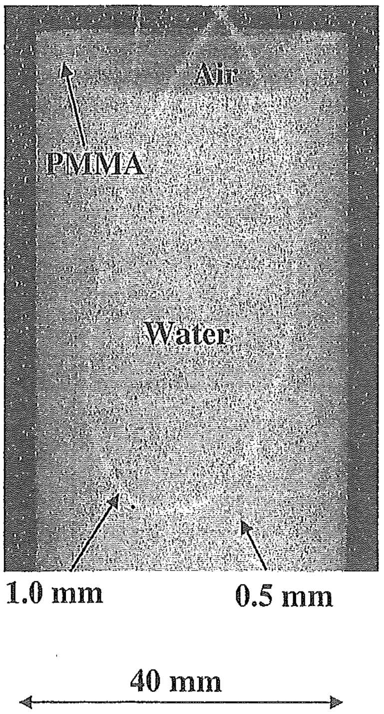

Fig. 10. Angiography of Teflon tube using contrast medium which contains approximately 65% gadodiamidehydrate.

PMMA case using a contrast medium which contains approximately 65% gadodiamidehydrate; the 0.5 mm tube can be observed easily. Figure 11 shows an angiogram of a rabbit head using gadolinium oxide powder, and fine blood vessels of approximately 100 μm were visible.

6. Discussion

In summary, we succeeded in producing K-series characteristic X-rays of tantalum and in performing K-edge angiography using gadolinium contrast media with a K-edge of 50.2 keV; this K-edge angiography could be a useful technique for reducing the dose absorbed by patients. Although we employed tantalum Kα (57.1 keV) and Kβ (approximately 65 keV) rays, Kβ rays should be absorbed using an ytterbium oxide filter in order to increase the image contrast of blood vessels. In addition, L-series characteristic rays should be absorbed before angiography using a tungsten or an ytterbium oxide filter. In these cases, the photon energies of the K-absorption edges of tungsten and ytterbium are 69.5 and 61.3 keV, respectively.

In cases where a high tube voltage beyond the critical excitation potential is applied, the optimum intensity for angiography can be controlled because the K-series characteristic intensity substantially increases with charging voltage. In this research, the generator-produced instantaneous number of K photons was approximately 1×10^9 photons/cm^2 per pulse at 1.0 m from the source.

Using this flash X-ray generator, because the photon energy of characteristic X-rays can be selected, quasi-monochromatic imaging, such as enhanced K-edge angiography using iodine contrast media and mammography, can be performed. In addition, steady-state monochromatic X-rays can be produced by a similar tube utilizing a hot cathode and a constant high-voltage power supply. In conjunction with the fine focusing technique, these monochromatic X-ray generators could be employed to perform K-edge angiography and X-ray phase-contrast radiography for edge enhancement.

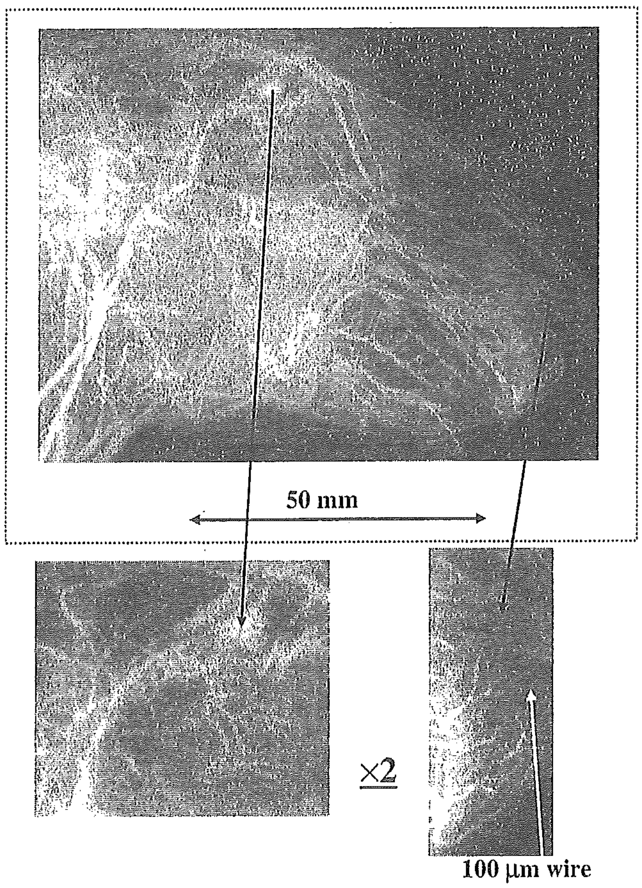

Fig. 11. Angiography of rabbit head using gadolinium oxide powder.

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Enhanced magnification angiography including phase-contrast effect using a 100-μm focus x-ray tube

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ABSTRACT

A microfocus x-ray tube is useful in order to perform magnification digital radiography including phase-contrast effect. The 100-μm-focus x-ray generator consists of a main controller for regulating the tube voltage and current and a tube unit with a high-voltage circuit and a fixed anode x-ray tube. The maximum tube voltage, current, and electric power were 105 kV, 0.5 mA, and 50 W, respectively. Using a 3-mm-thick aluminum filter, the x-ray intensity was 26.0 μGy/s at 1.0 m from the source with a tube voltage of 60 kV and a current of 0.50 mA. Because the peak photon energy was approximately 38 keV using the filter with a tube voltage of 60 kV, the bremsstrahlung x-rays were absorbed effectively by iodine-based contrast media with an iodine K-edge of 33.2 keV. Magnification angiography including phase-contrast effect was performed by three-time magnification imaging 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.

