THE CBS-THE MOST COST EFFECTIVE AND HIGH PERFORMANCE CARBON BEAM SOURCE DEDICATED FOR A NEW GENERATION CANCER THERAPY

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Abstract

The electron cooling technology applied to a heavy ion medical accelerator will open a world of high performance yet with considerably lower cost. The cold beam enables innovative beam extraction methods. It also enables high energy economic beam lines less power consumption. The highlight of the cold beam accelerator is a two axis rotating carbon gantry.

INTRODUCTION

A carbon ion beam is a superior tool to x-rays or a proton beam in both physical and biological doses in treating a cancer. A carbon beam has an advantage in treating radiation resistant and deep-seated tumors. One of its radiological effects is a mitotic independent nature. These features improve hypo fractionation, typically reducing the number of irradiations per patient from 35 to a few. It has been shown that a superior QOL (Quality Of Life) therapy is made possible by a carbon beam. The only drawback is its high cost. Nevertheless, tens of Prefectures and organizations are eagerly considering the possibility of having a carbon ion therapy facility in Japan. Germany, Austria, Italy, China, Taiwan and Korea also desire to have ones. A carbon beam accelerator at moderate cost is about 100 Million USD. With the "CBS", Cool/Cold Beam Synchrotron, a design philosophy, which will be described in this paper, the cost could be factor of 2 or 3 less, while improving its performance to better than that of the standard designs. Novel extraction techniques, a new approach to a high intensity beam and a new scanning method of a super micro beam is possible. The CBS will have an impact on the medical accelerator community.

CARBON IONS FOR CANCER THERAPY

It has been shown from a clinical study of more than 2000 patients at HIMAC of NIRS that Carbon ion beams are a powerful tool for treating various types of cancers. GSI in Germany also has accumulated excellent clinical results. Carbon ions compared with protons, have a better relative biological effect (RBE), an effective Linear Energy Transfer (LET) to a cancer tumour are more effective to treat radiation resistant tumours, a mitotic independence of the breakdown of a DNA double strands and so forth. Carbon ions have the property of lower scattering along their trajectory. The rms size of the cold carbon beam created in the CBS could be as small as 0.2 mm. It is like an invisible laser beam. Carbon ion beams of the CBS are so sharp and indeed can be called as "carbon knife" by an analogy of a gamma knife or cyber knife of X-rays.

COOL CARBON BEAM

There are couples of methods to create a cold carbon beam. The simplest way is to use a cold beam ion source where the ion itself is created. A laser based ion source could be one of candidates but the technology seems not to have matured yet. The MEDEBIS proposed by Reinard Becker could be such a possible candidate [1]. In his design parameter of the MEDEBIS, the numbers of C^{+6} are $2.2x10^{10}$ per spill in a pulsed mode. The emittance is said to be moderately small but reliable data of carbon ions was not available. This ion source produces a fully stripped carbon ion. Although the MEDEBIS may have a problem of impurity content of O^{+8}, this source could be a good candidate of CBS.

In medical applications, the price of a carbon synchrotron system is more expensive than a proton accelerator system. Therefore numbers of patients to be treated become of important. In this respect, the beam intensity of an accelerator system is of primary concern as it is highly correlated to a number of treatment rooms or treatment length of period. From this point of view we took a different approach from that of Becker. We chose an Electron Cooling system (EC hereafter) for a cold beam creation.

POWER OF ELECTRON COOLING

The electron cooling was invented by Budker at the Institute of Nuclear Physics (BINP)[2]. Since then an active and extensive development has been undertaken by V.Parkhomchuk and others at BINP [3,4]. In recent years BINP has made state of art coolers for SIS (GSI, Darmstadt, Germany), CSRe and CSRm rings (IMP, Lanzhou, China) and LEIR at CERN. As a base for our project we take the design of the cooler EC-300 for CSRe storage ring.

How does the cooling help? The logic is as follows: With a beam property of low scattering along ion trajectories of carbon ions, (i) The edge of irradiation zone should be less then 1 mm (ii) Beta function of a magnet system should be about β_x=3 m where the length from the last magnet to a patient is about 2 m. (iii) As result the emittance should meet the condition of $\beta\gamma\sigma^2/\beta_x<$ 0.3 mm mrad. Normal value of emittance without cooling usually is 2-3 mm mrad. By a collimating the beam it is possible to reach 0.3 mm mrad at the cost of losing ion beam intensity.

