HADRON ACCELERATORS IN MEDICINE

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ABSTRACT

The application of hadron accelerators (protons and light ions) in cancer therapy is discussed. After a brief introduction on the rationale for the use of heavy charged particles in radiation therapy, a discussion is given on accelerator technology and beam delivery systems. Next, existing and planned facilities are briefly reviewed. The Italian Hadrontherapy Project is then described in some detail, with reference to both the National Centre for Oncological Hadrontherapy and the design of different types of compact proton accelerators aimed at introducing proton therapy in a large number of hospitals.

1. INTRODUCTION

The applications of particle accelerators in biomedicine are many and varied. They range from the widespread use of small electron linacs for radiation therapy, to the application of 10-20 MeV proton cyclotrons for radionuclide production or elemental analyses, to the use of 50-60 MeV proton accelerators for neutron therapy or proton treatment of eye melanomas to the recent larger hadron accelerators installed in hospitals for cancer therapy with proton or ion beams. Here we will only focus our attention on the use of hadron accelerators in cancer radiation therapy.

The use of hadron beams (protons, neutrons and light ions such as carbon, oxygen and neon) in cancer radiation therapy has grown considerably since the first trials in the mid-fifties and now a few hospital-based facilites are in operation, with a few others being built or designed. A very comprehensive and up-to-date review of the field can be found in refs. [1, 2]. About 17,000 patients have been treated worldwide. Pathologies for which protons have a clear clinical indications are: uveal melanomas, chordomas and chondrosarcomas of the skull base, spinal and paraspinal tumours, parasellar meningiomas, optical nerve gliomas, craniopharyngiomas, acustic nerve schwannomas, artero-venous malformations and hypophysis adenomas. There are other classes of tumours for which protons should prove superior to conventional radiotherapy but still need clinical investigation.

The rationale for the use of protons in radiation therapy relies on the superior dose distribution which can be achieved with respect to photons and electrons: this is due to the low lateral scattering undergone by protons, their well-defined range and the increasing ionization (i.e., dose deposition) with increasing penetration in tissue, which produces the well known Bragg peak (Fig. 1). Ions heavier than protons show an even improved dose distribution and the additional advantage of an increasing biological effect (RBE) at the end of range [3]. The depth of the Bragg peak depends on the initial energy of the protons and its width on the energy spread of the beam. By varying the energy during the irradiation in a well controlled manner, one can superimpose many narrow Bragg peaks and obtain a Spread Out Bragg Peak (SOBP) to treat an extended region in depth. Although this implies an increase of the skin dose, the overall dose distribution is still far better than that achieved in conventional radiation therapy. Fig. 2 shows the range-energy curves in tissue for protons and various ions. From these curves it is easily seen that, in order to irradiate deep seated tumours, the minimum required energy for protons and carbon ions is 200 MeV and 400 MeV/u, respectively. All the clinical needs can be satisfied with proton energies in the range 60 - 250 MeV and beam intensities of the order of 5×10^{10} - 10^{11} protons/s. For light ions the corresponding figures are 120 - 400 MeV/u and 10^{9} - 10^{10} ions/s, according to the type of ion (carbon or oxygen). These requirements represent the basic specifications which must be met by a hadron accelerator designed for cancer therapy.

The actual specifications which must be met by a proton or ion radiation therapy facility to provide a reliable and effective tool against cancer are much stricter. The performance specifications of the accelerator and of the beam transport and delivery systems must satisfy a number of clinical requirements. These performance specifications are expressed in forms of energy range, energy variability, beam intensity, lateral penumbra, distal dose fall-off, source-to-surface distance, time structure of the extracted beam, raster scanning system specifications and beam abort time. A discussion on these specifications may be found in ref. [4].

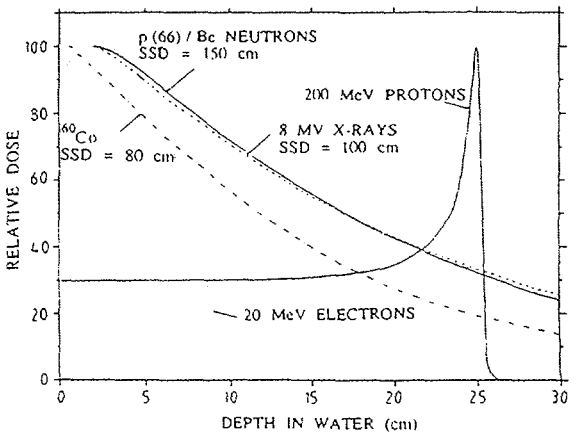

Fig. 1. Depth-dose curves for photons (from a Cobalt source and a 8 MV linear accelerator), neutrons, 20 MeV electrons and 200 MeV protons. With protons the highest dose is released near the end of their range in tissue giving raise to the Bragg peak. SSD is the source-to-skin distance.

