DEDICATED PROTON ACCELERATOR COMPLEX FOR A COMPREHENSIVE ONCOLOGY CENTRE

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Abstract

Beams of heavy charged particles represent the most advanced tools for external therapy of deep-seated tumours. The most important requirements for a medical accelerator are safety, reliability, beam stability, low energy consumption, and efficiency of beam delivery to the treatment room. Main peculiarity of the proposed dedicated accelerator complex is the machine design especially for the 'raster scanning' to provide the 3D conformal treatment of cancer tumours.

1 INTRODUCTION

Protons have excellent physical properties for radiation therapy, which permit one to control very precisely the shape of the dose distribution inside the patient's body. Specific arm of the project is to provide conformal 3D therapy using the 'raster scanning' technique [1] with an 'isocentric' gantry. These sophisticated 'active spreading systems' require changing during the treatment, at the same time the energy of the beam and, with magnets placed upstream, its direction. The 'raster scanning' technique is based upon an active energy- and intensity-variation within the treatment time. To realise this method of the tumor scanning, the energy of the beam must be varied from the accelerator and the accelerated particles should be extracted very slowly. The dedicated complex should consists of three treatment rooms with the 360-degree rotating 'isocentric' gantry and two rooms with horizontal beam.

2 MEDICAL ACCELERATOR COMPLEX

2.1 General Layout

The 3D 'active scanning' method allows one to construct dose distributions with complex shapes and to provide exact conformation of the dose to the target volume with variable range modulation. If desired, non-homogeneous dose distributions can be constructed. The flexibility of the 'active scanning' method can be used to correct the homogeneity errors in the dose distribution. Moreover, all the protons transported in the gantry beam-line are used directly in the patient. This reduces the activation of the collimation elements as well as the neutron background and the amount of the beam particles needed from the accelerator. This procedure requires extremely reliable accelerator operation guaranteeing spatial beam stability of better than a few millimeters at the target. Energy and intensity of the beam should be changed from pulse to pulse. The use of the slow resonant extraction extends the beam spill time sufficiently to perform on-line dosimetry at the patient and to switch the beam on and off according to the dose required.

Proton therapy requires beam energies in the range of 60 ÷ 220 MeV, changing with the step less than 0.4 MeV, to treat tumours at all depths. The energy accuracy should be 0.1 %. Beam intensities at least of 2×10^{10} protons per second are needed (~ 5 Gray/min) to reduce the treatment time to a few minutes. Intensities should be controlled over a wide dynamic range to the accuracy less than 2 %. The beam abort time must be less than *100 μsec*.

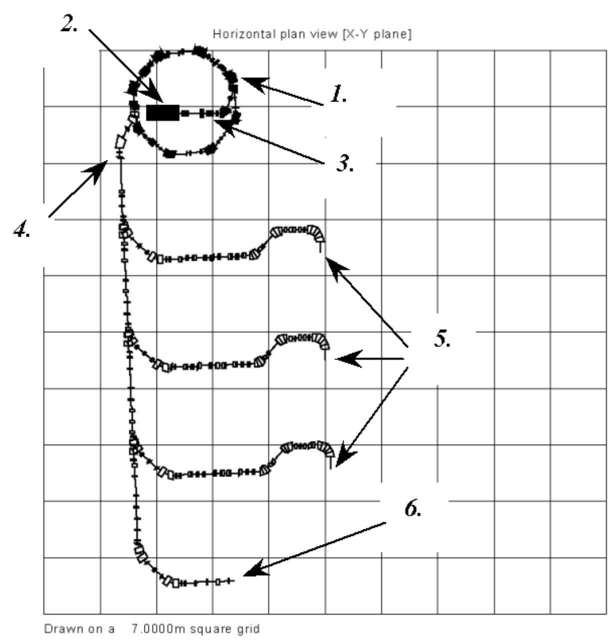

Figure 1: General layout of the Complex

Main elements of the complex are (see Fig.1): (1) proton synchrotron; (2) commercial injector; (3) injection

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beam transport system; (4) high-energy beam transport system; (5) three identical 'isocentric' gantries; (6) two horizontal beam-lines.

2.2 Main Elements of the Complex

(1) Dedicated Proton Synchrotron

The magnetic scanning of the extracted beam requires a smooth and slow extraction of the beam from the accelerator as well as a stable position of the beam spot during the extraction. Reasonable values for the rise time of the current in the deflection magnets need a stable extraction over a period of, at least, 400 ms in which one slice will be treated. It is necessary to provide an active energy variation decreasing the particle energies successively in steps on request from the scan system and to shift the particle range from slice to slice in the target volume. All beamline components must be adjustable to the actual beam rigidity within time duration between two beam pulses. These requirements directly determine the performance of the machine.