Keywords: high-contrast angiography, magnification digital radiography, microfocus x-ray tube, energy-selective imaging, phase-contrast effect

1. INTRODUCTION

Conventional flash x-ray generators utilizing condensers are useful in order to perform high-speed radiography including biomedical applications, and several different generators have been developed.[1-7] In particular, plasma flash x-ray generators[8-10] have been employed to produce clean K-series characteristic x-rays, and we have confirmed the irradiation of higher harmonic hard x-rays of Kα and Kβ lines. Without forming plasmas, demountable flash x-ray tubes can be employed to perform fundamental study on producing monochromatic x-rays,[11,12] and have succeeded in producing clean characteristic x-rays using angle dependence of bremsstrahlung x-ray distribution in Sommerfeld's theory. However, monochromatic flash radiography has had difficulties in increasing x-ray duration, and in performing magnification

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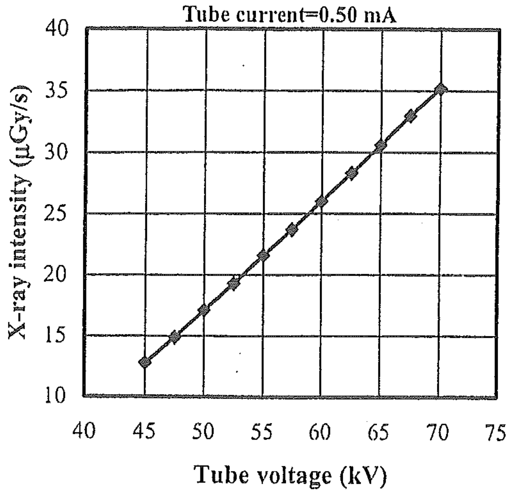


Figure 4: X-ray intensity (μGy/s) as a function of tube voltage (kV) with a tube current of 0.50 mA.

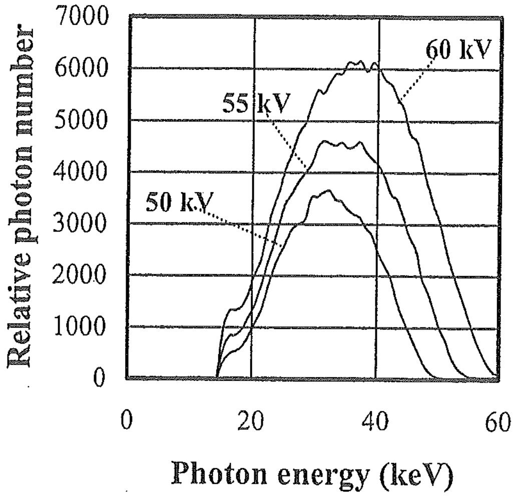


Figure 5. Bremsstrahlung x-ray spectra measured using a cadmium telluride detector with changes in the tube voltage.

4.3 *Magnification radiography*

The magnification radiography was performed by three-time magnification imaging using the CR system and the filter at a tube voltage of 60 kV, and the distance (between the x-ray source and the imaging plate) was 1.5 m (Fig. 6). Firstly, the spatial resolutions of conventional (cohesion) and magnification radiographies were made using a lead test chart. In the magnification radiography, 50 μm lines (10 line pairs) were clearly visible (Fig. 7). Subsequently, Fig. 8 shows radiograms of tungsten wires coiled around rods made of polymethyl methacrylate (PMMA). 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. Radiograms of one set of a bolt and a nut are shown in Fig. 9, the edge of a bubble in the bolt and the seam between the bolt and the nut are visible in magnification radiography.

4.4 *Enhanced magnification angiography*

The magnification angiography was performed at the same conditions using iodine microspheres of 15 μm in diameter, and the microspheres (containing 37% iodine by weight) are very useful for making phantoms of non-living animals used for angiography. Angiogram of a rabbit heart is shown in Fig. 10, and the coronary arteries are visible. Figure 11 shows angiograms of a larger dog heart using iodine spheres. Although the image contrast decreased slightly with increases in the thickness of the PMMA plate facing the x-ray source, the coronary arteries of approximately 100 μm were observed using a 100-mm-thick plate.

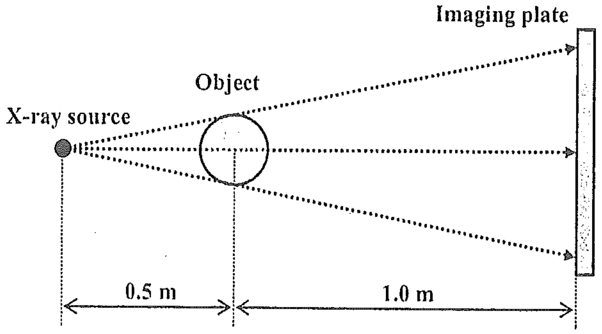


Figure 6. Three-time magnification imaging using an imaging plate in conjunction with a microfocus tube.

