# A novel beam optics concept in a particle therapy gantry utilizing the advantages of superconducting magnets

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## Abstract

**Purpose:** *A first order design of the beam optics of a superconducting proton therapy gantry beam is presented. The possibilities of superconducting magnets with respect to the beam optics such as strong fields, large apertures and superposition of different multipole fields have been exploited for novel concepts in a gantry. Since various techniques used in existing gantries have been used in our first design steps, some examples of the existing superconducting gantry designs are described and the necessary requirements of such a gantry are explained.*

**Methods:** *The study of a gantry beam optics design is based on superconducting combined function magnets. The simulations have been performed in first order with the conventional beam transport codes.*

**Results:** *The superposition of strong dipole and quadrupole fields generated by superconducting magnets enables the introduction of locally achromatic bending sections without increasing the gantry size. A rigorous implementation of such beam optics concepts into the proposed gantry design dramatically increases the momentum acceptance compared to gantries with normal conducting magnets. In our design this large acceptance has been exploited by the implementation of a degrader within the gantry and a potential possibility to use the same magnetic field for all energies used in a treatment, so that the superconducting magnets do not have to vary their fields during a treatment. This also enables very fast beam energy changes, which is beneficial for spreading the Bragg peak over the thickness of the tumor.*

## Neue Strahloptikkonzepte in Teilchentherapie-Gantries mit Nutzung der Vorteile der supraleitenden Magnete

## Zusammenfassung

**Zweck:** *Ein strahlenoptisches Design der ersten Ordnung von einer supraleitenden Protonentherapiegantry ist vorgestellt. Die Möglichkeiten der supraleitenden Magnete in Bezug auf die Strahlenoptik, z.B. starke Felder, große Aperturen und Überlappung der unterschiedlichen Multipolfelder, wurden für das neue Konzept der Gantry ausgenutzt. Da einige Techniken der existierenden Gantries für die ersten Schritte des Designs genutzt wurden, sind einige Beispiele der vorhandenen supraleitenden Gantries beschrieben und die notwendigen Anforderungen an solche Gantries sind dargelegt.*

**Methoden:** *Die Untersuchung des Strahloptikdesigns der Gantry basiert auf den supraleitenden Magneten mit kombinierten Funktionen. Die Simulationen wurden in der ersten Ordnung mit konventioneller Strahlentransport-Software durchgeführt.*

**Ergebnisse:** *Die supraleitenden Magnete geben die Möglichkeit, starke Dipol- und Quadrupolfelder zu überlagern. Die Nutzung dieser Überlagerung der Felder erlaubt die Einführung der lokal achromatischen Biegungsabschnitte ohne Zunahme der Gantrygröße. Die gründliche Umsetzung der neuen Strahlenoptikkonzepte in vorgestelltem Gantrydesign erlaubte eine dramatische Steigerung der Impulsakzeptanz verglichen mit Gantries mit normalleitenden Magneten. In unserem Design wurde*

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**Conclusions:** *The results show an improvement of its momentum acceptance. Large momentum acceptance in the gantry creates a possibility to implement faster dose application techniques.*

**Keywords:** Proton therapy, gantry, superconducting, beam optics

*diese große Akzeptanz durch die Implementierung von dem Degrader auf der Gantry ausgenutzt, was zur potentiellen Möglichkeit führt, das gleiche Magnetfeld für alle bei der Behandlung genutzten Energien zu verwenden, sodass das Feld der supraleitenden Magnete während der Behandlung nicht geändert werden muss. Das erlaubt sehr schnellen Energiewechsel, was vorteilhaft für die Ausweitung der Braggspitze über die Tiefe des Tumors ist.*

**Schlussfolgerungen:** *Die Ergebnisse zeigen ein signifikantes Potential der Verbesserung der Impulsakzeptanz der Gantry. Die hohe Impulsakzeptanz erschafft die Möglichkeit, die neuen Methoden der Dosisanwendung zu implementieren.*

**Schlüsselwörter:** Protonentherapie, Gantry, supraleitend, Strahloptik

# 1 Introduction

## 1.1 Scanning the proton beam over tumor tissue

Proton therapy makes use of the Bragg peak with its finite range to help provide conformal dose distributions, thus minimizing dose to healthy tissue. In the scanning pencil beam technique, a narrow proton beam is scanned in the two transverse directions and the depth of the Bragg peak is set by adjusting the energy of the pencil beam.

In this study we have chosen to examine a gantry design based on pencil beam scanning.

The size of the tumor projection in the plane perpendicular to the beam direction is usually much larger than the beam diameter. Scanning of the beam in both transverse directions is performed in most modern facilities via the deflection of the narrow 'pencil beam', a technique first demonstrated in NIRS [1] and LBNL [2] and first used in clinical routine in a gantry at PSI [3]. The beam is deflected via the scanning magnets, which are located before (upstream scanning) or behind (downstream scanning) the final bending magnet in the gantry, but also other possibilities exist [4,5].

The location in depth of the Bragg peak is set by choosing the beam energy. Presently, at almost all operating proton therapy facilities the beam is accelerated by a cyclotron with fixed extraction energy or by a synchrotron with adjustable energy. In cyclotron facilities the energy is being reduced by a so-called degrader – low Z material in the beam transport system – to the value required by the patient treatment plan. Such a system and the following magnets can be designed such that they allow energy variations to spread the dose in depth over the tumor thickness. The time needed to change energy accordingly, should be as short as possible to limit the total treatment time. Another advantage of the possibility to have fast energy changes is in the application of volumetric rescanning [6]. Currently, the energy change speed is limited by the time to change the magnetic fields between a degrader and the patient.

All magnets in the gantry must be ramped synchronously with the energy (proportionally to beam momentum) variation, which should be fast in order to minimize the treatment time. The PSI Gantry 2 magnetic field is changed on the order of $\Delta B/(B\Delta t) = 10\%$ per second. We consider this speed as a desirable specification for a gantry with superconducting magnets. However, for such magnets the ramping speed of is slower due to limitations of AC-losses and hysteresis effects.

In order to reduce the treatment time, it is useful to reduce the time it takes to make an energy change. In Section 1.4 this will be discussed in more detail.

*Proton therapy gantry high level requirements*

A gantry is a rotating system at the final section of the particle therapy facility beamline. It is composed of several dipole and quadrupole magnets, which are able to bend proton beams with a maximum energy of approximately 230–250 MeV or carbon ions of approximately 450 MeV/nucl, not taking particle beam imaging into account. Its rotation together with the movement of the patient table allows irradiating the tumor tissue from different directions. In the so-called iso-centric gantries the gantry and magnets direct the central (non-scanned) beam toward the iso-center, the common point in space, where the gantry rotational axis is crossed with the beams from all gantry directions. The irradiated tumor must coincide with the beam trajectory and can be placed at the gantry iso-center [7]. The scanning system deflects the beam in a lateral direction with respect to the direction of the central (non-scanned) beam.

