PROGRESS REPORT ON THE PROTON THERAPY FACILITY PROJECT AT NATIONAL CANCER CENTER, KOREA

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Abstract

A proton therapy facility is under construction at the National Cancer Center in Korea to utilize proton beams accelerated from a 230-MeV commercial cyclotron for radiation therapy. The building for therapy equipments will be completed in January 2005, and the site survey for equipment installation will begin a month earlier. The major components such as cyclotron and gantries have been pre-assembled and tested at the factory before being packed for shipping to the NCC. To prepare for the clinical operation in 2006, we are trying to attain permissions from the Korea Food and Drug Administration and the Korea Institute of Nuclear Safety. Also, we have worked on Monte Carlo simulations for therapeutic beam formation, and on dosimeter systems to measure dose distributions of dynamic beams.

INTRODUCTION

A proton therapy facility is under construction at the National Cancer Center in Korea to provide cancer patients with high quality radiation therapy [1]. The facility will be equipped with a 230-MeV cyclotron, two gantries and one horizontal fixed beam line. The beam line is also extended to the experimental area. The proton beam equipments are manufactured by a Belgian company IBA, and are considered as the next generation of the similar equipments currently in operation at the Northeast Proton Therapy Center (NPTC) in Boston [2]. The rigging of the cyclotron and two gantries will be started in January 2005 when the building is fully ready. The installation will take about one and a half year before accepting the first treatment room. The first patient is expected to be treated in the mid 2006.

The facility design asked for thorough investigations on radiation shielding. The shielding calculations have been performed using both analytic and Monte Carlo methods.

To prepare for the clinical acceptance of the equipments, we have performed some beam measurements at the Midwest Proton Therapy Center (MPRI) in the US and also Monte Carlo simulations. In addition, we are building a dose measurement system with a CCD camera and scintillation screen for scanning beams. Also, a large-area parallel plate chamber is under construction to be used at the experimental area.

FACILITY CONSTRUCTION AND MAJOR EQUIPMENTS

Construction of the proton therapy facility was started from June 2003, and is expected to be completed in January 2005. The layout of the facility is shown in Fig.1. The treatment rooms are located in the second floor of the basement as the first floor is occupied with several units of conventional electron linear accelerators and CT simulators used for treatment planning. Configuration of the facility essentially looks after the typical designs of proton therapy facilities like the one at Loma Linda University Medical Center and the NPTC. Two gantries will be installed in 2005, and a room is reserved for the third gantry to be installed in the future.

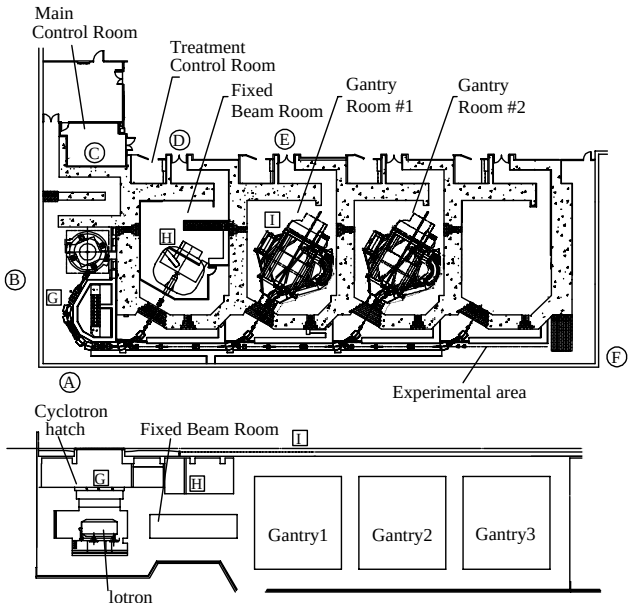

Figure 1: Layout and sectional view of the proton therapy facility.

The major components are: cyclotron, gantry, beam line elements, patient positioning system, and control system. The cyclotron is a fixed energy machine, and the beam energy is controlled to the range of 60-230 MeV using a graphite energy-degrader system followed by the slits and collimators to define the beam energy usually by less than 1%. Two similar cyclotrons are in operation at the NPTC and NCC, Kashiwa in Japan [3], and we expect improved performances in the aspects of beam extraction efficiency and rf power losses. The major parameters of the cyclotron are listed in Table 1. The gantry is of a large throw type having two dipole magnets with its diameter of 11 m. The beam line along the gantry is achromatic in both planes to deliver high quality beams at all angles.

