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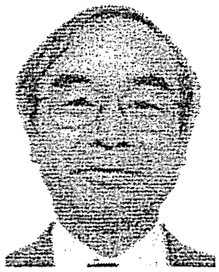

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Enhanced magnification angiography including phase-contrast effect using a 100-μm focus x-ray tube

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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

Laser-Generated, Synchrotron, and Other Laboratory X-Ray and EUV Sources, Optics, and Applications II Edited by Kyrala, Gauthier, MacDonald, Khounsary
Proc. of SPIE Vol. 5918, 591811, (2005) · 0277-786X/05/$15 · doi: 10.1117/12.620205

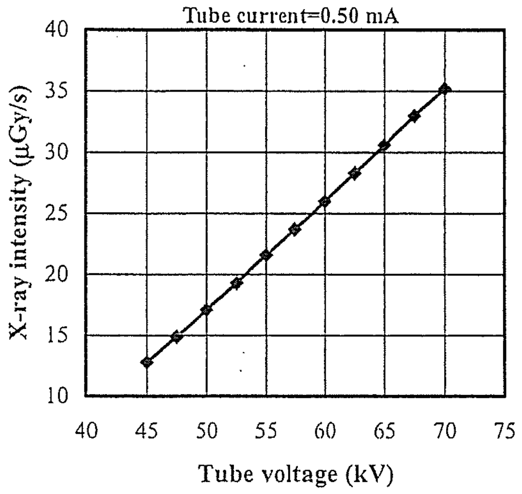


Figure 4: X-ray intensity (μGy/s) as a function of tube voltage (kV) with a tube current of 0.50 mA.

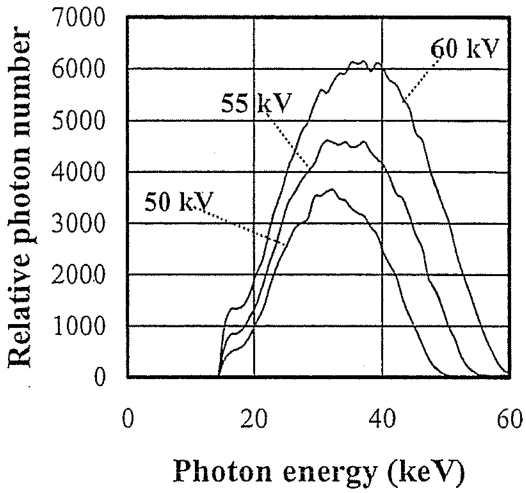


Figure 5. Bremsstrahlung x-ray spectra measured using a cadmium telluride detector with changes in the tube voltage.

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.

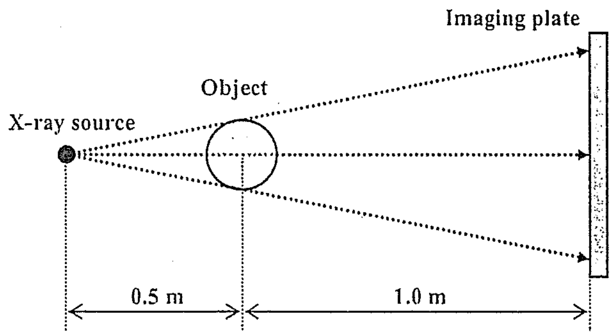


Figure 6. Three-time magnification imaging using an imaging plate in conjunction with a microfocus tube.

