LARGE MEDICAL GANTRIES

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Abstract

A combination of a Medical Gantry and Patient Positioning System is required to direct a particle therapy beam to a supine patient from a variety of angles. The requirements of beam direction and size for precision beam treatment result in stringent criteria for the magnetics and mechanical structure of the gantry. In this paper the requirements of large gantries for ion therapy will be discussed. A variety of ion beam gantries for beam energies up to and beyond 250 MeV have been designed, constructed and are being planned. The mechanical alignment requirements will be discussed and compared with actual performance where available. The beam optics requirements will also be discussed and compared amongst different gantry designs

I. INTRODUCTION

Since Wilson wrote the classic paper in 1946 [1] discussing the potential application of fast protons for radiotherapy, the use of ion beams for therapy has been proven an effective tool for a variety of treatments[2]. The development of advanced treatment planning codes[3] has allowed for the conceptualization of treatments that maximize dose delivered to the necessary volume and minimize the dose received by possible nearby critical structures. In parallel with that development, the hardware required to realize advanced treatment techniques has also been developing. The Gantry includes the structure which carries the necessary devices so as to allow the therapy beam to be directed from virtually any angle in a plane.

One eventual aim of a gantry system, and the most demanding, is to be capable of dynamic therapy. This technique requires the beam angle to be adjusted during a treatment, thus minimizing the beam exposure at the surface while concentrating on the target volume. One of the key advantages of protons is the precision with which one can potentially deliver a treatment. The treatment area can be selectively treated while minimizing the dose to a critical structure which can be a fraction of millimeter distant. To realize this possible precision, the hardware, for protons in particular, must be appropriately designed.

II. OPERATING GANTRIES and GANTRIES UNDER CONSTRUCTION

There are presently several ‘high energy’ gantries in operation or under construction. This paper does not include discussion of the conventional electron linear accelerator gantry system, which are widespread throughout the world. Rather the concentration is on gantries for therapy with heavier particles.

Consideration of Gantry geometry possibilities leads to several paths. It is interesting to note that, especially in the case of proton and neutron gantries, practically all these paths have been, or are being pursued. There has not developed a consensus about a preferred scheme, other than finding one which results in the least expensive overall system and yet meets all the clinical requirements.

One question to ask is whether the accelerator and the gantry should be decoupled or physically integrated. Of course the latter requires an accelerator capable of being attached to a gantry structure and rotated through 360°. This in fact has been done at Harper Hospital in Detroit[4] for neutron therapy, using superconducting cyclotron technology. There is no proven method to date of miniaturizing an accelerator for high energy proton therapy although a couple have been proposed. Another advantage of coupling the accelerator with the gantry is that the accelerator typically produces a non-symmetric phase space, which can be used if it is integrated in the gantry. A main reason to separate the gantry is to feed a beam delivery system with multiple beam lines.

Another issue is the relative location of the gantry and patient positioner relative to the isocenter. Should the patient positioner be required to move as the gantry rotates as implemented at the PSI [5]. In the case where the isocenter is a point along the axis of rotation of the gantry this is not necessary, but that requires the gantry to be large.

Last in this non-exhaustive list, is the question of how the beam will be shaped (spread) to match the target. The use of passive scattering vs. active scanning may affect the gantry geometry. In particular, the location of the scattering or scanning devices may be an important factor. It is desirable to have a large effective source to isocenter distance. This can be achieved by a large physical separation between the beam spreading system and the isocenter after the last gantry dipole, or by incorporating these systems within the gantry optics and using close to point to parallel focusing from the spreaders to the gantry output (as at PSI). This latter technique tends to require large apertures and therefore heavy dipole(s).

All of the above considerations have merit one way or the other, and have in fact been implemented one way or the other. Production of more than one of a kind seems to be left for commercial development. This was done in the case of neutron gantries by Scanditronix (Scand) [6,7,8] and The Cyclotron Corporation (TCC) [9,10]. Several gantries were built. For protons, Science Application International Corporation (SAIC) has built one for the Loma Linda University Medical Center (LLUMC) [11] and a team of Ion Beam Application (IBA) and General Atomics (GA) is presently building one for Massachusetts General Hospital (MGH) [12].

