THE ACCTEK PROTON MEDICAL ACCELERATOR AND BEAM DELIVERY SYSTEM FOR CANCER THERAPY AND RADIOGRAPHY *

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Abstract
ACCTEK has developed prototype magnets and vacuum chambers for its concept of the proton medical accelerator under grants from the U.S. National Cancer Institute. The synchrotron would accelerate H^- ions to 250 MeV and utilize charge exchange extraction. Under development also is a beam delivery system consisting of a raster scanning magnet system and a rotating 90^0 bending magnet system. Conceptual ideas and progress on these systems are discussed. Also discussed are the feasibility of proton radiography and computed tomography with accelerators designed for proton therapy.

1. Introduction

ACCTEK Associates has been awarded grants from the United States National Cancer Institute for research and development on its concepts of a proton medical accelerator and a proton beam delivery system for radiation treatment of cancer. These are under the Small Business Innovative Research program applicable to all large U.S. Agencies with the purpose of commercializing technology developed with Federal funds.

The conceptual design of the Proton Medical Accelerator has been reported previously [1]. Briefly it would be a synchrotron to accelerate H^- ions to 250 MeV in a ring of small straight magnets of relatively small aperture, and utilize charge exchange extraction in a thin foil at any desired energy. To accommodate H^- ions without substantial losses due to magnetic stripping and neutralization by the residual gas the accelerator would operate with a peak field of 5.6 kG and a vacuum of 10^{-10} Torr. The ring diameter would thus be relatively large, 13.7 m, and the ultra high vacuum would be achieved by nonevaporable Zr/Al getter strips, in an extruded aluminum vacuum chamber. Feedback from extracted beam current monitors to two fast orbit bumping magnets, would be utilized to control the extracted beam current. The nominal repetition rate would be 1 Hz, with 0.3 sec rise and fall time of the field and 0.4 sec at constant field to accommodate extraction for 0.3 sec. At this rate the average beam current would be 3nA. The magnets are designed for two times this rate of rise, 2 Hz with a 0.2 sec flat top or 3.3 Hz without flat top, with corresponding increase in the average current.

The SBIR grant on the Proton Medical Accelerator was completed in November 1989. Twelve small prototype magnets of accelerator quality were constructed at Argonne National Laboratory under a Work-For-Others contract. These are shown in Figure 1 in various stages of completion. They are arranged in an arc that is equal to the design value for the medical accelerator. Aluminum extrusion vacuum chambers were also produced and bent to the proper radius of curvature without serious distortion of water holes for baking and cooling nor distortion of the structure for mounting the getter strips on the inside radius. This was an important issue not at all certain in the initial design.

The low field and use of many small straight magnets for the ring has led to some interesting alternatives because of the ease of increasing the field and redesigning the lattice layout. Many of these have been discusses previously [2] and are not repeated here.

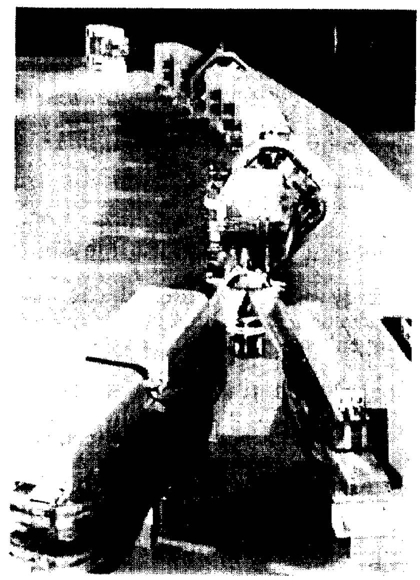

Fig. 1. Prototype magnets for H^- accelerator, arranged on an arc of a circle of 13.7 m diameter.

2. Raster Scanning System

Under the current grant from the National Cancer Institute ACCTEK is developing a raster scanning system for 250 MeV proton beam delivery. This consists of a fast horizontal scanning magnet operating at 660 Hz followed by a slow vertical scanning magnet at 3 Hz as shown in Figure 2. The field scanned, with a lever arm of 3 m from the center of the vertical magnet, is 25 cm horizontally and 35 cm vertically. The arrangement is similar to one being developed at Lawrence Berkeley Laboratory [3] for raster scanning of heavy ions, although the ACCTEK system scans at a much faster rate. The rate is determined by the desire to complete a single scan of the entire area in 0.17 sec to accommodate 2 Hz operation of the accelerator.

