The x-ray tube is a demountable diode type, as illustrated in Fig. 6. This tube is connected to the turbomolecular pump with a pressure of about 1 mPa and consists of the following major devices: a rod-shaped molybdenum 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. Because bremsstrahlung rays are not emitted in the opposite direction to that of electron trajectory, molybdenum Kα rays can be produced using a 20-μm-thick zirconium K-edge filter.

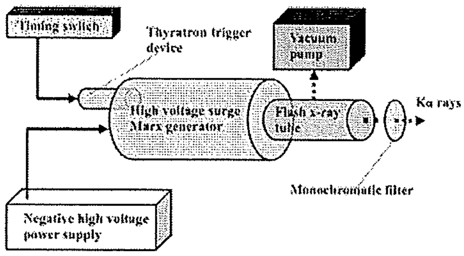

Fig. 5. Block diagram of the compact monochromatic flash x-ray generator.

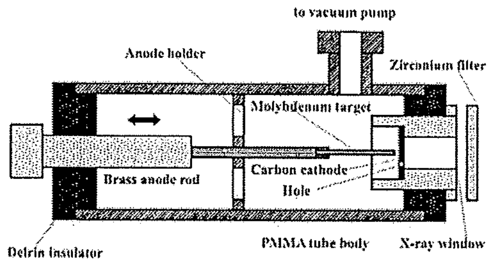

Fig. 6. Structure of the monochromatic flash x-ray tube with a PMMA tube body.

3.2 Characteristics

At a charging voltage of −70 kV, the maximum tube voltage and current were 120 kV and 1.0 kA, respectively. The x-ray pulse widths were approximately 70 ns, and the Kα intensity was approximately 70 μGy per pulse at 0.5 m from the source of 3.0 mm in diameter. In the spectrum measurement, clean molybdenum Kα lines were left using the zirconium filter, and the K-ray intensity substantially increased with increasing the charging voltage (Fig. 7).

3.3 High-speed radiography

The monochromatic flash radiography was performed by the CR system at 0.5 m from the x-ray source with the filter, and the charging voltage was −70 kV. The radiogram of water falling into a polypropylene beaker from a glass test tube is shown in Fig. 8. This image was taken with the slight addition of an iodine-based contrast medium. Because the x-ray duration was about 100 ns, the stop-motion image of water could be obtained.

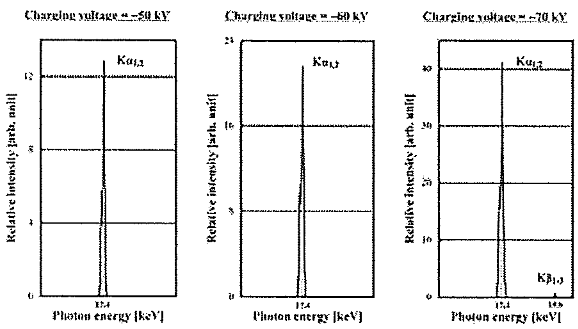

Fig. 7. X-ray spectra from the monochromatic flash x-ray tube with a molybdenum target.

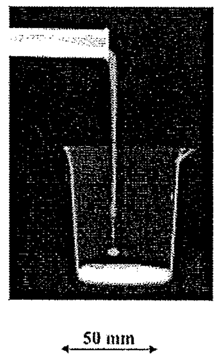

Fig. 8. Radiogram of water falling into a polypropylene beaker from a glass test tube.

4. SUPER-FLUORESCENT PLASMA X-RAY GENERATOR

4.1 Generator

Figure 9 shows a block diagram of a high-intensity super-fluorescent 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.

The x-ray tube is a demountable cold-cathode diode that is connected to the turbomolecular pump with a pressure of approximately 1 mPa. 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, at the target tip. Because bremsstrahlung rays are not emitted in the opposite direction to that of electron trajectory, tungsten Kα lines are left by absorbing Kβ lines using an ytterbium oxide filter.

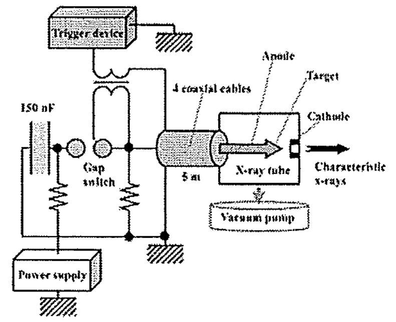

Fig. 9. Block diagram of the high-intensity super-fluorescent plasma flash x-ray generator

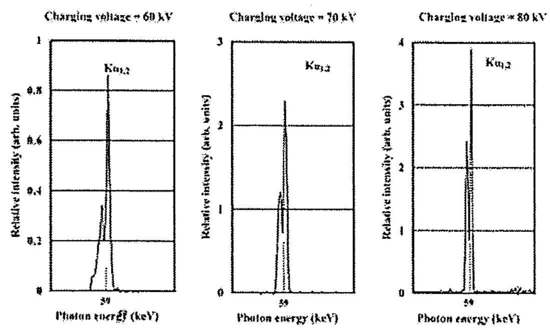

Fig. 10. X-ray spectra from the super-fluorescent plasma x-ray tube with a tungsten target.

4.2 Characteristics

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 are approximately 160 kV and 40 kA, respectively. The x-ray pulse widths were approximately 110 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 using the filter. When the charging voltage was increased using the filter, the characteristic x-ray intensities of tungsten Kα lines increased. The Kα lines were clean, and hardly any Kβ lines and bremsstrahlung rays were detected (Fig. 10).

4.3 High-speed gadolinium K-edge angiography

Figure 11 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 flash angiography was performed by the CR system at 1.2 m from the x-ray source, and the charging voltage was 70 kV. Figure 12 shows angiogram of a rabbit head using gadolinium oxide powder, and fine blood vessels of approximately 100 μm were visible.

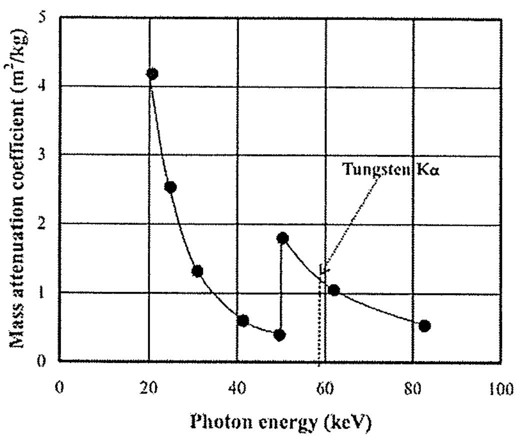

Fig. 11. Mass attenuation coefficients of gadolinium and the average photon energy of the tungsten Kα lines.

