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A PROTON ACCELERATOR FOR MEDICAL APPLICATIONS

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A PROTON ACCELERATOR FOR MEDICAL APPLICATIONS*

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The conceptual design of a proton synchrotron for radiation treatment of cancer in a hospital is described. The proton energy would be variable on a pulse-pulse basis up to 250 MeV with a maximum accelerated current of 60 nA average at a 5 Hz repetition rate. The ring would consist of a number of small straight magnets for ease of construction and flexibility of arrangement of dipoles, quadrupoles, and drift spaces. The ring would be 23' in diameter. The same magnets (2 times the number) could also be arranged in a ring of twice the diameter. The synchrotron would then accelerate helium ions to 250 MeV/ μ , suitable for the full range of radiation treatment. A major feature of this larger synchrotron, which is the one described below, is the ability to accelerate negative hydrogen ions to 250 MeV (with lower peak magnetic fields to avoid magnetic stripping of the ions) and utilize charge exchange extraction. The latter technique appears to be most promising for satisfying the requirements for dynamic raster scanning of narrow proton beams.

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1. Introduction

The use of medium energy protons for radiation treatment of some cancer sites has been pioneered by a number of physicians from Massachusetts General Hospital utilizing the 160 MeV cyclotron at Harvard University.^[1] Their success over the past 10 years has focused attention on the advantages of protons for cancer treatment, first pointed out by Wilson.^[2] Other facilities carrying on therapy programs with light ion beams include Lawrence Berkeley Laboratory (He), and proton facilities in the Soviet Union, Sweden, and Japan. This growing interest has stimulated the formation of the Proton Therapy Co-Operative Group (PTCOG)^[3] composed of interested physicians, medical physicists, and accelerator physicists. The Group has met several times to discuss the issues and requirements of proton (and light ion) radiation treatments, and, in particular, accelerators suitable for use in hospitals.

To be suitable for hospital use, the accelerator must be simple, very reliable, safe, and relatively inexpensive, compared to machines designed for physics applications. These characteristics appear achievable because of the much lower intensity requirements for medical applications. For instance, simple single-turn injection at relatively low injection energy can be utilized, as well as small aperture magnets. Instabilities and resonances, common to high current beams with strong space charge forces, can be avoided. Finally, conservative engineering design of the individual components can significantly enhance the overall reliability.

Accelerator concepts along these lines were originated by the author in 1974.^[4] These initial concepts were aimed at a proton diagnostic instrument capable of computerized tomography. Acceleration of negative hydrogen ions was envisioned, with slow extraction accomplished by passing a controlled portion of the beam through a very thin foil. The two electrons are removed from the proton in the process so that the proton, being of opposite charge, then exits the accelerator in passing through the next magnet. These concepts were not carried further because the medical interest and institutional support were not adequate at the time. Many of the same concepts, however, are applicable to a synchrotron for radiation treatment of cancer and are incorporated in the present design.

The practical application of H^- charge exchange as applied to synchrotrons was developed as an injection technique to avoid the limitations of Liouville's theorem, and fill a small storage ring to its space charge limit.^[5] It was subsequently utilized routinely by the ZGS and RCS (Rapid Cycling Synchrotron) at Argonne, by the boosters at Fermilab and KEK (Japan), by the AGS at Brookhaven, and more recently by the SNS at Rutherford (England), and the PSR at Los Alamos.

As mentioned, however, injection into a medical synchrotron can be a simple process. On the other hand, slow extraction with control of the beam current to $\pm 2\%$ might be required to achieve the necessary dose homogeneity for 3D dynamic raster scanning of the tumor site. The success of charge exchange injection has demonstrated the simplicity of the technique, and as much success can be anticipated by using this process for slow extraction. The alternative, resonant extraction, is a much more sophisticated process and could

prove to be more difficult to control with the precision and reliability required for this method of dose delivery to the patient.

Other methods of dose delivery, such as the double scattering method used at Harvard,^[6] or the wobbling beam method developed at Berkeley,^[7] do not require as much precision in beam current control. Each method has its advantages and disadvantages.