DESIGN OF RADIATION SHIELDING

The primary source of radiation produced by bombardment of proton beams on a metallic target is

Table 1: Major parameters of the cyclotron

Energy	230 MeV
Beam current	1-300 nA
Yoke diameter	4.3 m
Weight	220 tons
Intensity modulation	15 μ sec (20-80 %)
Radio frequency	106.1 MHz
Harmonic mode	4

neutron. In Monte Carlo calculations the target is assumed to be a copper block thick enough to stop the beam. The energy spectrums of neutrons are calculated with MCNPX [4] at a few representative energies. The spectrums for a 200 MeV proton beam are displayed in Fig. 2 at different angles. A comparison is made to the case of a polyethylene target, which shows substantially lower neutron productions. The shielding wall is made of regular concrete with a density of 2.3 g/cm^2, which is effective in neutron shielding and also structurally sound. The gamma radiation mostly having energy below 10 MeV is well shielded by the thick neutron shielding walls.

The proton beam losses are distributed along the therapy equipments. The major locations of the losses are: 1) inside of the cyclotron, 2) in the beam line of the energy degradation and energy selection, and 3) sections of therapeutic beam formation. Other locations are designed to have minimal beam losses besides the beam stops to interrupt the beam in emergency and to measure the beam currents. These losses are conservatively evaluated based on the experiences at the NPTC.

The radiation shielding was first evaluated with an analytic expression adopted by Tesch [5], which was also used for the shielding wall design at the NPTC [6]. The expression to evaluate the dose is given below:

$$H = \sum_i \sum_\theta N_{i\theta} H_{casc,i\theta} e^{-\frac{\rho T}{\cos(\alpha)\lambda_{i\theta}}} / d^2,$$

where $N_{i\theta}$ is the number of protons with energy E_i lost at θ with respect to the shield, $H_{casc,i\theta}$ is cascade neutron source term, ρ is shield density, T is shield thickness, α is angle between wall and ray connecting source and receptor, $\lambda_{i\theta}$ is shield attenuation length, and d is distance from source to receptor. Evaluations have been made around the facility, and in Table 2 are listed some results at the receptors indicated in Fig.1. The limit of radiation dose is 1 mSv/yr in public area. This limit is met by considering proper occupancy factors into each area.

In addition, we have performed Monte Carlo calculations with MCNPX to evaluate the doses outside of the walls more accurately as listed in Table 2. The MCNPX results are generally lower and known to be closer to the measurements [6]. Both mechanisms of neutron production in a target, evaporation and cascade are included in the MCNPX calculations, while evaporation is not considered in the analytic calculations at the receptors behind the walls thicker than 1.5 m.

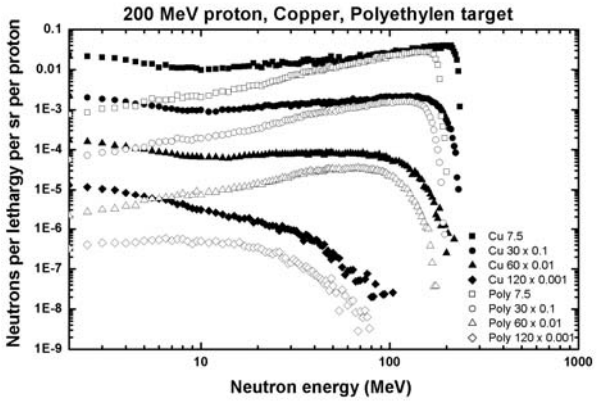

Figure 2: Neutron energy spectrums at different angles for two different target materials.

The usage pattern of the beams for therapy was assumed considering patient throughput, yearly operation hours, and so on. Some usage of the experimental area is also assumed.

Radiation monitoring system is located around the facility mainly to monitor neutrons. The detector is a BF3 proportional counter with a polyethylene sphere of 25.4 cm diameter as moderator.

Table 2: Analytical and MCNPX results at different receptors indicated in Fig.1.

Location	Analytic (mSv/yr)	MCNPX (mSv/yr)
A	0.79	0.36
C	<0.1	<0.1
D	1.7	0.3
E	2.4	3.1
F	1.7	0.41
G	2.4	3.3
H	0.57	0.19
I	3.0	2.7

MEASUREMENTS AND SIMULATIONS OF THERAPEUTIC PROTON BEAMS

The end of therapeutic beam line is named as nozzle. There are three methods of therapeutic beam formation: 1) scattering, 2) wobbling, and 3) scanning. We plan to use a double scattering method in the beginning, and then a wobbling method for the first gantry as the method becomes more solidly defined. The raster scanning nozzle will be installed for the second gantry as soon as the system is available, and eventually for the first gantry.

The therapeutic beams formed at the horizontal beam line of the MPRI have been measured [7] and also simulated using GEANT4 [8] as part of preparatory efforts. Figure 3 shows the formation of Spread-Out Bragg Peaks (SOBP) produced by a rotating modulation wheel together with the configuration of the beam line. The modulation wheel is made of layers of Lucite plates with its angular extension matched for the intensity of each Bragg peak as shown in Fig. 3. Dose distributions

were measured for different processes of therapeutic beam formation to be compared with simulation results.