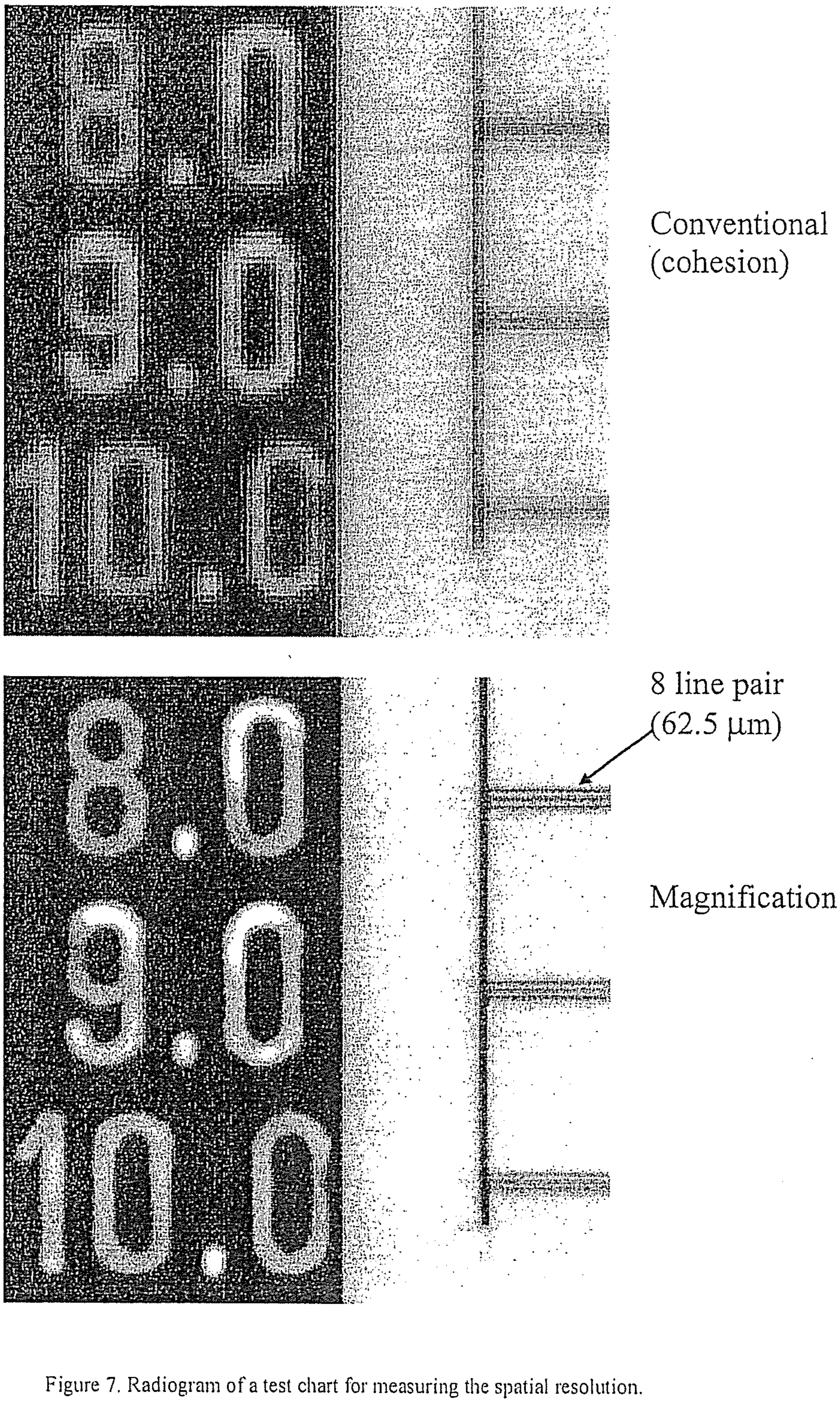


Figure 7. Radiogram of a test chart for measuring the spatial resolution.

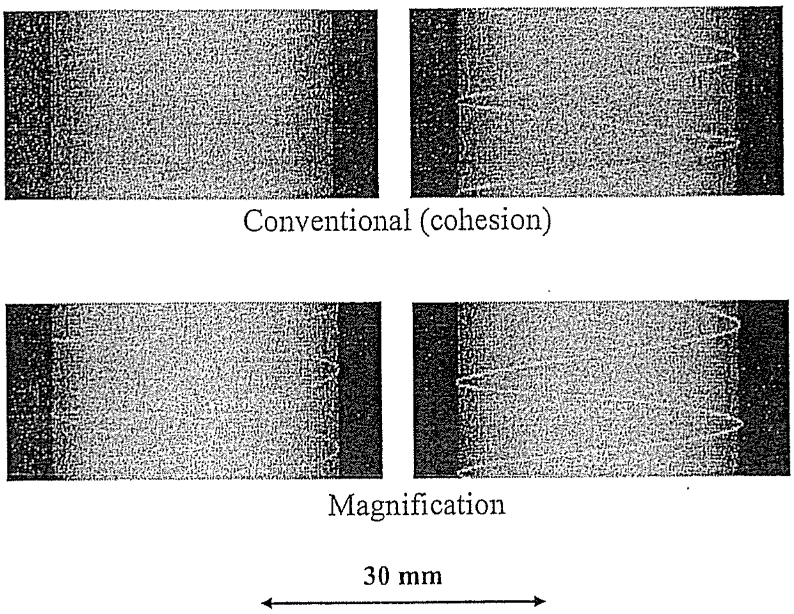


Figure 8. Radiograms of tungsten wires coiled around PMMA rods.