The choice of providing the option of accelerating H^- ions has two important consequences. First, the vacuum must be significantly better than if protons were accelerated in order to avoid the loss of the weakly bound electron by collisions with the residual gas atoms. Second, the magnetic field, which appears like an electric field in the rest frame of the ion, must be relatively weak in order not to remove the same electron. Both of these consequences will be discussed in the next section.

2. Design Issues

There is not a general consensus within the PTCOG on the feasibility or advantages of dynamic raster scanning. Three dimensional contouring can be accomplished in other ways, such as with multileaf collimators that can be varied during the treatment. Goitein and Chen^[8] have pointed out that the reduction in dose to healthy tissue in average cases might be 10%, not an overwhelming factor. On the more positive side, however, are a) the potential for achieving 90-95% efficiency of beam utilization (compared to 20-30% for the double scattering method presently in use at Harvard, and 50-70% for the wobbling beam method at Berkeley), and b) the very real possibility of reducing the lever arm between the last magnet and the patient, making the isocentric gantry much more practical (provided one can overcome the increased entrance dose resulting from a conical distribution emanating from a point source, such as with faster treatment planning, patient set-up, and alignment so that more fields can be utilized). The latter, in particular, could lead to a significant reduction in dose to normal tissue, reinforcing the choice to preserve the option of accelerating H^- ions.

The question of the required beam intensity has been the subject of continuing discussion in PTCOG meetings. Accelerator intensity requirements vary with the treatment site and volume, the mode and efficiency of beam delivery to the tumor, the desired treatment time, and the desired fractional dose. The discussion has been led by Goitein^[9] in order to guide the accelerator designers. A general conclusion has been that significantly higher intensity ($\times 10$) than presently available at Harvard would be desirable in order to treat larger volumes in shorter times. With a beam delivery efficiency of 20-30%, characteristic of the double scattering method, this results in an accelerated beam requirement of 10-20 nA average, or 2-6 nA effective into the tumor. The lower current would be applicable with beam delivery efficiencies approaching 100%, which might be possible with dynamic raster scanning beams. The latter, however, has not been demonstrated so that it might be prudent for a designer to accept the higher current requirement initially. A reasonable design might incorporate the ability to operate at 5-10 Hz and an average current of 10-20 nA, with the option of demonstrating slow extraction with raster scanning. The present design accomplishes these goals.

The H^- ion lifetime due to stripping in a magnetic field is a very steep function (exponential) of energy and magnetic field.^[10] At 250 MeV in a field of 0.56 T, the magnetic lifetime would be 20 sec (the actual lifetime in the medical synchrotron would be about 30 sec because the dipoles only occupy about 2/3 of the ring). The lifetime would be reduced an order of magnitude with only a 10% increase in magnetic field, so that the upper limit of the magnetic field for accelerating H^- ions to this energy was chosen as 5.6 kG. With peak energy and a constant field for 0.4 sec, during which time the circulating current will vary linearly from its full value to 0 as beam is extracted, the loss will be about 2/3 of one percent of the accelerated beam. There would be negligible loss due to magnetic stripping during the accelerating cycle, and the loss during extraction at an energy less than 250 MeV would be substantially lower than that calculated above.

With a maximum field of 5.6 kG at an energy of 250 MeV, the diameter of the ring would be about 45'. This relatively large size for a ring of this energy is considered by some as a serious drawback. However, the magnet cross-section is quite small and the potential advantages of H^- ion acceleration with charge exchange extraction seem sufficient to retain this option. When it is realized that the synchrotron can be easily and inexpensively installed in a tunnel under the hospital parking lot, with beam simply transported into the hospital, the relatively large size does not seem like a significant drawback.

The modular approach adopted by ACCTEK Associates, the design of small, straight dipole magnets for ease of construction and flexibility of layout, allows a simple resolution of these issues of desired beam intensity and synchrotron size. The same magnets operated at 11 kG, arranged in a ring of 23' diameter, could accelerate protons to 250 MeV at high repetition rates with a maximum current of 60 nA. The latter represents the space charge limit of such a synchrotron with 3 MeV injection, and corresponds to a tune shift of 0.25 due to space charge forces. The tune shift with an average current 20 nA should then be only 1/3 of the conventional limit, and therefore not introduce significant problems in the tune-up.

