J. Synchrotron Rad. (1998). **5**, 336–338

Design of a compact synchrotron light source for medical applications at NIRS

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(*Received 4 August 1997; accepted 15 October 1997*)

A synchrotron light source dedicated to medical applications is required to be compact for installation in limited spaces at hospitals. The NIRS storage ring, with a circumference of 44.8 m, is designed to accelerate electrons up to 1.8 GeV and to store a beam of 400 mA. The ring is composed of superconducting bending magnets for downsizing. A beam of 300 MeV is injected into the ring from a microtron operated at an L-band RF frequency. There are two superconducting multipole wigglers with nine poles and a maximum field of 8 T, which can produce a photon flux of about 1.4×10^{13} photons s^{-1} mrad^{-1} (0.1% bandwidth)$^{-1}$ at 33 keV used for coronary angiography.

Keywords: compact rings; superconducting multipole wigglers; coronary angiography; monochromatic X-ray computer tomography.

1. Introduction

Intense monochromatic X-rays provided by synchrotron light sources have many advantages for X-ray imaging and medical diagnoses. We have a plan to construct a synchrotron light source dedicated to medical applications at the National Institute of Radiological Sciences. The final goal of the plan is to install a compact light source in a hospital in order to use it practically for medical diagnoses or radiotherapy. A typical medical application of synchrotron radiation is intravenous coronary angiography (CAG). It has been clinically applied to examinees as a non-invasive diagnostic method for discovering coronary heart disease (Dix, 1995; Sugishita *et al.*, 1997). Compact rings dedicated to CAG have already been proposed (Wiedemann *et al.*, 1994; Oku *et al.*, 1993). However, they have not been well studied compared with those for industrial applications, mainly because 2 GeV rings are required to obtain a high flux of hard X-rays for most medical applications and it is usually difficult to make such a ring compact.

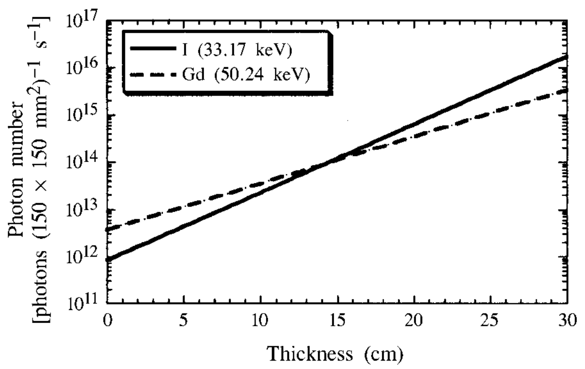

Figure 1
Photon flux required for energy-subtraction coronary angiography. Solid and dashed lines are for contrast agents I and Gd, respectively.

Table 1
Conditions on photon flux estimation for CAG.

Two-dimensional animation	30 frames s^{-1}
Each frame	Energy subtraction
Exposure of X-ray for each image	$<$2 ms
Interval between two exposures	$<$5 ms
Radiation field	150×150 mm
Minimum size of coronary artery	1 mm diameter
Contrast agents	Iodine, gadolinium
Concentration of contrast agents	2% at coronary arteries
Pixel size of detectors	0.25×0.25 mm

Our plan is motivated by heavy-ion radiotherapy, which has been performed with carbon beams at the NIRS since 1994 (Yamada, 1995). More precise heavy-ion irradiation can be achieved with the help of monochromatic X-ray computer tomography. Computer tomography with monochromatic X-rays makes the planning of the treatment of heavy-ion radiotherapy more accurate without the beam hardening effect. In addition, there are many other applications of synchrotron radiation, such as CAG, large-magnification X-ray radiography, fluorescent X-ray imaging, advanced mammography and bronchography (Thomlinson, 1995). These help discover cancers at an early stage and determine optimum treatments for the cancers.

