PROTON SYNCHROTRON FOR CANCER THERAPY AND ITS BREATH-SYNCHRONIZED OPERATION

S. Fukumoto, H. Tsujii, T. Inada, Y. Hayakawa, A. Maruhashi, Y. Takada, J. Tada, H. Tsuji, K. Endo*, Y. Irie*, I. Yamane*, K. Muto*, and T. Shintomi*

Proton Medical Research Center, University of Tsukuba
Tsukuba-shi, Ibaraki-ken, 305, Japan
*National Laboratory for high Energy Physics
Tsukuba-shi, Ibaraki-ken, 305, Japan

Abstract

Clinical trial of proton therapy is being carried out with high energy beams from the KEK accelerator complex. Since the results are so promising that a dedicated proton therapy facility is being designed. Its accelerator is a 230 MeV separated-function and slow-cycling synchrotron with a 10 MeV linac as its injector. Its layout is described and breath-synchronized operation is discussed.

Introduction

One of the most important factors of the radiation therapy is to give sufficient dose in tumor keeping dose provided in normal tissue as little as possible. R. Wilson pointed out high energy proton beams can be a good modality because of their dose distribution characteristics in water[1]. Investigation in Sweden showed that the relative biological effectiveness (RBE) of proton beams is almost equal to that of photons and electrons. Eye melanoma has been treated successfully at Massachusetts General Hospital (MGH) in Boston. Clinical trial of proton therapy started in July of 1983 at Tsukuba using 500 MeV protons of KEK 12 GeV proton synchrotron complex. Because of high energy advantage, deep-seated tumors such as liver, lung and esophagus have bean treated with much better results compared with conventional radiation therapy. To extend the study further and to treat patients of different situations, a dedicated proton therapy facility is being designed and its specifications are being discussed. It will be built next to the conventional radiation therapy facility at the University Hospital. Basic design of the facility was made[2,3,4]. Its has two treatment rooms, Room No.1 is equipped with a horizontal beam and two vertical beams, one from the upper direction and the other from the lower direction. Room No.2 is equipped with two vertical beams but no horizontal beam.

Fermilab designed and made a dedicated 250 MeV proton synchrotron for Loma Linda University Medical Center in California[5]. Patient treatment started there in fall of 1990. The 200 MeV Orsay synchrocyclotron is now fully dedicated to cancer therapy. At ITEP in Moscow, protons of about 200 MeV are fast extracted from the 10 GeV synchrotron and used for therapy. Proton therapy has begun recently at Clatterbridge in UK, Louvain-la-Neuve in Belgium and Nice in France. Two dedicated facilities are being designed in the U.S.A.. One is at MGH and the other is at UC Davis Cancer Center in Sacramento, California. Among heavy charged particles, protons are being recognized as the practical modality in the next generation.

