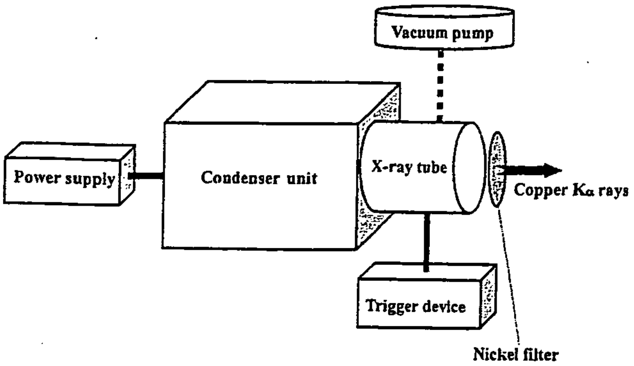


Fig. 1 Block diagram of high-intensity plasma flash x-ray generator.

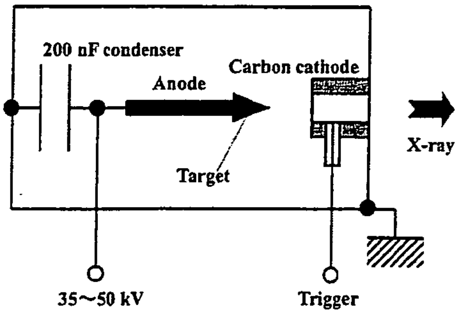


Fig. 2 Circuit diagram of generator.

generator as a trigger device, and a flash x-ray tube. In this generator, a low-impedance transmission line (Fig. 2) is employed to increase maximum tube current. The high-voltage main condenser is charged to 50 kV by the power supply, and electric charges in the condenser are discharged to the tube after triggering the cathode electrode with the trigger device. The plasma flash x-rays are then produced.

2.2 *X-Ray Tube*

The x-ray tube is a demountable cold cathode triode that is connected to the turbomolecular pump with a pressure of approximately 1 mPa (Fig. 3). This tube consists of the following major parts: a hollow cylindrical carbon cathode with a bore diameter of 10.0 mm, a brass focusing electrode, a trigger electrode made from copper wire, a stainless-steel vacuum chamber, a nylon insulator, a polyethylene terephthalate (Mylar) x-ray window 0.25 mm thick, and a rod-shaped copper target 3.0 mm in diameter with a tip angle of 60 deg. The distance between the target and cathode electrodes is approximately 20 mm, and the trigger electrode is set in the cathode electrode. As electron beams from the cathode electrode are roughly converged to the target by the focusing electrode, evaporation leads to the formation of a weakly ionized linear plasma, consisting of copper ions and electrons, around the fine target.

2.3 *Principle of Clean Kα-Ray Irradiation*

In the linear plasma, bremsstrahlung photons with energies higher than the K-absorption edge are effectively absorbed and are converted into fluorescent x-rays (Fig. 4). The plasma then transmits the fluorescent rays easily, and bremsstrahlung rays with energies lower than the K edge are also absorbed by the plasma. In addition, because bremsstrahlung rays are not emitted in the opposite direction to that of electron acceleration, intense characteristic x-rays are generated from the plasma-axial direction. Sub-

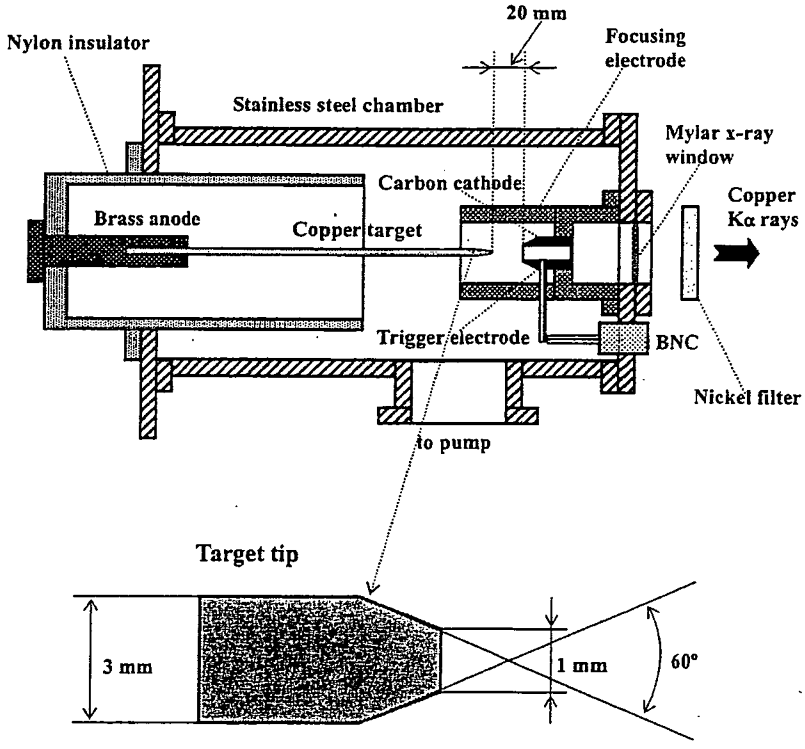


Fig. 3 Schematic drawing of flash x-ray tube with rod target.

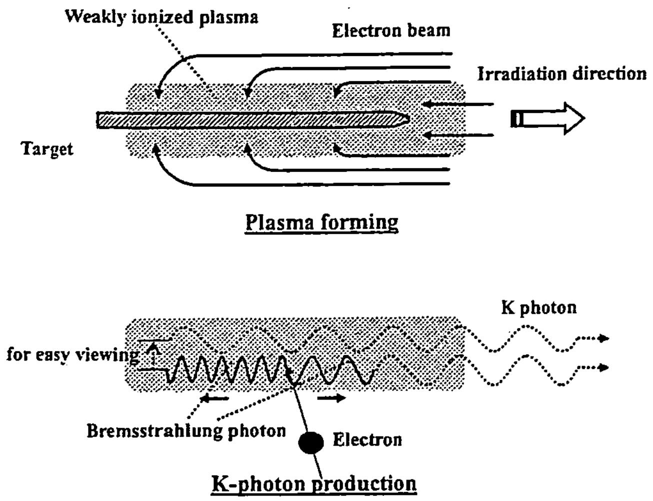


Fig. 4 K-photon irradiation from weakly ionized plasma.

sequently, Kβ rays (8.90 keV) are absorbed effectively using a 10-μm-thick nickel K-edge filter with an edge of 8.33 keV, and quite clean Kα rays (8.04 keV) are produced.

3 Characteristics

3.1 *Tube Voltage and Current*

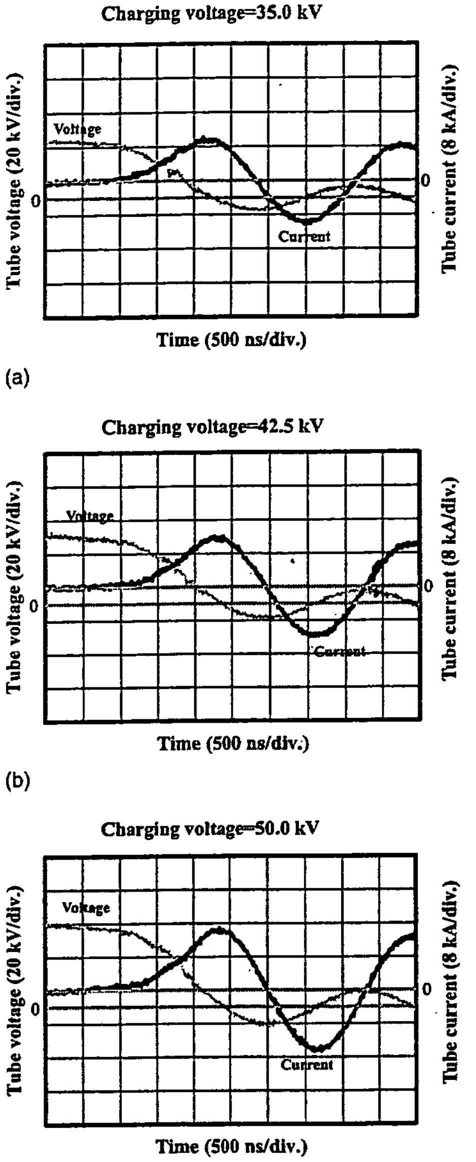
Tube voltage and current were measured by a high-voltage divider with an input impedance of 1 GΩ and a current transformer, respectively. Figure 5 shows the time relation for the tube voltage and current. At the indicated charging voltages, they roughly displayed damped oscillations. When the charging voltage was increased, both the maximum tube voltage and current increased. At a charging voltage of 50 kV, the maximum tube voltage was almost equal to the charging voltage of the main condenser, and the maximum tube current was approximately 15 kA.

3.2 *X-Ray Output*

An x-ray output pulse was detected using a combination of a plastic scintillator and a photomultiplier using a 10-μm-thick monochromatic copper filter (Fig. 6). The x-ray pulse height substantially increased with corresponding increases in the charging voltage. The x-ray pulse widths were about 700 ns, and the time-integrated x-ray intensity per pulse measured by a thermoluminescence dosimeter (Kyokko TLD Reader 1500 utilizing MSO-S elements without energy compensation) had a value of about 20 μC/kg at 1.0 m from the x-ray source, with a charging voltage of 50 kV.

