PRELIMINARY DESIGN OF A REDUCED COST PROTON THERAPY FACILITY USING A COMPACT, HIGH FIELD ISOCHRONOUS CYCLOTRON.

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ABSTRACT

It is generally agreed that, among the different kinds of radiations usable for radio-therapy, high energy proton beams exhibit the best ballistic specificity. However, the development of proton therapy has been hindered by the size, cost and complexity of high energy accelerators. We have therefore tried to design not only an accelerator, but a complete proton therapy facility where the size, the investment, the complexity and the cost of operation would be minimized.

For a clinical therapy unit, reliability is of paramount importance, and we believe that the simplicity and sturdiness of the design is the key to reliability. This is why we selected a non-superconducting isochronous cyclotron as accelerator. The magnet is a high field (3.09 Tesla peak field, 2.165 Tesla average field at extraction) deep valley design, using 190 kW in conventional coils. The complete cyclotron is split in two parts at the median plane. The upper half can be quickly raised by one meter, using hydraulic jacks, allowing an unrestricted access to all cyclotron elements. This design feature, combined with a rapid pump downtime (30 min.) should contribute to maintain downtime of the accelerator to very low values.

The cyclotron would then feed two or three isocentric gantries. IBA has developed a new concept leading to isocentric gantries of reduced size and cost [1].This new gantry design allows to reach infinite "source to patient distance" with a gantry not exceeding 2.5 m total maximum radius.

The dose delivery system has been reviewed to optimize dose accuracy, uniformity and speed of delivery. In the axial motion , the magnetic sweep is controlled by a dose integrator, guaranteeing an uniform dose irrespectively of minor intensity fluctuations.

1. Basic specifications for a proton therapy cyclotron

1.1. Energy

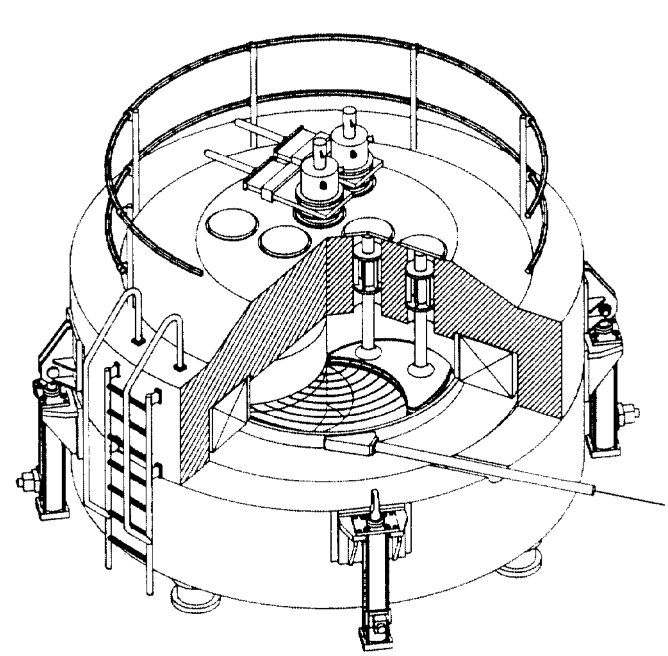
The energy required for a proton therapy cyclotron is entirely defined by the energy-range relation in tissues (quite equivalent to the range in water). The range at 175 MeV is approximately 20 cm, at 200 MeV, 27 cm and at 250 MeV the range is 37 cm. If passive beam flattening filters are used, some extra range must be provided to allow for the energy losses in the filters. However, in this case we intend to use beam scanning and such filters are not needed. Therefore, we feel that a maximum energy of 230 MeV, yielding a range of 32 cm is sufficient.

1.2. Intensities

Intensities needed for a proton therapy facility have been discussed by Goitein [2] [3]. To deliver the doses needed in one fraction (from 14 Gy for Choroidal Melanomas to 1.8 Gy for Pelvic tumors or Hodgkin's Disease) in one minute or less, using a scanned beam technique, an effective dose rate of 5 10^{12} protons/min or 13.5 nA is needed.