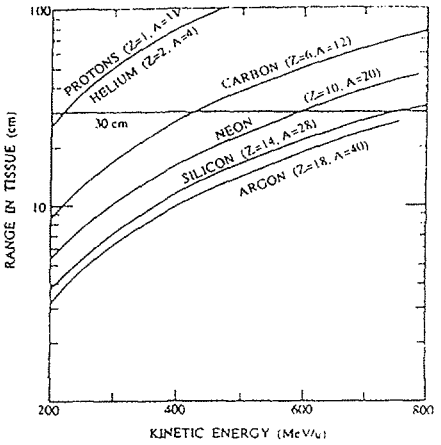

Fig. 2. Range-energy curves in tissue for protons and light ions.

2. ACCELERATOR TECHNOLOGY AND INSTRUMENTATION

The two types of hadron accelerator which are the best candidates for medical use are the cyclotron and the synchrotron, but the linac is also being studied, as discussed in Section 6. The obvious advantages offered by a conventional cyclotron working at fixed energy (i.e., fixed frequency and fixed field) are simplicity and reliability of operation and continuous beam. The dc beam yields beam intensities well exceeding the requirement for therapy and sufficient for other potential applications such as radionuclide production and neutron therapy. The disadvantage with a fixed energy accelerator is that the beam has to be degraded in energy down to the value corresponding to the maximum depth of penetration for the given treatment. This is done by means of an absorber placed just outside the extraction port of the accelerator, before the beam is transported to the treatment room. Degradation of beam energy implies degradation of beam quality; this has to be compensated by analyzing the beam in momentum, with loss of beam intensity and activation of components in the beam transfer line. Therefore, according to the amount of energy degradation required, for a given beam intensity to be delivered at the isocentre, a higher intensity must be accelerated in the machine. The proton intensity required for therapy is of the order of a few nA and cyclotrons can deliver extracted beams of several tens of μA, so that in this respect the above requirement does not constitute a restriction. Nevertheless since the extraction efficiency in a proton cyclotron is not particularly good (60-70%), the higher is the beam current accelerated in the machine, the higher are the prompt radiation generated and the induced activity in the vacuum chamber. In principle, the absorber can be placed in the treatment room at the end of the transfer line (which presents the advantage that the magnets in the beam line operate at fixed field), but in this case no clean-up in momentum can be made and reaction products from the degrader can reach the patient. This is particularly critical in the case of treatment of ocular melanoma, where only 70 MeV are required and degradation of the beam quality subsequent to the energy reduction would therefore be much pronounced.

Whereas a cyclotron is made up of a large magnet and the particle beam, in the course of the acceleration process, moves along a spiral trajectory starting from the machine centre, in a synchrotron the beam circulates along a fixed orbit and the magnetic and radiofrequency fields vary according to the acceleration process. The obvious example for the synchrotron solution is the facility at the Loma Linda University Medical Center (LLUMC) in California [5] described in the next Section, where a Radio Frequency Quadrupole (RFQ) injects protons at 2 MeV into a compact synchrotron of 20 m circumference which then accelerates the beam to a final energy in the range 70 - 250 MeV. The main advantage offered by a synchrotron is its energy variability. An interesting possibility is that the energy of the extracted beam can be varied from one cycle to the next in steps as small as 1 MeV, so that the modulation of the Bragg peak (see below) can be achieved without the use of absorbers, thus preserving beam

quality and reducing the amount of shielding required. In this case it is obvious that the same rapid adjustment of the magnetic field is necessary for the magnets in the beam line and gantry (see below). This possibility should not be underestimated since the use of absorbers causes scattering and straggling that increase the transverse and longitudinal size of the beam and blur its edges. On the other hand the accelerator is more complex than a cyclotron working at fixed frequency and fixed field, because of its peculiar operating principle relying on rapid field variation. However many synchrotrons far more complex that the ones required for therapy are in operation worldwide.