Table 1 contains some parameters for large gantries for a selection of facilities with or building them. Presently, large gantries are used for delivery of neutrons, or protons for therapy. To date, the large and expensive requirements for high energy *heavy ion* gantries have prevented them from being used in these applications. Superconducting technology has yet to be applied to this.

III. MAGNETIC GANTRY REQUIREMENTS

A number of parameters will determine the features and layout of the gantry without an integrated accelerator. These include:

1. *The distance between the last bending magnet and the patient.*
   The beam must be uniformly spread out to the desired treatment cross section. With the spreading devices after the last dipole, the effective Source to Isocenter Axis distance (SAD) will be smaller than the distance from the last magnet to isocenter. The spreading can be done before the last magnet; however that will increase the magnet size significantly. This distance must be large enough to reduce to an acceptable level, the ratio of the skin dose to the target dose, to provide the desired field size while not losing too much energy in the spreading..
2. *The space required for patient movement about the isocenter*.
   This will determine the minimum ‘gantry stay clear’ envelope for the patient support system and alignment devices.
3. *The magneto-optical properties and beam trajectory requirements*.
   This includes the beam particle and energy. It is related to beam spreading methods to be used and optimization of the gantry size and weight. Beam spreading methods can include Scattering and Scanning or combinations of both.
4. *Input Beam Requirements*
   Input beam should be achromatic, rotationally invariant (at least for scanning) at the coupling point to the gantry to achieve invariant beam properties independent of gantry angle.
5. *Mechanical Properties/Isocenter Requirements*
   This category includes mechanical deflections and reproducible, as well as non-reproducible, positioning errors, particularly of beam sensing

Table 1 - Facilities with Large Gantries

Gantry -Max Parameter	Essen	Riyadh	Clatterbridge	Seattle	NAC	Detroit	LLUMC	PSI	NPTC PLAN
Date	1982	1982	1985	1984	1988	1990	1992	1995	~1998
Incident Particle	d	p,d	p	p	p	d	p	p	p
Energy (MeV)	14.3	26,15	62	50.5	66	50	250	270	235
Treatment Particle	n	n	n	n	n	n	p	p	p
Energy (MeV)	6	10	26.9	21.3	28.8	21	270	85-270	70-235
SAD (m)	1.25	1.5	1.5	1.5	1.5	1.83	2.75	∞	2.25
Weight (tons)	5	5	55	43	50	50	96	120	~100
Rotation (deg)	220	240	270	380	370	360	370	370	370
FieldSize (cm)	20x20	20x20	30x30	33x29	29x29	20x30	40 diam	20x↔	30x40
Isocentricity (diam-mm)	2	2	4	4	2.5	2	0.7x1.6	2	2
Phase Space (π-mm•mr)							2	40	30/18
Manufacturer	TCC	TCC	Scand	Scand.	Scand	MSU	SAIC	PSI	GA
Beam delivery	Sn=Scan, Sc=Scatter, W=Wobble ↔=table move						**Sc/(Sn)**	***Sn***	Sc/W/

devices and resulting in beam pointing errors at the isocenter, during gantry rotation. The vacuum requirements are modest at the 10^{-2} torr level.

6. *Magnetic field switching speed*.

   For some applications it is required to change the beam energy transported by the gantry while treating a patient, either for depth modulation, or for changing treatment portals. The magnets on the gantry must be capable of this.

IV. NEUTRON THERAPY GANTRY

Neutron Gantries transport either protons or deuterons to a target (normally Beryllium) in the treatment head. The beam is converted to a neutron beam. The beam is 'modified' with devices such as beam flattening filters, wedges, and collimators to set the desired field size.

One example is operating at the University of Washington at Seattle [7]. This Scanditronix gantry is designed to provide a beam as symmetrical as possible in both transverse planes. The magnification of the optical system is near unity resulting in relatively weaker focusing requirements. The focusing is also designed to be point to point imaging. This has the effect of minimizing the beam trajectory angular requirements at the entrance to the gantry.

