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DESIGN OF A COMPACT PROTON ACCELERATOR FACILITY DEDICATED FOR CANCER THERAPY

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Abstract

A compact proton accelerator facility dedicated for cancer therapy is designed. A combined function synchrotron with the maximum energy of 250 MeV is adopted. A ferrite-loaded untuned cavity with the frequency range of 1.6-8 MHz is utilized for beam acceleration. Combination of a horizontal beam line and a gantry in the same treatment room will realize multi-port irradiation. The beam efficiency will be improved by positioning outside the treatment room even with a single treatment room. Scanning method for beam spreading also increases beam utilization efficiency.

1. Introduction

Recently it becomes to be known that cancer therapy with use of charged particles as proton and heavy ions is very promising from the view-point of quality of life of the patient due to the merit of no erasion. As the national center for cancer therapy with heavy ions, HIMAC has been constructed at National Institute of Radiological Sciences and clinical treatments have already been started with good results[1]. However, the number of patients who need such treatments is overwhelmingly large compared with the curable patients at such a limited facility and it is inevitable to realize a proton therapy facility of such a size as can be operated with an areal-center hospital of somewhat larger size. For this purpose, a proton accelerator facility entirely dedicated for cancer therapy has been designed. Because of the merit of variability of the output beam energy, proton synchrotron is adopted. In order to avoid the complex control of magnet power supplies, a combined function lattice is adopted[2]. As the RF accelerating cavity, a ferrite loaded untuned cavity with the frequency range of 1.6~8 MHz is adopted for the merit of easy operation[3]. For the medically dedicated machine, it should be operated with minimum possible down time without any professionals of accelerator. From this point of view, the RF power source of the injector linac is to be optimized.

2. Total Scheme

As the main accelerator, a proton synchrotron as shown in Fig. 1 is adopted because of the merit of energy variability[2]. The charged particle therapy is considered to orient the pin-point irradiation to the volume spread in three dimensions. In this case, the depth of the irradiation is to be controlled by the beam energy and then variable energy accelerator is inevitable. The pencil beam is to be scanned in two directions in the transverse phase spaces. In order to realize flat dose distribution by this scanning method, good time structure of the beam is required.