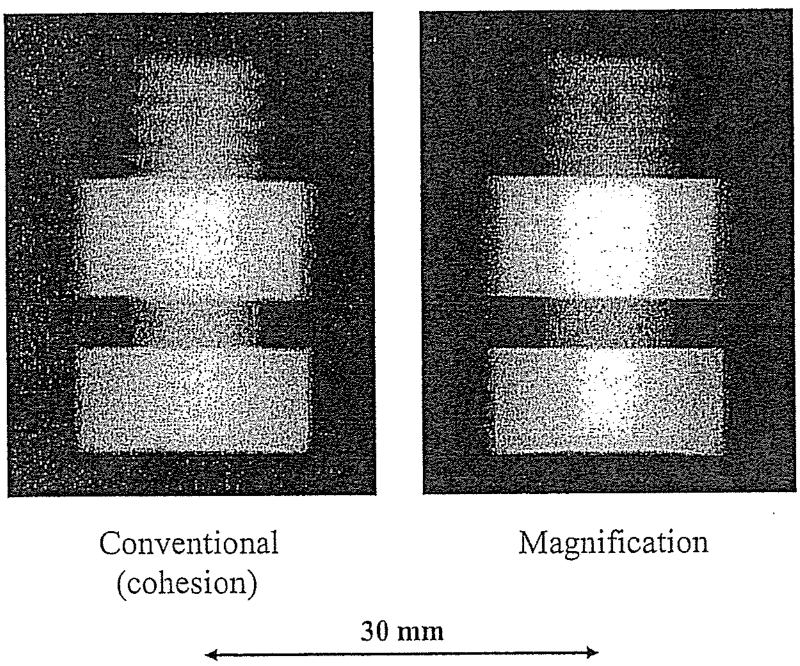
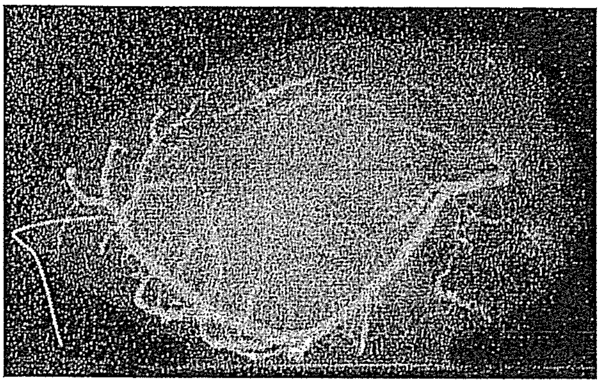


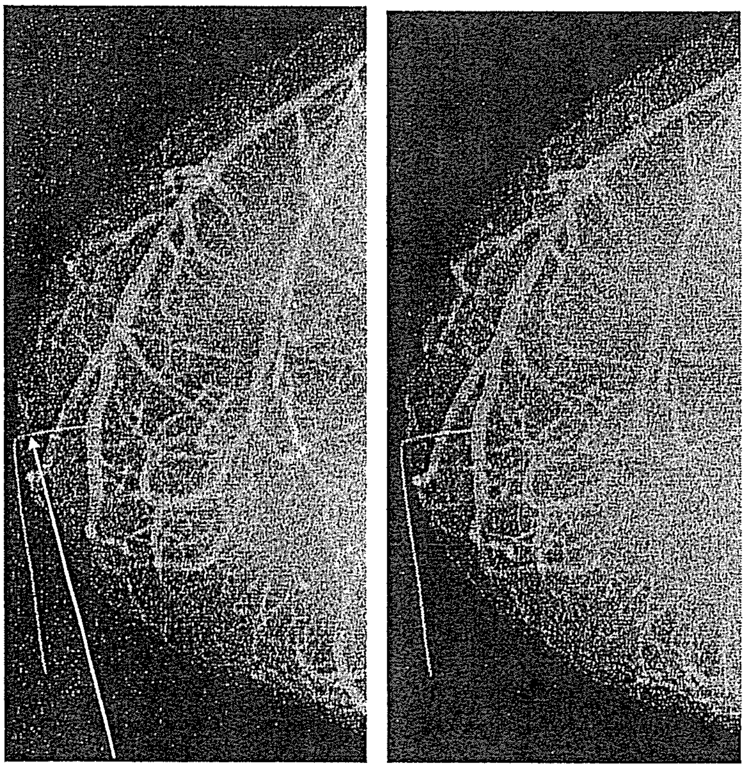
Figure 9. Radiograms of a set of a plastic bolt and a nut.