Tsukuba dedicated proton therapy facility plan

According to the experiences at Proton Medical

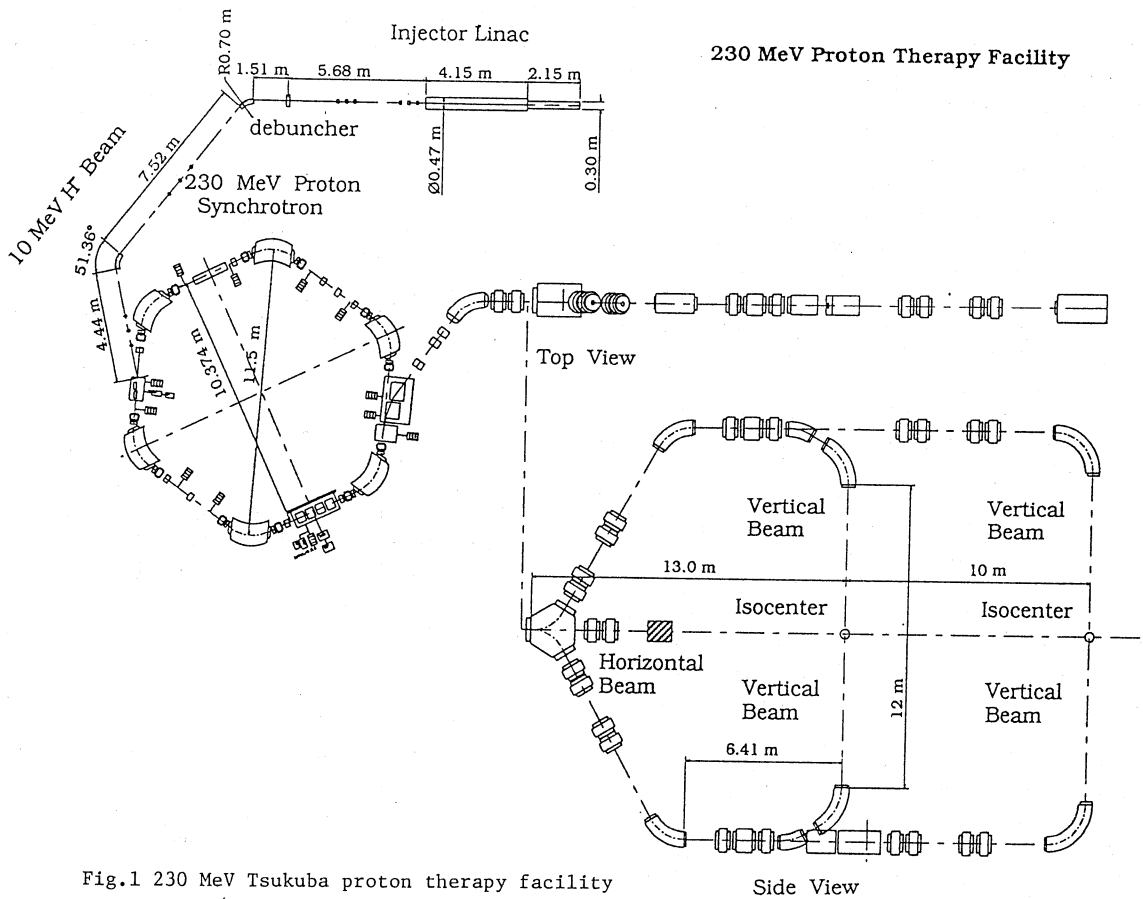


Fig.1 230 MeV Tsukuba proton therapy facility

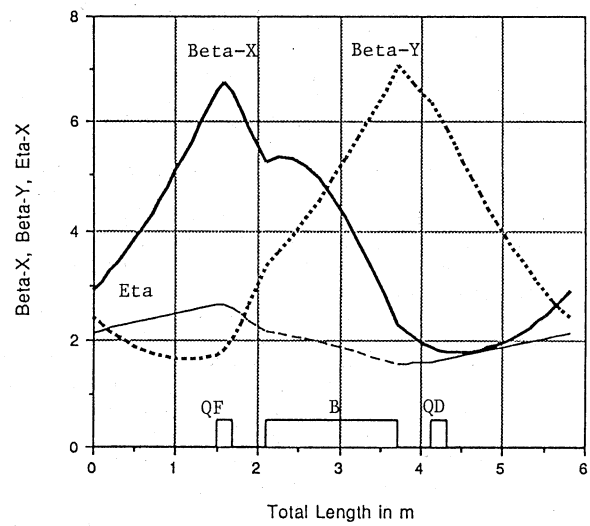
Research Center (PMRC), University of Tsukuba, the design goal of the energy and intensity is fixed to 230 MeV and 20 nA. The range of the 230 MeV protons is 32.7 cm in water and it seems sufficient for most Japanese patients. The beam intensity specification depends on the beam delivery system and is even sufficient for a passive beam spreading system with double scatters. Irradiation time is around 2 minutes with this intensity.

Synchrotron, linear accelerator[6] and cyclotron[7,8,9] can fulfill the specifications above mentioned. Although the synchrotron is not so simple in structure than the cyclotron and it needs more sophisticated beam handling than the linear accelerator, it is ready for construction without thorough R&D at the present. Thus a 230 MeV separated-function and slow-cycling synchrotron is designed. It has six superperiods and its main parameters and layout with beam transport lines are shown in Table 1 and Fig.1. It has 3 m long straight sections. Parameters of a synchrotron with straight section of 2 m long are also shown in Table 1. Beta function and dispersion function of both synchrotrons show no big difference between them. This means that the design of the bigger synchrotron can be easily modified to that of the smaller one.

