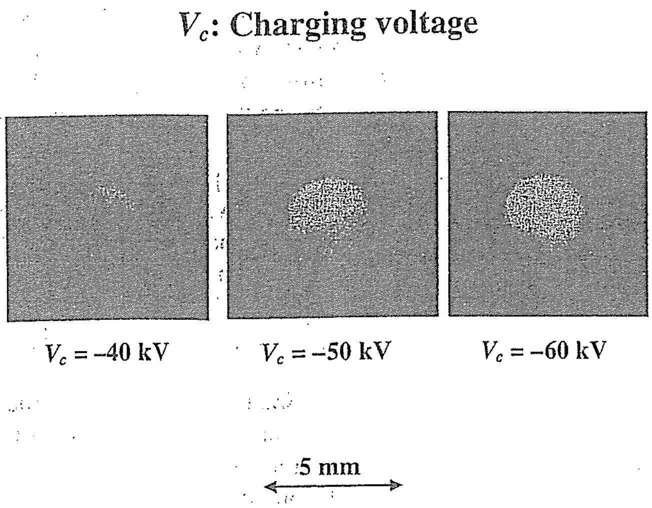

Fig. 8. Images of X-ray source with changes in charging voltage at constant space between target and cathode electrodes.

3.4 X-ray spectra

X-ray spectra were measured by a transmission-type spectrometer with a lithium fluoride curved crystal 0.5 mm in thickness. The spectra were measured using a computed radiography (CR) system[15] (Konica Regius 150) with a wide dynamic range, and relative X-ray intensity was calculated from Dicom digital data. Figure 9 shows the measured spectra from the molybdenum target. We observed sharp lines of K-series characteristic X-rays, while bremsstrahlung rays were hardly detected. The characteristic X-ray intensity of the $K\alpha$ and $K\beta$ lines substantially increased with increasing charging voltage.

4. Radiography

Flash radiography was performed using the CR system at 0.5 m from the X-ray source, and the charging voltage and the T–C space were −60 kV and 1.0 mm, respectively.

Firstly, rough measurements of spatial resolution were made using wires. Figure 10 shows radiograms of tungsten wires coiled around a pipe made of polymethyl methacrylate.

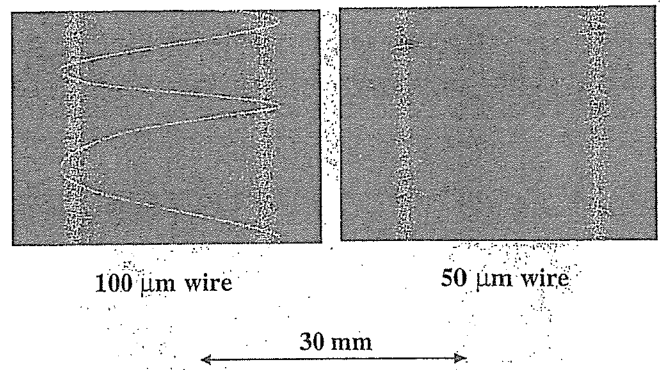

Fig. 10. Radiograms of tungsten wires of 50 and 100 μm in diameter coiled around pipes made of polymethyl methacrylate.

Although the image contrast increased with increasing wire diameter, a 50-μm-diameter wire could be observed.

An image of plastic bullets falling into a polypropylene beaker from a glass test tube is shown in Fig. 11. Because the X-ray pulse widths were approximately 60 ns, the stop-motion image of bullets could be obtained. Figure 12 shows an angiogram of a rabbit heart; iodine-based microspheres of 15 μm in diameter were used, and fine blood vessels of approximately 100 μm were visible.

5. Discussion

Concerning the spectrum measurement, sharp molybdenum K-series characteristic X-rays were obtained, and monochromatic $K\alpha$ lines can be obtained using a zirconium filter. The photon energies of characteristic X-rays are determined by the target element, and the X-ray intensity increases with increasing tube voltage by increasing the charging voltage. As compared with the plasma flash X-ray generator utilizing a molybdenum target triode,[13] bremsstrahlung X-rays were hardly observed at all even when higher tube voltages were applied to the diode, since the characteristic X-rays were produced from the target tip. Because the maximum tube voltage can be increased easily, and high-photon-energy K-series characteristic X-rays from

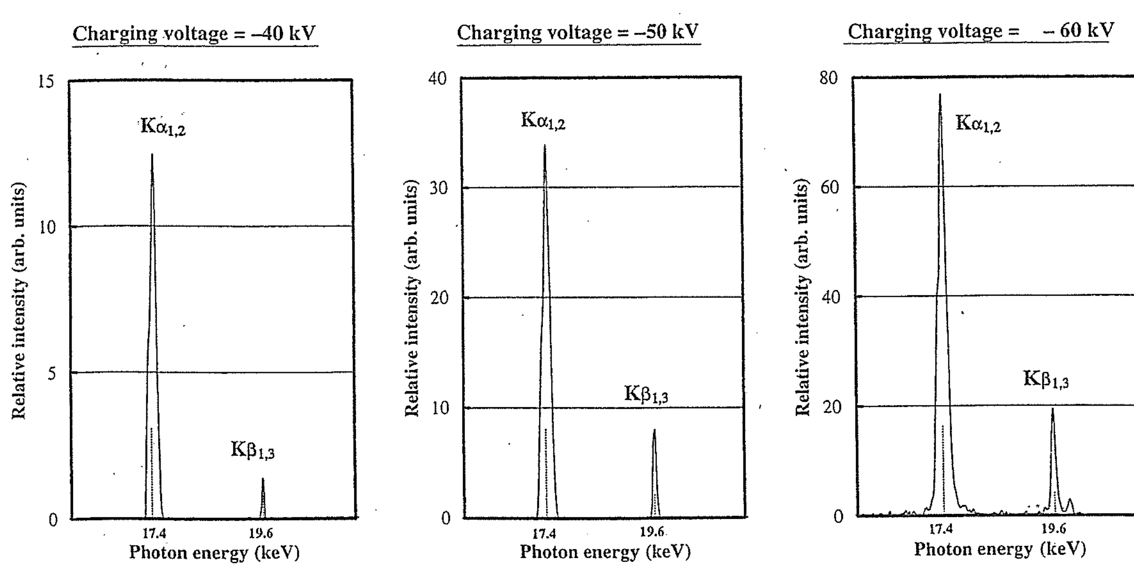

Fig. 9. X-ray spectra from weakly ionized molybdenum plasma according to changes in charging voltage with space of 1.0 mm.

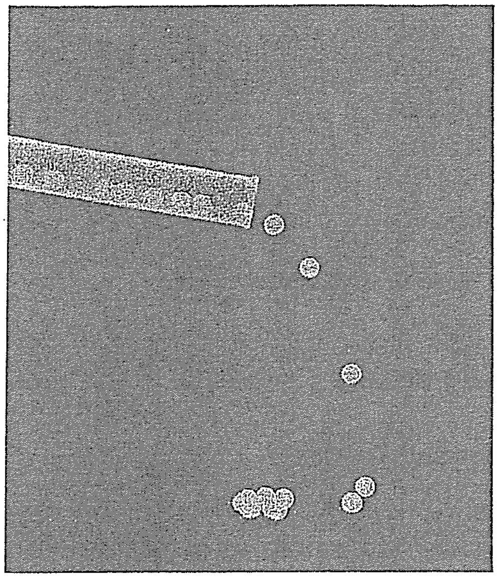

Fig. 11. Radiograms of plastic bullets falling into polypropylene beaker from glass test tube.