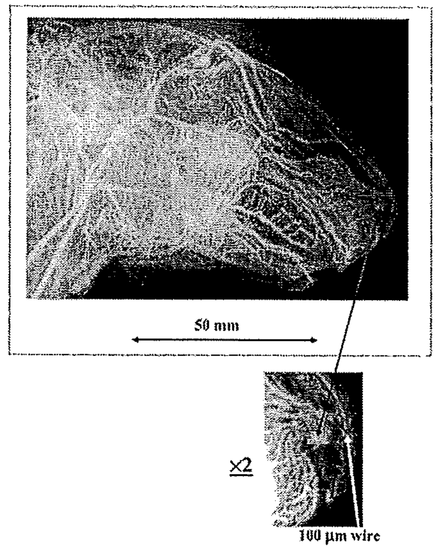

Fig. 12. Angiogram of a rabbit head using gadolinium oxide powder.

5. CERIUM X-RAY GENERATOR

5.1 Generator

The main circuit for producing x-rays is illustrated in Fig. 13, 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.

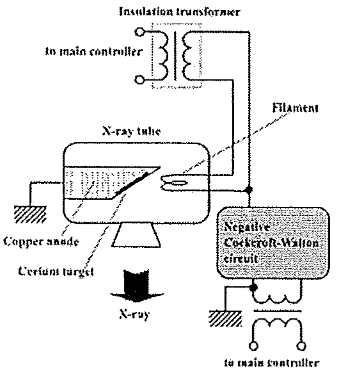

Fig. 13. Main high-voltage transmission line of the cerium x-ray generator.

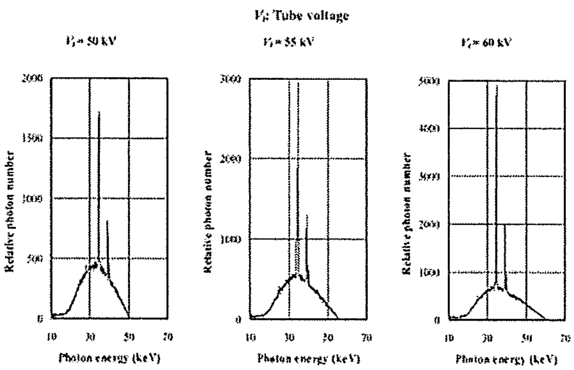

Fig. 14. X-ray spectra using a 3.0-mm-thick aluminum filter with changing the tube voltage.

5.2 Characteristics

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 (Fig. 14), and the x-ray intensity was 19.9 μGy/s at 1.0 m from the source with a tube voltage of 60 kV and a current of 0.40 mA.

5.3 High-speed radiography

Real-time cohesion radiography was performed using an image intensifier (I I) and a high-sensitive CCD camera (MLX) made by NAC Image Technology at a frame speed of 30 Hz and an image capture time (shutter speed) of 1 ms (Fig. 15). Radiograms from the I I are taken by the CCD camera, and digital video files are recorded by a personal computer through a video capture box.

Figure 16 shows the mass attenuation coefficients of iodine at the selected energies; the coefficient curve is discontinuous at the iodine K-edge. The average photon energy of the cerium Kα lines is shown just above the iodine K-edge. The average photon energy of Kα lines is 34.6 keV, and iodine contrast media with a K-absorption edge of 33.2 keV absorb the lines easily. Therefore, blood vessels were observed with high contrasts.

Figure 17 shows two frames (angiograms) of water falling into polypropylene beaker from a plastic test tube. These images were taken with a tube voltage of 60 kV, and an iodine-based contrast medium was added a little. Because the capture time was about 1 ms, the stop-motion images of water were obtained. Therefore, blood vessels can be seen with high contrasts.

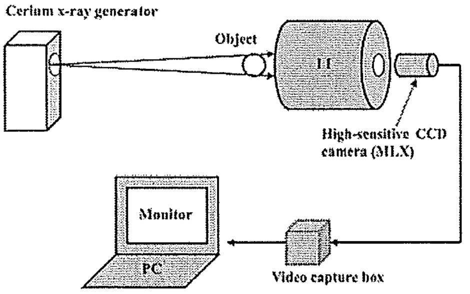

Fig. 15. Experimental setup for performing real-timeradiography with a short capture time utilizing an image intensifier and the MLX camera.

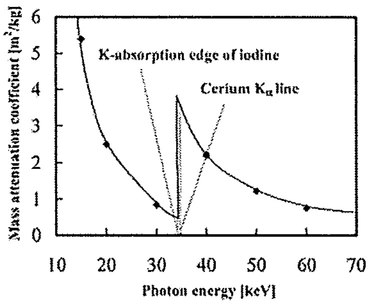

Fig. 16. Mass attenuation coefficients of iodine at the selected energies and the average photon energy of the cerium Kα lines.

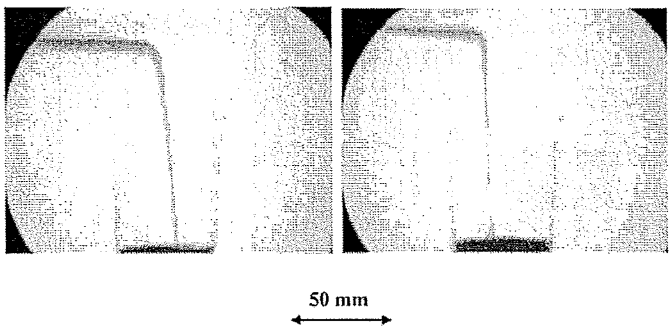

Fig. 17. Two angiograms (frames) of water falling into a polypropylene beaker from a plastic test tube using an iodine medium.

6. MICROFOCUS X-RAY GENERATOR

6.1 Generator and experimental setup

The microfocus x-ray generator consists of a main controller, an x-ray tube unit with a Cockcroft-Walton circuit, an insulation transformer, and a 100-μm-focus x-ray tube. The tube voltage, the current, and the exposure time can be controlled by the controller. The main circuit for producing x-rays employs the Cockcroft-Walton circuit in order to decrease the dimensions of the tube unit (Fig. 18). In the x-ray tube, the positive and negative high voltages are applied to the anode and cathode electrodes, respectively. 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 70 kV, and the tube current was regulated to within 0.50 mA (maximum current) by the filament temperature. The exposure time is controlled in order to obtain optimum x-ray intensity, and narrow-photon-energy bremsstrahlung x-rays are produced using a 3.0-mm-thick aluminum filter for absorbing soft x-rays.

6.2 Characteristics

At a constant tube current of 0.50 mA, the x-ray intensity increased when the tube voltage was increased. At a tube voltage of 60 kV, the intensity with the filter was 26.0 μGy/s. Subsequently, in order to measure x-ray spectra, we employed a cadmium telluride detector (XR-100T, Amptek). When the tube voltage was increased, the bremsstrahlung x-ray intensity increased, and both the maximum photon energy and the spectrum peak energy increased.