Table 1: Basic parameters of main ring with cooling and without cooling at various energies.

Beam size in mm	5 MeV/n	120 MeV/n	400 MeV/n
r.ms. size without EC	8.6	3.8	2.5
5σ aperture without EC	43.0	19	12.5
r.m.s size with EC	2.25	0.42	0.2
5σ aperture with cooling	11.0	2.1	1.0

As one can easily see in Table 1, the aperture of the main ring with cooling can be made at 3 times less at the injection energy and 10 times less at extraction range of energy of 120-400 MeV/n. This looks reasonable to make a main ring system with aperture ±10mm. In this case a 3 times decreasing aperture will decrease weight of magnet at 10 times. This decrease is especially important for the distribution channel and the flexible magnets of gantry system. That can be done with aperture +-5 mm.

COLD BEAM SYNCHROTRON

At full energy the beam size of five standard deviation is only 1 mm. It may be a too radical for a medical ion synchrotron. In the first CBS to be constructed, we will have a large aperture margin to avoid a risk as there are so many new challenges in the CBS. In the following, we show the CBS layout of possible two cases in Figure 1 and 2, specification and parameters.

(i) The general specifications for the CBS facility are based on the following premises for the clinical requirements;

Clinical spec: 2 fixed port (horizontal and 45 degree)

Type of particles: Carbon, proton (option)

Ion energy: 140-400 MeV/u, a variable beam energy from spill to spill

Average dose rate: 5 Gy/min

Field size: 15 cm x 15 cm

Dose uniformity: ± 4% of the prescribed dose over treatment field

Delivered dose accuracy:2%

Irradiation method: Revised spot scanning system with synchronization of respiration or equivalent irradiation system

(ii) Below we list the treatment technologies.

• Active energy variation: The control system of the main synchrotron enables a smooth variation of ion beam energy. This is a technique that can only be realized with synchrotron accelerators. For the precise ion beam energy, varying the electron cooler device can be applied.

• Active beam radius control: The beam radius is controlled by active focusing elements, not by a passive collimator. By using active focusing elements instead of collimators the spot radius can be changed from spill to spill.

• Beam intensity variation: Different beam intensities can be chosen for the treatment. Each intensity step can be set from spill to spill.

• Spot scanning: The spot scanner moves the beam transversely over the tumor from pixel to pixel, i.e. from one spot to the next. This technical development enables a precise irradiation of irregularly shaped tumors and an optimum saving of the surrounding healthy tissue.

• Synchronization with respiration or any other types of motion: This is necessary because the efficiency of beam utilization is greatly improved with keeping a precise irradiation of cancer tissues.

In Figure 1 the base variant of CBS injection scheme is presented. The alternate variant of injection with a tandem pre-accelerator is presented in Figure 2.

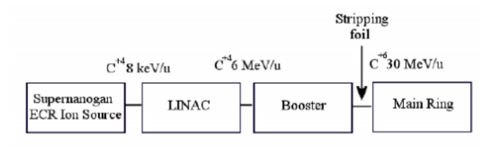

Figure 1: The base injection scheme.

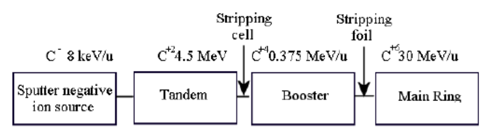

Figure 2: The alternate injection scheme.

For the base design of the injector, RFI of Linac System Inc. was chosen because of its high cost performance, low power consumption and capability of accelerating of both proton and carbon with high intensity [5].

In either case of the base design or alternate case, a rapid cycling booster is inserted between the injector and the main ring synchrotron. The addition of the booster greatly enhances beam intensity while keeping the additional cost modest. One more attractive feature of having the booster is the simultaneous acceleration of proton of the energy of 220 MeV, which is an appropriate energy of a proton cancer therapy. $^{12}C^{+4}$ is accelerated in the booster. The choice of 4+ instead of 6+ in the booster injection is to accommodate the main ring by raising a space charge limit at the injection of the main ring. In the following Table, some basic parameters of the main ring are listed.