To allow full flexibility in patient treatment, the accelerator should be coupled, via a beam transfer line, to an isocentric beam delivery system called "gantry". A gantry is a rotating mechanical structure which supports and positions the terminal tract of the beam transfer line (the bending, focusing and steering magnets, as well as the "nozzle") in order to focus the proton beam on the patient's tumour from any desired angle in the plane of rotation, as is routinely done in conventional radiotherapy. In this way the conformation of the dose distribution (especially for tumours having a complex shape) can be obtained and the dose delivered to the surrounding healthy tissue is reduced. The gantry can be isocentric or eccentric (in which case the gantry and the couch rotate around a common point, the isocentre), as the unit designed at the Paul Scherrer Institut (PSI), Villigen, Switzerland [6]. An isocentric gantry of the "corkscrew" type, similar to that installed at LLUMC, is shown in Fig. 3. With this geometry the longitudinal dimension is reduced as compared to a gantry with an in-plane configuration and is only fixed by the space needed by the movements of the patient couch in the horizontal plane for patient positioning. The radial dimension is determined by the source to isocentre distance, which should be large enough in order to minimize the skin dose. This geometry achieves a minimization of the overall swept volume. The outer diameter of the gantry shown in the figure is about 11 m with a distance between the exit of the last dipole and the isocentre of 3.4 m.

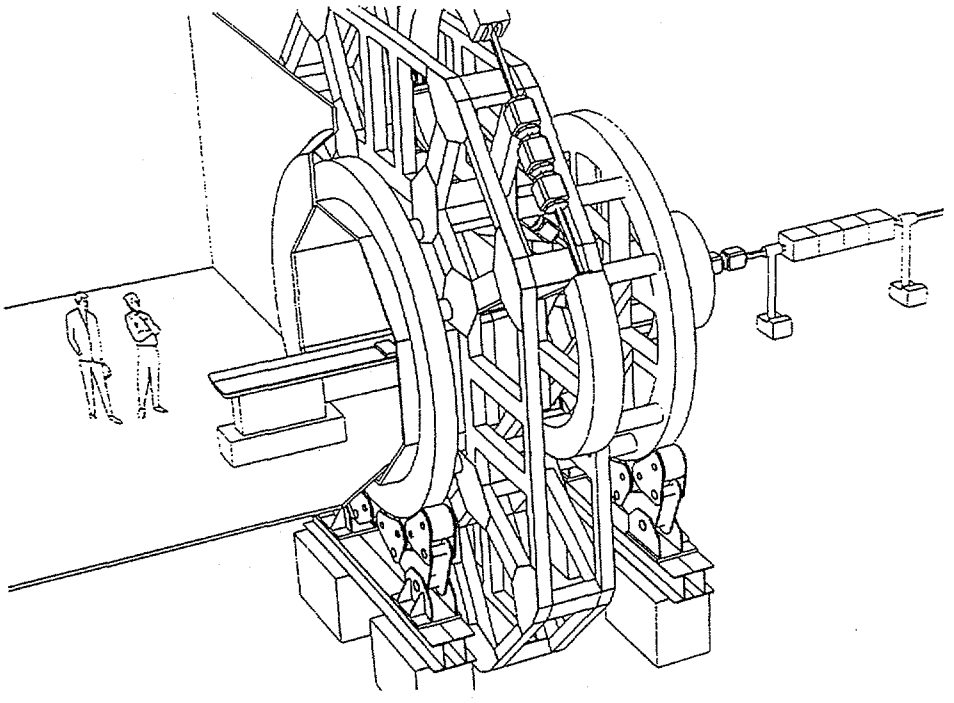

Fig. 3. "Corkscrew" isocentric gantry.

To achieve a conformal irradiation of the target volume, the beam has to be spread both longitudinally (i.e. in depth) and transversally. This can be achieved either with "passive" or "active" methods. Modulation of the beam energy is necessary to obtain a SOBP matching the entire tumour extension in depth. With a fixed energy accelerator this is obtained with a rotating absorber of variable thickness placed in the beam path, or via a so called range shifter. With a synchrotron energy variation can be achieved on a pulse to pulse basis, as discussed above. Lateral beam spreading can be achieved with a passive device. This system makes use of a double scatterer: the first spreads the incoming pencil beam, the second flattens the dose distribution in order to obtain a large and uniform irradiation field. Lateral and distal shaping are obtained by means of a collimator and a compensating bolus. Better results in conformal therapy can be achieved by an active scanning method, either voxel or raster. In this case an optimized dose delivery without the use of any patient specific hardware is obtained. Magnetic beam scanning fully

displays its advantages when, instead of irradiating regularly shaped areas, scanning is limited to the cross section of the target volume including only a small safety margin around it. With raster scanning the beam is scanned continously across a slice at a given depth and the dose delivery is controlled by the scanning speed and/or by the beam intensity.

