DESIGN AND APPLICATION OF A PROTON THERAPY ACCELERATOR

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It was pointed out long ago by R. R. Wilson[1] that proton beams are clinically valuable because of their well defined boundaries and the Bragg peaking of their energy deposition in matter at the end of the range. To make a proton therapy facility feasible one must, however, have precise information of the location of the treatment volume and the characteristics of the overlying tissue. It is only within the last decade that new imaging techniques, computed tomography (CT scanning), Magnetic Resonance Imaging (MRI), positron emission tomography (PET) and ultrasound, have been available to give adequate accuracy. Thus the time is now ripe for the construction of a full-time proton therapy facility in a major medical institution. This paper describes the design of the Proton Therapy Facility for the Loma Linda University Medical Center.

The Accelerator

A top energy of 250 MeV is needed for the beam to penetrate the maximum thickness of a human body. The peak beam current requirement is taken to be 10nA so as to limit the irradiation time of a treatment to no more than two minutes. A synchrotron is chosen because it can provide the requisite beam current and because of its easy energy variability to treat disease sites at different depths. Fig. 1 shows an isometric drawing of the accelerator and Table 1 lists the major parameters. We will discuss here only the unique features of the design.

1. As injector we chose an RFQ linac. The operation of an RFQ is simple and reliable, and the proton source is exposed and easily serviceable (compared with an electrostatic accelerator). Provisions are made for possible future addition of a drift-tube linac section to boost the injection energy to a higher value. A debuncher is used to reduce the energy spread and the one-turn injection is accomplished by vertically kicking the beam onto the closed orbit with an electric kicker.

Table 1. Accelerator Parameters

Synchrotron ring	
Injection energy	1.7 MeV
Extraction energy	70–250 MeV
Circumference	20.053 m
Number of sectors	4
Straight section lengths	0.5 m, 2.0 m
Number of dipoles	8
Dipole length	1.257 m
Dipole end angle	0.328 rad(18.8°)
Injection field	0.118T
Extraction field	0.77–1.52T
Betatron tunes	0.60(H), 1.32(V)
Magnet pulse shape rise/flat-top/return	0.5/1.0/0.5 sec
Injection	
Injector type	RFQ linac
Energy	1.7 MeV
RF frequency	425 MHz
Length	220.5 cm
Output beam current	50 mA
Injection scheme: Vertical, single turn injection with electric kicker	
Acceleration	
Cavity type	untuned, ferrite loaded
Harmonic number	1
Frequency range	0.899–9.17 MHz
Energy gain per turn	90.0 eV
Peak cavity voltage	330 V
Extraction	
Horizontal half-integer resonant extraction with electrostatic wire septum and Lambertson iron septum magnet	