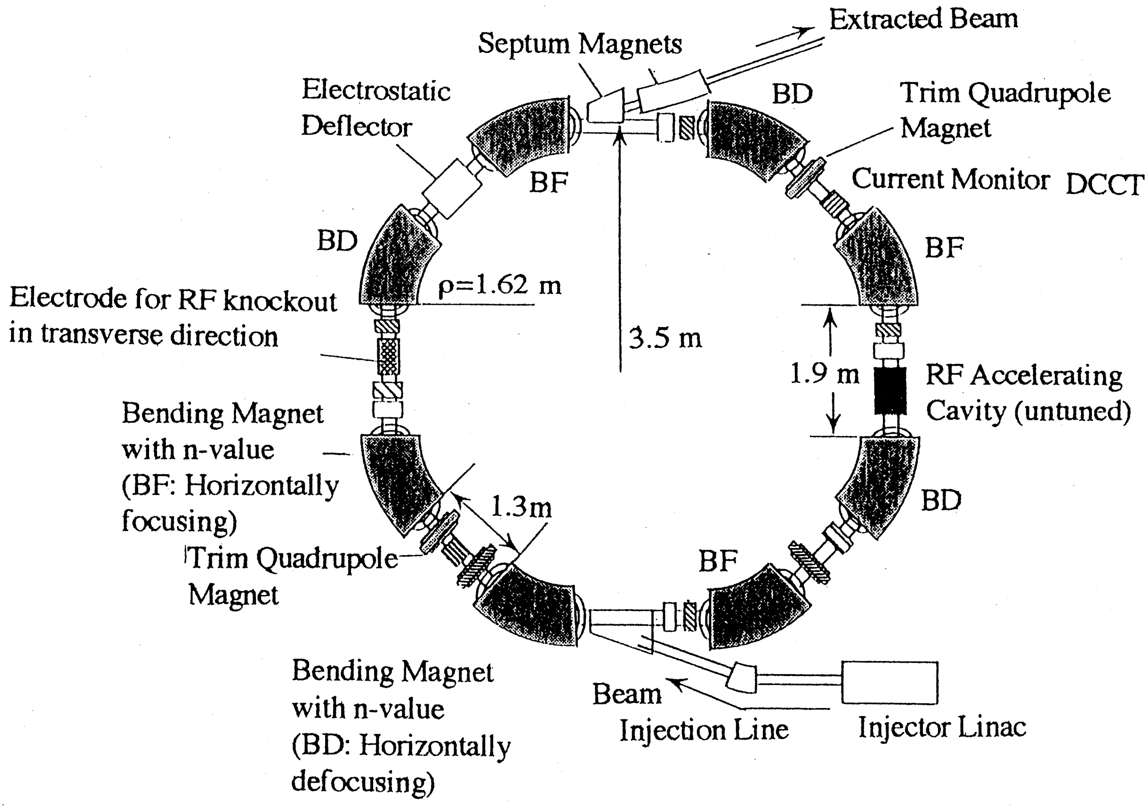

Fig 1. Layout of the Proton Accelerator System.

Table 1 Main Parameters of the Proton Therapy Facility

Particle	Proton
Beam Energy	70-250 MeV
Repetition Rate	0.5 Hz
Dose Rate	5 Gy per Min.(Max.)
Number of Treatment Room	1
Beam Ports	Horizontal and Rotatable
Method of Beam Spreading	Scanning

For the efficient use of the beam, positioning of the next patient is required to be performed in parallel with the present treatment. Usually, several treatment rooms are provided to make preparation in the other rooms. It, however, requires rather larger number of operators. So we propose such a system that positioning of the patients is performed outside the treatment room with use of X-ray CT and so on. After entering into the treatment room, the precise positioning of the patient to the beam is realized with mechanical alignment of the positioning stage with use of knock-pins[4]. Thus, only one treatment room will be provided. The required beam specification by the medical use is listed up in Table 1.

3. Injector Linac

In order to make the injector linac to be compact, rather higher frequency of 433 MHz is adopted. In order to attain the required maximum dose rate of 5 Gy per min., proton beam with the intensity of 5×10^{10} per pulse is needed for irradiation with scanning instead of scatterer. If this intensity of the proton beam is injected into the synchrotron ring with the kinetic energy of 7 MeV, the betatron tune shift due to space charge force is estimated to be 0.09 assuming that the emittance of the injected beam are 145π mm·mrad and 10π mm·mrad in horizontal and vertical directions, respectively. This tune shift seems tolerable and the injection energy of 7 MeV is acceptable. The scanning method has rather higher beam efficiency in producing wide irradiation field[5], which makes the beam injection into the synchrotron with necessary intensity much easier.

As the injector linac, we propose the combination of the RFQ and Alvarez Drift Tube linac, which are similar to the one already operated at ICR, Kyoto University since 1992[6]. As the RF source, the system based on the klystron as the existing one and that based on parallel planar triode arrays should be carefully compared from the point of view of reducing the beam down time as short as possible and attaining the easy operation. The latter system commercially available can generate up to 360 kW while the current proposed system needs ~500 kW of peak power. Some R&D is necessary for this power source.

4. Synchrotron

With use of combined function lattice, complexity related to the tracking between power supplies of dipole and quadrupole magnets can be avoided. Another element which requires complex control is an RF accelerating cavity, because its frequency changes according to the beam acceleration. In ordinary proton synchrotron, the resonant frequency of the accelerating cavity is tuned by the bias magnetic field of the ferrite loaded cavity. In the proposed case, the required voltage is fairly low and the possibility of the untuned cavity is studied. If the synchrotron ring as shown in Fig. 1 is operated as shown in Fig. 2, the acceleration voltage and revolution frequency should satisfy the relation shown in Fig. 3. Only a few hundreds volts is needed as the gap voltage and untuned RF cavity driven by a solid state amplifier is adopted to make the operation easy, which eliminates the control of the biasing current for the ferrite. A power model of the untuned cavity has been fabricated and its structure is shown in Fig. 4. We have devised a new driving method for the untuned cavity by coupling the RF power to each winding of the ferrite ring. Owing to the good efficiency of power feeding, necessary gap voltage of several hundreds volts can be attained with the power less than 1 kW which can be easily generated by a solid stale amplifier[7].