There is a big interest in the proton therapy research to reduce the cost of the proton therapy facilities in order to

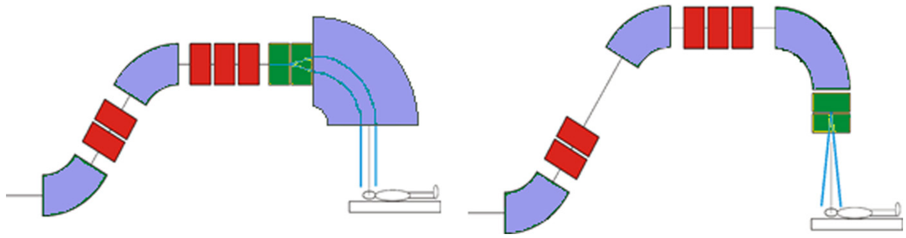

Fig. 1. Schematic layout of scanner magnets being located upstream (left) or downstream (right) of the final bending magnet.

make this type of therapy more affordable. It is perhaps possible that a reduction in the magnet weight can play a role in cost reduction, but this has not yet been proven. Since most proton therapy facilities are located in the densely populated areas in order to allow a good access of the potential patients to the facility, the footprint and the volume of the facility play an important role in the gantry acquisition decisions. Consequently, these parameters must as well be taken into account during the gantry design. Although these considerations will have an effect on decisions the direction of certain solutions, in this paper we concentrate on the design of the beam optics when using superconducting magnets.

The use of superconducting magnets is an emerging issue in the development of proton and ion therapy gantries. Designs of superconducting gantries have been developed worldwide by research institutes as well as companies. Presently several of these designs are under commissioning or construction [8–10]. A possible motivation to apply superconducting magnets in a gantry could be the possibility to reduce the gantry's weight, eventually up to an order of magnitude for heavy ion facilities. In case of the larger heavy ion gantries (e.g. a normal conducting Heidelberger Ionenstrahl-Therapiezentrum gantry $l = 25$ m and $r = 6.5$ m), the National Institute of Radiological Sciences design shows that the size could be reduced significantly. The implications of these factors for a proton therapy facility are not yet fully understood. We believe they could have positive benefit, but in the current study our concentration is on the possible performance benefits of using superconducting magnets.

### 1.2 Beam optics

The bending of the beam in every dipole magnet causes the trajectory of particles with a non-nominal momentum to deviate from the nominal axis of the beam. This chromatic phenomenon is called dispersion and can be described by the so-called dispersion function. The combination of gantry magnet apertures and the maximum amplitude of this trajectory determines the maximum momentum deviation that can be accepted by the gantry. In existing gantries with global chromatic correction a gantry accepts a momentum band of about ±0.5–1%.

In the present paper we distinguish global achromatic correction, meaning suppression of the transverse and angular dispersion at the iso-center, and local achromatic correction suppressing transverse and angular dispersion (see below) by a subgroup of magnets within the gantry. In both cases this is a first order achromaticity. The beam position is independent of energy, but chromatic focusing errors have not yet been taken into account. The gantry optics is simulated in first order with Transport [11,12] software. In Transport the effect of the beamline elements on the beam envelope is described by so called "transform matrices" ($R$, with each matrix element represented by $R_{ij}$) describing the beam optical elements (magnets and drift spaces).

### 1.3 Upstream and downstream transverse scanning

Scanning in the plane transverse to the depth direction is performed by "scanning magnets", which are located upstream or downstream of the final bending section (see Fig. 1).

In the case of downstream scanning the gantry radius is larger than for a comparable upstream scanning gantry, since sufficient drift space for the beam after the scanner magnets is needed. One advantage of the downstream scanning is that the final bend dipole can have a smaller aperture and hence it can be lighter, easier to manufacture and may have more relaxed requirements for the field quality. Another advantage could be a larger clinical field size can be obtained.

In case of upstream scanning, the scanning magnets can be imaged to the iso-center as "point to parallel". A deflection of the scanning magnet then causes a parallel displacement of the pencil beam at isocenter, i.e. an infinite Source-to-Axis-Distance (SAD). In the terms of optical imaging, this is achieved by arranging for $R_{22}$ and $R_{44}$ between the scanning magnets location and the iso-center to be small, i.e. zero for the infinite SAD. Another advantage of the upstream scanning is a possible reduction of gantry radius. On the other hand, it can lead to an increase of the gantry length, due to the location of the scanning magnets in the horizontal section of the gantry. But, since gantry length contributes linearly to the volume and footprint area, and the radius contributes with its square to the volume, upstream scanning is still advantageous for the gantry volume. In addition, the gantry radius contributes with a factor of two to the total height of the needed vertical space in the building.

In case of upstream scanning, the final bending dipole must have a relatively large aperture to enable a reasonable scanning field size in both $x$ and $y$ plane. For example in case of parallel scanning the aperture of the final bend must be at least as large as the sum of field size at the iso-center plus the beam spot size in the magnet. This may impose a complicated (due to large aperture and field homogeneity requirements) and costly design of the magnet [7,13], and especially in this respect superconducting magnets are expected to be of advantage.

It is also possible to position one scanning magnet before and one after the final bend, combining the upstream and downstream scanning [5]. When using the first scanning magnet for scanning in the bending plane of the following bending magnet, this magnet only needs a large aperture in the bending direction. Scanning in the dispersive plane can be performed with the scanning magnet behind the bending magnet. For example, the "racetrack" dipole layout has a much larger aperture in the dispersive plane than in the transverse one and is a good candidate for such a scanning option.

In IBA's gantry Proteus I the scanning magnets are located in between the bending magnets of the last bending section, which also has reduced the gantry radius [14].

### 1.4 Energy modulation and magnetic ramping

Tumor tissue is scanned in depth by varying the energy of the beam and hence the penetration depth of the Bragg peak. Given the width of the Bragg peak in proton beams, scanning is performed in layers of approx. 5 mm thickness. The corresponding required change of the beam momentum per layer is of the order of ~1%. The time needed for a momentum change between the layers should be kept as short as possible to reduce the treatment time. It varies from 1 s in existing commercial gantry models to 0.1 s at PSI Gantry 2, including a pause for the system to stabilize [15]. In this gantry the full range of energy change (70–230 MeV, which corresponds to ~50% in the momentum) needs approx. 2.5 s. The fields of all magnets downstream of the degrader are changed synchronously, proportional to the momentum of the particles. Since each step (of ~1% in momentum) requires an accurate synchronization and several checks, effectively such a step takes 80–100 ms. Of course a requirement is that the gantry should be able to "follow" such an energy change: there should be no change in position or spot shape in the transverse plane at isocenter when energy is changed. An achromatic layout is of great advantage in this respect. We have adopted this requirement for the specifications of the planned superconducting gantry as well.

For the gantries with large ferromagnetic yokes the effect of the magnetic hysteresis must be considered as well. One way to deal with the effect is to perform the energy change only in one direction, e.g. from the higher to the lower energy, so that the impact of hysteresis becomes reproducible. Allowing energy changes in both directions would require a complicated algorithm taking into account the recent magnetic ramping history.

For the superconducting magnets achieving a high ramp rate might become a serious problem because of the eddy currents. Eddy currents in the normal conducting part of the cables generate heat and increase the probability of the magnetic quenching. Eddy currents in the superconducting cables are remanent, hence they can lead to field shape changes. If it is necessary to include rapid (e.g. <1 s) magnetic field changes, a particular focus must be laid on the design of a suitable magnet with large aperture, the choice of the optimal superconductor material and its filament-layout, as well as on the consideration of cooling possibilities (e.g. cryo-coolers, liquid helium etc.).