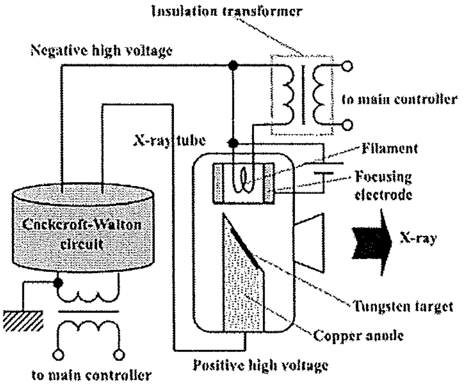

Fig. 18. Circuit diagram of the microfocus x-ray generator.

6.3 High-speed radiography

Real-time magnification radiography was performed by twofold magnification imaging using the I I, the camera, and the filter. The tube voltage, the capture time, and the distance between the x-ray source and the I I were 60 kV, 1 ms, and 1.0 m, respectively. Figure 19 shows radiograms of a metronome, and stop-motion images of a pendulum were observed.

6.4 High-speed enhanced magnification angiography

Figure 20 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 media with a K-absorption edge of 33.2 keV absorb the rays easily, blood vessels were observed with high contrasts.

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. Angiograms of a dog heart on a turn table is shown in Fig. 21, and the fine coronary arteries are not visible due to the spatial resolution of I I. In addition, neo-capillaries in a rabbit cancer are clearly visible using a high-resolution I I (Fig. 22).

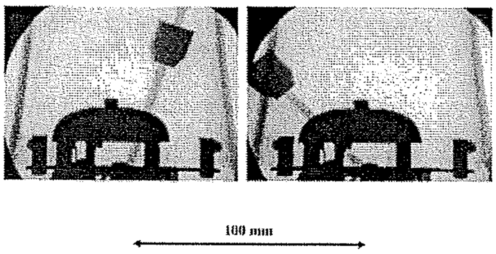

Fig. 19. Radiograms of a metronome.

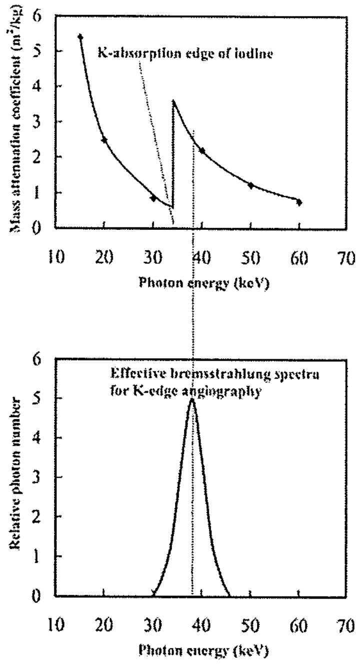

Fig. 20. Mass attenuation coefficients of iodine and effective bremsstrahlung x-rays for enhanced K-edge angiography.

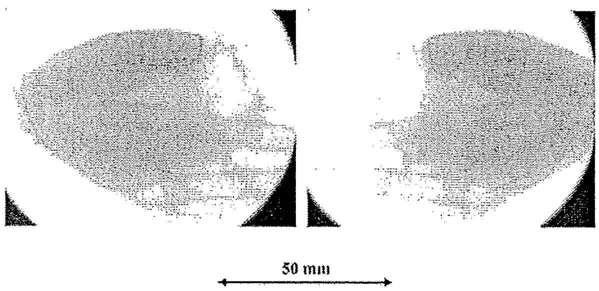

Fig. 21. Angiogram of an extracted dog heart using iodine microspheres on a turn table.

Fig. 22. Angiograms of a rabbit cancer using a high-resolution I I.

7. CONCLUSION AND OUTLOOK

We have developed various x-ray generators corresponding to specific radiographic objectives, and the x-ray duration ranges from approximately 100 ns to continuous exposure. In the weakly ionized plasma formation, extremely clean and intense K lines were produced, since the bremsstrahlung x-rays were absorbed effectively by the plasma. In particular, the harmonic bremsstrahlung rays survived and waved due to the x-ray resonation in the plasma.

Although most flash x-ray generators utilize Marx surge generators and produce hard bremsstrahlung x-rays, monochromatic flash x-ray generators have been employed to observe aluminum grains in studies on space debris on the earth. In addition, because the monochromatic tubes realize uniform monochromatic x-ray intensity distributions, the absorber thickness can be calculated easily.

To perform enhanced K-edge angiography using iodine-based contrast media, the cerium, samarium and gadolinium targets are very useful because K lines from these targets are absorbed effectively by iodine media. Therefore, using the x-ray I I in conjunction with the MLX camera with short capture times of approximately 1 ms, stop-motion images of fine blood vessels can be almost seen. Using this image intensifying system, the image quality slightly fell with decreases in the capture time. However, stop-motion image can be obtained when the capture time is decreased.

We employed an x-ray generator with a 100-μm-focus tungsten tube and performed real-time twofold magnification radiography (fluoroscopy) using the I I and the MLX camera. To perform angiography, we employed narrow-photon-energy bremsstrahlung x-rays with a peak photon energy of approximately 35 keV, which can be absorbed easily by iodine-based contrast media. Although we obtained mostly absorption-contrast images, the phase-contrast effect may be added in cases where low-density media are employed.

Because the focus diameter of the tube has been decreased to 10 μm using a rotating anode microfocus tube developed by Hitachi Medical Corporation, a high-resolution and high-speed magnification radiography system will become possible.

ACKNOWLEDGMENTS

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High-sensitive radiography system utilizing a pulse x-ray generator and a night-vision CCD camera (MLX)

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ABSTRACT

High-sensitive radiography system utilizing a kilohertz-range stroboscopic x-ray generator and a night-vision CCD camera (MLX) is described. The x-ray generator consists of the following major components: a main controller, a condenser unit with a Cockcroft-Walton circuit, and an x-ray tube unit in conjunction with a grid controller. The main condenser of about 500 nF in the unit is charged up to 100 kV by the circuit, and the electric charges in the condenser are discharged to the triode by the grid control circuit. The maximum tube current and the repetition rate are approximately 0.5 A and 50 kHz, respectively. The x-ray pulse width ranges from 0.01 to 1.0 ms, and the maximum shot number has a value of 32. At a charging voltage of 60 kV and a width of 1.0 ms, the x-ray intensity obtained without filtering was 6.04 μGy at 1.0 m per pulse. In radiography, an object is exposed by the pulse x-ray generator, and a radiogram is taken by an image intensifier. The image is intensified by the CCD camera, and a stop-motion image is stored by a flash memory device using a trigger delay device. The image quality was improved with increases in the x-ray duration, and a single-shot radiography was performed with durations of less than 1.0 ms.