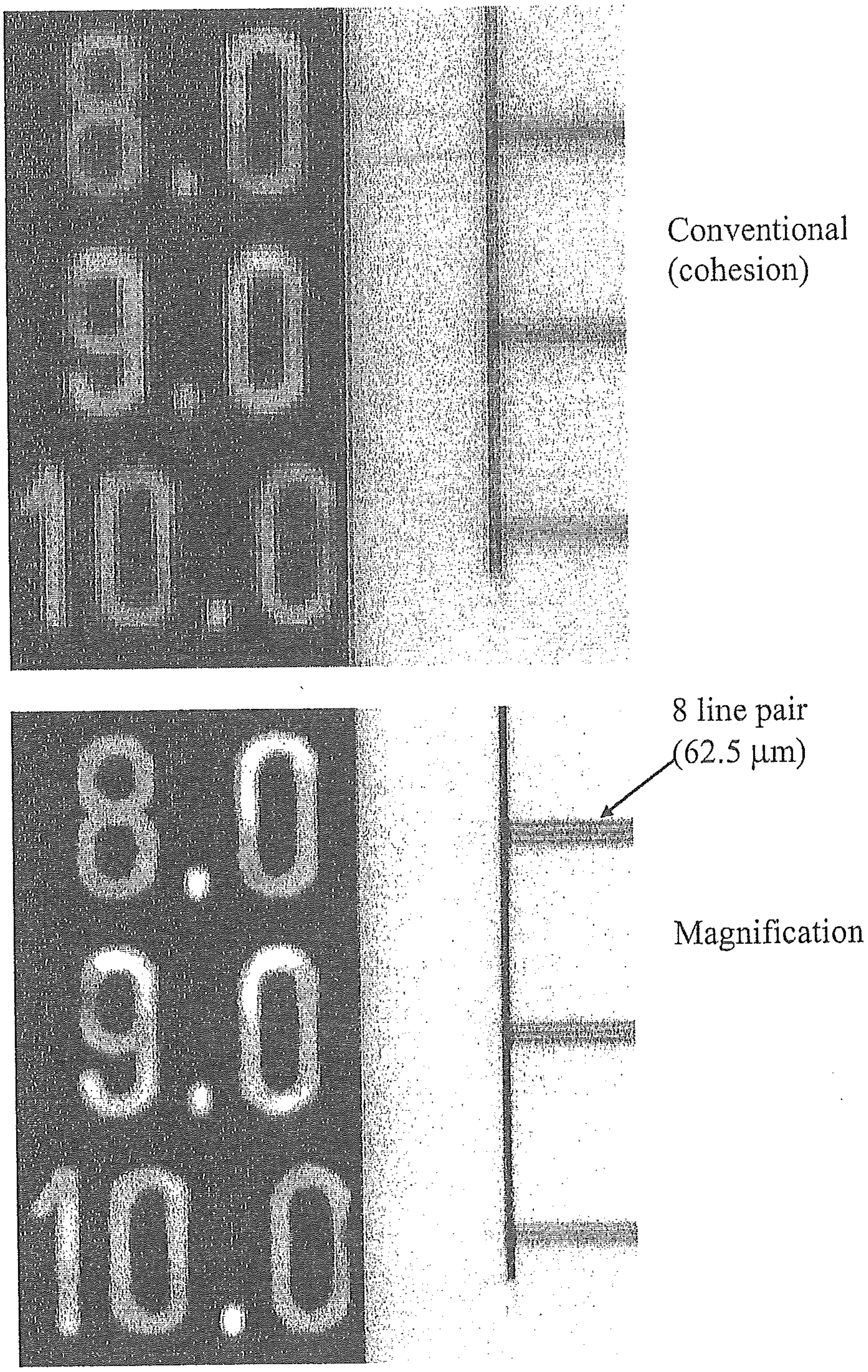
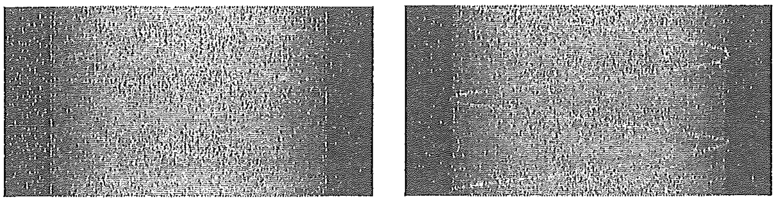
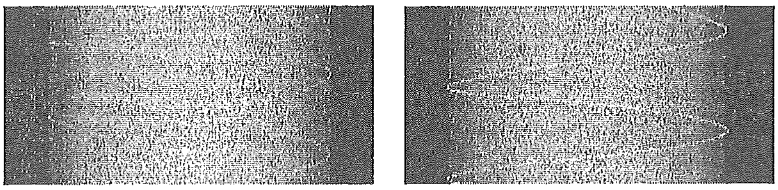


Figure 7. Radiogram of a test chart for measuring the spatial resolution.

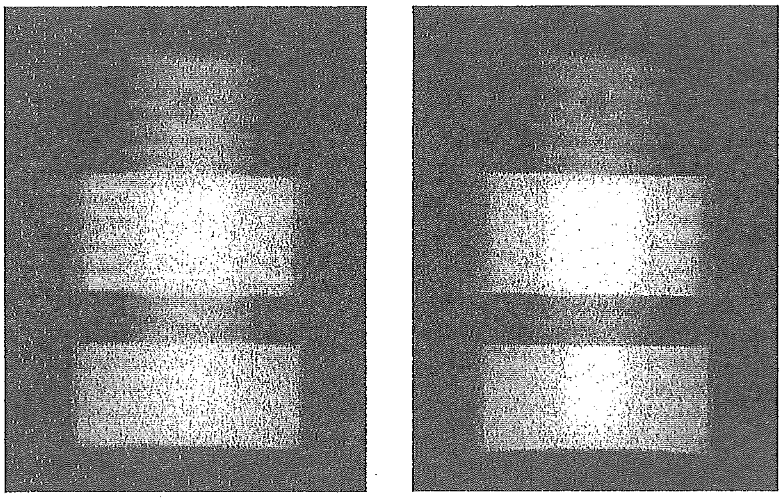


Conventional (cohesion)



Magnification

Figure 8. Radiograms of tungsten wires coiled around PMMA rods.



Conventional (cohesion)

Magnification

Figure 9. Radiograms of a set of a plastic bolt and a nut.

