Preliminary Design of a Dedicated Proton Therapy Linac

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Abstract

The preliminary design has been completed for a low current, compact proton linac dedicated to cancer therapy. A 3 GHz side-coupled structure accelerates the beam from a 70 MeV drift tube linac using commercially available S-band rf power systems and accelerating cavities. This significantly reduces the linac cost and allows incremental energies up to 250 MeV. The short beam pulse width and high repetition rate make the linac similar to the high energy electron linacs now used for cancer therapy, yet produce a proton flux sufficient for treatment of large tumors. The high pulse repetition rate permits raster scanning, and the small output beam size and emittance result in a compact isocentric gantry design. Such a linac will reduce the facility and operating costs for a dedicated cancer therapy system.

I. INTRODUCTION

The use of electron linacs to treat cancer has experienced an explosive growth during the past two decades, with more than 5,000 accelerators estimated to be in use worldwide. Although it is widely recognized that proton beams have major advantages over these photon and electron beams, their widespread use has not been realized due to the lack of dedicated facilities. However, more than 9,000 patients have been treated worldwide with ion beams at institutions with physics research accelerators[1]. The success of these programs has led to significant interest within the medical community in dedicated proton treatment facilities. The first such system, a 250 MeV synchrotron at Loma Linda Medical Center in California, is now treating patients[2].

As pointed out by Mandrillon[3], the key requirements for a dedicated proton therapy accelerator are: (1) it must be compact enough for installation in a large hospital; (2) it must be highly reliable and easy to operate; (3) it must be compatible with beam scanning and isocentric gantries; and (4) it must have a maximum energy greater than 200 MeV with an intensity sufficient to treat large tumors in short irradiation times. Other important considerations include safety, ease of maintenance and costs. The costs include the equipment, the facility, and the operating costs. While the synchrotron facility at Loma Linda appears to be technically viable, its cost and complexity have caused radiation therapy groups to consider other approaches.

Conventional proton linacs are routinely used for injecting beams into synchrotrons, but they are considered too expensive and too powerful for low current, high energy proton acceleration. However, a "proton version" of the conventional S-band electron linacs used for radiation therapy satisfies all of the requirements for a dedicated proton therapy accelerator. It is possible to use an S-band side-coupled structure for this application due to the very low current required from the linac. The use of multiple side-coupled (SCL) linac tanks and rf power amplifiers allows variable output energy. A compact, high frequency drift tube linac (DTL) can be used as the injector to the SCL at 70 MeV, with a conventional radio frequency quadrupole (RFQ) linac as its input.

II. LINAC SPECIFICATIONS

The linac proposed as a dedicated proton therapy accelerator is shown schematically in Figure 1. Specifications for this compact high energy proton linac, the AccSys Model PL-250, are listed in Table I. By using the RFQ linac as a "chopper" for the pulsed proton beam from a small duoplasmatron ion source, the current per pulse in the linac could be controlled using the beam pulse width. By varying the beam injected into the RFQ using an electrostatic lens and varying the beam pulse width and pulse repetition rate, the average output beam intensity can be varied from a few nA up to a few hundred nA. Goitein[4] has indicated that these currents are adequate for treating a wide range of tumors. Similarly, by turning on one or more of the SCL klystrons, the output beam energy can be varied in eleven steps from 70 to 250 MeV of approximately 18 MeV each. Continuous beam energies can be achieved between these steps with little loss in intensity by using degrading foils.

The key to the reliability, economy, and clinical compatibility of this proton therapy linac is the availability of the SCL cavities and rf power systems from an established medical linac manufacturer. Using such proven components will make the proton linac configuration similar to existing high energy medical electron linacs. The same pulse structure would be used and the output current would be varied by adjusting the repetition rate and injected peak current. Compatibility with existing electron linac protocol could be developed. The high repetition rate would be compatible with beam scanning, although the beam current would be more than adequate for treatment of large tumors using the beam scattering technique.

