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LIGHT ION LINACS FOR MEDICAL APPLICATIONS

J. N. Bradbury, E. A. Knapp, D. E. Nagle
University of California
Los Alamos Scientific Laboratory
Los Alamos, NM 87544

Summary

Recent advances in linear accelerator technology point to the feasibility of designing and developing practical medical linacs for producing protons, neutrons, or pi mesons for the radiation therapy of cancer. Additional uses of such linacs could include radioisotope production and charged particle radiography. For widespread utilization medical linacs must exhibit reasonable cost, compactness, reliability, and simplicity of operation. Possible extensions of current accelerator technology which might provide these characteristics are discussed in connection with linac design, fabrication techniques, materials, power sources, injectors, and particle collection and delivery systems. Parameters for a medical proton linac for producing pions are listed.

Introduction

Several types of fast particles including protons, neutrons, and pions are currently being evaluated for use in the radiation therapy of cancer. There is already evidence to indicate that some of these particles will prove superior to conventional x rays or γ rays in certain applications. Almost all studies with these new radiations are utilizing accelerators that were originally designed and constructed for physics research; for a variety of reasons these accelerators are not appropriate for adaptation to widespread medical use. It appears likely that as the clinical evaluation programs near completion a demand will be created for medically practical sources of some particles. Consequently it is desirable to examine different classes of accelerators, which could produce therapy beams, in terms of compactness, construction cost, reliability, and simplicity of operation.

The particle energies and currents for some useful therapy beams (the order of 100 R $\min^{-1}$) are listed in Table 1.

Table 1

Primary Beam	Therapy Beam	Average Accelerated Current (μA)	Energy (MeV)
deuteron	→ neutron	15	50
proton	→ neutron	40	70
proton	→ proton	< 1-10 (1)	200
alpha	→ alpha	< 1-10 (1)	700
proton	→ pion	30-300 (2)	500
electron	→ pion	1000-10,000 (2)	500

(1) Depends upon radioisotope production capability.

(2) Depends on collection efficiency of magnetic channel.

This paper will address the potential of proton linear accelerators for producing therapy beams of protons, neutrons, and pions. The therapeutic efficacy of alphas as compared with protons probably will not justify the considerably larger expense of their acceleration with a linac. Also although pion producing electron linacs are feasible the very high beam powers required present serious control and possible component damage problems. In addition to providing radiotherapy beams the proton → proton machine and the proton → pion machine could be used for charged particle radiography. Any of the high current machines would prove useful for radioisotope production.

General Considerations for Medical Linacs

Linacs are capable of accelerating large currents with very low beam loss and generally exhibit high a.c. power to beam power conversion efficiencies. Commercial medical electron linacs in the 4 to 20 MeV range are now available whose operation is simple, reliable, and economical. For primary beams with energies of 200 MeV or less a post-coupled drift tube accelerator can be designed. The proton → proton machine (see Table 1) would need three resonant accelerator tanks, the proton → neutron machine only one tank and one power amplifier. For the proton accelerator to produce pions the drift tube structure would be followed by a side-coupled waveguide structure similar to that employed at LAMPF.

In order to keep the proton accelerator reasonably compact, and thus compatible with typical hospital sites, an energy gradient substantially higher than the currently established 1 to 2 MeV/m is essential. Accelerator economics dictate operating near the breakdown limit and using a low duty cycle. A low duty cycle poses no problem for radiotherapy or radioisotope production but does affect the counting aspects of radiographic or pion visualization applications. High frequency operation will provide a more efficient accelerator with shorter fill times and will simplify system fabrication and surface processing. Overall design of the accelerator and control system must emphasize reliability, economy, and ease of operation. The several types of medical linacs under consideration must to some extent be considered separately (rather than as extensions or modifications of one another) in order to fully optimize parameters from a practical point of view. However much of the technology development required will be applicable to all the linacs and can readily be shared between different detailed designs.