Present facilities still make use of passive systems for conforming the dose to the tumour. We shall briefly discuss the dosimetry of therapeutical proton beams with reference to such systems. The dosimetry instrumentation depends on the time structure of the proton beam, which is a function of the type of accelerator employed (i.e., a cyclotron, a synchrotron or a linac). Dosimetry of the therapeutical beam is needed for:

1) control of position and intensity of the proton beam emerging from the modulation and spreading devices;
2) control of the uniformity of the proton fluence in the cross-section of the therapeutic beams;
3) absolute beam calibration, i.e. knowledge of the absorbed dose in a tissue element exposed to the beam at the isocentre under standard physical conditions;
4) knowledge of the relative tri-dimensional distribution of the absorbed dose in the irradiated region;
5) knowledge of the values of microdosimetric quantities in the irradiated region.

The primary purpose of a monitoring system is to ensure the correspondence between prescribed and delivered dose at a reference point, within a given overall uncertainty which needs to be +/- 2.5%, according to clinical requirements. Dose delivery must be promptly stopped as soon as the prescribed dose is reached. The agreement between nominal and actual proton range in water has to be periodically tested with a relative system. Instrumentation for on-line monitoring of the beam includes ionisation chambers, multiwire ion chambers, secondary emission monitors (SEM) and silicon diodes. Instrumentation for reference dosimetry includes a water calorimeter, ionisation chamber and Faraday cup.

Medium energy proton medical accelerators produce secondary radiation, mainly neutrons, from the beam interaction with accelerator components (the magnets in particular), beam delivery devices (such as collimators) and the patient. Although the beam intensity is much lower than that typical of research accelerators, the fact that such accelerators may be installed inside a hospital or in highly populated areas calls for a shielding design which should virtually reduce to zero the radiological impact on the external environment. Typical shielding thickness ranges from about 1.5 m of ordinary concrete (for shielding of the treatment room from the beam transfer line, where beam losses are limited) up to 4 m (for shielding the forward secondary emission in the case of a treatment room with a fixed horizontal beam). Shielding thickness obviously also depends on whether one should ensure dose equivalent limits compatible with occupationally exposed workers or with the general public.

As stated above, a particular situation with a medical proton facility is the use of an isocentric gantry in one or more treatment rooms. The fact that the beam can be rotated 360 degrees in one plane imposes specific requirements on the shielding walls of the room, as well as accurate assumptions on the use factor of the shields. Access to the treatment rooms needs to be a two or three leg maze, several metres long, to avoid the use of a massive shielding door which would not be compatible with the clinical use of the facility. At most, a polyethylene door like those used in conventional radiation therapy units provided with electron linacs may be used. Interlocks must obviously prevent access to the accelerator hall and to any room when the beam is present, but the possibility of fast beam abort and immediate entrance into the treatment room must be ensured in case the patient needs assistance.

3. FACILITIES

At present, protontherapy centres are located in the United States, in Russia, in Japan, in South Africa and in Europe. Most of the clinical experience has so far been obtained at research institutions which have devoted, at some stage, part of the accelerator time to medical uses. Loma Linda is the first hospital-based proton radiotherapy facility for treatment of deep-seated tumours (Fig. 4). The Northeast Proton Therapy Center (NPTC) to be built at MGH (Massachusetts General Hospital, Boston, USA) will be the next, where patient treatments should start in 1998 [7]; it will have 2 or 3 isocentric gantries with an in-plane configuration but with equivalent performances of the "corkscrew".

The only centre with experience in light ion therapy is the Lawrence Berkeley Laboratory (LBL) in the USA, which terminated its clinical activity in February 1993 [8]. LBL was equipped with a fixed horizontal beam and with a "wobbler" beam spreading system. A hospital-based centre for light ion radiotherapy (HIMAC) has recently started operation in Chiba (Japan) [9]; it is equipped with a combination of fixed horizontal and vertical beams. Together with GSI at Darmstadt (see below), this centre will provide during the next years crucial information on the effectiveness of ion therapy and will define the needed protocols.

Let us now briefly review the situation in Europe. Hospital-based centres for treatment of ocular melanomas are in operation at the Centre Antoine-Lacassagne (Nice, France) and at the Douglas Cyclotron Unit (Clatterbridge, UK). Eye treatments are also given at non-hospital facilites: PSI Villigen (Switzerland), CPO in Orsay (France), UCL in Louvain-la-Neuve (Belgium) and GWI in Uppsala (Sweden). At PSI a medical beam line provided with an isocentric gantry, a new room and a medical annex building have been constructed; the beam tests are completed