3.3 *X-Ray Source*

To measure images of the Kα source, we employed a pinhole camera with a hole diameter of 100 μm (Fig. 7). When the charging voltage was increased, the plasma x-ray source grew, and both spot dimension and intensity increased. Because the x-ray intensity is the highest at the center of the spot, both the dimension and intensity decreased according to both increases in the thickness of a filter for absorbing x-rays and decreases in the pinhole diameter.



(a)

(b)

(c)

Fig. 5 Tube voltages and currents with charging voltage of (a) 35.0 kV, (b) 42.5 kV, and (c) 50.0 kV.

3.4 *X-Ray Spectra*

X-ray spectra from the plasma source were measured using a transmission-type spectrometer with a lithium fluoride

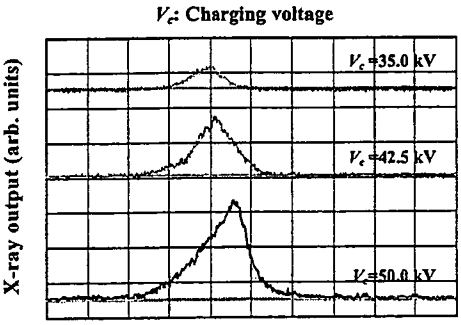


Fig. 6 X-ray outputs measured by plastic scintillator with changes in charging voltage.

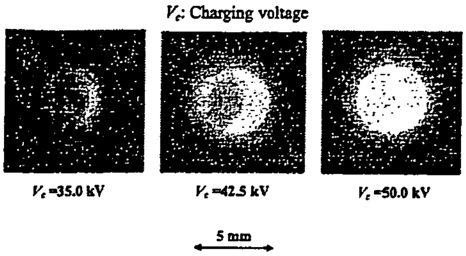


Fig. 7 Images of Kα x-ray source measured by pinhole of 100 μm from plasma axial direction.

curved crystal 0.5 mm in thickness. The spectra were taken by a computed radiography (CR) system[21] (Konica Regius 150) with a wide dynamic range, using the filter, and relative x-ray intensity was calculated from Dicom digital data. Figure 8 shows measured spectra from the copper target using the filter. In fact, we observed clean Kα lines such as lasers, and confirmed the significant filtering effect, while bremsstrahlung rays were hardly detected at all. The characteristic x-ray intensity of the Kα lines substantially increased with corresponding increases in the charging voltage, and the Kβ line was absorbed by the filter. Although this spectrometer has sufficient energy resolution for measuring $K\alpha_1$ and $K\alpha_2$ lines, we could observe only a single line.

4 Radiography

Plasma radiography was performed by the CR system without using the filter, and the distance between the x-ray source and imaging plate was 1.2 m.

First, rough measurements of image resolution were made using wires. Figure 9 shows radiograms of 50-μm-diam tungsten wires coiled around a pipe, and a rod made of polymethyl methacrylate with a charging voltage of 50 kV. Although the image contrast increased using the pipe, 50-μm-diam wires could be observed.

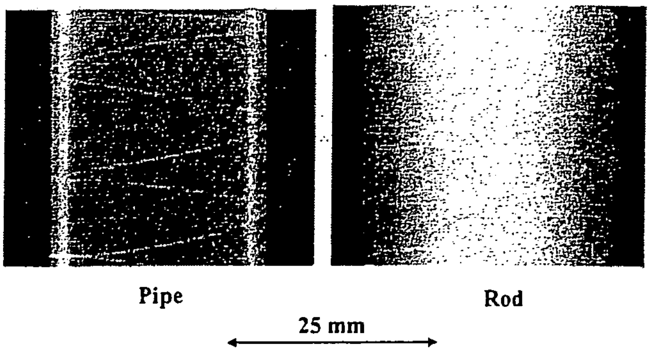


Fig. 9 Radiograms of tungsten wires 50 μm in diameter coiled around pipe, and rod made of polymethyl methacrylate.

The image of water falling into a polypropylene beaker from a glass test tube is shown in Fig. 10. This image was taken with a charging voltage of 45 kV, with the slight addition of an iodine-based contrast medium. Because the x-ray duration was about 1 μs, the stop-motion image of water could be obtained.

Figure 11 shows an angiogram of a rabbit heart; iodine-based microspheres of 15 μm in diameter were used with a charging voltage of 50 kV, and fine blood vessels of about 100 μm were visible.

5 Discussion

Concerning the spectrum measurement, we obtained fairly clean Kα lines from a weakly ionized linear plasma x-ray

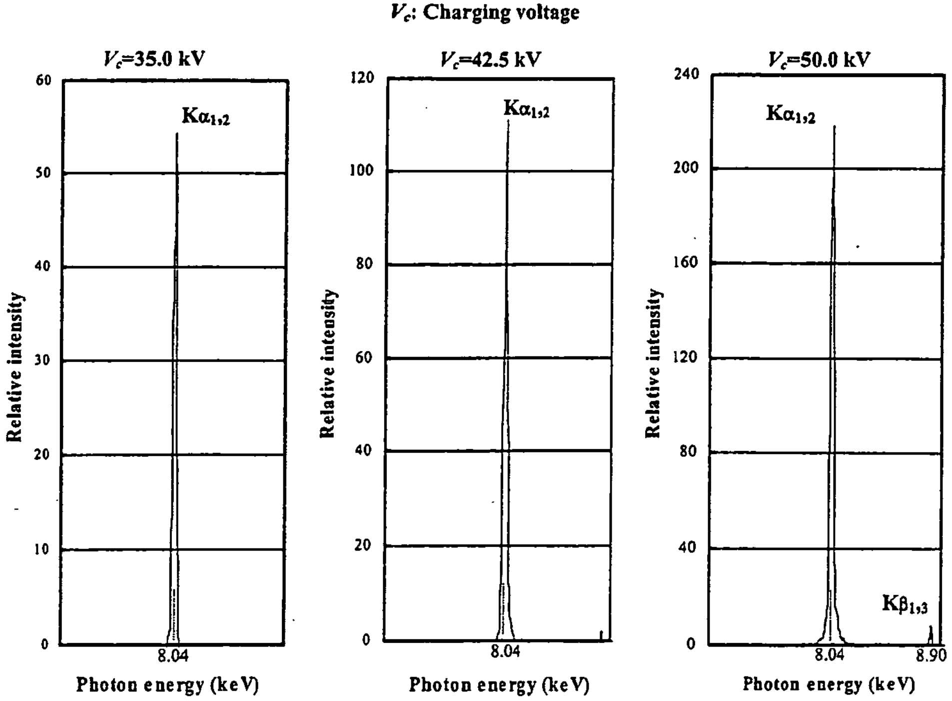


Fig. 8 X-ray spectra from weakly ionized copper plasma according to changes in charging voltage and to insertion of nickel K-edge filter.

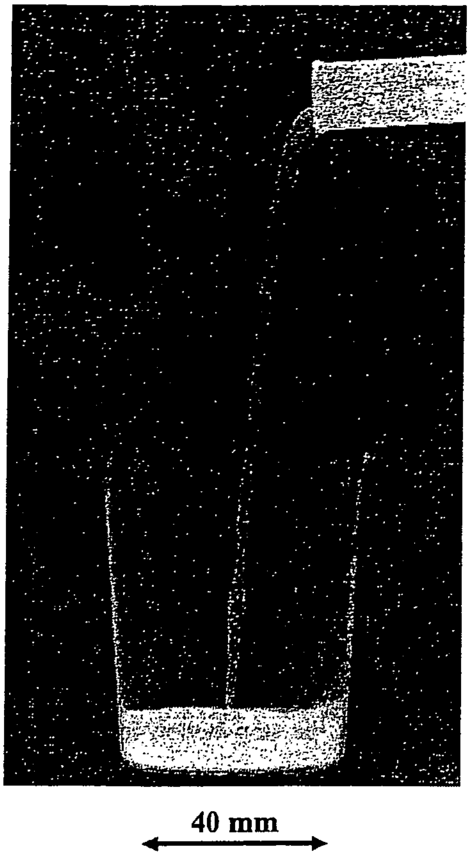


Fig. 10 Radiogram of water falling into polypropylene beaker from glass test tube.

source by absorbing Kβ lines using the K-edge filter. The lines are produced by x-ray enhancement by spontaneous emission, and the coherence can be increased by development of a resonator or by pulse laser irradiations from the plasma axial direction to produce higher harmonics. In a medical application, cerium Kα rays (34.6 keV) are absorbed effectively by an iodine-based contrast medium, and high contrast microangiography can be performed.

In this research, we obtained sufficient characteristic x-ray intensity per pulse for CR radiography using the filter, and the generator-produced number of characteristic Kα photons was approximately 5×10^{13} photons/cm$^2\cdot$s at 1.0 m from the source. In addition, since the photon energy of characteristic x-rays can be controlled by changing the target elements, various quasimonochromatic high-speed radiographies, such as high-contrast microangiography and parallel radiography using an x-ray lens, will be possible.

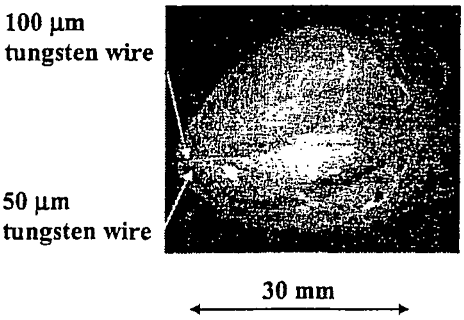


Fig. 11 Angiograms of rabbit heart.