## 2 Examples of existing superconducting gantries

Some interesting proposals of superconducting gantries have been published [16,10]. In the current chapter some examples of gantries with superconducting technology use are discussed. The chosen examples are based on designs or use components which we have used as starting points or have regarded as interesting concept for our study. In particular, the combination of dipole and quadrupole fields within one magnet in the currently commissioned NIRS gantry and the achromaticity and compactness of the two bending sections in the ProNova SC360 gantry have served as starting reference of the design presented here.

### 2.1 ProNova SC360 Gantry

In 2011 a design of a superconducting gantry for therapy with 230 MeV protons was proposed by the company ProNova [17,9]. This design consists of two superconducting bending sections. Each of them contains two superconducting dipoles with a superconducting quadrupole triplet between them (in Fig. 2 marked green). Before the first superconducting section and in between the two sections four normal conducting quadrupoles are located. This design performs downstream scanning (see Section 4.1), and makes use of superconducting separated-function magnets (i.e. separate dipoles and quadrupoles) in the bending sections. The optics of each bending section has been designed such that each is *locally achromatic*. This gantry is commercially available, but at the moment of writing no clinically working system was in operation yet.

The beam optics design of a superconducting gantry for 350 MeV protons, based on ProNova SC360, was developed in 2012 [18]. Such a gantry could be useful for proton radiography and for the treatment of smaller tumors [19,20].

The SC sections in this design were achromatic, resulting in a momentum acceptance of the gantry of $\Delta p/p > \pm 3\%$. This larger momentum acceptance has initiated some studies at PSI on the potential possibilities of a gantry with even larger momentum acceptance in general and has triggered the design presented in this paper.

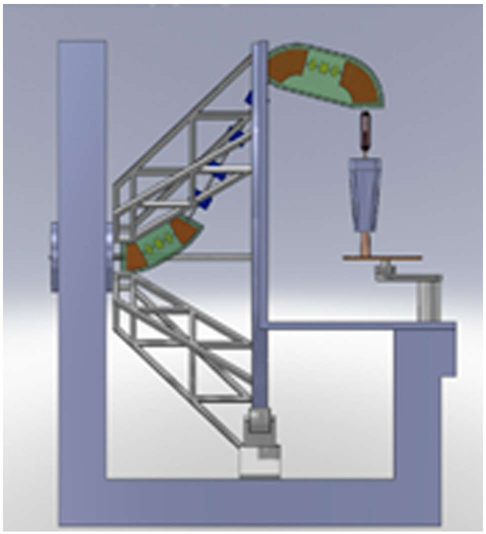

Fig. 2. Layout of the ProNova superconducting SC360 Gantry [9].

### 2.2 NIRS superconducting carbon ion gantry

In particle therapy carbon ions are also used for tumor irradiation. The maximum energy of the carbon ions required for therapy is in the order of 450 MeV/nucl, so that the magnetic rigidity $B\rho$ of the carbon beam is approximately a factor 3 higher than a proton beam with equal penetration range. This has resulted in very a large gantry for carbon therapy, the first one has been designed by Pavlovic and is in clinical operation at HIT in Heidelberg [21] since 2009. So especially for carbon therapy the very strong fields which are possible with superconducting magnets are expected to have a considerable effect in reducing the size and weight of the gantry.

The first superconducting gantry has been developed by Toshiba and is being commissioned at the National Institute of Radiological Science (NIRS) in Japan for carbon ion therapy [8,22,23]. It is an iso-centric gantry with a layout similar to PSI's Gantry 2 [7] and gantry at GSI, having first bending sections with the dipoles of smaller aperture and a scanning system upstream of the final bending section. Therefore this last bending section of 90° has a larger aperture to match to the needed lateral space for the beam. In the NIRS gantry the superconducting dipoles of each bend are of the so called "cosine theta" type and have been designed such that they are combined with quadrupole fields with independent and adjustable strengths (see Fig. 3) [8]. Control of the beam size, dispersion suppression and focusing of the beam at the iso-center is performed simultaneously with bending. In this design the gantry is globally achromatic and local achromaticity has not been applied in the different bending sections.

## 3 Technical aspects

We have started the design of a gantry for scanning proton beams of maximum 230 MeV, with superconducting magnets in the bending sections as previously described.

### 3.1 Size and weight of the gantry

Contrary to the possibilities in an ion gantry, it is not to expect that the diameter of a proton therapy gantry can be reduced drastically by utilizing superconducting magnets, since a reduction of the bending radius of the proton beam from approximately 1 m to, say, 0.5 m, would only decrease the total gantry radius with not more than 0.5 m. In proton gantries the major factor influencing the diameter is the choice between scattering, upstream, downstream or combined scanning methods. Superconducting magnets, however, offer more possibilities for upstream scanning due to the larger aperture of the magnets of this type.

It can be expected that usage of magnets with no or little iron yoke can reduce the weight of the gantry substantially by a factor of 5 to 10. This can be illustrated by the comparison of the ProNova SC360 gantry weight (∼25 tons [9]) with the weight of PSI Gantry 2 (∼200 tons).

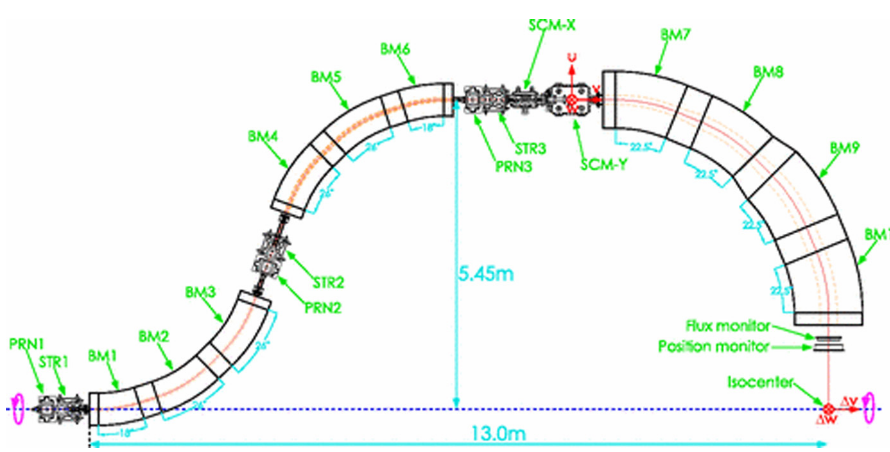


Fig. 3. Layout of NIRS superconducting carbon ion gantry [5].

### 3.2 Dealing with superconducting magnet technology in a clinical environment

The fractionation scheme mostly used in a radiation treatment with proton therapy and the needed high availability allows only maintenance shutdowns of 1–2 days. Therefore the cryogenic system and the magnets should be designed such that they do not need to be warmed up on a time scale of 10 or more times per year and a quench of the superconducting system cannot be accepted in the clinical environment which would cause a treatment interruption of several days.

Superconducting magnets can be designed to have a very strong magnetic field. One should take care that the stray field of the magnets is not larger than 0.5 mT at the iso-center [24]. This is a level which is considered to be safe for patients with pacemakers, and of other critical devices next to the magnets may cause a gantry angle dependence of the field quality in the magnets due to the presence of iron in the gantry's environment (e.g. in the concrete). Therefore shielding with additional coils or some iron will be necessary.