This smaller machine is labelled PAMA-1 and will require a conventional extraction system for protons, either fast (single turn) extraction, or resonant extraction. The maximum energy to which H^- ions could be accelerated in this machine would be 130 MeV (12 cm penetration in soft tissue). This energy could be useful for some applications and would allow investigation of the potential of charge exchange extraction and dynamic raster scanning.

These magnets (two times the number) arranged in a ring of twice the diameter is called PAMA-2, and is the synchrotron designed under the grant. It would allow acceleration of H^- ions to 250 MeV at a peak field of 5.6 kG. At 11.2 kG, it would be capable of accelerating helium ions to 250 MeV/u, the energy required for the full range of radiation treatment. Energy deposition by helium ions has the same biological effect as that of protons, but the reduced multiple coulomb scattering could justify the higher cost for those cases in which sensitive tissue is to be avoided and sharpness of dose cutoff is very important.

To estimate the vacuum requirements for H^- ion acceleration, the total cross-section for conversion of the H^- to either neutral or positive ions by

collision with residual gas molecules is used.^[11] In this calculation, the loss is estimated for a vacuum of 10^{-10} Torr, with the residual gas composed of 95% hydrogen and 5% oxygen. This composition is equivalent to that measured in high vacuum tests. With this composition, the total cross-section drops rapidly from $3.8 \times 10^{-18} \text{ cm}^2$ at 10 MeV to $1.4 \times 10^{-19} \text{ cm}^2$ at 250 MeV. It is estimated that about 1/6 of 1% of the accelerated beam is lost by this process during the 0.4 sec extraction time, and perhaps double this amount during the 0.3 sec acceleration time. Such losses are quite tolerable, although they would begin to be of concern if the pressure were an order of magnitude higher, or if the residual gas contained a larger percentage of oxygen or similar molecules. Since 10^{-10} Torr appears able to be achieved in a reliable way, the vacuum requirements will not be an impediment to the acceleration of H^- ions.

3. Conceptual Design

The design described is that of PAMA-2 for H^- ions at 250 MeV (although the magnets can operate at higher fields and repetition rates). Some of the parameters of the accelerator are listed in Table 1.

The synchrotron consists of 64 small bending magnets, each 52 cm long (including coils), arranged in a ring of 45' diameter. The ring contains 16 quadrupole magnets for focusing. All magnets are arranged in eight groups, separated by 70 cm straight sections. A schematic of this layout is shown in Fig. 1. Extraction at any energy is possible from any one of the eight straight sections, unless the equipment in a given straight section precludes this (one straight section is occupied by the injection magnet and another by the RF acceleration system). During normal operation, beam is injected when the magnetic field is at the proper low value, acceleration takes place for 0.3 sec while the field is increased to its final value, and beam is extracted during 0.4 sec while the field is held constant. The field is then reset to the injection value in 0.3 sec. The entire cycle requires 1 sec with external beam available for about 40% of the time.

The dipoles, with outside dimensions of $20.3 \times 26.7 \text{ cm}^2$ have 7.5 cm wide pole faces and a vertical gap of 3.2 cm. The magnetic field is uniform to within 0.1% over an area of $2 \times 4 \text{ cm}$. The magnets are straight, with iron length 38.9 cm, and the coils are off the midplane for simplicity of construction. A current of 361.8 A is required in the 20 turn coil to achieve a field of 5.6 kG. The quadrupoles also have the coils off the midplane, and the 9-turn coil has a current of 330.3 A for a gradient of 8.25 T/m with a "stay-clear" aperture to accommodate the elliptical vacuum chamber of the dipoles ($3.2 \text{ cm} \times 9.5 \text{ cm}$).