Keywords: high-sensitive radiography, image intensification, high-sensitive CCD camera, pulse x-ray generator

1. INTRODUCTION

With advances in high-voltage pulse technology, high-photon-energy flash x-ray generators[1,2] have been developed utilizing multi-stage Marx generators, and the maximum photon energy has been increased up to approximately 1 MeV for military applications. In contrast, we have developed low-photon-energy flash x-ray generators[3-6] with photon energies of lower than 150 keV, and have performed high-speed soft radiographies including biomedical applications.

To produce extremely clean K-series characteristic x-rays such as lasers, we have developed three characteristic flash x-ray generators[7-15] and have succeeded in producing clean K lines. In particular, bremsstrahlung x-rays are absorbed effectively by weakly ionized metal plasmas. Subsequently, we have developed steady-state characteristic x-ray

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generators[16] and have succeeded in producing clean K lines utilizing angle dependence of the bremsstrahlung x-rays.
In the biomedical field, because there are no ultra-high-speed movements, a condenser-discharge stroboscopic x-ray generator[17,18] has been developed. In this generator, the x-ray duration can be controlled from 10 μs to 1.0 ms, and the maximum repetition rate is approximately 50 kHz. In conjunction with a computed radiography (CR) system, short-duration and multi-shot radiographies are possible. In addition, the velocity of a high-speed object can be calculated easily by measuring the length of blurring because the x-ray duration can be controlled correctly within 1.0 ms.
Recently, because an extremely high-sensitive color CCD camera (MLX) has been developed by NAC image technology, we are very interested in intensifying the x-ray image signals using the MLX camera in conjunction with a pulse x-ray generator. Using image intensifying, the absorbed dose can be reduced from patients.
In this research, we employed a stroboscopic x-ray generator and performed a preliminary study on the intensification of image signal, utilizing an image intensifier and the high-sensitive CCD camera.

2. PULSE X-RAY GENERATOR

Figure 1 shows the block diagram of a 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. The main condenser of about 500 nF in the unit is charged up to 100 kV by the circuit, and the electric charges in the condenser are discharged to the triode by the grid control circuit. Although the tube voltage decreased during the discharging for generating x-rays, the maximum value was equal to the initial charging voltage of the main condenser.
The x-ray tube is a glass-enclosed hot-cathode triode and is composed of the following major parts: an anode rod made of copper, a tungsten plate target, an iron focusing electrode, a tungsten hot cathode (filament), a tungsten grid, and a glass tube body. The electron beams from the cathode are accelerated between the anode and cathode electrodes and are converged to the target by the focusing electrode. The tube is set in the metal case filled with insulation oil, and the diaphragm regulates the irradiation field.

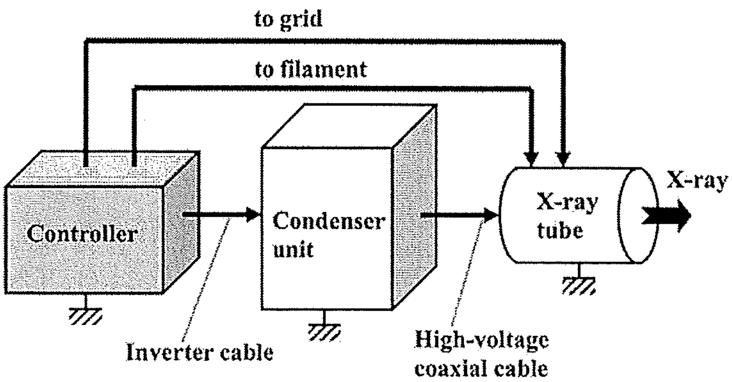

Fig. 1. Block diagram of the kilohertz-range stroboscopic x-ray generator.

3. CHARACTERISTICS

3.1 X-ray output

The x-ray output was detected by a pin diode, and the output voltages from the diode were measured by a digital storage scope (Fig. 2). When the charging voltage was increased, the pulse height increased substantially. Using this generator, the pulse width can 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 3 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 x-ray duration. At a constant pulse width of 1.0 ms, the intensity increased with increasing the charging voltage. At a charging voltage of 60 kV and a width of 1.0 ms, the x-ray intensity was 6.04 μGy per pulse at 1.0 m from the source.

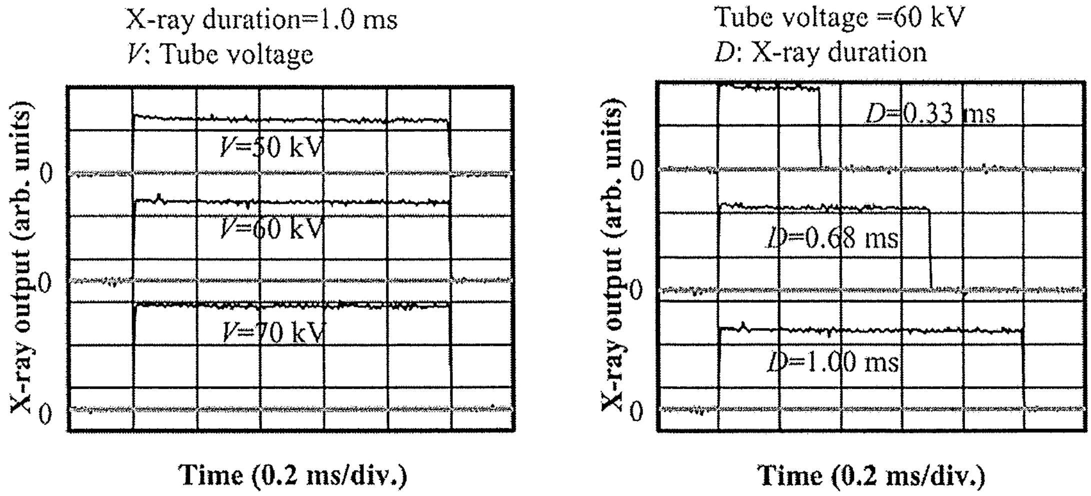

Fig. 2. X-ray outputs at indicated conditions.