Among them, CAG is one of the best studied applications. It has many techniques in common with the other X-ray diagnosis methods and requires a higher photon flux than the other applications. Therefore, the synchrotron light source designed for CAG may be applicable to other medical applications. Thus, we started designing the synchrotron light source for the purpose of performing energy-subtraction CAG.

2. Photon flux required for CAG

The basic idea is the digital subtraction of two images produced by two types of monochromatic X-rays (Rubenstein *et al.*, 1981). One type of X-ray has an energy higher than that of the K-edge of iodine, which is used as a contrast agent injected into coronary arteries through veins. The coronary arteries are enhanced in this image because of strong absorption of the X-ray. The other type of X-ray has lower energy than that of the K-edge, so that the arteries have similar tones in the image as other organs and bones. The subtraction of the two images gives a good contrast image for the arteries. Under the conditions listed in Table 1, the photon flux required for the CAG was calculated for two contrast agents, iodine and gadolinium. Here, we empirically assume the signal-to-noise ratio of the subtracted image to be 2, and a mixing ratio of scattered X-rays and direct X-rays to be 100%. Iodine is used as the contrast agent for angiography, which has a K-edge at 33.17 keV. On the other hand, gadolinium is used for magnetic resonance imaging, which has a K-edge at 50.24 keV. The calculated photon flux is shown in Fig. 1. For a body 20 cm thick, which is clinically a typical case, 6.4×10^{14} and 3.5×10^{14} photons s^{-1} (10 mrad)$^{-1}$ with 0.45% bandwidth are required for 33 and 50 keV, respectively.

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Table 2
Design parameters.

Beam energy	1.8 GeV
Beam current	400 mA
Circumference	44.8 m
Bending magnets	
Number of magnets	8
Bending angle	45°
Bending radius	1.33 m
Maximum field	4.5 T
Beam emittance	
(with wigglers)	4.9×10^{-7} m rad
(without wiggler)	7.3×10^{-7} m rad
Energy loss per turn (in total)	1200 keV
RF frequency	508 MHz
RF voltage	2.0 MV
Microtron	
Beam energy	300 MeV
Peak current	10 mA
Multipole wiggler	
Number of wigglers	2
Number of poles	9
Maximum field	8 T
Period length	400 mm

3. Design and layout of the synchrotron light source

The design parameters are listed in Table 2 and the system layout is shown in Fig. 2. The storage ring has an octagonal shape equipped with two insertion devices at the straight sections. A microtron is used as an injector to downsize the whole system. There are two insertion-device beamlines for clinical uses and a few bending-magnet beamlines for basic research and development of new techniques.

3.1. *Injector and storage ring*

For our purpose, a race-track microtron is preferable to a linear accelerator because it requires less space. The microtron accelerates electrons up to 300 MeV after 21 recirculations. It is operated at an L-band microwave frequency instead of S-band for stable operation of a high-current beam. Particle-tracking simulation shows that the beam-transfer efficiency is almost 100% when the emittance of the beam injected from the electron gun is less than 50π mm mrad. The microtron is operated at a repetition rate of 10 Hz with a duty factor of 0.003%. The 300 MeV beam has 1π mm mrad emittance and $\pm 0.1\%$ momentum spread, which meet injection conditions for the storage ring.

The ring consists of eight superconducting bending magnets. The bending magnets and quadrupole magnets form a double-bend achromat that makes the straight sections dispersion-free. Superconducting multipole wigglers installed at the straight sections produce a strong magnetic field, so that their edge effect largely changes the betatron tunes. The quadrupole magnets therefore have to be continuously varied in their strength to keep the tunes constant while the wigglers are being excited. Then the ring superperiods become two from four. The β functions and the dispersion function in this case are shown in Fig. 3. QF1 and QD1, located at both sides of a wiggler, are operated independently of QF and QD in order to keep the β_y function as constant as possible. $\nu_x = \nu_y = 3.25$ was found to be a good operating point for obtaining wide dynamic apertures. There are three sextupole magnets at a dispersive section to correct the chromaticities.