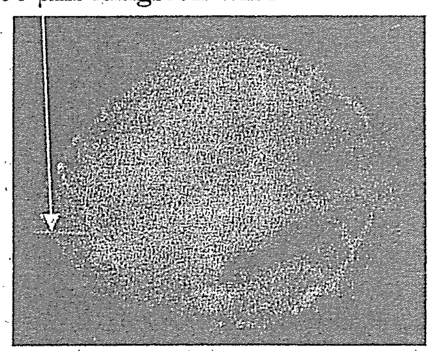

Fig. 12. Angiogram of rabbit heart.

the cerium or tungsten target can be produced. In particular, the cerium target is very useful in order to perform microangiography using iodine-based contrast mediums.

In this research, although the number of generator-produced characteristic K photons was approximately 7×10^{14} photons/cm^2·s at 0.5 m from the source, the number can be increased easily by increasing the electrostatic energy in the condensers.

Using this generator, because the photon energies of characteristic X-rays can be selected, various quasi-monochromatic high-speed radiographies, such as high contrast microangiography and photon-counting radiography for decreasing noise from radiograms, will be possible.

Acknowledgment

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Demonstration of enhanced *K*-edge angiography using a cerium target x-ray generator

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The cerium target x-ray generator is useful in order to perform enhanced *K*-edge angiography using a cone beam because *K*-series characteristic x rays from the cerium target are absorbed effectively by iodine-based contrast mediums. The x-ray generator consists of a main controller, a unit with a Cockcroft-Walton circuit and a fixed anode x-ray tube, and a personal computer. 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 were 65 kV and 0.4 mA, respectively, and the focal-spot sizes were 1.0 $\times$ 1.3 mm. Cerium $K\alpha$ lines were left using a barium sulfate filter, and the x-ray intensity was 0.48 μC/kg at 1.0 m from the source with a tube voltage of 60 kV, a current of 0.40 mA, and an exposure time of 1.0 s. Angiography was performed with a computed radiography system using iodine-based microspheres. In coronary angiography of nonliving animals, we observed fine blood vessels of approximately 100 μm with high contrasts. © *2004 American Association of Physicists in Medicine.* [DOI: 10.1118/1.1803433]

Key words: x-ray source, x-ray tube, x-ray spectra, attenuation coefficient, angiography

I. INTRODUCTION

Synchrotrons generate monochromatic parallel x-ray beams using single crystals. These beams with photon energies of approximately 35 keV have been employed to perform enhanced *K*-edge angiography,[1–4] since the beams are absorbed effectively by iodine-based contrast mediums. However, it is difficult to increase the irradiation field, due to the parallel beam, and to obtain sufficient machine times for various research projects, including medical applications.

Currently, flash x-ray generators utilize cold-cathode radiation tubes and produce extremely short x-ray pulses of less than 1 μs. So far, several different flash x-ray generators have been developed,[5] and the generators with photon energies of lower than 150 keV[6–11] can be employed to perform biomedical radiography. In order to produce monochromatic x rays, plasma flash x-ray generators are useful, since quite intense and clean characteristic x rays have been produced from weakly ionized linear plasmas of nickel, copper,[12] and molybdenum,[13] while bremsstrahlung rays are hardly detected at all. Using these generators, the characteristic x-ray intensity substantially increased with corresponding increases in the charging voltage.

Since *K*-series characteristic x rays from cerium target are absorbed effectively by iodine-based contrast mediums, a cerium-target x-ray tube is very useful in order to perform high contrast angiography. On the other hand, cerium is a rare earth element and has a high reactivity, and it is difficult to design the target. However, we are very interested in pro-

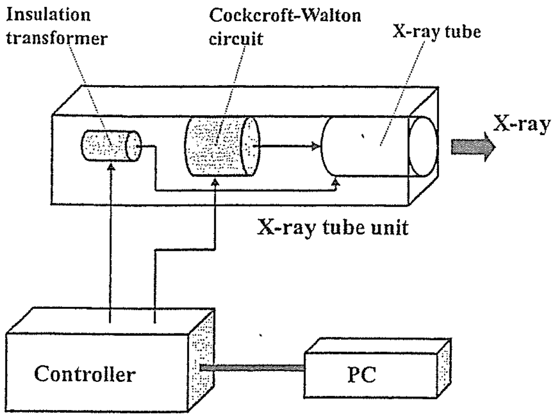

FIG. 1. Block diagram of the compact x-ray generator with a cerium-target radiation tube, which is used specially for K-edge angiography using iodine-based contrast mediums.

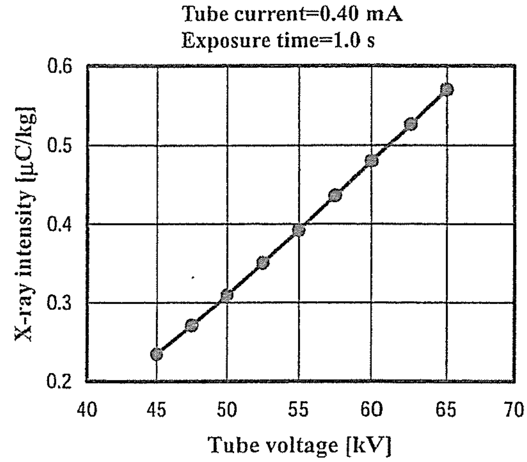

FIG. 3. X-ray intensity measured at 1.0 m from the x-ray source according to changes in the tube voltage.

ducing cerium characteristic x rays to perform cone beam angiography because the irradiation field can be increased easily.

In the present research, we developed a compact x-ray generator with a cerium target tube, and used it to perform a preliminary study on enhanced K-edge angiography achieved with cerium $K\alpha$ rays.

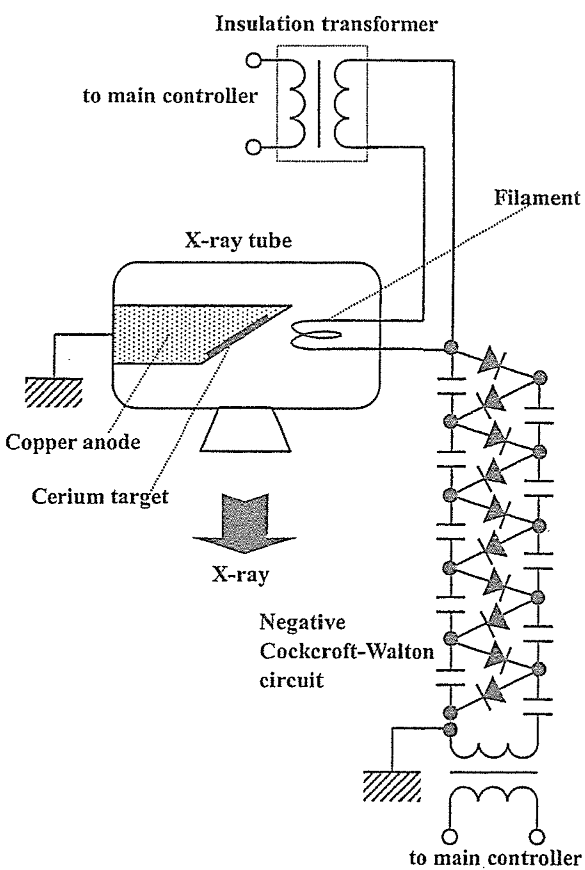

FIG. 2. Main circuit of the x-ray generator.

