33.2 keV absorb the lines easily. Therefore, blood vessels were observed with high contrasts.

The angiography was performed by the CR system[21] (Konica Regius 150) using the filter with a tube voltage of 60 kV, and the distance (between the x-ray source and the imaging plate) was 1.5 m. First, rough measurements of spatial resolution were made using wires. Figure 5 shows radiograms of tungsten wires coiled around rods made of polymethyl methacrylate. Although the image contrast decreased somewhat with decreases in the wire diameter, due to blurring of the image caused by the sampling pitch of 87.5 μm, a 50-μm-diameter wire could be observed.

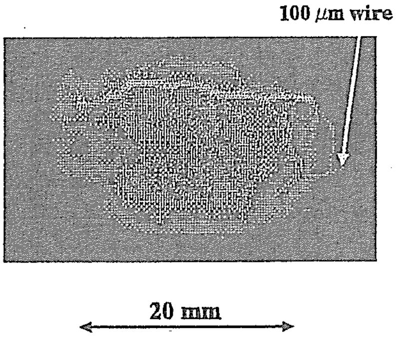

Fig. 6: Angiograms of an extracted rabbit heart using iodine microspheres.

An angiograms of a rabbit heart is shown in Fig. 6. This image was obtained using iodine microspheres of 15 μm in diameter. Fine blood vessels in the coronary arteries in the heart were visible. Figure 7 shows an angiogram of a larger dog heart using iodine spheres, and blood vessels of approximately 100 μm in diameter were visible.

5. Discussion

In summary, we employed an x-ray generator with a cerium-target tube and succeeded in producing cerium K-series characteristic x-rays, which can be absorbed easily by iodine-based contrast media. In the spectrum measurement, high-photon-energy bremsstrahlung x-rays beyond cerium K-edge (40.4 keV) were absorbed effectively.

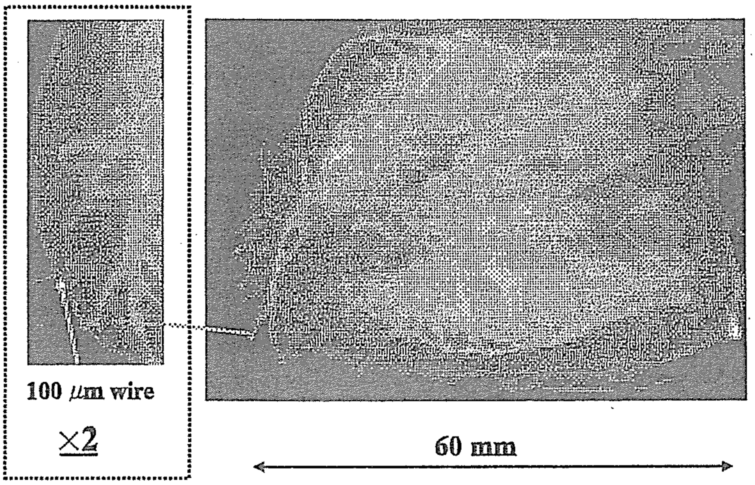

Fig. 7: Angiograms of an extracted dog heart using iodine microspheres.

In angiography, fine blood vessels were observed with high contrast with a spatial resolution of approximately 100 μm; the resolution was almost equal to the sampling pitch (87.5 mm) of the CR system. Therefore, the pith should be minimized, and magnification digital radiography including phase-contrast effect should be employed in order to improve the spatial resolution.

Although the cerium x-ray generator used in this research produces both the characteristic and the bremsstrahlung x-rays, bremsstrahlung intensity can be decreased effectively by considering the angle dependence without using the filter, since bremsstrahlung rays are not emitted in the opposite direction to that of electron trajectory. Subsequently, the generator produced maximum number of estimated characteristic photons was approximately 5×10^7 photons / ($\mathrm{cm}^2\cdot\mathrm{s}$) at 1.0 m from the source, and the photon count rate can be increased easily by improving the target.

Acknowledgment

This work was supported by Grants-in-Aid for Scientific Research (13470154, 13877114, 16591181, and 16591222) and Advanced Medical Scientific Research from MECSST, Health and Labor Sciences Research Grants (RAMT-nano-001, RHGTEFB-genome-005 and RHGTEFB-saisei-003), Grants from the Keiryo Research Foundation, The Promotion and Mutual Aid Corporation for Private Schools of Japan, Japan Science and Technology Agency (JST), and The New Energy and Industrial Technology Development Organization (NEDO, Industrial Technology Research Grant Program in '03).

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X-ray Spectra from a Characteristic X-ray Generator with a Molybdenum Tube

Eiichi Sato[a], Etsuro Tanaka[b], Hidezo Mori[c], Toshiaki Kawai[d], Takashi Inoue[e], Akira Ogawa[e], Kiyomi Takahashi[f], Shigehiro Sato[f] and Kazuyoshi Takayama[g]

(Received October 31, 2005)

Abstract

This generator consists of the following components: a constant high-voltage power supply, a filament power supply, a turbomolecular pump, and an x-ray tube. The x-ray tube is a demountable diode which is connected to the turbomolecular pump and consists of the following major devices: a molybdenum rod target, a tungsten hairpin cathode (filament), a focusing electrode, a polyethylene terephthalate x-ray window 0.25 mm in thickness, and a stainless-steel tube body. In the x-ray tube, the positive high voltage is applied to the anode (target) electrode, and the cathode is connected to the tube body (ground potential). In this experiment, the tube voltage applied was from 22 to 36 kV, and the tube current was regulated to within 100 μA by the filament temperature. The exposure time is controlled in order to obtain optimum x-ray intensity. The electron beams from the cathode are converged to the target by the focusing electrode, and x-rays are produced through the focusing electrode. Using a lithium fluoride curved crystal, clean K-series characteristic x-rays were observed without using a filter. However, bremsstrahlung x-rays were observed using a cadmium telluride detector.

Keywords: demountable x-ray tube, quasi-monochromatic x-rays, K-series characteristic x-rays, Sommerfeld's theory, curved crystal, cadmium telluride detector

1. Introduction

Most flash x-ray generators employ high-voltage condensers and cold-cathode x-ray tubes,[1-5] and the plasma x-ray source has been growing with increases in the electrostatic energy in the condenser. By

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[a] Department of Physics, Iwate Medical University, 3-16-1 Honchodori, Morioka 020-0015, Japan
[b] Department of Nutritional Science, Faculty of Applied Bio-science, Tokyo University of Agriculture, 1-1-1 Sakuragaoka, Setagaya-ku 156-8502, Japan
[c] Department of Cardiac Physiology, National Cardiovascular Center Research Institute, 5-7-1 Fujishirodai, Suita, Osaka 565-8565, Japan
[d] Electron Tube Division #2, Hamamatsu Photonics K. K., 314-5 Shimokanzo, Iwata 438-0193, Japan
[e] Department of Neurosurgery, School of Medicine, Iwate Medical University, 19-1 Uchimaru, Morioka 020-8505, Japan
[f] Department of Microbiology, School of Medicine, Iwate Medical University, 19-1 Uchimaru, Morioka 020-8505, Japan
[g] Shock Wave Research Center, Institute of Fluid Science, Tohoku University, 2-1-1 Katahira, Sendai 980-8577, Japan

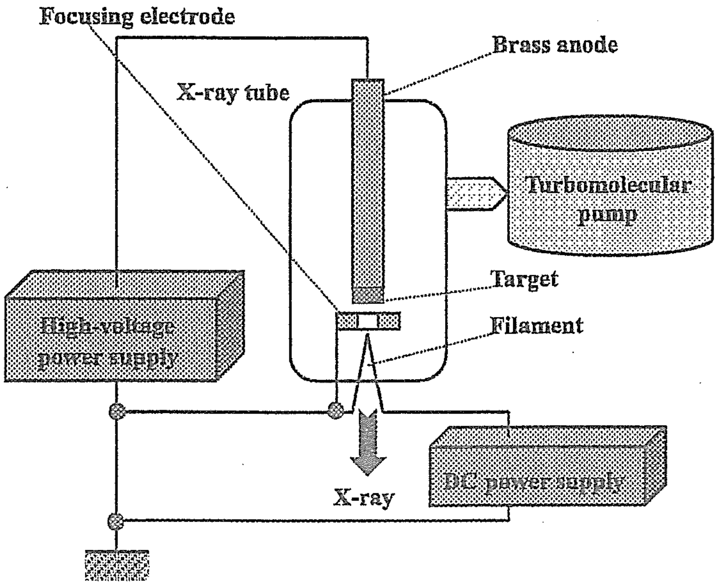

Fig. 1: Block diagram including the main transmission line of the compact x-ray generator with a quasi-monochromatic diode.

