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# 'RIESENRAD' ION GANTRY FOR HADRON THERAPY PART III*

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## *Abstract*

When using accelerator beams for cancer therapy, the three-dimensional freedom afforded by a *gantry* helps the treatment planner to spread out surface doses, avoid directions that intercept vital organs and irradiate a volume that is conformal with the tumour. The general preference is for an iso-centric gantry turning 360° in the vertical plane around the patient bed with sufficient space to be able to orientate the patient through 360° in the horizontal plane. For hadrontherapy, gantries are impressive structures of the order of 10 m in diameter and 100 tons in weight and to date only proton gantries have been demonstrated to operate satisfactorily. The increased magnetic rigidity of say carbon ions will make ion gantries more difficult and costly to build. For this reason, exo-centric gantries and, in particular the so-called 'Riesenrad' gantry with a single 90° bending magnet, merit further attention. The power consumption is reduced and the heavy magnets with their counterbalance weight are reduced and are kept close to the axis. The treatment room, which is lighter, is positioned at a larger radius, but only the patient bed requires careful alignment. An optics module called a 'rotator' is needed to match an incoming dispersion vector to the gantry in order to have an achromatic beam at the patient. A practical design is described that assumes the beam is derived from a slow-extraction scheme in a synchrotron and that the beam sizes are controlled by modules in the transfer line. Magnetic scanning is integrated into the gantry optics for both transverse directions.

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# 1 INTRODUCTION

The quality of cancer treatment with accelerator beams is greatly improved by the use of a *gantry* that can deliver the beam to the patient from any wanted direction. The three-dimensional freedom afforded by the gantry allows the treatment planner to spread out surface doses, avoid directions that intercept vital organs and to irradiate a volume that is conformal with the tumour. The general preference is for an iso-centric gantry turning 360° in the vertical plane around the patient position with sufficient space about the iso-centre to be able to orientate the patient through 360° in the horizontal plane. For hadrontherapy, gantries are impressive structures of the order of 10 m in diameter and 100 tons in weight and maintaining sub-millimetre precision for the transfer line elements supported by these structures is not trivial. To date, only proton gantries have been built and demonstrated to function satisfactorily [1,2,3]. The increase in beam rigidity from protons to say carbon ions (~ ×5) will make these gantries more difficult and costly to build.

In an isocentric gantry the beam must first be bent away from the axis before being directed back, towards the patient. Typically, a conical gantry requires 180° (45° up and 135° down) of bending. In order to limit the overall diameter, the scanning magnets are best positioned upstream of the last dipole. However, this requires a large aperture magnet which creates the problem of supporting, with high precision, a large mass in an off-axis position. For this reason, exo-centric gantries and, in particular the so-called 'Riesenrad' gantry, merit further attention [3]. In this structure, the geometry is inverted, by deflecting the beam with a single 90° dipole which rotates about the incoming (horizontal) beam axis. The patient is positioned in a lightweight treatment room at a larger radius, as indicated in Figure 1.

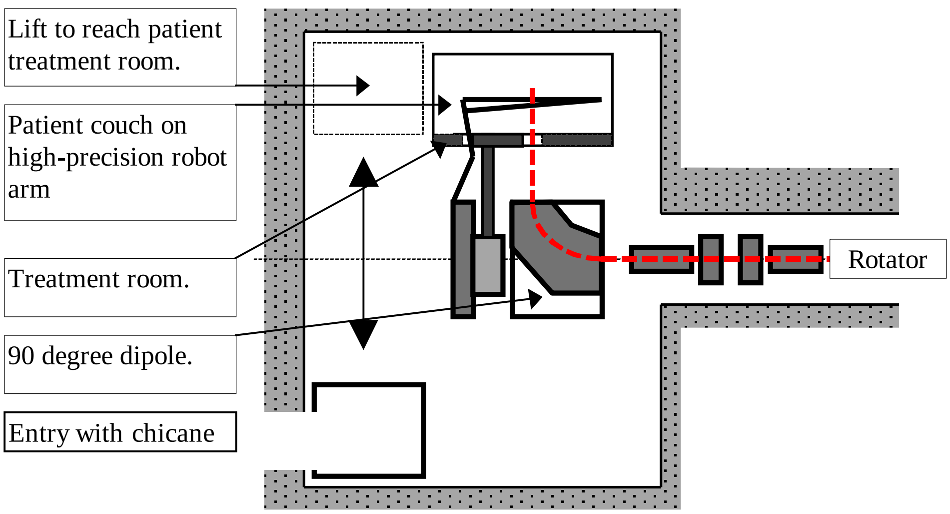



Figure 1: Schematic layout of a 'Riesenrad' ion gantry.

