A PLAN OF SYNCHROTRON LIGHT SOURCE FOR MEDICAL APPLICATIONS AT NIRS

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Abstract

We are designing a synchrotron light source dedicated to medical applications including an intravenous coronary angiography for practical uses at a hospital. The synchrotron light source system is required to be compact for installation in a limited space of the hospital. The system is also required to be operated easily as medical diagnostic devices. For the energy subtraction coronary angiography, required design parameters of a storage ring are circumference of about 45 m, 1.8 GeV electron energy and a stored beam currents of 400 mA. A multipole wiggler with superconducting coils is considered to be one of key devices to realize a required high photon intensity. A microtron with higher energy than that of commercially obtainable ones could be a good candidate of an injector for downsizing the whole system. Two optional beam lines are planned to be used for clinical diagnoses, and other beam lines of bending magnets are prepared for biology, physics, and other experiments to support the clinical diagnoses. We report the design of the compact system for the medical uses.

1 INTRODUCTION

Medical applications of synchrotron radiation have been researched at various synchrotron light source facilities in the world. Intense monochromatic X rays provided by the synchrotron light sources take many advantages on X ray imaging and medical diagnoses. A coronary angiography is one of well studied techniques, and has been clinically applied to examinees at HASYLAB[1] and KEK[2].

The applications are researched mainly in the facilities which have storage rings of electron energy more than 2 GeV. Because most medical applications require not only high energy X rays, but also intense photon flux compared with basic researches, such as a structural biology. High energy and large current storage rings are necessary for satisfaction of the both conditions. Compact rings dedicated to the coronary angiography were proposed in a few papers[3,4]. But the compact ring for the medical applications is not well studied in comparison with those for idustrial applications.

Our plan is motivated by a heavy ion radiotherapy which has been being carried out with carbon beams at NIRS since 1994[5]. More precise heavy ion irradiation can be achieved with help of a monochromatic X ray CT scans. Because the CT scans with the monochromatic X ray make a treatment planning of the heavy ion radiotherapy more accurate by giving more precise CT-

values. Conventional X ray tubes cannot produce enough photon flux to provide with intense monochromatic X rays. Another new techniques studied at NSLS[6] on a mammography and a bronchography are expected to give better quality images which help to find out cancers at early stages. Strongly focused X ray sources are able to make magnification of X ray radiography larger than the conventional one, so microscopic investigations is possible on examinees. There are many options of the medical applications, but, of course, they must be developed in complement with many other medical fields.

We mention here estimation of the photon flux for the medical applications and the basic design, and show its layout.

2 ESTIMATION OF PHOTON FLUX

Intravenous coronary angiography using synchrotron radiation is a powerful and non-invasive diagnostic method for finding out the coronary heart disease. This technique is one of the best studied method in the medical applications. It has also many common techniques with other diagnoses, and requires intense photon flux compared with the other applications. At these backgrounds we took the energy subtraction coronary angiography as the start point of the design.

Photon flux for the coronary angiography was estimated with conditions as follows.
1) Two dimensional animation of 30 frames per second.
2) Exposure area is at least 150 mm × 150 mm.
3) Each frame is produced by subtraction of two images taken at a higher energy X ray exposure and a lower one with respect to a K-edge energy of a contrast agent, respectively.
4) Exposure time of each image is less than 2 ms, and an interval time between two exposures is less than 5 ms.
5) At least 50 % stricture of a coronary artery can be clinically recognized.
6) Concentration of the contrast agent is more than 20 % at the coronary arteries.

The first condition is as same as the conventional angiography. The third one is necessary to avoid artifacts from the image due to cardiac impulses. The last condition is a case of catheter injection of the contrast agent from a central venous. Iodine is clinically used as a contrast agent, which has K-edge at an energy of 33.17 keV. The calculated photon flux is 1.8×10^{15} photons/sec/15 mrad with 0.3 % band width of 33 keV at 20 cm water equivalent body thickness. Higher energy photons have an advantage of less exposure dose on an

examinee because of low absorption rate. Gadolinium could be a good candidate of the contrast agent of the angiography, which has the K-edge at 50 keV. We calculated the flux at 8×10^{14} in case of the Gd with the same conditions. The Gd is used as the contrast agent for MRI, but so far the concentration is too low to use for the angiography. Nevertheless it is worth taking the new contrast agent into consideration for the design. Rough estimations of photon flux for the other applications, the large magnification X ray radiography and monochromatic X ray CT scan, preliminarily showed that flux was order of 1 or 2 less than those of the angiography. It means that about 2×10^{15} photon flux is supposed to cover the amount of photon required by other medical applications we will try in future.