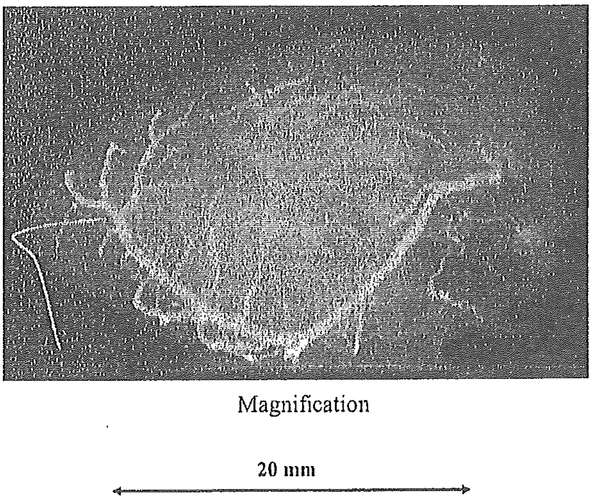
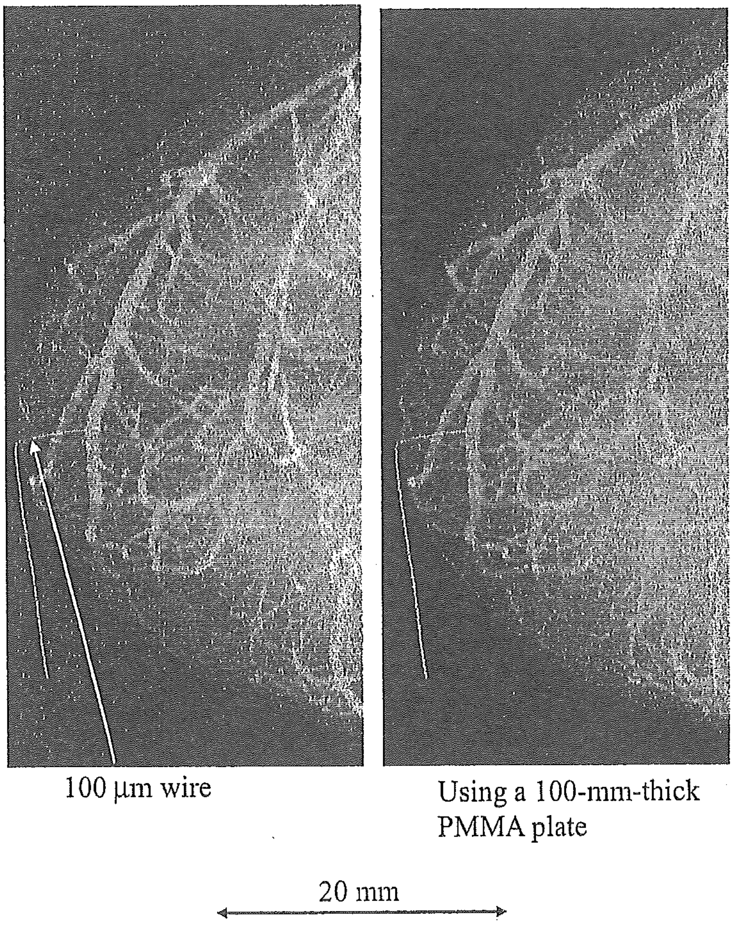


Figure 10: Angiogram of an extracted rabbit heart using iodine microspheres.


100 µm wire

Using a 100-mm-thick PMMA plate

Figure 11. Angiograms of an extracted dog heart.

5. CONCLUSIONS AND OUTLOOK

In summary, we employed an x-ray generator with a 100-μm-focus tungsten tube and performed enhanced magnification angiography including phase-contrast effect using narrow-photon-energy bremsstrahlung x-rays with a peak photon energy of approximately 38 keV, which can be absorbed easily by iodine-based contrast media. The bremsstrahlung x-ray intensity substantially increased with increases in the tube voltage, and the tube voltage was determined as 60 kV in order to increase the image contrast. In enhanced angiography, although we obtained almost absorption-contrast images, phase-contrast effect may be added in cases where low-density media are employed.

Because the sampling pitch of the CR system is 87.5 μm, we obtained spatial resolutions of approximately 50 μm using 3-time magnification imaging even when a 100-μm-focus tube was employed. In order to observe fine blood vessels of less than 100 μm, the spatial resolution of the CR system should be improved to 43.8 μm (Konica Minolta Regius 190), and the iodine density should be increased.

At a tube voltage of 60 kV and a current of 0.50 mA, the maximum number of photons was approximately 4×10^7 photons/cm^2·s at 1.0 m from the source, and the photon count rate can be increased easily using a rotating anode microfocus tube developed by Hitachi Medical Corporation. Recently, the maximum electric power of the microfocus x-ray tube has been increasing, and the kilowatt-range tube can be realized. Therefore, the dynamic magnification radiography is possible using a flat panel detector with a pixel size of less than 100 μm.

ACKNOWLEDGMENTS

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Monochromatic x-ray generator utilizing angle dependence of bremsstrahlung x-ray distribution

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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 (Wehnelt) 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 clean Kα rays are produced through the focusing electrode using a 20-μm-thick zirconium filter. The x-ray intensity was 12.1 μGy/s at 1.0 m from the x-ray source with a tube voltage of 30 kV and a tube current of 100 μA, and monochromatic radiography was performed using a computed radiography system.

Keywords: demountable x-ray tube, electron-impact source, monochromatic x-rays, Kα rays, Sommerfeld's theory

1. INTRODUCTION

Recently, we have developed several different flash x-ray generators[1-6] corresponding to specific radiographic objectives, and the plasma x-ray source has been growing with increases in the electrostatic energy in the condenser. By forming weakly ionized linear plasma[7-10] using rod targets, we confirmed irradiation of clean K-series characteristic x-rays such as hard x-ray lasers from the plasma axial direction using a table-top flash x-ray generator. This super fluorescence has been employed to perform cone-beam monochromatic radiography such as iodine K-edge angiography.[11] Furthermore, because higher harmonic hard x-rays have been produced from the copper plasma, we have to confirm the irradiations of higher harmonics with charges in the target element.

At present, brilliant monochromatic parallel x-ray beams from synchrotron radiation are used in various fields including medical imaging,[12-15] and large-scale x-ray free electron laser sources are constructing as a new-generation radiation

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source for producing monochromatic coherent x-rays. In contrast, small-scale steady-state monochromatic parallel and cone beams can be employed to perform medical imaging including phase-contrast radiography and K-edge angiography[16,17] in hospitals.

In this paper, we developed a monochromatic x-ray generator, used to perform a preliminary experiment for generating clean molybdenum Kα rays by angle dependence of the bremsstrahlung x-rays.