This configuration has the following advantages:

- All magnetic equipment is on or close to the axis which reduces mechanical deflections and thermal effects in the support structure.
- The total beam bending angle is an absolute minimum which implies minimum power consumption.
- Only the patient couch requires high precision alignment relative to the final dipole; the positioning of the treatment room is not critical.

A practical design for a ‘Riesenrad’ gantry for carbon ions is described. It is assumed that the beam is derived from a third-integer resonant-slow extraction scheme in a synchrotron and that the beam sizes are controlled by dedicated modules in the transfer line [4]. Magnetic scanning is integrated into the gantry optics for both transverse directions. The design of the scanning system and the main bend is discussed.

## 2 OPTICAL CONCEPT

### *2.1 Matching*

Unlike conventional gantries in cyclotron based facilities, the ‘Riesenrad’ ion gantry is strongly influenced by two factors:

Firstly, the slow extraction from a synchrotron delivers a beam with high asymmetry between the transverse phase spaces [5]. In the vertical plane, the phase-space distribution is essentially the same as that circulating in the ring. In the horizontal plane, the distribution is near-rectangular, with a uniform particle density and has a small emittance. The rectangular shape in horizontal phase space has become known as the ‘bar’ of charge, which is considered as a diameter across an unfilled ellipse of much large area, as shown in Figure 2.

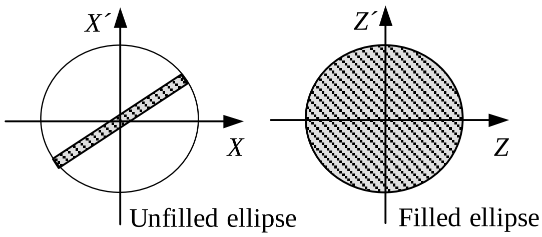



Figure 2: Normalised phase-space plots of a third-integer slow-extracted beam.

Secondly, the dispersion created by the 90° bending magnet has to be compensated by an incoming dispersion vector so as to deliver an achromatic beam to the patient. This dispersion vector is conveniently created by the bending module that deviates the beam from the main extraction line towards the gantry room.

With normal matching techniques it would not be possible to create an achromatic beam at the patient, independent of the gantry rotation angle. An optics module called a rotator, that maps the incoming beam as well as the dispersion vector to the normal modes of the gantry is used [6,7]. The rotator is a quadrupole lattice with phase advances of $2\pi$ and $\pi$ and one-to-one and one-to-minus one transfer matrices for the horizontal and vertical planes respectively that is rotated physically by half the gantry angle.

