

A Hospital-Based Hadrontherapy Complex

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Abstract

The main design issues of the feasibility study for a hospital-based hadrontherapy facility to be built in Italy (Progetto Adroterapia) are described. The facility will aim at the treatment of 1000 patients/year in 5 treatment rooms served by a H⁻/light ion synchrotron. Generation of neutron beams for boron neutron capture therapy and production of positron emitting radionuclides for PET diagnostics is also foreseen. Two of the treatment rooms will be equipped with a rotating isocentric irradiation unit.

1. INTRODUCTION

A feasibility study has just been completed for a hospital-based hadrontherapy facility to be built in Italy [1]. The name *hadrontherapy* has been introduced to indicate *collectively* the use of beams of protons, neutrons and light ions in radiation therapy. This *Hadrontherapy Centre* should be a "centre of excellence" and it is conceived to provide, within a hospital which has already available all other facilities, the techniques and the tools that are related to state-of-the-art radiation therapy. The facility will aim at the treatment of 1000 patients/year and, to maintain at a reasonable level the costs of therapy, it will be provided with several treatment rooms.

Due to their physical properties, hadron beams are characterized, with respect to photon and electron beams, by a more favourable dose distribution in tissue (see, for example, ref. [2]), particularly enhanced for light ions which, in addition, show a higher radiobiological effectiveness. The technological complexity and the costs of hadrontherapy (especially for light ions) call for the definition of a priority scale of the clinical indications. For proton beams four categories of pathologies have been individuated on the basis of the type and of the importance of the expected advantages. An estimate of the patient afflux to the Hadrontherapy Centre, based on the epidemiological data reported in the Tumour Registries of several Italian provinces has yielded an overall figure of about 2500 - 3000 patients expected from the North of Italy.

An accelerator designed for hospital use must meet a number of requirements which are substantially different from

those of a machine to be operated in a research environment. Reliability, flexibility (in order to satisfy the future trends in hadrontherapy), safety and simplicity of operation from the user's point of view represent the basic aspects against which the choice of the accelerator system should be balanced, along with the capital and running costs.

2. FACILITY LAYOUT

The facility will be provided with:

- 1) two treatment rooms equipped with an isocentric gantry (see below) capable of transporting protons up to 250 MeV;
- 2) one treatment room equipped with one horizontal beam and one vertical beam pointing downward, also for 250 MeV proton beams;
- 3) one room equipped with two horizontal beam lines, one for irradiations of eye tumours and one mainly devoted to head and neck treatments;
- 4) one room with one horizontal beam for experimental activities with both protons and light ions (dosimetry, radiobiology, calibrations, etc.);
- 5) one room devoted to future light ion treatments;
- 6) two smaller rooms served by the 11 MeV proton beam from the injector, one for the production of positron emitting radionuclides for PET uses (¹¹C, ¹³N, ¹⁵O and ¹⁸F), the other for thermal neutron production for boron neutron capture therapy (BNCT).

The complex will consist of two buildings: an underground, heavily shielded area (the "bunker") housing the accelerators and the treatment rooms, and a surface building above ground with conventional facilities and office space. The bunker will have a surface area of about 3500 m²; the maximum height will be about 15 m to accommodate the gantry rooms. Besides the basement of the gantry rooms, the lowest floor houses the rooms for BNCT and for radionuclide production.

3. ACCELERATOR DESIGN

The accelerator is a H⁻ synchrotron (which will also accelerate protons) capable of providing 60-250 MeV proton beams with an average intensity of about 10 nA. The injection energy in the synchrotron is 11 MeV. The injector

is an RFQ + DTL structure, delivering average currents of 50-100 μA , sufficient for producing positron emitting radionuclides for PET diagnostics [3] and thermal and epithermal neutrons for BNCT [4].

The design includes the possibility of upgrading the complex to accelerate fully stripped light ions up to ^{16}O to a final energy in the range 120-400 MeV/u with minor interventions on the ring and the addition of a second ion source and injector. A synchrotron has been preferred as it easily provides pulse-to-pulse energy variability over fine steps, as required by the clinicians. In addition, the cyclotron and linac options are ruled out by the request of future upgrade to light ion acceleration.