II. GENERATOR

Figure 1 shows the block diagram of the x-ray generator, which consists of a main controller, an x-ray tube unit with a Cockcroft-Walton circuit, and a cerium-target tube, and a personal computer. The tube voltage, the current, and the exposure time can be controlled by both the controller and the computer. The main circuit for producing x rays is illustrated in Fig. 2, and employed the Cockcroft-Walton circuit in order to decrease the dimensions of the tube unit. In the circuit, the condensers are always in series, and are charged serially. 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 which is used for isolation from the high voltage from the Cockcroft-Walton circuit. In this experiment, the tube 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 in order to obtain optimum x-ray intensity. Monochromatic $K\alpha$ lines were left using a 5-mm-thick barium sulfate filter in which barium sulfate powder was mixed with polymethyl methacrylate (PMMA) resin, since both the bremsstrahlung and the $K\beta$ rays were absorbed effectively by the filter. 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 the PMMA resin. In this case, the density was 7.6 mg/cm^2.

III. CHARACTERISTICS

A. 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 with an exposure time of 1.0 s (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 inten-

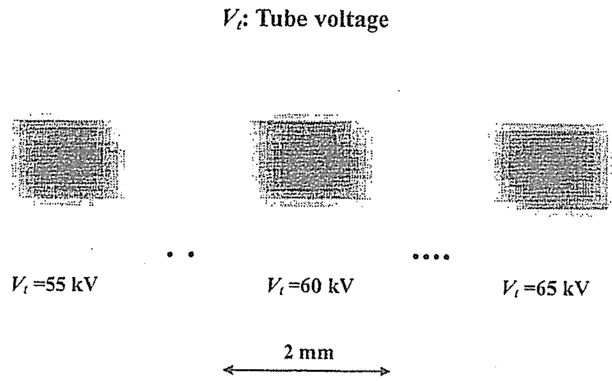

FIG. 4. Effective focal spots with changes in the tube voltage.

sity with a tube voltage of 60 kV and a current of 0.40 mA was 0.48 μC/kg at 1.0 m from the source with errors of less than 0.2%.

B. Focal spot

In order to measure images of the x-ray source after the barium sulfate filtration, we employed a pinhole camera with a hole diameter of 50 μm (magnification ratio of 1:1) in conjunction with a computed radiography (CR) system[14,15] with a sampling pitch of 87.5 μm. When the tube voltage was increased, spot dimensions seldom varied and had values of 1.0×1.3 mm (Fig. 4).

C. X-ray spectra

In order to measure x-ray spectra, we employed a cadmium tellurium detector (CDTE2020X, Hamamatsu Photonics Inc.) (Fig. 5). Compared with a germanium detector, this detector has a lower energy resolution of 1.7 keV. When the tube voltage was increased, the characteristic x-ray intensities of $K\alpha$ lines increased, and both the maximum photon energy and the intensities of bremsstrahlung x rays increased. The barium sulfate filter significantly attenuate the spectra above the barium K-edge energy of 37.399 keV. The areas under the spectral curves correlate closely to the total x-ray intensities shown in Fig. 3.

IV. 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 $K\alpha$ 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 $K\alpha$ 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. Subsequently, in angiography testing, we usually employ nonliving animal phantoms using microspheres.

The angiography was performed by the CR system (Konica Regius 150) using the filter, and the distance (between the x-ray source and the imaging plate) was 1.5 m.

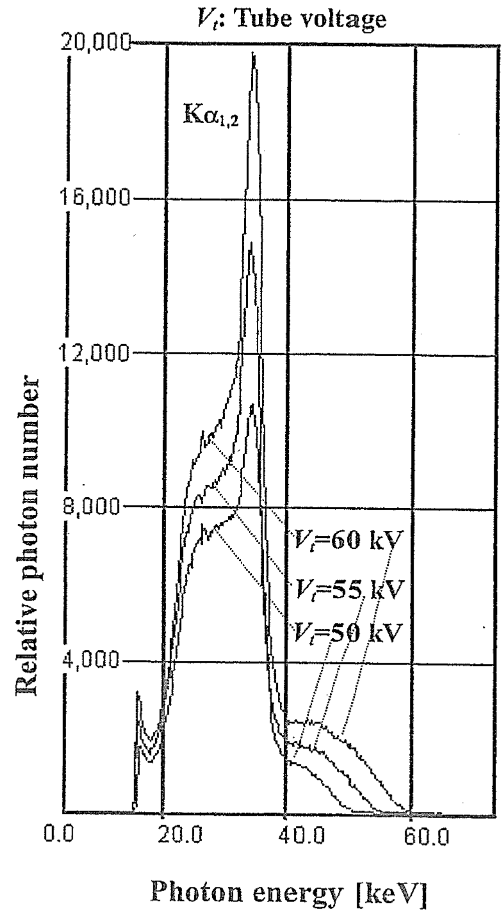

FIG. 5. X-ray spectra measured by a cadmium tellurium detector with changes in the tube voltage.

First, rough measurements of image resolution were made using wires. Figure 7 shows radiograms of tungsten wires in a rod made of PMMA with a tube voltage of 60 kV. 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.

Angiograms of rabbit hearts are shown in Fig. 8. These two images were obtained using iodine and cerium micro-

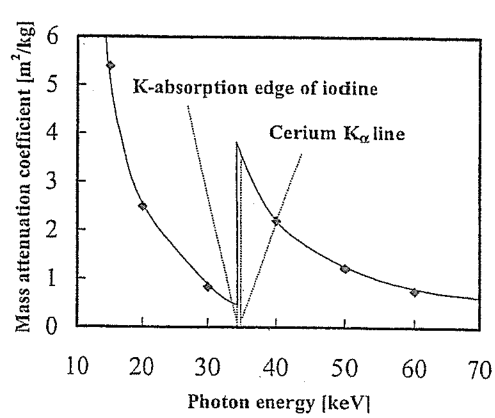

FIG. 6. Mass attenuation coefficients of iodine, and the average photon energy of the cerium $K\alpha$ lines is shown just above the iodine K edge.