New Directions in Linac Technology

Accelerator Gradient

At present proton linac accelerator gradients are in the range 1 to 2 MeV/m whereas electron linacs for medical purposes operate above 12 MeV/m. It is necessary to increase the proton linac gradient in order to provide the compactness required for widespread medical use. In the commercial electron linacs there are short capture section cavities which demonstrate that gap length is not a critical parameter leading to the high gradients observed. The most important factor in avoiding difficulties at high power levels is the provision of clean, pure, de-gassed drift tube surfaces which implies a high degree of cleanliness in the vacuum technology. Bakeable cavity structures with highly polished surfaces are required. Increasing the operating frequency of the post-coupled drift tube accelerator section reduces dimensions, and thereby makes high vacuum technology more readily applicable; the sparking limit also increases with frequency. A geometry using elliptically shaped drift tubes may also improve the sparking limit.[1] A short duty cycle reduces the average heat power to be removed per unit length. While electron linac gradients may not be realized for proton linacs, gradients in the region 5 to 7 MeV/m (surface fields of 15 to 20 MeV/m) are probably achievable and would result in substantial economy. A gradient of 4.6 MeV/m has been achieved in a prototype drift tube structure.[2]

Operating Frequency

The LAMPF drift tube accelerator operates at the historical frequency of 200 MHz with the high energy side-coupled cavity structure operating at 800 MHz. If the operating frequency of the medical linac drift tube accelerator were raised to 400 MHz (and the side-coupled structure to 1200 MHz) significant improvements in performance and fabrication cost could be achieved. The higher frequency allows a higher energy gradient with concomitant decrease in length and cost and provides a higher shunt impedance implying a more efficient structure with reduced power requirements; smaller diameter tanks, of the order of 40 cm rather than 80 cm, simplify the tank fabrication tremendously and allow high temperature bakeout. The filling time will also decrease by a factor of almost three to the order of 40 μs which makes pulse lengths of about 100 μs feasible. Shunt impedances for a 400 MHz post-coupled drift tube linac accelerating protons to 200 MeV will range from about 20 MΩ/m to 53 MΩ/m where the lower value is not too different from the lowest value for acceleration to 100 MeV at 200 MHz.

The primary difficulties resulting from higher frequency operation are the larger number of drift tubes required and the need for increased radial focusing with decreased space available for focusing elements. These problems are discussed below.

Injector

A high injection energy in the range 2 to 4 MeV ameliorates the problem of radial containment of the beam and allows beam aperture and drift tube spacing requirements of the 400 MHz linac to be more easily satisfied; on the other hand less longitudinal damping is provided than with conventional injection energies of 0.5 to 0.8 MeV. It may prove expedient to use high energy injection for the single frequency system (final energy less than 200 MeV) and low energy injection for the two-frequency pion producing accelerator to improve beam bunching and acceptance characteristics. Final choices must be based on detailed beam dynamics studies. Candidates for the low energy injector include pulsed transformer systems and Cockcroft-Walton generators; pelletrons or Van de Graaff generators could serve as reliable relatively economical high energy injectors.

Focusing

A major cost item in the fabrication of proton linacs is the quadrupole electromagnets needed for radial containment of the beam. In the case of the LAMPF linac the set points of these magnets are exactly those indicated by calculation; if permanent quadrupole magnets could be developed considerable economy could be realized in the fabrication of large numbers of drift tubes. These magnets must be compact and capable of withstanding a high temperature bakeout cycle. A family of very stable magnetic materials has been developed[3] which should meet the requirements of most of the linac. Properly constructed magnets could provide field gradients up to perhaps 5 kG/cm. Calculations for an injection energy of 2 MeV, a frequency of 400 MHz, and a ++-- magnet configuration indicate that the first quadrupole may require a gradient of about 7 kG/cm; consequently the first few focusing elements may have to be electromagnets but the majority could be permanent magnets.

Side-Coupled Waveguide Structure

Proton linacs for producing pions require beam energies of at least 500 MeV. At an energy between 150 and 200 MeV the shunt impedance of the drift tube type accelerator structure becomes sufficiently low that a transition is indicated to the side-coupled waveguide structure which combines high shunt impedance with a wide tolerance of tuning errors. Transition energies near 200 MeV simplify the matching operation but will result in somewhat less overall efficiency. To avoid acceptance limitations the side-coupled accelerator frequency should not be more than 1200 MHz; the effective shunt impedance will be about 50 MΩ/m.