The proposed scanning method is to scan a fixed field on each pulse and modulate the dose by controlling the extracted beam current. At a given depth the beam would be turned "on" when the leading edge of the sweeping beam intersected the desired contour of 100% dose at that depth and "off" when the trailing edge intersected the opposite boundary of that contour. The term "off" might mean reduction to a low value, say 5%, in order that the time of arrival of the beam at the end of the sweep can be detected by scintillators. The latter is but one link of a multielement safety system that is being

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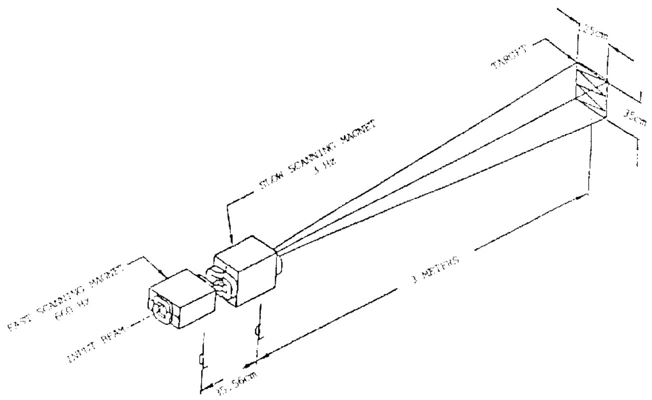

Fig. 2. Raster Scanning System

studied but will not be described. The transition from full dose to a low level is then about one beam diameter beyond the specified contour, with corrections for multiple coulomb scattering.

The scanning magnets being constructed are made of grain-oriented steel and are tape wound on a mandrel. The core is then cut, pole faces machined, and C cores assembled into H type magnets. The fast scanning magnet is made of 4 mil Selectron tape and the magnet steel is 8 3/4" wide, 4" high and 8" long. The gap is 1 3/4" x 0.787" high. The vertical scanning magnet is made of 12-mil Selectron tape and the magnet is 10" wide, 8 1/4" high, and 7 3/4" long. The gap is 3 1/4" wide by 2 1/32" high.

The proposed ability to control the extracted beam current with some precision has greatly simplified the design (and reduced the cost) of the power supply for the horizontal scanning magnet. Rather than producing a linear ramp, requiring a power supply with 200 kVA ratings and switching problems at zero current, it appears that a simpler resonant circuit will be adequate. The horizontal sweep speed will then vary with time so that the beam current will have to have a sinusoidal variation in order that the radiation dose remain uniform with time. This can be accomplished by inserting an ac wave in the feedback loop to the magnets that control the circulating beam position at the stripping foil in the accelerator.

The power supply for the vertical scanning magnet does not present a difficult problem because the power requirement is much lower at the slower sweep rate, and a jitter in zero crossing of the current is not nearly as critical. For this supply a linear ramp would be adequate. It is also possible to reduce the deviation from a straight line of the horizontal sweep by an order of magnitude (to 35μ for a 1.5 mm vertical travel during one horizontal sweep) by the introduction of a small second harmonic current of the proper phase and amplitude into the vertical scan magnet supply.

The design of these power supplies is not yet complete but it is planned to test these ideas with a low energy beam.

3. Rotating Beam Delivery System

Also being developed as part of the SBIR grant is a low cost alternative to the 360° vertical gantry. It is proposed that a vertically downward directed beam, able to be rotated through an angle up to 60°, along with horizontal beams in the same and other treatment rooms, could serve most of the functions of the gantry and at much lower cost. Capital and operating cost are seen as critical issues in the widespread development and utilization of proton therapy.

A schematic of the layout of such a system is shown in Figure 3. A 90° bending magnet system is capable of rotation about the incident beam axis to direct the beam vertically downward to a treatment room below beam level or to either of two horizontal beam treatment rooms at the beam level. A counterweight extends above the axis of rotation to balance the system.

It is believed, although a detailed study has not been carried out, that the shielding of each of the rooms is adequate that they could be utilized independently so that setup of a patient in one room could be performed while treatment was carried out in another. Such a flexibility of multiple treatment rooms is important for the economic feasibility of the proton treatment facility whereas the gantry system services only one treatment room per gantry. With the magnets in the 90° bending system turned off the beam emerging through the center of rotation could be transported downstream to be utilized in other treatment areas, possibly in a duplication of the layout shown.

