ROTATIVE GANTRY FOR DOSE DISTRIBUTION IN HADRONTHERAPY

M. Bajard, UCBLyon/IN2P3/Institut de Physique Nucléaire, Villeurbanne F-69622 France
F. Kircher, CEA/IRFU/SACM, CEA/Saclay F-91191 Gif-sur-Yvette

Abstract

Tumour treatments with high velocity ion beams or protons are characterised by a great depth precision (Bragg peak) and a low divergence for dose delivery in very small volumes. In order to spare normal tissues before and around the tumour it is necessary to have the choice of the beam incidence into the patient. Different devices have been built mainly exocentric and isocentric. Many others are being studied [1] [2]. Cryogenic solutions are considered in order to reduce the total mass in rotation. For example it would be very interesting to choose a superconductive solution for the last 90° dipole.

INTRODUCTION

This work has been done in the frame of the ETOILE project for a national hadrontherapy center in France [3]. Laboratories concerned are CEA/IRFU/SACM, IN2P3/IPNL, ETIC/INSA/LYON, GANIL/SDA. It concerns mainly the feasibility study of a gantry for carbon ions with superconducting magnets and the associated gantry.

WHY A GANTRY

A gantry can drive the beam to the patient at any angle optimized by the treatment planning (over a range of 360°).

The measurements of the beam intensity and profile and the control of the dose distribution are in the nozzle. Inside the gantry system both the position of the patient and his rotation in an horizontal plane can be possible.

The dose deposition is optimized in multiport conditions (Fig. 1):

- multiport allow to spare vital organs,
- most of the proton centers have gantries,
- all hadrontherapy projects are interested in this aspect,
- the only existing design for carbon ions at Heidelberg must be upgraded because it is too large and too expensive.

TWO MAIN TYPES

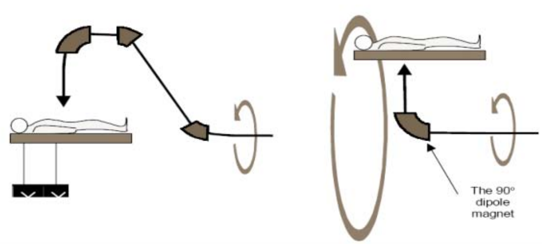

Figure 1: The isocentric gantry on the left is preferred.

MAIN REQUIREMENTS

- Beam penetration depth in water: 27 cm
- Minimization of the weight, the design and the cost for a carbon gantry
- Distance between last magnet and patient: 2 m
- Scanning before the last dipole
- Treatment zone 20 x 20 cm
- Active treatment, spot scanning
- Minimization of the diameter
- Beam dimension 4 to 10 mm

➔ Aperture of last dipole 20 x 20cm!
Consequence on building and structure

HIT DESIGN AT HEIDELBERG

The beam line [4] and the mechanical structure are described below (see Fig.2 and 3).

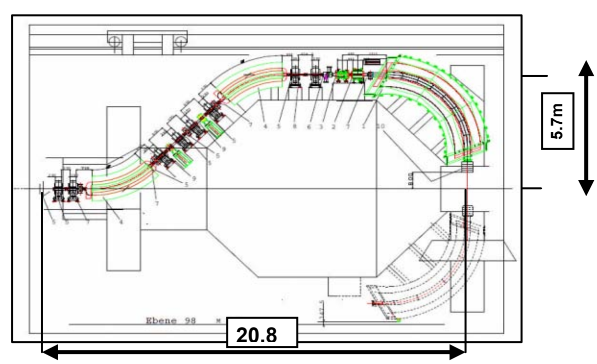

Figure 2: Beam line.

Weight of the dipole: 70 t. Total weight: 650 t
For C^{6+}: 430 MeV/u
Power supply: 622 kW/h (+ cooling!)

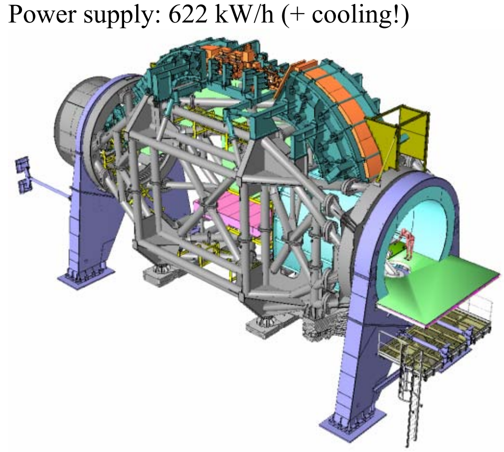

Figure 3: Mechanical support.

WHY A SUPERCONDUCTING GANTRY

The development of accelerators for medical applications in Japan, USA and E.C. has revived this subject.

Main advantages are:

- reduction of gantry weight by en factor ~ 3 and of gantry diameter (see Fig.4),
- expected reduction of the total cost (realization, operation and building),
- the use of superconductivity for such devices was suggested a long time ago (F. Kircher, Magnet Technology 1983) without any realization for about twenty years.

