CYCLOTRON VERSUS SYNCHROTRON FOR PROTON BEAM THERAPY

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Two categories in the beam delivery system of proton therapy, scattering and scanning, are closely related to the temporal beam structure, in turn, type of accelerators. Since the synchrotron is well matured technically, efficient and dependable routine operation will be achieved if it is combined with a scatterer system. When more sophisticated scanning system is sought for better dose distribution, suitable *cw* beams are delivered by a cyclotron. Obviously it can be combined with any scattering system.

1 Introduction

The maximum energy for proton beam therapy is ranging $230MeV - 250MeV$ depending on regions or countries. The proton range in tissue is almost same as the range in water. The ranges of $230MeV$ and $250MeV$ protons are $32.7cm$ and $37.7cm$ respectively. The beam intensity specifications widely accepted for dedicated accelerators are $10nA - 20nA$ for dose deposition of more than $2Gy$ into a target volume in a few minutes. Most of the protons delivered from the accelerators, however, have been lost in a beam delivery system and usually a current of less than $1nA$ penetrates into a patient for cancer treatment.

Proton beams have a linear energy transfer (LET) of around $5keV/\mu m$ except near the end of their path in water. Although this figure is apparently bigger than that of photons, about $2keV/\mu m$, the biological effects of both protons and photons are very similar. Thus they are classified into the low-LET radiation. Advantage of protons over photons as a treatment modality is characterized by plateau and Bragg peak, *i.e.*, a long, low dose deposition path and a large peak near the end of their ranges. Other heavy particle beams such as neutrons or heavy ions are characterized by large relative biological effectiveness(RBE) due to their high linear energy transfer.

The pioneering works on proton beam therapy were made in U.S.A. and in Sweden. Synchrocyclotrons had been used exclusively until a synchrotron started to supply protons for the purpose in Moscow in 1969. Their energies were more than $160MeV$. Since eye melanoma has been treated successfully with Harvard Cyclotron protons in Boston, several isochronous cyclotrons for nuclear physics research, isotope production or fast neutron therapy have been used for irradiation of these tumors. Their energies are $80MeV$ or less, so that deep-seated tumor in a body can not be treated with these cyclotrons. What kind of tumor is frequent or not is in some cases depends on regions and races. The eye melanoma is very rear in Japan. On the other hand, liver cancer is much more frequent than in Europe or in North America.

The KEK booster synchrotron was originally designed and built as the injector of the $12GeV$ proton synchrotron. It is a rapid-cycling, combined function synchrotron of $20Hz$ repetition and injects 9 pulses of $500MeV$ protons into the main ring. After the injection completes, about 40 pulses are switched and delivered to the proton therapy facility, spallation neutron target or meson production target until the next injection starts. The $500MeV$ protons are degraded to $250MeV$, then transported to the treatment rooms. Based on advantage of the high energy, deep-seated tumors have been extensively treated at Tsukuba. Proton Medical Research Center, University of Tsukuba (PMRC) is responsible for patient treatment which started in 1983. Number of treated patients so far is 444, about 1/4 of which are liver cancer patients. Since clinical results, especially for liver cancer, are so promising that a new dedicated facility is being proposed. A synchrotron was designed originally at the first version, but it was replaced by a commercially available compact cyclotron later.[1], [2]

The first dedicated proton therapy facility was built at Loma Linda University Medical Center in California, U.S.A.. Its accelerator is a $250MeV$ edge-focusing synchrotron which was designed and built by Fermilab. A rotating gantry of $10m$ diameter was installed simultaneously, then other two have become operational sequentially. Most of space of the facility is devoted to treatment equipments such as rotating gantries, so that size of the accelerator itself seems less important than it was thought at its design. Patient treatment started in 1990 and patient totals by April 1995 are reported to be 1,262,[3] in which prostate cancer patients are predominant. The worldwide patient totals with proton beams are more than 15,000, which have been treated by 17 facilities. Another dedicated facility is being built at Northeast Proton Therapy Center (NPTC) in Boston.[4] The compact cyclotron is to be installed there.