In order to limit the cryogenic complexity of a rotating system with liquid helium, the use of cryo-coolers is also considered. This has been implemented on the gantry at NIRS [8]. The cryo-coolers must be designed and mounted such that they can be subject to the gantry rotational movement and their choice has to be feasible from the commercial point of view.

### 3.3 Design related to aspects of beam properties

#### 3.3.1 Beam phase space at the coupling point and at the iso-center

We aim for a design of the ion-optics that is independent of gantry angle. The phase space of the beam is assumed to be symmetric between the transverse planes $x$ and $y$ and free of momentum dispersion, at the coupling point, so that there is no correlation between energy and position in the beam spot. This insures that the beam shape in the gantry is independent of gantry's rotation angle. Installation of a circular collimator at the coupling point can further constrain the beam properties. Such a collimator could limit the beam size transported to iso-center and decouple beam misalignment errors in the fixed beamline from the iso-center in such a way that would only lead to intensity changes rather than beam position changes at iso-center. At the iso-center the spot is desired to be a symmetric, Gaussian shape with a radius of 2.5 mm ($1\sigma$). The beam size at iso-center is determined by the beam (or collimator) size at the coupling point, the beam optics in the gantry, the scattering in materials such as the vacuum window at the magnet exit, air, the monitors, possible range shifters and in the nozzle and in the air between the vacuum window. The scattering contributions dominate at low energies.

The optics calculations presented here, have all been performed at high energies, neglecting distortions due to scattering, deviations from ideal magnetic fields (homogeneous fields and gradients and sharp magnet boundaries) and higher order effects. These distortions are currently under investigation.

#### 3.3.2 Ion optical design and introduction of a collimator

The first order optics is designed such that the collimator at the coupling point is imaged to the iso-center. In the optics transport matrix [25,26], the matrix elements $R_{11}, R_{12}, R_{33}$ and $R_{34}$ relate the particle location $x_{\mathrm{isoc}}$ and $y_{\mathrm{isoc}}$ at iso-center to the particle location $x_{\mathrm{coupl}}$ and $y_{\mathrm{coupl}}$ and particle angle $\theta_{\mathrm{coupl}}$ and $\varphi_{\mathrm{coupl}}$ at the coupling point:

$$x_{\mathrm{isoc}} = R_{11} \cdot x_{\mathrm{coupl}} + R_{12} \cdot \theta_{coupl} \quad \text{and}$$
$$y_{\mathrm{isoc}} = R_{33} \cdot y_{\mathrm{coup}} + R_{34} \cdot \varphi_{\mathrm{coupl}}$$

The imaging of the coupling point to the iso-center (point-to-point imaging) requires the elements $R_{12}$ and $R_{34}$ to be zero, so that:

$$R_{12} \cdot \theta_{\mathrm{coupl}} = R_{34} \cdot \varphi_{\mathrm{coupl}} = 0$$

Since we require the imaging to be achromatic as well, we obtain the situation that the magnifications in $x$ and $y$ ($R_{11}$ and $R_{33}$) are determining the particle location according to first order optics:

$$x_{\mathrm{isoc}} = R_{11} \cdot x_{\mathrm{coupl}} \quad \text{and} \quad y_{\mathrm{isoc}} = R_{33} \cdot y_{\mathrm{coupl}}$$

It is useful for the elements $R_{11}$ and $R_{33}$ to be equal so that a symmetric beam spot size at the coupling yields a symmetric beam spot size at the iso-center, hence facilitating the treatment planning.

The advantage of these imaging conditions is that, in first order, errors in beam divergence and entering angle at the coupling point do not result in pencil beam displacement or spot size increase at the iso-center. Also, beam offset or beam size errors at the coupling point result only in larger losses on the collimator at this location, but not in the beam offset or spot size increase at the iso-center. This results in conditions at iso-center becoming less sensitive to alignment errors in the beamline upstream of the gantry.

An additional collimator at an intermediate focus before the last bending section is also included. The design is such that a similar imaging exists between this collimator and the iso-center. Also, if a degrader is included immediately before the second collimator the pencil beam size at isocenter will be determined mostly by the diameter of this degrader-collimator.

#### 3.3.3 Ion optical design/achromaticity

Dispersion in $x$-direction (plane of the bending magnets) can be represented by the transport matrix element $R_{16}$. The beam size $x$ at any location is given by

$$x = \sqrt{x_{\text{monochrom}}^2 + (R_{16}\Delta p/p)^2}$$

where $x_{\text{monochrom}}$ is the beam size without taking momentum deviation into account. At the coupling point the dispersion value must be zero in order to ensure the rotational symmetry of the gantry. At the iso-center the dispersion value must be kept low to minimize the beam size and to prevent any correlation of beam penetration depth with the horizontal position. It is also beneficial to keep the beam size value in the gantry itself as small as possible in order to have a larger momentum acceptance through the gantry, and one way to do that is to minimize the dispersion function.

#### *3.3.4 Beam losses*

It is important, especially in the design of the static beamline upstream of the gantry, to concentrate beam losses only at particular locations which are specifically designed for it, such as collimators after the degrader or the energy selection slit. These areas must be sufficiently shielded, so that the losses do not affect the patients and the personnel working in the neighboring rooms. Also beam losses should be prevented in the superconducting magnets, since these could lead to quenching and consequently to a longer down time.

Therefore already in first order optics design, choices of distances and magnet strengths have been made such that he beam size and deflection due to scanning will have a limited amplitude within the gantry magnets, so that some space will be available for beam size increase due to the scattering and higher order effects.

### 3.4 Beam optics requirements resulting from scanning considerations

The maximal area at the iso-center, which can be covered by the scanning method without mechanical movement of the patient or the rotation of the gantry, is called *field size.* The requirement for the transverse clinical field size varies and can reach 30 cm × 40 cm. Since this field size has major consequences on the magnet aperture for an upstream scanning system and given the constraints on these apertures the field size requirement has been relaxed to 20 cm in each direction. All desired positions within a rectangular or elliptic field must be reachable with the scanner magnets for all energies. The modulation in depth should cover the full water equivalent range from ∼0 to 35 cm. The range between 0 and 4 cm depth can be covered by a pre-absorber (range shifter plates) in the nozzle, just before the patient, since the transmission through the degrader for the energy reduction below 70 MeV would be too low. Therefore we have assumed that the energies of the protons to be transported in the gantry magnets are in the range 70–250 MeV.

It is helpful to ensure that the spot size and shape do not change when transverse scanning is performed. In order to achieve that a high homogeneity of the final bend magnetic field must be ensured. Also, the focusing in the bend should not be dependent on the beam's transverse position and a sufficient compensation of the higher order effects, if present, must be achieved.

