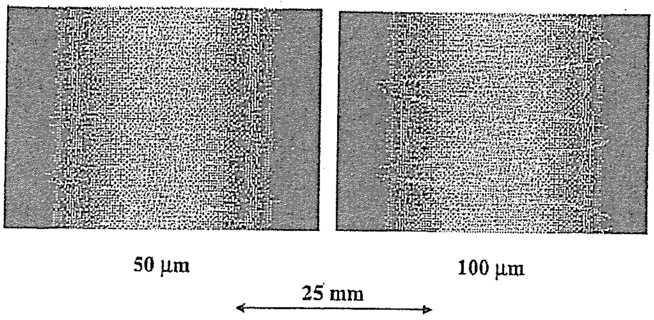

Figure 8: Radiograms of tungsten wires coiled around rod made of polymethyl methacrylate.

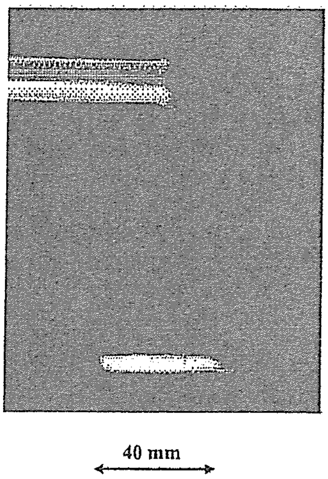

Figure 9: Radiogram of water falling into polypropylene beaker from glass test tube.

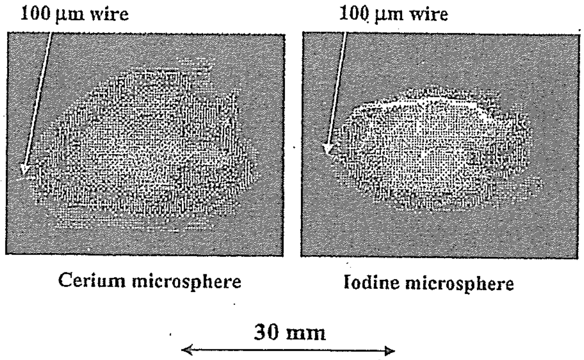

Figure 10: Angiograms of rabbit hearts using iodine and cerium microspheres.

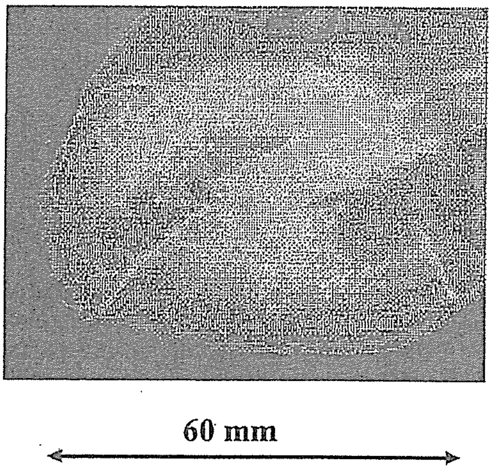

Fig. 11 Angiograms of extracted heart of dog.

6. DISCUSSION

Concerning the spectrum measurement, we obtained fairly clean cerium Kα and Kβ lines. Therefore, we are very interested in the measurement the characteristic rays from nickel, copper, molybdenum, silver, and tungsten targets; the target element should be selected corresponding to the radiographic objectives.

In this research, the generator produced instantaneous number of K photons was approximately 5×10^8 photons/cm^2 per pulse at 1.0 m from the source. Subsequently, the intensity can be increased by increasing the electrostatic energy in condenser, and monochromatic Kα lines are produced using a barium oxide filter with a barium K-edge of 37.4 keV.

Using this flash x-ray generator, as high output voltages can be produced using cables, high-photon-energy K-series characteristic x rays can be produced by increasing the atomic number of the target element. 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 from ytterbium, tantalum, and tungsten targets, will be a 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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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 4×10^7 photons/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-

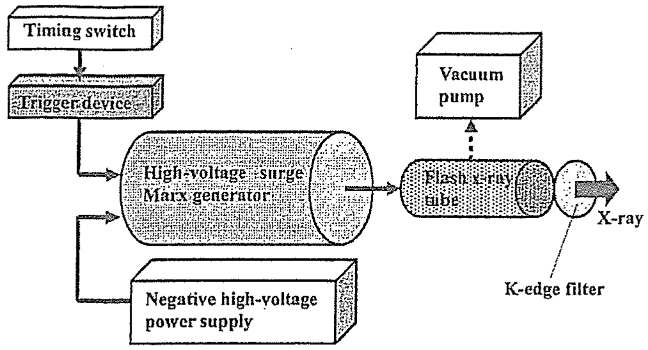

Fig. 1 Block diagram of the compact quasimonochromatic flash x-ray generator.

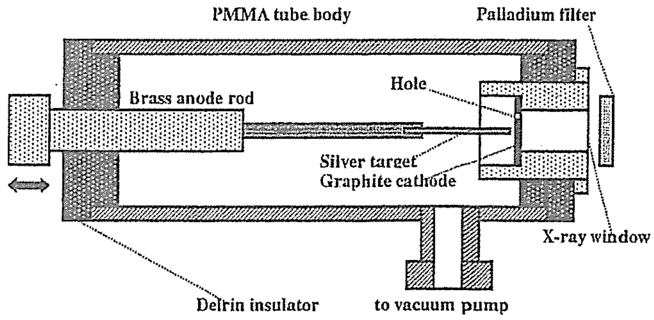

Fig. 3 Schematic drawing of the flash x-ray tube.