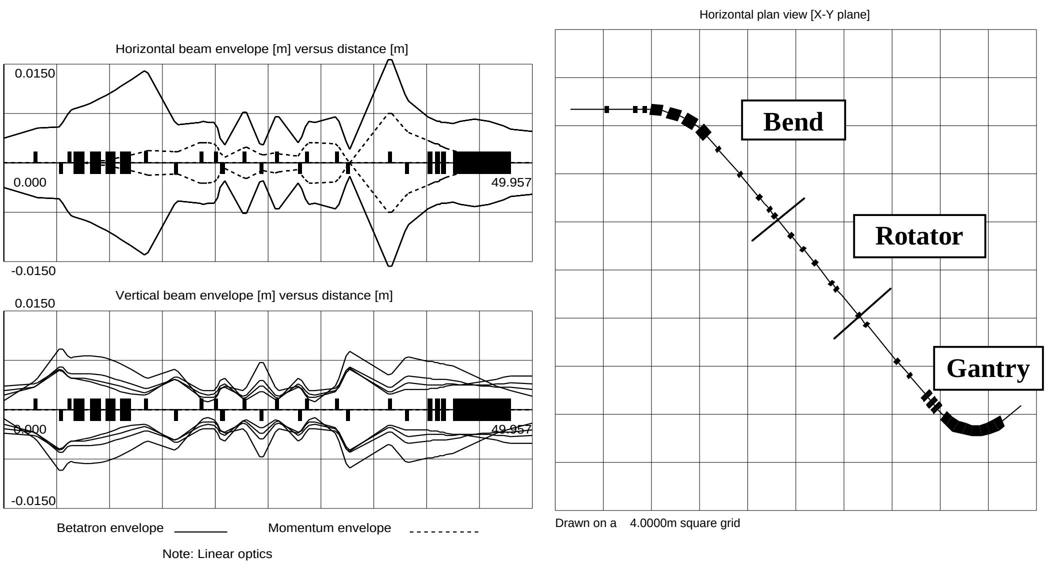
*2.2 Beam size control*

In the present design, it is assumed that the beam size control is performed in the upstream transfer line and the details of this philosophy are given in [4]. In brief, the bending module, rotator and gantry have constant focusing gradients independent of the beam sizes. In the horizontal plane, the betatron amplitude function is constant and the beam size is changed by rotating the 'bar' of charge with an external module called phase shifter. In the vertical plane the overall transfer matrix is a constant magnification telescope. The incoming betatron amplitude function is therefore mapped directly to the patient with a fixed magnification. The beam size is then changed by altering the vertical betatron amplitude function upstream with a module called a 'stepper'.

*2.3 'Riesenrad' optics*

The philosophy for the beam size control and the dispersion makes it necessary to design the bending module, rotator and gantry as an integral unit. In the example given, the lattice functions at the entry to the bending module are: $\beta_x = 3$ m, 1 m $< \beta_z <$ 12.5 m, $\alpha_x = \alpha_z = 0$ and zero dispersion. The medical specifications require beam sizes of 4 to 10 mm at full width half maximum.

For the horizontal plane, with an unfilled emittance ellipse of 4.5 $\pi$ mm mrad (total geometric, energy independent), the betatron amplitude function required at the patient is 5 m (remember, in the horizontal plane the beam distribution is rectangular as mentioned in Section 2.1). In the vertical plane, the beam size is controlled in the usual way by varying the betatron amplitude function. In the example chosen, the normalised rms emittance is 0.75 $\pi$ mm mrad. The adiabatic damping over the energy range 120 to 400 MeV/u together with the medical specifications require a range of 2 to 25 m for the vertical betatron amplitude function at the patient. Thus the bending module, rotator and gantry must provide a magnification by a factor two. Figure 3 shows the beam sizes as described above and the geometry with the rotator and gantry at zero angle (horizontal plane).



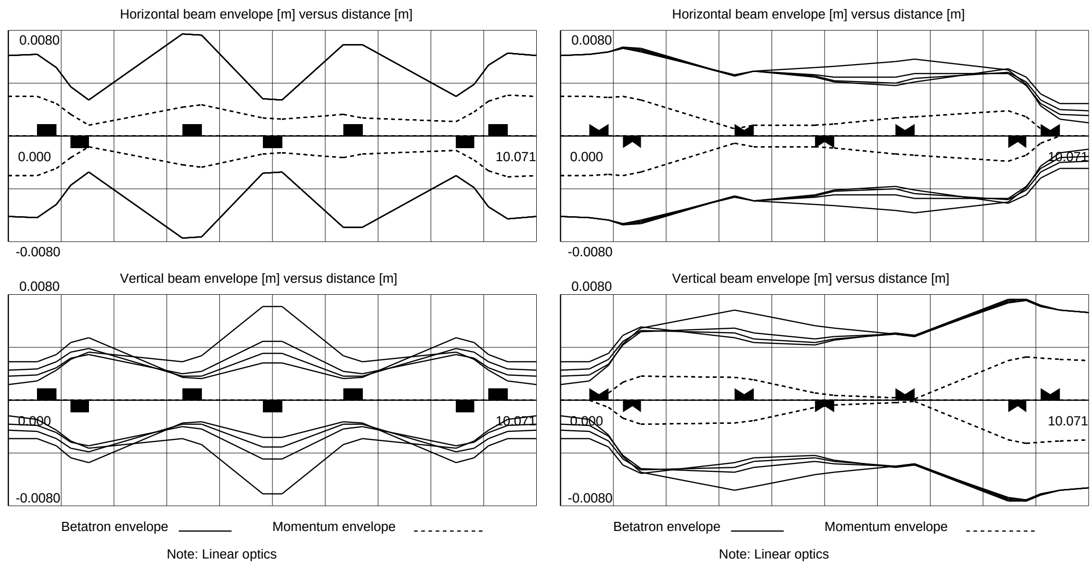

(a) (b)

Figure 3: (a) Beam sizes in bending module, rotator and gantry (b) Geometry of the structure.