As one can see in Fig. 5, the scanning SAD of the gantry is not infinite, but is in the order of 3 m. This value does not provide a parallel scanned beam but is larger than some common values of 1–2 m SAD in the downstream scanning gantries. Also, with the current optics the shift of the beam induced by the scanning magnets deflection is not too large in the final bend, so that the aperture of the final bend is not overly large. The shift of the scanned beam in the final bend is 82% and 44% of beam shift at the iso-center for the $x$ plane ($x_{\text{iso}}$, dispersive) and $y$ plane ($y_{\text{iso}}$, transverse), respectively. This means that, given the $2\sigma$ beam size of approximately 7 mm at the final bend exit and a magnet aperture radius $A_{\text{magn}}$ of 125 mm, the maximum extent of the beam edge of a scanned beam in each direction at the iso-center is

$$x_{\text{max, iso}} = \frac{A_{\text{magn}} - 2\sigma_x}{0.82} = 143\,\text{mm}$$

$$y_{\text{max,iso}} = \frac{A_{\text{magn}} - 2\sigma_y}{0.44} = 268\,\text{mm}$$

A yielding a total maximal (elliptic) scanning field of 286 mm × 536 mm. This indicates that the proposed design has still a good margin of bending magnet aperture for the stated 20 cm × 20 cm clinical field size requirement. This margin can be used to increase the momentum acceptance of the bend (see Section 4.3).

To summarize, the design of proton gantry optics should aim to fulfill the requirements stated in Table 1.

Table 1
Summary of parameter requirements for the superconducting gantry design.

| Requirement | Comment |
|---|---|
| Gantry radius $r$ | ≤3.2 m |
| Gantry length $l$ | ≤8.9 m |
| Gantry weight | ≪100 tons |
| Beam spot size at the iso-center | 2–3 mm ($1\sigma$) in vacuum at high energies |
| Scanning field size at iso-center | 20 cm × 20 cm |
| Magnetic field at the iso-center | 0.5 mT |
| $R_{12} = R_{34} = 0$ | Imaging between coupling point and iso-center |
| $R_{11} = R_{33} =$ const. | Imaging magnification factor |
| $R_{16} = R_{26} = 0$ | Achromaticity between coupling point and iso-center |
| Beam energy range | 70–230 MeV |
| Time for energy modulation 230 → 70 MeV | <2.5 s |
| Maximal duration of regular maintenance | 1–2 days |

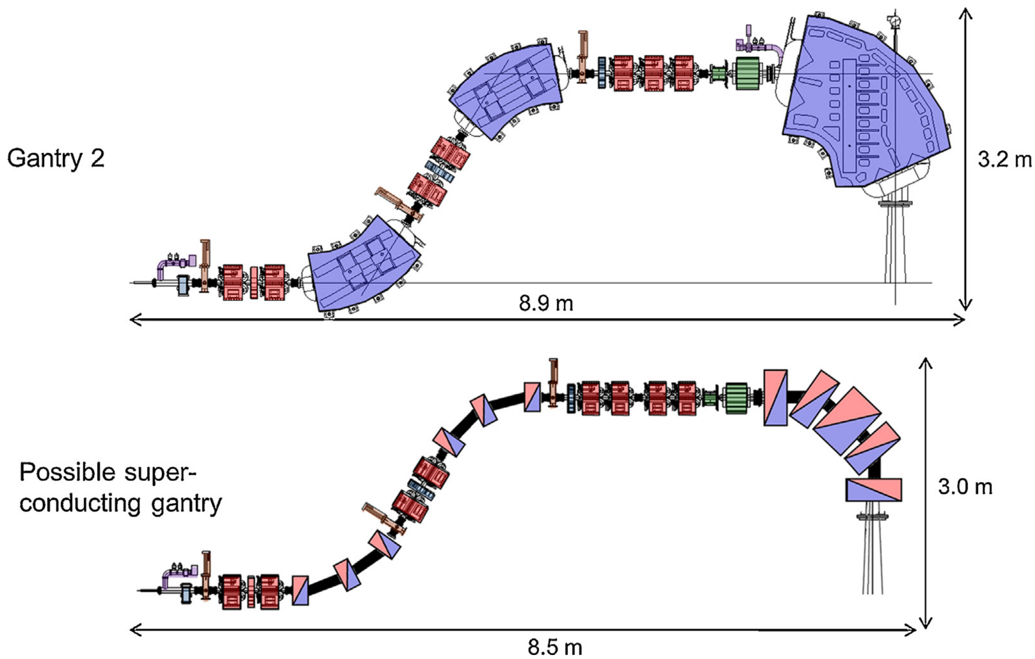


Fig. 4. Schematic illustration of Gantry 2 (top) and proposed superconducting gantry (bottom) with their dimensions. The dipole magnets are shown in blue, the quadrupoles in red, the combined function magnets (dipole and quadrupole) as a combination of blue and red and the scanning magnets in green.

## 4 Proposed optics design of the gantry

A beam optics design of a superconducting gantry based on the requirements specified in Section 3 has been developed and is presented in the current section. A brief description of the magnets considered for the current design is given in Section 6.

### 4.1 Gantry layout

Fig. 4 shows the layout of the proposed superconducting gantry (bottom) compared with PSI Gantry 2 (top). In both designs there are two bending sections of 60° and one of 90°. However, in the superconducting gantry design the bending sections consist of several superconducting combined function magnets with overlapping dipole and quadrupole fields. Additionally, the design contains eight normal conducting quadrupoles before and between the bending sections.

Scanning has been implemented upstream of the final bend, requiring a relatively large aperture of the final bend magnets.

### 4.2 Gantry optics and imaging

The specifications of the beam phase space area at the coupling point are assumed to be 1.75 mm × 4.25 mrad × $\pi$ = 7.44 $\pi$ mm mrad (1$\sigma$ values) in both transverse planes. The simulation of beam characteristics while passing through the gantry is shown in Fig. 5. The simulation has been performed as a first order calculation with Transport software for a beam of 250 MeV energy ($B\rho = 2.4$ Tm). In the figure the horizontal axis represents the z-position along the beam from coupling point to gantry iso-center. The vertical axis above zero indicates two standard deviations value of the vertical beam size ($2\sigma_y$). The vertical axis below zero indicates the value of the horizon beam size ($2\sigma_x$). The dotted line shows the dispersion ($0.01 \times R_{16}$) – the beam trajectory deviation for a momentum offset of $\Delta p/p = 1\%$. The red and blue curves show the trajectory of the beam when deflected by the scanning magnets in transverse and dispersive planes, respectively. In the calculations the available magnet aperture was not decided upon yet, so the deflection angles have been chosen so that the offset of the beam at the iso-center is 100 mm in each plane to achieve a rectangular scanning field of size 20 cm x 20 cm. The bending sections are marked in green.

As mentioned before, each bending section has been made achromatic to first order. The two 60° bending sections at the entry of the gantry consist of three 20° bending magnets, each having 30 mm aperture radius and 283 mm length (AMLA, AMLB, AMLC and AMLD, AMLE, AMLF in Fig. 5). In these sections the respective central magnets (AMLB and AMLE) also have strong quadrupole gradients (field index $\neq 0$), focusing the beam in x-plane and hence reducing the dispersion. The magnets have a dipole field of 3 T and a quadrupole gradient of up to 45 T/m, based on what can be achieved realistically [6]. The final 90° bending section has been split into five 2 T dipoles with 125 mm aperture radius and combined with overlapping quadrupole gradients of up to 33 T/m for the dispersion suppression and focusing.