Magnification

20 mm

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



100 μm wire

Using a 100-mm-thick PMMA plate

20 mm

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

This work was supported by Grants-in-Aid for Scientific Research (13470154, 13877114, 16591181, 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 the New Energy and Industrial Technology Development Organization (NEDO, Industrial Technology Research Grant Program in '03).

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原子核研究
Vol. 49 No. 5
研究会報告

Conventional enhanced K-edge angiography utilizing cerium x-ray generator

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Abstract

The cerium-target x-ray tube is useful in order to perform cone-beam K-edge angiography because Kα rays from the cerium target are absorbed effectively by iodine-based contrast media. The maximum tube voltage and current were 65 kV and 0.40 mA, respectively, and the focal-spot sizes were approximately 1×1 mm. Sharp cerium Kα lines were left using a barium sulfate filter, and the x-ray

intensity was 16.8 μGy/s at 1.0 m from the source with a tube voltage of 60 kV and a current of 0.40 mA. Angiography was performed using iodine-based microspheres 15 μm in diameter. In angiography of non-living animals, we observed fine blood vessels of 100 μm or less.

1. Introduction

Synchrotrons generate monochromatic parallel x-ray beams using single crystals. These beams with photon energies of approximately 35 keV have been employed to perform enhanced K-edge angiography,[1-3] since the beams are absorbed effectively by iodine-based contrast media.

In order to perform high-speed medical radiography, although several different flash x-ray generators[4-9] utilizing cold-cathode tubes have been developed, plasma flash x-ray generators[10-13] are useful to produce quasi-monochromatic x rays without using a K-edge filter. Therefore, we have performed a demonstration of cone-beam K-edge angiography[14] utilizing a cerium plasma generator, since K-series characteristic x rays from the cerium target are absorbed effectively by iodine. Recently, we have developed a steady-state x-ray generator utilizing a cerium-target tube, and have demonstrated enhanced K-edge angiography utilizing a barium sulfate filter.[15] In this research, Kα lines (34.6 keV) were left by absorbing Kβ lines (39.2 keV).

In the present research, we describe a preliminary study on cone-beam K-edge angiography achieved with cerium Kα rays using a barium sulfate filter.

2. Generator

Figure 1 shows the block diagram of the x-ray generator, which consists of a main controller, a cerium-target x-ray tube unit with a Cockcroft-Walton circuit and an insulation transformer, and a personal computer. The tube voltage, the current, and the exposure time can be controlled by both the controller and the computer. The main circuit for producing x rays is illustrated in Fig. 2, and employs 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. Monochromatic Kα lines were left using a 5-mm-thick barium sulfate filter in which barium sulfate powder was mixed with polymethyl methacrylate (PMMA) resin, since both the bremsstrahlung and the Kβ rays were absorbed effectively by the filter. In designing the filter, the surface density of the barium sulfate powder is important, since the x rays are absorbed effectively by the powder as compared with the PMMA resin. In this case, the density was approximately 10 mg/cm^2.

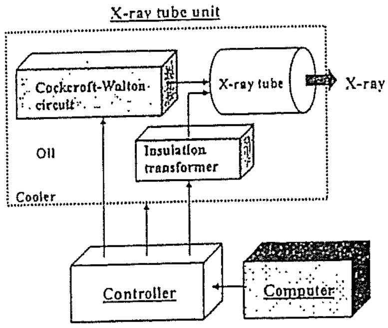

Fig. 1: Block diagram of compact x-ray generator with cerium-target tube.

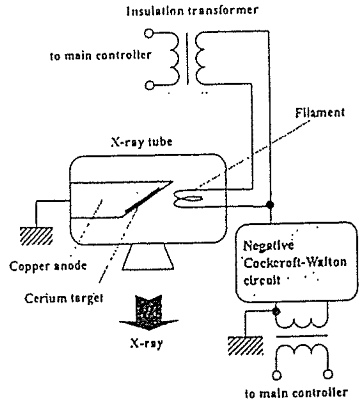

Fig. 2: Main circuit of x-ray generator.