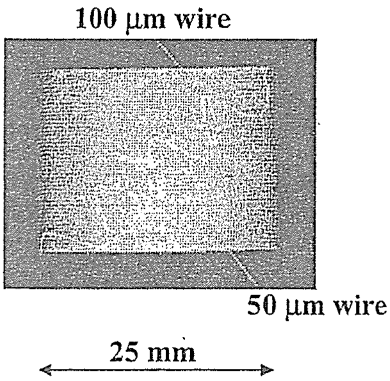

FIG. 7. Radiograms of tungsten wires in a PMMA rod with a tube voltage of 60 kV.

spheres of 15 μm in diameter at a tube voltage of 60 kV. The iodine spheres contained 37% iodine by weight, and the cerium spheres contained 18% cerium by weight. The concentration of spheres in the blood varies with the filling rate, and the estimated densities of the iodine and the cerium of blood are less than 0.44 and 0.17 g/cm^3, respectively. In the case where the cerium spheres were employed, the coronary arteries were barely visible. Figure 9(a) shows an angiogram of a larger dog heart using the cerium target at a tube voltage of 60 kV using iodine spheres. For comparison, we performed angiography with a tungsten x-ray tube at a tube voltage of 60 kV [Fig. 9(b)].

If we assume that the filling rate of the iodine microspheres in a blood vessel is constant, the image contrast of the blood is in inverse proportion to the vessel diameter. Next, the density ratios (maximum density divided by minimum density) obtained by the cerium and tungsten tubes were 4.3 and 2.7, respectively. In angiography using the tungsten target, blood vessels of approximately 100 μm were hardly observed at all.

V. DISCUSSION

In summary, we developed a x-ray generator with a cerium-target tube and succeeded in producing cerium $K\alpha$ lines, which can be absorbed easily by iodine-based contrast mediums. Both the characteristic and bremsstrahlung x-ray

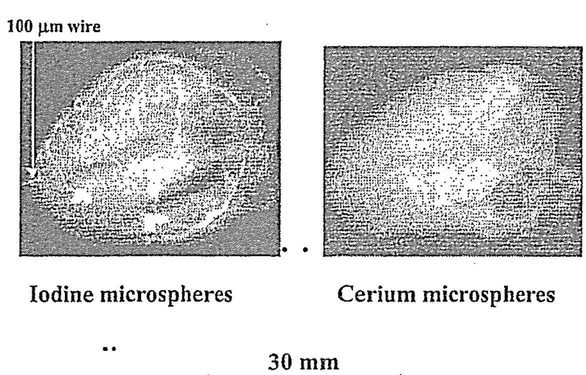

FIG. 8. Angiograms of extracted rabbit hearts using iodine and cerium microspheres with a tube voltage of 60 kV.

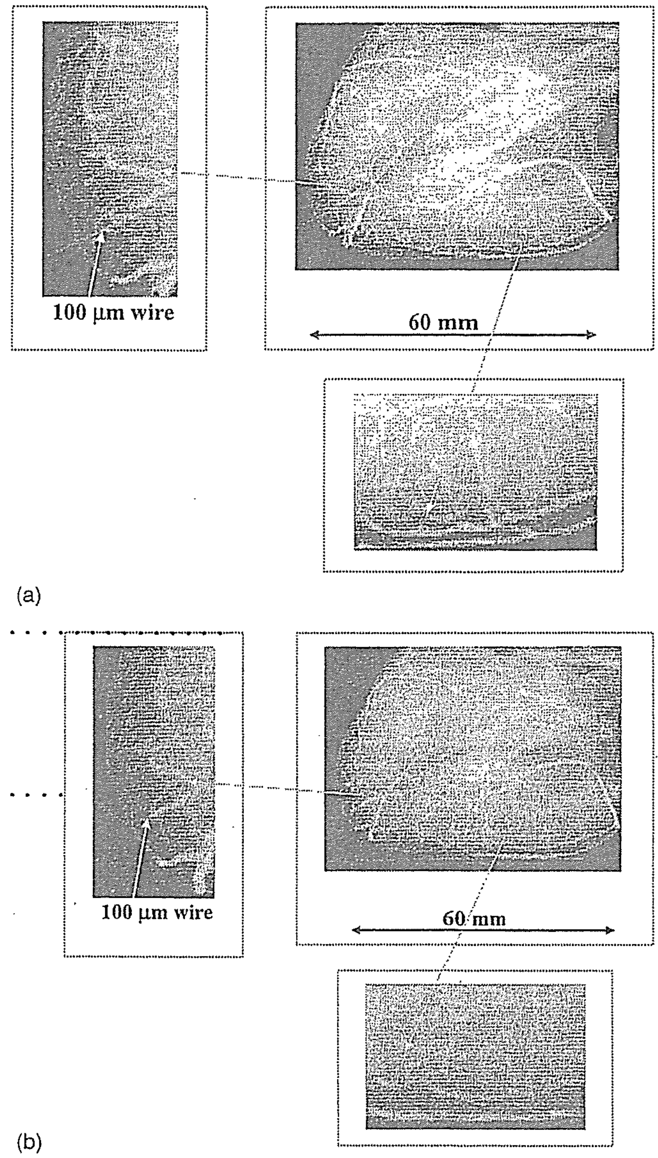

FIG. 9. Angiograms of an extracted dog heart achieved with (a) cerium and (b) tungsten target x-ray tubes using iodine microspheres with a tube voltage of 60 kV.

intensities increased with increases in the tube voltage, and $K\beta$ lines were absorbed effectively by the barium sulfate filter. The x-ray intensity was limited because the thermal contact between the target and the anode was not good. However, the intensity can be increased by welding the target or using a cerium-alloy target.

In this preliminary experiment, although the maximum tube voltage and current were 65 kV and 0.40 mA, respectively, the voltage and current could be increased. Subsequently, the generator produced maximum number of characteristic 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.

Since the sampling pitch of the CR system is 87.5 μm, we obtained resolutions of approximately 100 μm, and high-contrast blood vessels could be observed using a CR system. In order to observe fine blood vessels of less than 100 μm, the image resolution of the CR system should be improved as much as possible, and a flat panel system is useful to observe blood flows for cases of cardiovascular disease.

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Bremsstrahlung X-ray Spectra for Enhanced K-edge Angiography

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(Received November 5, 2004)

Abstract

Energy-selective enhanced K-edge angiography utilizing a conventional x-ray generator is described. The x-ray generator is SOFRON NST-1005, and the maximum tube voltage and current are 100 kV and 5 mA, respectively. In the present research, the tube voltage ranged from 45 to 65 kV, and the tube current was regulated to optimum values. The exposure time is controlled in order to obtain optimum x-ray intensity. At a charging voltage of 60 kV, the x-ray intensity rate obtained using an aluminum and a barium sulfate filters were 58.4 and 51.6 μGy/s at 0.7 m per pulse, respectively, and the dimensions of the focal spot were approximately 1 × 1 mm. Angiography was performed using both the aluminum and the barium sulfate filters with a charging voltage of 60 kV.

Keywords: angiography, aluminum filtering, barium sulfate filtering, quasi-monochromatic x rays, iodine-based contrast medium

1. Introduction

Monochromatic parallel radiography using a synchrotron in conjunction with single crystals continues to be the major tool used in x-ray phase imaging[1,2] and enhanced K-edge angiography.[3,4] In cases where the phase imaging is employed, the spatial resolution can be improved, and the number of tissues which can be observed using x rays increases.