When the wigglers operate, the beam emittance becomes small due to radiation damping, as seen in Table 2. The maximum beam size is estimated to be 2.4×0.8 mm at the wiggler magnets. The quantum lifetime is more than 100 h. The Touchek lifetime is more than 100 h even at a beam current of 1 A. The lifetime due to Coulomb scattering with residual gases is about 50 h for an average pressure of 1.3×10^{-7} Pa.

Thermal loading is a serious problem for the compact ring. The energy loss is 700 keV per turn at the bending magnets and 500 keV per turn at the wigglers. In the wiggler magnets the radiation fans out horizontally with an angle of 84 mrad, so that the beam ducts should be wide enough to reduce the thermal loading due to the wigglers. A thermal dumper and effective cooling system are also necessary for the beam ducts of bending magnets.

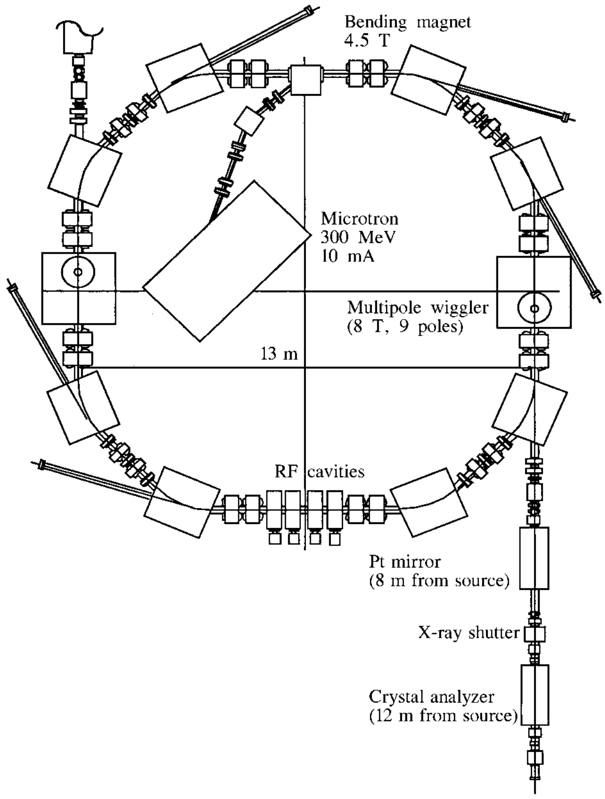

Figure 2
System layout of the compact synchrotron light source for medical applications.

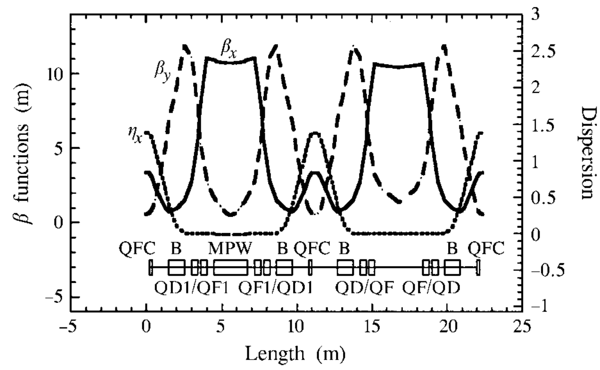

Figure 3
β functions (β_x, β_y) and dispersion function η_x of half of the ring when the wiggler is excited. MPW is the superconducting multipole wiggler.

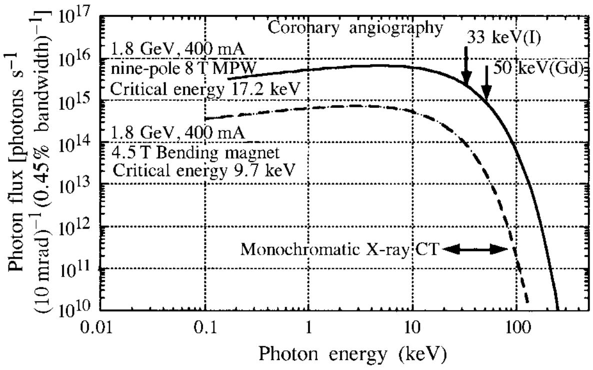

Figure 4
Photon flux spectra produced by an 8 T nine-pole multipole wiggler and a bending magnet.