Beam quality for therapy is different from that of other ordinary utilization for which monoenergetic protons are focused on a small target. At least a field of $10cm \times 10cm$ is required laterally with intensity uniformity of $\pm 2.5\%$. To spread Bragg peak to cover the tar-

get depth, protons of widely different energies should be injected into a target. The beam delivery system tailors accelerator beams to meet such requirements.[5] There are two methods of beam spreading, scattering and scanning. These are related deeply to temporal beam structure, in turn, to the type of accelerators.

2 Beam Delivery System and Temporal Beam Structure

Narrow and sharp accelerator beams are spread three-dimensionally to give uniform dose deposition over a target volume. A high-z thick scatterer, *e.g.* $6mm$ thick Pb plate, expand the impinging beam laterally by Coulomb multiple scattering. A single scatterer produces quasi Gaussian distribution laterally, in this case, beam utilization efficiency is about 10% to get uniformity of $\pm 1.0\%$. It is improved typically by a factor of three by a double scatterer system.

To make a spread-out Bragg peak (SOBP), a ridge filter or a range modulator is put just down stream of the scatterer. A system of a scatterer combined with ridge filters is completely static, so that it can accept beams of any temporal structure. It has favorable characteristics for routine patient treatment, *i.e.* simple and dependable. Since a ridge filter consists of many parallel metal ridges, the beam emittance should be large enough to achieve uniform dose distribution at the target. This inevitable large emittance causes large penumbra after protons pass through a final collimator.

When the beam is cw or periodic one with sufficient high repetition rate such as synchrocyclotron beams, a range modulator produces SOBP by changing thickness of low-z material, *e.g.* acrylic resins, quickly. It can be used with beams of any emittance. Protons, which pass through a scatterer and a ridge filter or a range modulator, drift to a final collimator and a bolus near a patient. The collimator cuts protons outside the tumor whereas the bolus, which is a kind of absorber, makes lateral energy distribution to irradiate complicated tumor and not to irradiate normal tissue beyond the tumor. The scattering systems have been well established and used widely for clinical treatment.

Although it fulfills crucial requirement of safety and reliability, the scatterer(s) consumes proton energies of $10MeV - 20MeV$, its drift space between the scatterer(s) to the patient should be long enough to get a large field, and some parts of SOBP near the proximal peak fall in normal tissue outside of the tumor. To overcome these shortages, various kinds of beam scanning system have been studied.

There are two types in the scanning system. One is the system in which protons are scanned magnetically to produce a large field. The scatterer(s) is replaced by scanning magnets, but a set of a final collimator and a bolus still remains. This is denoted by Scanning System I thereafter. The other is the spot scanning system denoted by Scanning System II in which a narrow pencil beam is scanned magnetically and computer-aided conformal therapy is intended.

One method in Scanning I is that proton beams are scanned by two orthogonal magnetic fields with sinusoidal exciting currents of different frequencies to produce a Lissajous pattern. The next is raster scanning in which the beam is scanned fast in x direction and slowly in y direction. The other is wobbling in which proton beams are circularly scanned. In these cases the proton beams must be cw or pulses of much higher repetition compared with the scanning frequency to achieve uniform dose deposition in a target volume. Obviously the beam intensity must be kept constant during scanning. Otherwise uniformity of dose distribution should be examined carefully for individual cases.

The spot scanning could deposit sufficient dose to a target volume keeping minimum dose to surrounding normal tissue. It requires more sophisticated beam control than that of Scanning I. A small volume in the tumor is irradiated to a predetermined level by a narrow beam. Then the beam is stopped, directed to a next small volume, and it is delivered until assigned level to this volume is attained. No bolus is needed. Clinical trial with this system will be soon performed at PSI.

3 Cyclotron vs. Synchrotron for Proton Beam Therapy

Pioneering works of proton beam therapy had been done by using synchrocyclotron beams. The synchrocyclotron was only one accelerator type which could deliver high energy protons for therapy. Now synchrotrons, isochronous cyclotrons or linear accelerators can deliver such protons. The linear accelerators are characterized by their high intensity capability, which may be too much for proton therapy. Since the existing proton linacs are pretty expensive, they seem not competitive to synchrotrons or cyclotrons even if they would be designed specially.