Note that in the horizontal plane the curves represents the maximum beam size, with the 'bar' of charge aligned with the horizontal axis in phase space. Since the 'bar' is rotated by the phase shifter, the actual beam size will vary throughout the line but still within this envelope. In the vertical plane the beam envelopes for 4, 6, 8 and 10 mm beam sizes at the patient are shown.

### *2.4 Rotator optics*

When the gantry turns, the situation is basically the same except that inside the rotator the beam is coupled and the beam sizes change locally. Figure 4 shows the beam sizes in the rotator module at 0° (gantry at 0°, rotator at 0°) and when rotating the beam by 90° (i.e. gantry at 90°, rotator at 45°). The incoming beam sizes correspond to those in Figure 3. In Figure 4a, the rotator is acting as a normal transfer line since the gantry is at zero angle. In Figure 4b, the dispersion and the beam sizes exchange planes and are mapped to the gantry which is now at an angle of 90°. In this case, the rotator is at 45° and the quadrupoles appear as skew quadrupoles. A more detailed analysis at intermediate angles shows that the beam remains well-behaved during the rotation.




(a) (b)

Figure 4: Beams sizes inside the rotator, (a) rotator at 0°, beam rotation zero, (b) rotator at 45°, beam rotation 90°.

### *2.5 Scanning optics*

The gantry structure is designed to accept a beam scanning system that will provide an irradiation field of $20 \times 18$ cm$^2$ at the patient. In order to keep the overall size of the installation within reasonable limits, the scanning magnets are positioned upstream of the main 90° dipole. This reduces not only the strength of the scanning magnets (longer lever arm) but also allows the patient to be positioned closer to the main dipole. The optics of the gantry and the positions of the scanning magnets were chosen to ensure parallel scanning in the horizontal plane and near parallel (divergence < 0.3°) in the vertical plane. For this the phase advances from the scanning magnets to the treatment volume are close to π/2. In the horizontal plane, a kick of ±26.18 mrad and, in the vertical plane, a kick of ±12.28 mrad are required. This is achieved by using three identical units, two for the horizontal and one for the vertical deflection. Figure 5 shows the maximum scanning positions in the horizontal and vertical planes respectively for the smallest beam size.