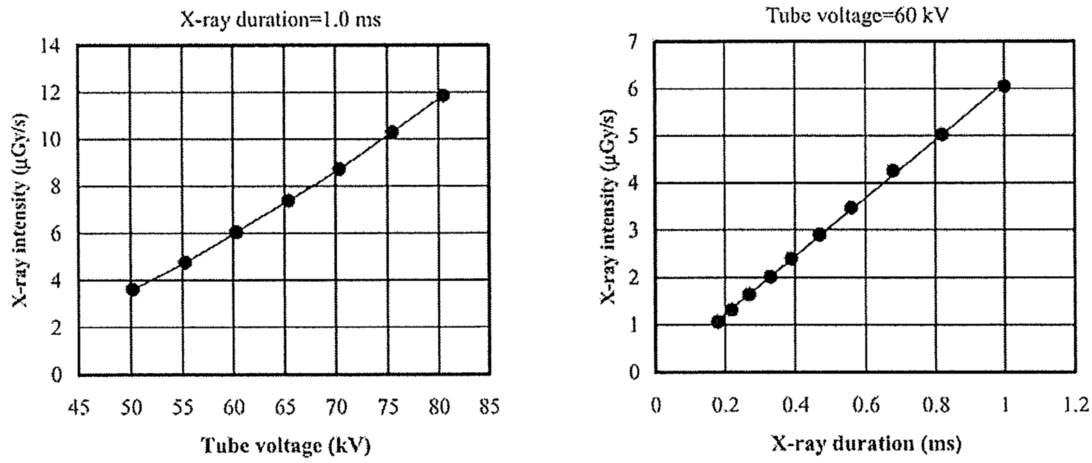

Fig. 3. X-ray intensities at 1.0 m from the x-ray source.

4. RADIOGRAPHY

Figure 4 shows the experimental setup for intensifying x-ray image signals using the MLX camera. An object is exposed by the pulse x-ray generator, and a radiogram is taken by an image intensifier. Then, the image is amplified by the CCD camera, and a stop-motion image is stored by a flash memory device using a trigger delay device with a delay time of 50 ms. The image quality improved with increases in the x-ray duration, and single-shot radiography was performed with durations of less than 1.0 ms.

First, rough measurements of spatial resolution were made using wires. Figure 5 shows a radiogram of a 200-μm-diameter tungsten wire coiled around a pipe made of polymethyl methacrylate. In this radiography, the wire was observed with blurring, and image quality improved with increases in the x-ray duration. Next, two radiograms of a metronome are shown in Fig. 6, and stop-motion images of a pendulum are visible. In radiography of plastic bullets, spherical bullets were clearly observed (Fig. 7). Finally, the image of water falling into a polypropylene beaker from a plastic test tube is shown in Fig. 8. This image was taken with the slight addition of an iodine-based contrast medium. Because the x-ray duration was 1 ms, the stop-motion image of water could be obtained.

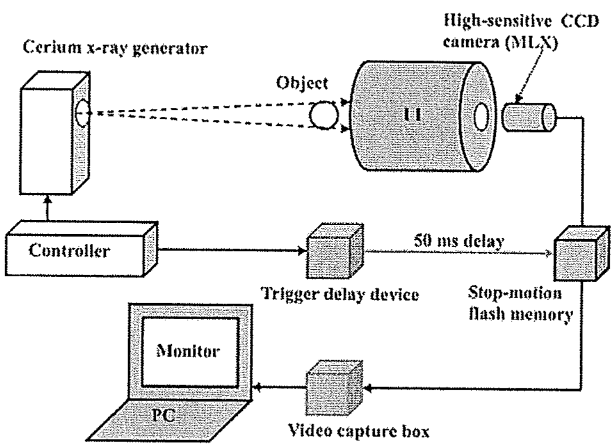

Fig. 4. Experimental setup for performing real-time radiography utilizing the MLX camera.

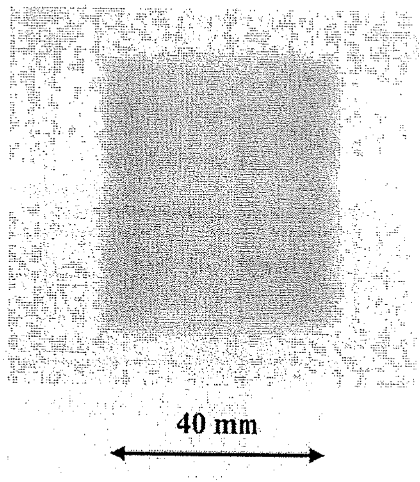

Fig. 5. Radiogram of a 200-μm-diameter tungsten wire coiled around a pipe made of polymethyl methacrylate.

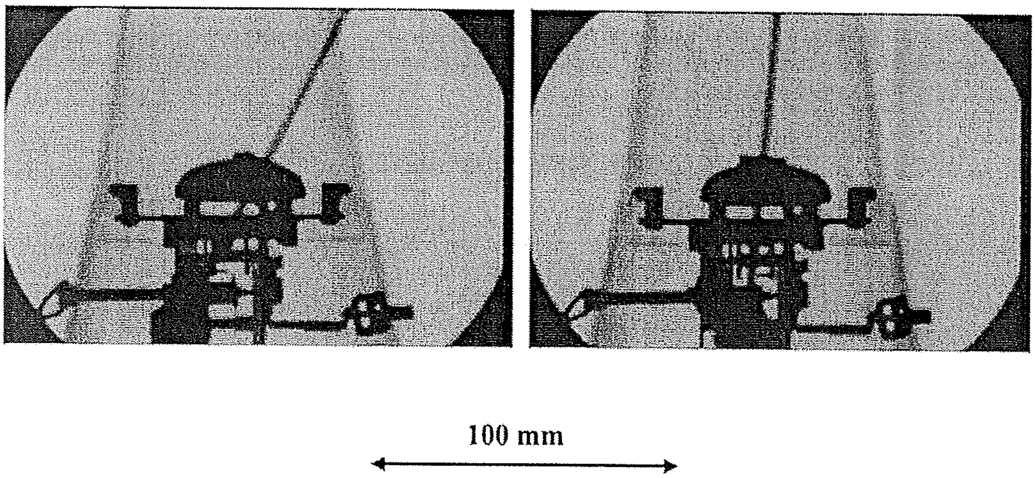

Fig. 6. Two radiograms of a metronome.

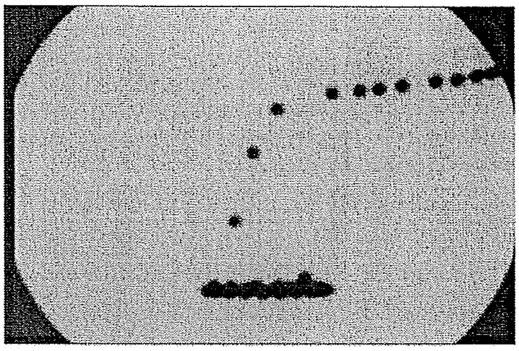

Fig. 7. Radiogram of plastic bullets falling into a polypropylene beaker from a plastic test tube.