Allowing for losses in the beam delivery system (energy selection) the extracted beam intensity should exceed 30 nA, with the internal beam intensity being around 50 nA.

Delivering this kind of intensity with an isochronous cyclotron is, of course a trivial problem : isochronous cyclotrons can very easily achieve extracted beam intensities of tons of microamperes i.e. thousand time more than needed. The fact that we operate the accelerator so much below his maximum capacities is a key factor in achieving a very high level of reliability. This is in contradiction with synchrotrons working with low energy injectors where space charge limits are quite close for the highest intensities needed for radiotherapy.



1.3. Variable versus fixed energy

To locate the Bragg peak at the required depth within the tumor, it is necessary to vary the energy of the incident beam accordingly. This energy variation needs to be very fast, because the Bragg peak is so narrow that the irradiation of one field requires several successive passes with progressively decreasing energies.

In order to optimize the simplicity and reliability of the system, we have selected the solution of a very simple accelerator working continuously at the maximum energy, i.e. 230 MeV. Between the accelerator and the isocentric gantry, the energy is degraded to the value corresponding to the deepest range for the patient being currently irradiated. This first and largest energy reduction is done in such a way as to reduce the beam degradation caused by the scattering and straggling of the beam.

The modulation of the range is controlled by a second cylindrical rotating absorber located at the exit of the isocentric gantry.

2. *The "CYCLONE 230"*

The CYCLONE 230 cyclotron proposed by IBA is an isochronous, high field, non-superconducting cyclotron able to accelerate protons to 230 MeV

2.1. The magnet system

The basic question in designing the magnet for a 230 MeV isochronous cyclotron is whether to use superconducting coils or not. In both solutions the field levels are rather similar : 3.09 Tesla on the hills, 0.985 Tesla in valleys, 2.165 Tesla average field at extraction radius, 1.74 Tesla average field at the center. Such field levels are mostly dictated by the requirement of obtaining adequate vertical focusing without having to use excessive spiraling of the sectors. The dimensions of both designs are therefore rather similar.

The advantages of a superconducting system are the reduction (possibly the suppression) of the iron for the return magnetic flux, a reduction of the electrical power required by the cyclotron in operation and, possibly, a reduction in the total investment costs.

On the disadvantage side of a superconducting system is the mechanical complexity of a split-coil cryostat, with very large forces involved, the potential problems associated with the liquid helium supply (from an attached helium liquefier or from batch transfers) and the very long thermal time constants of such a system : for a large mass, highly thermally insulated cryostat, warm-up or cool-down times are in excess of one week, yielding excessive down-time for a medical device, in case of problem.
The initial choice of the author is, therefore, to go for a classical, non superconducting design. Using the deep-valleys concept pioneered by IBA for non superconducting cyclotrons with the CYCLONE 30 series, it is possible to design a high field, non superconducting magnet for a 230 MeV cyclotron using no more than 185 kW of electrical power. The magnet section is illustrated in fig. 1, and the main parameters of the magnet are described in table 1. A number of steps have been taken to minimize the size and power consumption of the magnet, without compromising the optical qualities of the internal beam. The magnet gap in the hill regions decreases with radius, from 9.6 cm at the center to 0.5 cm at extraction, following an elliptical law [4]. The very small magnet gap at extraction is designed to provide a very sharp radial field fall-off, in order to simplify the beam extraction. For the same reason, the shape of the outer edge of the hills is patterned according to the shape of the orbit close to extraction and is not a circumference centered on the center of the machine like in other cyclotrons.

Another new technical feature of the CYCLONE 230 is that the azimuthal shape of the hills is not rectangular, like in other cyclotrons, but trapezoidal. The wider base of the hills reduces the magnetic saturation. This, in turn, results in an improved flutter and in a decreased power in the magnet coils.

To reduce also the main-coils power, the gap in the valleys was reduced to 60 cm. This value was found to be a reasonable compromise between the reduction of coil power and a convenient dimensioning of the R.F. system in the valley.

