DESIGN OF X-BAND MEDICAL LINEAR ACCELERATOR WITH MULTIPLE RF FEEDS AND RF PHASE FOCUSING*

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Abstract

A design of 6 MeV X-band 9.3 GHz medical linear accelerator is presented. It is composed of four separate clusters of accelerating cavities, where a coherent RF excitation is provided separately to each cluster. The use of multiple accelerating sections with multiple RF feeds permits the use of inexpensive RF sources. The first cluster is Alternate Phase Focusing (APF) RF cavities, providing radial and longitudinal beam focusing without the use of heavy and bulky magnets or solenoids. The three other clusters used for acceleration are composed of multiple standing wave sections operating in the Pi-mode. Each section has been designed and optimized for high shunt impedance by means of 2D SUPERFISH code and 3D CST code. A two dimensional code, named PTCC, was developed to facilitate design and analysis of the different parts of the accelerating structure.

INTRODUCTION

There are many advantages of using X-band accelerating structures which include small size, light weight, higher shunt impedance, short fill time, higher accelerating gradient and higher breakdown level. So X-band accelerating structure has the broad application for generation of high power beams for cancer therapy, material processing and sterilization [1], Schonberg.

Our Design of X-band linac aims to produce a 6 MeV electron beam needed to generate high energy X-ray for the purpose of radiotherapy, see linac output parameters in Table 1. Linac consists of four cavity modules: a buncher and three accelerating sections Fig. 1. These modules are fed with separate but coherent RF sources. The electron gun generates a 10 KeV DC electron beam to be then bunched and accelerated by first cavity module which is 8 cells buncher module. Then the bunches formed are accelerated to 6 MeV energy by three identical accelerating cavity modules each is a 7 cells module operating in π mode. The π mode has been chosen as the operating mode since it provides the highest shunt impedance and so it gives the highest energy gain for a given power.

BUNCHER DESIGN

Buncher is responsible for forming bunches from the DC gun beam, and also accelerating those formed bunches to relativistic energy 1 MeV ($\beta \simeq 1$). At the same time, the buncher combats the defocusing effect due to space charge, which is more pronounced at low beam energy.

Table 1: Linac Main Parameters

Parameter	Value
Electron beam energy (MeV)	6.1
Beam Spot size (mm)	< 3.75
Input beam current (mA)	15
Input beam energy (KeV)	10
Total Length (cm)	48.2
Operating frequency (GHz)	9.3
Operating Mode	π

In order to design the buncher, a one dimensional program is developed to track the on-axis particles (position, momentum and energy) for different phases [3] to maximize average energy and number of output particles at end of buncher section. Figure 2 shows particles behavior at different RF phase in terms of whether they reach the end of the linac and with how much energy.

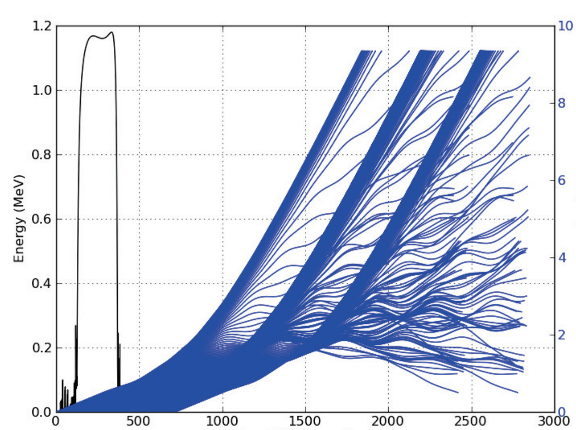

Figure 2: Phase plot and energy of bunch from buncher simulation using buncher 1D tool.

To minimize the size and weight of linac structure while maintaining the required beam spot size beside an acceptable level of beam transmission, RF fields are used for radial focusing instead of the external magnetic focusing devices, thereby reducing the weight of structure significantly. This method of focusing beam is called Alternate Phase Focusing (APF) [4].

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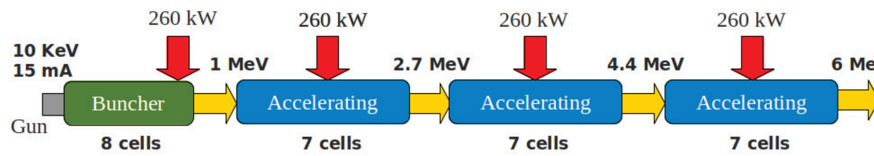

Figure 1: Schematic layout of the X-Band Medical Lianc.