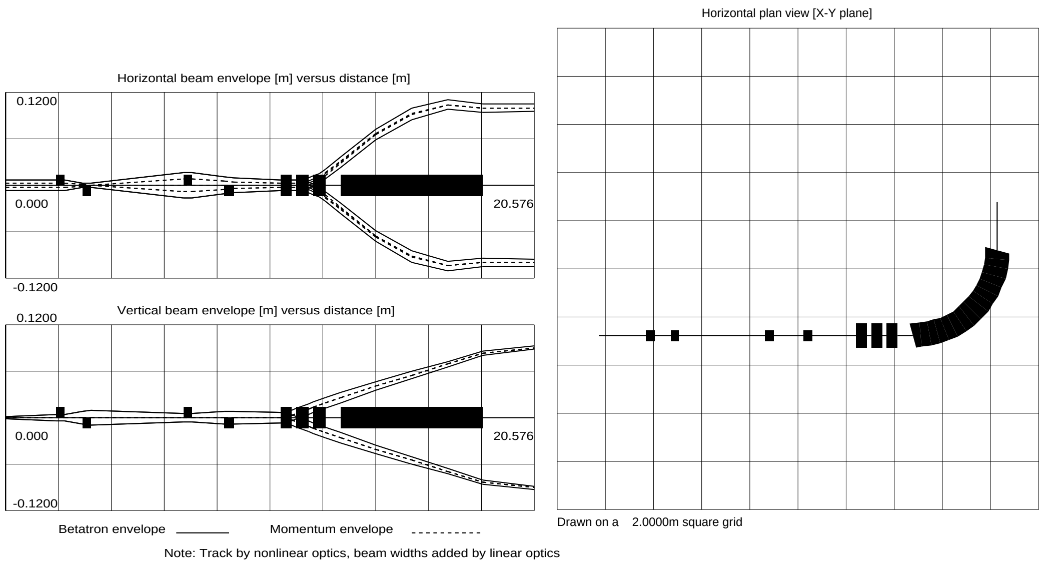



Figure 5: Magnetic scanning system with maximum deflection (field size 20 x 18 $cm^2$) and geometry of gantry in horizontal position.

## 3 MAGNETS AND CONVERTERS

### *3.1 Scanning system*

The three scanning magnets are identical. The design is based on the largest aperture requirement and is of the window-frame type. The yokes are laminated and the vacuum chamber should be either ceramic or thin-walled in a high-resistivity non-magnetic steel. Since the maximum fields are identical in all units, the scanning system can be run by three identical power converters. The system is designed to give a maximum scanning speed of 10 m/s for carbon ions at the highest energy. The parameters of the magnets and the bipolar power converters are summarised in Table1.

| **Parameter** | **Value** | | **Parameter** | **Value** |
|---|---|---|---|---|
| Aperture, $w \times h$ [$cm^2$] | 14 × 10 | | Current range I [A] | ±306 |
| Good-field $w \times h$ [$cm^2$] | 6 × 8 | | dI/dt max [kA/s] | 30.6 |
| Field range $B$ [T] | ±0.2 | | Voltage max. dc [V] | 4.01 |
| d$B$/d$t$ max. [T/s] | 20 | | Voltage max. ac [V] | 74.5 |
| Eff. magnetic length [m] | 0.45 | | Max. power [kW] | 2.39 |
| Inductance [mH] | 2.43 | | Converter type | Switch mode |
| Resistance [mΩ] | 13.29 | | Switching frequency [kHz] | 50 kHz |

Table 1: Scanning magnet and power converter parameters.

### *3.2 Main 90° bending magnet*

In order to keep the gantry as compact as possible, the 90 degree bend is built as a single unit with edge angles to help with the betatron matching and, in particular, to provide the optimum phase advances for the scanning system. The beam position at the patient must remain within ±10 % of the smallest spot size (i.e. ±0.4 mm) with all errors sources considered. If the field quality in the main dipole were the only source of error, this would translate into a specification for the field uniformity of $\Delta B/B < 8{\cdot}10^{-5}$. This is considered to be impractical in such a large, high-field magnet. Equally, standard measurement systems cannot assure this precision. Hence it will be necessary to use the beam with a position monitor to calibrate the system and create a correction map which is then applied via the scanning magnets* . Based on three-dimensional magnet simulations, the design value for the integrated field quality was chosen as $\Delta(B\ell)/(B\ell) < 5{\cdot}10^{-4}$. Figure 6 shows a perspective view of the magnet.

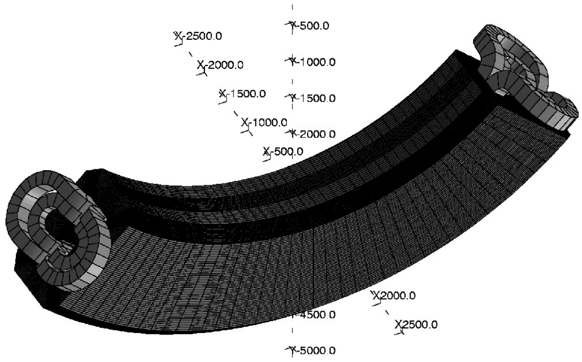



Figure 6: Perspective view of the 90° main bending magnet [8].

An indication of the field quality is given in Figure 7, at the maximum field level of 1.8 T in the centre of the magnet. Although the return yokes have been designed to take account of the magnet curvature, a slight asymmetry of the curves is still visible due to the uneven saturation of the two sides, which is detected by the three-dimensional calculation of the magnet.

