SUPERCONDUCTING GANTRY DESIGN FOR PROTON TOMOGRAPHY*

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Abstract

Precise proton therapy planning can be assisted by augmenting conventional medical imaging techniques with proton computed tomography (pCT). For adults this requires an incident proton energy up to around 330 MeV, requiring superconducting magnets if an imaging gantry is to replace a conventional 230-250 MeV gantry in the same space. Here we present optics considerations for a superconducting gantry to deliver 330 MeV protons within the context of the future Christie Hospital proton therapy centre, where it is proposed to increase the initial 250 MeV proton energy in the future with a 3 GHz 54 MV/m booster linac.

INTRODUCTION

The precise dose delivery achievable with proton-based radiotherapy requires accurate treatment planning to obtain the greatest benefit. Presently, margins defined around treatment plan volumes are larger than they might be; these margins account for uncertainties in translating densities from CT scans. Proton CT (pCT) may reduce this error by directly measuring proton stopping via the loss of energy as protons pass completely through an imaged structure; margins could be reduced from as much as 10 mm to as little as 2 mm using this technique, and several proton-counting detector technologies may be employed. Whilst head-and-neck and paediatric pCT is within the energy reach of current 230-250 MeV proton therapy machines such as cyclotrons, full adult pCT will require up to 350 MeV protons to maintain the Bragg peak beyond the imaged patient. The Cockcroft Institute is developing solutions to obtain 330 MeV protons using either FFAG or cyclinac approaches, constructing a prototype linac to be tested later in 2017 [1,2]; this is planned to be tested on a research beamline presently under construction at the Christie Hospital NHS proton therapy centre (see Fig. 1). Eventually a 330 MeV proton beam will be delivered to patients, and this paper examines a compact gantry design to achieve that.

GANTRY DESIGN

Isocentric Gantries

The isocentric gantry is today the most widely-used beam delivery arrangement in treatment centres around the world; the patient is maintained typically in a supine orientation with the treatment volume placed at the isocentre of rotation of the gantry [3]. In virtually all gantry designs the protons are directed normal to the axis of gantry rotation, and increasingly magnetic spot scanning of the delivered protons is employed where the scanning elements are placed either downstream or upstream of the final dipole bending magnets; upstream scanning over a significant transverse treatment field (e.g. 30×30 cm) necessitates a larger magnet aperture which may limit the achievable dipole field and can give rise to field variations during spot scanning that requires involved feedback correction to maintain the desired uniformity of spot properties. Hence, we favour downstream scanning although this increases the overall nozzle length from dipole magnet to isocentre; most commercial gantry solutions employ downstream scanning.

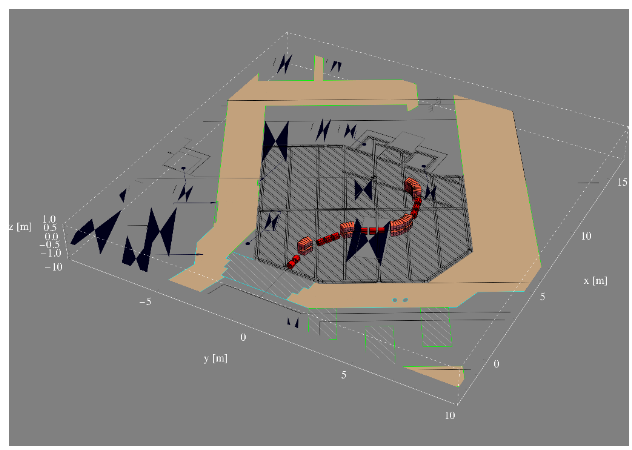

Figure 1: 3D visualisation of the pCT gantry in the research room at the Christie Hospital.

Gantry Design for pCT

Whilst compact, normal-conducting gantry designs exist for ‘treatment’ protons up to 250 MeV with dipole fields up to about 1.8 T (corresponding to treatment depths for the proton Bragg peak up to ~33 cm), the next generation of compact gantries will employ larger magnetic fields to bring down both size and weight. Moreover, optical designs increasingly aim to enlarge the momentum acceptance so that the gantry magnets need not be adjusted during scanning over a given depth range of the Bragg peak. Static gantry magnets allow for very rapid energy scanning in the case of high dose rate delivery methods.

There is not yet a compact design suitable for 330 MeV protons, for which the maximal beam rigidity increases from 2.43 Tm at 250 MeV to 2.84 Tm. To maximise the momentum acceptance in a given aperture we propose the use of a two-achromat optical design utilising 2.8 T dipole magnets; NIRS have demonstrated a cryogen-free design at this field for their carbon-ion treatment gantry [4]. Compared to our

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Table 1: Main parameters of the 330 MeV pCT achromat gantry design.

Gantry Length	7.6 m
Gantry Radius	3.5 m
Main Dipole Max Field	2.8 T
Dipole Type 1 Length/Angle	0.52 m / 30 deg
Dipole Type 2 Length/Angle	0.65 m / 37.5 deg
Quadrupole Length/Max Gradient	0.35 m / 25 T/m

previous ten-dipole Pavlovic-style design [5], this new achromatic layout requires fewer dipoles (six) and has a c.3.5 m radius (4.5 m previously). Nine 25 T/m normal-conducting quadrupoles accomplish beam matching to the isocentre after the coupling point; it may be possible to reduce this number later. Overall parameters are given in Table 1; the layout is shown in Figure 2.

