A COMPACT SUPERCONDUCTING 330 MeV PROTON GANTRY FOR RADIOTHERAPY AND COMPUTED TOMOGRAPHY*

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Abstract

The primary advantage of proton beam therapy as a cancer treatment is its ability to maximize the radiation dose delivered to the target volume and minimize the dose to surrounding healthy tissue, due to the inherently narrow Bragg peak at the end of the proton range. This can be further enhanced by imaging the target volume and surrounding tissues using proton computed tomography (pCT), which directly measures the energy loss from individual protons to infer the relative stopping power of the tissues. Proton energies up to 330 MeV are required for pCT. We describe a superconducting gantry design which can deliver protons for both therapy and pCT with a similar size to existing treatment gantries. The use of ten identical combined-function superconducting dipole magnets minimizes the weight and technical development required. Based on experience with superconducting magnets for carbon gantries it should be possible to change the magnetic field sufficiently quickly to perform spot-scanning over successive layers without inducing quenching. It is envisaged that a combination of cryogenic cooling and cryogen-free cooling will be used to achieve the required operating temperature for the magnet windings.

INTRODUCTION

Because of the narrow Bragg peak at the end of the proton range, it is important in proton therapy to accurately determine the extent of the target volume. This is because an error of a few millimetres could result in the distal edge of the irradiated volume either falling short of the full extent of the tumour or extending beyond it. This is particularly important for proton therapy where such errors could induce significant under- or over-dosing to the tumour and/or sensitive nearby structures.

Whilst (x-ray) CT provides adequate tissue density information for planning x-ray treatments, conversion to relative stopping power using the Hounsfield scale is not accurate enough for optimal proton treatment planning [1]. One obvious solution is to image the target volume and surrounding tissues using pCT, directly measuring the energy loss from individual protons to calculate the tissue relative stopping power [2, 3, 4].

PROTON BEAM REQUIREMENTS FOR pCT

In order for the energy loss from the proton beam passing through the patient to be measured, the Bragg peak must be located beyond the patient in a suitable

detector. Thus the proton energy must be larger than that required for treatment; protons exit the patient with a residual energy which when measured can be compared to the entrance energy and used to determine the integral of the patient tissue density along the line between entrance and exit [5, 6].

The current generation of proton CT detectors are heterogeneous calorimeters fabricated from, for example, silicon strip detectors with aluminium degrader material.

The energy resolution of any type of calorimeter is inversely proportional to the square root of the energy of the particle being measured, and so the resolution improves with increasing incident energy [7]. It has been stated [8] that the energy resolution of a calorimeter used in proton radiography should not exceed 1%, which is of comparable order to the energy straggling width induced in the patient, and requires a high residual energy of roughly 100 MeV [9]. Given the rough average energy loss of a proton when traversing a patient, this relatively high residual energy requirement is what drives the requirement for an incident energy up to around 330 MeV.

Standard tomographic reconstruction from many incident beams is undertaken to determine the relative stopping power. This can be translated directly into the required proton energy as a function of angle for treatment. This method will reduce the range error if the proton energy can be measured accurately enough.

With the considerably smaller beam current required for tomography and the location of the Bragg peak outside the patient, the dose due to imaging with pCT is likely to be small when compared to x-ray CT [10].

GANTRY CONCEPT

The isocentric gantry is now the most widely-used delivery arrangement in treatment centres around the world. The patient is kept in a supine position with the centre of rotation of the gantry passing in the horizontal plane through the patient and the incident beam rotated in the vertical plane around the patient. A number of dipoles are used on the gantry to create an offset between the beam and the gantry rotation axis and finally turn the beam toward the patient.

The size of the gantry is dictated by the choice of magnet technology. Normal-conducting dipole magnets have a maximum field on the beam axis of around 1.8 T. Given that the beam rigidity of protons at 250 MeV is 2.43 Tm, this implies a bending radius of 1.35 m; this bending radius, along with the space needed for the beam-spreading system to cover the treatment field, sets the overall size of the gantry, which is typically 5 to 6 metres in radius with a mass up to 200 tonnes. For protons at 330 MeV, the beam rigidity is about 17% larger, implying a bending radius of 1.58 m. Thus most existing treatment

rooms cannot accommodate a normal-conducting gantry of conventional design that could be used for pCT.

One solution would be to employ superconducting dipole magnets. Superconducting gantry designs for carbon ion therapy are underway at ETOILE and NIRS [11]. However, superconducting dipole magnets are also a way of designing a 330 MeV gantry for pCT that can be retro-fitted into existing treatment rooms.

MAGNET DESIGN

The NIRS gantry design consists of ten combined-function superconducting magnets (along with scanning and steering magnets). The gantry proposed in this paper uses a similar layout but with certain significant differences:

- the use of ten identical magnets;
- it features a novel scanning system after the final dipole.