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 Kα rays can be produced using a 30-μm-thick palladium K-edge filter. In the K-series characteristic x-ray irradiation, Kα rays are left by absorbing Kβ 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

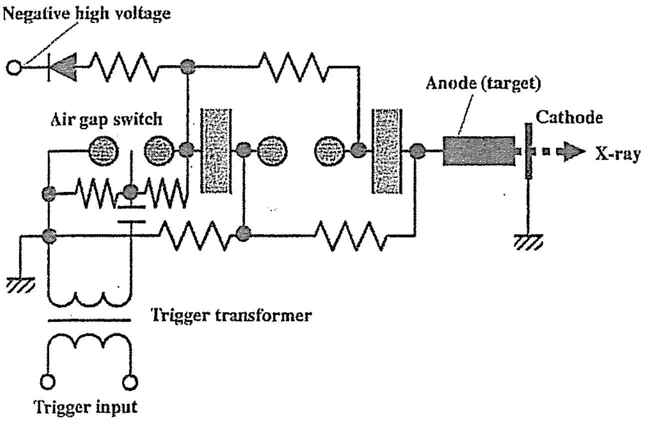

Fig. 2 Circuit diagram of the flash x-ray generator.

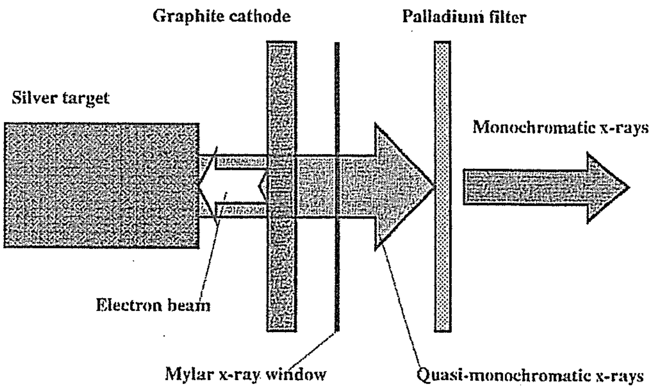

Fig. 4 Irradiation of silver Kα rays using a palladium K-edge filter.

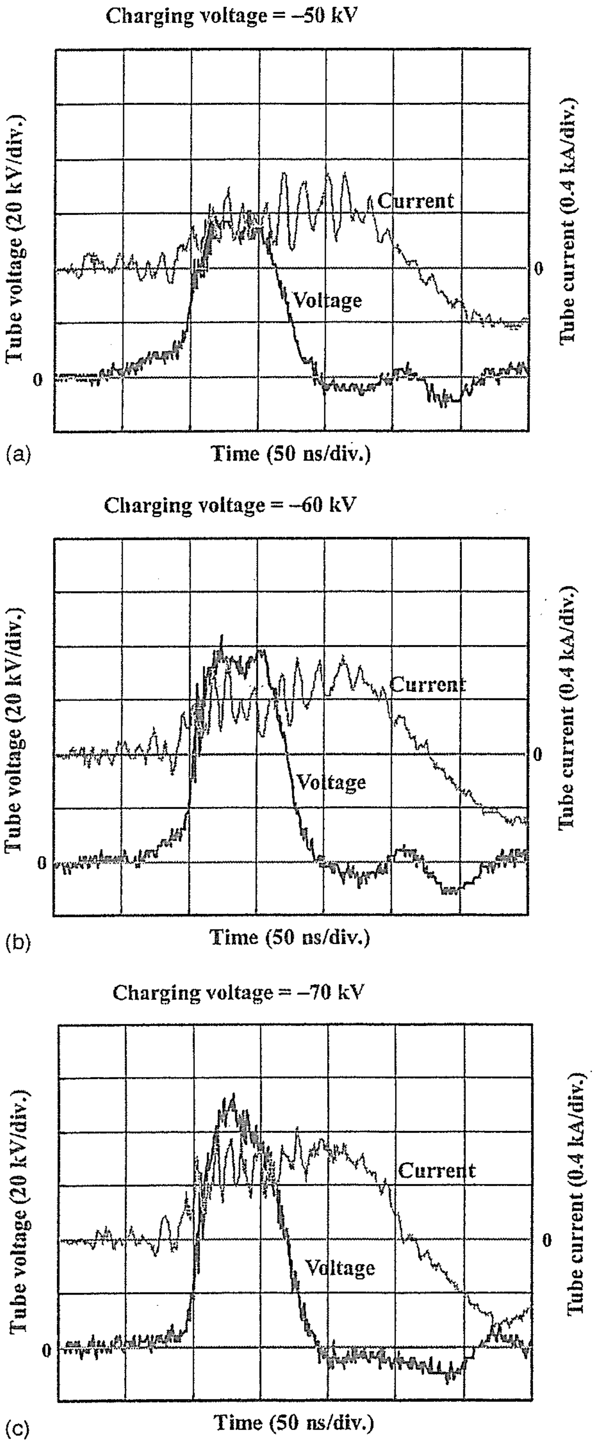

Fig. 5 Variations in tube voltage and current with charging voltages of (a) −50 kV, (b) −60 kV, and (c) −70 kV.