There is sufficient steering capability in the beam line to adjust the beam trajectory entering the gantry properly and there is a beam profile monitor at the entrance to the gantry. A corrector dipole in the gantry accounts for gantry deflection effects for different gantry rotation.

The neutron gantry does not have to be achromatic, since the neutron conversion target is after the last dipole and the resultant neutron spectrum has a large spread in energy. The Scanditronix design incorporates an xy wobbler which produces a circular wobbling pattern within the gantry primarily to reduce the thermal effects of the approximately 50uA beam on the conversion target. Note that the patient positioning device rotates about the isocenter below the gantry head.

The gantry head is cantilevered out and therefore the chair can be mounted below the path of gantry rotation. The beam points to the nominal isocenter within a circle of confusion of diameter 4mm. The reproducibility is better than this but not required for neutron treatments. The TCC gantries are smaller and have a reduced isocenter diameter.

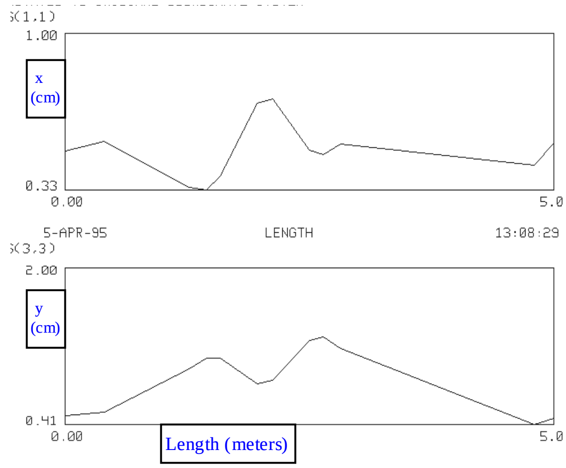

Figure 1 Beam envelope through Neutron Gantry

V. THE CORKSCREW GANTRY

In this gantry geometry, introduced by Enge and Koehler [13], the major bending occurs in the plane of rotation so that these magnets do not sweep out a large volume of space as the gantry is rotated as shown in figure 2. The three dimensional aspect of the beam trajectory led to the name "*Corkscrew* Gantry".. The result is that the building to house the gantry can be smaller. Disadvantages are the reduced volume available for the patient positioner motion and for access to patient within the gantry open space, and the greater amount of bend..

The Corkscrew gantry has one set of two 45° dipoles with a quadrupole triplet in between and another set of 135° dipoles and triplet. The first set bends the beam out of the plane of the beam line. The second set bends the beam in the orthogonal plane toward the isocenter. Since this system involves two orthogonal bends, and the whole system is required to be achromatic, then each set must be

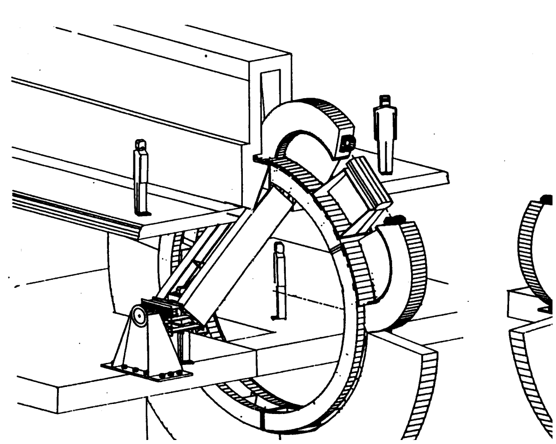

Figure 2. Schematic of Koehler Corkscrew Gantry

achromatic. In order to achieve achromaticity, four parameters are required, leaving two more parameters in addition to the pole edge rotations. This is adequate to ensure a reasonable beam size to be transmitted, but does not leave much flexibility for subsequent optics changes. The corkscrew gantry implemented at Loma Linda University Medical Center uses four quadrupoles per bend plane.

Figure 3 shows the results of a TRANSPORT calculation of the beam envelope in the Koehler gantry system. This optics tune produces a waist at 50cm from the exit of the last dipole. The input is presumed to be a symmetric beam with up to 24π mm-mrad phase space area and ±0.5% ΔP/P.