and patient treatment will begin in Spring 1996 [6]. The facility is intended for experimental cancer therapy and should treat about ~100 patients per year. A similar project is under way at the Gesellschaft für Schwerionenforschung (GSI) in Darmstadt (Germany), but treatments will be performed with light ion beams, probably carbon [10]. Radiobiological experiments and tests of instrumentation are under way. The project has been funded by the German government; the medical beam line, the instrumentation and the medical building are under completion and patient treatments should start in 1996. The facility will treat approximately 70 patients per year. A proposal also exists for setting up a beam line dedicated to proton therapy at the cooler synchrotron COSY in Jülich (Germany) [11]. The facility will be equipped with an isocentric gantry. At the Hahn-Meitner-Institute in Berlin (Germany) a project was started in 1992 to build a proton therapy facility for the treatment of ocular melanomas using the 72 MeV beam of the VICKSI cyclotron [12]. The cyclotron building is being modified and the treatment of the first patients is scheduled for October 1996. Another project is to upgrade the existing proton therapy facility at the Douglas Cyclotron Unit (Clatterbridge, UK). A design has been completed for a booster linac to raise the proton energy from 62 MeV to 200 MeV [13]. Plans also exist for setting up proton therapy at KVI in Groningen (The Netherlands) using the AGOR cyclotron and at Munich University (Germany) using the Tritron cyclotron [14]. Research on ion cancer therapy is also part of the AUSTRON project [15]. The only plan for a large hospital-based facility is the project for the Hadrontherapy Centre to be built in Italy and discussed in the next Section.

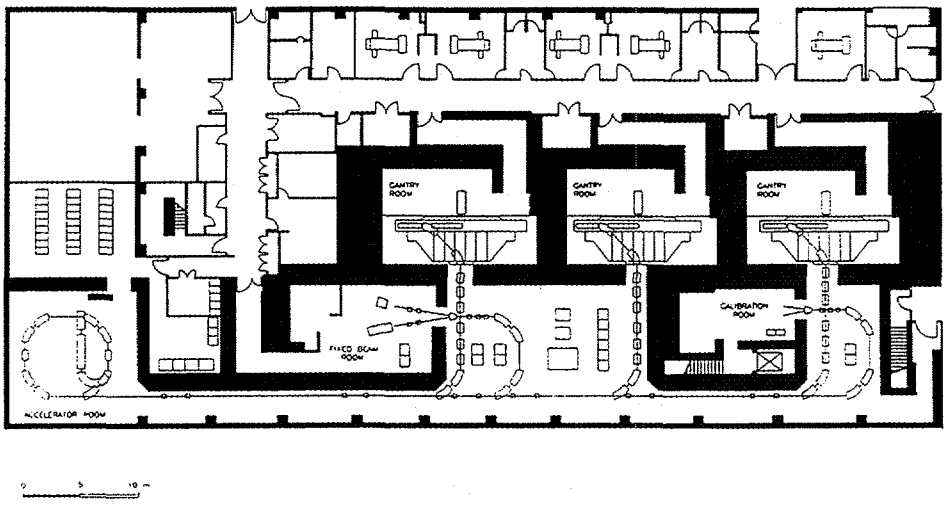

Fig. 4. Loma Linda University Medical Center, California, USA (adapted from ref. [5]).

4. THE TERA PROJECT

The aim of the TERA Project (Hadrontherapy Project) [16] is to set up a nationwide network of protontherapy centres which has been named RITA (Rete Italiana Trattamenti Adroterapici, Italian Network for Hadrontherapeutical Treatments). The centre of the network is occupied by the National Centre for Oncological Hadrontherapy (Centro Nazionale di Adroterapia Oncologica, CNAO) which will have four rooms for proton treatments. This hospital-based *Hadrontherapy Centre* should be a "centre of excellence" and it is conceived to provide 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 is designed with a relatively easy upgrading path to ion treatments. The other nodes of the RITA network are various *Protontherapy Centres*, which should make use of relatively "cheap" and "compact" proton accelerators to be installed, due to their reduced space requirement, in a number of hospitals distributed over the entire nation. Multimedia connections established among the centres will allow physicians and medical physicists to discuss the cases by exchanging CT, MRI and other diagnostic images and possibly plan the best treatment at a distance. Following such preparatory work, the patients will be referred to the closest or more convenient centre for hadron treatment. Some patients may be treated locally with conventional radiation and receive elsewhere only a proton (or ion) boost.

5. THE HADRONTHERAPY CENTRE

The CNAO (Fig. 5) [16, 17] will be provided with: 1) three treatment rooms equipped with an isocentric gantry capable of transporting protons up to 250 MeV; 2) one room equipped with two horizontal beam lines, one for irradiation of eye tumours and one mainly devoted to head and neck treatments; 3) one room with one horizontal

beam for experimental activities with both protons and light ions (dosimetry, radiobiology, calibrations, etc.); 4) one smaller room served by the 11 MeV proton beam from the injector, for thermal neutron production for boron neutron capture therapy (BNCT). The possibility also exists for the production of positron emitting radionuclides for PET diagnostics (^{11}C, ^{13}N, ^{15}O and ^{18}F).