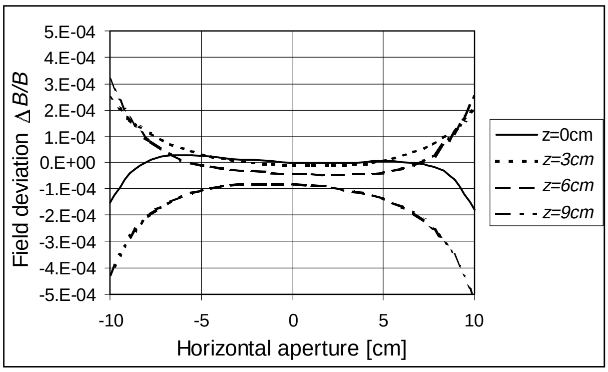



Figure 7: Field quality across the aperture at different vertical positions in the centre of the magnet.

* However, it should be noted that this method implies that the stability and reproducibility of the field must still be better than $8{\cdot}10^{-5}$ to make the correction via the scanning magnets valid.

In contrast to the scanning magnets, the gantry units will only need to change slowly as the treatment progresses from distal (high energy) to proximal (lower energy) slices of the tumour. Thus the power converter is essentially a dc unit. However, the precision and ripple must be $\Delta I/I_{max} < 4{\cdot}10^{-5}$ in order to respect the specification for the stability of the beam spot, as mentioned above. The principal parameters of the magnet and its power converter are given in Table 2.

| Parameter | Value | | Parameter | Value |
|---|---|---|---|---|
| Aperture, $w \times h$ [cm$^2$] | 26 × 20 | | Max. current [A] | 4150 |
| Good-field, $w \times h$ [cm$^2$] | 20 × 18 | | Current range [A] | 2128-4150 |
| Max. field [T] | 1.8 | | Max. current ramp [A/s] | 300 |
| Bending radius [m] | 3.53 | | Current ripple $\Delta I/I_{max}$ | $\pm 4{\cdot}10^{-5}$ |
| Iron weight [t] | 51.5 | | Current precision $\Delta I/I_{max}$ | $\pm 4{\cdot}10^{-5}$ |
| Copper weight [t] | 7.7 | | Max. voltage dc [V] | 82 |
| Inductance [mH] | 86.3 | | Max. voltage ac [V] | 26 |
| Resistance [mΩ] | 19.6 | | Max. power dc [kW] | 340 |

Table 2: Main 90° dipole magnet and power converter parameters.

## 7 CONCLUSIONS

A complete solution for a light-ion gantry, operating with a synchrotron using resonant slow extraction has been presented. This solution, known as the ‘Riesenrad’, is an exo-centric gantry that keeps the heavy magnetic equipment on, or close to, the axis while the patient is positioned in a light treatment room which can move at a larger radius. Only the patient couch inside the treatment room needs precision alignment with respect to the magnetic equipment on the axis. Such a configuration with a single 90° dipole magnet offers the minimum possible bending and hence reduces weight and power dissipation that are the major concerns for gantry designers. The advantage of the structure is that the impact of mechanical and thermal distortions of the support structure is reduced as compared to an equivalent iso-centric gantry.

Optically, the ‘Riesenrad’ differs from an iso-centric gantry since it is not possible with a single 90° magnet to create a closed-dispersion bump within the gantry itself. Therefore, the dispersion caused by this dipole has to be corrected upstream in the transfer line in order to deliver an achromatic beam to the patient. The dispersion matching is trivial as long as the gantry is not rotated (horizontal plane). However, with the gantry at an angle, the dispersion provided in the horizontal plane has to be rotated to the new gantry angle. This is solved by the use of a rotator that matches the fixed transfer line exactly to the rotating gantry whatever the angle. It should be noted that the rotator also transfers the phase spaces to the rotated gantry which is essential for the asymmetric phase spaces of the slow-extracted beam. The only constraint is that the transfer line, rotator and gantry have to be designed as an integrated module. The gantry is designed to give parallel scanning in both transverse planes. The beam size control is handled by the upstream transfer line.

The ‘Riesenrad’ concept is motivated by engineering constraints but should also have a cost and size advantage over iso-centric gantries. Although the configuration is unfamiliar it satisfies all medical specifications.

## ACKNOWLEDGEMENTS

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