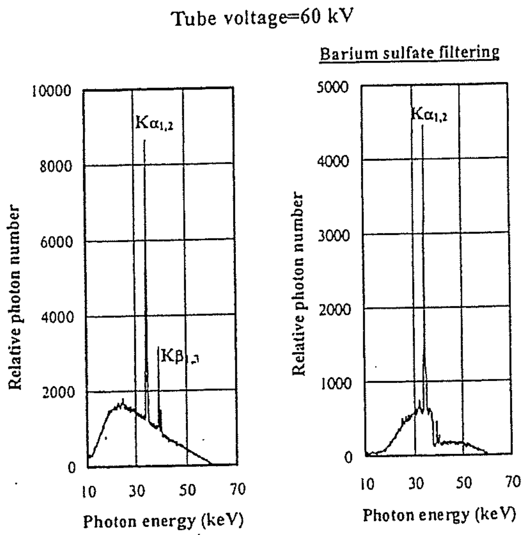

Fig. 3: X-ray spectra measured using germanium detector and filter.

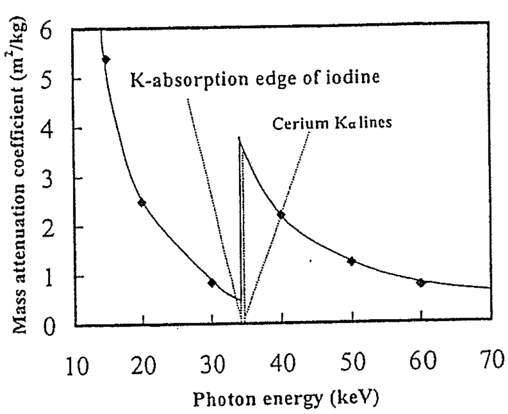

Fig. 4: Mass attenuation coefficients of iodine, and average photon energy of cerium Kα lines.

3. Characteristics

The x-ray intensity rate was measured by a Victoreen 660 ionization chamber at 1.0 m from the x-ray source. At a constant tube current of 0.40 mA, the x-ray intensity increased when the tube voltage was increased. In this measurement, the intensity with a tube voltage of 60 kV and a current of 0.40 mA was 16.8 μGy/s with errors of less than 0.2%.

In order to measure images of the x-ray source, we employed a pinhole camera with a hole diameter of 50 μm in conjunction with a Computed Radiography (CR) system[16] with a sampling pitch of 87.5 μm. When the tube voltage was increased, spot dimensions increased slightly and had values of approximately 1×1 mm.

In order to measure x-ray spectra, we employed a germanium detector (GLP-10180/07-P, Ortec Inc.) (Fig. 3). When the tube voltage was increased, the Kα intensity substantially increased, and both the maximum photon energy and the intensities of bremsstrahlung x rays increased.

4. Angiography

Figure 4 shows the mass attenuation coefficients of iodine at the selected energies; the coefficient curve is discontinuous at the iodine K-edge. The average photon energy of the cerium Kα lines is shown just above the iodine K-edge. Cerium is a rare earth element and has a high reactivity; however, the average photon energies of Kα is 34.6 keV, and iodine contrast mediums with a K-absorption edge of 33.2 keV absorb the lines easily. Therefore, blood vessels were observed with high contrasts.

The angiography was performed using the CR system (Konica Regius 150), iodine microspheres of 15 μm in diameter, and the filter. The distance (between the x-ray source and the imaging plate) was 1.5 m, and the tube voltage was 60 kV. Figure 5 shows angiograms of an extracted dog heart. Because the size of the dog heart is almost the same as human heart, human coronary arteries can be observed.