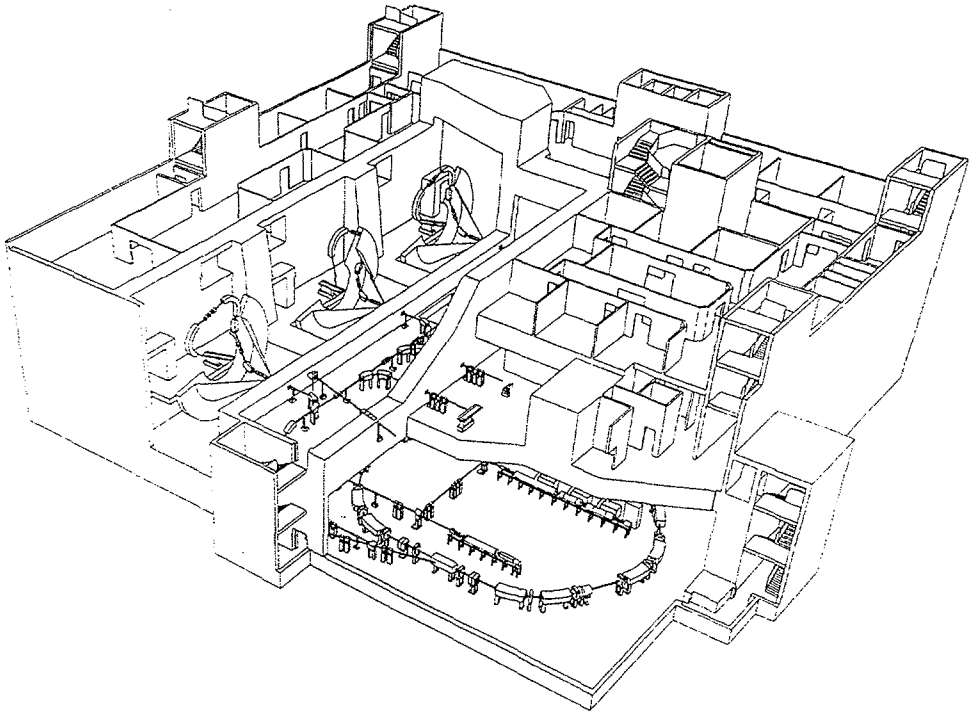

Fig. 5. 3-D view of the National Centre for Oncological Hadrontherapy (CNAO).

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 has a surface area of about 3000 m^2. The clinical area is located at a different level than the accelerator and services. The advantages of this solution are in terms of optimized patient flow and decoupled clinical and research/service activities. The building is made up of 4 floors or levels (numbered -1 to -4 top to bottom), with the two lower levels occupied by the accelerators (synchrotron and injectors), an experimental area and services (power supplies, ventilation, cooling, workshops, etc), and the two upper levels by the clinical area. The high energy beam is brought from level -4 to the main treatment area on level -2 by a single beam line. 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.

The main accelerator is a synchrotron capable of providing 60-250 MeV proton beams with an average intensity of about 10 nA [16-19]. 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 [20] and thermal and epithermal neutrons for BNCT [21, 22]. The synchrotron can also accelerate H^- ions and extract them by charge exchange in a foil [23]. The possibility exists 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 synchrotron design has been modified with respect to the description given in refs. [16, 18, 19]. A multi-turn injection scheme has been adopted [17], instead of single-turn. The present version of the lattice presents the following features: 1) implementation in one long straight section of multi-turn injection of protons, light ions and H^- ions. This requires the installation of two electrostatic septa and four fast magnets to provide a bump of variable amplitude corresponding to the exit of one of the two septa; 2) implementation in one long straight section of: a) resonant extraction of protons and fully stripped light ions by means of an electrostatic septum and two DC magnetic septa of increasing thickness in cascade; b) stripping extraction of H^- ions by means of a beryllium foil

placed inside a short dipole magnet; the latter provides a kick to the orbit of the stripped ions sufficient to channel them into the magnetic septa used for resonant extraction.

Proton treatments will most likely start by using passive beam delivery systems, i.e. the scattering foil technique. This method is basically not sensitive to the time structure of the beam, so that resonant extraction will be adequate. When raster scanning is implemented, an effort will be made to ensure beam stability over a time scale of μs or tens of μs. If resonant extraction does not provide the required beam stability, the linac and synchrotron operation can be converted to accelerate H^- ions and extract them by the charge-exchange technique. The magnets to produce the orbit bump and to separate the stripped protons from the circulating H^- beam are included in the lattice.