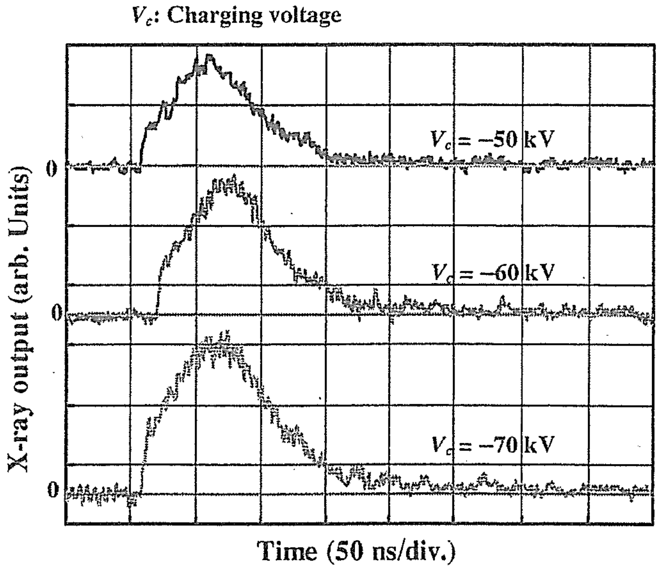

Fig. 6 X-ray outputs at the indicated conditions using the filter.

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.

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

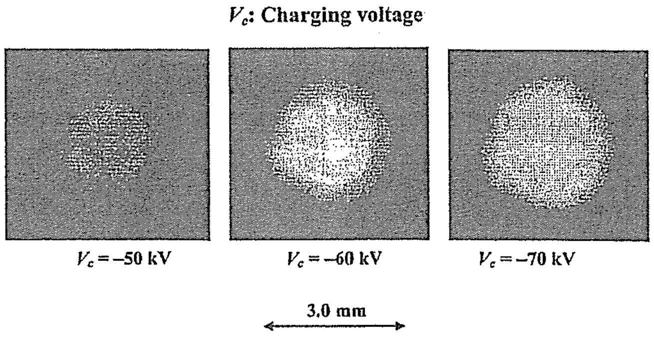

Fig. 7 Images of x-ray sources of Kα lines with changes in the charging voltage.

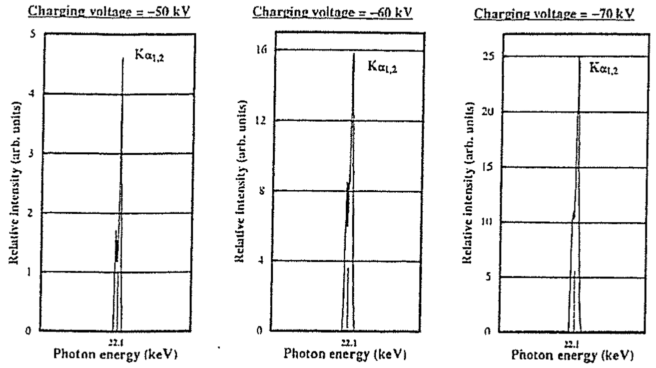

Fig. 8 X-ray spectra from a silver target with the filter.

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 $K\alpha$ lines, while bremsstrahlung rays were hardly detected. When the charging voltage was increased, the instantaneous tube voltage and current increased, and the $K\alpha$ intensity substantially increased.

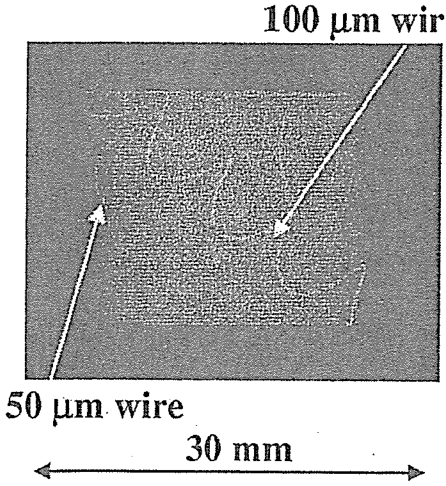

Fig. 9 Radiograms of tungsten wires of 50 and 100 μm in a rod made of polymethyl methacrylate.

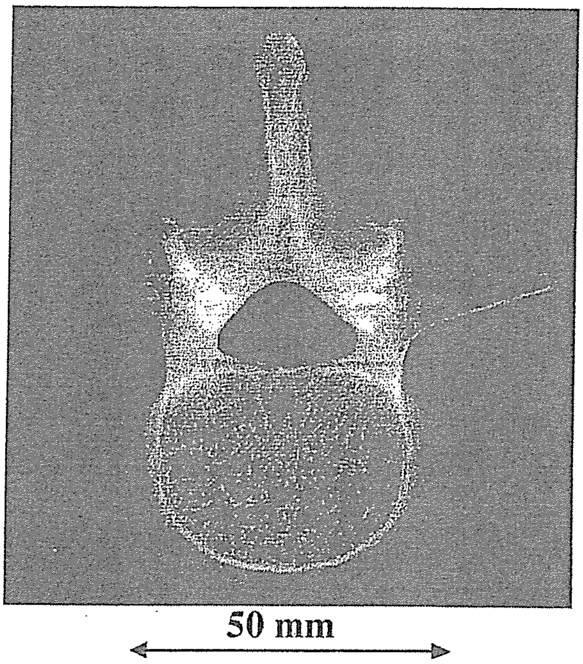

Fig. 10 Radiogram of a vertebra.