The vacuum chamber will be an extruded aluminum chamber with wall thickness of 1/8 inch, except on each side, where wall thickness will be increased to include a channel along its entire length for baking and cooling. The elliptical chamber will have outer dimensions of 3.0 cm vertically and 9.5 cm horizontally. A Zr/Al getter strip will be attached inside the chamber on the inner radius. When activated, this strip will have a pumping speed of about 1 liter/sec/cm² of surface area, sufficient to maintain the chamber, initially chemically cleaned and baked in-situ, below a pressure of 10^{-10} Torr. Twelve 11 liter/sec ion pumps will be distributed around the ring to pump noble gases and organic molecules such as methane, which are not affected by the getter

material. Two small turbomolecular pumping stations will be used for pumpdown and during bakeout. The latter, which consists of heating the chamber walls to drive entrapped gases from off the surface, will be carried out by passing hot water under pressure through the channels of the chamber to achieve a temperature of 150° C. The outer surface of the chamber will be covered with 1 mm of insulation to reduce heat losses during bakeout.

The RF accelerating system is required to produce an energy gain of 300 eV/turn at a frequency synchronized to the increasing velocity of the protons in the rising magnetic field. The frequency varies from 0.5 MHz with 1.5 MeV injection to 5.4 MHz at 250 MeV. The relatively large frequency swing is produced by a ferrite-loaded cavity. Two different types of ferrite rings provide the frequency response to cover the range.

The beam is injected through a fast-pulsed magnet onto the center line of the accelerator ring for a time equal to 1/2 revolution, 1.1 μ sec. The magnetic field must then be reduced to zero before the head of the beam bunch passes through it a second time. An injected current of 3 mA with a pulse repetition rate of 1 Hz then leads to an average current of 3 nA, sufficient for most treatment requirements if efficient raster scanning beam delivery is utilized.

The 1.5 MeV injected H^- beam is produced with a source in the terminal of a National Electrostatic Corporation (NEC) tandem accelerator, 5SDH-2. The latter has a proven record of reliability in commercial use. The use of a tandem machine as a preaccelerator provides a significant degree of flexibility to the injection system in that protons can be accelerated to 3 MeV and helium ions to 4.5 MeV. Injection of 3 MeV protons would allow double the beam current in the main accelerator when raster scanning was not required.

The major feature of the medical accelerator designed is the simplicity, flexibility, and reliability of the extraction process at full energy while the magnetic field is held constant at its peak value. Extraction makes use of two small magnets located one octant before and after the extraction straight section. Energizing these two magnets with equal currents moves the beam between the magnets, with maximum amplitude at the extraction straight section, while the beam position in the remaining 3/4 of the ring does not change. In this way, a very small and controlled portion of the circulating beam can be made to pass through a very thin foil (1 mg/cm²), one magnet upstream of the extraction straight section. Signals from a current monitor in the extracted beam can be connected back to the two fast magnets controlling the beam position at the foil. This technique can provide control of the extracted beam current to the desired value with considerable precision.

The simplicity and low cost of this extraction method allows it to be duplicated in any straight section in which other equipment (such as the RF accelerating cavity and the injection magnet) does not preclude it. Therefore, the beam can be extracted in at least four of the eight straight sections, and this provides a significant flexibility in the use of the accelerator. Furthermore, foils placed in all eight such locations around the ring, with sections on the inside, outside, top, and bottom of the beam aperture (like a window frame around the beam), can insure that any component failure or mistuning of the accelerator will result in extracted beam rather

than ions striking the vacuum chamber wall. Specialized shielding can then minimize radiation problems due to accidents, and the accelerator shielding cost will be minimized. This feature could result in a synchrotron with unusually low residual radiation levels.

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Table 1. Parameters of the Medical Accelerator, PAMA-2

Type of Ion	H ⁻
Max. Energy (MeV)	250
Injection Energy (MeV)	1.5
Extraction Type	Chg. Exch.
Duty Cycle (%)	40
Repetition Rate (Hz)	1
Avg. Current (nA)	3
Number of Dipoles	64
Peak Field (kG)	5.6
Good Field Region (cm)	2 x 4
Number of Quadrupoles	16
Gradient T/m	8.2
Beta Functions, max/min (m)	8.8/1.3
Eta Function, max (m)	2.5
Tunes ν/h	2.44/2.17
Ring Diameter (m)	13.7
Magnet Power (kW)	77

Figure Caption

Figure. 1. Schematic layout of the 64 dipoles and 16 quadrupoles of the Proton Accelerator for Medical Applications (PAMA). The diameter of this version is 45'.