Acknowledgments

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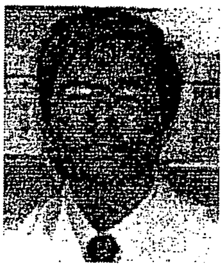

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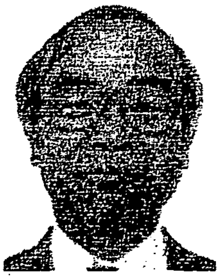

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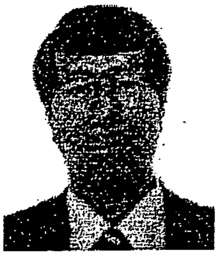

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Monochromatic flash x-ray generator utilizing a disk-cathode silver tube

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Abstract. The high-voltage condensers in a polarity-inversion two-stage Marx surge generator are charged from −50 to −70 kV by a power supply, and the electric charges in the condensers are discharged to an x-ray tube after closing gap switches in the surge generator with a trigger device. The x-ray tube is a demountable diode, and the turbomolecular pump evacuates air from the tube with a pressure of approximately 1 mPa. Clean silver Kα lines are produced using a 30-μm-thick palladium filter, since the tube utilizes a disk cathode and a rod target, and bremsstrahlung rays are not emitted in the opposite direction to that of electron acceleration. At a charging voltage of −70 kV, the instantaneous tube voltage and current are 90 kV and 0.8 kA, respectively. The x-ray pulse widths are approximately 80 ns, and the instantaneous number of generator-produced Kα photons is approximately 4×10^7photons/cm^2 per pulse at 0.3 m from the source 3.0 mm in diameter. *© 2005 Society of Photo-Optical Instrumentation Engineers.* [DOI: 10.1117/1.2049248]

Subject terms: x-ray source; x-ray tube; x-ray spectra; rapid imaging; x-ray beam filtration; monochromatic x-rays.

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1 Introduction

Energy-selective monochromatic radiography is a useful method for medical radiography, and quasimonochromatic x-rays have been produced using a K-edge filter when conventional medical x-ray tubes are employed. In contrast, monochromatic parallel x-ray beams have been produced using synchrotrons in conjunction with silicon single crystals, and have been applied effectively to enhanced K-edge angiography,[1–3] achieved with x-rays with a photon energy of approximately 35 keV.

In high-speed radiography, we have developed several different flash x-ray generators[4–9] utilizing cold cathode x-ray tubes, and intense and clean K-series characteristic x-rays have been produced from the axial direction of weakly ionized linear plasma[10–14] of nickel and copper using a plasma flash x-ray generator. In the plasma, bremsstrahlung spectra with photon energies of higher than the K-absorption edge are effectively absorbed and are converted into fluorescent x-rays. The plasma then transmits the fluorescent rays easily. Subsequently, the photon ener-

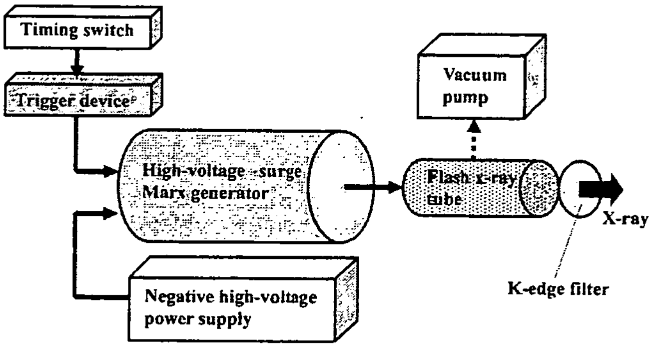

Fig. 1 Block diagram of the compact quasimonochromatic flash x-ray generator.

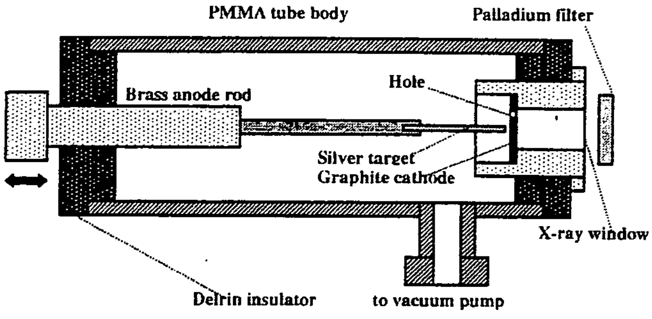

Fig. 3 Schematic drawing of the flash x-ray tube.

gies of the characteristic x-rays are determined by the target element, and the tube voltage should be increased in cases where the high-photon-energy characteristic x-rays are selected. However, it is difficult to increase the photon energies, since the maximum tube voltage is approximately 60 kV without using a high-voltage gap switch.

To increase the maximum tube voltage, a multistage surge Marx generator[15,16] is useful, because the output voltage is equal to the value of the condenser charging voltage multiplied by the stage number. In addition, characteristic x-rays can be produced by considering the angle dependence of bremsstrahlung x-rays when a cold cathode diode, in conjunction with the surge generator, is employed.

In this study, we developed a compact flash x-ray generator utilizing a silver-target radiation tube, and used it to perform a preliminary experiment for producing clean monochromatic x-rays.

2 Experimental Setup

2.1 *High-Voltage Circuit*

Figure 1 shows a block diagram of a compact monochromatic flash x-ray generator. This generator consists of the following components: a constant high-voltage power supply, a surge Marx generator with a capacity during main discharge of 425 pF, a thyratron trigger device for the surge generator, a turbomolecular pump, and a flash x-ray tube. Since the electric circuit of the high-voltage pulse generator employs a polarity-inversion two-stage Marx line[13,14] (Fig. 2), the surge generator produces twice the potential of the condenser charging voltage. When two condensers inside of the surge generator are charged from −50 to −70 kV, the ideal output voltage ranges from 100 to 140 kV.

2.2 *X-ray Tube*

The x-ray tube is a demountable diode type, as illustrated in Fig. 3. This tube is connected to the turbomolecular pump with a pressure of approximately 1 mPa and consists of the following major devices: a rod-shaped silver target 3.0 mm in diameter, a disk cathode made of graphite, a polyethylene terephthalate (Mylar) x-ray window 0.25 mm in thickness, and a polymethyl methacrylate (PMMA) tube body. The target-cathode space was regulated to 1.0 mm from the outside of the x-ray tube by rotating the anode rod, and the transmission x-rays are obtained through a 1.0-mm-thick graphite cathode and an x-ray window. In Sommerfeld's theory,[17] because bremsstrahlung rays are not emitted in the opposite direction to that of electron trajectory (Fig. 4), silver $K\alpha$ rays can be produced using a 30-μm-thick palladium K-edge filter. In the K-series characteristic x-ray irradiation, $K\alpha$ rays are left by absorbing $K\beta$ rays to perform the preliminary experiment for producing clean monochromatic x-rays and to confirm the filtering effect.

3 Results and Discussion

3.1 *Tube Voltage and Current*

Tube voltage and current were measured using a high-voltage divider with an input impedance of 10 kΩ and a

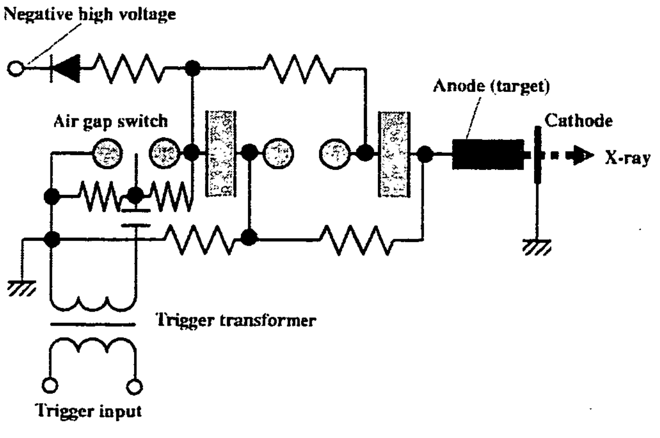

Fig. 2 Circuit diagram of the flash x-ray generator.

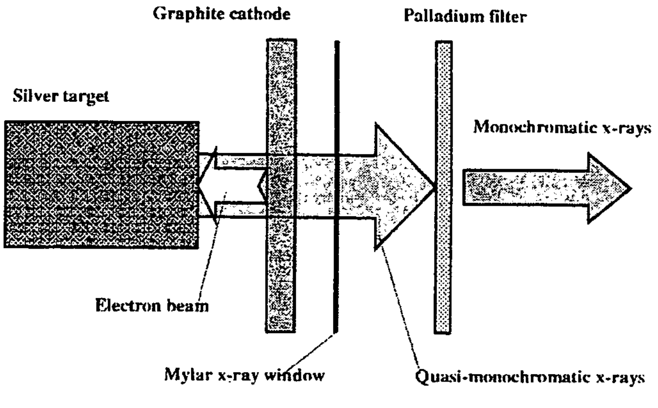

Fig. 4 Irradiation of silver $K\alpha$ rays using a palladium K-edge filter.