Table 1 Main parameters of 230 MeV proton synchrotron for therapy

Lattice	
Circumference	34.939 m (28.939)*
Structure	DOFB
Superperiod	6
Long Straight Section	3.0 m (2.0)
Bending Radius	1.55 m
Bending Magnet Length	1.623 m
Tune(H)	1.80
Tune(V)	1.85
Beta Function(H)	1.752-6.740 (1.249-6.211)
Beta Function(V)	1.630-7.078 (1.149-5.958)
Maximum Dispersion	2.64 (2.31)
Natural Chromaticity(H)	-0.2135 (-0.3458)
Natural Chromaticity(V)	-2.6175 (-2.365)
Gamma at Transition	1.722 (1.746)
Bending Magnet	
Injection(10 MeV)	0.296 T
Extraction(230 MeV)	1.498 T
Deflection Angle	60 deg.
Edge Angle	30 deg.
Gap	6.5 cm
Width	28 cm
Quadrupole Magnet	
Aperture	11.6 cm
Length	20 cm
Field Strength(H)	2.6172 m^{-2} (3.3459)
Field Strength(V)	-0.6886 m^{-2} (-1.5026)
Radiofrequency Acceleration (10-230 MeV)	
Frequency Range	1.24-5.11 MHz (1.50-6.17)
Voltage	450-300 V
Stable Phase	20-30 deg
Repetition Rate	0.5-1 Hz
Injector: 425 MHz Linac	
Input Energy	30 KeV
Output Energy	10 KeV
RMS Normalized Emittance	0.01 π cm.mrad
Ion Source: Multicusp H^-	

* ( ) is the parameters of 2 m long straight section design.



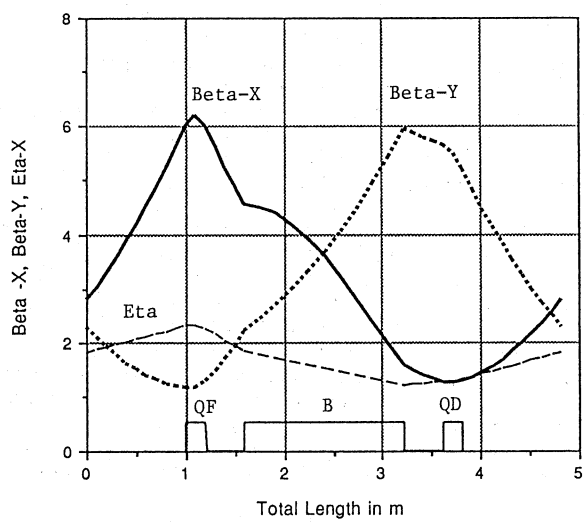


Fig.2 Beta and Eta function for long straight section of 3 m(upper) and 2 m(lower).

Since 500 MeV protons are extracted fast from the KEK booster, several ten nS beams are used at RMRC. The fast extraction system is simpler in operation and tuning than the slow extraction system although it requires special kickers. The new facility will be operated in the fast extraction mode at least just after its completion. The short beam bursts produce shock waves in tissue. The waves are expected to be used to detect Bragg peak position during irradiation[10]. Proton beams are broaden laterally to irradiate a big tumor by Coulomb multiple scattering in double scatterers and their momentum is spread by a ridge filter to make a spread-out Bragg peak. This passive system needs a slit and a bolus which taylor the beam to match the tumor volume. The passive beam broadening system is simple, reliable and not dependent on beam time structure, but the bolus and the slit must be made for each patient, moreover, most of protons are lost by hitting the slit.

Another beam manipulation system is a three dimensional (3D) scanning system. A narrow beam is deflected laterally in two directions and its energy is adjusted to irradiate the tumor volume. Many computer simulations have been made to develop a 3D system for a fixed target. Some organs, such as lung and liver, however, moves with breath during irradiation. Since it is more difficult to develop a 3D scanning system for a sometimes irregularly

moving target than for a fixed target, we will try to find a different approach based on the present status at PMRC.

Breath-synchronized operation

There are several possibilities to irradiate a moving target;
a) to deflect the beam and adjust its energy following the target,
b) to move the bed to fix the target spatially,
c) to extract the beam from the synchrotron which is keeping the beam in storage mode when the target is in some assigned position,
d) to start acceleration by a trigger from the target and irradiate with fast extracted beams.

a) and b) do not affect irradiation time, but might be not reliable because patient breath is not completely regular. c) is straight forward, but both irradiation time and magnet power loss of the synchrotron increase. Although d) requires modification of previous design, it seems worthwhile to examine in detail.