Table 2. Basic parameters of the main ring synchrotron

Type of particle		$^{12}C^{+6}$
Injection energy	MeV/u	30
Extraction energy	MeV/u	140-400
Magnet rigidity	Tm	6.362
Circumference	m	80.565
Revolution frequency, min/max,	MHz	1/2.86
Betatron tunes	H/V	3.42/2.43
Bending radius	m	3.98
Magnetic field	T	0.4/1.6
Magnet gap	mm	36

COOLING

The electron cooling process can be applied both to the injection energy and the final energy. For the scenario of working cycle with a Linac, the cooling is a very ineffective process with large numbers of ions in the bunch. The main task for the cooler is the cooling of the ion beam at the top energy (140 –400 MeV/u). The ion beam with a small transverse size is suitable for the pellet extraction described below. In this case the requirements for the kicker are reduced.

In Fig.3, the evolution of the normalized transverse emittanse and longitudinal momentum spread for E_i = 400 MeV per unit are presented. The simulation was made for these parameters: the beta-function in the cooling section is 12.5 m, the electron beam radius is 0.3 cm, and the electron current is I_e = 0.6 A, the length of ion bunch is equal to the synchrotron perimeter. As it is seen, the ion beam emittance and momentum spread are effectively decreased during about 200 ms.(It is 100 ms at injection.)

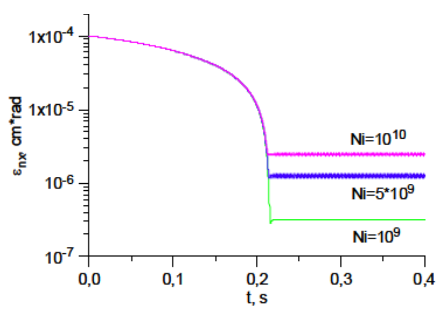

Figure 3: Time evolution of a normalized transverse emittance for different beam intensity. E_i = 400 MeV/u.

We have invented two kinds of extraction methods: The first one is a pellet extraction. There the bunched beam is divided into 1000 pellets(small number of ions) by two steps. In the first step, a second harmonic RF is introduced. 1/100 of the ions of fundamental bucket leak out into the second RF bucket. In the second step, we provide an ultra fast kicker of 10 kHz repetitions. A solid state switch with a rise time of 2 ns is available. Due to a cold beam property, the power consumption of the 10 kHz kicker is only 1 W. The second extraction method is to use EC to attach an electron to the C^{+6}. The probability to change to C^{+5} is very low and a small fraction of C^{+6} undergo a kick in the first dipole magnet of the arc and are then extracted. We call this a “recombination extraction”. A smooth slow beam extraction of the carbon ion is expected.

Finally, due to the extremely small beam size, the weight of the magnet of the gantry becomes tiny, thus enabling us to make a string of special flexible magnets. The length of vacuum duct also can be made variable so that an irradiation of the beam along a body of a patient at various angles is possible as well as around the axis. We call this a two axis rotating gantry. This optional gantry is shown in a layout of Fig.4.

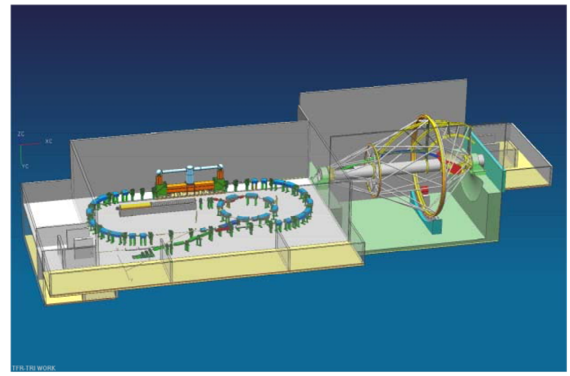

Figure 4: General layout of the CBS with rotating gantry.

CONCLUSION

The CBS can deliver much higher dose than any other carbon accelerator system under proposal by factor of several or more. Furthermore, the beam size of the main ring and high energy beam transport line could be reduced a considerable amount by EC. However, details of pellet extraction, a recombination extraction and 2 axis rotating gantry, could not be described. More details of the extraction and a novel treatment planning concept will be described in PTCOG42 to be held in June 8-10, in Tokyo.

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