The maximum duty factor of this linac is dictated by the SCL rf power systems, but it is more than adequate for the maximum current required. As shown in Table I, even

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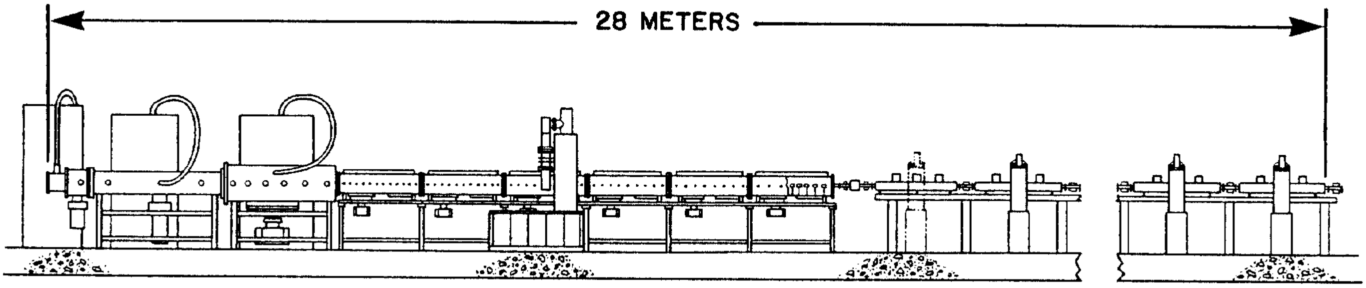

Figure 1. Schematic Layout of Model PL-250 Proton Therapy Linac.

though the peak rf power requirement is large, the maximum input ac power is much lower than other types of accelerators at this energy. Much of the input power is required for the rf systems and, since they will be turned off between treatments, the stand-by power is very low.

Table I
Preliminary Specifications for
a Dedicated Proton Therapy Linac

Accelerated particle	H^+	
Maximum beam energy	250	MeV
Minimum beam energy	70	MeV
No. energy increments	11	
Peak beam current	100-300	μA
Beam pulse width	1-3	μsec
Repetition rate	100-300	Hz
Average intensity	10-270	nA
Beam emittance (norm.)	<0.1	πmm-mrad
Beam energy spread	±0.4	%
Max. rf duty factor	0.125	%
Peak rf power	62	MW
Maximum input power	350	kW
Stand-by power	25	kW
Accelerator length	28	m

III. Linac Description

A. RFQ Injector

The initial stage of the Model PL-250 is a 3 MeV RFQ linac operating at a frequency of 499.5 MHz. A compact duoplasmatron ion source and electrostatic einzel lens inject up to 1 mA of 30 keV protons into this 2.43 m long structure. Operating with a vane-to-vane voltage of 50 kV, it requires a peak rf power of 0.2 MW, and has calculated beam transmission of 96%.

B. DTL Linac Section

The RFQ linac is closely coupled to the first tank of the DTL, which also operates at 499.5 MHz. The first four quadrupole magnets in this tank are used to transversely match the beam from the RFQ into the DTL. It has an output energy of 12.5 MeV, is only 1.87 m long, with 26 cells, and requires a peak rf power of 1.0 MW. There is no longitudinal matching into this DTL tank, so the beam rotates in longitudinal phase space. The synchronous phase of the structure is ramped instead to produce a good longitudinal match into the main DTL tank, which is operated at 999 MHz. The last four quadrupole magnets are used to transversely match the beam into the main DTL tank which accelerates the proton beam from 12.5 MeV to 70.4 MeV over a length of 7.92 m. It has a total of 98 accelerating cells and requires a peak rf power of 8.2 MW. The synchronous phase is constant at -30° and there is a PMQ assembly in every other drift tube, with the magnetic field gradient tapered down the length of the DTL. The bore in the drift tubes in both sections of the DTL is only 1 cm diameter, but the beam transmission is calculated to be 100%, even if the quadrupole magnets have random displacements as large as ±0.005 inch.

C. Side Coupled Linac Section

The largest section of the Model PL-250 Proton Therapy Linac is the SCL, which operates at 2997 MHz. Between the DTL and SCL is a 1 m matching section of two quadrupole magnets and a single SCL tank operating at a synchronous phase of -90°. This short tank is 0.2 m long and requires 0.3 MW of rf power.

The SCL consists of ten accelerator modules, each powered by a 7 MW S-band klystron system. Each module consists of 4 accelerating tanks brazed together using 11 equal length accelerating cells with a beam bore of 4 mm diameter. The four accelerating sections are connected with $3\beta\lambda/2$ bridge couplers to allow a single permanent magnet quadrupole to be placed between them. The quadrupole focusing gradient is ramped in steps along the length of the modules. The total length of the SCL is 14 m.

The calculated 250 MeV beam profile through the DTL and SCL is shown in Figure 2. No beam is lost during the acceleration. Calculations also show that the 70 MeV beam from the DTL can be transmitted through the SCL without any losses, or it could be deflected in the transition region to a separate treatment area.