The main advantages of accelerating H^- are:

- 1) the simplicity of the ejection system based on a thin target (stripping foil);
- 2) the very small transverse emittances (of the order of 0.1 π mm-mrad) of the extracted beams which can be obtained with a small stripping target. This allows the design of lighter and cheaper isocentric gantries as compared to conventional units, due to the smaller required aperture of the magnets;
- 3) the possibility of controlling the extracted beam intensity by a feedback from a monitor in the beam line to the dipoles driving the beam to the stripping foil. The allowed time constant for such direct feedback is fundamentally much shorter than for resonant extraction;
- 4) the fact that, in principle, the extracted beam intensity can be determined as well by measuring the current due to the electrons stripped at the target.

Acceleration of H^- ions instead of protons requires an ultra-high vacuum (of the order of 10^{-10} torr) and a low magnetic field (which translates into a ring of relatively large radius) to prevent beam losses caused by collisional electron detachment and magnetic stripping [5]. These two requirements do not constitute major constraints in the present design, as acceleration of ions needs a residual pressure as low as 10^{-9} torr and a ring of comparable size.

The main parameters of the synchrotron are listed in Table 1. Two dispersion free straight sections are required to reduce the effect of the magnet ripple on the intensity and on the emittance of the beams extracted both by the charge exchange [6] and the resonant method. A short bending magnet has been added in both the straight sections to provide two H^- extraction ports.

Only single-turn injection over half circumference has presently been studied for both H^- and light ions, because it is simpler than multi-turn injection and likely requires a smaller magnet aperture. The same injection elements designed for H^- (septum and fast kicker magnet) will also be adequate for light ions. The space-charge limit, for a Laslett incoherent tune shift $\Delta Q = -0.25$ and an emittance of 20 π mm-mrad at injection, is 4.9×10^{11} particles for 11 MeV H^- and 3.3×10^{10} particles for 3 MeV/u $^{16}\text{O}^{+8}$ ions, higher than the design intensity values required for radiotherapeutical applications.

Injection into a matched bucket and RF capture and acceleration on the fundamental harmonic of the revolution frequency have been considered. The RF accelerating system is

based on a cavity of the type developed at the Laboratoire National Saturne at Saclay (France) [7]; the cavity is loaded with amorphous iron (VITROVAC 6025 F from VACUUMSCHMELZE GmbH, Hanau, Germany) rather than ferrite. The magnetic sensitivity of VITROVAC is 20 to 40 times higher than that of ferrites and therefore the bias current requirements and the number of turns of the bias circuit are 20 to 40 times lower. This cavity can easily achieve a frequency swing of 10 in the range of frequencies of interest and therefore a single unit would be sufficient to accelerate both H^- and light ions.

Table 1. Synchrotron parameters.

	H^- / protons / $^{16}\text{O}^{+8}$
Injection energy [MeV/u]	11 / 11 / 3
Minimum extraction energy [MeV/u]	60 / 60 / 120
Maximum extraction energy [MeV/u]	250 / 300 / 400
Required average current [pA]	10 / 10 / 0.21
Circumference [m]	60.677
Focussing scheme	FODO
Number of FODO cells	10
Number of superperiods	2
Residual pressure [torr]	10^{-10}
Repetition rate [Hz]	2 / 2 / 1
Typical cycle (for maximum energy)	
Acceleration [s]	0.15 / 0.2 / 0.4
Flat-top [s]	0.25 / 0.2 / 0.3
Fall [s]	0.1 / 0.1 / 0.3
Emittance at injection [π mm-mrad]	20
Normalized emittance [π mm-mrad]	3.07 / 3.07 / 1.61
$\Delta p/p$ at injection [%]	± 0.35
Maximum $\Delta p/p$ during acceleration [%]	± 0.6
Frequency at injection [MHz]	0.75 / 0.75 / 0.40
Frequency at extraction [MHz] ^(a)	3.03 / 3.22 / 3.53
	horizontal/vertical
Betatron frequency	2.29 / 2.40
γ at the transition energy	2.085
Maximum values of the β functions [m]	10.05 / 9.50
Maximum value of the dispersion [m]	5.30 / 0
Acceptance with dispersion [π mm-mrad]	79 / 56

^(a) At the maximum energy.