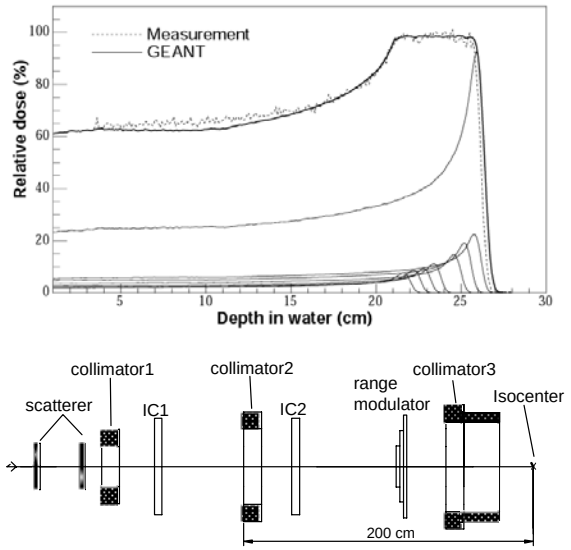

Figure 3: Formation of a 5 cm SOBP with a rotating modulation wheel and arrangement of beam-modifying elements in the horizontal fixed beam line of the MPRI.

PROTON BEAM DOSIMETRY

A film can be used for radiation dosimetry in 2D, but the image is not in real time, and it is known that the dose response is significantly nonlinear especially for proton beams. On the other hand a CCD camera-scintillator system [9] can measure dose distributions in real time. Mechanism is similar to that of the electronic portal imaging device used for conventional photon fields.

Detailed design of the system is underway as schematically shown in Fig. 4, and it will soon be assembled. A beam test is scheduled to be carried out at the Korea Institute of Radiological Sciences housing a 50 MeV proton cyclotron this year. In addition to the Bragg peak measurements, images will be taken for different arrangements of apertures and collimators with the CCD camera, and will be compared with Monte Carlo calculations. Absolute doses will be measured using standard parallel-plate Markus and Farmer chambers, which will be used to calibrate the CCD device at different energies.

The scintillating screen to be firstly used is Gd_2O_2S:Tb. A major difficulty of the phosphor scintillator is that there is quenching of light emission at high doses, which appears at the Bragg peak. The peak dose is not correctly reproduced, although it can be corrected. A mixture of different light-emitting phosphors is known to correct this nonlinearity. We plan to test various types of phosphors.

The camera system is thought to be a useful tool especially for Quality Assurance of moving beams. But it is still a 2D system, so it doesn't concurrently show the doses in different energies. The system is currently designed to be capable of shifting the range manually by placing different thicknesses of plastic phantoms in the upstream of the screen. The water phantom with bellow has been thought for remote control, but it has not been implemented. A real 3D system of dose measurement has been devised using multi layers of pixel chambers, but the usage in a hospital environment is found to be still not convenient.

As the hadron beams stop inside a patient, real time imaging of dose distribution during treatment can only be done with reaction products coming from the patient body. Two techniques 1) PET and 2) prompt gamma, have been discussed, and we are looking into the possibility of a prompt gamma detection system, but large neutron background and thus bulky shielding structure seems to make this technique not practical. We are trying to have a better signal to noise ratio and optimize the shielding design.

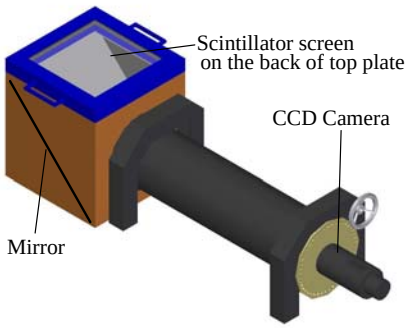

Figure 4: Schematic view of a 2D dose measurement system composed of CCD camera and scintillating screen.

CONCLUDING REMARKS

The proton therapy building is almost complete, and the major equipments like cyclotron and gantry will be rigged in early January 2005. Installation and clinical commissioning for the first treatment room will take over one year, and commissioning of the next two treatment rooms will need a few more months. The first patient is expected to be treated in the mid 2006. We are preparing own dose measuring devices to help commissioning the proton therapy equipments.

REFERENCES

[1] J. Kim, J. of Korean Phy. Soc., V42 (2003) 50.
[2] J. Flanz et al., Proc. 15th Int. Conf. on Cyclotrons and their Applications, (1998) 319.
[3] Y. Jongen et al., EPAC'98, Stockholm, (1998) 2354.
[4] MCNPX User's Manual Version 2.4.0. Los Alamos National Lab. Report LA-CP-02-408 (2002).
[5] T. Tesch, Radiation Protection Dosimetry, V11 (1985) 165.
[6] W. Newhauser et al., NIM A476 (2002) 80.
[7] V. Anferov et al., Proc. 16th Int. Conf. on Cyclotrons and their Applications, (2001) 27.
[8] S. Agostinelli et al., NIM A506 (2003) 250.
[9] S. Boon et al., Med. Phys. 27 (10) (2000) 219.