The patient breath period is usually ranging from 3 to 5 seconds. The beam intensity of 20 nA is to be attained by 1 Hz repetition in the original design. If this synchrotron accelerates protons not with the 1 Hz repetition but synchronized to the patient breath, the irradiation extends more than three times longer. A rapid cycling synchrotron such as the KEK booster may improve the situation, because the organ is stationary more than half of the breath period. Another option is a slow cycling synchrotron with a repetition of 0.3 Hz. In this case there is no big difference in irradiation time for with and without breath-synchronization. Number of protons per pulse must be three times larger than the original design. If an equal emittance is assumed in the horizontal and vertical planes, the emittance and the space-charge limited current at 10 MeV relates as shown in Fig.2. If the same COD contribution to the aperture is assumed, the bending magnet gap increases from 6.5 cm to 8.5 cm due to the vertical emittance growth for H^- charge-exchange injection. Bending magnet current and voltage are estimated as shown in Table 2. Although the magnets and their exciting current increase, the maximum voltage reduces to about a half of the original design. Thus the initial cost of the accelerator seems not seriously affected by the modification.

Conclusion

A dedicated 230 MeV proton synchrotron is designed for a new therapy facility at Tsukuba. It is a separated-function and slow cycling synchrotron with a 10 MeV linac as its injector. To irradiate effectively a tumor in an organ such as liver or lung which moves with patient breath, change of the repetition rate from 1 Hz to 0.3 Hz is examined. The irradiation time of the smaller repetition rate machine has not big difference between for the fixed target and for the moving target synchronized with patient breath. Although its beam intensity per pulse must be three times higher than that of the larger repetition rate machine, the initial cost of the both accelerators seems not seriously different.

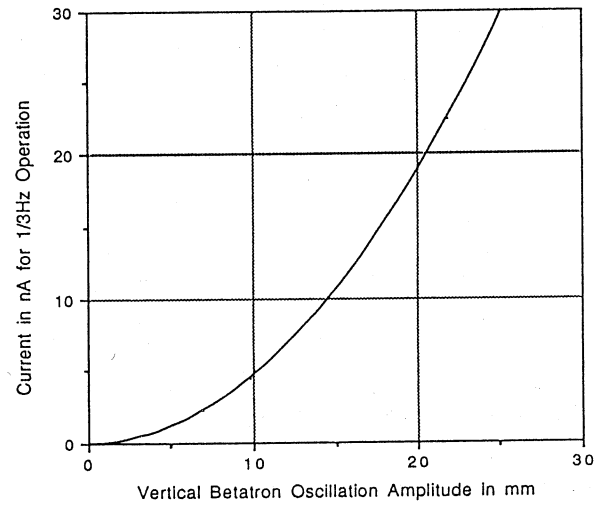


Fig.3 Vertical betatron oscillation amplitude vs. space charge limited current for 0.3Hz operation.

References

1. R. Wilson, Radiology, 47, 487, 1946.
2. K. Endo, S. Fukumoto, K. Muto, T. Kitagawa, T. Inada, Y. Takada, A. Maruhashi, Y. Hayakawa, T. Arimoto, J. Tada, M. Sato, Medical European Particle Accelerator Conference, 1459 (1989).
3. S. Fukumoto, K. Endo, K. Muto, M. Akisada, T. Inada, H. Tsujii, A. Maruhashi, Y. Takada, Particle Accelerators, Vol. 33, 153 (1990).
4. S. Fukumoto, K. Endo, K. Muto, M. Akisada, T. Kitagawa, T. Inada, H. Tsujii, A. Maruhashi, Y. Hayakawa, Y. Takada, J. Tada, Proc. International Heavy Particle Workshop, Paul Scherrer Institute, 70 (1990).
5. Proton Therapy Facility Engineering Design Report, Fermi National Accelerator Laboratory, February, 1987.
6. R. W. Hamm, K. R. Crandall, J. M. Potter, Preliminary Design of a Dedicated Proton Therapy Linac, to be published in Proc. 1991 IEEE Particle Accelerator Conference.
7. Y. Jongen, 12th PTCOG Meeting, May, 1990.
8. H. Blosser, Nucl. Instr. Meth. in Physics Research, B40/41 (1989), 1326.
9. T. Takayama, KURRI-TR-342 (1991), Kyoto University.
10. J. Tada, Y. Hayakawa, K. Hosono, T. Inada, Properties of Acoustic Signals Generated by Pulsed Proton Beam Irradiation in Water and in Soft Tissue, to be published in Medical Physics.

Table 2 Maximum exciting current and voltage of bending magnets for 1 Hz (upper line) and 0.3 Hz (lower line) repetition rate.

Rise time (sec)	Bend. gap (cm)	Max. curr. (A)	Inductance (mH)	Max. voltage (V) Ldi/dt	Ri	Total
0.5	6.5	2,550	87.0	382	130	512
1.5	8.5	3,335	67.0	97.4	170	267.4