2. GENERATOR

Figure 1 shows a block diagram of a compact 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 rod target 3.0 mm in diameter, a tungsten hairpin cathode (filament), a focusing (Wehnelt) 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 (Fig. 3), clean molybdenum Kα rays can be produced using a 20-μm-thick zirconium filter.

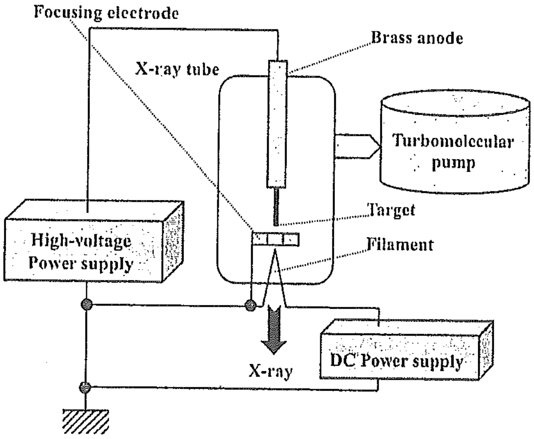


Figure 1: Block diagram including the main transmission line of the compact x-ray generator with a monochromatic diode.

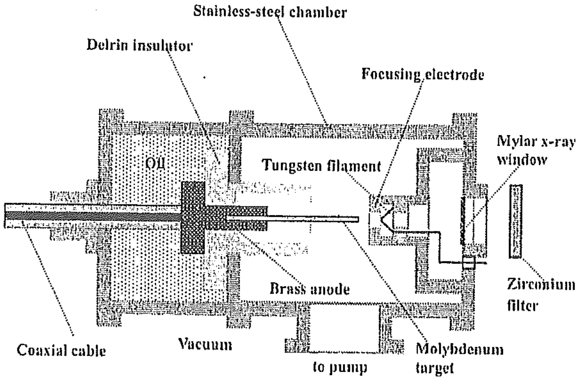


Figure 2: Schematic drawing of the monochromatic x-ray tube.

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. 4). At a constant tube current of 100 μA, the x-ray intensity increased when the tube voltage was increased. In this measurement, the intensity with a tube voltage of 30 kV and a current of 100 μA was 12.1 μGy/s at 1.0 m from the source.

3.2 X-ray source

In order to measure images of the x-ray source, we employed a pinhole camera with a hole diameter of 100 μm in conjunction with a computed radiography (CR) system[18] (Fig. 5). When the tube voltage was increased, the spot diameter slightly increased and had a maximum value of approximately 2.3 mm.

3.3 X-ray spectra

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 (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; the data was scanned by Dicom viewer in the film-less CR system. Subsequently, the relative x-ray intensity as a function of the data was calibrated using a conventional x-ray generator, and we confirmed that the intensity was proportional to the exposure time. Figure 6 shows measured spectra from the molybdenum target using the filter. We observed clean Kα lines, while bremsstrahlung rays were hardly detected. The Kα intensity substantially increased with increases in the tube voltage.

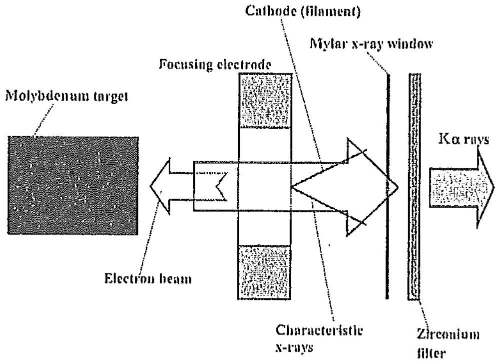


Figure 3: K-photon irradiation from the x-ray tube.

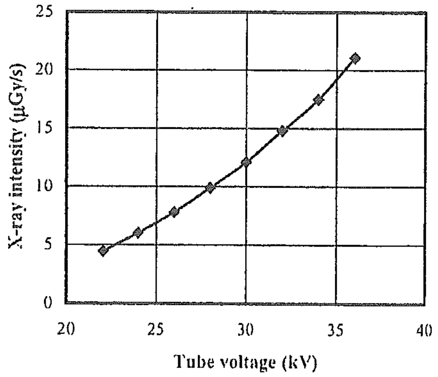


Figure 4: X-ray intensity at 1.0 m from the x-ray source according to changes in the tube voltage.

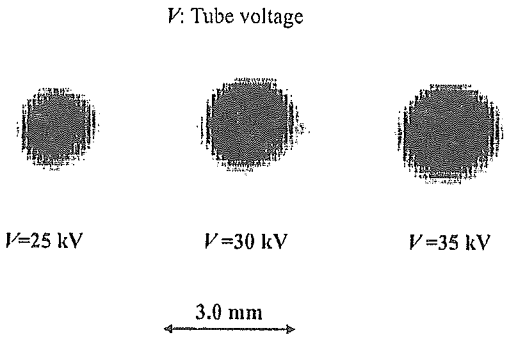


Figure 5: Images of the characteristic x-ray source obtained using a pinhole camera with changes in the tube voltage.