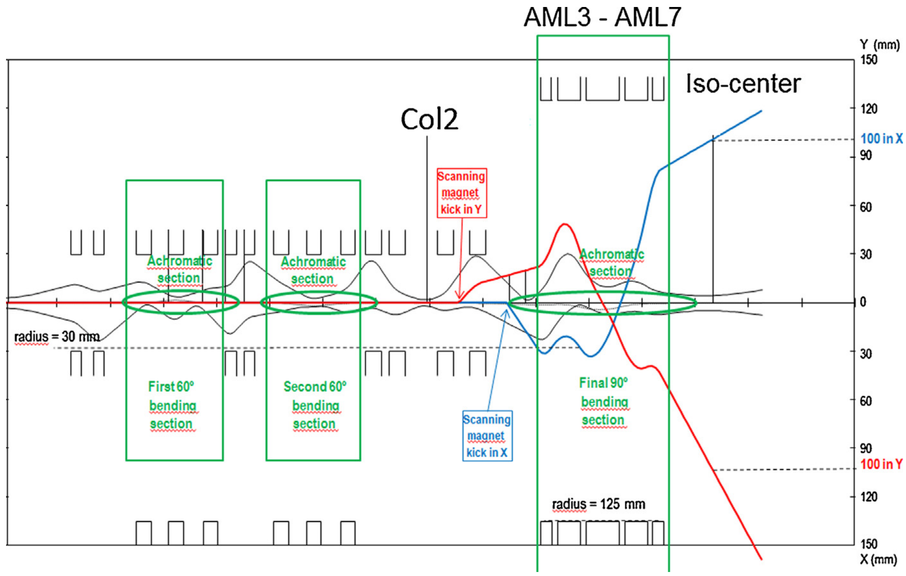


Fig. 5. Transport simulation result for the proposed superconducting gantry beam optics.

In the calculation result presented here non-linear effects on the pencil beam size due to the scanning have not been taken into consideration. Such tracking simulations have been carried out for a variation of this gantry and can be found in [27]. These show that the sextupole compensation might be required to achieve a sufficiently small beam size for the scanned beam.

There are several parameters which need separate, dedicated simulations, such as field variations due to hysteresis and time dependent settings of the superconducting magnets and an imperfect alignment of the magnets due to a non-perfect rigidity of the gantry mechanics, which can be gantry angle dependent.

### 4.3 Momentum acceptance and use of a collimator and degrader

As discussed at the end of Section 1, all existing normal conducting gantries are achromatic as a whole, but the achromaticity is not restored within each individual bending section. In the proposed superconducting gantry design each bending section is achromatic by itself (“local achromaticity”). Such achromaticity in the beam optics is achieved using strong quadrupole fields focusing in the dispersive plane, located within each bending section. Because of that, the dispersion function never reaches a high value. Using this feature, a gantry design is presented here, with a momentum acceptance of $>\pm 10\%$. This means that without a change of the currents in the superconducting magnets, a beam with a momentum offset of up to $\pm 10\%$ (corresponding to the energy offset of almost $\pm 20\%$) can still pass through the aperture of the gantry magnets and the vacuum pipe.

The combination of a large momentum acceptance downstream of a degrader can lead to interesting clinical beam delivery modalities that might otherwise not be achievable in a superconducting magnet gantry system. Firstly, the momentum spread of the beam entering such a superconducting gantry does not need to be limited to the $\pm 0.5$–$1\%$, required by the apertures in most normal conducting gantries. This can be of potential benefit to transport a larger momentum spread at low energies (when the energy spread is too small for easy longitudinal spreading), or as otherwise clinically acceptable.

In the described design, the speed of energy changes is only determined by the mechanical speed of the degrader. An energy step of 2% can be made within several ms if the degrader is mounted at a location where the beam size is small. In that case the mechanical motions in the degrader can also be small (and can thus be fast). To minimize the emittance growth the beam should have a small diameter when entering a degrader. The round collimator aperture at the coupling point at the entrance of the gantry is imaged to the iso-center, but with an intermediate image at a second collimator between the second and the third bend (Col2). The ($1\sigma$) beam size at this collimator is $1.25\,\text{mm} \times 1.25\,\text{m}$, which is also imaged to the iso-center in such a way that the beam spot size there is $2.5\,\text{mm} \times 2.5\,\text{mm}$ ($1\sigma$).

Due to the large momentum acceptance, the bending magnets then need not change their fields when the degrader is set to another energy. However, first order imaging to iso-center is blurred by chromatic errors. If the energy changes are implemented just after the collimator before the last bend, one can correct the chromatic imaging errors. This can be done by applying an energy dependent correction in some quadrupoles behind the degrader. The optical feasibility of such correction has been demonstrated above on the example