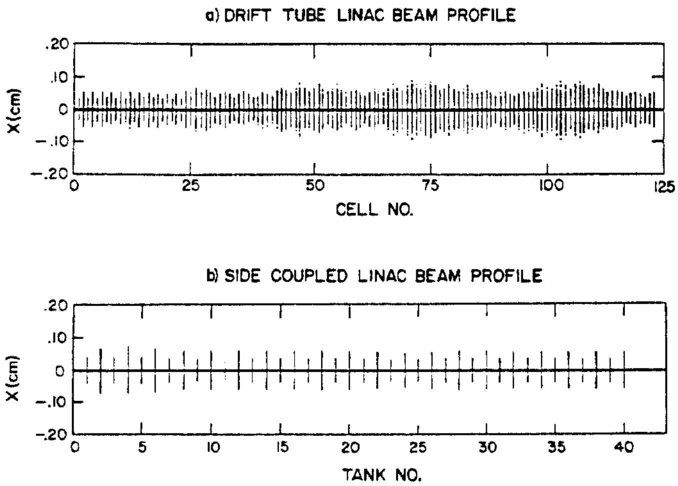

Figure 2. Beam Envelopes Through the Model PL-250 Linac.

D. RF System

Each section of the linac is powered by a separate rf power amplifier driven by a master oscillator and timing circuit. The RFQ and first DTL tank are powered by rugged planar triode amplifiers, developed by AccSys for use with rf structures. The main DTL tank is powered by a single 10 MW klystron. This klystron would be a commercial L-band unit modified to operate at 999 MHz.

The largest number of rf power systems would be required for the SCL. These 10 S-band klystron systems would be production models manufactured for medical electron linacs. The modulator is capable of pulse widths up to 5 μsec and it can be operated at a duty factor of 0.0012. Each klystron typically requires 25 kW of input ac power when operating at its maximum duty factor. It is a self-contained unit that can easily be interfaced to the master driver chassis.

E. Vacuum System

The ion injector and RFQ linac will be pumped by commercial cryopump systems to effectively handle the hydrogen gas load from the ion source (<0.5 sccm). The DTL structure will be pumped by clean, low maintenance ion pumps, since the gas load in this section will be minimal and it will seldom be opened to atmospheric pressure. The SCL modules will also be pumped by ion pumps, just as in electron linacs, but with slightly larger pumps because the modules will not be sealed. The rf power windows will be mounted in the middle bridge coupler of each SCL module so they are field replaceable. Vacuum valves will localize sections for repair or service, which will speed up pump down and rf conditioning time.

F. Control

The Model PL-250 Proton Therapy Linac can be operated as a large version of a modern medical electron linac, since it is very stable and rugged. The level of control necessary to achieve this type of operation is minimal, since all of the timing and rf power stabilization is hardwired into the system. It can easily be operated by an upgraded version of the PC-based system used by AccSys for its smaller linacs. After initial start-up each day, control of the linac could be transferred to the treatment control system for delivery of the beam to various treatment areas and switching on and off of the rf systems to control the patient dose. This stand-by mode would greatly reduce the required input power to the accelerator while not affecting the reliability or instantaneous availability of the beam.

G. Beam Delivery System

Beam delivery to treatment areas can be achieved with modest transport elements, since the diameter of the output beam from the linac is less than 2 mm and the emittance in both planes of less than 0.1 πcm-mrad. Since a magnet gap of 1.0 cm or less would be required for the beam, a conventional isocentric gantry can also be constructed using very small magnets, less than 1 ft^3 in size[5]. Future studies will include the design of a compact isocentric gantry to take advantage of the linac beam and satisfy the medical requirements for patient irradiation.

IV. SUMMARY

The Model PL-250 Proton Therapy Linac will provide an economical, reliable, easy to use accelerator for dedicated cancer therapy facilities. Since all of the technology for such a system has been developed, the risks are minimal. For example, the side-coupled structure and its rf power systems (the major portion of the accelerator) are similar to those used by Siemens Medical Laboratory, one of the world's largest manufacturers of medical electron linacs. This significantly reduces the cost of the linac, with the preliminary cost estimated to be less than $8.0M. In addition, system studies show that the building costs and beam delivery system costs are significantly reduced by the use of this linac system. Further studies are planned to finalize the design of this linac and a beam delivery system to make them commercially available for proton therapy.

V. REFERENCES

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