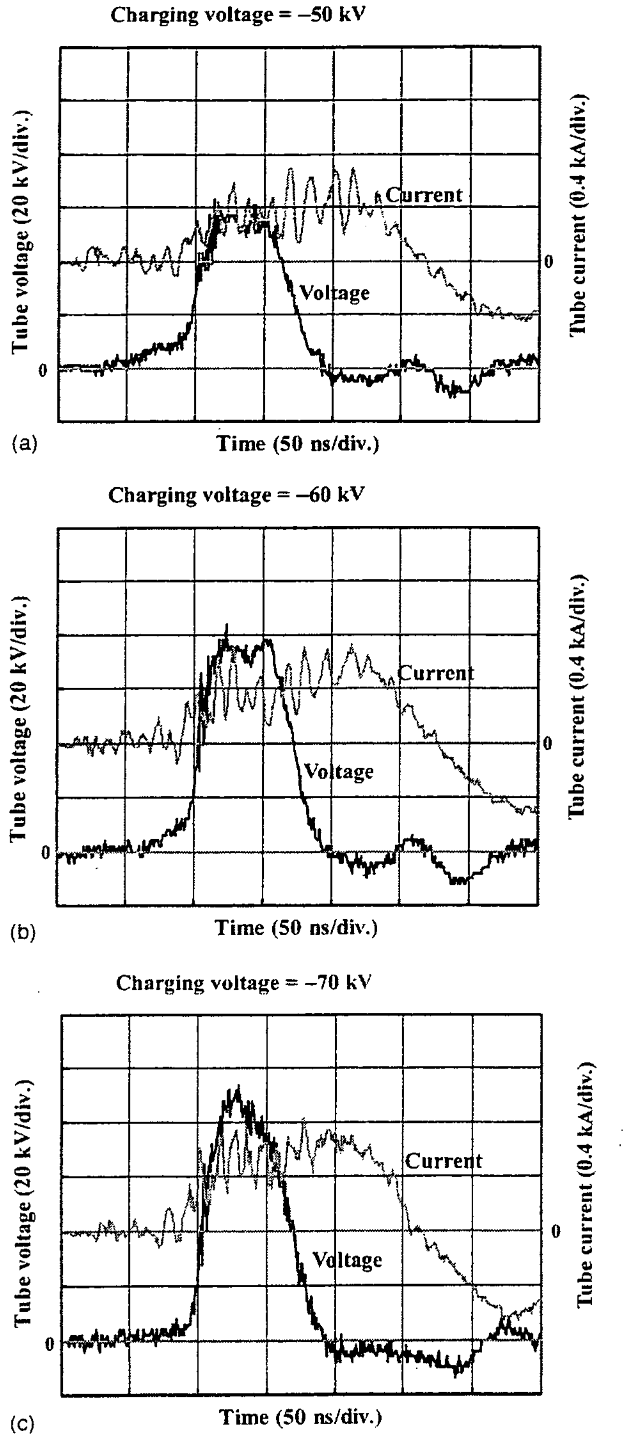

Fig. 5 Variations in tube voltage and current with charging voltages of (a) -50 kV, (b) -60 kV, and (c) -70 kV.

current transformer, respectively (Fig. 5). The voltage and current displayed roughly damped oscillations, because the discharge resistance in the tube varied rapidly from infinity to approximately 0 Ω during the discharge. Thus, at the first quarter cycle of the oscillations, when the voltage decreased, the current increased. Within twice the potential of the condenser charging voltage, the instantaneous voltage increases according to increases in the charging voltage and to increases in the target-cathode space. On the other hand, the instantaneous current increases with increases in the charging voltage and decreases in the space. At a space of 1.0 mm, the instantaneous voltage and current increased with increases in the charging voltage, and the voltage and current were approximately 90 kV and 0.8 kA, respectively, at a charging voltage of –70 kV.

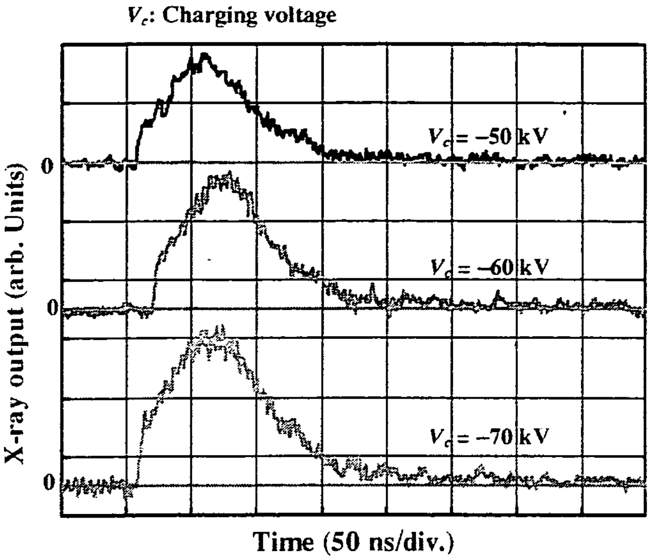

Fig. 6 X-ray outputs at the indicated conditions using the filter.

3.2 *X-ray Output*

An x-ray output pulse was detected using a combination of a plastic scintillator, a photomultiplier, and the filter (Fig. 6). When the charging voltage was increased, the pulse height increased, but the width seldom varied. The widths were approximately 80 ns, and the time-integrated x-ray dose measured using a thermoluminescence dosimeter (Kyokko TLD Reader 1500 having MSO-S elements without energy compensation) had a value of approximately 90 μGy per pulse at 0.3 m from the x-ray source with a charging voltage of –70 kV. In the dose measurement, five elements (detectors) with a diameter of 2.0 mm and a length of 11 mm were set at 0.3 m from the x-ray source, and the dose was the average value of ten shots of flash

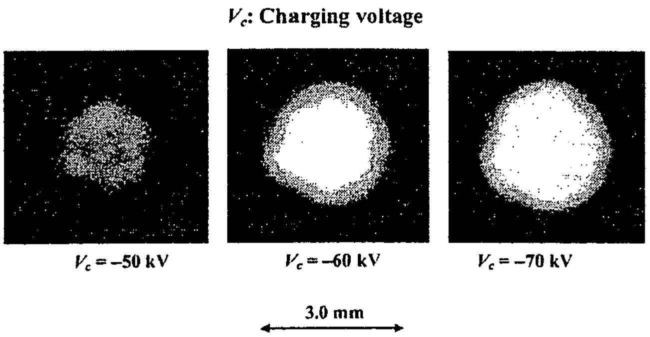

Fig. 7 Images of x-ray sources of Kα lines with changes in the charging voltage.

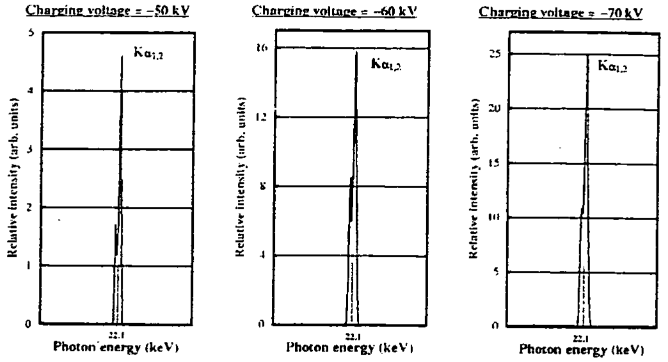

Fig. 8 X-ray spectra from a silver target with the filter.

x-rays and five elements. Using this high-voltage pulse generator, the maximum repetition rate of flash x-rays was approximately 50 Hz. At a charging voltage of −70 kV and the maximum rate, the electric power of the flash x-ray generator and the dose rate are estimated at approximately 200 W and 400 μGy/s at 1.0 m, respectively.

3.3 *X-ray Source*

To observe the x-ray source, we employed a 100-μm-diam pinhole camera, an x-ray film (Polaroid XR-7), and the filter (Fig. 7). When the charging voltage was increased, the spot intensity increased, and the intensities corresponded well to the x-ray pulse height. The dimension was almost equal to the target diameter and had a value of approximately 3.0 mm.

3.4 *X-ray Spectra*

X-ray spectra were measured by a transmission-type spectrometer with a lithium fluoride curved crystal 0.5 mm thick. The spectra were taken using a computed radiography (CR) system[18] (Konica Regius 150) with a wide dynamic range, and relative x-ray intensity was calculated from Dicom digital data. Figure 8 shows the measured spectra from the silver target with the filter. We observed clean Kα lines, while bremsstrahlung rays were hardly detected. When the charging voltage was increased, the instantaneous tube voltage and current increased, and the Kα intensity substantially increased.

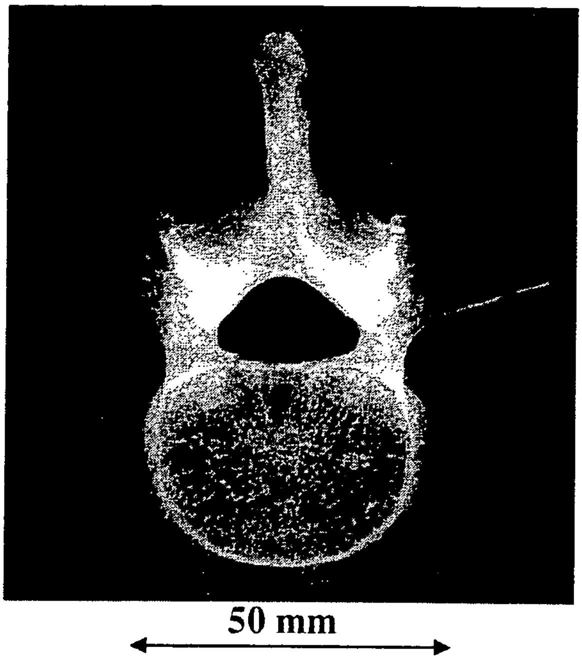

Fig. 10 Radiogram of a vertebra.