forming weakly ionized linear plasma[6-9] using a rod-target triode, we confirmed irradiation of clean K-series characteristic x-rays such as hard x-ray lasers and their higher harmonic hard x-rays from the plasma axial direction. Because the plasma transmits high-photon-energy bremsstrahlung x-rays, it is difficult to produce high-photon-energy characteristic x-rays. In view of this situation, we have developed a super-fluorescent x-ray generator[10-13] by forming weakly ionized plasma at the target tip and have succeeded in producing comparatively clean K-series characteristic x-rays of molybdenum, cerium, tantalum, and tungsten. In particular, the cerium target is useful for performing iodine K-edge angiography, and gadolinium K-edge angiography can be performed using tantalum and tungsten targets.

Steady-state K-series characteristic x-rays left by filters have been employed to perform mammography using a molybdenum target and to perform K-edge angiography[14-17] using a cerium target. In addition, the rays are also useful for performing real-time radiography achieved with a flat panel detector. Because the characteristic x-ray intensity decreases with increases in the filter thickness, the development of a characteristic x-ray generator utilizing the angle dependence of bremsstrahlung x-rays has been wished for.

In the spectrum measurements, we employ a cadmium telluride detector and a lithium fluoride curved crystal. The detector is useful to measure the total spectra including scattering beams. On the other hand, the spectra from only the x-ray source can be measured using the crystal by selecting Bragg's angle.

In this paper, we describe an x-ray generator developed and used to perform a preliminary experiment for generating clean K-series characteristic x-rays by angle dependence of the bremsstrahlung x-rays and measurement of the x-ray spectra using two methods.

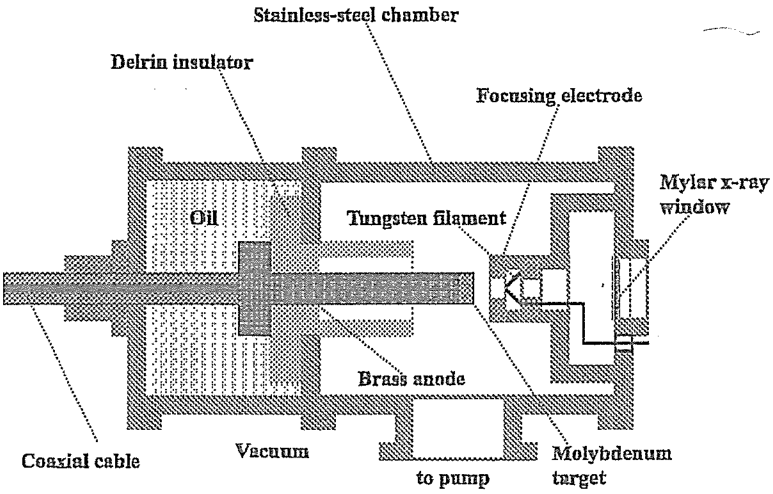

Fig. 2: Schematic drawing of the quasi-monochromatic x-ray tube.

2. Generator

Figure 1 shows a block diagram of a compact characteristic (quasi-monochromatic) x-ray generator. This generator consists of the following components: a constant high-voltage power supply (SL150, Spellman Inc.), a DC filament power supply, a turbomolecular pump, and an x-ray tube. The structure of the x-ray tube is illustrated in Fig. 2. The x-ray tube is a demountable diode which is connected to the turbomolecular pump with a pressure of approximately 0.5 mPa and consists of the following major devices: a molybdenum plate target, a tungsten hairpin cathode (filament), a focusing electrode, a polyethylene terephthalate x-ray window 0.25 mm in thickness, and a stainless-steel tube body. In the x-ray tube, the positive high voltage is applied to the anode (target) electrode, and the cathode is connected to the tube body (ground potential). In this experiment, the tube voltage applied was from 22 to 36 kV, and the tube current was regulated to within 100 μA by the filament temperature. The exposure time is controlled in order to obtain optimum x-ray intensity. The electron beams from the cathode are converged to the target by the focusing electrode, and x-rays are produced through the focusing electrode. Because bremsstrahlung rays are not emitted in the opposite direction to that of electron trajectory, clean molybdenum K-series x-rays can be produced without using a filter.

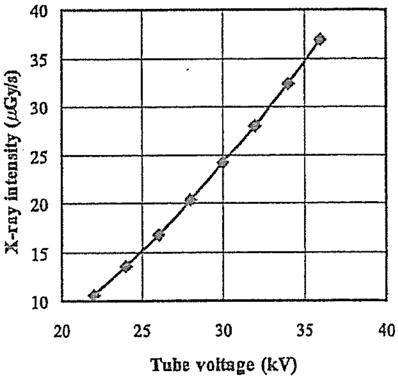

Fig. 3: X-ray intensity at 1.0 m from the x-ray source according to changes in the tube voltage.

3. Characteristics

3.1 X-ray intensity

X-ray intensity was measured by a Victoreen 660 ionization chamber at 1.0 m from the x-ray source (Fig. 3). At a constant tube current of 100 μA, the x-ray intensity increased when the tube voltage was

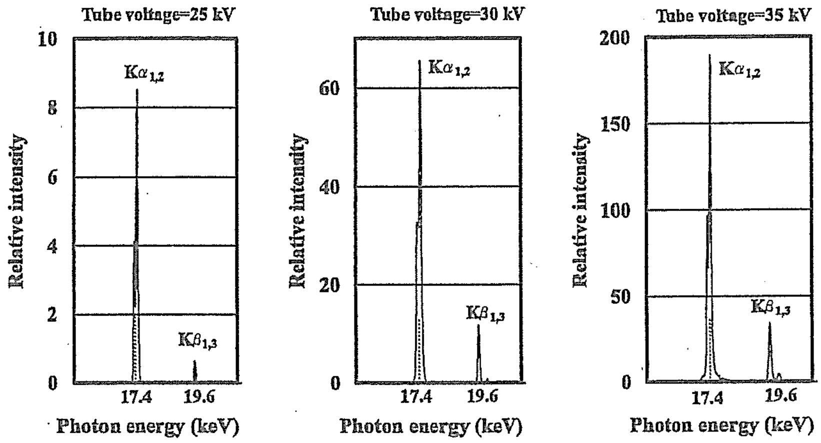

Fig. 4: X-ray spectra from the molybdenum target measured using a transmission type spectrometer with a lithium fluoride curved crystal.

increased. In this measurement, the intensity with a tube voltage of 30 kV and a current of 100 μA was 24.2 μGy/s at 1.0 m from the source.

3.2 *X-ray spectra*

First, x-ray spectra were measured using a transmission-type spectrometer with a lithium fluoride curved crystal 0.5 mm in thickness. The x-ray intensities of the spectra were detected by an imaging plate of the CR system[18] (Konica Minolta Regius 150) with a wide dynamic range, and relative x-ray intensity was calculated from Dicom original digital data corresponding to x-ray intensity. Figure 4 shows measured spectra from the molybdenum target. We observed clean K lines, while bremsstrahlung

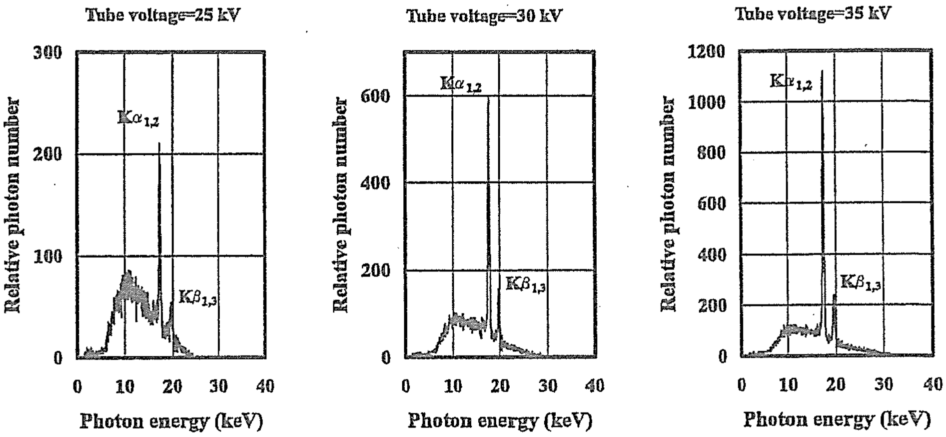

Fig. 5: X-ray spectra from the molybdenum target measured using a cadmium telluride detector.