Using those values, we found that a good vertical focusing of the beam could be found with a 60° spiral angle of the sectors at extraction.

To facilitate maintenance, the cyclotron is split in two halves at the median plane. Using hydraulic jacks the upper half can be quickly (15') raised by one meter, so as to give an unrestricted access to the cyclotron median plane. Using this construction the cyclotron can be opened for maintenance or repair, closed, pumped down and restarted for operation in less than two hours. This is quite different from current superconducting cyclotron designs where an operation requiring access to the median plane implies a major dismantling and a loss of beam time for several days.

Magnet system

number of sectors		4	
sector angle at the center		36°	
sector angle at extraction		53°	
hill field		3.09	T
valley field		0.985	T
average field at extraction		2.165	T
average field at center		1.739	T
magnetic induction		$5.248\ 10^5$	A.t
apparent current density in coils		153	A/cm^2
actual current density in coils		214	A/cm^2
power per coil		92	kW
weight of one coil		13.3	tons
weight of the iron		165	tons
Spiral angle	at the center	0°	
	at extraction radius	60°	

Table 1

2.2. The R.F. acceleration system

In a previous paper [5], we had proposed a "cyclotron without dees", i.e. a R.F. system where a pair of opposite hills would play the role of R.F. resonating structure. However, such a structure would lead to a sub-optimum design for the magnet and was, therefore, not selected in the present design.

Instead, we present a more classical R.F. system in which two dees are located in two opposite valleys. Those dees resonate on the fourth harmonic of the ions orbital frequency, i.e. 25.527 MHz x 4 = 102.11 MHz. This frequency falls in the F.M. broadcasting frequency range. It is therefore possible to find inexpensive standard broadcasting transmitters meeting the requirements of the system.

In the present design, each dee is supported by four vertical pillars, located symmetrically above and below the median plane. This design optimizes the simplicity, the mechanical stability of the dees and the cooling. In addition, a careful optimization of the diameter and length of each pair of opposed pillars allows to shape the amplitude of dee voltage versus radius.

R.F. system

harmonic mode	H = 2	
dee voltage	100	kV
frequency	102.1	MHz
resonating system	2 dees in opposite valleys	
length of resonator	60	cm
capacitive loading estimated (each resonator)	58	pf
total RF power estimated for 100 kV	65	kW

Table 2

2.3. Vacuum system

Due to the moderate vacuum requirements of a positive ion cyclotron, a quite classical vacuum system can be used. In this case, we plan to reach the operational pressure of 5 10^{-6} mbar by a combination of 4 x 2000 l/sec oil-diffusion pumps located below the cyclotron and 4 x 3000 l/sec valved cryopumps located above the cyclotron. Due to the large pumping speed installed, in regard of the volume to be evacuated and of the surfaces to be outgassed, it is expected that an operational vacuum of less than 10^{-5} mbar can be achieved less than 30 minutes after pump-down if the cyclotron has been vented to dry nitrogen.

2.4. Injection

The protons will be produced at the center of the cyclotron by a hot filament P.I.G. source introduced axially, through an air-lock. The life time of an ion source filament is expected to be in excess of one month, considering the very small arc currents required.
Beam pulsing for the voxel scanning will be obtained by pulsing the arc voltage. Experiments have shown that rise and fall times of
.5 µsec or less could be achieved by this method.

2.5. Extraction

High efficiency extraction of high energy protons of an isochronous cyclotron is well known, but still quite difficult problem. However, in this case, three considerations allow significant simplification of the extraction system :

- the gap of the hills is decreased to 5 mm at extraction, allowing to reach a very steep magnetic field fall-off for the extracted beam
- as the extracted current is low (30 nA), a moderate extraction efficiency can be tolerated without causing an excessive machine activation
- unlike some physics experiments, proton therapy does not require the ultimate in beam emittance or energy resolution.