Alternate Phase Focusing

Using RF phase focusing, the lengths of buncher cells have been modified to introduce focusing/bunching at specific cells of buncher. By increasing cavity length than standard one (the length at which particle reach end of cell at phase exactly π), bunch will reach the center of next cell ahead of crest (+ve synchronous phase) seeing positive radial field causing radial focusing effect. On the other hand by decreasing cavity length than standard one, bunch will reach the center of next cell behind of crest (-ve synchronous phase) spending much time in longitudinal bunching region (negative slope longitudinal field) giving longitudinal bunching effect. This can be shown graphically in Fig. 3 and Fig. 4.

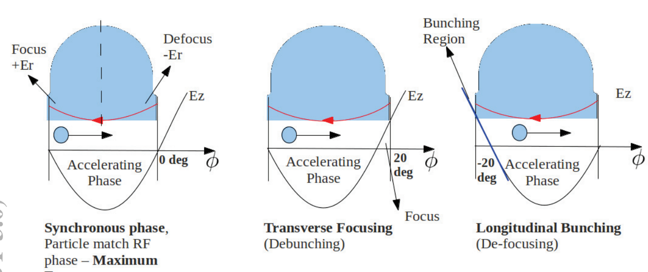

Figure 3: Different bunch phase relative to crest effect.

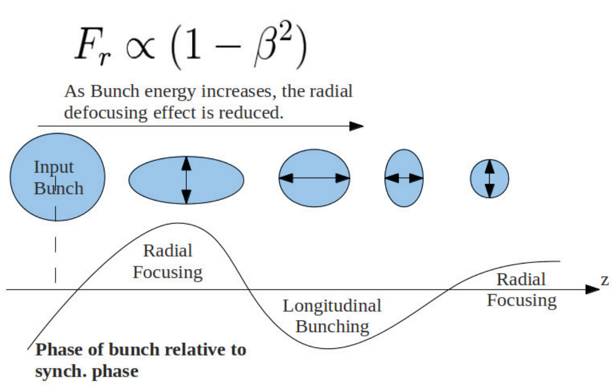

Figure 4: Alternate Phase Focusing.

Figure 5 shows linac buncher design using APF to achieve a beam spot size at output of buncher less than 3.75 mm. The length of each cavity has been optimized to have focusing effect in the first few cells, then bunching effect in next cells. The radius of each cavity is modified to have resonant frequency of each cell equal operating frequency of 9.3 GHz.

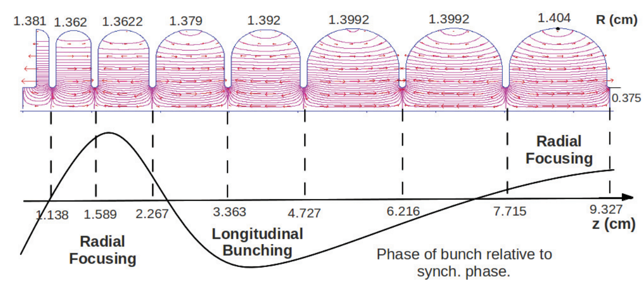

Figure 5: Buncher dimensions to achieve phase focusing.

ACCELERATING SECTION

Accelerating section is composed of 7 identical cells each with $\lambda/2$ length. Drift tubes separate between cavity clusters to eliminate coupling between accelerating sections (independent sections fed by synchronous RF sources). The single cell dimensions (Fig. 6) have been optimized to achieve maximum shunt impedance using a 2D electromagnetic solver code SUPERFISH.

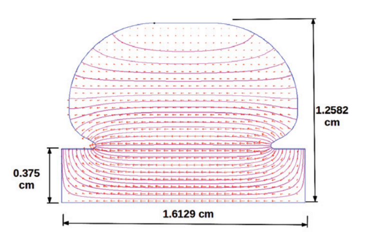

Figure 6: Accelerating Cell Cavity Shape.

Field Flatness in Accelerating Section

In accelerating structure with identical cells, field level at cells can not be equal (called field flatness) due to field leakage in end cells drift tubes connected to end cells. Field flatness can be achieved by slightly reducing end cells radius, called tuning [5]. Field flatness has the advantage of increasing the shunt impedance of the cavity section compared to the unflate untuned section. Figure 7 shows the field along accelerating section before and after field flatness tuning of end cells.