3.5 *Radiography*

Monochromatic flash radiography was performed using the CR system 0.3 m from the x-ray source with the filter, and the charging voltage was −70 kV. First, rough measure-

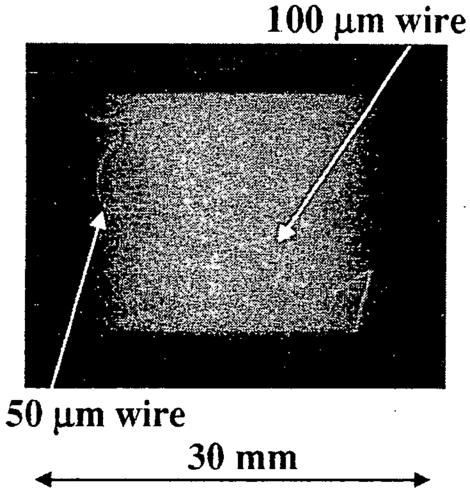

Fig. 9 Radiograms of tungsten wires of 50 and 100 μm in a rod made of polymethyl methacrylate.

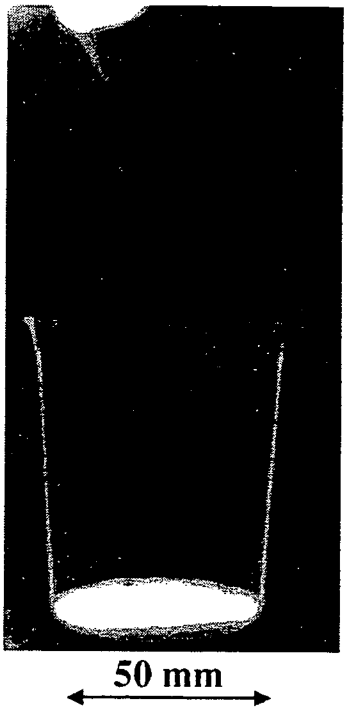

Fig. 11 Radiogram of water falling into a polypropylene beaker from a glass test tube.

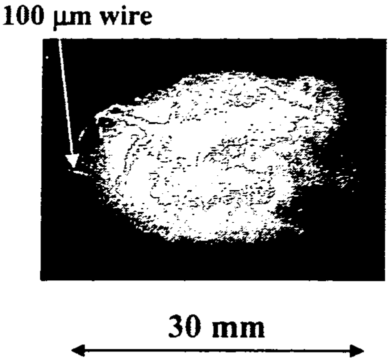

Fig. 12 Angiogram of a rabbit heart.

ments of spatial resolution were made using wires. Figure 9 shows radiograms of tungsten wires in a rod made of polymethyl methacrylate. Although the image contrast increased with increases in the wire diameter, a 50-μm-diam wire could be observed.

Figure 10 shows a radiogram of a vertebra, and fine structures in the vertebra are observed. Next, the image of water falling into a polypropylene beaker from an injector is shown in Fig. 11. This image was taken with the slight addition of an iodine-based contrast medium. Because the x-ray duration was approximately 80 ns, the stop-motion image of water could be obtained. Figure 12 shows an angiogram of a rabbit heart; iodine-based microspheres 15 μm in diameter were used, and fine blood vessels of approximately 100 μm were visible.

4 Conclusion and Outlook

Concerning the spectrum measurement, we obtained fairly clean silver Kα rays (22.1 keV). Therefore, we are very interested in the measurement of the Kα rays from cerium (34.6 keV) and tungsten (58.9 keV) targets. The target element should be selected corresponding to the radiographic objectives. In medical applications, K-series characteristic x-rays of cerium are absorbed effectively by an iodine-based contrast medium with a K-edge of 33.2 keV, and K-edge angiography can be performed.

In this research, the instantaneous number of generator-produced Kα photons was approximately 4×10^7 photons/cm^2 per pulse 0.3 m from the source. However, the intensity can be increased by increasing the electrostatic energy in condensers in the surge generator, and quasimonochromatic x-rays of both Kα and Kβ (24.9 keV) lines are produced without using the palladium filter with a K-edge of 24.3 keV.

Using this flash x-ray generator, because the photon energy of characteristic x-rays can be selected, a high-speed photon-counting radiography system can be performed to decrease noise from radiograms. As compared with a steady state x-ray generator, since the target element can be changed easily using this demountable PMMA tube, demonstrations of monochromatic radiography will be accomplished easily.

Acknowledgment

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X-ray spectra from a cerium target and their application to cone beam K-edge angiography

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Abstract. The cerium-target x-ray tube is useful for 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 glass-enclosed diode with a cerium target and a 0.5-mm-thick beryllium window. The maximum tube voltage and current are 65 kV and 0.4 mA, respectively, and the focal-spot sizes are 1.3×0.9 mm. Cerium K-series characteristic x-rays are left, using a 3.0-mm-thick aluminum filter, and the x-ray intensity is 19.9 μGy/s at 1.0 m from the source with a tube voltage of 60 kV and a current of 0.40 mA. Angiography is performed with a computed radiography system using iodine-based microspheres 15 μm in diameter. In angiography of nonliving animals, we observe fine blood vessels of approximately 100 μm with high contrasts. *© 2005 Society of Photo-Optical Instrumentation Engineers.* [DOI: 10.1117/1.2049268]

Subject terms: x-ray tube; cerium target; quasimonochromatic x-rays; characteristic x-rays; K-edge angiography.

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1 Introduction

Monochromatic parallel x-ray beams are the basis of radiography using synchrotrons in conjunction with single crystals, and these beams have been employed to perform enhanced K-edge angiography[1-3] and x-ray phase imaging.[4-6] In angiography, the beams with photon energies of approximately 35 keV are absorbed effectively by iodine-based contrast media. However, it is difficult to obtain sufficient machine times for various research projects, including medical applications. Subsequently, monochromatic cone beams with energies of approximately 35 keV are useful for increasing the irradiation field for K-edge angiography.

To perform high-speed medical radiography, although several different flash x-ray generators[7-13] utilizing cold-cathode tubes have been developed, plasma flash x-ray generators[14-18] are useful for producing quasimonochromatic x-rays without using a K-edge filter. Therefore, we have performed a demonstration of cone beam K-edge

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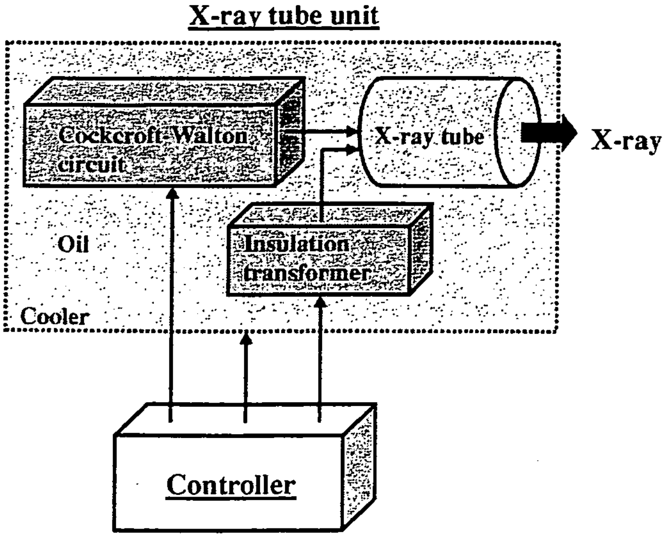

Fig. 1 Block diagram of the compact x-ray generator with a cerium-target radiation tube, which is used specially for K-edge angiography using iodine-based contrast media.

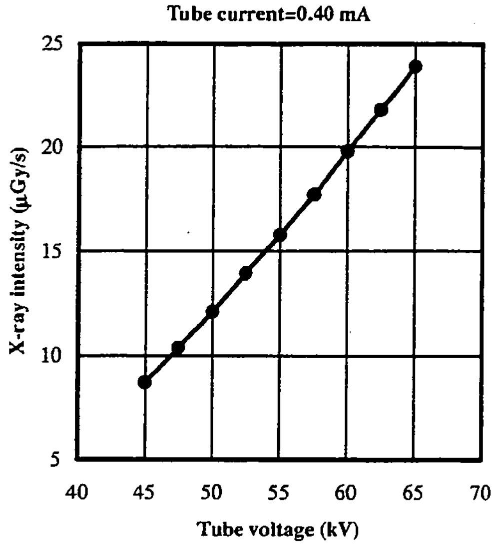

Fig. 3 The x-ray intensity (μGy/s) as a function of tube voltage (kV) with a tube current of 0.40 mA.

angiography[19] utilizing a cerium plasma generator, since K-series characteristic x-rays from the cerium target are absorbed effectively by iodine.