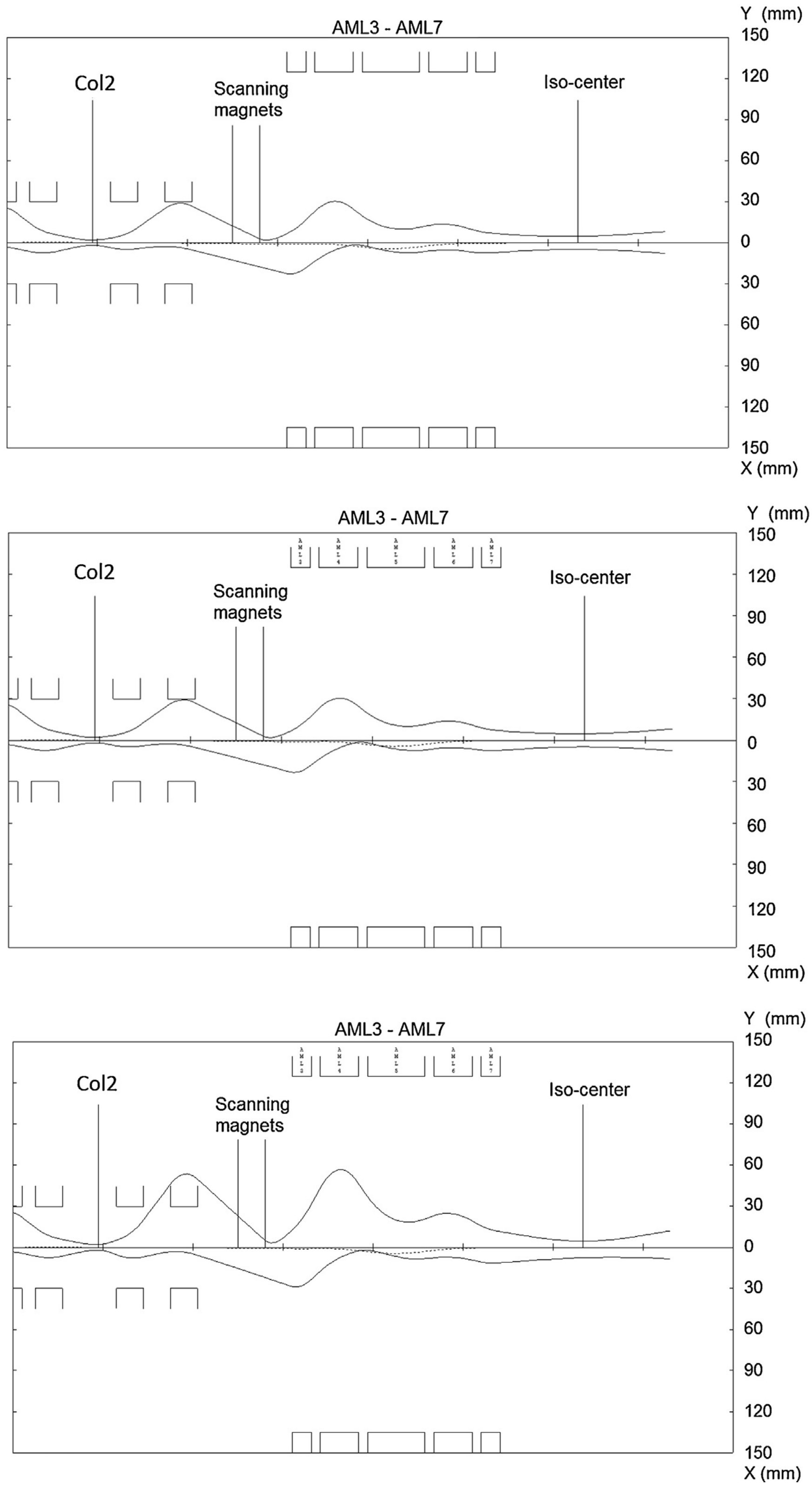


Fig. 6. Beam optics for the proposed superconducting gantry after the collimation point Col2 without momentum reduction in the collimator (top), with momentum reduction of 10% (center) and with a corresponding beam divergence increase of 21 mrad (bottom).

of energy reduction of $\Delta p/p = -10\%$ (see Fig. 6), however it will require fast power supplies for the normal conducting quadrupole magnets.

In the present design the beam forms a waist at the position of the collimator, which is designed such that the beam divergence is high by strong focusing. This high divergence ensures that multiple scattering in a degrader will not increase the beam divergence too strongly, since the angular spread from the multiple scattering adds quadratically to the original beam divergence. For the worst-case-scenario of degradation from 70 MeV to 53 MeV the divergence would increase approx. from 15 mrad to 25 mrad in both transverse directions.

The generally known limited ramping speed of the superconducting magnets might be a bottleneck for the speed of energy scanning. However, if a treatment can be performed with zero, only one or maybe two field settings per gantry angle, the treatment time can be strongly reduced.

#### *4.3.1 Challenges*

Fig. 6 demonstrates the effect of such a range shifter installed in the collimator point between the second and the third bend (Col2). The diagram in Fig. 6 top shows the beam envelope downstream of Col2 with a momentum matched to the fields of the magnets (same situation as for Fig. 5). In the central diagram the beam transport with a momentum change at Col2 of $\Delta p/p = -10\%$ has been simulated. The fields of normal conducting quadrupoles have been modified for the changed energy to keep the focus and the imaging at the iso-center at this modified energy, while the fields of the superconducting magnets have been left unchanged. The focusing and imaging function works well in this case. However if one includes also the effect of the emittance growth produced by the range shifter for the worst case of beam energy reduction from 70 MeV to 53 MeV (see Fig. 6, bottom), the beam size at the location of the normal conducting quadrupoles and at the iso-center would become too large. Therefore, also with a gantry designed for large momentum acceptance, one still needs the pre-absorber in the nozzle to achieve very low energies for dose applications in and just below the skin.

We have observed that for the scanning with the momentum reduction via the range shifter the offset of the beam at the iso-center is dependent on beam energy for a given scanning magnet strength or deflection angle. The achromatic condition $R_{16} = R_{36} = 0$ is not fulfilled for all the variable bending angles of the scanning magnets, meaning that this first-order effect due to the scanning angle can defy the dispersion suppression. However, we found that it is possible to correct the corresponding beam shift at the iso-center by modifying the scanning magnet's deflection angle. This first order chromaticity error induced by the scanning magnets will also increase the spot size at isocenter. However, we found that this can be corrected for by readjustment of a quadrupole between the second collimator (Col2) and the scanning magnets. Since these magnets are designed for fast scanning, such corrections will be included in the setting tables generated in the treatment planning. Additionally we are studying the possibility of modifying the exit angle of the scanning magnet pole faces, to obtain a scanning-angle related focusing correction.

### 4.4 Other beam optics concepts

In this section several alternative possibilities in the design are discussed. These alternative solutions motivate the choices made for the here presented design of a gantry with superconducting magnets.

#### *4.4.1 First two bends as achromatic section*

It has been examined whether the first two bends could be used as one achromatic section instead of each one of them being achromatic by itself. The advantage of such arrangement would be that it would not be necessary to separate the magnets of the first two bending sections in three segments with different (or alternating) quadrupole gradients, but it would just be sufficient to have two 60° bending magnets with homogeneous fields. However, in that case the maximum amplitude of the dispersion would be significantly higher in this layout compared with the case with two separate achromatic sections. This is due to the longer bending magnet (60° instead of 20°) in that case. For the quadrupole magnets an aperture radius of 30 mm has been assumed, hence such arrangement would pose a limitation to the momentum acceptance and in these two bending sections it would be reduced from $\Delta p/p = \pm 10\%$ to $\Delta p/p \approx \pm 2\%$. However, since most energy spread will be due to the degrader mounted at Col2 in the gantry and the momentum spread in the incoming beam is smaller than $\pm 1\%$, this could be acceptable.

#### *4.4.2 Downstream scanning with 30 mm aperture radius dipole*

A very serious option is to use the same superconducting magnets with 30 mm radial aperture used for the first two bends also for the final bending magnets. The smaller aperture would ensure less stray fields and shorter fringe fields, resulting in an easier beam optics design and reducing the necessity for extensive shielding of the magnets. Also, the dipole field of the magnets could be strongly increased, hence making the bend significantly shorter.

However, this would only be possible with downstream scanning. In that case the scanning field is not limited by the aperture of the last bending magnet. The earlier mentioned experience at PSI with upstream scanning has demonstrated the feasibility of such approach and the desire to keep the gantry radius as low as possible has motivated our preference for a system with upstream scanning.

Table 2
List of parameters for the superconducting magnets used in the gantry design.

| Bend number | Dipole field in the center | Max. quadrupole gradient | Top pole field | Full aperture |
|---|---|---|---|---|
| 1 & 2 | 3 T | 45 T/m | 3.35 T | 60 mm |
| 3 | 2 T | 33 T/m | 6.17 T | 250 mm |

### 4.4.3 *Using quadrupoles between the dipoles*

In the design of the achromatic bending sections it has been considered to use two dipoles and a few (e.g. 3) quadrupoles between them, to give more possibilities for focusing, while keeping the system achromatic. This has also been done in ProNova SC360 Gantry design [9,18]. However, all quadrupoles between the dipoles influence the dispersion as well as the focusing. In addition to this, the total length of the achromatic bending section would increase. Also, due to their larger length, there is less dispersion focussing in the bending magnets, so that the momentum acceptance will not be as large. From magnet technology [27] we experienced that it can be relatively easy to add quadrupole fields that overlap the bending dipole field. When using five instead of three quadrupole fields, this conveniently adds degrees of freedom for the design of the beam focussing and dispersion suppression. As will be discussed in the next section, the CCT magnet design is supporting this possibility.

## 5 Magnet design

A brief overview of the requirements set for the superconducting magnets for the described gantry design.