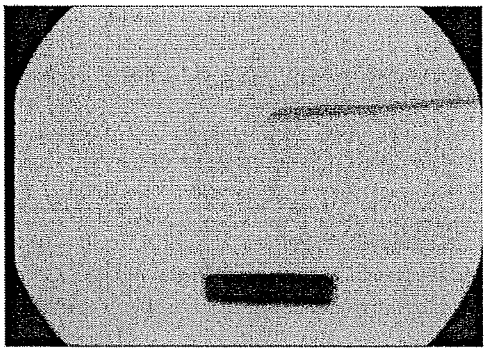

Fig. 8. Radiogram of water falling into a polypropylene beaker from a plastic test tube using an iodine medium.

5. DISCUSSION

We performed a fundamental study on the intensification of the x-ray image signal using the MLX camera, and the image quality improved with increases in the x-ray duration. The spatial resolution was primarily determined by the resolution of I I and the pixel number of the camera. Therefore, the resolution improves with improving the I I resolution and with increasing the pixel number.

Without considering the absorbed dose, short-duration real-time radiography is possible by decreasing the image capture time of the camera using a steady-state x-ray generator. In cases where a microfocus x-ray generator is employed, the spatial resolution improves using magnification radiography, and the real-time radiography with a capture time of 1 ms can be performed by image intensifying.

In this experiment, although we performed only single-shot radiography, high-speed dynamic radiography could be possible using a high-speed high-sensitive video camera synchronizing to the repetitive x-ray output, and the repetition rate can be increased to approximately 50 kHz.

ACKNOWLEDGMENTS

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K-edge magnification digital angiography using a 100-μm-focus tungsten tube

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Abstract. A microfocus x-ray tube is useful to perform magnification digital radiography, including phase-contrast effects. 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 are 105 kV, 0.5 mA, and 50 W, respectively. Using a 3-mm-thick aluminum filter, the x-ray intensity is 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 is approximately 35 keV using the filter with a tube voltage of 60 kV, the bremsstrahlung x-rays are absorbed effectively by iodine-based contrast media with an iodine K-edge of 33.2 keV. Magnification angiography is performed by three-fold magnification imaging with a computed radiography system using iodine-based microspheres 15 μm in diameter. In angiography of nonliving animals, we observe fine blood vessels approximately 100 μm with high contrasts.

[DOI: 10.1117/1.2542285]

Subject terms: high-contrast angiography; magnification digital radiography; microfocus x-ray tube; energy-selective imaging; phase-contrast effect.

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1 Introduction

Conventional flash x-ray generators utilizing condensers in conjunction with cold-cathode tubes are useful to perform high-speed radiography, including biomedical applications, and several different generators have been developed.[1–7] In particular, linear-plasma x-ray generators[8–10] utilizing triodes have been employed to produce clean K-series characteristic x-rays of nickel and copper, and we have confirmed the irradiation of higher harmonic hard x-rays of K-series characteristic x-rays. Without forming plasmas, a flash x-ray diode with a disk cathode can be employed to perform a fundamental study on producing characteristic x-rays,[11,12] and we have succeeded in producing clean K-series lines using the 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 radiography, including the phase-contrast effect.

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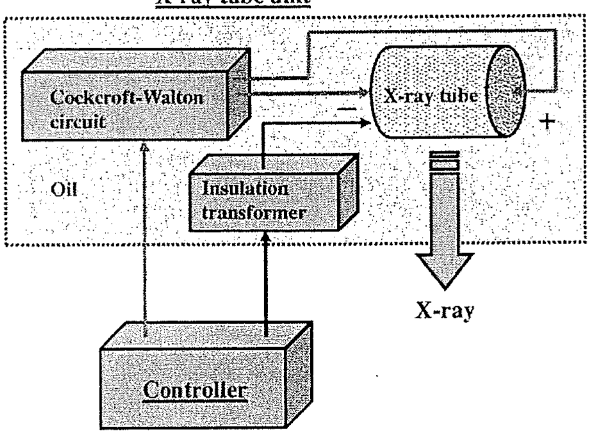

Fig. 1 Block diagram of the x-ray generator.

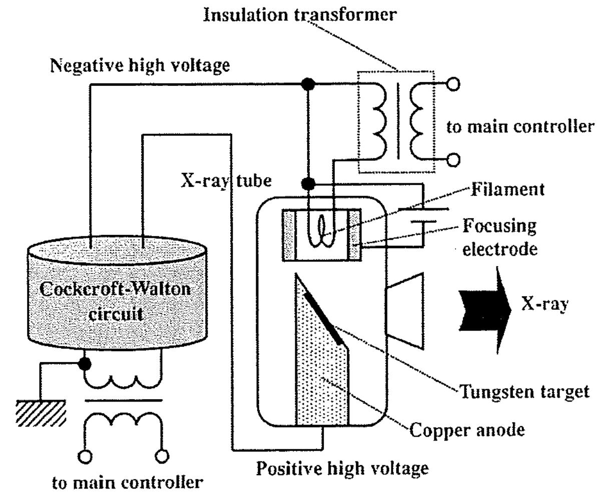

Fig. 2 Electric circuit of the x-ray generator.

Synchrotrons are capable of producing high-dose-rate monochromatic parallel x-ray beams using a monochrocollimator, and the beams have been applied to phase-contrast radiography[13,14] and enhanced K-edge angiography.[15,16] In angiography, monochromatic x-rays with photon energies approximately 35 keV have been employed, because the rays are absorbed effectively by iodine-based contrast media with an iodine K edge of 33.2 keV.

Without using synchrotrons, phase-contrast radiography for edge enhancement can be performed using a microfocus x-ray tube. Magnification radiography, including the phase-contrast effect, has been applied in mammography achieved with a computed radiography (CR) system[17] using a 100-μm-focus molybdenum tube.[18] Subsequently, we have developed a cerium x-ray generator[19,20] to perform enhanced K-edge angiography using cone beams, and have succeeded in observing fine blood vessels and coronary arteries with high contrasts using cerium $K\alpha$ rays of 34.6 keV. However, it is difficult to design a small focus cerium tube for angiography.

Magnification radiography is useful to improve the spatial resolution in digital radiography, and the phase contrast may come into effect in edge enhancement of comparatively large objects, including thick blood vessels filled with low-density contrast media. Therefore, narrow-photon-energy bremsstrahlung x-rays with a peak energy of approximately 35 keV from a microfocus tungsten tube are useful to perform high-contrast high-resolution angiography.

In the present research, we employed a 100-μm-focus tungsten tube, used to perform enhanced magnification angiography by controlling bremsstrahlung x-ray spectra using an aluminum filter.