To perform high-speed radiography, several different flash x-ray generators have been developed,[5] and soft generators[6-10] with photon energies of lower than 150 keV can be employed to perform biomedical

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radiography. In order to produce monochromatic x rays, plasma flash x-ray generators[11-15] are useful, since quite intense and sharp characteristic x rays such as lasers have been produced from weakly ionized linear plasmas of nickel, copper and molybdenum.

Parallel beams with photon energies of approximately 35 keV have been employed so as to perform angiography, since these beams are absorbed effectively by an iodine-based contrast medium. Subsequently, K-series characteristic x rays with energies of approximately 35 keV are also useful, and fine blood vessels were observed with high contrasts. In view of this situation, we have developed x-ray generators with cerium-target tubes[16,17] which can produce $K\alpha$ rays of 34.6 keV. Because bremsstrahlung x rays of approximately 35 keV also useful in order to perform high-contrast angiography, the development of optimum filters for producing narrow-energy-latitude bremsstrahlung x rays are desired in cases where a conventional tungsten target is employed.

In this research, we employed a tungsten-target x-ray tube and performed a preliminary study on enhanced angiography achieved with bremsstrahlung x rays with narrow-photon-energy latitudes produced by filtering in conjunction with a computed radiography (CR) system.[18]

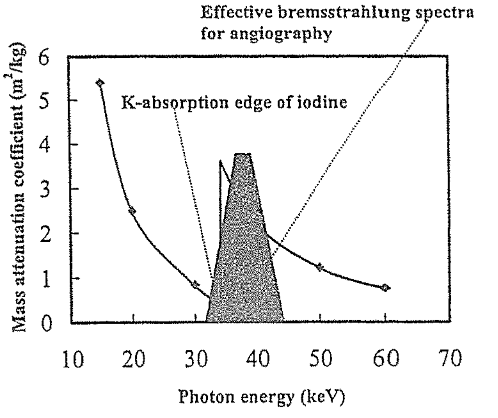

Fig. 1: Mass attenuation coefficients of iodine and effective bremsstrahlung x rays for enhanced K-edge angiography.

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 effective bremsstrahlung x-ray spectra for K-edge angiography are shown above the iodine K-edge. Because iodine contrast mediums with a K-absorption edge of 33.2 keV absorb the rays easily, blood vessels were observed with high contrasts.

3. Experimental setup

A steady state x-ray generator (SOFRON NST-1005) is shown in Fig. 2, and the maximum tube voltage and current are 100 kV and 5 mA, respectively. In this experiment, the tube voltage applied was from 45 to 65 kV, and the tube current was regulated to within 5.0 mA (maximum current) by the filament

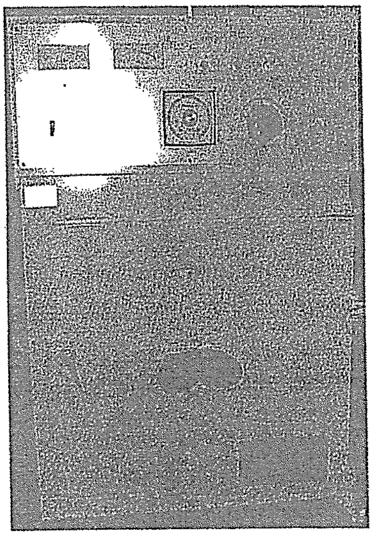

Fig. 2: X-ray generator.

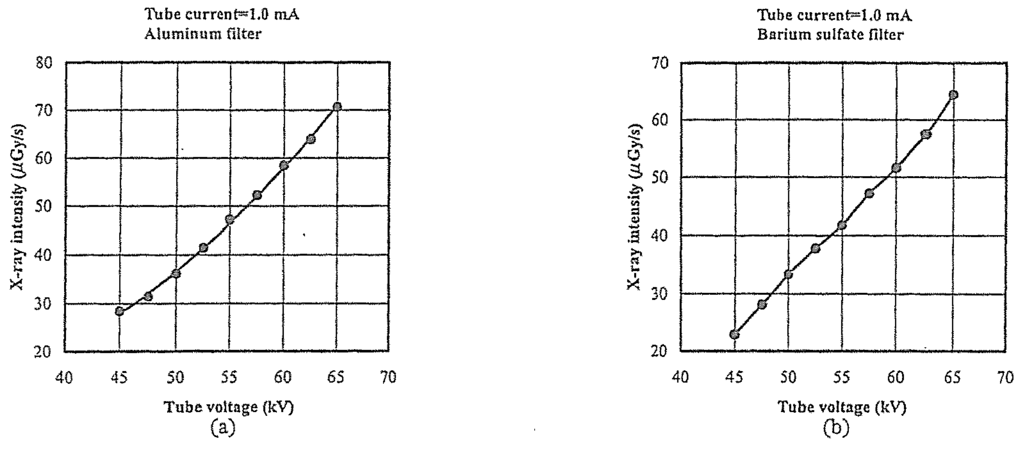

Fig. 3: X-ray intensity at 1.0 m per pulse with changing charging voltage (a) using aluminum filter and (b) using barium sulfate filter.

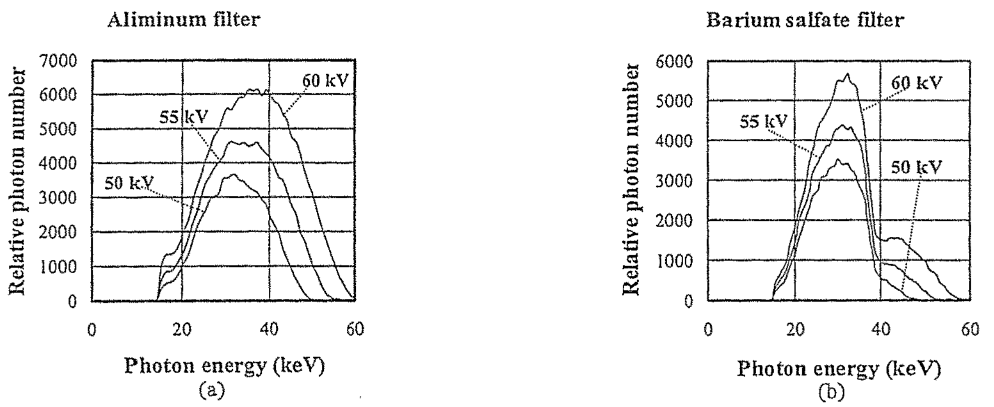

Fig. 4: X-ray spectra using (a) aluminum filter and (b) barium sulfate filter.

temperature. The exposure time is controlled in order to obtain optimum x-ray intensity. 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 the PMMA resin.

4. Characteristics

4.1 X-ray intensity

Figure 3 shows the x-ray intensity at 0.7 m per pulse measured by a Victoreen 660 ionization chamber. The x-ray intensity increased with increasing the tube voltage. At a tube voltage of 60 kV, the x-ray intensities obtained using an aluminum filter and a barium sulfate filter were 58.4 and 51.6 μGy/s, respectively, at 0.7 m from the x-ray source.