To get a small transverse emittance of the extracted beam a momentum-amplitude selection technique one can use [2]. The beam during the slow extraction is moved towards the stationary resonance by a betatron core. For improving of the spill quality it is possible to use a RF-empty bucket technique.

The focusing structure of the dedicated synchrotron should meet the specific requirements [3]. Main limitation of the ring circumference is determined by the maximum magnetic field of the ring dipole magnets (less than 1.4 Tesla to avoid steel saturation effects). Time variation of main parameters of the synchrotron are chosen to get maximum efficiency of the capture/acceleration/extraction process and to change the output energy from maximum till minimum level by request of the control system, connected with a dose measurement system.

Main elements of the synchrotron have been designed to meet technological and economic requirements. The ring has enough space to install diagnostic elements for measurement and correction beam parameters.

(2) Commercial Injector

The injection energy is 7 MeV, then the Model PL-7i linac (AccSyS Inc., USA) is chosen as the injector for the dedicated proton synchrotron. This commercial linear accelerator provides the necessary operational parameters, stability and reliability for use with the proton therapy synchrotron. The circumference of the synchrotron is enough to install the *Model PL-7i* linac inside of the ring to minimize the overall area of the accelerator hall.

(3) Injection Beam Transport System

Calculation of the injection beamline including influence of the space-charge effects, the chromatic aberrations and emittance growth in the debunching RF-cavity is performed. The obtained transfer efficiency from the Injector till the synchrotron is better than *3%*. The total length of the injection beamline from the injector till last bump-magnet (including the ring segment) is equal to *12.9m*.

(4) High-Energy Beam Transport System

In the extraction beamline for the hadron therapy accelerator complex, a rather special situation is met in which a fixed transfer line must be matched to a section of line, called the *Gantry*. In the case of the 'raster scanning' the beamline till the patient should be designed without any spreading systems (used for the 'passive' method). It means that the isocentric gantry (as main element of HEBTS) has to be able to rotate through a full 360^0 without affecting the beam spot at the patient. Main peculiarities of the beamline are connected with *non-symmetrical beam* after the synchrotron, *caused by the resonant extraction*.

Particle density of the extracted beam in the horizontal phase space for a given momentum will be quasi-constant over the '*rectangle*' with a constant emittance. The vertical transverse emittance one can describe as a *'Gaussian' beam* distribution changing adiabaticly for different final energy. The particle distribution of the extracted 70 MeV beam after the electrostatic deflector in the horizontal phase-plane is presented in Figure 2.

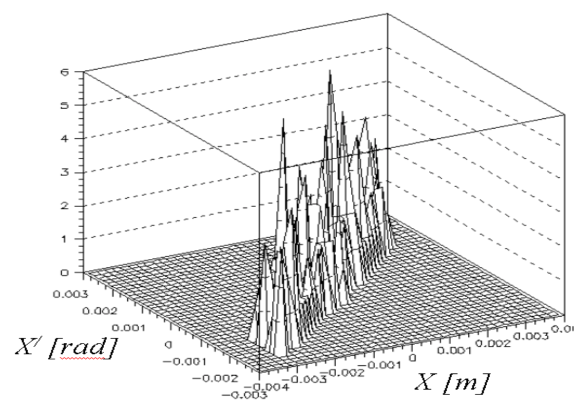

Figure 2: Particle distributions after the electrostatic deflector (70MeV)

The beamline from the ring till a single gantry consists of the following main parts: matching section between the ring and the extraction beamline; 'chopper' region; zero-dispersion bend; 'rotator' section. The beam envelopes along this line, corresponding to the 60MeV beam, are presented in Figure 3.

To eliminate dependence of the beam spot at the patient on the beam delivery angle, some special module (so-called a 'rotator' section) should be placed just before the 'gantry'. This module has to be physically rotated by half of the gantry angle. The 'triplet' rotator allows getting enough small ratios of the betatron amplitude fluctuation during variation of the rotation angle from 0^0 till 90^0.

(5) 'Isocentric' Gantry

To realize the 'active' technique with 'quasi-parallel' scanning the double deflection method should be considered in the 'gantry' design. It means, that the 'gantry' structure should have two dipole deflectors, installed before the last bending magnet. The spreading area should be $20\times 20\ cm^2$.