The mechanical design of the LAMPF waveguide structure requires several sophisticated machining and brazing processes.[4] Other side-coupled geometries which might be cheaper to fabricate yet equivalent in performance have been suggested.[5,1] Forging and coining techniques may be utilized, and Cu plated parts might be employed. The area of structure fabrication needs considerable research and development.

It may prove advantageous to insert a 180° bend in the linac at the transition region to shorten the effective length.[6] The cost and effect on beam of an achromatic and isochronous 180° bending system needs to be evaluated in terms of the advantages of a shorter facility.

Tank Coupling

A scheme of accelerator tank coupling which may substantially increase system reliability has been proposed.[7] The entire side-coupled linac is coupled together with bridge couplers and driven at discrete points by high power rf amplifiers. This technique may result in reduced phase and amplitude sensitivity of the system while removing the requirement of 100% rf system reliability for acceleration operation. These advantages would be very important for a practical medical linac.

Power Sources

Experience at LAMPF and elsewhere indicates that the klystron power source is very suitable for applications in which several tank systems have to be phased together and operated under stringent amplitude control requirements. Klystrons are available at 400 MHz which produce peak power of 20 MW and average power of 300 kW. Three such tubes could provide power for the 200 MeV drift tube linac producing 100 μA average current at a duty factor of 5×10^{-3}. A reasonable pulse length is 80 μs (60 pps) if the fill time can be decreased to 20μs. Seven klystrons operating at 1200 MHz producing peak and average powers of 10 MW and 50 kW respectively would be suitable for the side-coupled structure section.

Control System

Reliability, safety, and simplicity of operation are essential features for accelerators intended for medical use and must be incorporated into the control system design. Regulating systems must be optimized to handle the expected parameter variations and operating ranges with a minimum of operator manipulation. Closed-loop regulation of such system parameters as field level and phase in multiple amplifier systems need to be developed to provide consistency of beam quality, decreased sensitivity to the effects of component drift and aging, and simplified operation. Much of the control system and fast-acting protective circuitry developed for modern accelerators such as LAMPF can serve as prototypes for a medical linac.

Beam Channels

In the case of proton and neutron beams the channel design is relatively straightforward with shielding being the major complication. Primary proton beams should be sufficiently intense to enable the linac to serve several treatment areas. A portion of the proton beam could also be delivered to a radiography station. The beam stop area can be used for radioisotope production.

Table 2

Medical Proton Linac for Producing Pions

Energy	500 MeV
Average current	100 μA
Peak current	20 ma
Pulse length	50-100 μs
Duty factor	0.005
Drift Tube Section	
Frequency	400 MHz
Injection energy	~ 1 MeV
Length	40 m
No. tanks and rf systems	3
No. drift tubes	158
Focusing	permanent and electro-magnet quads.
Tank rf power (peak)	45 MW
Beam rf power (peak)	4 MW
Total rf power (peak)	50 MW
Accelerator gradient	5 MeV/m
Side-Coupled Section	
Frequency	1200 MHz
Transition energy	200 MeV
Length	40 m
No. tanks and rf systems	7
Focusing	doublets
Tank rf power (peak)	62 MW
Beam rf power (peak)	6 MW
Total rf power (peak)	70 MW
Accelerator gradient	7.5 MeV/m

A channel with very large acceptance is required for the pion producing linac in order to keep primary beam currents sufficiently low to simplify control system requirements, reduce facility cost, and avoid serious induced radioactivity problems. Conventional magnetic channels of the order of 60×10^{-3} sr imply primary pion currents of several hundred μA in order to produce desirable radiotherapy beams of pions. A 1 sr acceptance channel (e.g., the superconducting channel developed at Stanford,[8] if practical) could be utilized with a primary proton beam of about 30 μA. Another possibility is a pulsed lens system since the linac has a low duty cycle.

The pion-producing proton linac could use two production targets to serve four treatment areas. Again, the beam stop area would serve as a radioisotope production area of great potential. It may also prove possible to divert a small fraction of the 200-MeV proton beam at the end of the drift tube structure to provide the facility with proton therapy and proton radiography capability.

Parameters of a Proton Linac for Producing Pions

The parameters for a practical medical proton linac for producing radiotherapy pion beams are indicated in Table 2. A proton current of 100 μA is assumed which would require a pion channel acceptance of at least 0.2 sr yielding about 10^9 pions s^{-1}.

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