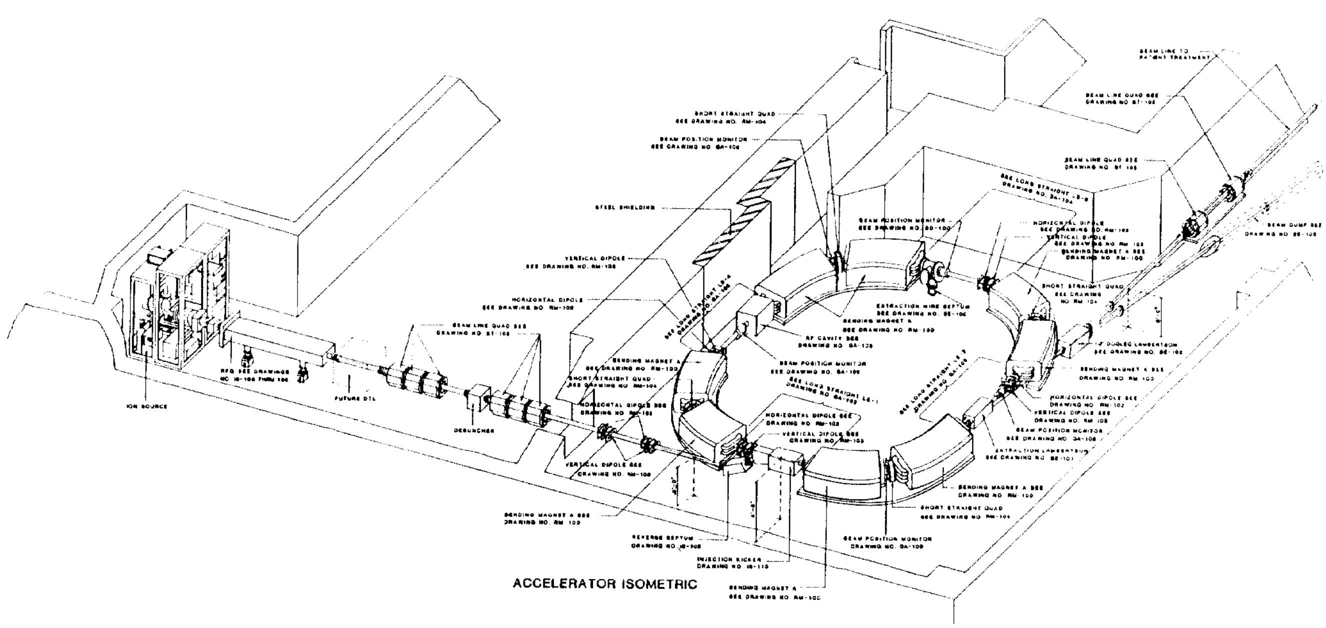

Figure 1: Isometric drawing of the accelerator system.

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2. The synchrotron has a weak-focusing lattice using only wedge focusing at dipole ends. For a small synchrotron with only four cells, alternating gradient focusing is not necessary in order to obtain a phase advance of some 70° per cell to minimize β_{max}, thereby minimizing the required aperture. The weak focusing lattice gives a large frequency dispersion which leads to high instability thresholds. The expected intensity is $\sim 3\text{x}10^{11}$ protons/pulse.

3. A magnet cycle is composed of a 0.5 sec linear rise, a 1.0 sec flat-top and a 0.5 sec return to start. Acceleration to 250 MeV in 0.5 sec requires an energy gain of only 90 eV/turn. However, to create adequate bucket size to contain the rather large momentum spread of the beam requires a peak voltage of 330V. The frequency swing is large, a factor 10.2 to 1. The cavity is untuned, but it is ferrite loaded to maintain a uniform impedance of 50 Ω over the entire frequency range. This rather large uniform cavity impedance also necessitates high beam-instability thresholds.

4. Half-integer slow extraction over a 1 sec flat-top is provided for a dynamic beam-spreading system. This will be discussed in greater detail below.

The Treatment Facility

While the design of the accelerator system is more-or-less conventional, very little experience is available for the design of the beam-transport and beam-delivery systems. Therefore, by necessity, more innovations are incorporated in these systems.

By far the largest part of the time a patient spends in a treatment room is used for accurate alignment of the treatment volume and the beam. As a consequence, multiple treatment rooms are needed to match the capability of the accelerator. A total of five treatment rooms is planned. Two of the rooms are equipped with only fixed beams. The three principal all-purpose treatment rooms are identical, each equipped with a gantry on which the beam transport line is mounted so that the beam can be rotated and directed to enter the disease volume from different directions, thereby minimizing damage to the healthy overlying tissue. Fig. 2 gives the plan layout of the entire facility showing all five treatment rooms and Fig. 3 is an isometric drawing showing the beam lines and beam-line magnets. In addition to the accelerator, this drawing shows the extracted beam trunk line, the two fixed beams with 180° bend each and the three gantry beams each containing a total bend of 360°. The first horizontal bending magnet of each branch beam acts as the switching magnet. When it is turned off the beam goes through a hole in the yoke straight on to the switching dipole of the next branch beam.