2 Experimental Setup

Figure 1 shows the block diagram of a microfocus x-ray generator used in this experiment, and the generator consists of a main controller, an x-ray tube unit with a Cockcroft-Walton circuit, an insulation transformer, and a 100-μm-focus x-ray tube. The tube voltage, current, and exposure time can be controlled by the controller. The main circuit for producing x-rays is illustrated in Fig. 2, and employs the Cockcroft-Walton circuit to decrease the dimensions of the tube unit. In the x-ray tube, positive and negative high voltages are applied to the anode and cathode electrodes, respectively. 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 70 kV, and the tube current was regulated to within 0.50 mA (maximum current) by the filament temperature. The exposure time is controlled to obtain optimum x-ray intensity, and narrow-photon-energy bremsstrahlung x-rays are produced using a 3.0-mm-thick aluminum filter for absorbing soft x-rays.

3 Results and Discussion

3.1 *X-Ray Intensity*

The x-ray intensity was measured by a Victoreen 660 ionization chamber 1.0 m from the x-ray source using the filter (Fig. 3). At a constant tube current of 0.50 mA, the x-ray

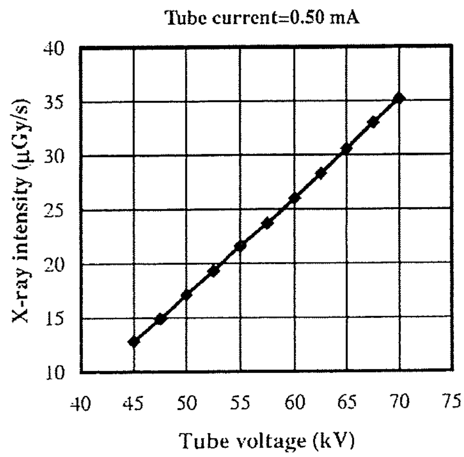

Fig. 3 X-ray intensity (μGy/s) as a function of tube voltage (kV) with a tube current of 0.50 mA.

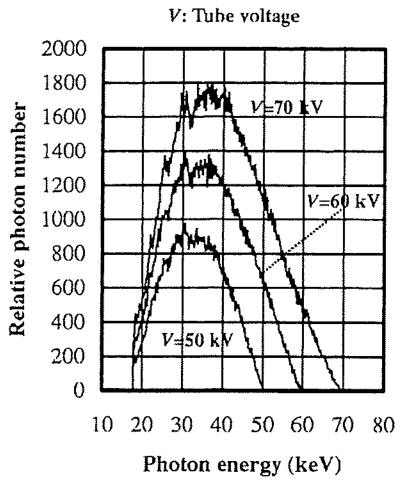

Fig. 4 Bremsstrahlung x-ray spectra measured using a cadmium telluride detector with changes in the tube voltage.

intensity increased when the tube voltage was increased. At a tube voltage of 60 kV, the intensity with the filter was 26.0 μGy/s.

3.2 *X-Ray Spectra*

To measure x-ray spectra, we employed a cadmium telluride detector (XR-100T, Amptek) (Fig. 4). When the tube voltage was increased, the bremsstrahlung x-ray intensity increased, and both the maximum photon energy and the spectrum peak energy increased.

To perform K-edge angiography, bremsstrahlung x-rays of approximately 35 keV are useful, and the high-energy bremsstrahlung x-rays decrease the image contrast. Using this filter, because bremsstrahlung x-rays with energies higher than 60 keV were not absorbed easily, the tube voltage for angiography was determined as 60 kV by considering the filtering effect of radiographic objects.

3.3 *Magnification Radiography*

Magnification radiography was performed by threefold magnification imaging using the CR system and the filter at a tube voltage of 60 kV. The distance between the x-ray source and the imaging plate was 1.5 m (Fig. 5). First, the spatial resolutions of conventional (cohesion) and magnification radiographies were made using a lead test chart. In the magnification radiography, 62.5-μm lines (eight line pairs) were visible (Fig. 6). Subsequently, Fig. 7 shows radiograms of tungsten wires coiled around rods made of

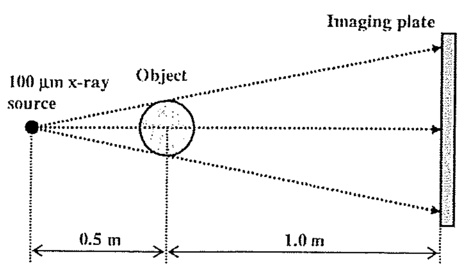

Fig. 5 Threefold magnification imaging using an imaging plate in conjunction with a microfocus tube.

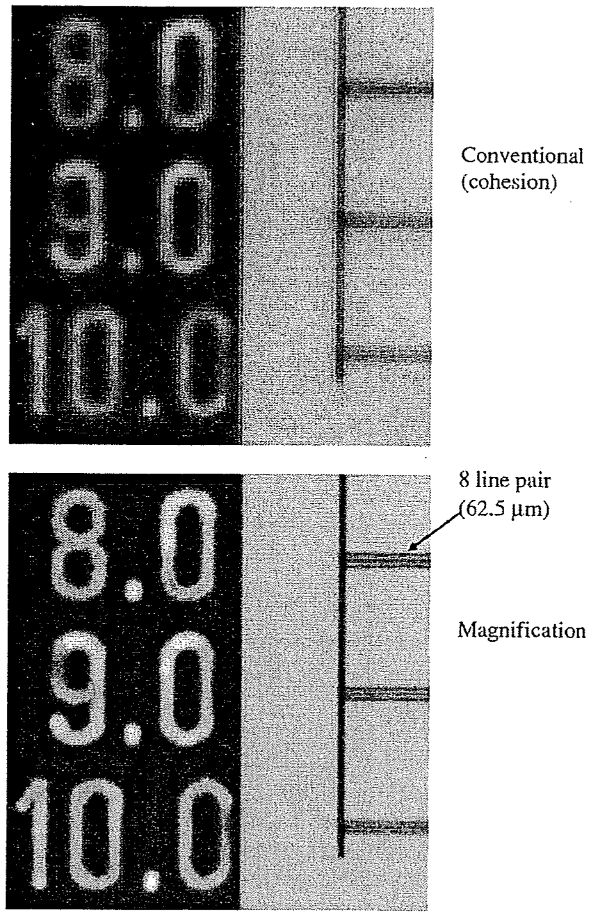

Fig. 6 Radiograms of a test chart for measuring the spatial resolution.

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-diam wire could be observed. Radiograms of one set of a bolt and a nut are shown in Fig. 8. The edge of a bubble in the bolt and the seam between the bolt and the nut are visible in magnification radiography.

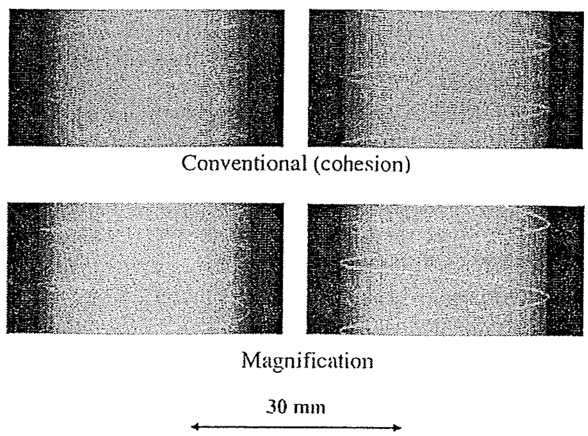

Fig. 7 Radiograms of tungsten wires coiled around PMMA rods.