However, unlike cyclotron for physics, it is very necessary to obtain an extraction tuning that would be very uncritical and highly reproducible from day to day.
It is therefore the opinion of the authors that high extraction efficiency methods that are based on a high turn separation and on a very critical turn positioning in respect to the extraction devices should not be used in this case.
On the opposite, we plan to allow the acceleration of a rather large radial emittance and of a wide phase angle. The beam can then be considered as radially continuous at extraction, without any turn structure apparent anymore.

The extraction system based on such a beam structure has obviously a lower efficiency but this lower efficiency is obtained by very uncritical tunings.
The extraction system considered would use a septum magnet, followed by passive gradient correctors.

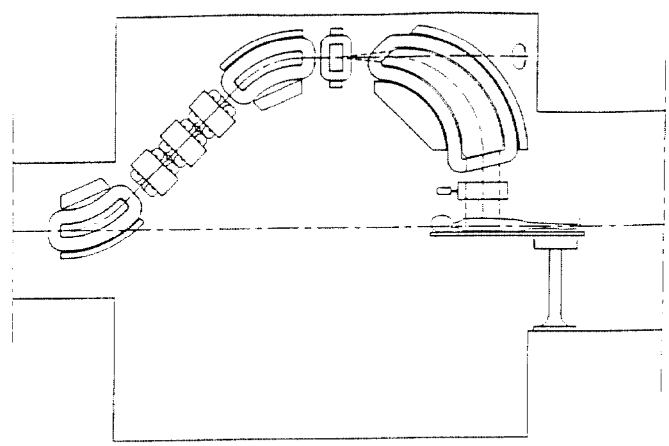
2.6. Control

The proposed cyclotron is a very simple, stable, uncritical and continuous device. The level of control required to achieve unattended operation of the cyclotron is therefore minimal, and the cyclotron should be considered as a mere accessory of the beam delivery system.

For the cyclotron control we plan to use a high level PLC based control system, basically similar to the system used in the lower energy CYCLONE 30 series. The human interface is made through graphical displays on a color screen, a keyboard and two virtual knobs (optical encoders) that can be allocated by software to any variable cyclotron parameter. The normal cyclotron operation is foreseen to be unattended, the cyclotron control being normally relayed to the beam delivery control desk.

3. *Isocentric gantry*

As a part of this proton therapy facility design, IBA has developed the concept of a more compact and simple isocentric gantry. This gantry is more fully described in another paper [1]. To cover the irradiation field, the beam is scanned in three dimensions. Range scanning is achieved by a cylindrically shaped, rapidly rotating absorber. The beam pulses are accurately synchronized to the absorber position so as to deliver beam only in the specific range.
Axial scanning is achieved by a sweep magnet located upstream of the final 90° magnet. Thanks to the optical features of the system, the beam reaching the patient is swept parallely with a maximum amplitude of 40 cm. The effective source to isocenter distance is therefore infinite. The axial scanning speed is controlled by a dose integrator allowing an automatic correction of small beam intensity variations.
The transverse beam scanning is obtained by a slow rotation of the final 90° magnet around the axis of the incoming beam. By combining this motion with a slight gantry motion, a parallel beam translation, or even an arc therapy is also possible, reducing the skin radiation dose.
A conceptual mechanical structure has been designed. The reduced diameter of this gantry (5 meter, 16.4 feet) allows its installation in the therapy room of much more reasonable size.
Further calculation and engineering studies will be needed to confirm that the proposed structure is able to meet safely all design specifications.



4. *Acknowledgements*

Proton therapy is today a small and close knit community. Every progress is therefore more or less an achievement of the whole community and not of some groups or individuals. As newcomers in this community we would like to thank specially Michael Goitein, Monroe Rabin, Bernard Gottschalk and Ken Gall at MGH, Anders Brahme at the Karolinska, Eros Petroni and Hans Blattmann at P.S.I.and Pierre Scaillet at Middelheim, Antwerpen. It was through the informations they supplied to us and through the educating and simulating discussions we had together that we could improve and refine the design of this proton therapy facility.

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