Before entering a gantry the beam first passes through a series of six rotating quadrupoles. The 6–quad chain produces transfer matrices unity in one plane and minus-one in the other plane. When the quad-chain is rotated by angle $\theta/2$ it rotates the phase planes of the beam by θ to match a θ-angle rotation of the gantry. Thus with fixed tuning of the gantry magnets the focal condition of the beam at the isocenter can remain unchanged for all gantry angles.

To irradiate the whole treatment volume the narrow proton beam must be spread laterally and ranged longitudinally. The beam spreading and ranging can be either passive or active. Passive spreading and ranging are achieved by placing scatterers and energy absorbers in the beam. In an active system the beam is scanned laterally by two dipoles sweeping the beam in orthogonal directions and ranged longitudinally by varying the accelerator energy. Only the passive system has been used in actual treatment. It is likely that the passive system will be employed during the initial operation. As more operating experience is gained on the facility, the more efficient and more precise active system will be developed and applied.

The Controls

The success of the full-scale application of proton therapy in a hospital environment depends crucially on the ease, the flexibility and the reliability of the control system. As designed, the control system consists of multiple distributed microprocessor-based systems

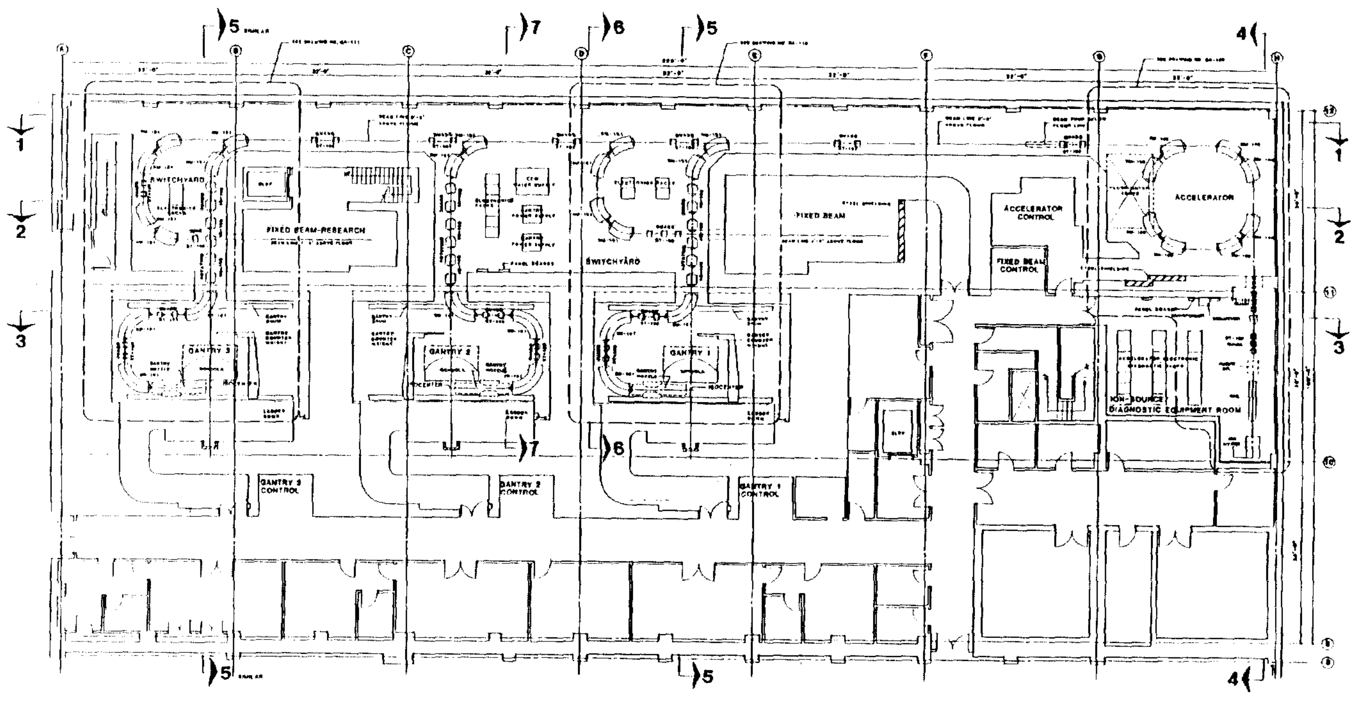

Figure 2: Plan view showing the layout of the total facility.