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-

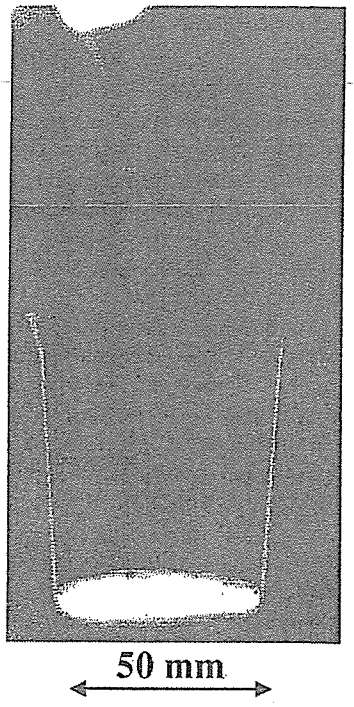

Fig. 11 Radiogram of water falling into a polypropylene beaker from a glass test tube.

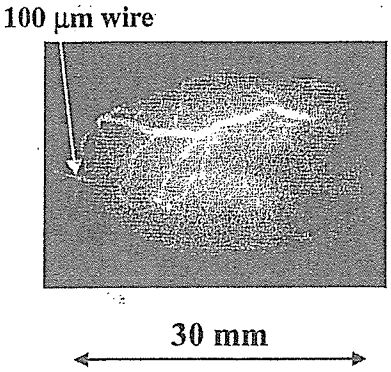

Fig. 12 Angiogram of a rabbit heart.

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

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 the 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 2003).

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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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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.

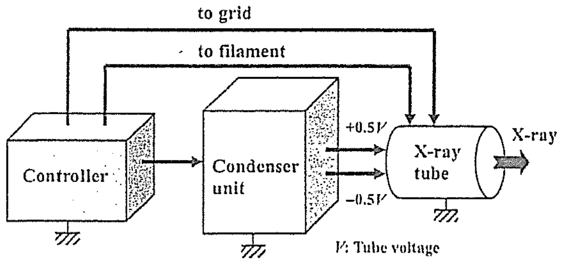


Figure 1: Block diagram of the kilohertz-range stroboscopic x-ray generator.

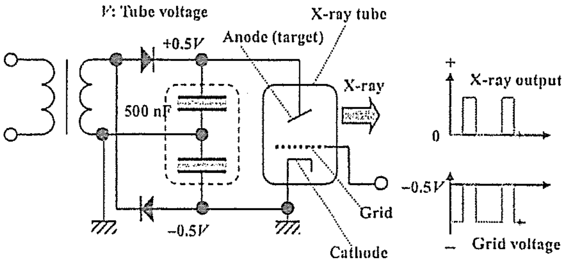


Figure 2: Main circuit of the kilohertz-range stroboscopic x-ray generator.

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.

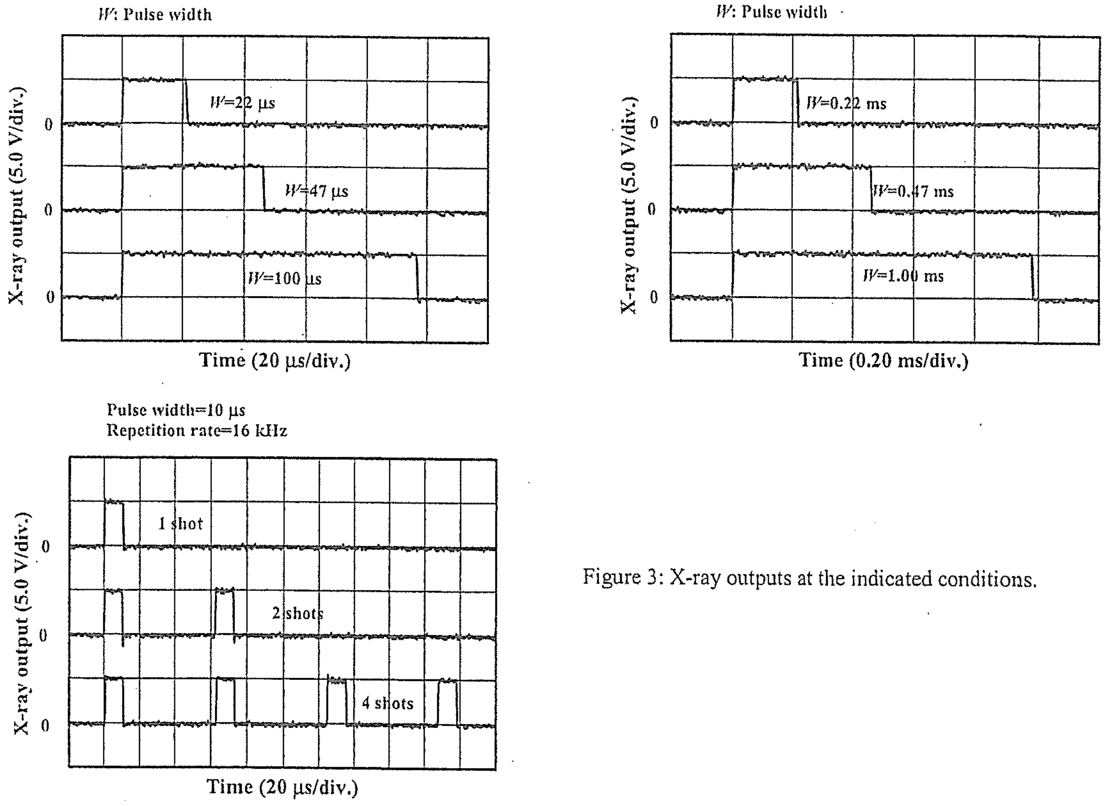


Figure 3: X-ray outputs at the indicated conditions.