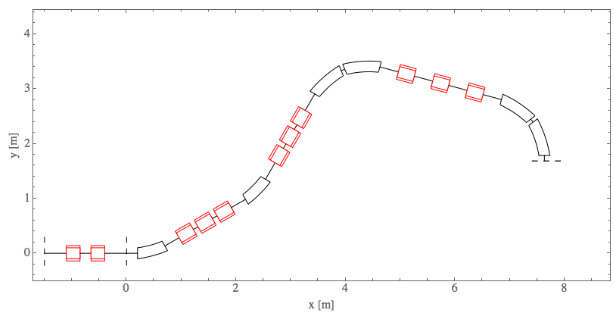

Figure 2: Preliminary layout of the compact pCT gantry design; dipoles shown (curved) in black, quadrupole triplets shown in red.

Optical Properties of the Design

The set of nine quadrupoles allows use to achieve a reasonably-small dispersion (and hence a larger momentum acceptance) whilst controlling optical functions; see Figure 3. The lack of upstream scanning alleviates the need for a 90-degree phase advance between scanning magnets and isocentre, although in the present design the source-to-axis distance (SAD) is quite small (about 1.8 m). Magnification/demagnification from the entrance coupling point is likewise straightforward to achieve. Typical beam optics parameters are shown in Figure 4.

Comparison with Other 330/350 MeV Designs

The present design has several advantages and disadvantages when compared to other designs. LBNL/PSI have presented a Pavlovic-style gantry utilising a single large-aperture superconducting dipole to achieve a reduced outer radius but still a significant overall length [6,7]. The large-aperture final magnet facilitates beneficial upstream/parallel scanning at the isocentre but at the expense of likely requiring involved optical correction when scanning within the field of the final SC dipole; however, the LBNL/PSI design predicts a larger momentum acceptance (up to 20%)

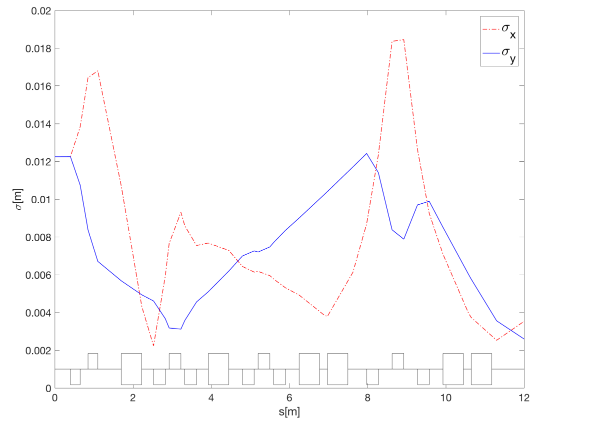

Figure 3: The one σ beam size for emittance ε = 10 μm and energy spread $\Delta E/E$ = 0.5%.

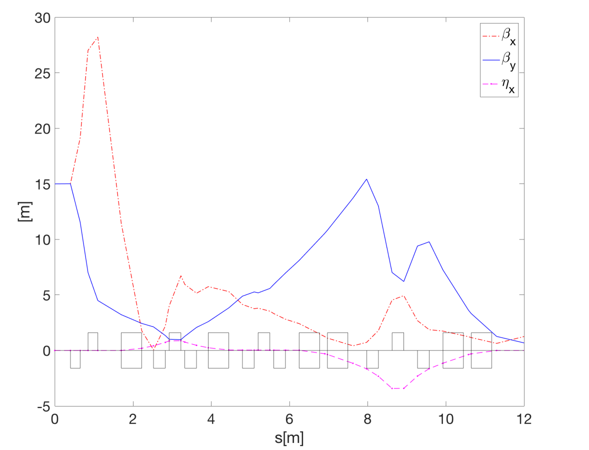

Figure 4: Typical optical functions achievable in the compact pCT gantry design.

compared to existing gantries, allowing for rapid energy scanning.

FFAG designs have been proposed by several authors that propose a very large energy acceptance without magnet adjustment (as large as 70 MeV to 250 MeV or even greater) and a compact size if superconducting magnets are utilised [8]. However, FFAG designs require very strong gradient combined-function magnets and consequently a complex optical design and verification of the constructed magnetic elements. We consider that the achromatic design achieves many of the same advantages in a simpler design.

FUTURE PLANS

The optics design presented in this paper satisfies the key requirements for a gantry suitable for proton CT that can replace a conventional proton therapy gantry in the same space. It may be further improved by increasing the SAD to a larger value; i.e. reducing the size of the achromats by incorporating a quadrupole gradient into the superconducting dipole design. A forthcoming magnet feasibility study should also answer the question whether increasing the magnetic field above 2.8 T for the required aperture is possible. Additionally, the gantry setup in practice should be examined, i.e. locations for diagnostic screens, an assessment of scanning system, as well as incorporation of collimation and energy selection systems into the design. It is then to be followed by beam tracking studies and inclusion of the booster linac into the final gantry design.

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