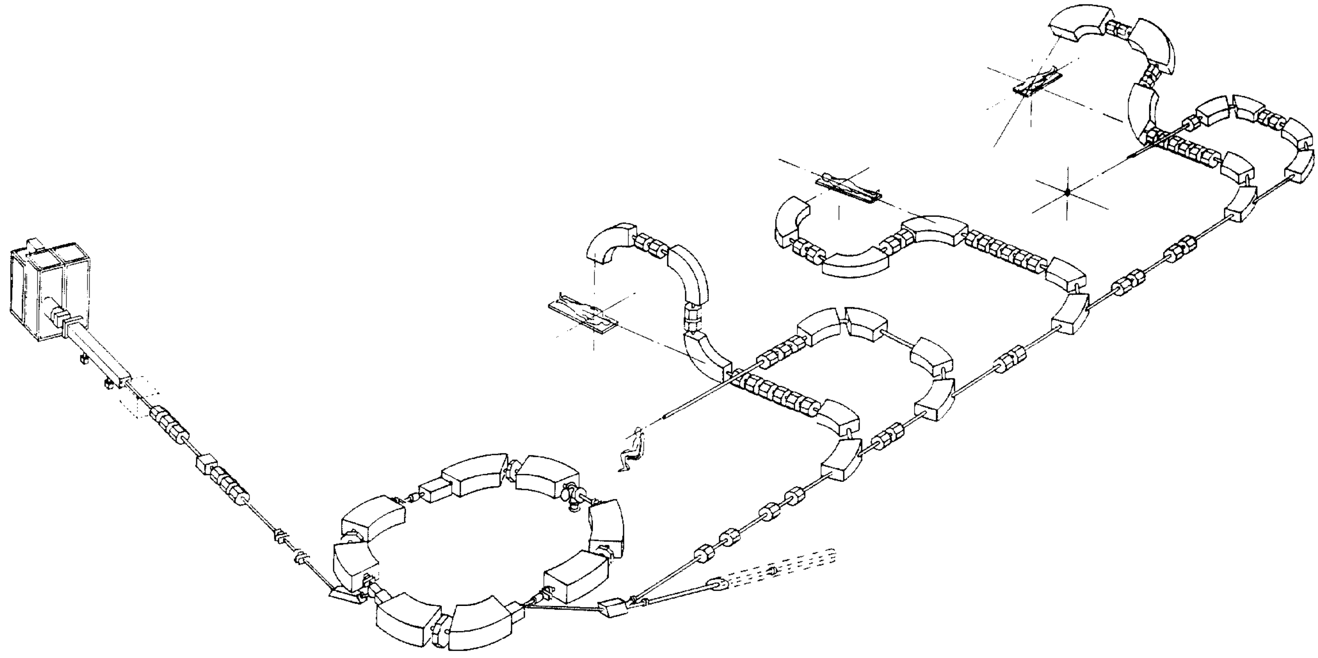

Figure 3: Isometric diagram showing the treatment facility beam lines and beam line magnets.

networked together and to the central MASSCOMP computer using the IEEE-802.5 Local Area Network (LAN) standard. When the active beam-spreading is used the beam spill from the accelerator should be uniform and controllable to an accuracy of $\pm$(1-2)% and the energy should be able to be reset from pulse to pulse with similar accuracy. A table of operational parameters such as scan-line positions and limits, beam energy, required beam intensity at each energy step, etc. must be set up for each treatment and stored on the patients' floppy disks. In addition to the active management of all treatment operations, the control system must also perform a great number of monitoring, updating, scheduling, recording, reporting and display tasks.

When fully developed it is estimated that this facility will be able to provide more than 20,000 treatments each year.

Reference

[1]R. R. Wilson, Radiobiology 47, 487 (1946)