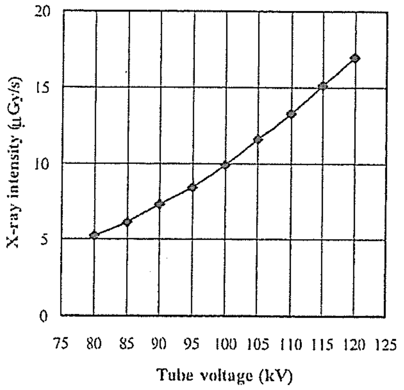


Figure 4: X-ray intensities at 1.0 m per pulse with changes in the charging voltage with an exposure time of 1.0 ms.

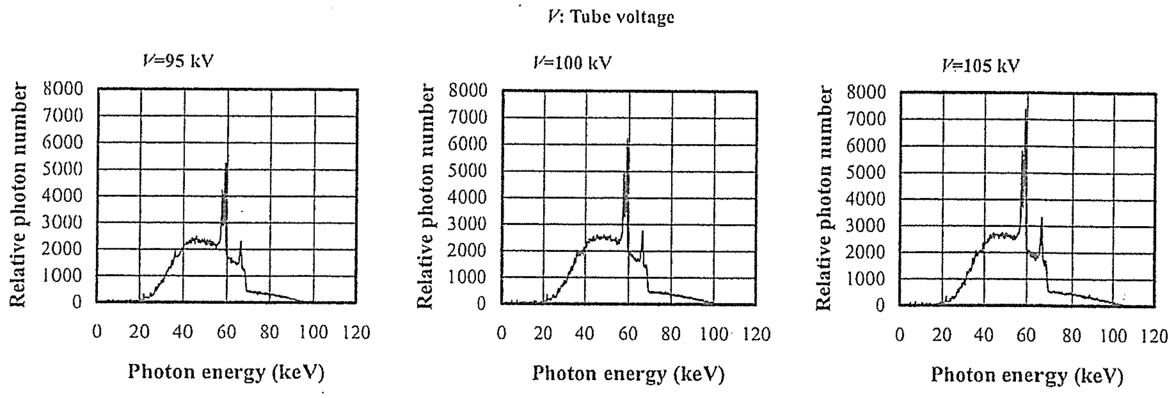


Figure 5: X-ray spectra at the indicated conditions.

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).

The image of water (20% gadolinium oxide suspension) falling into a polypropylene beaker from a plastic test tube is shown in Fig. 9. Because the x-ray duration was 1.0 ms, the stop-motion image of water could be obtained.

Figure 10 shows an angiogram of a polytetrafluoroethylene (Teflon) tube in a PMMA case using a contrast medium which contains approximately 65% gadodiamidehydrate with a duration of 1.0 ms, and a high-contrast tube with a bore diameter of 1.0 mm is observed. Figures 11 and 12 show angiograms of a rabbit ear and head using gadolinium oxide powder with a duration of 1.0 ms, and fine blood vessels of approximately 100 μm were visible.

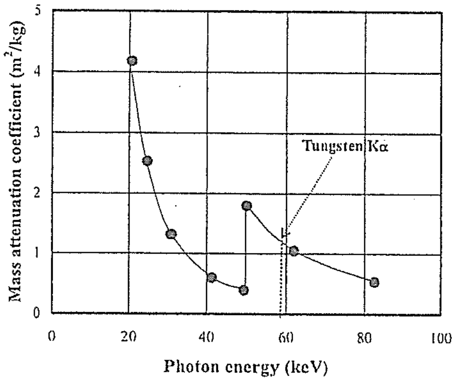
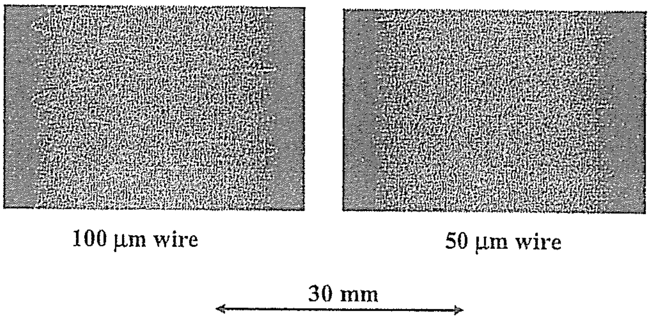


Figure 6: Mass attenuation coefficient of gadolinium and the average photon energy of tungsten Kα lines is shown above gadolinium K edge.



100 μm wire

50 μm wire

Figure 7: Radiograms of tungsten wires coiled around PMMA rods.