The required residual pressure of 1×10^{-10} torr will be achieved by means of Non Evaporable Getter strips (type St-707 manufactured by SAES Getters Inc., Milano, Italy) installed in the dipole vacuum chambers. Sputter ion pumps will be used to pump noble gases and methane.

The synchrotron is provided with multiple extraction: two charge exchange extraction systems for H^- , one resonant extraction for protons and one fast extraction port to be used as beam abort. The extraction of H^- ions is achieved by smoothly driving the circulating beam against a small

beryllium or carbon foil, by generating a local orbit bump by means of two small dipoles. An equivalent pair of vertical steerers is used to control the vertical position of the beam. The ions traversing the foil lose their two electrons and are converted to protons. A short bending magnet placed just after the stripping foil separates the protons from the circulating H⁻ ions. The two extraction channels can be operated on a pulse-to-pulse basis and independently at different energies.

Each of the two stripping foils is located in correspondence of a vertical focussing quadrupole in a straight section of the synchrotron, where the dispersion and the angular dispersion vanish. It has been calculated that small horizontal and vertical emittances (of the order of 0.1 π mm-mrad) can be obtained with this scheme, by using a target a few μ m thick and a few tens of μ m wide in the vertical direction [6]. The absence of dispersion and angular dispersion at the target position eliminates any correlation between the momentum of a beam particle and its probability of being extracted. Effects such as slewing of the average momentum of the extracted beam during the spill and modulation of the extracted beam current at the frequency of the ripple of the magnetic field in the dipoles are therefore strongly suppressed. Control of the extracted beam intensity is provided by two fast horizontal steering magnets.

The extraction of protons and light ions is achieved by exploiting the third integer resonance line $3Q_x = 7$. The working point is moved to this line by a fast air-core quadrupole and the resonance is excited by means of an adequate set of sextupoles. The resonant extraction system for protons will include an electrostatic septum and a magnetic septum placed about 120° downstream in betatron phase. The upgrading to light ions will likely require the addition of a second septum.

A fast beam abort system can also be implemented by means of a fast extraction system consisting of a fast kicker and a septum magnet located about 120° downstream in betatron phase.

4. GANTRY DESIGN

As stated above, two treatment rooms will be equipped with an isocentric gantry. This unit allows to rotate the terminal tract of the beam line 360° around the patient in order to vary the direction of irradiation as is done in conventional radiotherapy. It has been decided that the gantry should be isocentric rather than eccentric (such as the unit designed at the Paul Scherrer Institut (PSI), Villigen, Switzerland [8]). This decision is mainly dictated by the choice of avoiding a translation of the patient couch which, while permitting a reduction of the gantry radius, has obvious shortcomings from a clinical point of view. An isocentric gantry of the "corkscrew" type, similar to that installed at the Loma Linda University Medical Center (LLUMC) in California [9], has been chosen. With this geometry the longitudinal dimension is considerably reduced as compared to a gantry of conventional design and is only fixed by the space needed by the movements of the patient couch. The radial dimension is determined by the minimum source to isocentre

distance, which should be at least 2 m as imposed by the necessity of minimizing the skin dose. This geometry achieves a minimization of the overall swept volume.

The outer diameter of the proposed gantry is about 11 m with a distance between the exit of the last dipole and the isocentre of 3.4 m. The gantry optics is achromatic and designed to fully exploit the small beam size. The estimated vertical aperture of the dipole magnets is 2 cm. The weight of the magnets is estimated to be about 6 t and the overall weight of the gantry is expected not to exceed 25 t, versus a value of 95 t for the LLUMC design and 120 t for the PSI design. Some steering magnets will be needed for final adjustment of the beam position.

5. CONCLUSIONS

After a sufficient clinical experience has been gained with proton treatments, the accelerator can be upgraded to start treatments with light ions. After initial operation with one horizontal beam line, the building can be expanded with the addition of one or two additional treatment rooms for ions.

6. ACKNOWLEDGMENTS

We are indebted to P. Lefèvre, G. Petrucci, A. Susini and M. Weiss (CERN) for many useful discussions and advices. The building and plants have been designed with the support of CERN ST division, for which we wish to thank F. Ferger and A. Scaramelli. Part of the design has been carried out by AccTek Associates (La Grange, Illinois) on a consultancy basis.

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