The advantage of accelerating H^- ions lies in the very simple extraction scheme, in which a slow orbit bump, produced by two small dipoles, drives the beam against a thin stripping foil (e.g., beryllium or carbon); the H^- ions undergo charge exchange in the foil, are turned into protons, separated from the circulating beam by a short dipole placed downstream the foil and extracted. This extraction mechanism is simpler than resonant extraction and should allow a better control of the therapeutic beam, which is, as mentioned above, particularly important when a scanning technique is adopted.

Acceleration of H^- ions 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 [24]. 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.

6. THE COMPACT ACCELERATOR PROJECT

Another main goal of the TERA project is the development of a "compact" proton accelerator for hospital installation. Such an accelerator should satisfy the following requirements (or at least most of them): 1) it should accelerate a minimum of 2×10^{10} protons/s to at least 190 MeV; 2) it should be built (including ancillary systems) in less than 300 m^2 (shielded area and service space); 3) it should consume less than 250 kW and 4) it should cost, with one external beam (but without civil engineering) less than 10 M$; this figure should include the cost of controls and beam delivery, but the cost of the injector can be excluded if it is also used to produce PET radionuclides for the same hospital. Four options have been considered: 1) a synchrotron using pulsed magnets with a peak field of 4 T; 2) a linear accelerator; 3) a superconducting cyclotron and 4) a weak focussing synchrotron of the LLUMC type but of reduced circumference. The demonstration of the feasibility of one (or more) of these designs would represent a significant technology transfer from the research field to industry and the medical field. The four designs are thoroughly discussed and compared in ref. [25] and a brief overview is given in ref. [26]. Here we will only discuss the two designs which are most advanced, i.e. the linac and the superconducting cyclotron.

Linear accelerator. Medical electron linear accelerators (6-25 MeV) are commonly used for radiation therapy, where they have largely replaced cobalt-60 units. The typical RF frequency of these machines is 3 GHz, for which many RF power supplies are commercially available. To develop a compact proton linac one is therefore tempted to use this frequency, although it is unusual for accelerators, as such a high frequency is not compatible with a large bore aperture (i.e., high proton intensity). However, the proton current required for therapy is so low (a few nA) that the use of a 3 GHz structure becomes feasible.

The linac (Fig. 6) is designed for a maximum energy of 200 MeV. The accelerator structure is made of: 1) an ion source and low energy beam transport; 2) an RFQ operating at 750 MHz, accelerating protons to 5 MeV, with total length of 2.6 m; 3) a 3 GHz Side-Coupled-Drift-Tube-Linac divided in 6 sections, to accelerate protons to 70 MeV; each section, less than 2 m long, consists of a number of tanks and requires less than 1.5 MW peak power. Focussing is achieved by permanent magnet quadrupoles; 4) an achromatic magnetic system to bend the beam 180° (either in the horizontal or in the vertical plane - the latter solution is preferred to minimize the surface area required), made up of 2 dipoles, 2 quadrupoles and 2 re-bunching cavities; 5) a Side-Coupled-Linac accelerating protons from 70 to 200 MeV, with a peak power requirement of 30 MW. The SCL is made up of 30 tanks, each consisting of 14 cells. The total length of the SCL structure is 12 m. The facility should be equipped with one beam at a fixed energy of 70 MeV (from the exit of the SCDTL) for eye treatments and one gantry transporting the 200 MeV beam from the exit of the SCL. The energy should be variable in steps by switching off some of the SCL tanks.

Superconducting cyclotron. A superconducting cyclotron is an interesting solution for a hospital-based accelerator, since size and cost are inversely proportional to the magnetic field, operation costs are much reduced with respect to a room-temperature cyclotron and superconductivity is a well-established technique, already used in hospitals (NMR imaging). A superconducting cyclotron accelerating deuterons to 50 MeV has been designed and built at the Michigan State University and is in use at the Harper Grace Hospital in Detroit for neutron therapy [27]. A small superconducting cyclotron (12 MeV) for the production of PET radionuclides is marketed by Oxford Instruments [28].