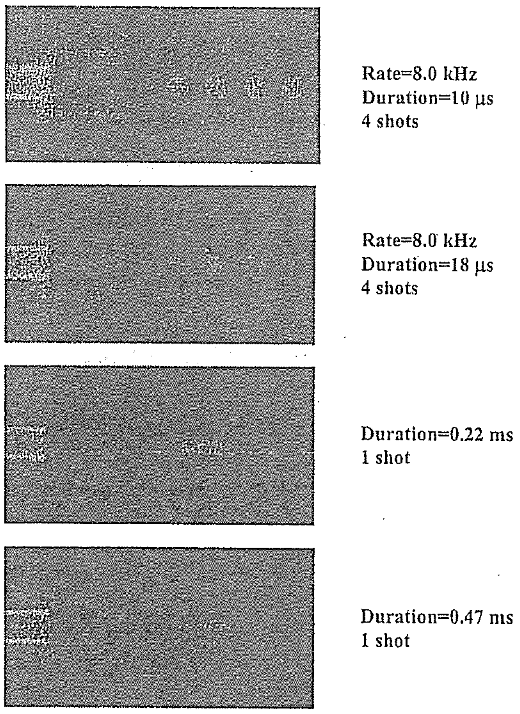


Figure 8: Radiograms of plastic bullets from an air gun at the indicated conditions.

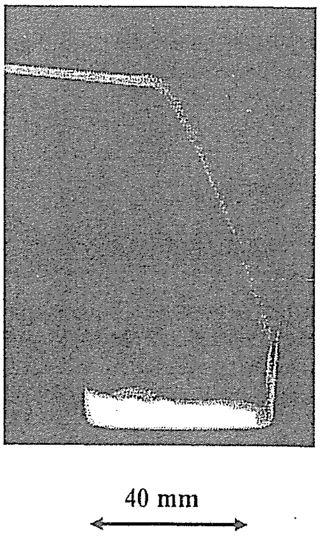


Figure 9: Radiogram of water falling into a polypropylene beaker from a glass test tube.

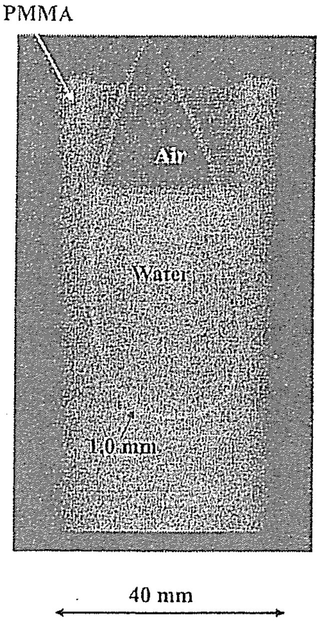


Figure 10: Angiography of a Teflon tube using a contrast medium which contains approximately 65% gadodiamidehydrate.

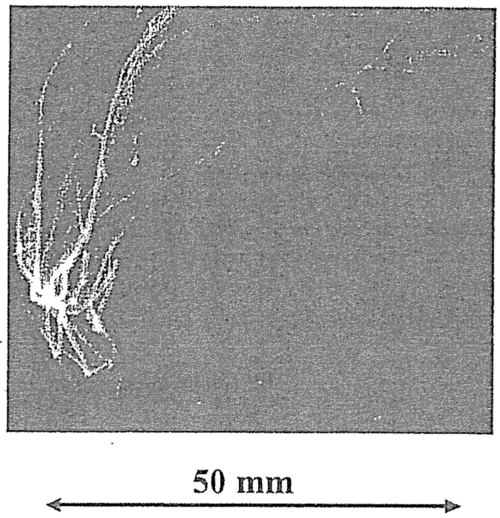


Figure 11: Angiography of a rabbit ear using gadolinium oxide powder.

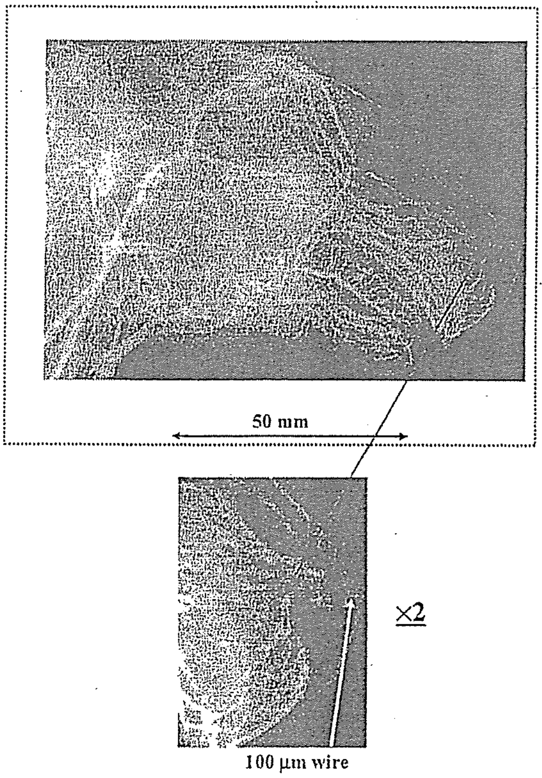


Figure 12: Angiography of a rabbit head using gadolinium oxide powder.