Based on the principle of phase stability, the synchrotron has no limitation on its accelerating energy. If a transition energy is reached on the course of acceleration, where the isochronous condition is fulfilled, proton bunch(es) becomes out of control of the accelerating rf field. Therefore, synchrotron designers try to avoid the transition energy. On the other hand, the cyclotron works always under the isochronous condition. Then restriction on the magnetic field is much more rigorous than that of the synchrotrons. Many synchrotrons have been built so far. As far as the single particle model is applicable, they are technically well matured, and design

energies are certainly attained. Potentially no beam loss occurs by fast extraction and around 90% extraction efficiency has been attained by slow extraction. At every big accelerator facilities excellent beam spills with little ripple or fluctuation have been realized. To get them, however, the power supplies of magnets must be very stable at the flat tops. Moreover every starting of a synchrotron needs time consuming tuning for such good beam spills. If a dedicated accelerator will start every morning, realize good beam spills and stop at night, it might be very inefficient even with the-state-of-the-art. This means a combination of a synchrotron with a scanning system is not practical. On the other hand, a combination of a synchrotron with a scatterer system including ridge filters is a good choice for an operation in which many patients are to be treated routinely.

If a cylindrical target volume is assumed and its base is perpendicular to the beam, there is no advantage of a spot scanning system over a scatterer system in dose distribution as far as the SOBP width is equal to the cylinder height. If the target is a sphere of several cm in diameter, SOBP, whose width is equal to the diameter, overflows from the target volume to normal tissue by a static system. However, the plateau dose reaches to around 80% of SOBP in this case, so that multi-portal irradiation should be selected to spare skin which is sensitive to radiation. If a tumor is irradiated through anterior and lateral portals, 100% dose region in normal tissue is greatly reduced. This is shown by a simple model in figure 1 where the plateau dose is constant and 80% of the SOBP dose. If the spherical target is irradiated through one portal, the volume of fully irradiated (100%) normal tissue is a half or 50% of the target volume. If the target is irradiated through anterior and lateral portals, it reduces to only 6.8%. If this target is irradiated through opposite directions, no more 100% dose region exists in normal tissue. It is important clinically to deposit more than a dose which kills tumor cells into the target and at the same time to deposit less than a dose which ensures recovery of irradiated cells into the surrounding normal tissue. It is apparent that the sophisticated spot scanning system would give better dose distribution than a scattering system has done. But clinical superiority should be confirmed in the future.