Recently, we have developed a steady-state x-ray generator utilizing a cerium-target tube, and have demonstrated enhanced K-edge angiography utilizing cerium Kα lines (34.6 keV).[20] In this research, Kα lines were left by absorbing Kβ lines (39.2 keV) using a barium sulfate filter with a barium K edge of 37.4 keV. However, because cerium Kβ lines are also absorbed effectively by iodine, both Kα and Kβ lines can be selected to increase the x-ray intensity for angiography. In measurements of x-ray spectra, although we usually employed a cadmium telluride detector with a photon energy resolution of 1.7 keV, the resolution should be minimized to measure the characteristic x-ray intensity.

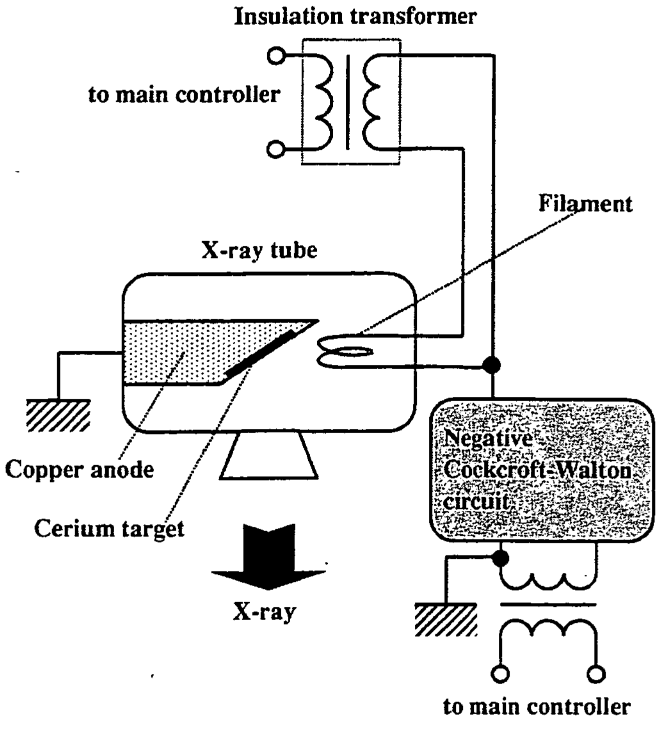

Fig. 2 Main circuit of the x-ray generator.

In the present research, we measured the x-ray spectra from a cerium-target tube using a germanium detector, and performed a preliminary study on cone beam K-edge angiography achieved with cerium characteristic x-rays without using a K-edge filter.

2 Experimental Setup

Figure 1 shows the block diagram of the x-ray generator, which consists of a main controller and an x-ray tube unit 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 employs the Cockcroft-Walton circuit to decrease the dimensions of the tube unit. In the x-ray tube, the negative high 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. In this experiment, the tube

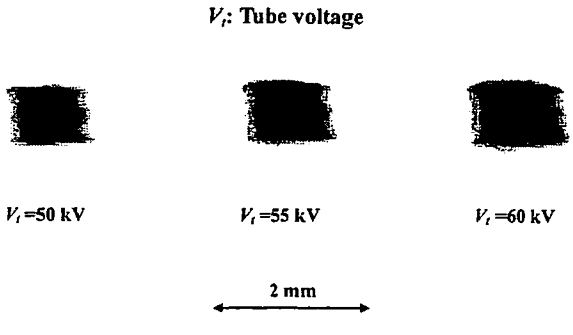

Fig. 4 Effective focal spots with changes in the tube voltage.

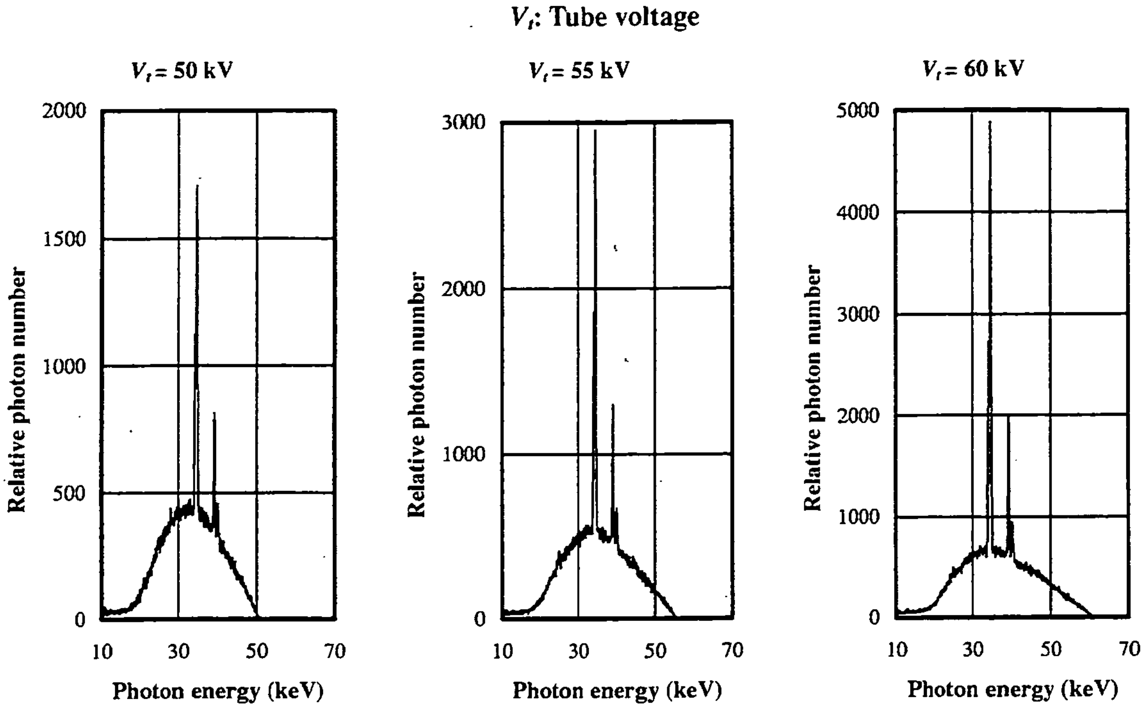

Fig. 5 X-ray spectra measured using a germanium detector with changes in the tube voltage.

voltage applied 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 to obtain optimum x-ray intensity. Quasimonochromatic x-rays are produced using a 3.0-mm-thick aluminum filter for absorbing soft bremsstrahlung rays.

3 Results and Discussion

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 using the filter (Fig. 3). At a constant tube current of 0.40 mA, the x-ray intensity increased when the tube voltage was increased. In this measurement, the intensity with a tube voltage of 60 kV and a current of 0.40 mA was 19.9 μGy/s at 1.0 m from the source, with errors of less than 0.2%.

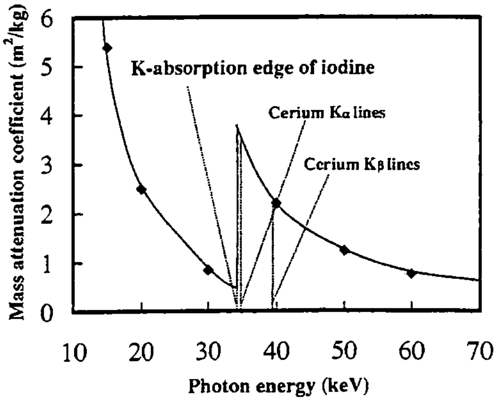

Fig. 6 Mass attenuation coefficients of iodine, and the average photon energies of cerium $K\alpha$ and $K\beta$ lines.

3.2 *Focal Spot*

To measure images of the x-ray source after the aluminum filtration, we employed a pinhole camera with a hole diameter of 50 μm (magnification ratio of 1:2) in conjunction with a computed radiography (CR) system[21] with a sampling pitch of 87.5 μm. When the tube voltage was increased, spot dimensions increased slightly and had values of 1.3×0.9 mm (Fig. 4).

3.3 *X-ray Spectra*

To measure x-ray spectra, we employed a germanium detector (GLP-10180/07-P, Ortec Incorporated) with a photon energy resolution of approximately 0.12 keV (Fig. 5). When the tube voltage was increased, the characteristic x-ray intensities of $K\alpha$ and $K\beta$ lines substantially increased, and both the maximum photon energy and the intensities of bremsstrahlung x-rays increased. Because the widths of the lines were approximately 1 keV, the photon energy resolution of this detector was an optimum value. In an empirical equation, because the characteristic x-ray in-

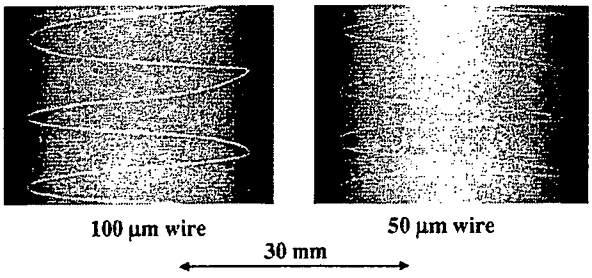

Fig. 7 Radiograms of tungsten wires coiled around PMMA rods with a tube voltage of 60 kV.

