposes particular care during winding to avoid creating gaps among turns. For the compaction of windings, the racetrack shaped coil cannot take advantage from the hoop stress present in conventional circular solenoids. Here the situation is even worse, because the coils are curved, and this implies that one of the two legs is concave.

By analyzing the magnetic system with CAD, a particularity of the racetrack curved geometry suggested and made possible an alternative approach for the winding that appears attractive and much easier from the winding point of view.

The concave profile of the coil, included half of the development of each coil head, is exactly identical to the convex profile of the other leg of the coil, head excluded. This means the concave profile can be replaced with an equivalent convex one, without affecting the overall turn length.

This geometric property allows the winding to be made with a simple and innovative approach: the coil is wound with the two legs convex and symmetrical, mirrored with respect the coil longitudinal axis. Once all the turns have been wound, the convex leg is pushed and moved gradually toward its final concave profile. This procedure is much more simpler from the mere winding point of view but, as usual, the solving of one problem has to be paid with the rising of another problem, in principle less severe, that is how to safely push the leg in the concave position.

VII. CRYOGENIC LAYOUT AND LOSSES

The superconducting magnet is designed to operate between 4.5 K and 5.4 K (in any case below 6 K). It will be indirectly cooled through the aluminum alloy mechanical structure (thermal conductivity ≈ 0.1 W/cm K at 4.5 K) by means of cryocoolers, so that the use of cryogenic fluids will be avoided. The reasons of this choice are two: first, the design of a cryogenic vessel is complicated by the rotation and, second, the gantry will operate in a medical environment where a “cryogen free” magnet is preferred for a matter of safety and because it does not need periodic refill. The Gifford–McMahon cryocoolers, largely used for cryogen free systems, are preferred because they are simple to use and fairly reliable (12–18 months maintenance-free continuous operation).

The magnet will be partially charged and discharged during the clinical treatment, so it cannot work in persistent mode via a superconducting switch but has to be supplied continuously. As the current leads are always connected to the magnet giving the main contribute to the heat losses, they must end with HTCS bars working between 4.5 K and about 50 K. The first stage of the cryocoolers ($T = 50$ K) will be connected to the intermediate stage of the current leads (metal-superconductor interface)

and to the thermal shields. The different contributes to the heat load P at 4.5 K (P_1) and 50 K (P_2) are shown in Table II.

It must be noted that, among the commercial cryocoolers, only one is reported to have 2.25 W cooling power at 4.2 K (supplied with 6.4 kW), so that the maximum heat load is limited by that value. Due both to this reason and to the fact that the current leads are the larger source of heat, the maximum current must be not higher than 1000 A. So, in normal operation, only one cryocooler is needed but during the magnet charge and discharge, as well as during the cool down, higher power is required. To supply the additional power a second cryocooler is foreseen. The two cryocoolers will operate in parallel during both the transients and the magnet cooling or, alternately, in case of failure or maintenance of one of them. A single stage cryocooler operating at 50 K could be necessary to compensate the steady state heat load at 50 K mainly due to the heat conduction through the metal sections of the current leads.

When the current of the bending dipole is changed, the field variation at the winding produces heat dissipations.

We can identify the following mechanisms: 1) hysteretic losses in S/C filaments, 2) coupling losses in the multi-filamentary strand, 3) eddy currents in the conductor stabilizer.

In order to evaluate the losses, we have to refer to the coil-operating mode. We have considered a coil operation as follows: i) The starting bending field is 4 T, ii) The field is decreased in step of 0.025 T. The time to reduce the field is 1 s. iii) The field is kept constant for 1 s between two variations. iv) Step by step, the field decreases down to a minimum of 3 T.

We will give two different values of the losses for each mechanism: The average dissipation, in $W(P)$ and the total release of energy, in $J(Q)$. We think that the latter is the most significant value, because it can be associated to an adiabatic increase of the coil temperature. In this case, the safety condition is that during the process the coil temperature is always less than the sharing temperature.

The total energy release to ramp down the field from 4 T to 3 T in steps of 0.025 T is $Q_{tot} = 616$ J. Supposing that this energy is adiabatically deposited in the winding only, the temperature increases up to 5.8 K (i.e., less than the sharing temperature $T_g = 6.0$ K).

A more detailed report of the dipole design, especially on the winding techniques can be found in the internal INFN note [6].

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