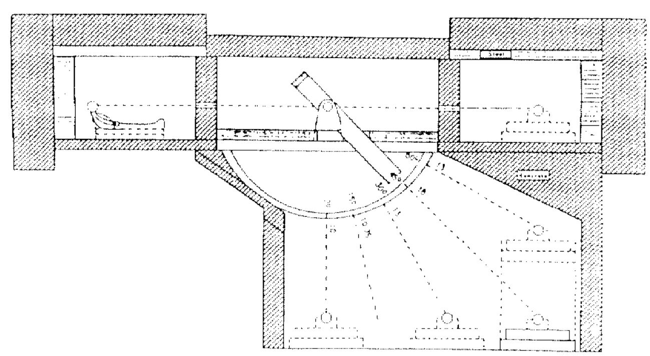

Fig. 3 90° Rotating Magnet System for Beam Delivery to Three Treatment Rooms

Not shown, but important to the efficient utilization of this layout is a horizontal beam into the vertical treatment room at the patient level. In what might become the most widely used mode, treatment in this room could be carried out in 3 fields, two horizontal beams 180° apart (obtained by 180° rotation of the patient in the horizontal plane) and one vertical beam.

In the vertical treatment room variation in the incident angle through 0-45° requires patient motion in the horizontal plane. It is proposed to accomplish this by having the patient cart mounted on rails (lowest support), able to be rotated 360° (next support level), and finally laterally along the patient axis (upper support level). If treatment at 45-60° from the vertical is desired it might be accomplished by utilizing a vertical elevator as shown in order not to further extend the lever arm for treatment. It is believed that this patient motion can be accomplished with the speed and precision with which a vertical gantry can be positioned and that multiple field treatment will be sufficient to reduce the dose to healthy tissue so that the missing flexibility (compared to a gantry) of treatment from below is not a serious drawback.

The raster scanning system described above, and diagnostics not described, would be carried on the rotating system and thus serve all three treatment rooms. It is designed for the minimum distance, 3 m, so that the longer distance, 5.5 m at 45°, can easily be handled. This variation of lever arm with angle does not present a problem to the raster scanning system, which will be under computer control.

A prototype of this rotating support system is under construction and its angular reproducibility will be demonstrated. Six small straight bending magnets of a type similar to those designed for the accelerator will be mounted on the support to provide 90° of bending, along with one quadrupole and the scanning magnet system described above. The total magnet weight is about two tons so that the total system with counterweight is just over four tons. It is anticipated that it can be rotated 90° in one minute.

4. Proton Radiography

The application of protons to medical radiography was pioneered by A. Koehler and V.W. Steward at Harvard [4]. The initial work was extended at Argonne National Laboratory[5], at Los Alamos National Laboratory[6], and with alpha particles and heavier ions at Lawrence Berkeley National Laboratory[7]. The results clearly demonstrated the superiority of density resolution in soft human tissue of ion beams over x-rays at equivalent deposited radiation dose. This application of protons has not been pursued vigorously, however, because suitable proton beams did not exist to make this a practical technique. With the development of dedicated proton machines for cancer therapy, and especially of fast raster scanning beams, the advantages need to be reexamined.

While 300 MeV protons would be desirable for this application, the therapy beam of 250 MeV, with a range in soft tissue of about 38 cm water equivalent, is adequate for complete penetration of the head and most sections of average size humans. The low intensities required for a projection radiograph, a few times 10^8 protons, is sufficiently low that high quality proton beams can be produced at most machines by collimation. The capability for proton radiography can therefore be usefully developed at any dedicated therapy facility. Perhaps the most serious limitation of proton radiography is the loss of spatial resolution due to multiple coulomb scattering. The problem is minimized on the average, however, by knowledge of the transverse position of the entrance beam rather than that of the position at exit because of the cumulative buildup of angular divergence and beam size as the protons penetrate deeper into tissue.

Projection radiographs with protons in situ just prior to therapy could prove to be useful. It would directly check the patient alignment, and the calibration and operational status of the accelerator, transport, scanning system, and monitors. It could detect quantitatively any changes from previous treatment fractions, and any temporary problems that would affect the precision of dose delivery.

The potential for proton CT scanning also seems clear with raster scanning of a high quality proton beam. With a scanned beam size of 1 x 2 mm^2 and 360 line scans at 1° increments good computed cross sections should be obtained with 10^3 protons/2mm^2. The estimated dose for this procedure, provided scanning planes were no closer than 5 mm apart, is about 0.5 rem, comparable to or lower than that for an x-ray CT exposure. For upright patients, particularly applicable to images of the head, the time for a proton CT is only limited by the rate of rotation of the patient. A rotation period of 10 or 20 seconds seems reasonable. For reclining patients, of course, a gantry would be required for proton CT.

Not only is it anticipated that the proton CT scan will give more precise density data than x-ray scans, but it measures proton stopping power directly. The potential exists therefore of improving treatment planning by eliminating the use of CT conversion numbers with their uncertainty because of x-ray beam hardening.

Finally the use of proton CT as a stand alone technique independent of therapy seems sure to find wide application once the capability exists.

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