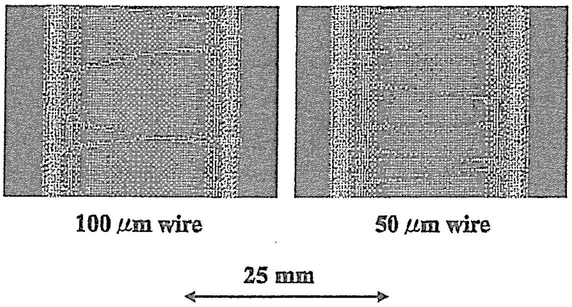

Fig. 6: Radiograms of tungsten wires of 50 and 100 μm in diameter coiled around pipes made of polymethyl methacrylate. A 50-μm-diameter wire could be observed.

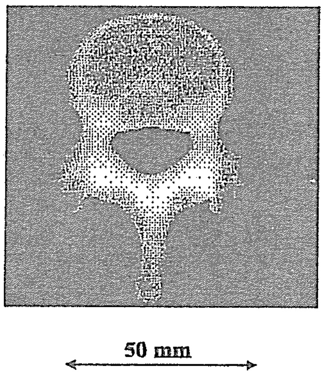

Fig. 7: Radiogram of a vertebra. Fine structure of the vertebra were visible.

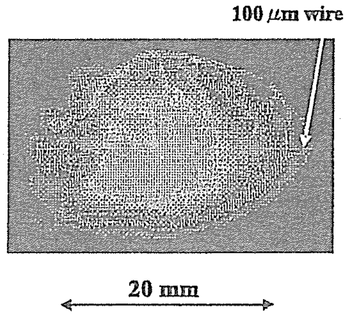

Fig. 8: Angiograms of a rabbit heart. Coronary arteries were visible.

rays were hardly detected. The characteristic x-ray intensity substantially increased with increases in the tube voltage.

The measured spectra using a cadmium telluride detector are shown in Fig. 5. Using the detector, we observed low intensity continuous x-rays. When the tube voltage was increased, both the characteristic x-ray intensity and the maximum photon energy increased.

4. Radiography

The monochromatic radiography was performed by the CR system at 1.0 m from the x-ray source with the filter, and the tube voltage was 30 kV.

First, rough measurements of image resolution were made using wires. Figure 6 shows radiograms of tungsten wires coiled around pipes made of polymethyl methacrylate (PMMA). Although the image contrast increased with increases in the wire diameter, a 50-μm-diameter wire could be observed.

A radiogram of a vertebra is shown in Fig. 7, and the fine structure of the vertebra was observed. Next, angiography was performed using iodine microspheres of 15 μm in diameter. Figure 8 shows an angiogram of a rabbit heart, and we obtained high contrast images of coronary arteries and fine blood vessels.

5. Conclusions and Outlook

In summary, we developed a new characteristic x-ray generator with a molybdenum-target tube and measured clean molybdenum K lines using the crystal spectrometer. However, continuous x-rays were detected using the detector. In both measurements, the characteristic x-ray intensity increased with increases in the tube voltage, and monochromatic Kα lines were left by a zirconium filter. Because we could measure bremsstrahlung x-rays[19] from a transmission-type molybdenum target using the crystal, the bremsstrahlung intensity was low as compared with that obtained using conventional molybdenum tubes.

In this preliminary experiment, although the maximum tube voltage and current were 36 kV and 100 μA, the voltage and current could be increased to 100 kV and 1.0 mA, respectively. Under the pulsed operation, the current can be increased to approximately 1 A without considering the target evaporation. Subsequently, the generator produced maximum number of characteristic photons was approximately 1×10^6 photons/(cm$^2 \cdot$ s) at 1.0 m from the source, and the photon count rate can be increased easily by increasing the current.

Using this x-ray generator, because it is not easy to produce high-photon-energy K-series characteristic x-rays, we are very interested in increasing the energy by changing the electrode configuration between the target, cathode, and focusing electrodes.

Acknowledgments

This work was supported by Grants-in-Aid for Scientific Research (13470154, 13877114, 16591181, and 16591222) and Advanced Medical Scientific Research from MECSST, Health and Labor Sciences Research Grants (RAMT-nano-001, RHGTEFB-genome-005 and RHGTEFB-saisei-003), Grants from the Keiryo Research Foundation, The Promotion and Mutual Aid Corporation for Private Schools of Japan, Japan Science and Technology Agency (JST), and The New Energy and Industrial Technology Development Organization (NEDO, Industrial Technology Research Grant Program in '03).

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Variations in Cerium X-ray Spectra and Enhanced K-Edge Angiography

Eiichi SATO, Etsuro TANAKA[1], Hidezo MORI[2], Toshiaki KAWAI[3], Takashi INOUE[4], Akira OGAWA[4],
Akira YAMADERA[5], Shigehiro SATO[6], Fumihito ITO[7], Kazuyoshi TAKAYAMA[8], Jun ONAGAWA[9] and Hideaki IDO[9]

Department of Physics, Iwate Medical University, 3-16-1 Honchodori, Morioka, Iwate 020-0015, Japan
[1]*Department of Nutritional Science, Faculty of Applied Bio-science, Tokyo University of Agriculture, 1-1-1 Sakuragaoka, Setagaya-ku, Tokyo 156-8502, Japan*
[2]*Department of Cardiac Physiology, National Cardiovascular Center Research Institute, 5-7-1 Fujishiro-dai, Suita, Osaka 565-8565, Japan*
[3]*Electron Tube Division #2, Hamamatsu Photonics K.K., 314-5 Shimokanzo, Iwata, Shizuoka 438-0193, Japan*
[4]*Department of Neurosurgery, School of Medicine, Iwate Medical University, 19-1 Uchimaru, Morioka 020-8505, Japan*
[5]*Department of Radiological Technology, School of Health Sciences, Hirosaki University, 66-1 Honcho, Hirosaki, Aomori 036-8564, Japan*
[6]*Department of Microbiology, School of Medicine, Iwate Medical University, 19-1 Uchimaru, Morioka 020-8505, Japan*
[7]*Digital Culture Technology Corp., Kanno The 2nd Bldg., 3-17-7 Chuo-dori, Morioka 020-0021, Japan*
[8]*Shock Wave Research Center, Institute of Fluid Science, Tohoku University, 2-1-1 Katahira, Aoba-ku, Sendai 980-8577, Japan*
[9]*Department of Applied Physics and Informatics, Faculty of Engineering, Tohoku Gakuin University, 1-13-1 Chuo, Tagajo, Miyagi 985-8537, Japan*

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A cerium-target X-ray tube is useful in performing cone-beam K-edge angiography because K-series characteristic X-rays from the cerium target are absorbed effectively by iodine-based contrast media. The X-ray generator consists of a main controller and a unit with a high-voltage circuit and a fixed anode X-ray tube. The tube is a 1.0-mm-focus diode with a cerium target and a 0.5-mm-thick beryllium window. The maximum tube voltage and current were 65 kV and 0.4 mA, respectively. Cerium $K\alpha$ rays were selected out using a barium sulfate filter, and the X-ray intensities without filtering and with a barium sulfate filter were 209 and 16.8 μGy/s, respectively, at 1.0 m from the source with a tube voltage of 60 kV and a current of 0.40 mA. Angiography was performed with an X-ray film using the filter and iodine-based microspheres 15 μm in diameter. In the angiography of nonliving animals, we observed fine blood vessels approximately 100 μm in diameter with high contrasts.
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KEYWORDS: X-ray tube, cerium target, monochromatic X-rays, $K\alpha$ rays, K-edge angiography

1. Introduction

Monochromatic parallel X-ray beams have been used to perform enhanced K-edge angiography[1–4] using iodine-based contrast media because the X-rays with photon energies of approximately 35 keV are absorbed easily by iodine with a K-edge of 33.2 keV. In conjunction with a high-resolution camera, fine blood vessels of approximately 50 μm can be observed.[3] Although the parallel beams have also been employed to perform phase-contrast radiography,[5–7] weakly absorbing materials have been observed with high contrasts.