4.2 X-ray spectra

In order to measure x-ray spectra, we employed a cadmium tellurium detector (CDTE2020X, Hamamatsu Photonics K.K.) (Fig. 4). Compared with a germanium detector, this detector has a lower energy resolution of 1.7 keV. When the tube 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. The barium sulfate filter, with a surface density of approximately 10 mg/cm^2, significantly attenuate the spectra above the barium K-edge energy of 37.4 keV. The areas under the spectral curves correlate closely to the total x-ray intensities shown in Fig. 3

5. Angiography

The angiography was performed by the CR system (Konica Regius 150) using the filters with a tube voltage of 60 kV, and the distance between the x-ray source and the imaging plate was 0.7 m. The image contrast hardly varied even when the filter was changed.

Figure 5 shows radiograms of tungsten wires coiled around a rod made of polymethyl methacrylate using the aluminum filter. Although the image contrast increased with increases in the wire diameter, a 50-μm-diameter wire could be observed.

Figures 6 and 7 show angiograms of a rabbit thigh (barium sulfate filter) and a dog heart (aluminum filter), respectively. In angiography, iodine-based microspheres of 15 μm in diameter were used, and fine blood vessels of approximately 100 μm were visible.

6. Discussion

Concerning the spectrum measurement, we obtained bremsstrahlung x rays with narrow energy latitudes using both the aluminum and the barium sulfate filters. When the aluminum filter was employed with a tube voltage of 60 kV, the peak photon energy of spectra was approximately 35 kV.

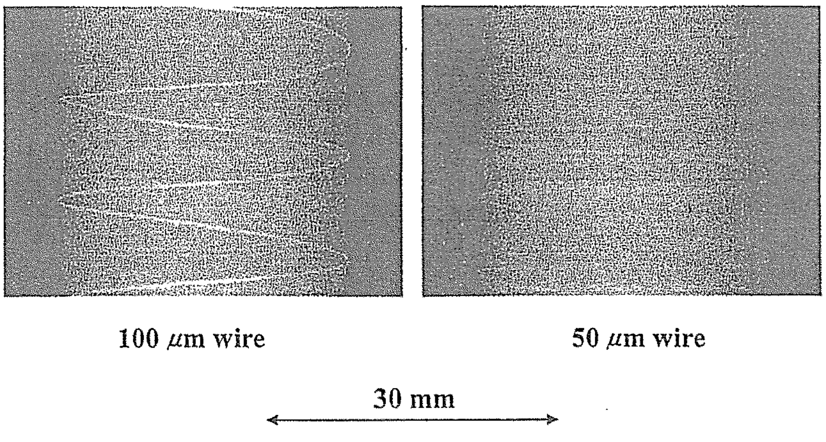

Fig. 5: Radiograms of tungsten wires coiled around rod made of polymethyl methacrylate using aluminum filter.

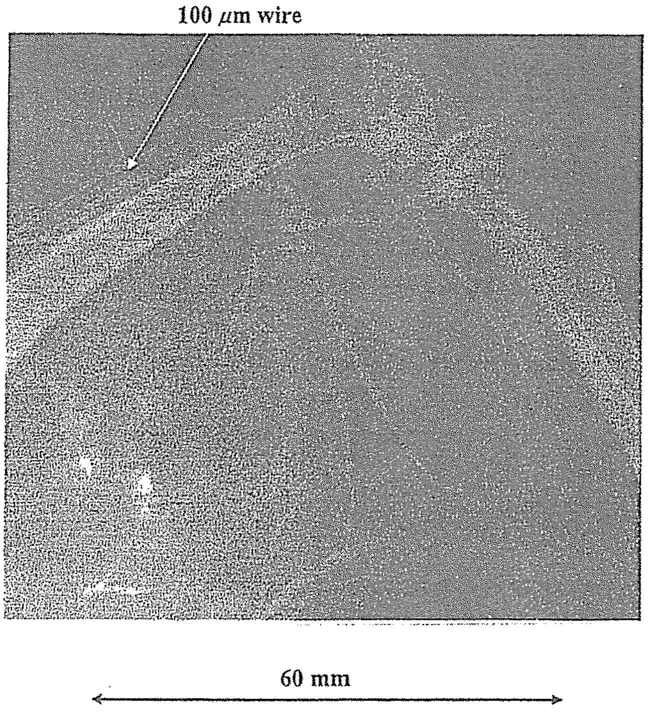

Fig. 6: Angiograms of rabbit thigh achieved with barium sulfate filter

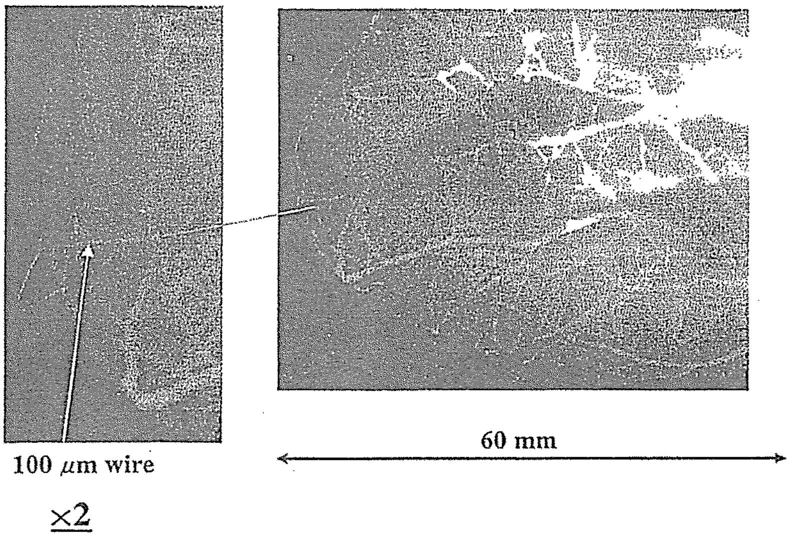

Fig. 7: Angiogram of dog heart with aluminum filter.

Therefore, the filter thickness should be increased as much as possible to decrease bremsstrahlung x rays of lower than K-absorption edge of iodine. Subsequently, using the barium sulfate filter, because the peak photon energy was nearly equal to the K-edge, aluminum filtering should be employed. In addition, cerium oxide filter is also useful in order to increase the peak energy and to decrease the low-photon-energy bremsstrahlung x rays.

In the present research, we employed a low-dose-rate x-ray generator in order to measure the x-ray spectra using a semiconductor detector. However, conventional medical x-ray generators with high dose rates can be employed to increase the tube current and to decrease the exposure time at a constant tube voltage.

With recent advances in angiography using MRI, if the density of gadolinium-based contrast media increases, enhanced K-edge angiography utilizing monochromatic x-ray generators, which produce Kα rays of ytterbium, tantalum, and tungsten, will be fairly useful technique to decrease the absorbed dose during angiography.

Acknowledgment

This work was supported by Grants-in-Aid for Scientific Research (13470154, 13877114, 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 Keiryo Research Foundation, The Promotion and Mutual Aid Corporation for Private Schools of Japan, Japan Science and Technology Agency (JST), and New Energy and Industrial Technology Development Organization (NEDO, Industrial Technology Research Grant Program in '03).

