BOOSTER LINEAR ACCELERATORS FOR PROTON THERAPY

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Abstract

Radiotherapy using proton beams of energies of order 200MeV is now accepted as a feasible cancer treatment technique whose widespread use has so far been limited by the high costs of proposed facilities. AEA Technology have identified a low-cost solution using a linear accelerator to boost existing hospital cyclotrons. The present project status and the design of the booster linear accelerator are discussed.

Introduction

Approximately 1 million new cancer patients worldwide are referred annually for conventional radiotherapy using X-rays, but a major limiting factor is the potential for damage to surrounding healthy tissue for tumours located close to critical organs or radio-sensitive body tissues. A significant advance, originally used as early as 1955, is to use a beam of energetic (~ 200MeV) protons to provide the radiation dose. Protons deliver the dose at a relatively uniform low level until they have lost a significant fraction of their energy, at which point the dose increases reaching a sharp peak close to the end of the proton range. The exact position of this Bragg peak in dose can be controlled by steering or collimating the proton beam for transverse movement and by modifying the impact proton energy for longitudinal movement. This allows fine control of the dose distribution and significantly reduces the potential for damage to surrounding tissues. Over 10000 patients have now been treated by this method, and for a number of tumours dramatic improvements in success rates have been clearly demonstrated. Despite the obvious advantages inherent in treating tumours with less damage to surrounding tissue, proton therapy has only been adopted on a clinical basis at a very small number of centres worldwide due to the very high cost (typically > £10M) required for a dedicated high energy clinical facility. AEA Technology propose to investigate the novel idea of using **linear accelerators** to boost the energy of existing medical cyclotrons (of which there are an estimated 24 suitable devices throughout the world). This will provide the required high energy proton therapy facilities at a suitably reduced cost.

Accelerator Options

The requirement for high energy proton therapy is to construct a hospital-based accelerator capable of providing a 200MeV H^+ beam of 10-30nA averaged over a 30-60 second treatment time. Variation in beam energy using the accelerator is an advantage, but energy degradation using foils is possible for fixed energy machines. Table 1 summarises the status of possible accelerator options for high energy proton facilities, from which it can be seen that no option is fully satisfactory, and all options suffer from high cost for installation into hospitals.

TABLE 1
Proton Therapy Accelerator Options

Option	Status
Cyclotron	• Compact • Technology well proven • High operating power • Energy not variable • High cost
Superconducting Cyclotron	• Low operating power • Technology not proven • Long down times • High cost
Synchrotron	• Compact • Medium operating power • Energy variable • Proven designs - too low current • High current design not proven • High cost
Linac	• Technology well proven • Medium average power • High current/low emittance • Long length • Energy step variable • High cost

AEA Technology have proposed the novel solution of adding a booster linac to an existing hospital low energy (62.5MeV) cyclotron used for low energy proton therapy. The resultant 200MeV accelerator is very cost-effective as the first third of the acceleration is already in existence, together with the considerable hospital infrastructure required for proton therapy. Present (rough) estimates suggest a cost of order £5M ($10M) for production of a 200MeV two room treatment facility equipped with static horizontal and vertical beams.

Proposed Booster Linear Accelerator

AEA Technology have formed a European partnership to design, fabricate and test a booster linac at Clatterbridge Hospital (Department of Oncology) in the UK. Figure 1 shows a schematic of the proposed device.

A 1.28 GHz SCCL will bunch and accelerate the beam from an existing Scanditronix MC60 cyclotron that will be modified to produce a 100μA pulsed (approx 10μsec) beam. Table 2 shows the input and output specifications for the linac.

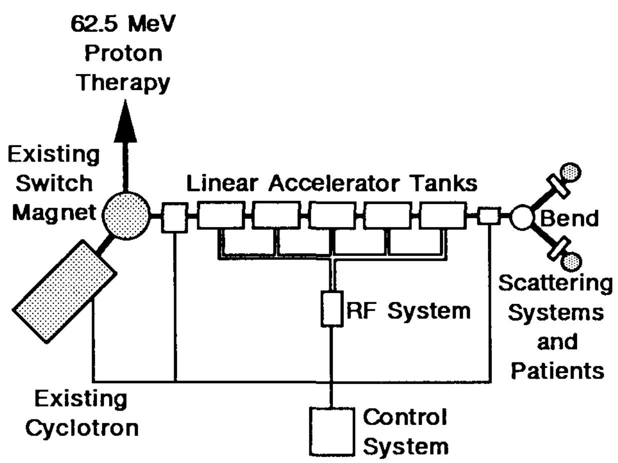

Fig.1 Schematic of the Proposed Clatterbridge 200 MeV Proton Therapy facility

TABLE 2
Clatterbridge Booster Linac Specification

Input Beam	Energy Current Emittance (4RMS UN)	62.5MeV 100μA 9.3 πmm mrad
Output Beam	Energy Average Current (per RF pulse) Average Current (per Treatment)	200MeV 25μA 25nA
Accelerator	Type Frequency Capture Efficiency Synchronous Phase Kilpatrick Factor Duty Cycle	SCCL 1.284GHz 20% -30° 1.35 0.1%

The specification shown applies to the initial baseline design used to demonstrate the feasibility of the proposed application. The main aim of this initial design is to verify whether such a booster linac can meet the design objectives of total length and peak RF power of less than 20 metres and 20MW (ie. 20kW average) respectively with the aim of using a single 20MW klystron.