Flash radiography of biomedical tissues has been investigated for a number of years, and several different flash X-ray generators have been developed corresponding to specific radiographic objectives.[8–11] The advantages of flash radiography include the use of K-series characteristic X-rays and their relatively good imaging contrast. However, monochromatic flash radiography[12–15] has encountered difficulties in increasing X-ray duration, and in performing X-ray computed tomography (CT).

Recently, a steady-state X-ray generator utilizing a cerium-target tube[16] has been developed, and has been employed to perform enhanced K-edge angiography achieved with cerium $K\alpha$ rays and iodine-based contrast media, since $K\alpha$ rays (34.6 keV) are absorbed effectively by iodine. In this case, because the sampling pitch of a computed radiography system (Konica Minolta Regius 150)[17] is 87.5 μm, a spatial resolution of approximately 100 μm has been obtained. Therefore, the resolution should be minimized by using a film or decreasing the pitch.

In the above-mentioned preliminary experiment,[16] we employed a cadmium tellurium detector with a photon energy resolution of 1.7 keV to measure X-ray spectra from the cerium target. However, the resolution should be minimized to measure the characteristic X-ray intensity and to confirm the K-edge effect of a barium sulfate filter for absorbing $K\beta$ and bremsstrahlung X-rays, because the photon energy width of the K-series lines is approximately 1 keV.

In the present research, we measured the X-ray spectra from a cerium target tube using a germanium detector with a photon energy resolution of 0.12 keV and performed a preliminary study on enhanced K-edge angiography achieved with cerium $K\alpha$ rays.

2. Generator

Figure 1 shows a block diagram of the X-ray generator, which consists of a main controller and an X-ray tube unit

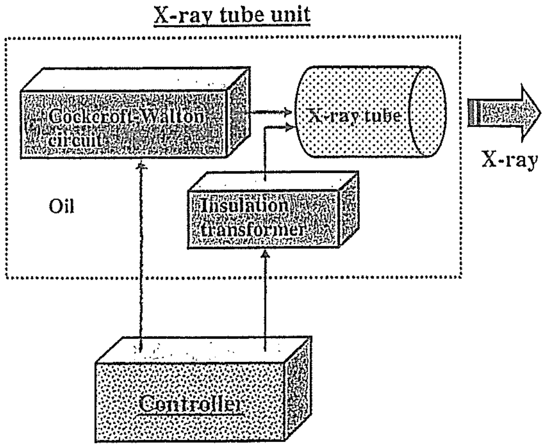


Fig. 1. Block diagram of compact X-ray generator with cerium-target radiation tube, which is used particularly for K-edge angiography using iodine-based contrast media.

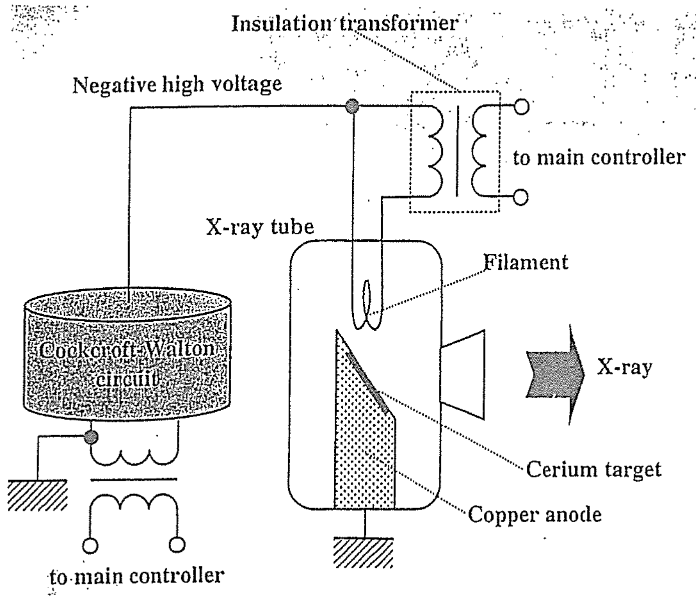
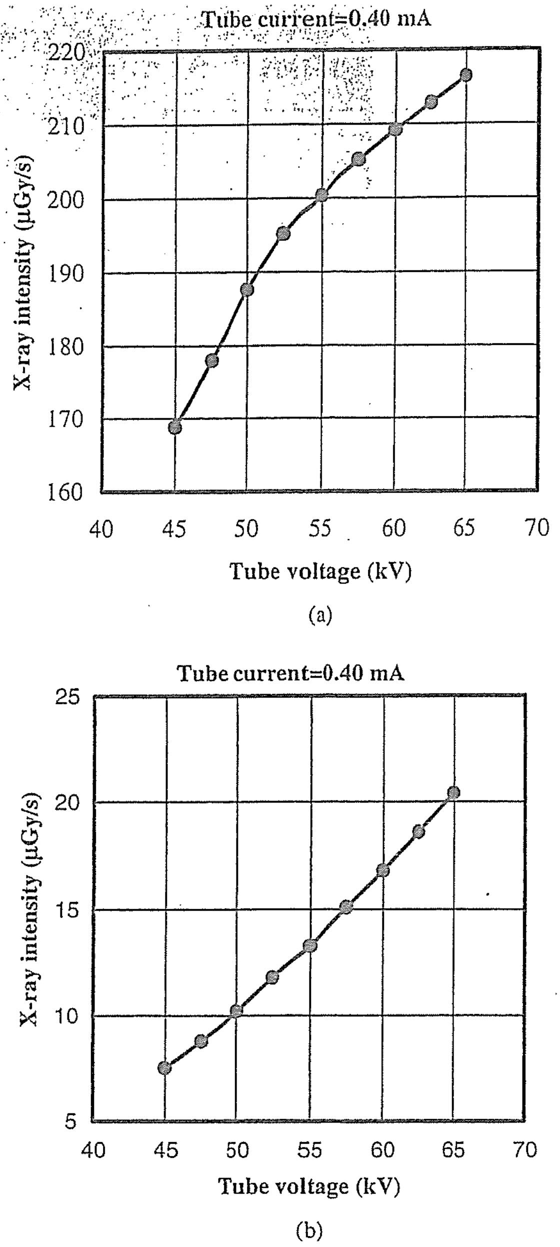


Fig. 2. Main circuit of X-ray generator.



(a)

(b)

Fig. 3. X-ray intensity measured at 1.0 m from X-ray source according to changes in tube voltage (a) without filtering and (b) using barium-sulfate filter.

with a Cockcroft–Walton circuit and a cerium-target tube. The tube voltage, the current, and the exposure time can be controlled by the controller. The main circuit for producing X-rays is illustrated in Fig. 2, and it employed the Cockcroft–Walton circuit in order to decrease the dimensions of the tube unit. In the X-ray tube, a high negative voltage is applied to the cathode electrode, and the anode (target) is connected to the tube unit case (ground potential) to cool the anode and the target effectively. The filament heating current is supplied by an AC power supply in the controller in conjunction with an insulation transformer. The tube is a conventional diode with a plate cerium target, a 1.0 mm focus, a take-off angle of 22°, and a 0.5-mm-thick beryllium window. In this experiment, the tube voltage was from 45 to 65 kV, and the tube current was regulated to within 0.40 mA (maximum current) by the filament temperature. The exposure time is controlled in order to obtain optimum X-ray intensity. Monochromatic Kα rays are selected out using a barium sulfate filter for absorbing bremsstrahlung and Kβ rays. In designing the filter, the surface density of the barium sulfate powder is important, since the X-rays are absorbed effectively by the powder as compared with poly(methyl methacrylate) (PMMA) resin. In this case, the density was approximately 30 mg/cm^2.