The field qualities required for the stated gantry optics are described in Table 2. Since the final bend has a large aperture due to the upstream scanning system, the requirements of the central field and the field gradient can be more relaxed than for the first two bends: approximately 3.4 T. Due to the addition of the field gradient and the dipole field the maximal pole field, however, is significantly higher with 6.2 T in the final bend.

Such fields are rather difficult to achieve with reasonable field quality using normal conducting magnets. Despite the fact, that such a locally achromatic design would not be impossible with normal conducting magnets, the size and weight of the normal conducting magnets implementing the presented design and local achromaticity would be significantly larger than for the current globally achromatic gantries. For example, the corkscrew downstream scanning design by the Harvard group [28] and first installed at Loma Linda, has local achromaticity in the bending sections and has a diameter of approximately 12 m. However these gantries cannot be modified easily into a gantry with upstream pencil beam scanning. Hence, it is the local achromaticity within each bend that becomes more practical with usage of superconducting magnets and is a particular feature facilitated by their use.

The proposed design considers magnetic fields below the technically achievable limits. Stronger magnetic fields would mean shorter magnets for the given beam rigidity $B\rho$. However, this would lead to a larger relative influence of the fringe fields at the entrance and the exit of the magnet. Nevertheless, the gantry design for carbon ions or for protons with 350 MeV energy even stronger magnetic fields can be considered.

The required field quality, here defined as the difference between the required field and the actually obtained field, for the magnets in the full aperture region is estimated to be on the order of $10^{-4}$ to $10^{-3}$. Detailed tracking calculations following the pencil beam along its different trajectories through the bending system will give a more accurate field quality constraint. We are working on a process to define the requirements more clearly than the ones defined by a simple number specifying the field homogeneity. We are continuing our work on a further refined optical layout that minimizes higher order aberrations to correct beam size and shape distortions and enhances the effective momentum acceptance.

The relatively short length of approximately 1 m causes particular challenges of the design of such magnets, such as the effects on the field quality due to magnet curvature and as in case of the final bend magnets, the large aperture. Also, due to the rotation of the gantry the liquid helium cooling method is rather difficult to implement. The helium flow must be guaranteed under different angles and during the rotation. Quenching (e.g. due to a vapor in the cooling fluid) has to be kept to minimum to provide the required high availability. The connection of the liquid helium channels must not be disrupted due to the rotation. Hence, in case of the liquid helium cooling

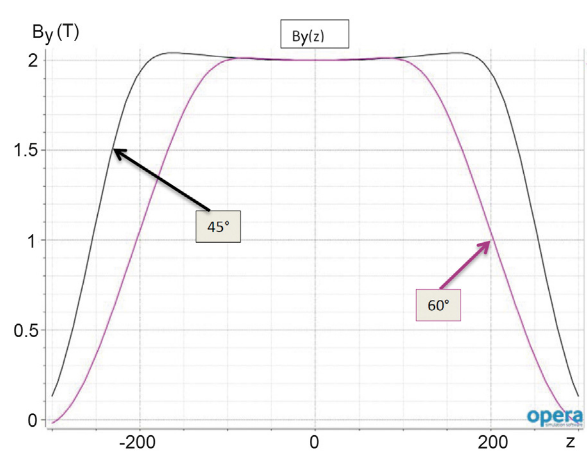


Fig. 7. The vertical B-field along the Z-axis (beam direction) for the coils winded under the angle of 45° and 60° with the vertical axis [29].

a gantry with a limited rotation angle of a bit more than 180°, may offer technical possibilities when combined with a the patient table which can be rotated horizontally [7].

The detailed layout analysis and the tracking simulations recently performed in Canted–Cosine–Theta (CCT) magnets [27] show that these are a viable option to satisfy many of the requirements, since they offer a relatively good beam quality over a large aperture. A consequence of the large aperture is a longer extension of the fringe fields. The sharp cut off fringe field ("SCOFF") model, which is frequently used in beam optics codes, cannot be applied and track simulations are the only valid tool in that case. To study the extent and shape of the CCT fringe fields a preliminary study has been performed [29]. Fig. 7 demonstrates this in the example of two different coil windings.

## 6 Conclusions and outlook

We believe that the use of superconducting magnets is an important research topic in the field of proton and ion therapy gantries. The beam optics of a superconducting gantry has been developed. The optics design presented here is partially advancing concepts of some other currently existing superconducting gantry designs described in Section 2.

In this contribution new options for the beam optics are shown, enabled by the use of superconducting bending sections. These include locally strong dispersion suppression, meaning that a first order achromatic correction becomes possible within each single bend. The superconducting magnets enable the combination of the required strong quadrupole fields within the bending magnets. The local dispersion suppression keeps the maximum value of dispersion low. In the design presented here this property has been used to accept a very large energy spread, so that in many cases the transport of a beam with varying energy through the last bending section of the gantry can be performed without adjusting the bending field of this section. The fast energy variation is accomplished by a degrader mounted at an intermediate focus before the last bending section in the gantry. This enables a beam energy modulation on much faster timescale than at the currently existing facilities, so that a scanning of the total tumor volume can be done within several seconds. The large aperture and the strong gradients of the superconducting magnets also help to transport the beam with a relatively large emittance far from the aperture of the magnet, hence minimizing the beam losses in the gantry.

The combination of the strong fields with a large aperture enabled by the superconducting magnets can be used for the beam optics purposes. The mentioned magnet properties could ease the implementation of the upstream scanning, which would help to minimize the gantry radius. The proposed gantry is approximately 3.0 m in radius and 8.5 m in length including an estimated outer magnet radius, but without cryo-modules, free space and eventual shielding. This is to be compared with 3.2 m radius and 8.9 m length of PSI Gantry 2. In the first order calculation the beam spot size and shape are unaltered when the beam is scanned through the final bend, but additional higher order beam optics calculations are essential. The challenges include the correction of chromatic focusing errors and the control of the system's imaging properties at large scanning amplitudes. Beam position and focusing errors depend on the field quality across the large magnet aperture.

Several points have been mentioned, which require further investigation and study. The following list summarizes these points:

List of further studies

- Superconducting magnet design:
  - Cooling of the magnets, incl. during gantry rotation.
  - Mutual influence of the large aperture magnets with a strong field.
  - Superconducting magnet ramping.
  - Quench protection, incl. the effect of beam losses.
  - Optimal superconductor material:
    - The thermal acceptance.
    - Mechanical properties.
    - Cost of the material.
  - Effect of fringe fields on the beam optics and the stray fields at the patient location.
- Beam optics:
  - Chromatic effects at large transverse scanning angle.
  - Higher order imaging effects due to large beam divergence and large magnet aperture.
  - Higher order imaging at large beam momentum spread and large momentum offset.
  - Tracking of scattering in the range shifter, nozzle and the air.
  - Possibilities to increase SAD.
- Mechanics
  - Design of mechanical support.
  - Search for a layout allowing convenient imaging and patient table positioning

We have reported on the results of a preliminary study. While challenges remain to be worked out, it is believed that no insurmountable problems have been identified in the still open issues. Therefore the here presented gantry design of the new concept (degrader on gantry in combination with an energy acceptance such to allow a constant magnetic field during the treatment) can be considered a promising next step in the development of particle therapy.

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