Given the low beam current produced by the cyclotron, space charge effects for the linac will be negligible. The first key issue for the linac design is therefore the high input emittance. Modelling at Culham therefore began using the approach adopted by Wangler [1] to examine the periodic focusing structure. Assuming a FODO structure of acceptance twice the input emittance and transverse phase advance of 70° and using a Fourier representation of the focusing lattice the maximum beam radius was calculated as a function of the number of SCCL cells per tank for 62.5, 100, 150 and 200MeV beam energy. For this initial linac physics design it is assumed that minimisation of the cavity length and power is the main design objective, and so a low linac bore radius of 10mm has been adopted. The lattice analysis therefore shows that an average of 22 cells will be allowable per tank. (During the detailed design phase, once the costs of RF splitters and bridge couplers are available, the bore radius will be further reduced where possible to improve the shunt impedance).

SUPERFISH modelling was then carried out on the MJN 33 MHz PC-compatible computer at Culham to model 62.5, 100, 150 and 200MeV cavities.

TABLE 3
SUPERFISH Results

	62.5 MeV	100 MeV	150 MeV	200 MeV
g/L	0.381	0.413	0.429	0.441
Q	13790	16311	18250	19462
T	0.810	0.832	0.848	0.855
Z(MΩ/m)	51.1	62.9	71.8	77.2
ZT^2(MΩ/m)	33.5	43.5	51.6	56.4
E_{max}/E_o	4.00	3.95	4.02	4.02
1/2 cavity Power (W) (E_o = 1MV/m)	397.2	397.5	412.0	428.0

Table 3 shows the resultant values for the gap to cell length ratio, cavity Q, transit time factor, shunt impedance, field ratio and half-cavity power. The initial design is based upon that of Chang et al [2] with an 85mm tank radius, 20° nose angle and 2.5mm/5mm nose radii (see Fig.2). At each energy the gap length was varied to achieve the desired resonant frequency.

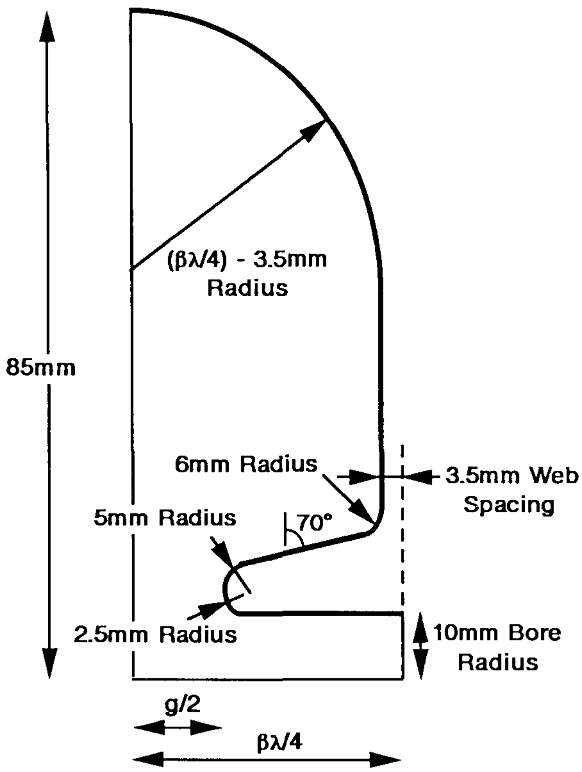

Fig. 2 Generic SCCL Cavity Geometry

Using polynomial fits to the derived values for focusing lattice parameters, half cavity power, E_o and T, the linac parameters shown in Table 4 have been derived assuming 15% power loss in the side cells and the bridge couplers and (3βλ/2) spacing between tanks.

TABLE 4
Linac Parameters

Total Energy Gain	137.5MeV
Linac Power (SUPERFISH)	30.8MW
Linac Length (SUPERFISH)	17.5m
No of Cells	323
No of Tanks	16
Approx Total Intertank Length	2.4m
Approx Side Cell/ Bridge Coupler Power	4.6MW
TOTAL LENGTH	19.9m
TOTAL POWER	35.4MW

The length is slightly lower than the 20m limit allowed, whilst the power is much higher than the 20MW target. The use of two 20MW klystrons will therefore be required.

Conclusions

The analysis proves that it is feasible to design a linac capable of producing an output beam of the properties required for proton therapy using a cyclotron as a pre-accelerator. The resultant linac is 20m long and will require an RF power of 35.4MW (assuming power losses in the side cells and bridge couplers of 15%).

The design proposed, however, is only an initial feasibility study. Further physics design is now required to:

(a) further optimise the cell geometry of Figure 2 to increase shunt impedance at low energies.

(b) examine phase ramping and/or pre-bunching as a means of increasing the capture efficiency. This will lower activation levels in the linac and lower the required duty cycle, but increase both the length and the power consumption.

(c) perform the trade-off between bore radius and number of tanks.

(d) look at switching off individual tanks as a means of varying the output energy.

The initial design feasibility study is now largely complete. UK Government funding is now awaited to lead to a prototype test at Clatterbridge Hospital in late 1995.

References

[1] T P Wangler "Space Charge Limits in Linear Accelerators", Los Alamos note LA-8388 (1980).

[2] C R Chang, R Bhandari, W Funk, D Raparia and J Watson "Design Studies of SSC Coupled Cavity Linac", IEEE Particle Accelerator Conference p.2993 (1991)