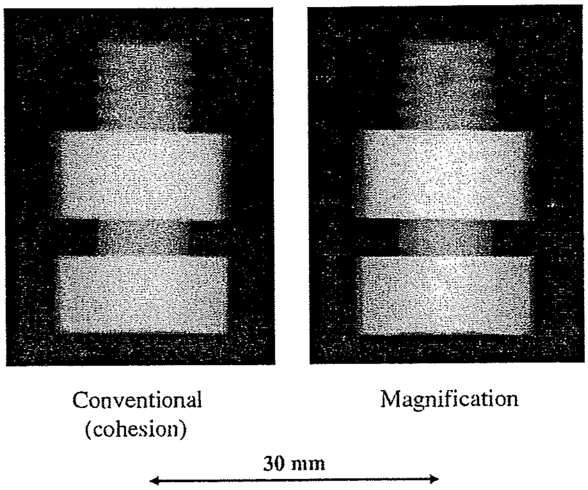

Fig. 8 Radiograms of a plastic bolt and nut.

3.4 *Enhanced Magnification Angiography*

Figure 9 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 media with a K-absorption edge of 33.2 keV absorb the rays easily, blood vessels were observed with high contrasts.

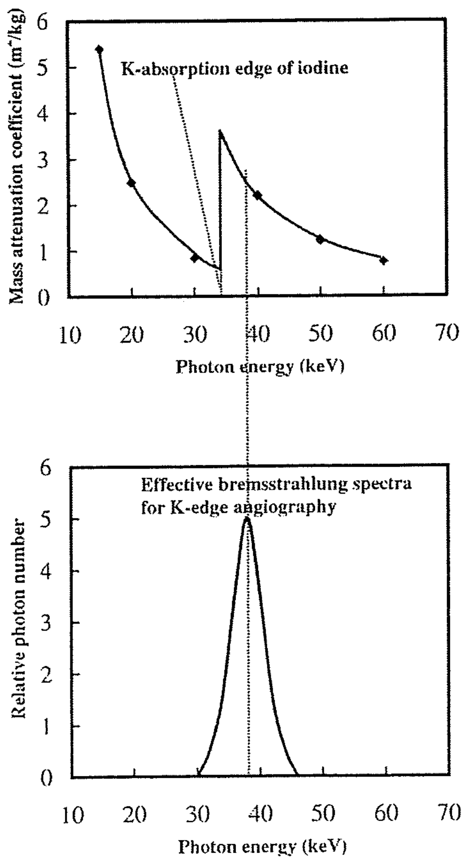

Fig. 9 Mass attenuation coefficients of iodine and effective bremsstrahlung x-rays for enhanced K-edge angiography.

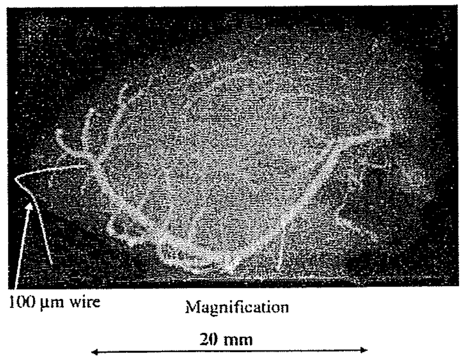

Fig. 10 Angiogram of an extracted rabbit heart using iodine microspheres.

Magnification angiography was performed at the same conditions using iodine microspheres of 15 μm in diameter. The microspheres (containing 37% iodine by weight) are very useful for making phantoms of nonliving animals used for angiography. An 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, coronary arteries of approximately 100 μm were observed using a 100-mm-thick plate.

4 Conclusion and Outlook

We employ an x-ray generator with a 100-μm-focus tungsten tube and perform enhanced K-edge magnification angiography using narrow-photon-energy bremsstrahlung x-rays with a peak photon energy of approximately 35 keV, which can be absorbed easily by iodine-based contrast me-

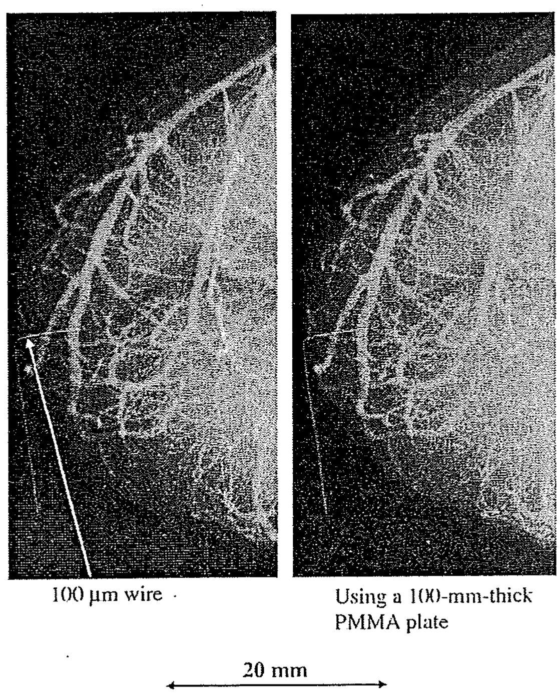

Fig. 11 Angiograms of an extracted dog heart.

dia. The bremsstrahlung x-ray intensity substantially increases with increase in the tube voltage, and the tube voltage is determined as 60 kV to increase the image contrast. In magnification angiography, although we obtain mostly absorption-contrast images, the 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 obtain spatial resolutions of approximately 50 μm using threefold magnification imaging, even when a 100-μm-focus tube is employed. 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. Based on experimental results, the maximum magnification rate without blurring is approximately threefold using a 100-μm-focus tube, and the rate increases with decreasing the focus diameter. In addition, the rate should be minimized to decrease the exposed dose from patients.

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/$(\mathrm{cm}^2 \cdot \mathrm{s})$ at 1.0 m from the source, and the photon count rate can be increased easily using a rotating anode microfocus tube. Recently, the maximum electric power of the microfocus x-ray tube has been increasing, and the kilowatt-range tube can be realized. Furthermore, since a 10-μm-focus rotating anode tube has been developed by Hitachi Medical Corporation, dynamic high-resolution angiography is possible using a flat panel detector with a pixel size of less than 100 μm. Finally, this high-resolution, high-contrast angiography could be very useful for observing fine blood vessels in regenerative medicine, coronary arteries, and irregular capillaries in cancers.

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