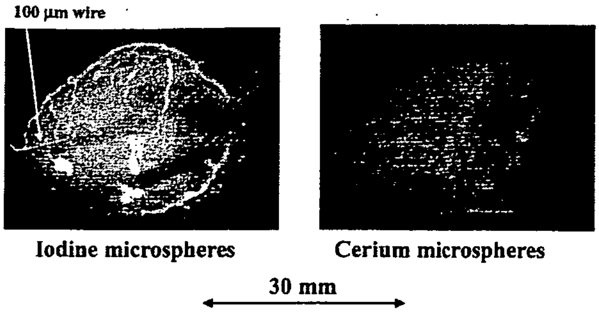

Fig. 8 Angiograms of extracted rabbit hearts using iodine and cerium microspheres with a tube voltage of 60 kV.

tensity is proportional to approximately 1.5 power of the voltage difference between the tube voltage and the critical excitation voltage, the measured intensities of the characteristic x-rays corresponded well to the equation.

3.4 *K-Edge Angiography*

Figure 6 shows the mass attenuation coefficients of iodine at the selected energies; the coefficient curve is discontinuous at the iodine K edge. The average photon energy of the cerium Kα and Kβ lines are shown just above the iodine K edge. The average photon energies of Kα and Kβ lines are 34.6 and 39.2 keV, respectively, and iodine contrast media with a K-absorption edge of 33.2 keV absorb the lines easily. Therefore, blood vessels were observed with high contrasts.

The angiography was performed by a CR system (Konica Minolta Regius 150) using the filter, 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 7 shows radiograms of tungsten wires coiled around a rod 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-diam wire could be observed.

Angiograms of rabbit hearts are shown in Fig. 8. These two images were obtained using iodine and cerium microspheres of 15 μm in diameter at a tube voltage of 60 kV.

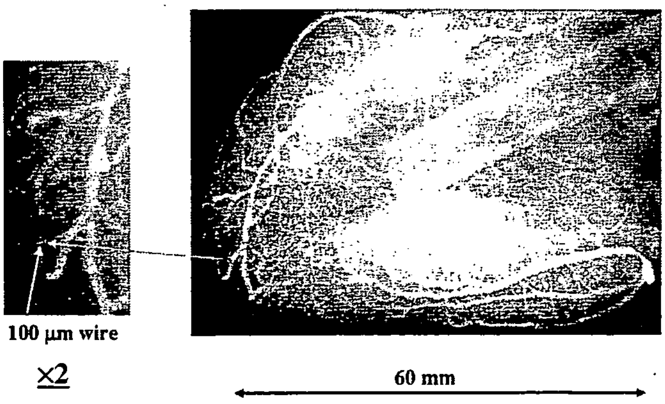

Fig. 9 Angiograms of an extracted dog heart using iodine microspheres with a tube voltage of 60 kV.

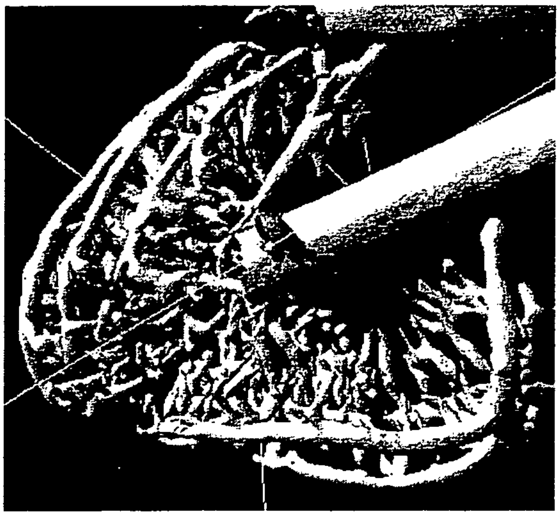

Fig. 10 3-D image of coronary arteries constructed from x-ray CT images by Pascal.

The microspheres are very useful for making phantoms of nonliving animals used for angiography. The iodine plastic spheres contained 37% iodine by weight, and the cerium plastic spheres were chemically stable and contained 18% cerium by weight. In the case where the cerium spheres were employed, the coronary arteries were barely visible, since the cerium spheres transmitted cerium characteristic x-rays easily. Figure 9 shows an angiogram of a larger dog heart at a tube voltage of 60 kV using iodine spheres. For comparison, we show a 3-D image of the coronary arteries constructed from x-ray CT images by Pascal (Digital Culture Technical Corporation) with a tungsten x-ray tube (Fig. 10). Using this imaging technique, fine blood vessels were not observed at all.

4 Conclusion and Outlook

In summary, we employ an x-ray generator with a cerium-target tube and succeed in producing cerium characteristic x-rays, which can be absorbed easily by iodine-based contrast media. The characteristic x-ray intensities increase with increases in the tube voltage, and low-photon-energy bremsstrahlung rays are absorbed effectively by the filter.

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 in Sommerfeld's theory.[22] Subsequently, the generator-produced maximum number of characteristic photons is approximately 3.5×10^7 photons/cm$^2 \cdot$s at 1.0 m from the source, and the photon count rate can be increased easily by improving the target.

The x-ray intensity is limited because the thermal contact between the target and the anode is not good. However, the intensity can be increased by welding the target or using a cerium-alloy target. In addition, a rotation anode tube can be developed by the sputtering of cerium.

Compared to the 3-D blood images constructed from x-ray CT images by Pascal, fine blood vessels are visible.

Because the sampling pitch of the CR system is 87.5 μm, we obtain spatial resolutions of approximately 100 μm. To observe fine blood vessels of less than 100 μm, the spatial resolution of the CR system should be improved to approximately 50 μm (Konica Minolta Regius 190). In addition, the spatial resolution can be improved easily to approximately 50 μm or less in cases where an x-ray film (Fuji Ix 100) is employed.

Acknowledgment

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High-speed enhanced K-edge angiography utilizing cerium plasma x-ray generator

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Abstract. The cerium target plasma flash x-ray generator is useful to perform high-speed enhanced K-edge angiography using cone beams, because K-series characteristic x-rays from the cerium target are absorbed effectively by iodine-based contrast mediums. In the plasma generator, a 200-nF condenser is charged up to 60 kV by a power supply, and flash x-rays are produced by the discharging. The x-ray tube is a demountable triode with a trigger electrode, and the turbomolecular pump evacuates air from the tube with a pressure of approximately 1 mPa. Target evaporation leads to the formation of weakly ionized linear plasma, consisting of cerium ions and electrons, around the target, and intense flash x-rays are produced. At a charging voltage of 55 kV, the maximum tube voltage is almost equal to the charging voltage of the main condenser, and the maximum current is approximately 20 kA. When the charging voltage is increased, weakly ionized cerium plasma forms, and the K-series characteristic x-ray intensities increase. The x-ray pulse widths are about 500 ns, and the time-integrated x-ray intensity has a value of about 40 μC/kg at 1.0 m from the x-ray source with a charging voltage of 55 kV. In the angiography, we employ a filmless computed radiography (CR) system and iodine-based microspheres.

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Subject terms: plasma x-ray; cerium target; weakly ionized cerium plasma; characteristic x-ray; K-edge angiography.

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1 Introduction

Flash x-rays are useful to perform high-speed radiography, and various generators have been developed to correspond to specific radiographic objectives.[1-5] In the cases of multishot and cine radiographies, we have developed several different repetitive-flash[6-10] and stroboscopic x-ray generators.[11-17] Although most flash x-ray generators have cold-cathode tubes, the stroboscopic generators utilize hot-cathode tubes.

In conjunction with single crystals, synchrotrons generate monochromatic x-rays. These rays play important roles in parallel radiography and have been employed to perform high-contrast K-edge angiography[18] and x-ray phase imaging.[19,20] However, it is difficult to obtain sufficient machine times for various research projects, including medical applications.

As for angiography using iodine-based contrast mediums, K-series characteristic x-rays of cerium are extremely useful, since the rays are absorbed easily by iodine. In particular, since fairly intense and sharp characteristic x-rays have been produced from weakly ionized linear plasmas[21-24] of nickel, copper, and molybdenum, the development of a cerium-target x-ray tube for angiography is highly desirable.

In this research, we developed a single flash x-ray generator with a cerium-target plasma tube and performed a preliminary study on weakly ionized cerium plasma angiography.

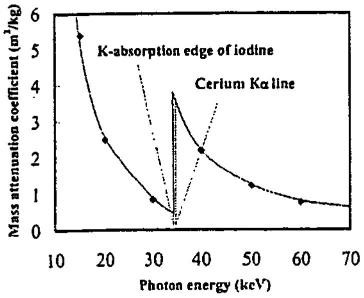

Fig. 1 Relation between mass attenuation coefficient of iodine and average photon energy of cerium Kα lines.

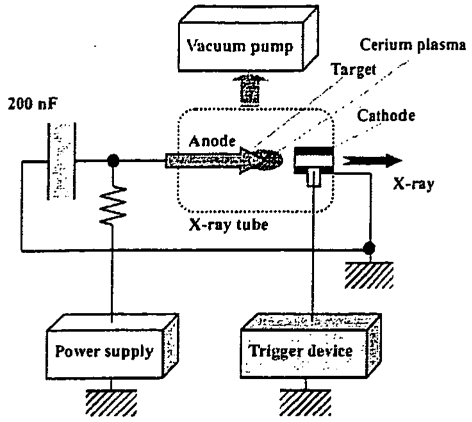

Fig. 2 Block diagram of high intensity plasma flash x-ray generator.

2 Principle of K-Edge Angiography

Figure 1 shows the mass attenuation coefficients of iodine at the selected energies; the coefficient curve is discontinuous at the iodine K-edge. The average photon energy of the cerium Kα lines is shown just above the iodine K-edge. Cerium is a rare earth element and has a high reactivity; however, the average photon energy of Kα lines is 34.566 keV, and iodine contrast mediums with a K-absorption edge of 33.155 keV absorb the lines easily. Therefore, blood vessels were observed with high contrasts.