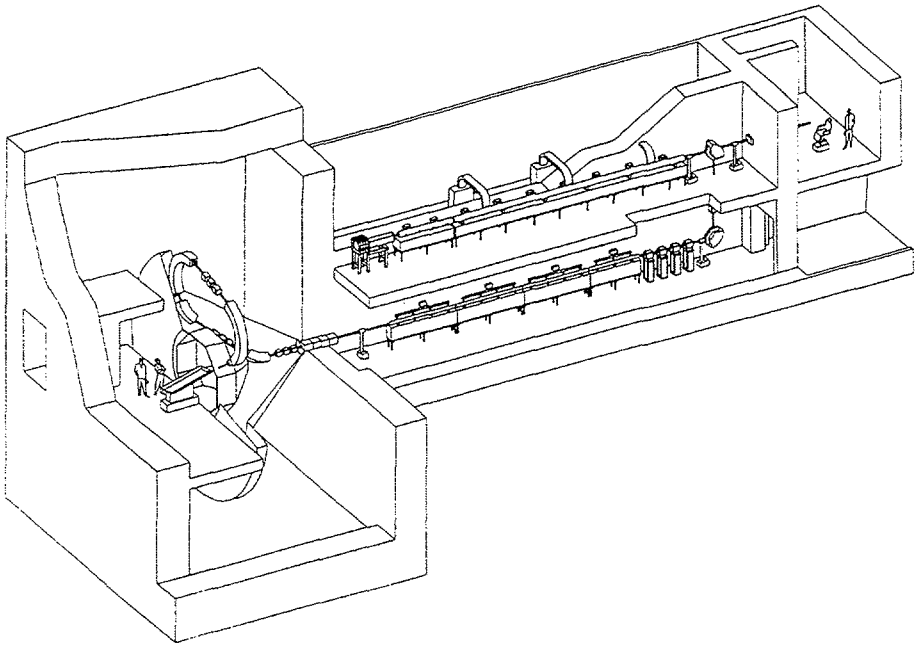

Fig. 6. The 200 MeV linac, folded vertically, supplying two treatment rooms.

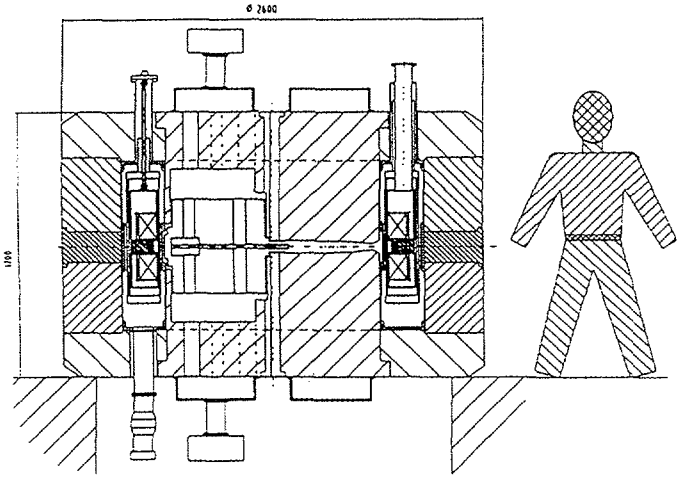

Fig. 7. Schematic view of the superconducting cyclotron.

The cyclotron presently under study is a joint project between the University of Milan and the Centre Antoine Lacassagne in Nice. The cyclotron (Fig. 7), designed for a fixed energy of 200 MeV, has a three sector configuration and uses three RF cavities located in the valleys. The extraction radius is 0.7 m. To simplify the construction and the operation of the cyclotron and to reduce cost, the following solutions have been adopted: 1) the isochronous field is obtained by shimming the poles, thus avoiding the necessity of trim coils; 2) the superconducting coils are operated in the persistent mode by shortening them with a superconducting cable; 3) there is no LHe liquifier, but the cryostat only requires a periodic refilling (about once a fortnight); 4) acceleration is achieved with 3rd RF harmonic; 5) use is made of an internal proton source. The estimated power consumption of the cyclotron is 120 kW. The magnetic field at the centre is 2.53 T. The extracted proton current for therapy application is limited to 500 nA.

7. CONCLUSIONS

Advantages and disadvantages of different options for a medical proton accelerator, whether it should be in particular a cyclotron or a synchrotron have been largely addressed, but no definitive answer has been given, and is likely to not exist. There are several factors which have to be balanced: fixed versus variable energy extraction, time structure of the extracted beam (continuous versus pulsed), flexibility versus simplicity of operation, reliability, space requirements, capital and running costs, etc. As a point of fact, a synchrotron was chosen at LLUMC, and a cyclotron for the NPTC to be built at the Massachussetts General Hospital in Boston. For ion acceleration, the synchrotron seems to be the only choice, as has been demonstrated by the European study EULIMA.

The linac may prove to be a successful candidate, while the use of a superconducting magnet can drastically reduce size, weight and the possible cost of a cyclotron. The next few years will hopefully bring interesting technological developments to this field, not only on the accelerator side, but also in the beam delivery system. The development of more compact gantry systems is as important as the construction of compact proton accelerators. If these advances do come about, it is possible that proton (and possibly ion) therapy will become a widespread clinical tool.

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