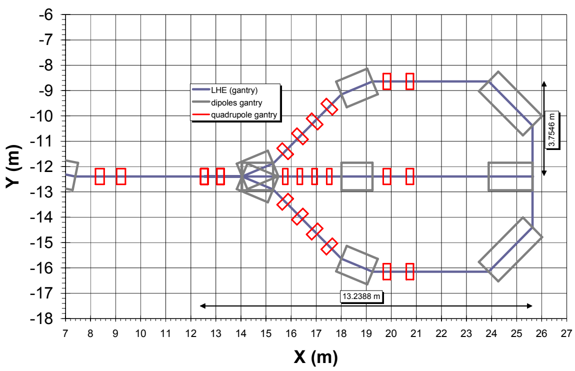

Figure 4: Design of beam line.

MECHANICAL DESIGN OF THE GANTRY

Various solutions have been studied, with a maximum deformation allowed from +/- 0.5 mm to +/- 5 mm.

The present solution (see Fig 5), compared to a solution with conventional magnets (into parenthesis) shows:

- length: 13.5 m (21 m),
- radius: 4 m (6 m),
- total weight (with magnets): 240 t (650 t),
- deformation: < 1 mm.

Precautions to be taken:

- must use non magnetic material
- temperature regulation in the room $\Delta T < 5°C$
- Cryogenics
- use of cryocoolers for operation
- first cooling from room temperature to 4°K with external fluids use of HTS current leads

Estimated weight for the 90° magnet alone: 17 t

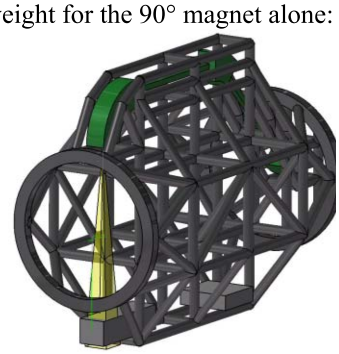

Figure 5: Superconducting gantry structure.

THE 90° SUPERCONDUCTING DIPOLE

This is the most challenging element of the gantry.

Main points of the technical specification:

- $B*\rho$ = 6.45 T x m,
- $\int$ Bdl/B0 $\leq$ 2 10-3 in the useful aperture (200 x 200 mm^2),
- must withstand the beam energy change:
  - equivalent to a decrease of field of 0.03 T/sec during 1.5 sec, then 1sec flattop, for a total time of 3 minutes then 20 minutes until the next treatment,
- fringing field at the patient head $\leq$ 5 G..

Main options chosen:

- magnetic structure (see Fig. 6):
  - flat race track coils (x 12),
  - active shielding (no magnetic iron),
  - B_0 = 3.22 T (ρ = 2 m).
- Conductor:
  - low working point with respect to critical current (large safety margin),
  - NbTi cable soldered in a copper support,
  - mean increase of temperature during beam energy change: 0.6°K (total temperature margin: 1.45°K).

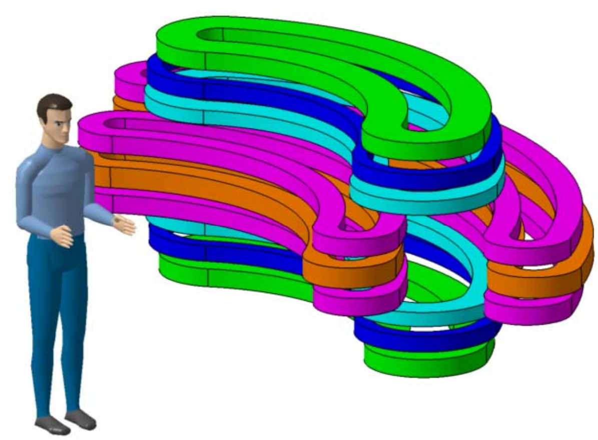

Figure 6: Twelve coils for the 90° SC magnet.

THE USE OF THE SUPERCONDUCTIVITY IN THIS CONTEXT IS NOT TRIVIAL

It is frequently suggested that superconducting magnets should be used to reduce gantry weight and size.

In practice, it is not trivial to:

- build large bending angle superconducting dipoles, especially with large apertures,
- either use flexible cryogenic lines that roll on and off a drum on the gantry axis, or to mount the helium liquefier on the gantry,
- ensure a continuous flow of cryogens at all gantry angles,
- ramp superconducting magnets quickly.

If the superconducting dipole is iron-free, then the stray field is problematic for the detectors.

If the superconducting dipole has an iron yoke, then the reduction of weight is smaller

There are also security risks when cryogenic fluids are used:

- quenching could frighten the patients when the release valve opens,
- there is a risk of oxygen deficiency if large quantities of helium are released,
- there is a risk of cold burns (especially for the lungs) if large volumes of vapor are released.

A reduction in the magnetic rigidity or in the treatment field would bring a more direct advantage [5] [6].

CONCLUSIONS

Many different approaches are possible for gantry design and beam delivery.

The choice of a particular gantry design depends on:

- the type of beam delivery: passive spreading, voxel or raster, scanning,
- whether the beam is spread before or after the last dipole i.e. parallel of divergent scanning,
- the transverse beam distributions: slow extracted beams or fast extracted beams,
- the magnetic rigidity: protons or light ions,
- the method chosen for the rotational optics matching.

Critical parameters for the cost of the design are the size of the treatment field and the maximum beam penetration.

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