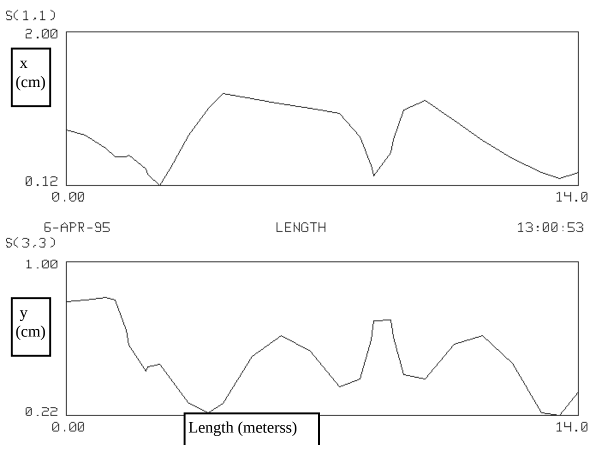

Figure 3. Beam envelope through Corkscrew Gantry

In the LLUMC implementation the gantry is primarily supported by large rings. The bearing is supported from the floor and one wall by support struts. The gantry is a cone shaped structure made up a of 7 foot circular ring at one end and a 16 foot ring at the larger end. The plates are connected by struts. The assembly is fabricated in sections small enough to transport into the gantry room. The magnets at LLUMC are aligned to 0.2 mm individually and 0.4mm gantry overall. Measurements of beam pointing accuracy to isocenter result in an isoshape of less than 1.6mm diameter [14].

A variation of the Corkscrew gantry is the *Supertwist* gantry suggested by Francis Farley [15]. This starts with the corkscrew physical layout concept and departs from the two orthogonal bend solution by stretching out the corkscrew in such a way that the total gantry length is longer and each magnet twists the beam in a trajectory through this path. Much of the focusing is done with pole edge rotations and the resulting system is achromatic at the end.

VI. LARGE THROW GANTRY

A conventional gantry is being built for MGH/NPTC [16] by IBA and GA. The rotating elements begins with 4 quadrupoles in the plane of the beam switchyard which rotate with the gantry. They match the beam from the symmetric waist produced by the beam line to the gantry optics with an emittance of up to 32 mm-mrad..

The beam is deflected through 45°, focused by five quadrupoles before it is bent through 135° and directed towards the isocenter. The distance from the output of the 135° dipole to the isocenter is 3.0 meters. The gantry quadrupoles can be adjusted to produce a waist with a diameter of 12mm at the isocenter. For scattering and wobbling, the quadrupoles can be tuned to produce a 10mm radius waist at the center of the range modulator which is about 20cm from the last dipole as shown in figure 4.

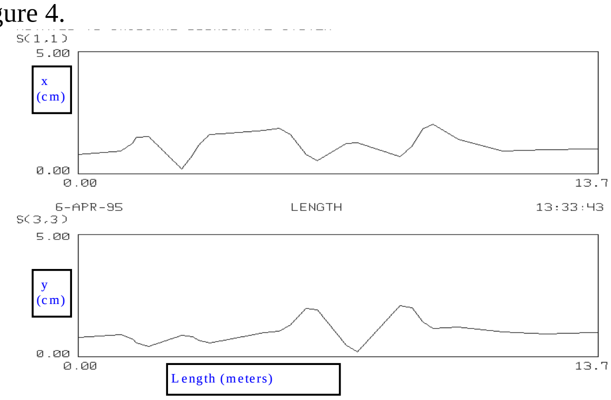

Figure 4. Beam envelope through NPTC Gantry

There is sufficient space in the gantry to include beam position and profile monitors which are capable of determining and correcting the beam trajectory angle and position to the required tolerance within the beam modification elements of the Nozzle. This tolerance is basically sub millimeter precision at the location of the scatters for scattering, or at the isocenter for scanning. The magnets accept a momentum spread of ±0.5% ΔP/P.