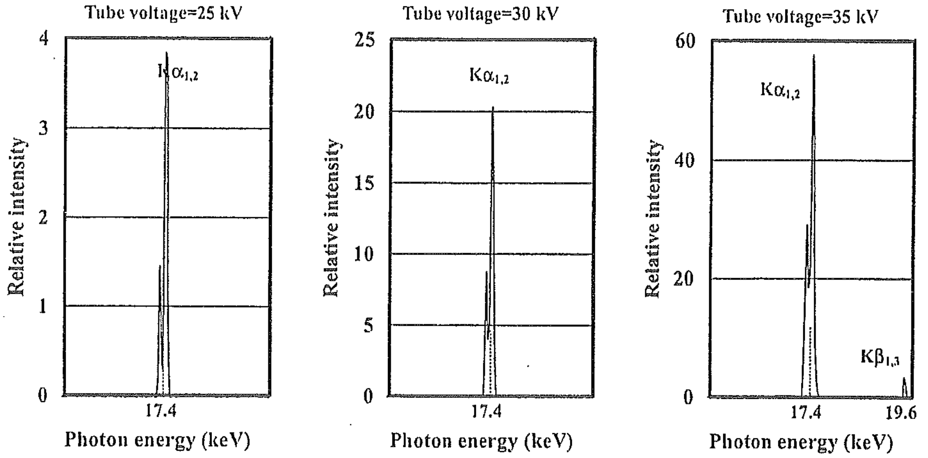


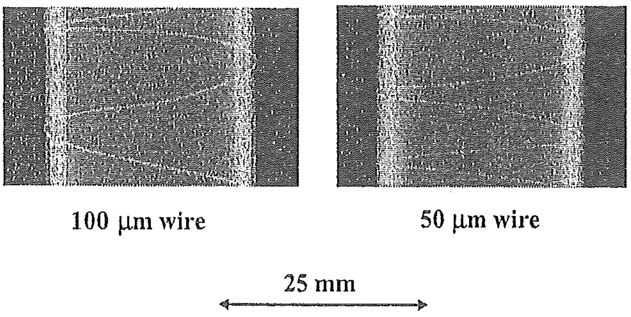
Figure 6: X-ray spectra from the molybdenum target. The spectra were measured using a transmission type spectrometer with a lithium fluoride curved crystal.

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.

Firstly, rough measurements of image resolution were made using wires. Figure 7 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. 8, and the fine structure of the vertebra was observed. Next, angiography was performed using iodine microspheres of 15 μm in diameter. Figures 9 and 10 show angiograms of a rabbit heart and thigh, respectively, and we could obtained high contrast images of coronary arteries and fine blood vessels.


100 μm wire

50 μm wire

Figure 7: 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.

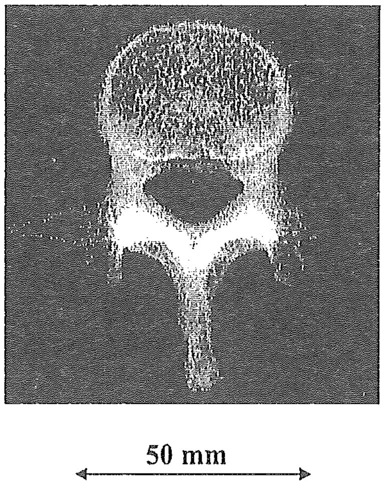


Figure 8: Radiogram of a vertebra. Fine structure of the vertebra were visible.

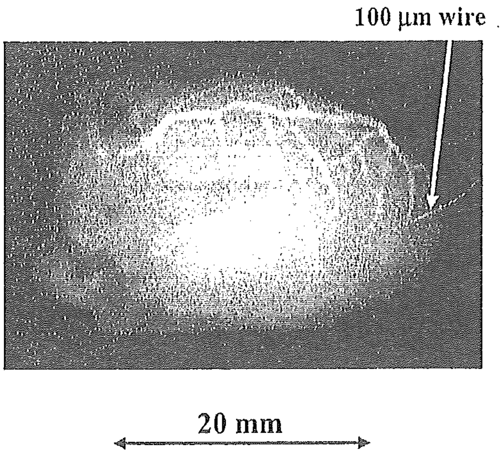


Figure 9: Angiograms of a rabbit heart. Coronary arteries were visible.

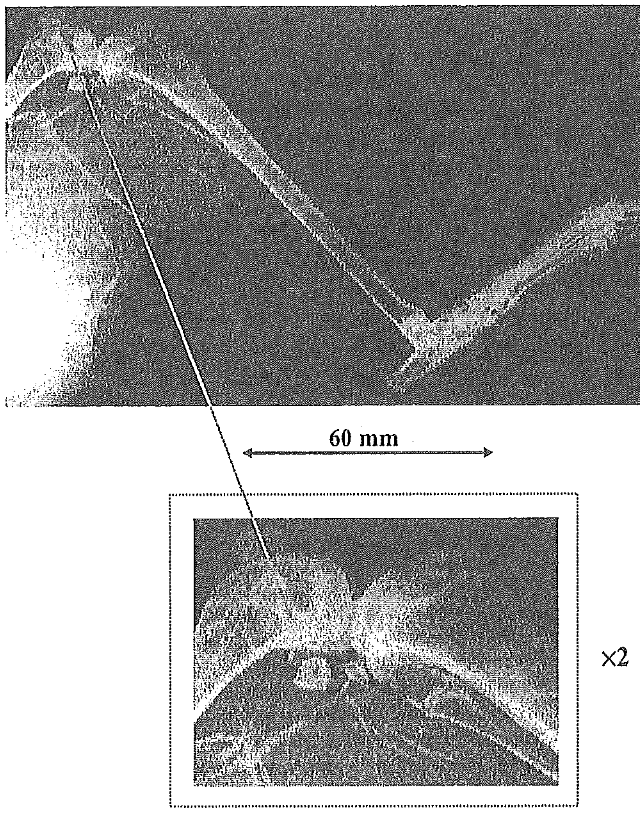


Figure 10: Angiogram of a rabbit thigh. Fine blood vessels of approximately 100 μm in diameter were visible.

5. CONCLUSIONS AND OUTLOOK

We developed a new monochromatic x-ray generator with a molybdenum-target tube and succeeded in producing clean molybdenum Kα lines. The Kα intensity increased with increases in the tube voltage, and monochromatic Kα rays were left by the zirconium filter. Without using the filter, bremsstrahlung x-rays were hardly observed.

In this experiment, although the maximum tube voltage and current were 36 kV and 0.10 mA, 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 maximum number of characteristic photons was approximately 5×10^6 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.