3 Generator

3.1 *High-Voltage Circuit*

Figure 2 shows a block diagram of a high-intensity plasma flash x-ray generator. This generator consists of the following essential components: a high-voltage power supply, a high-voltage condenser with a capacity of about 200 nF, a turbomolecular pump, a krytron pulse generator as a trigger device, and a flash x-ray tube. The high-voltage main condenser is charged up to 60 kV by the power supply, and electric charges in the condenser are discharged to the tube after triggering the cathode electrode by the trigger device. The plasma flash x-rays are then produced.

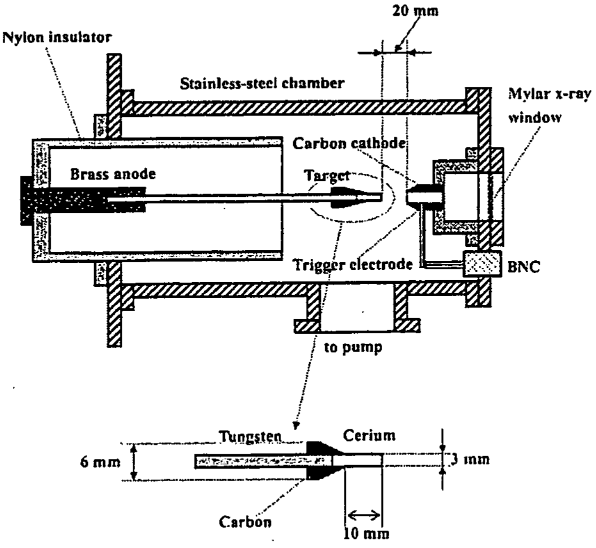

Fig. 3 Schematic drawing of flash x-ray tube.

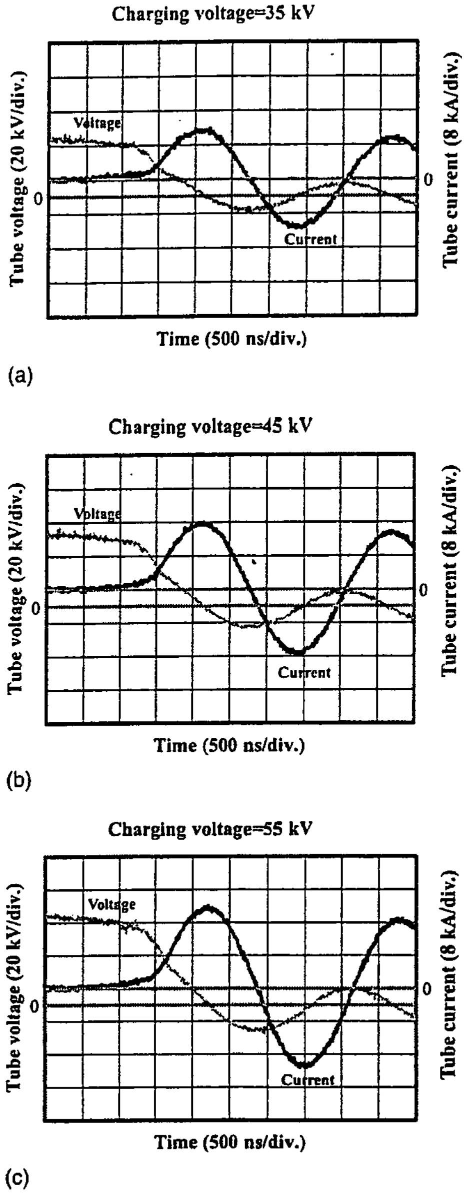

Fig. 4 Tube voltages and currents with charging voltage of (a) 35, (b) 45, and (c) 55 kV.

3.2 *X-Ray Tube*

The x-ray tube is a demountable cold-cathode triode that is connected to the turbomolecular pump with a pressure of approximately 1 mPa (Fig. 3). This tube consists of the following major parts: a hollow cylindrical carbon cathode with a bore diameter of 10.0 mm, a trigger electrode made from a copper wire, a stainless-steel vacuum chamber, a nylon insulator, a polyethylene terephthalate (Mylar) x-ray window of 0.25 mm, and a rod-shaped cerium target of 3.0 mm in diameter. The target tip is embedded in the carbon rod to absorb the characteristic x-rays of carbon by the window. The distance between the target and cathode electrodes is approximately 20 mm, and the trigger electrode is

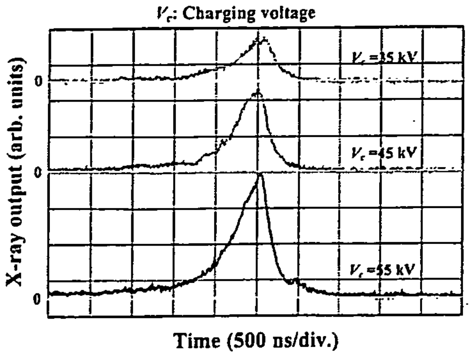

Fig. 5 X-ray outputs at indicated conditions.

set in the cathode electrode. As electron beams from the cathode electrode are roughly converged to the target by an electric field in the tube, the weakly ionized plasma, which consists of cerium ions and electrons, forms around the target by evaporating.

4 Characteristics

4.1 *Tube Voltage and Current*

Tube voltage and current were measured by a high-voltage divider with an input impedance of 1 GΩ and a current transformer, respectively. Figure 4 shows the time relation between the tube voltage and current. At the indicated charging voltages, they roughly displayed damped oscillations. When the charging voltage was increased, both the maximum tube voltage and current increased. At a charging voltage of 55 kV, the maximum tube voltage was almost equal to the charging voltage of the main condenser, and the maximum tube current was about 20 kA.

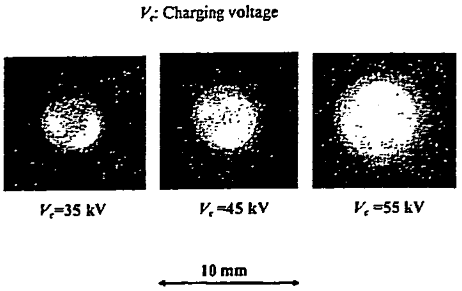

Fig. 6 Images of plasma x-ray source.

4.2 *X-Ray Output*

An x-ray output pulse was detected using a combination of a plastic scintillator and a photomultiplier. The x-ray pulse height substantially increased with corresponding increases in the charging voltage (Fig. 5). The x-ray pulse widths were about 500 ns, and the time-integrated x-ray intensity measured by a thermoluminescence dosimeter (Kyokko TLD Reader 1500 utilizing MSO-S elements without energy compensation) had a value of about 40 μC/kg at 1.0 m from the x-ray source with a charging voltage of 55 kV.

4.3 *X-Ray Source*

To measure images of the plasma x-ray source, we employed a pinhole camera with a hole diameter of 100 μm (Fig. 6). When the charging voltage was increased, the plasma x-ray source grew, and both spot dimension and intensity increased. Because the x-ray intensity is the highest at the center of the spot, both the dimension and intensity decreased according to both increases in the thickness of a filter for absorbing x-rays and decreases in the pinhole diameter.

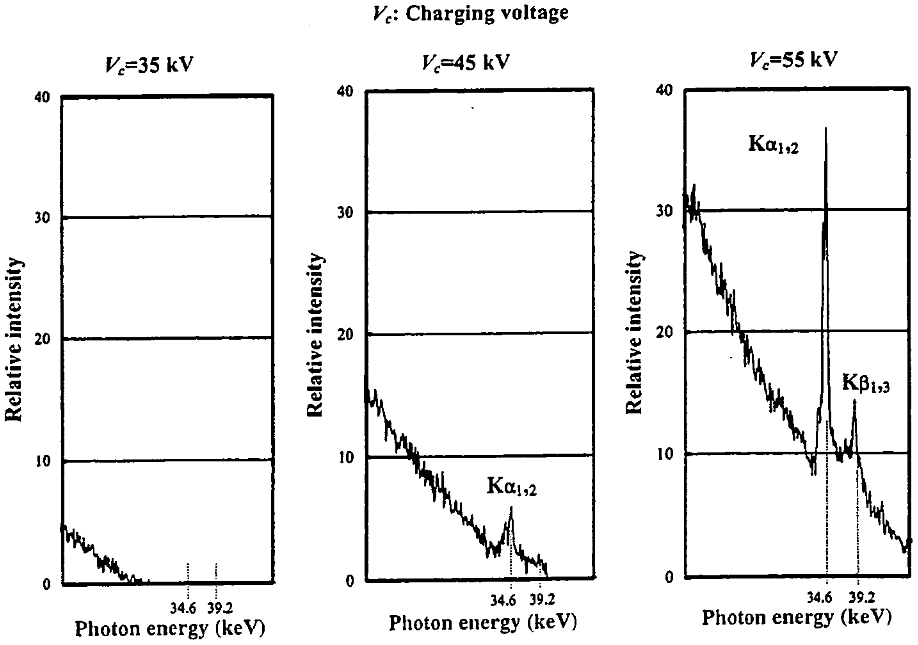

Fig. 7 X-ray spectra from weakly ionized cerium plasma.