5. DISCUSSION

In summary, we succeeded in performing high-speed enhanced angiography utilizing tungsten K-series characteristic x-rays and gadolinium contrast media. As compared with angiography using iodine media, the absorbed dose could be decreased utilizing angiography achieved with gadolinium media.

Concerning the spectrum measurement, we obtained K-series characteristic x-rays using the tungsten filter. When the filter was employed with a charging voltage of 100 kV, the peak photon energy of the spectra was approximately 50 kV. Therefore, the filter thickness should be increased in order to decrease bremsstrahlung x-rays with energies lower than the K-absorption edge of tungsten. In the imaging, we have to consider the filtering effect of human body. Subsequently, Kβ rays should be absorbed using an ytterbium oxide filter in order to improve the image contrast of blood vessels.

Using this filter with a charging voltage of 100 kV and a pulse width (exposure time) of 1.0 ms, although we obtained the x-ray intensities of approximately 10 μGy at 1.0 m per pulse, the intensity should be maximized by increasing the tube current in order to improve the image quality using the CR system.

Nowadays, because flat panel detectors are very useful in order to perform real-time dynamic imaging with high spatial resolutions of 100 μm or less, stop-motion images of blood flows can be obtained using gadolinium media.

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Enhanced K-edge plasma angiography achieved with tungsten Kα rays utilizing gadolinium-based contrast media

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ABSTRACT

The tungsten plasma flash x-ray generator is useful in order to perform high-speed enhanced K-edge angiography using cone beams because Kα rays from the tungsten target are absorbed effectively by gadolinium-based contrast media. In the flash x-ray generator, a 150 nF condenser is charged up to 80 kV by a power supply, and flash x-rays are produced by the discharging. The x-ray tube is a demountable diode, and the turbomolecular pump evacuates air from the tube with a pressure of approximately 1 mPa. Since the electric circuit of the high-voltage pulse generator employs a cable transmission line, the high-voltage pulse generator produces twice the potential of the condenser charging voltage. At a charging voltage of 80 kV, the estimated maximum tube voltage and current were approximately 160 kV and 40 kA, respectively. When the charging voltage was increased, the characteristic x-ray intensities of tungsten Kα lines increased. Using an ytterbium oxide filter, the Kα lines were clean, and hardly any Kβ lines and bremsstrahlung rays were detected. The x-ray pulse widths were approximately 60 ns, and the time-integrated x-ray intensity had a value of approximately 50 μGy at 1.0 m from the x-ray source with a charging voltage of 80 kV. Angiography was performed using a film-less computed radiography system and gadolinium-based contrast media. In angiography of non-living animals, we observed fine blood vessels of approximately 100 μm with high contrasts.

Keywords: angiography, gadolinium-based contrast media, characteristic x-rays, monochromatic x-rays, tungsten Kα rays

1. INTRODUCTION

Ultrafast X-Ray Detectors, High-Speed Imaging, and Applications, edited by Stuart Kleinfelder,
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Monochromatic parallel beams from synchrotron orbital radiation have been employed in phase-contrast radiography[1,2] and enhanced K-edge angiography.[3,4] In particular, the parallel beams with photon energies of approximately 35 keV have been employed to perform iodine K-edge angiography, because the beams are absorbed effectively by iodine-based contrast media with a K-absorption edge of 33.2 keV. Without using a synchrotron, we have developed an x-ray generator utilizing a cerium-target tube,[5,6] and have performed cone-beam K-edge angiography achieved with cerium Kα rays of 34.6 keV.

Gadolinium-based contrast media with a K-edge of 50.2 keV have been employed to perform in MRA, and the gadolinium density has been increasing. In view of this situation, ytterbium Kα rays (52.0 keV) are useful for enhanced K-edge angiography, because the Kα 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. However, because ytterbium is a lanthanide series element and has a high reactivity, Kα rays of tantalum (57.1 keV) and tungsten (58.9 keV) are also useful to perform angiography.

In order to perform high-speed biomedical radiography, several different flash x-ray generators[7-10] with photon energies less than 150 keV have been developed, and plasma flash x-ray generators[11-13] have been developed to perform a preliminary experiment for producing hard x-ray lasers. From weakly ionized plasma, clean K-series characteristic x-rays of nickel and copper and their higher harmonic hard x-rays have been produced. Furthermore, high-photon-energy monochromatic flash x-ray generators[14-17] have been developed to produce K-series characteristic x-rays of molybdenum, cerium, ytterbium, tantalum, and tungsten, because bremsstrahlung rays are not emitted in the opposite direction to that of electron trajectory in Sommerfeld's theory.

In this article, we describe an intense monochromatic plasma flash x-ray generator with a tungsten target tube, and used it to perform a preliminary study on angiography achieved with tungsten Kα rays using an ytterbium oxide filter.

2. PRINCIPLE OF K-EDGE ANGIOGRAPHY

Figure 1 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.

3. GENERATOR

3.1 High-voltage circuit

Figure 2 shows a block diagram of a high-intensity plasma flash x-ray generator. The generator consists of the following essential components: a high-voltage power supply, a high-voltage condenser with a capacity of approximately 150 nF, an air gap switch, a turbomolecular pump, a thyratron pulse generator as a trigger device, and a flash x-ray tube. In this generator, a coaxial cable transmission line is employed in order to increase maximum tube voltage using high-voltage reflection. The high-voltage main condenser is charged up to 80 kV by the power supply, and electric charges in the condenser are discharged to the tube through the four cables after closing the gap switch with the trigger device.