The molybdenum K-series characteristic x-rays are useful for mammography, and the photon energies of characteristic x-rays can be selected by the target element. In particular, enhanced K-edge angiography can be performed using a cerium target because cerium Kα rays (34.6 keV) are absorbed easily by iodine-based contrast media with an iodine K-edge of 33.2 keV. Furthermore, low-dose enhanced K-edge angiography can be performed utilizing a tungsten target in conjunction with gadolinium media.

Using these angiographies, coronary arteries and fine blood vessels formed in regenerative medicine may be observed with high contrasts. Furthermore, a flat panel detector is useful to observe blood flows for cases of cardiovascular disease.

ACKNOWLEDGMENTS

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Energy-selective gadolinium angiography utilizing a stroboscopic x-ray generator

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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 120 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 50 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 100 kV and a width of 1.0 ms, the x-ray intensity obtained using a 50-μm-thick tungsten filter was 9.88 μGy at 1.0 m, and the dimensions of the focal spot had values of approximately 1 × 1 mm. Angiography was performed using the filter at a charging voltage of 100 kV.

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

1. INTRODUCTION

Flash x-ray generators are capable of producing high-dose rate short x-ray pulses, and have been applied to high-speed radiography in various fields.[1] To produce hard flash x-rays with maximum photon energies of approximately 1 MeV, multistage Marx surge generators have been developed. Furthermore, induction linear accelerators have been developed

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(SPIE, Bellingham, WA, 2005) · 0277-786X/05/$15 · doi: 10.1117/12.621056

and improved to produce 10-MeV-order flash x-rays.[2] In contrast, 100-kV-order flash x-ray generators have been developed and applied to biomedicine. Subsequently, soft x-ray lasers have been produced using a gas-discharge capillary,[3-5] and clean K-series characteristic x-rays[6-9] and their higher harmonic hard x-rays have been produced from weakly ionized linear plasma.

In high-speed medical radiography, the repetition rate is one of the technical key parameters in real-time dynamic radiography. In view of this situation, we have developed two stroboscopic x-ray generators[10] and have succeeded in producing repetitive x-rays with a maximum repetition rate of approximately 50 kHz. These generators employ 500 nF condensers and hot-cathode tungsten tubes, and the duration can be controlled form 10 µs to 1.0 ms

Recently synchrotrons generate monochromatic parallel x-ray beams using a monochromator, and these beams have been employed to perform enhanced K-edge angiography.[11-13] To perform angiography, the beams with photon energies of approximately 35 keV have been used, because iodine contrast media with a K-absorption edge of 33.2 keV absorb the beams effectively. In view of this situation, we have developed x-ray generators with cerium-target tubes[14,15] which can produce $K\alpha$ rays (34.6 keV). Subsequently, we have performed energy-selective high-speed angiography[16] using quasi-monochromatic x-rays produced by the aluminum filtering.

Gadolinium-based contrast media with a K-edge of 50.2 keV have been employed to perform angiography in MRI, and the gadolinium density has been increasing. In view of this situation, $K\alpha$ rays of tantalum (57.1 keV)[17] and tungsten (58.9 keV) are also useful to perform angiography, because the $K\alpha$ rays are absorbed effectively by gadolinium media. As compared with angiography using iodine media, the absorbed dose can be decreased considerably utilizing angiography achieved with gadolinium media.

In this research, we employed a tungsten-target x-ray tube and performed a preliminary study on high-speed gadolinium angiography achieved with quasi-monochromatic x-rays produced by the tungsten 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 (Fig. 2). The main condenser of approximately 500 nF in the unit is charged up to 120 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. In this generator, positive and negative high voltages are applied to the anode and cathode electrodes, respectively.

The x-ray tube is a glass-enclosed hot-cathode triode and is composed of the following major parts: a rotating anode tube with a tungsten target, a focusing electrode, a hot cathode (filament), a 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 irradiation field.

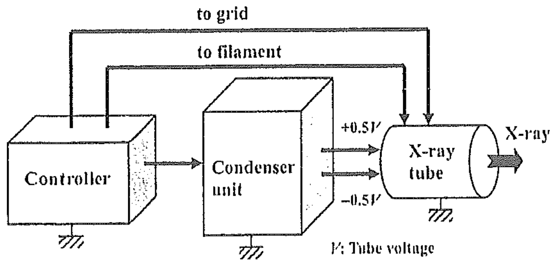

Figure 1: Block diagram of the kilohertz-range stroboscopic x-ray generator.

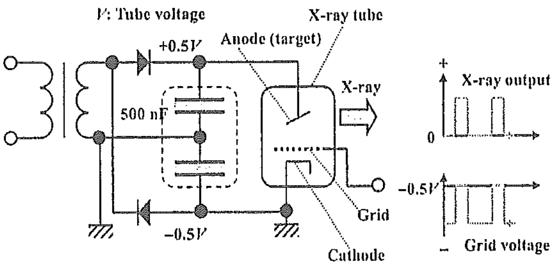

Figure 2: Main circuit of the kilohertz-range stroboscopic x-ray generator.

3. CHARACTERISTICS

3.1 X-ray output

The x-ray output signal was measured by a digital storage scope (Fig. 3) at the indicated conditions. 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.

3.2 Time-integrated x-ray intensity

Figure 4 shows the time-integrated (absolute) value of the x-ray intensity 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 with increases in the charging voltage. At a charging voltage of 100 kV and a width of 1.0 ms, the x-ray intensity obtained using a 50-μm-thick tungsten filter was 9.88 μGy per pulse 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 1 × 1 mm.