3.2. *Insertion device and photon flux*

We have optimized both the wiggler and the ring under the conditions that enough photon flux should be obtained at the primary energies of 33 and 50 keV, but the higher harmonics, which deteriorate image qualities, be as little as possible. The optimum wiggler has nine poles with a period length of 400 mm and a maximum field of 8 T, which gives the critical energy ε_c = 17.2 keV at 1.8 GeV. At 400 mA stored current, it can produce a photon flux of 2.2×10^{15} photons s^{-1} (10 mrad)$^{-1}$ within a 0.45% bandwidth at 33 keV, and 1×10^{15} photons s^{-1} (10 mrad)$^{-1}$ at 50 keV. The ratios of the third harmonics to the primary components are 3.5 and 0.5% at 33 and 50 keV, respectively. Energy spectra of the photon flux produced by the wiggler and the bending magnet at 1.8 GeV are shown in Fig. 4.

3.3. *Beamlines*

The multipole wiggler beamline is composed of a mirror coated with Pt, an Si(311) asymmetry reflection monochromator, beam shutters, carbon degraders and windows. When the primary energy of the X-rays is 33 keV, the Pt mirror is used for cutting higher harmonics. The (311) reflecting plane forbids the reflection of the second harmonic. Reflectivities of the Pt mirror with 3 Å surface roughness are calculated to be 86% at 33 keV and 0.2% at 99 keV. In passing through 20 cm of water-equivalent thickness, the 33 keV X-rays are attenuated by about 1/800, which is 40 times larger than that of 99 keV X-rays. As a result, the amount of the third harmonic becomes 0.2% of that of the primary component behind the body. In the case of 50 keV X-rays, the attenuation rates of the primary and third-harmonic components are not very different in the body, and so the third-harmonic component is about 1% of the primary without the Pt mirror. Thus, the deterioration of image qualities is low enough in both cases to obtain clear contrast in the images. In order to obtain a radiation field of 150 × 150 mm, the synchrotron radiation fan beam is expanded vertically by employing an asymmetry reflection technique of the monochromator (Akisada *et al.*, 1986). For 33 keV X-rays the vertical beam size is expanded by 40 times from 3.8 to 150 mm, when the angle between the crystal surface and the (311) reflecting plane is 6.2°. For 50 keV X-rays the asymmetry reflecting angle of 4.1° gives the same radiation field. When the angle between the surface of the monochromator and the incident beam oscillate sinusoidally around 6.5° with an amplitude of 0.55 mrad, the energy of the reflected X-rays varies as

$$\varepsilon(t) = 33.17 + 0.16\sin(\omega t) \quad (\mathrm{keV}).$$

For example, 71.4 Hz is a suitable frequency for the energy subtraction using iodine as the contrast agent because the energy can change between 33.33 and 33.01 keV. Then, the averaged energies of 33.32 and 33.02 keV are obtained with ΔE = ±0.01 keV during 2 ms at the peaks and valleys of the oscillation at 5 ms intervals, respectively. The total transmission efficiency of the beamline is about 30% for 33 keV X-rays and less for 50 keV X-rays.

4. Summary

The system presented here can provide enough photon flux for CAG. The photon flux required for monochromatic X-ray computer tomography is estimated as $\sim10^{11}$–10^{12}, much less than for CAG. Therefore, the system may be applicable to other medical applications. The higher harmonic components are suppressed by less than 1% of the primary ones at the detectors. The transmission efficiency of the beamline is 30% for 33 keV X-rays.

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