3 DESIGN AND LAYOUT OF THE LIGHT SOURCE

3.1 Design parameters and Layout

We studied a few cases to search suitable design parameters as listed in Table 1. We took a space limitation into the consideration of the design. So the dimension of the ring is limited less than 15×15 m^2.

Table 1 Design parameters

Items	Designed values	
Beam energy	1.8	GeV
Beam intensity	400	mA
Circumference	44.8	m
Bending magnet	8	
bending angle	45°	
radius	1.33	m
maximum magnetic field	4.5	T
Beam emittance	4.9×10^{-7}	m·rad
Energy loss per turn (total)	1200	keV
RF frequency	508	MHz
RF voltage	2.0	MV
Microtron		
beam energy	300	MeV
peak current	10	mA
Multipole wiggler	2	
number of pole	9	
maximum magnetic filed	8	T
period length	400	mm

The system layout is shown in Figure 1. The main ring has no booster ring for preacceleration, but has a microtron as an injector. The ring has four free straight sections for insertion devices as well as for injection device and acceleration RF cavities. There are two beam lines from the insertion devices and a few lines from bending magnets.
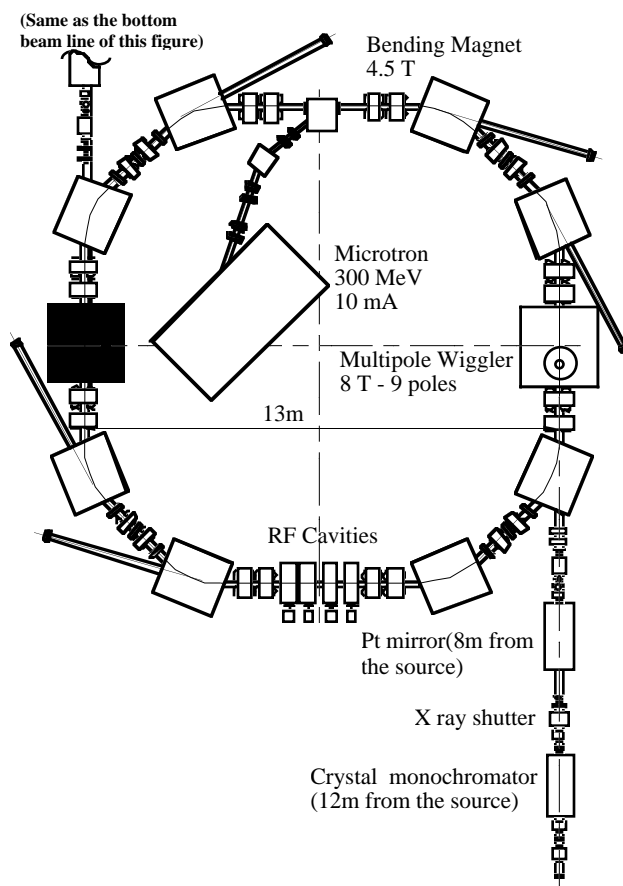

Figure: 1 System layout of the compact synchrotron light source for the medical applications.

3.2 Injector and Ring

A race-track microtron is preferred over a linear accelerator because of the space limitation. In this design 300 MeV microtron through 21 recirculations with 10 mA current was studied. The microtron is operated in L-band microwave instead of S-band for stable operation and higher stored energy. According to a numeric simulation, beam transportation efficiencies are almost 100% when the emittance of injected beams from an electron gun is less than 50π mm·mrad. The final beam conditions are chosen to be 0.1π mm·mrad emittance and $\pm1\%$ momentum resolution to meet ring injection conditions

The ring consists of eight superconducting bending magnets to reduce the circumference. The bending magnets and quadrupole magnets form a double bend achromatic lattice and make a dispersion free at the straight sections. Superconducting multipole wigglers give strong magnetic field at the straight sections. Their edge effect changes the beam tunes during raising up their excitation currents. The strength of quadrupole magnets have to be continuously varied to keep the tunes constant following the wiggler excitations. QF and QD at both sides of the wiggler are operated independently with another QF and QD in order to keep β_y function symmetry with respect to the wiggler. The operating point is

$\nu_x = \nu_y = 3.25$. There are three sextupole magnets between a pair of bending magnets to correct a chromaticities.

When the wigglers are operated, the beam emittance will become 0.5π mm·mrad due to radiation dumping. The maximum beam size is estimated 2.4 mm × 0.8 mm in the wiggler magnets. A quantum life time is more than 100 hours when the RF peak voltage is more than 1.9 MV. Touchek life time is more than 100 hours at 1 A of 1.8 GeV operation. A life time due to collisions with the residual gases in the ring is about 50 hours under an average pressure of 1.3×10^{-7} Pa.