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Compact x-ray generator utilizing cerium-target tube for angiography

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ABSTRACT

The cerium-target x-ray tube is useful in order to perform cone beam K-edge angiography because K-series characteristic x rays from the cerium target are absorbed effectively by iodine-based contrast mediums. 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 were 65 kV and 0.4 mA, respectively, and the focal-spot sizes were 1.3×0.9 mm. Cerium K-series characteristic x rays were left using a 3.0 mm-thick aluminum filter, and the x-ray intensity was 0.59 μC/kg at 1.0 m from the source with a tube voltage of 60 kV, a current of 0.40 mA, and an exposure time of 1.0 s. Angiography was performed with a computed radiography system using iodine-based microspheres 15 μm in diameter. In angiography of non-living animals, we observed fine blood vessels of approximately 100 μm with high contrasts.

Key words: x-ray tube, cerium target, quasi-monochromatic x rays, characteristic x rays, K-edge angiography

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 mediums. However, it is difficult to obtain sufficient machine times for various research projects, including medical applications.

In order to perform high speed medical radiography, although several different flash x-ray generator[7-13] utilizing

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cold-cathode tubes have been developed, plasma flash x-ray generators[14-18] are useful to produce quasi-monochromatic x rays without using a K-edge filter. Therefore, we have performed a demonstration of cone-beam K-edge 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,[20] and have demonstrated enhanced K-edge angiography utilizing a barium sulfate filter. In the spectrum measurement, although Kα lines were produced, bremsstrahlung x rays with photon energies of lower than the barium K-edge (37.4 keV) were also observed. Therefore, optimum filtering for K-edge angiography should be selected to increase image contrast of fine blood vessels.

In the present research, we employed a compact x-ray generator with a cerium target tube, and used it to perform a preliminary study on cone beam K-edge angiography achieved with cerium characteristic x rays utilizing an aluminum filter.

2. GENERATOR

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 employed the Cockcroft-Walton circuit in order 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 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 in order to obtain optimum x-ray intensity. Quasi-monochromatic x rays are produced using a 3.0 mm-thick aluminum filter for absorbing soft bremsstrahlung rays.

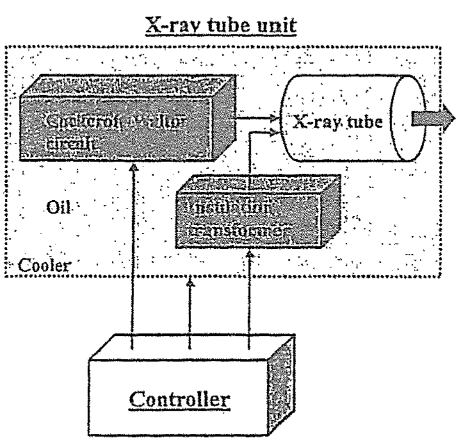

Fig. 1. Block diagram of compact x-ray generator with cerium-target radiation tube, which is used specially for K-edge angiography using iodine-based contrast mediums.

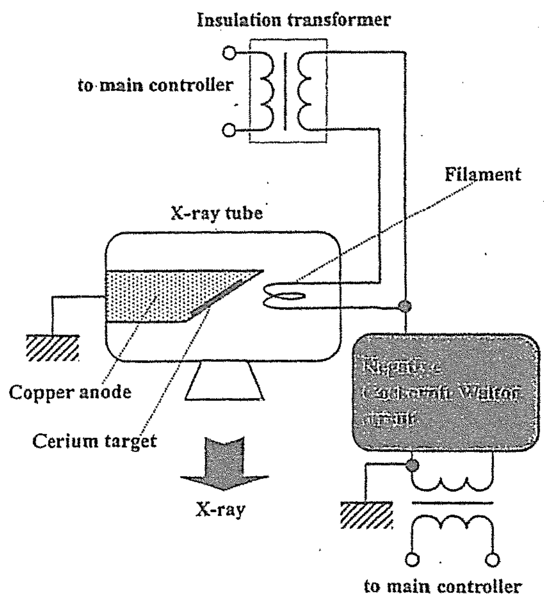

Fig. 2. Main circuit of x-ray generator.

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 using the filter with an exposure time of 1.0 s (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 0.59 μC/kg at 1.0 m from the source with errors of less than 0.2%.

3.2 Focal spot

In order 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

In order to measure x-ray spectra, we employed a cadmium tellurium detector (CDTE2020X, Hamamatsu Photonics Inc.) (Fig. 5). Compared with a germanium detector, this detector has a lower energy resolution of 1.7 keV. When the tube voltage was increased, the characteristic x-ray intensities of Kα and Kβ lines increased, and both the maximum photon energy and the intensities of bremsstrahlung x rays increased.

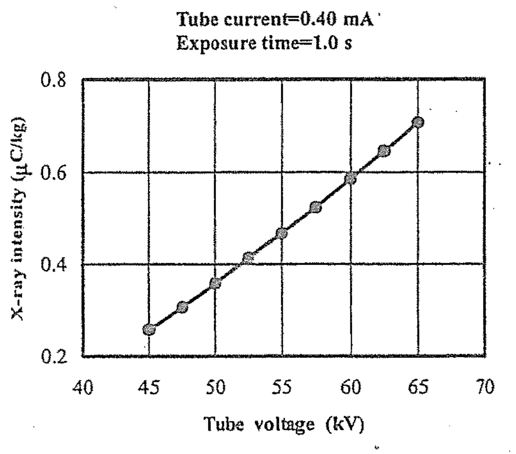

Fig. 3. X-ray intensity measured at 1.0 m from x-ray source according to changes in tube voltage.

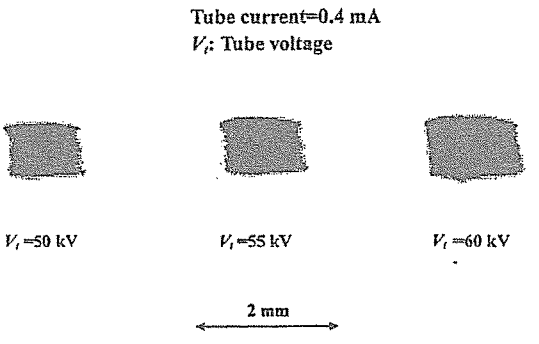

Fig. 4. Effective focal spots with changes in tube voltage.

4. 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. Cerium is a rare earth element and has a high reactivity; however, the average photon energies of Kα and Kβ lines are 34.6 and 39.2 keV, respectively, 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.

The angiography was performed by the CR system (Konica Regius 150) using the filter, and the distance (between the x-ray source and the imaging plate) was 1.5 m. Firstly, 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-diameter 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. In the case where the cerium spheres were employed, the coronary arteries were barely visible. Figure 9 shows an angiogram of a larger dog heart using the cerium target at a tube voltage of 60 kV using iodine spheres. For comparison, we show 3-dimensional image of the coronary arteries constructed from x-ray CT images by Pascal (Digital Culture Tech. Corp.) with a tungsten x-ray tube (Fig. 10). Using this imaging technique, fine blood vessel were not observed at all.