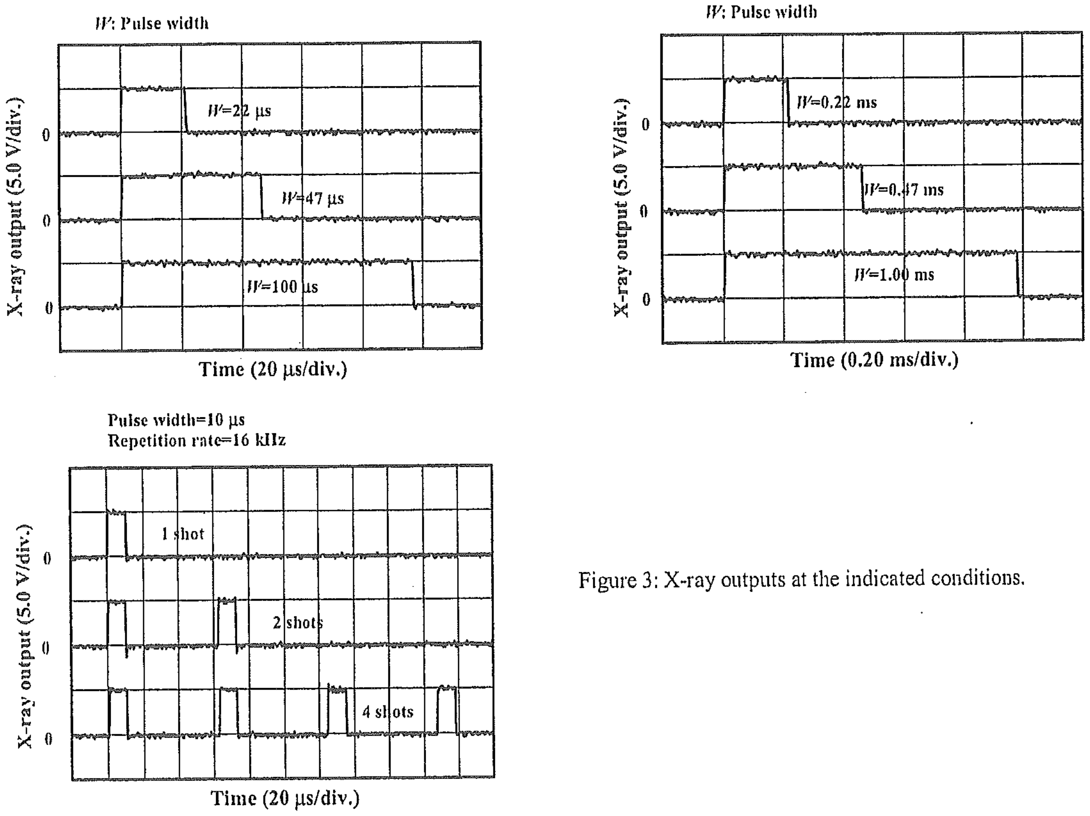

Figure 3: X-ray outputs at the indicated conditions.

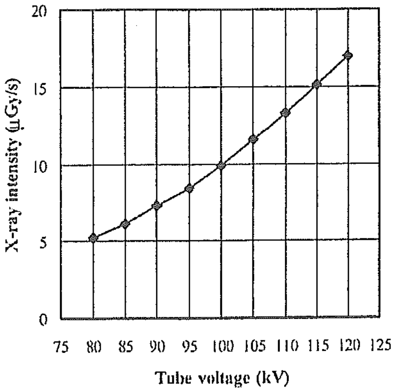

Figure 4: X-ray intensities at 1.0 m per pulse with changes in the charging voltage with an exposure time of 1.0 ms.

V: Tube voltage

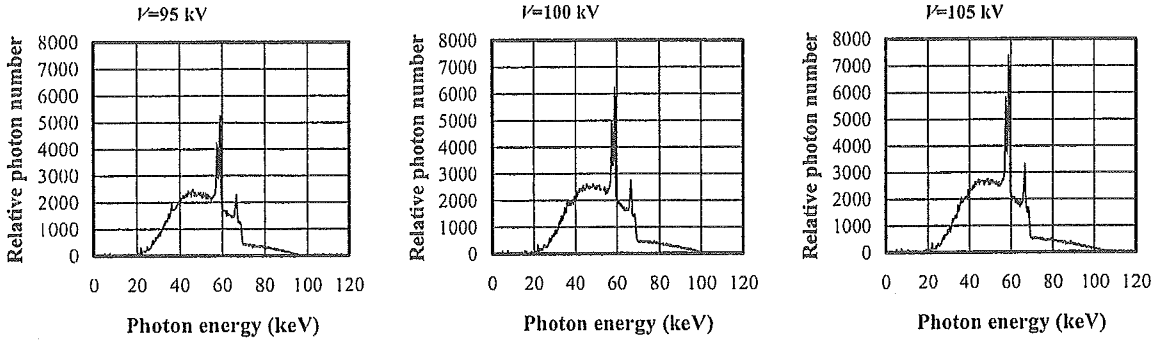

Figure 5: X-ray spectra at the indicated conditions.

3.4 X-ray spectra

In order to measure x-ray spectra with the filter, we employed a cadmium telluride detector (XR-100T, Amptek Inc.) (Fig. 5). 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 areas under the spectral curves correlate closely to the total x-ray intensities shown in Fig. 4.

4. ANGIOGRAPHY

Figure 6 shows the mass attenuation coefficients of gadolinium at the selected energies; the coefficient curve is discontinuous at the gadolinium K-edge. The average photon energy of the tungsten Kα lines is shown just above the gadolinium K-edge. The average photon energy of tungsten Kα lines is 58.9 keV, and gadolinium contrast media with a K-absorption edge of 50.2 keV absorb the lines easily. Therefore, blood vessels were observed with high contrasts.

The radiography was performed by the CR system using the filter with a charging voltage of 100 kV, and the distance between the x-ray source and the imaging plate was 1.0 m. The image contrast hardly varied even when the filter was changed.

Firstly, rough measurements of spatial resolution were made using wires. Figure 7 shows radiograms of tungsten wires coiled around rods made of polymethyl methacrylate (PMMA). Although the image contrast increased with increases in the wire diameter, a 50 μm-diameter wire could be observed. Next, the time resolutions were roughly observed using a plastic bullet from an air gun. Although we obtained completely stop-motion images of a bullet utilizing multi-shot radiography with a duration of 10 μs, the average velocity could be measured with durations of sub-milliseconds (Fig. 8).