The reason for this is to make the design and construction as simple and therefore cost-effective as possible. Only one magnet design needs to be developed; for this single design to have sufficient aperture to be used throughout the gantry without being unfeasibly large the scanning system must be positioned after the final dipole. It is proposed to use direct-wind superconducting technology, with the dipole and quadrupole coils electrically isolated within the magnet and connected to separate power supplies. This will allow each field component to be independently excited and will provide the flexibility needed to match the beam parameters from a range of accelerators to the required values at the isocentre. Each magnet has a magnetic length of 400 mm and a dipole field of 2.8 T. A schematic layout of this design is shown in Fig 1 along with the outline of the inside of the shielding in a typical treatment room.
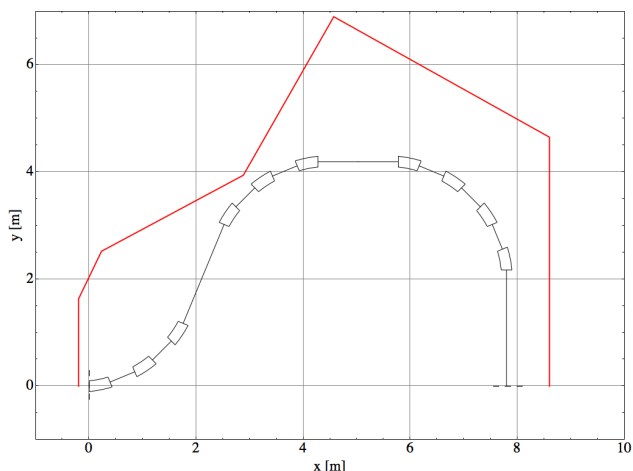

Figure 1: A schematic layout of the gantry design within a typical treatment room envelope.

The key parameter affecting the size of the magnets is the aperture, or more precisely the size of the good-field region. In order to calculate this it is necessary to match a range of reasonable input Twiss parameters (taken from a range of existing gantry designs) to a range of reasonable

final Twiss parameters at the isocentre. Thus the design is not restricted to any particular accelerator/nozzle combination. The combinations chosen are shown in Table 1.

The one sigma beam size (σ_i) throughout the gantry for each of these combinations is calculated using the standard formula, where β_i is value of the beta function and η_i is the value of dispersion at every point:

$$\sigma_i = \sqrt{\beta_i \varepsilon_i + \eta_i^2 \frac{\Delta E^2}{E^2}} \qquad (1)$$

The emittance (ε_i) and energy spread ($\Delta E/E$) used are the largest values from a review of accelerators for proton beam therapy [12], 10 mm mrad and 0.5 % respectively. The required magnet aperture is then the largest value of σ_i at every point in the beamline, across all eight matching scenarios.

Figure 2 shows the largest value of the one sigma beam size throughout the gantry for all the combinations of initial and final beta functions shown in Table 1. As only one of the matched solutions has a beam size that extends significantly beyond 20 mm, this is a pragmatic choice for a single magnet design good-field region.

Table 2 lists the maximum quadrupole gradients required and the maximum beam size that the magnet must accommodate.

Table 1: Twiss Parameters used for Matching

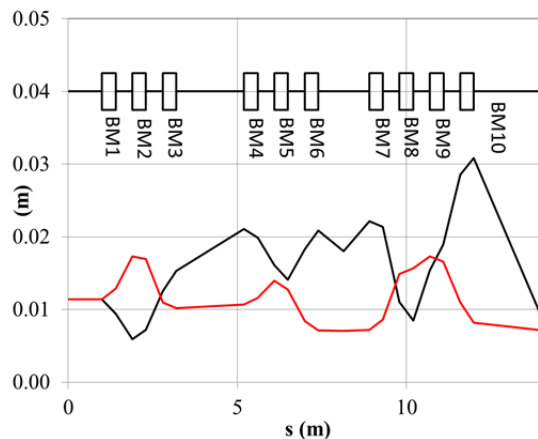| Initial β_x or β_y (m)	Final β_x or β_y (m)	σ_i at isocentre (mm)*
0.5	0.5	2.2
1	0.5	2.7
1	1	3.0
5	1	3.2
10	1	3.1
10	3	5.5
13	3	5.5
13	5	7.1

*for ε_i = 10 μm, $\Delta E/E$ = 0.5 %

Figure 2: The largest value of the one σ beam size.

Table 2: Magnet Parameters

Magnet	Maximum Gradient (m^{-2})	Maximum 1σ beamsize (mm)
BM1	3.11	13
BM2	-1.71	17
BM3	3.50	15
BM4	2.58	21
BM5	-2.09	16
BM6	2.67	21
BM7	3.50	22
BM8	0.69	16
BM9	1.60	19
BM10	1.97	31

OPERATIONAL ISSUES

The use of conventional liquid helium-filled cryostats is not possible as liquid movement during gantry rotation would lead to quenching, so it is necessary to use cryocoolers to obtain the low temperatures required in the magnet windings. These cryogen-free heat pumps will allow rotation of the gantry magnets, but they have a more limited capacity than cryogenic-liquid cooling systems. A pre-cooling procedure using liquid nitrogen can be used to reduce the temperature to ~70 K, prior to the cryocoolers cooling the magnets to 4 K. This will reduce the total time to get from room temperature to 4 K from several weeks to several days. Superconducting magnets are also prone to quenching when new (until a period of training has been undertaken) and also when required to change field rapidly. Tests at NIRS [13] have shown that quenches can be recovered from immediately; although realistically it may take up to two hours. It is not yet clear how quenching would be managed if this happened part-way through a treatment fraction.

CONCLUSION

The requirement for a high-energy gantry is primarily driven by the desire to image the whole body with pCT and the resolution limitations of the detection system. Although delivering a 250 MeV proton beam (which is the current gantry state-of-the-art) is sufficient for pCT for head-and-neck treatments and for many paediatric treatments, higher energies are needed to image for all treatment types.

We have designed a superconducting gantry suitable for retro-fitting into existing proton-therapy treatments rooms that can deliver protons at 330 MeV and is thus capable of supporting pCT, which offers significant advantages for treatment planning; particularly for tumours in close proximity to vital structures in the body.

ACKNOWLEDGMENT

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