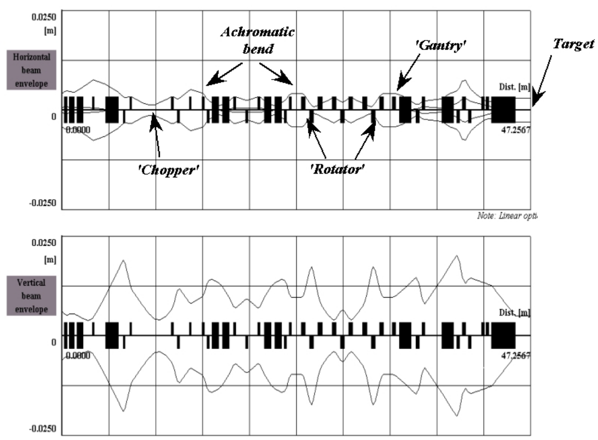

Figure 3: Beam envelopes along the HEBT (60MeV)

Radiation oncologists clearly prefer 'isocentric' systems to 'eccentric' ones. The main reason is to keep the 'isocentre' fixed with respect to the well- defined room coordinate system which makes the routine patient positioning and checking of the treatment angle easier.

The scanning system consists of one horizontal and one vertical scanning magnets. General view of the 'iso-centric' gantry from the matching point to the rotator and till the patient as a target is shown in Figure 4.

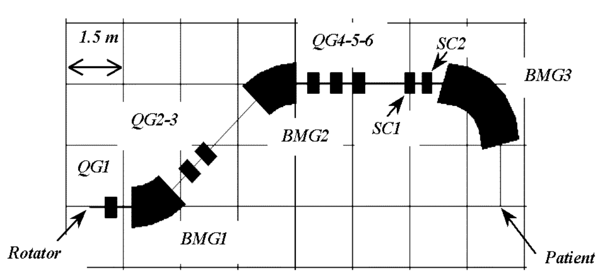

Figure 4: Layout of the 'iso-centric' gantry

Drift space between the edge of last dipole magnet and the isocentre should be ~ *1.5 m*. In this case the 'gantry' radius is equal to *3.0 m.* The 'gantry' length is equal to *10.76 m*.

In order to achieve the 'quasi-parallel' (with the angles less than 1^0) beam scanning, the scanning magnets *SC1&SC2* should be placed near the beam focuses in the horizontal and vertical planes. To get the 'quasi-parallel' beam after the last bending magnet, this dipole magnet should be the element with the edge focusing. The gap size of the magnet should be about *30cm* in the case of the double planes scanning of large treatment area ($20\times 20cm^2$). Parameters of six quadrupole magnets (*QG1-QG6*) and two 'small' bending magnets (*BMG1-BMG2*) are chosen to provide achromatic feature of the focusing structure, matching with the 'rotator' section and enough small beam size in the gantry.

The *FWHM* spot sizes of the 60 MeV beam on the target are equal to *4 mm* both the vertical co-ordinate and the horizontal co-ordinate (Figure 5). It means, that the developed high-energy beamline from the dedicated synchrotron till the 'target' (patient) meets the special medical requirements. The beam transport system can be refined, but this will not affect the beamline concept.

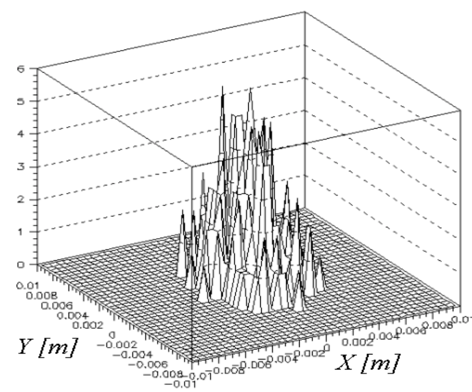

Figure 5: Beam profile at the 'isocentre' of the Gantry

Matching to the next gantry room and rooms with fixed horizontal beam has been designed. To minimize the cost of the Complex main parts of the matching sections should be identical. This section of the high-energy beam line is based on the regular FODO structure. The length of the matching section is determined by the transverse dimension of the gantry hall and shielding area.

3 COST ESTIMATION

Cost assessment of the Complex is about 20M$ including linac (6%), injection beamline (5%), synchrotron (21%), extraction beamline (3%), beamlines till treatment rooms (22%), three rotating gantries (36%), two rooms with horizontal beam (7%). This Complex could be used to cure about 1'000 patients per year.

REFERENCES

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