Thermal loading can be a serious problem in case of the compact ring. The energy loss is 700 keV/turn at the bending magnets and 500 keV/turn at the wigglers. Power density can be estimated about 8.4 kW/cm^2 at the bending sections. In the wiggler magnets the radiation horizontally fans out 84 mrad, so parts of the power dissipate on the side wall of the beam ducts. Some special structured beam duct which has thermal dumper and effective cooling system is necessary at the bending magnets. The thermal loading of the wigglers can be reduced by widening lateral size of the beam ducts.

3.2 Insertion device and photon flux

The multipole wiggler is an essential component to obtain high photon flux. Design of the wiggler depends on design of the ring, and vice versa. We studied some cases to determine the designs of the wiggler and the ring under conditions of enough photon flux and less contamination of higher harmonic components of 33 and 50 keV which deteriorate image qualities.
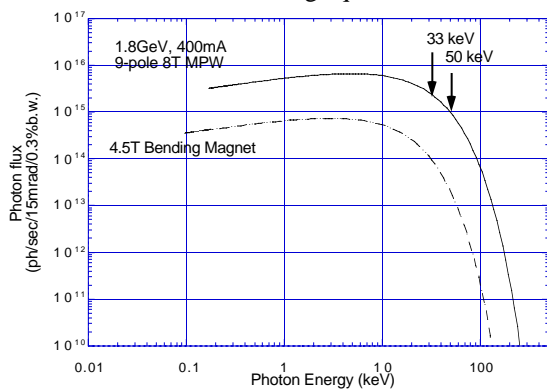

Figure: 2 Photon flux spectra produced by the 8T, 9-pole multipole wiggler and the bending magnet.

We put a limitation of the contamination of third harmonics at 5 %. The wiggler has 9-pole with period length is 400 mm and maximum field 8 T, that gives a critical energy $\varepsilon_C = 17.2$ keV at 1.8 GeV ring operation. The flux of 33 keV X ray is 2.2×10^{15} photons/sec within 0.3% band width and 15 mrad, and 1×10^{15} photons at 50 keV. The third harmonics is about 3.5 % and 0.5 % of the amounts of 33 keV and 50 keV photon, respectively. Energy spectra of the photon flux produced by the wiggler

and the bending magnet at the full electron energy are shown in Figure 2.

3.3 Beam lines for medical use

There are two beam lines from the multipole wigglers for clinical uses, and a few lines from the bending magnets for basic researches and developments of new technique.

Large silicone crystal of (311) as a monochromator generates X rays with a few tenth % energy resolution with asymmetry reflection. It broaden the beam vertically to 150×150 mm^2 wide at the examinee position. Another mirror is installed upstream of the monochromator to remove the higher harmonic components to less than 0.1 %. In this design, the transmission efficiency is less than 60 % for 33 keV X ray. Introduction of a parabolic mirror should be taken into the consideration to cover wider solid angle than 15 mrad.

4   SUMMARY

The light source designed here is applicable to the higher energy angiography using the gadolinium contrast agent. It meets to a tide of using higher energy photon in the medical application to give the low exposure dose. But there are some technical problems to be solved, such as thermal loading. This design should be elaborated more to overcome the problems and to realize the stable operation with the long lifetime for the clinical use.

5   ACKNOWLEDGEMENTS

REFERENCES

[1] Dix. W-R: `Intravenous coronary angiography with synchrotron radiation', Prog. Biophys. Molec. Biol. 63, 159(1995).
[2] Y. Sugishita, S.Otsuka, Y. Itai, T. Kakeda, M. Ando and K. Hyodo: `Synchrotron radiation coronary angiography and its application', Abstracts of 10th. meeting of the Jap. Soc. for Synchrotron Radiation. Research, pp.1,(1997).
[3] H. Wiedemann, M.Baltay, R Carr, M. Hernandez and W. Lavender: `A compact radiation suorce for digital subtractive angiography', Nulc. Instrum. Methods A347,515(1994).
[4] Y. Oku, K. Aizawa, S. Nakagawa, M. Ando, K. Hyodo, S. Kamada and H. Shiwaku: `Conceptual design of a compact electron storage ring system dedicated to coronary angiography', Proc. of the 1993 IEEE PAC, Vol.2, 1468.
[5] S. Yamada: `Commissioning and performance of the HIMAC medical accelerator', Proc. of the 1995 IEEE PAC, Vol.1 pp.5.
[6] W. Thomlinson: `Current status and perspectives of synchrotron radiation in medicine', Proc. of Int. Conf. on Applications of Physics in Medicine and Biology, Trieste, Italy 1996.