3. Characteristics

3.1 X-ray intensity

The X-ray intensity rate was measured by a Victoreen 660 ionization chamber at 1.0 m from the X-ray source (Fig. 3). At a constant tube current of 0.40 mA, the X-ray intensity increased when the tube voltage was increased. At a tube voltage of 60 kV and a current of 0.40 mA, the intensities without filtering and with the filter were 208 and 16.8 μGy/s, respectively, with errors of less than 0.2%. The X-ray intensity was limited because the thermal contact between the target and the anode was not good.

3.2 X-ray spectra

In order to measure X-ray spectra, we employed a germanium detector (GLP-10180/07-P, Ortec Inc.) (Fig. 4). Without filtering, when the tube voltage was increased, the X-ray intensities of cerium K-series characteristic lines increased, and both the maximum photon energy and the bremsstrahlung X-ray intensity increased. Using the filter, both the Kβ lines and the bremsstrahlung X-rays with photon energies higher than the barium K-edge of 37.4 keV were absorbed effectively, and sharp Kα lines were left. With increases in the tube voltage, the Kα intensity substantially increased, and the maximum photon energy increased.

In order to perform K-edge angiography, the Kα rays are useful, and the high-energy bremsstrahlung X-rays decrease the image contrast. Using the filter, because bremsstrahlung X-rays with energies higher than 60 keV were not absorbed easily, the tube voltage for angiography was determined to be 60 kV. Subsequently, low-energy bremsstrahlung rays

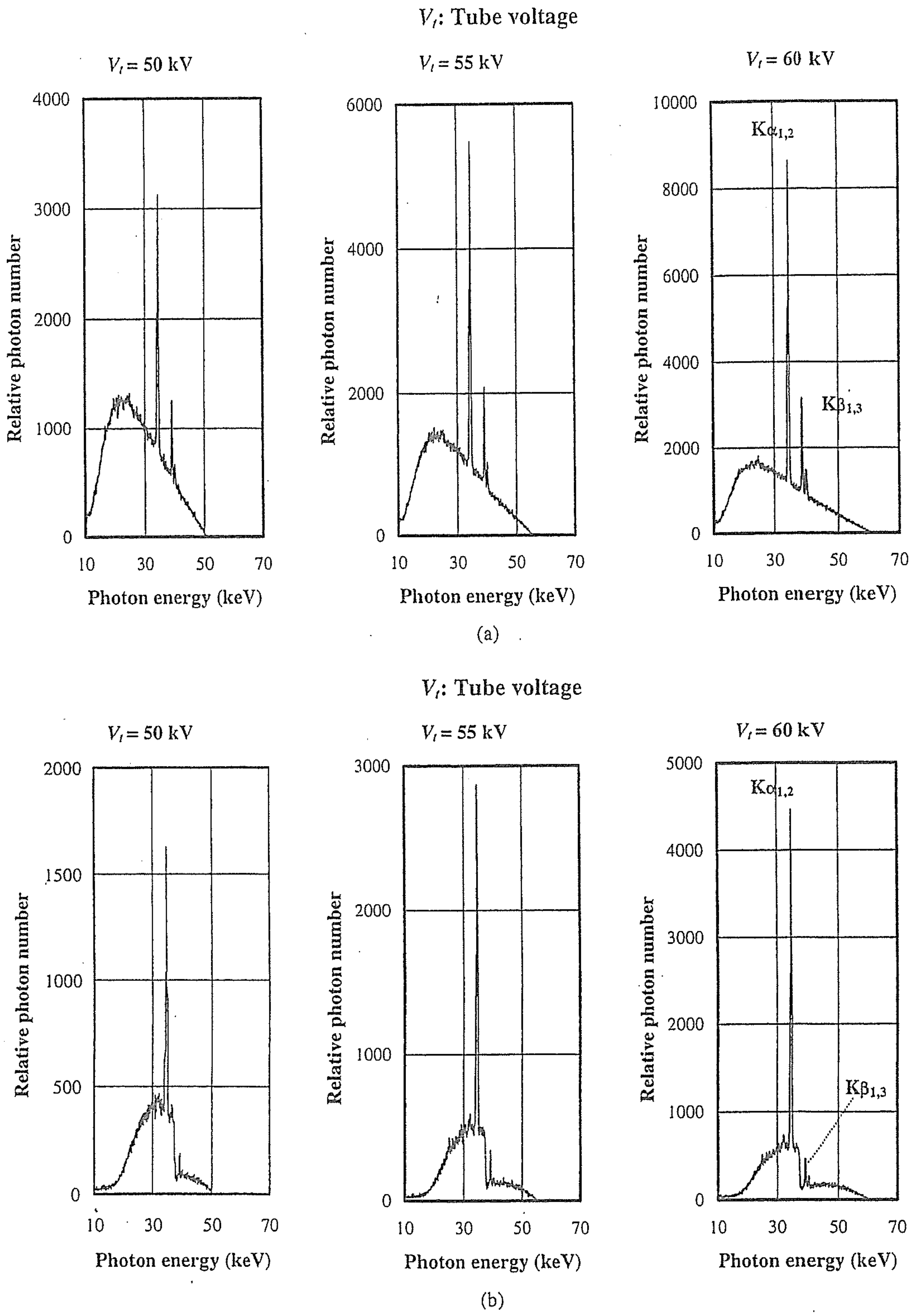


Fig. 4. X-ray spectra measured using germanium detector with changes in tube voltage (a) without filtering and (b) using barium sulfate filter.

with energies lower than the K-edge should be minimized using the filter or an aluminum filter to increase the blood-vessel contrast, since the iodine contrast media transmit the rays easily.

4. K-edge Angiography

Because the average photon energy of $K\alpha$ is 34.6 keV, iodine contrast media with a K-absorption edge of 33.2 keV absorb the $K\alpha$ lines easily. Therefore, blood vessels were observed with high contrasts. In order to observe fine blood vessels approximately 50 µm in diameter, the angiography was performed using an X-ray film (Fuji IX 100), iodine microspheres 15 µm in diameter, and the filter. The distance between the X-ray source and the imaging plate was 1.5 m, and the tube voltage was 60 kV. First, rough measurements of spatial resolution were made using wires. Figure 5 shows radiograms of tungsten wires coiled around rods made of PMMA with an X-ray exposure time of 300 s. Although the

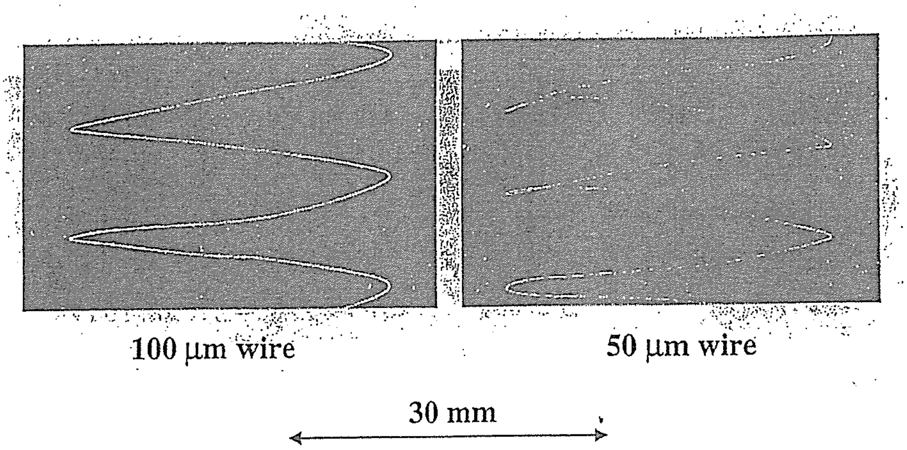


Fig. 5. Radiograms of tungsten wires coiled around PMMA rods.