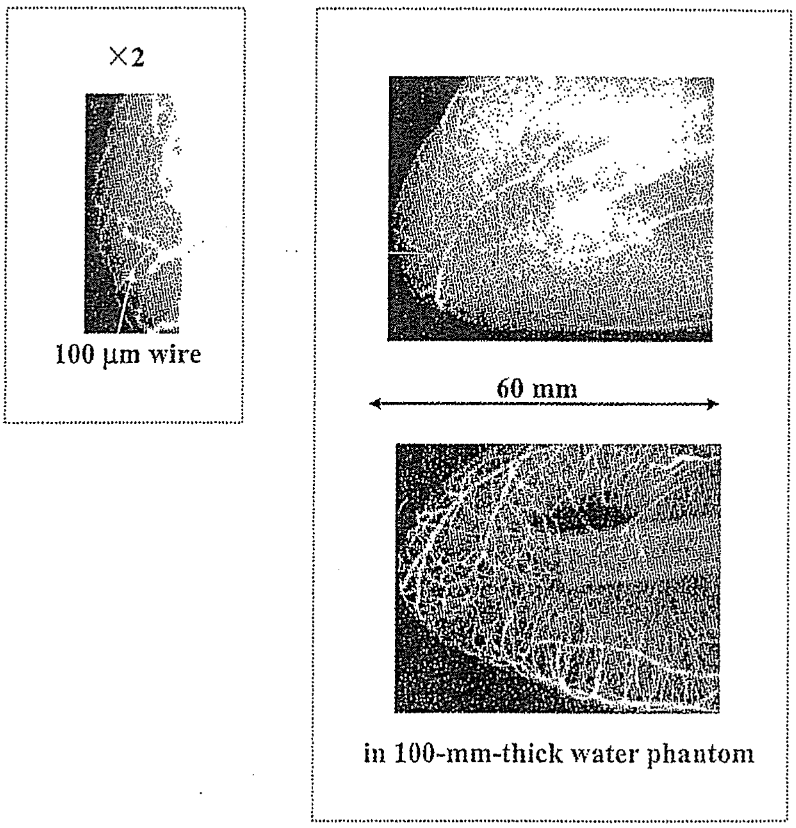

Fig. 5: Angiogram of extracted dog heart using iodine microspheres.

5. Discussion and Conclusions

In summary, we developed a new x-ray generator with a cerium-target tube and succeeded in producing cerium Kα lines, which can be absorbed easily by iodine-based contrast media. Both the characteristic and bremsstrahlung x-ray intensities increased with increases in the tube voltage, and Kβ lines were absorbed effectively by the barium sulfate filter.

In this preliminary experiment, although the maximum tube voltage and current were 65 kV and 0.40 mA, respectively, the voltage and current could be increased. Subsequently, the generator produced maximum number of Kα photons was approximately 3×10^7 photons/cm^2·s at 1.0 m from the source, and the photon count rate can be increased easily by improving the target.

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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Optical Engineering 44(9), 096502 (September 2005)

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

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0091-3286/2005/$22.00 © 2005 SPIE

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

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

Paper 040733R received Oct. 6, 2004; revised manuscript received Mar. 8, 2005; accepted for publication Mar. 11, 2005; published online Sep. 16, 2005. This paper is a revision of a paper presented at the SPIE conference on X-Ray Sources and Optics, August 2004, Denver, CO. The paper presented there appears (unrefereed) in SPIE Proceedings Vol. 5537.

1 Introduction

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

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

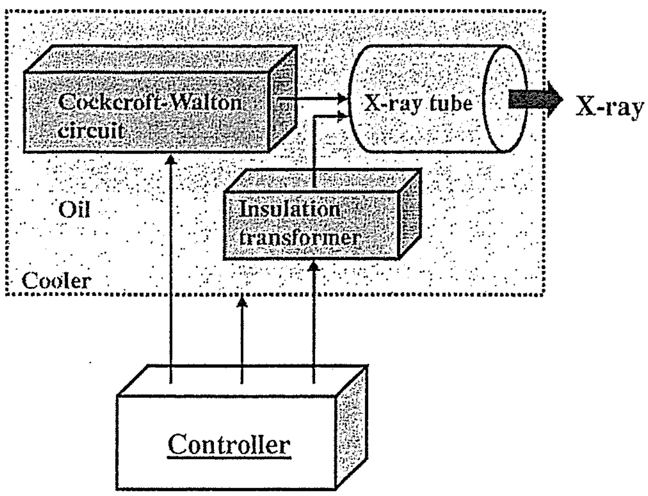

Fig. 1 Block diagram of the compact x-ray generator with a cerium-target radiation tube, which is used specially for K-edge angiography using iodine-based contrast media.

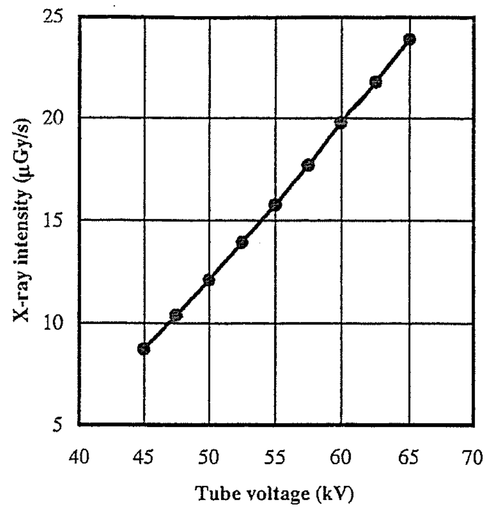

Fig. 3 The x-ray intensity (μGy/s) as a function of tube voltage (kV) with a tube current of 0.40 mA.