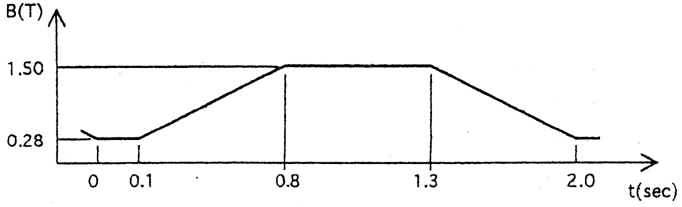

Fig. 2 Excitation Pattern of the Synchrotron.

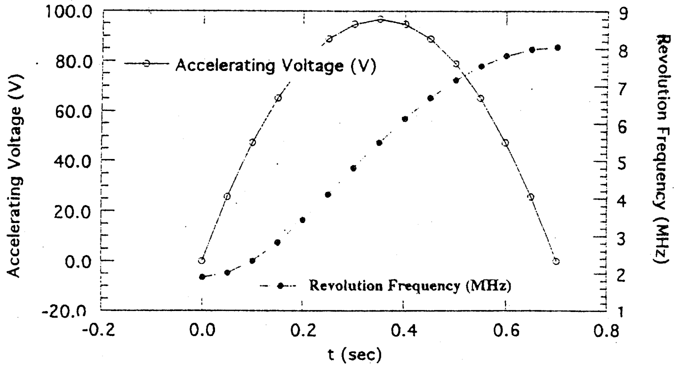

Fig. 3 Sweeping Pattern of RF frequency and Voltage.

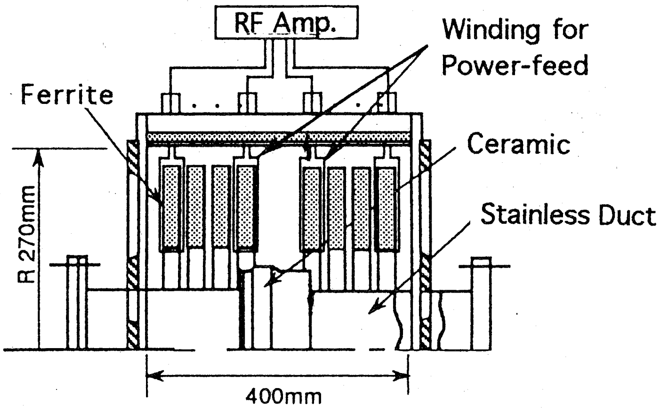

Fig. 4 The really fabricated untuned cavity.

Further a digital synthesizer is utilized as the signal source of the RF because its purity in frequency component is known to contribute easy acceleration control from the experience of HIMAC[8]. It is expected that no feed back for phase and radial position differences might be necessary if the B dot clock for frequency control is precise enough (~0.2G), once the sweep pattern of the RF frequency is fixed. So the RF feed back with beam signal will be utilized only for fixing the new excitation pattern, which makes daily operation very easy.

In order to provide the beam with long duration from the synchrotron, slow beam extraction with third order resonance is to be utilized. In the proposed system, the new method to utilize transverse RF electric field is to be utilized[9,10]. The betatron amplitude of the beam which initially locates inner side of the separatrix is enlarged by this RF electric field. This method is effective to realize the small emittance extracted beam and synchronization with breathing.

In Table 2, the main parameters of the proton synchrotron are listed up.