The rotating structure utilizes a configuration of rings, truss, and shell elements to support the magnets in a "space frame". The structure is stiff and engineered to minimize its weight. The front ring is axially constrained. The truss elements are removable for ease of transport, assembly and possible repair. The structure is supported on both rings using a "wiffle-tree" assembly. Figure 5 shows a schematic view.

Gantry structure parts deflect under load and this deflection varies with rotation angle. The rings deform and the nozzle bends. The important quantity to consider, since

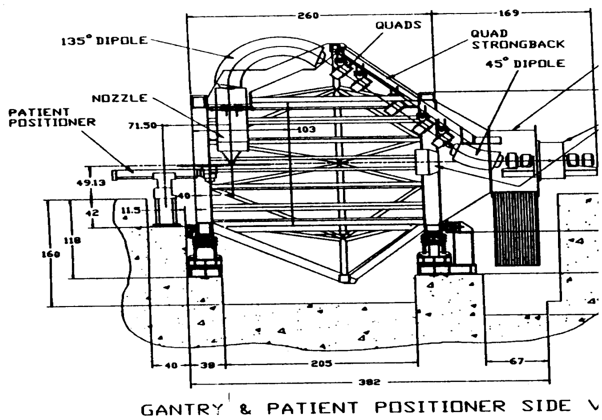

Figure 5. NPTC Gantry Layout

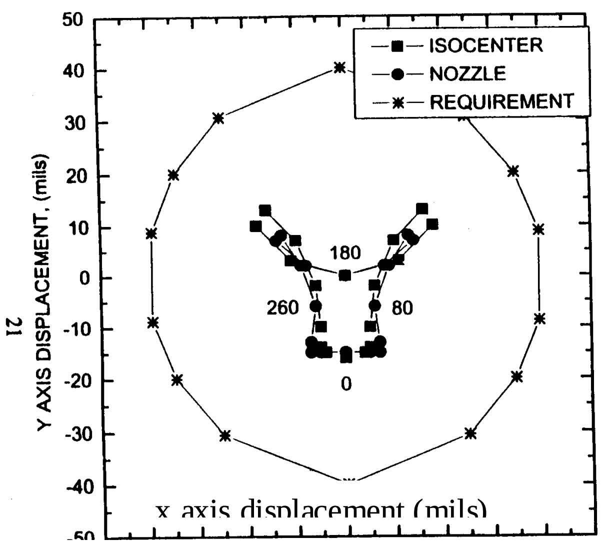

Figure 6. Calculated Nozzle Point Error

beam monitors are mounted in the nozzle, is the change in the beam position relative to the ideal isocenter as a function of gantry rotation. Figure 6 shows the results of a calculation of this quantity for the NPTC gantry.

VII. COMPACT GANTRY

The PSI Gantry design [5] results from the special requirements of the beam at PSI and a general intention to design a gantry for potential users with inadequate space for a large throw gantry. A large beam phase space results from degrading the beam energy significantly (from 590 MeV to between 270 and 85 MeV) just before the gantry. The gantry is a '*compact*' style which spans a diameter of only 4m and is designed to accommodate the spot scanning technique [5] to be use at PSI. The end of the last dipole coincides with the gantry axis of rotation. The patient table is mounted directly on the front wheel of the gantry and moves with the gantry to keep the distance between the isocenter and gantry axis constant. The beam is focused to achieve unity magnification and achromaticity. The apertures can accommodate up to ±1% ΔP/P. There are seven quadrupoles in this system to achieve the desired conditions. The sweeper magnet is located before the last 90 degree dipole and is the limiting aperture of the system. Another condition requires that the scanning magnet to output optics satisfy point to parallel focusing. This results in a near infinite effective SAD. This allows the distance between the exit of the last dipole and the isocenter to be minimized. This is achieved using pole edge rotations on the last dipole. Sextupole aberrations are present and are corrected by introducing sextupole components in some quadrupoles.

VIII. PATIENT POSITIONER

Although not described in this paper, the patient positioning system is very important. It is half of the overall beam delivery/positioning system and potentially half the source of what could be a sub-mm positioning error budget. A system approach is very important; it must be engineered at the same time as the gantry.

IX. ACKNOWLEDGMENTS

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