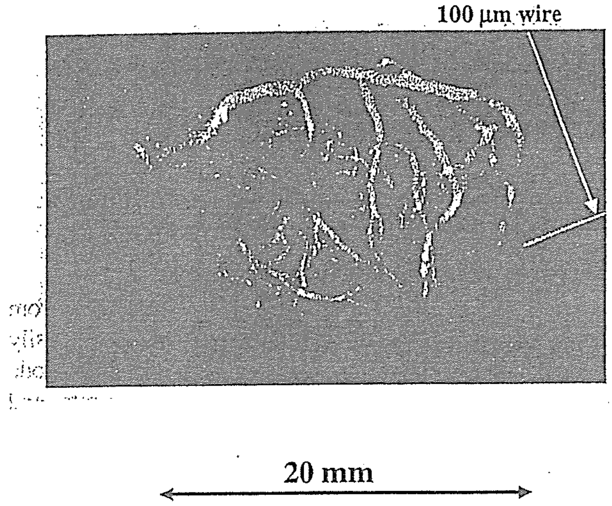


Fig. 6. Angiograms of extracted rabbit heart using iodine microspheres.

image contrast hardly varied with decreases in the wire diameter, a 50-µm-diameter wire could be observed clearly.

Figures 6 and 7 show angiograms of a rabbit heart and a thigh, respectively, with an exposure time of 300 s. The coronary arteries in the heart and fine blood vessels in the thigh with diameters of approximately 100 µm were visible. Figure 8 shows an angiogram of a dog heart in a 100-mm-thick water phantom with an exposure time of 1,500 s. Because the size of the dog heart is almost the same as that of a human heart, human coronary arteries can be observed. For comparison, we show a three-dimensional (3D) image of the coronary arteries constructed from X-ray CT images taken by Pascal (Digital Culture Tech. Corp.) with a tungsten X-ray tube (Fig. 9). This heart was the same as that used in K-edge angiography and was observed from the same direction by rotating the three-dimensional (3D) image; CT angiography was performed without using the water phantom. Using this 3D angiography achieved with a multislice helical CT, fine blood vessels were not observed at all.

5. Discussion

In the present research, we employed an X-ray generator with a cerium-target tube and succeeded in producing cerium characteristic X-rays, which can be absorbed easily by iodine-based contrast media. Both the characteristic and bremsstrahlung X-ray intensities increased with tube voltage

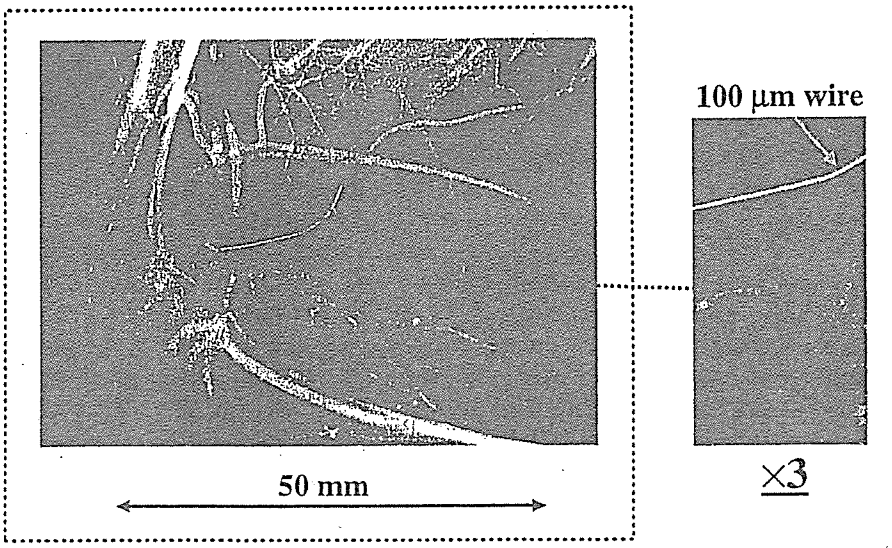


Fig. 7. Angiogram of rabbit thigh using iodine microspheres.

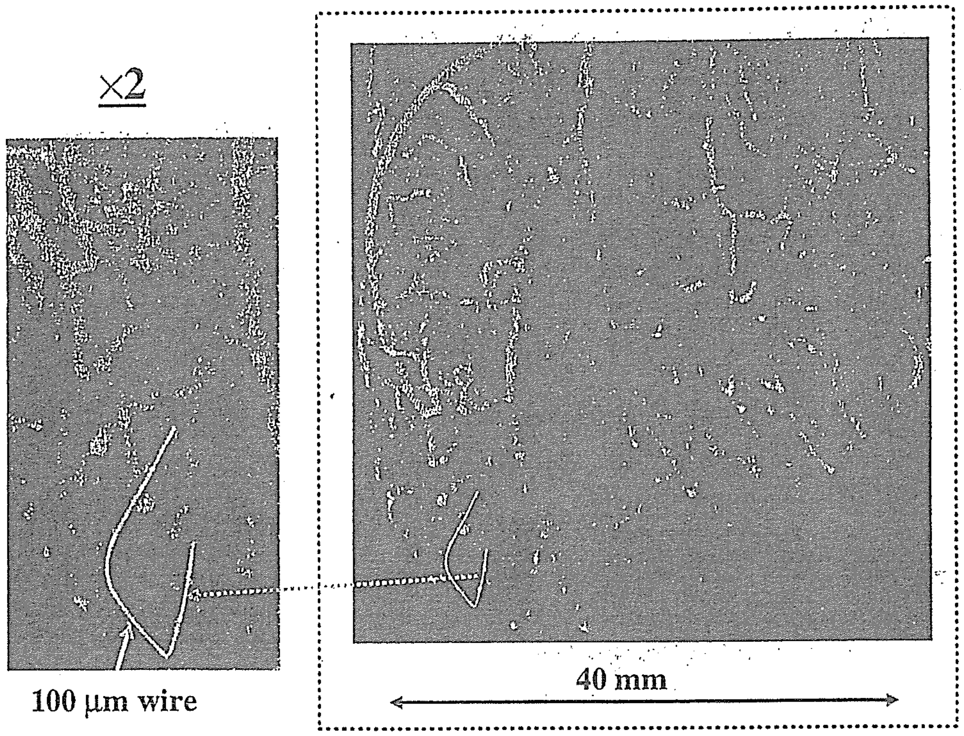


Fig. 8. Angiogram of extracted dog heart in 100-mm-thick water phantom using iodine microspheres.

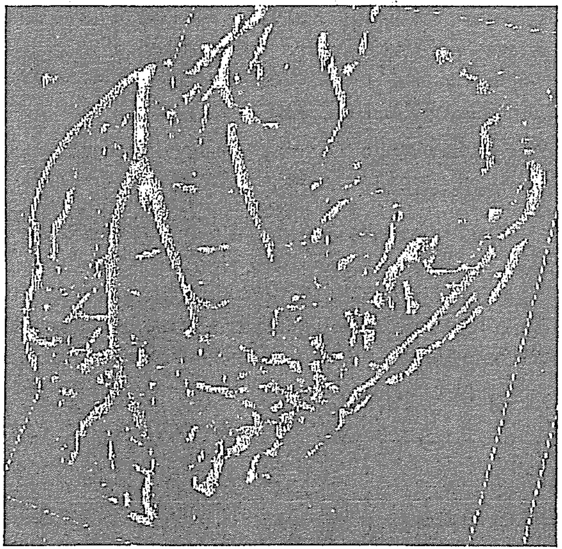


Fig. 9. Three-dimensional image of coronary arteries constructed from X-ray CT images taken by Pascal.

without filtering. Using the filter to absorb Kβ and bremsstrahlung X-rays, Kα rays were left, and we performed K-edge angiography using the filter with a tube voltage of 60 kV. To produce clean characteristic X-rays without using the filter, the angle dependence of the bremsstrahlung intensity should be considered, since bremsstrahlung rays are not emitted in the direction opposite that of the electron trajectory in Sommerfeld's theory.[18]

Currently, angiography is performed using both the bremsstrahlung and characteristic X-rays produced from a tungsten X-ray tube. However, enhanced K-edge angiography in this work was primarily performed using cerium Kα rays. Using the filter, the maximum number of Kα photons was approximately 3×10^7 photons/(cm^2·s) at 1.0 m from the source, and the photon count rate can be increased easily by improving the target. For example, a new rotation anode tube has been designed to increase the X-ray dose rate, and the rate can be increased by increasing the anode diameter.