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

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

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

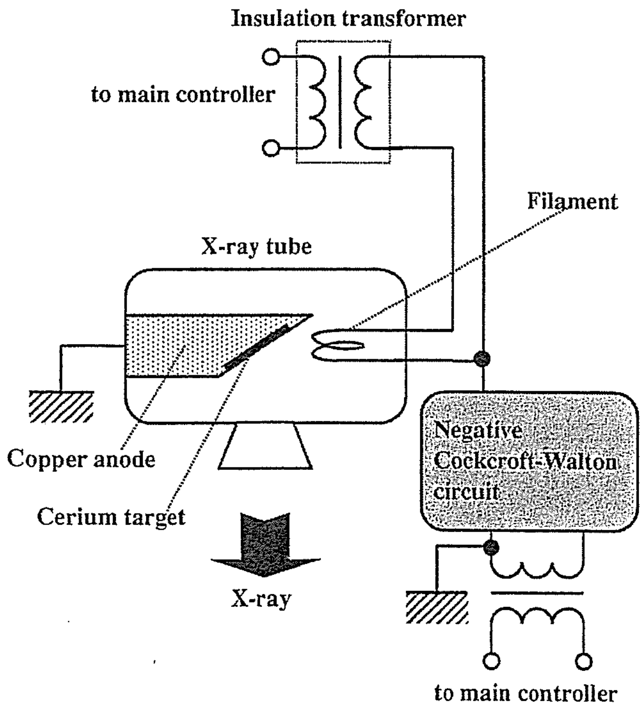

Fig. 2 Main circuit of the x-ray generator.

2 Experimental Setup

Figure 1 shows the block diagram of the x-ray generator, which consists of a main controller and an x-ray tube unit with a Cockcroft-Walton circuit and a cerium-target tube. The tube voltage, the current, and the exposure time can be controlled by the controller. The main circuit for producing x-rays is illustrated in Fig. 2, and employs the Cockcroft-Walton circuit to decrease the dimensions of the tube unit. In the x-ray tube, the negative high voltage is applied to the cathode electrode, and the anode (target) is connected to the tube unit case (ground potential) to cool the anode and the target effectively. The filament heating current is supplied by an AC power supply in the controller in conjunction with an insulation transformer. In this experiment, the tube

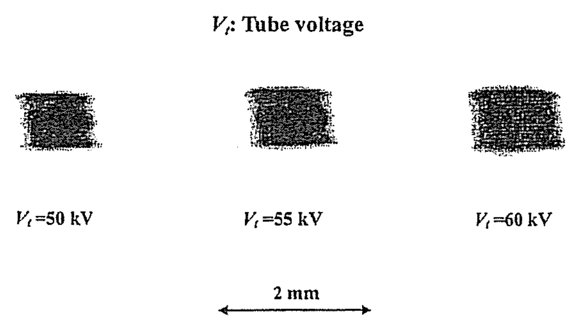

Fig. 4 Effective focal spots with changes in the tube voltage.