Cyclotron cw beams can be used not only for a scattering system but also for any of scanning systems if they can be quickly on or off in a switching time of $\sim 100 \mu sec$. The lungs and liver move sometimes up to $5cm$ when a person breaths. Beam switching capability is also required for a breath synchronized irradiation method, in which the tumor is irradiated only at a fixed phase of breath motion, although the switching speed is low ($\sim 100msec$). The switching may be possible by deflecting full energy protons, however, it can be done
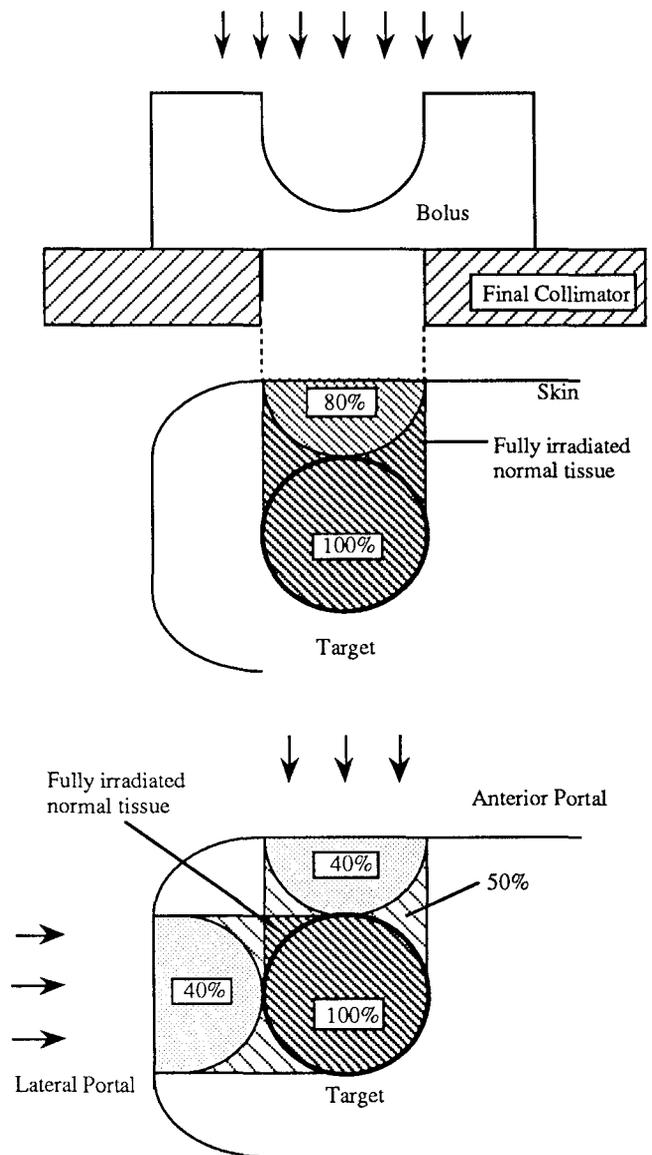
Figure 1: Dose distribution of simplified models irradiated through a set of bolus and final collimator. The plateau dose is assumed constant and 80% of the SOBP dose. Upper: The spherical target is irradiated through one portal. The volume of fully irradiated normal tissue is 50% of the target volume. Lower : The target is irradiated through anterior and lateral portals. The fully irradiated normal tissue reduces to only 6.8%.

easily by an external ion source system without producing unnecessary radioactivity. This system enables to install a standby ion source. More than 200 isochronous cyclotrons are in the world, but existing ones of more than $200MeV$ protons are only 7. Although merits of cyclotron beams have been recognized, magnets of an ordinary $200MeV$ isochronous cyclotron was estimated to be more than $1,000tons$ and it seemed too huge to install in a hospital. Although a H^- cyclotron can be energy variable, but it would be huger.

A $230MeV$ compact cyclotron of $200tons$ was proposed jointly by IBA in Belgium and SHI in Japan. The first one is being fabricated for NPTC in Boston. Fixed energy has demerits of unwanted neutrons for patients and smearing the distal edge by path length straggling in an additional energy absorber. However, they might not be so serious practically. According to an estimation at PMRC, where $250MeV$ protons have been used exclusively, unwanted irradiation due to neutrons which emerge in the scatterer and other parts of the beam delivery system is less than 1% compared with dose deposited in a target volume with protons in unit of neutron-Sv/proton-Gy.[6] It is impossible to determine deep-seated tumor edge definitely by CT imaging. Then medical doctors add margin on CT images of a tumor and direct to irradiate the tumor including the margin. In this case, the distal edge smearing becomes less important. The generally accepted minimum energy of $70MeV$ is for eye melanoma treatment, which has been treated with $200MeV$ synchrocyclotron beams at CPO in Orsay, France.

4 Conclusions

There are many criteria for accelerator selection, such as capital and operation costs, size, reliability, easy maintenance and so on. In addition to these items, a preferred beam delivery system, and in turn, temporal beam structure are closely related to the type of accelerator to be installed into a dedicated proton beam therapy facility.

The synchrotron is well matured technically. High extraction efficiency reduces radiation shielding. Energy variable capability is inherent. If it is combined with a well established beam delivery system of a scatterer and ridge filters, efficient and dependable routine operation would be achieved.

An attractive compact cyclotron is being fabricated for NPTC in Boston. Cyclotron cw beams are acceptable not only for any scatterer system but also for the most sophisticated scanning system aiming better dose localization, if they could be switched quickly. Fixed energy might not be a big disadvantage.

References

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