In energy-selective imaging including K-edge angiography, the filtering effect of the absorber should be considered, and the X-ray spectra using the filter at a tube voltage of 60 kV hardly varies with changes in the thickness of the water phantom according to the spectrum estimation. Due to the absorption coefficient, Kβ rays are also useful for angiography, and both the Kα and Kβ rays can be left using a cerium oxide filter with a surface density of approximately 10 mg/cm^2. In addition, an aluminum filter with a thickness of approximately 3.0 mm is useful in absorbing unnecessary bremsstrahlung X-rays with energies lower than the K-absorption edge.

Acknowledgment

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Energy-selective high-speed radiography utilizing stroboscopic x-ray generator

Eiichi Sato*[a], Etsuro Tanaka[b], Hidezo Mori[c], Toshiaki Kawai[d], Shigehiro Sato[e], Hidenori Ojima[f],
Kazuyoshi Takayama[f] and Hideaki Ido[g]

[a] Department of Physics, Iwate Medical University, 3-16-1 Honchodori, Morioka 020-0015, Japan
[b] Department of Nutritional Science, Faculty of Applied Bio-science, Tokyo University of Agriculture, 1-1-1 Sakuragaoka, Setagaya-ku, 156-8502, Japan
[c] Department of Cardiac Physiology, National Cardiovascular Center Research Institute, 5-7-1 Fujishirodai, Suita, Osaka 565-8565, Japan
[d] Electron Tube Division #2, Hamamatsu Photonics K. K., 314-5 Shimokanzo, Toyooka Village, Iwata-gun 438-0193, Japan
[e] Department of Microbiology, School of Medicine, Iwate Medical University, 19-1 Uchimaru, Morioka 020-8505, Japan
[f] Shock Wave Research Center, Institute of Fluid Science, Tohoku University, 2-1-1 Katahira, Sendai 980-8577, Japan
[g] Department of Applied Physics and Informatics, Faculty of Engineering, Tohoku Gakuin University, 1-13-1 Chuo, Tagajo 985-8537, Japan

ABSTRACT

Energy-selective high-speed radiography utilizing a kilohertz-range stroboscopic x-ray generator and its application to high-speed angiography are described. This generator consists of the following major components: a main controller, a condenser unit with a Cockcroft-Walton circuit, and an x-ray tube unit in conjunction with a grid controller. The main condenser of about 500 nF in the unit is charged up to 100 kV by the circuit, and the electric charges in the condenser are discharged to the triode by the grid control circuit. Although the tube voltage decreased during the discharging for generating x rays, the maximum value was equal to the initial charging voltage of the main condenser. The maximum tube current and the repetition rate were approximately 0.5 A and 32 kHz, respectively. The x-ray pulse width ranged from 0.01 to 1.0 ms, and the maximum shot number had a value of 32. At a charging voltage of 80 kV and a width of 1.0 ms, the x-ray intensities obtained without filtering, using an aluminum filter, and using a barium sulfate filter were 14.8, 5.48 and 5.05 μGy per pulse, respectively, at 1.0 m, and the dimensions of the focal spot had values of 3.5×3.5 mm. Angiography was performed using both the aluminum and the barium sulfate filters at a charging voltage of 60 kV.

Keywords: energy-selective radiography, bremsstrahlung x rays, filtering, stroboscopic x-ray, pulse x-ray, angiography

1. INTRODUCTION

Modern high-speed x-ray generators are capable of producing short x-ray pulses with high dose rates, and have been applied to radiography in various fields. To produce hard flash x rays with maximum photon energies of approximately 1 MeV, multistage Marx surge generators have been developed.[1] Furthermore, induction linear accelerators[2] have been developed and improved to produce 10-MeV-order flash x rays. In contrast, 100-kV-order flash x-ray generators have been developed and applied to biomedicine.[3,4]

In the cases of multiple-shot and cine radiographies, we have developed several different repetitive-flash[5-8] and stroboscopic x-ray generators.[9-11] Although most flash x-ray generators have cold-cathode tubes, the stroboscopic generators utilize hot-cathode tubes. Particularly, although a 50 kHz stroboscopic generators have been manufactured, the repetition rate can be increased to MHz order.

Recently synchrotrons generate monochromatic parallel x-ray beams using a monochromator, and these beams have been employed to perform enhanced K-edge angiography[12,13] and x-ray phase imaging.[14,15] To perform angiography, the beams with photon energies of approximately 35 keV have been used, because iodine contrast mediums with a

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K-absorption edge of 33.155 keV absorb the beams effectively. In view of this situation, we have developed x-ray generators with cerium-target tubes[16,17] which can produce Kα rays of 34.6 keV.

In this research, we employed a tungsten-target x-ray tube and performed a preliminary study on high-speed angiography achieved with quasi-monochromatic x rays produced by filtering in conjunction with a computed radiography system.

2. GENERATOR

Figure 1 shows the block diagram of the kilohertz-range stroboscopic x-ray generator. This generator consists of the following major components: a main controller, a condenser unit with a Cockcroft-Walton circuit, and an x-ray tube unit in conjunction with a grid controller (Figs. 2 and 3). The main condenser of about 500 nF in the unit is charged up to 100 kV by the circuit, and the electric charges in the condenser are discharged to the triode by the grid control circuit. Although the tube voltage decreased during the discharging for generating x rays, the maximum value was equal to the initial charging voltage of the main condenser.

The x-ray tube is a glass-enclosed hot-cathode triode and is composed of the following major parts: an anode rod made of copper, a tungsten plate target, an iron focusing electrode, a tungsten hot cathode (filament), a tungsten grid, and a glass tube body. The electron beams from the cathode are accelerated between the anode and cathode electrodes and are converged to the target by the focusing electrode. The tube is set in the metal case filled with insulation oil, and the diaphragm regulates the radiation field.

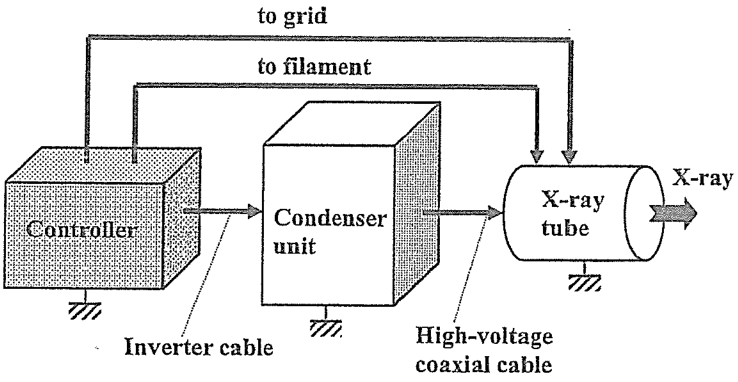


Figure 1: Block diagram of kilohertz-range stroboscopic x-ray generator.

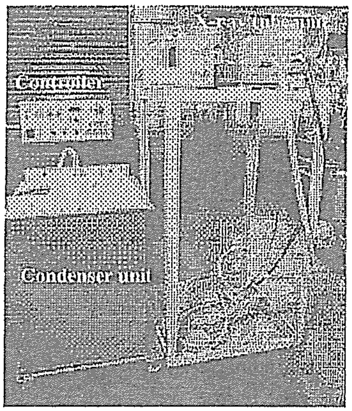


Figure 2: Stroboscopic x-ray generator.

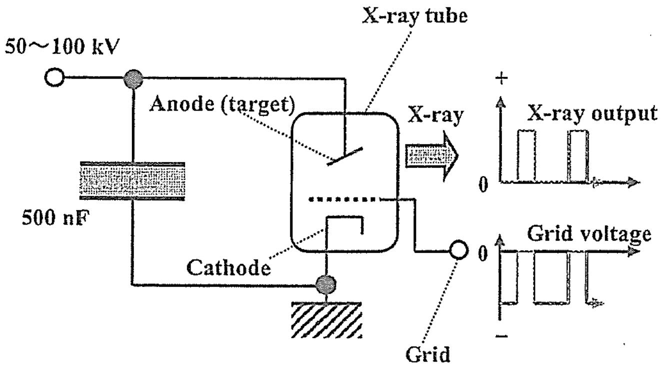


Figure 3: Main high-voltage circuit of x-ray generator.