5. DISCUSSION

In summary, 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 mediums. The characteristic x-ray intensities increased with increases in the tube voltage, and low-photon-energy bremsstrahlung rays were 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 acceleration. Subsequently, the generator produced maximum number of characteristic photons was approximately 35M 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.
As compared with 3-dimensional blood images constructed from x-ray CT images by Pascal, fine blood vessels were visible. Because the sampling pitch of the CR system is 87.5 μm, we obtained spatial resolutions of approximately 100 μm. In order to observe fine blood vessels of less than 100 μm, the spatial resolution of the CR system should be improved as much as possible.

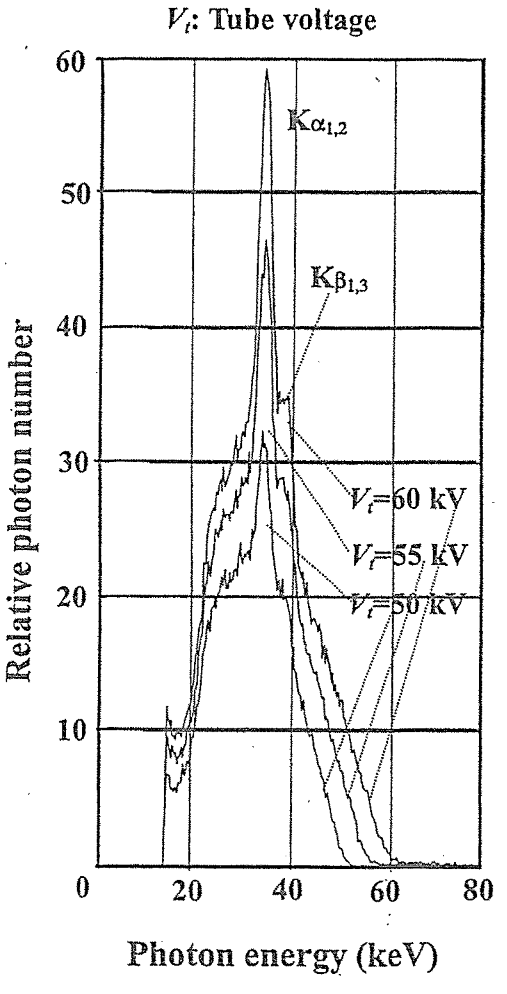

Fig. 5. X-ray spectra measured using cadmium tellurium detector with changes in tube voltage.

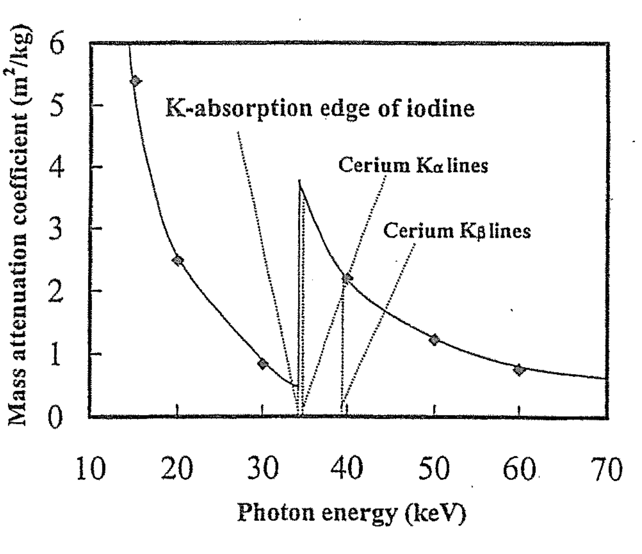

Fig. 6. Mass attenuation coefficients of iodine, and average photon energies of cerium Kα and Kβ lines.

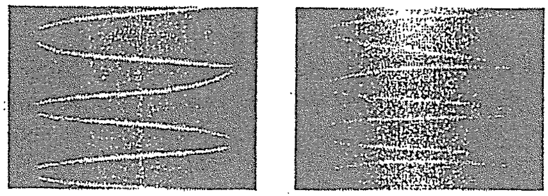

Fig. 7. Radiograms of tungsten wires in PMMA rod with tube voltage of 60 kV.

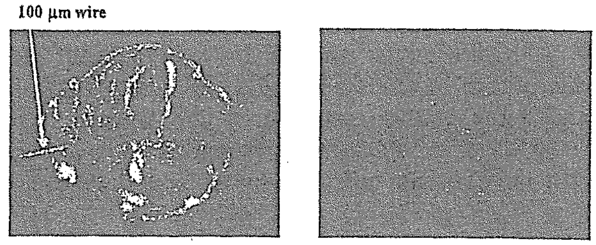

Fig. 8. Angiograms of extracted rabbit hearts using iodine and cerium microspheres with tube voltage of 60 kV.

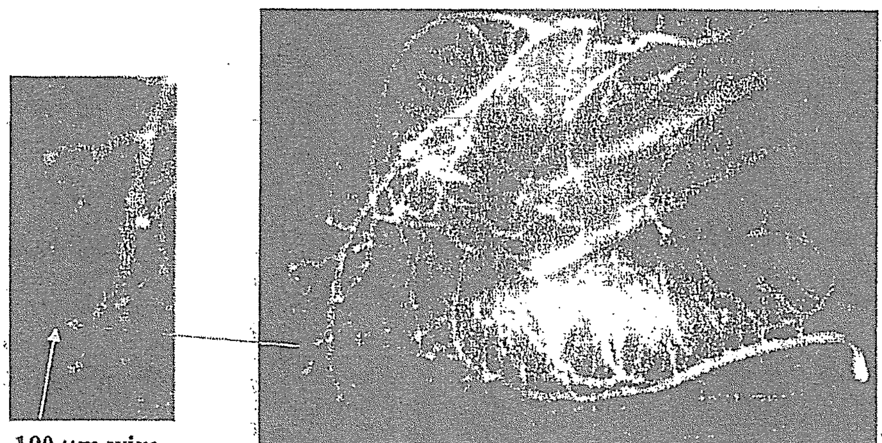

Fig. 9. Angiograms of extracted dog heart using iodine microspheres with tube voltage of 60 kV.

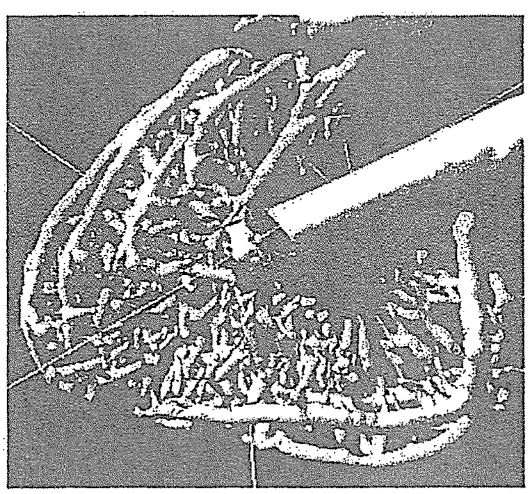

Fig. 10. 3-dimensional image of coronary arteries constructed from x-ray CT images by Pascal.

ACKNOWLEDGMENTS

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