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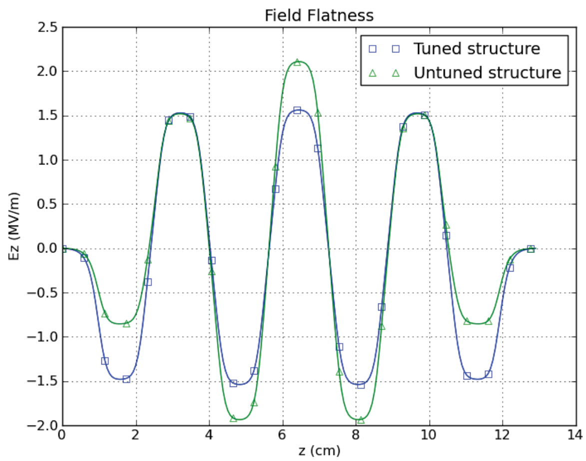

Figure 7: Field flatness after and before tuning.

POWER COUPLER

The RF power is coupled to the accelerator structure by means of magnetic coupling through aperture (or slot) applied to the cylindrical wall of a cavity center cell. The magnetic field of the waveguide TE_{01} propagating mode couples the circular flux lines of the input cavity magnetic field. The dimension of the coupler window (w) and the central cell radius (R_c) (Fig. 9) have been tuned in order to obtain simultaneously a coupling coefficient $\beta \simeq 1$ (zero reflection $S_{11} = 0$ at operating frequency) and resonant frequency of the whole system (cells+coupler) equal to 9.3 GHz. This matching process has been done using the 3D electromagnetic simulation code CST Microwave Studio.

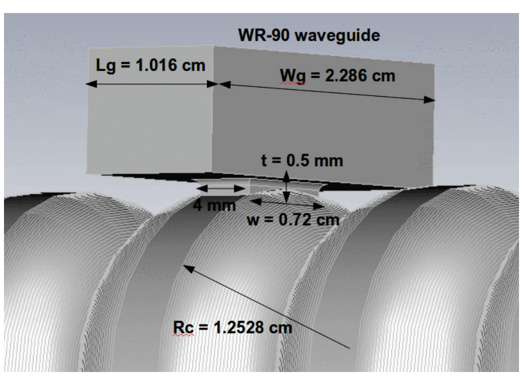

Figure 8: Power coupler through rectangular waveguide.

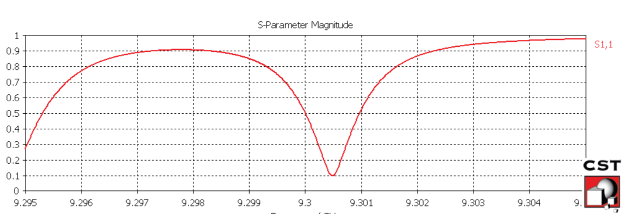

Figure 9: Reflection Coefficient - S_{11}.

BEAM DYNAMICS

A two dimensional particle tracking code called PTCC [3] is developed to simulate the dynamics of the beam bunches in different sections. Figure 10 shows three bunches at the output of Linac from beam dynamics simulation. As can be seen the spot size at the end of Linac is about 3 mm. The analysis is for accelerator design without using magnetic focusing.

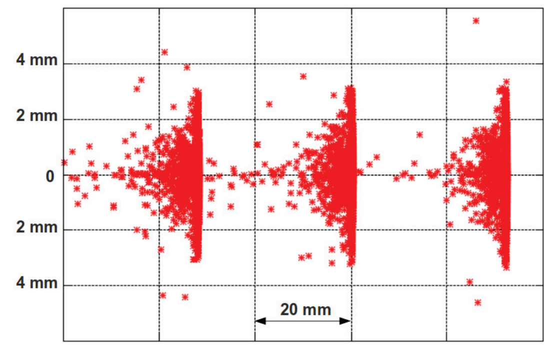

Figure 10: 6 MeV output bunches of Linac.

CONCLUSION

A complete design of an X-band Linac for use in medical radiotherapy is presented. The design is composed of four sections; the first is a buncher and the other three are standing RF accelerating sections excited with the π-mode. The buncher is an alternate phase focusing, that achieve both radial and longitudinal focusing. To facilitate the buncher design, a 2D code is developed. The overall behavior of the system is assessed using a developed 2D simulation code named PTCC.

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