3. CHARACTERISTICS

3.1 X-ray output

The x-ray output was detected by a pin diode, and the output voltages from the diode were measured by a digital storage scope (Fig. 4). Using this generator, the pulse width could be controlled correctly and ranged from 10 μs to 1.0 ms. The maximum repetition rate was approximately 50 kHz, and stable repetitive x-ray pulses were obtained. When the charging voltage was increased, the pulse height increased substantially.

3.2 Time-integrated x-ray intensity

Figure 5 shows the time-integrated (absolute) value of the x-ray intensity (exposure) at 1.0 m per pulse measured by a Victoreen 660 ionization chamber. The intensity was proportional to the driving pulse width. At a constant pulse width of 1.0 ms, the intensity increased in proportion to approximately the second power of the charging voltage. At a charging voltage of 80 kV and a width of 1.0 ms, the x-ray intensity obtained without filtering, using an aluminum filter, and using a barium sulfate filter were 14.8, 5.48, and 5.05 μGy per pulse, respectively, at 1.0 m from the source.

3.3 X-ray source

The image of the x-ray source was measured using a pinhole camera with a hole diameter of 50 μm and a computed radiography (CR) system (Konica Regius 150)[18] with a sampling pitch of 87.5 μm. When the charging voltage was increased, the dimensions hardly varied, and were approximately 3.5×3.5 mm.

3.4 X-ray spectra

In order to measure x-ray spectra, we employed a cadmium tellurium detector (CDTE2020X, Hamamatsu Photonics Inc.) (Fig. 6). Compared with a germanium detector, this detector has a lower energy resolution of 1.7 keV.

When the charging voltage was increased, both the maximum photon energy and the intensities of bremsstrahlung x rays increased, and the photon energy of the spectrum peak also increased. The 3-mm-thick aluminum filter attenuated the low-photon-energy bremsstrahlung x rays. Subsequently, the barium sulfate filter, with a surface density of approximately 10 mg/cm^2, significantly attenuated the spectra above the barium K-edge of 37.4 keV. The areas under the spectral curves correlate closely to the total x-ray intensities shown in Fig. 4.

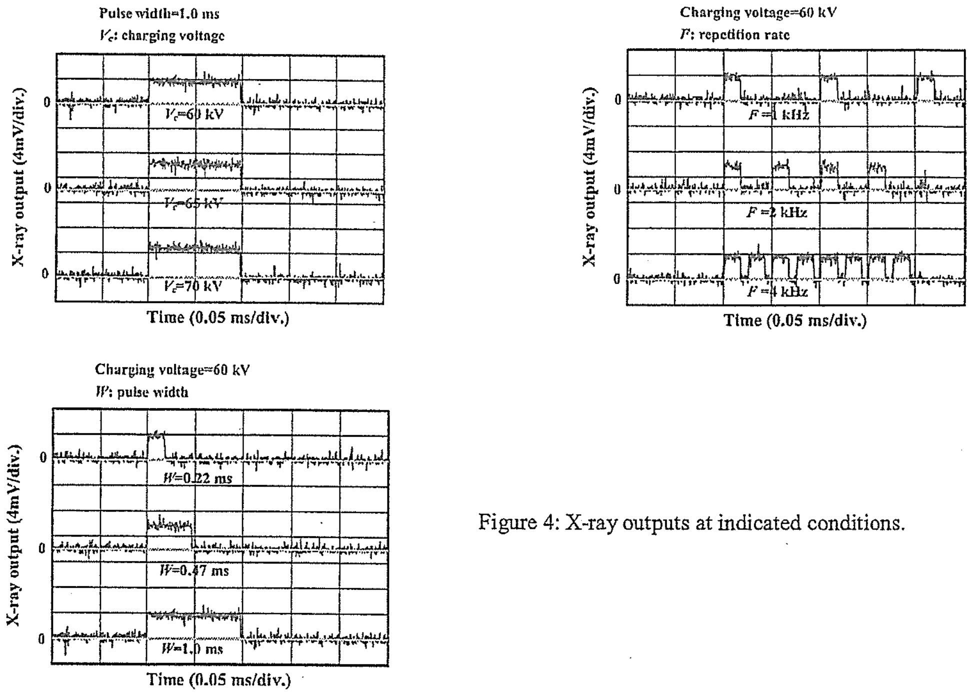
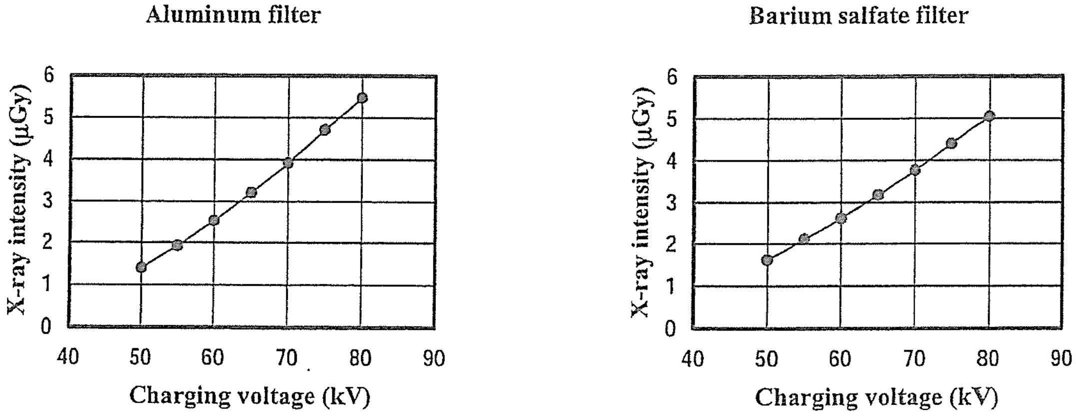


Figure 4: X-ray outputs at indicated conditions.



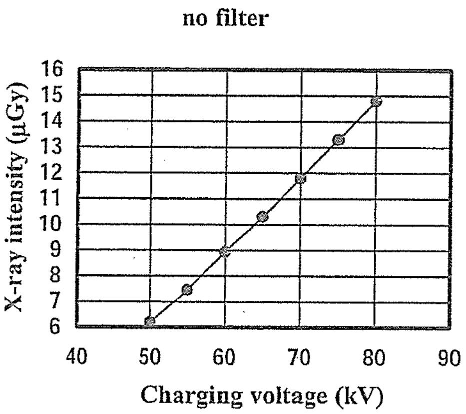


Figure 5: X-ray intensities at 1.0 m per pulse with changing charging voltage with exposure time of 1.0 ms.

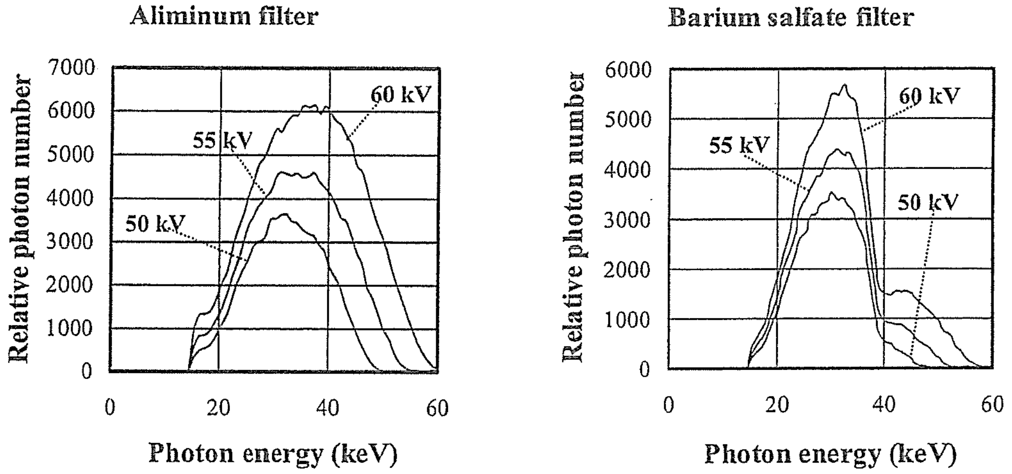


Figure 6: X-ray spectra at indicated conditions.