3.2 X-ray tube

The x-ray tube is a demountable cold-cathode diode 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 ring-shaped graphite cathode with an inside diameter of 4.5 mm, a stainless-steel vacuum chamber, a nylon insulator, a polyethylene terephthalate (Mylar) x-ray window 0.25 mm in thickness, and a rod-shaped tungsten target 3.0 mm in diameter. The distance between the target and cathode electrodes can be regulated from the outside of the tube, and is set to 1.5 mm. As electron beams from the cathode electrode are roughly converged to the target by the electric field in the tube, evaporation leads to the formation of weakly ionized plasma, consisting of tungsten ions and electrons, around the target. Because bremsstrahlung rays are not emitted in the opposite direction to that of electron acceleration (Fig. 4), tungsten Kα rays can be produced using an ytterbium oxide filter with a surface density of 20 mg/cm^2.

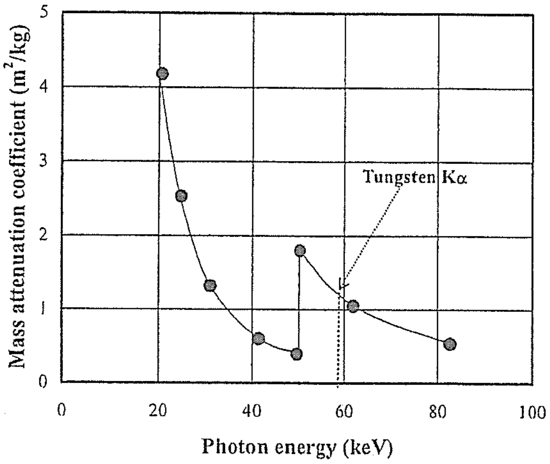

Figure 1: Mass attenuation coefficient of gadolinium and the average photon energy of tungsten Kα lines is shown above gadolinium K edge.

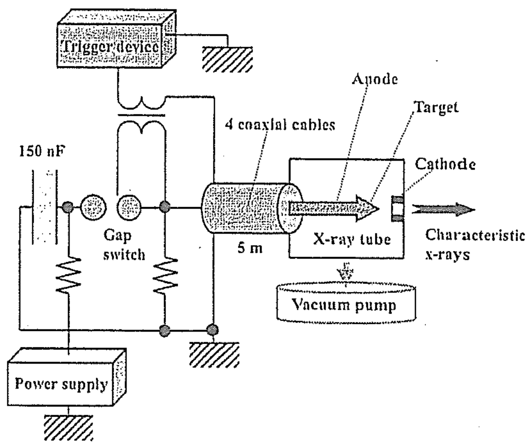

Figure 2: Block diagram including high-voltage circuit of the intense monochromatic plasma flash x-ray generator with a tungsten-target tube.

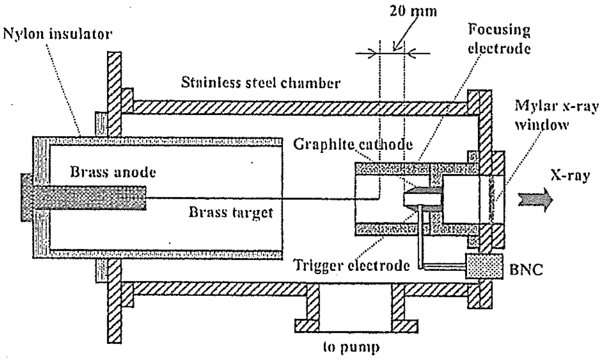

Figure 3: Schematic drawing of the flash x-ray tube with a rod-shaped tungsten target.

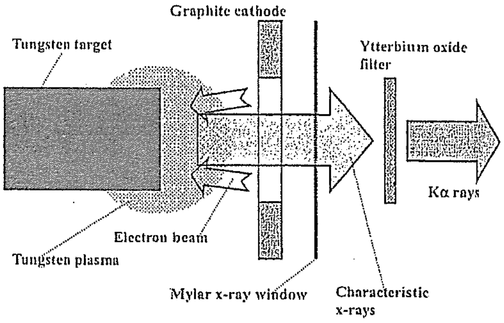

Figure 4: Irradiation of K-series characteristic x-rays of tungsten.

4. CHARACTERISTICS

4.1 Tube voltage and current

In this generator, it was difficult to measure the tube voltage and current since the tube voltages were high, and there was no space to set a current transformer for measuring the tube current. Currently, the voltage and current roughly display damped oscillations. When the charging voltage was increased, both the maximum tube voltage and current increased. At a charging voltage of 80 kV, the estimated maximum values of the tube voltage and current were approximately 160 kV (2 times the charging voltage) and 40 kA, respectively.

4.2 X-ray output

X-ray output pulse was detected using a combination of a plastic scintillator and a photomultiplier (Fig. 5). The x-ray pulse height substantially increased with corresponding increases in the charging voltage. The x-ray pulse widths were approximately 60 ns, and the time-integrated x-ray intensity measured by a thermoluminescence dosimeter (Kyokko TLD Reader 1500 having MSO-S elements without energy compensation) had a value of approximately 50 μGy at 1.0