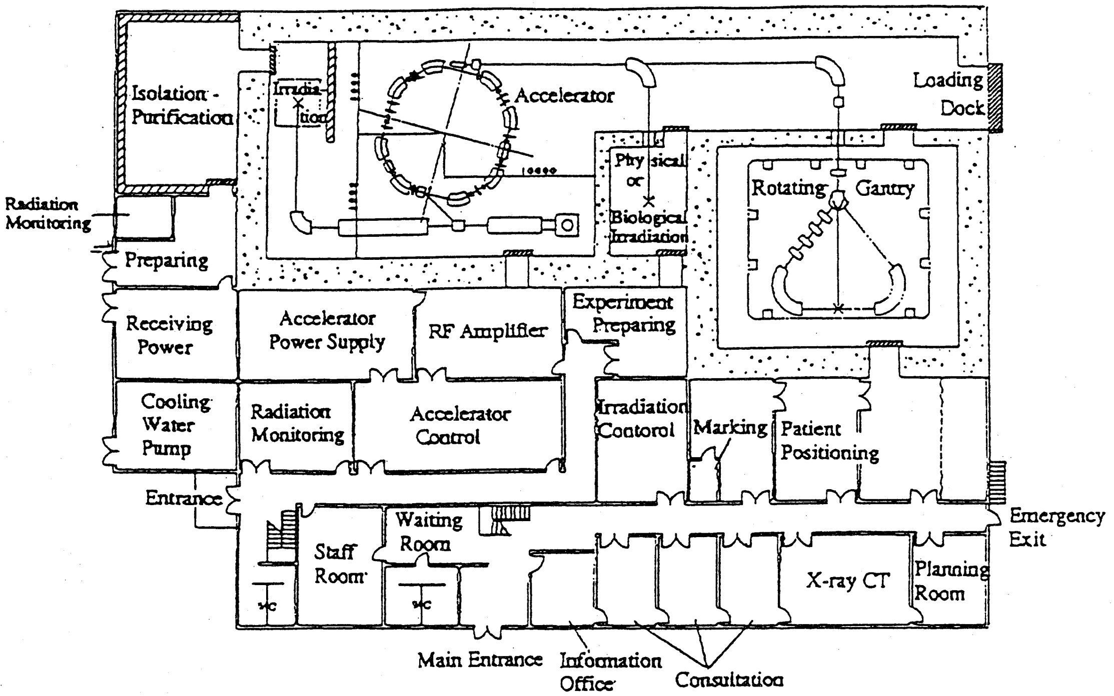

Fig. 5 Layout of the proposed KUMPF as the example of the irradiation facility with a single treatment room.

5. Irradiation System

It is needed to enlarge the irradiation field to some size as 15 cm in diameter to cover the region of the tumor to be treated. For this purpose, we propose to utilize the scanning method2). The scatterer reduces the beam utilization efficiency to ~1/4, while the present method is expected to utilize the extracted beam more than 85 %5). This reduced the necessary beam intensity from the synchrotron to 5 x 10^{10} at the needed dose rate of 5 Gy/min. Due to this reduction of the beam intensity, the space charge tune shift is under 0.01 even if the injection energy is 7 MeV, which is considered to contribute the injector system to be compact.

In order to make efficient irradiation to the tumor with smallest possible damage to the surrounding normal cells, multi-port irradiation is inevitable. For this purpose, we propose a combination of a single horizontal beam line and a rotating gantry in the same treatment room4). With this configuration, concentration of the irradiated dose to the tumor is considered to be well realized.

Table 2 Main Parameters of the Proton Synchrotron

Injection Energy	7 MeV
Maximum Energy	250 MeV
Circumference	22.9 m
Superperiod	4
Number of Betatron Oscillations per Turn	
Horizontal Direction	1.75
Vertical Direction	0.85
Magnet System	
Dipole Magnet (combined function)	
Number of Magnets	8
Deflection Angle	45°
Radius of Curvature	1.62 m
Maximum Field Strength	1.5 T
n-value	
at the horizontally focusing section	-0.599
At the vertically focusing section	0.904
RF System	
Frequency Range	1.59-8.04 MHz
Acceleration Voltage	~100 V
Cavity	Ferrite Loaded-untuned
Signal Source	Digital Synthesizer

As already mentioned, a single treatment room equipped with both a horizontal and a rotatable gantry is provided which keeps the needed operator as small as possible. The positioning of the patients will be performed outside the treatment room relying on the mechanical precision of alignment. In Fig. 5, the layout of the KUMPF is shown as an example of such medically dedicated facility.

The present authors at Kyoto University are proposing to realize such a facility at first at their university in order to make R&D research with real facility4).

6. Acknowledgements

The authors would like to present their sincere thanks to the group of the HIMAC at National Institute of Radiological Sciences in Chiba for providing them a lot of valuable information based on the experience of operation of the HIMAC. They are also grateful to Prof. K. Ono at Research Reactor Institute, Kyoto University for his fruitful discussion on the KUMPF. This work is supported by Grant-in-Aid for Scientific Research from the Ministry of Education, Science and Culture.

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