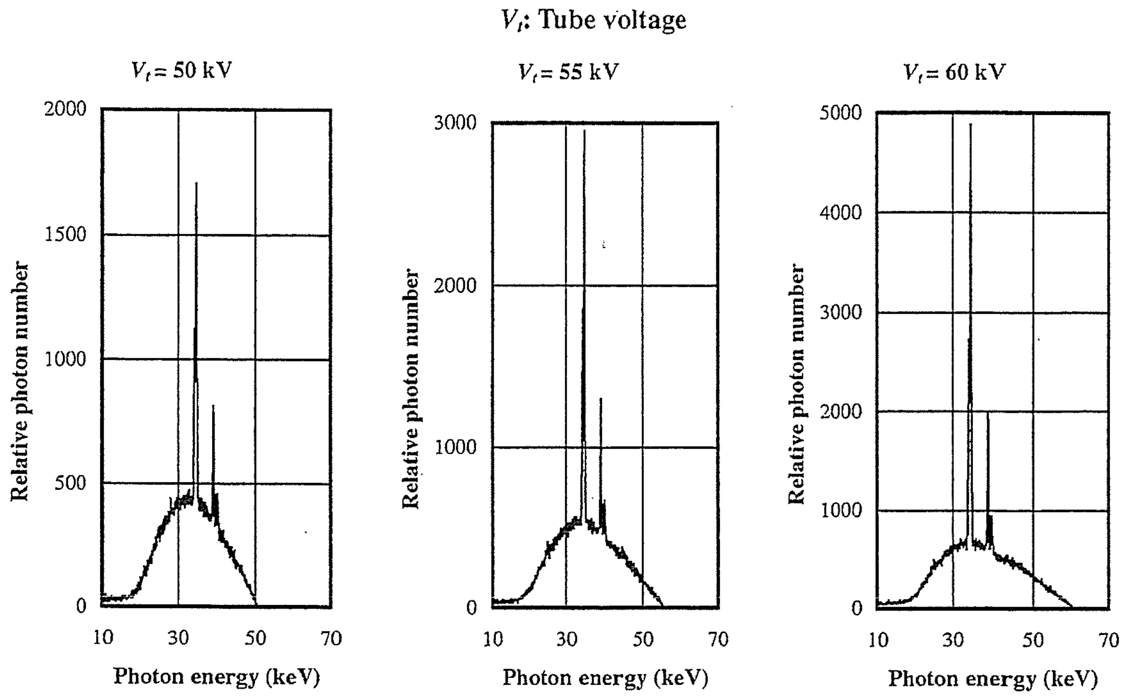

Fig. 5 X-ray spectra measured using a germanium detector with changes in the tube voltage.

voltage applied was from 45 to 65 kV, and the tube current was regulated to within 0.40 mA (maximum current) by the filament temperature. The exposure time is controlled to obtain optimum x-ray intensity. Quasimonochromatic x-rays are produced using a 3.0-mm-thick aluminum filter for absorbing soft bremsstrahlung rays.

3 Results and Discussion

3.1 *X-ray Intensity*

X-ray intensity was measured by a Victoreen 660 ionization chamber at 1.0 m from the x-ray source using the filter (Fig. 3). At a constant tube current of 0.40 mA, the x-ray intensity increased when the tube voltage was increased. In this measurement, the intensity with a tube voltage of 60 kV and a current of 0.40 mA was 19.9 μGy/s at 1.0 m from the source, with errors of less than 0.2%.

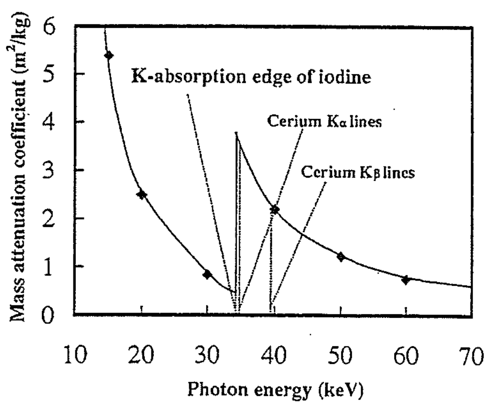

Fig. 6 Mass attenuation coefficients of iodine, and the average photon energies of cerium Kα and Kβ lines.

3.2 *Focal Spot*

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

3.3 *X-ray Spectra*

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

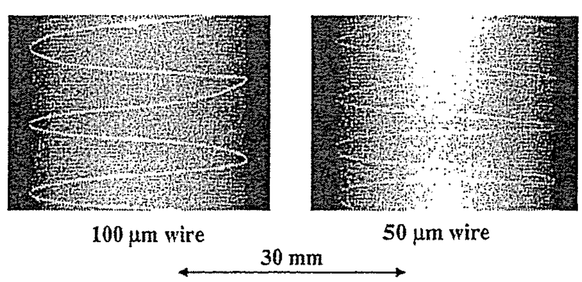

Fig. 7 Radiograms of tungsten wires coiled around PMMA rods with a tube voltage of 60 kV.

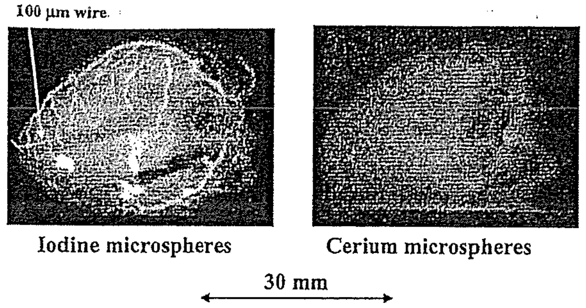

Fig. 8 Angiograms of extracted rabbit hearts using iodine and cerium microspheres with a tube voltage of 60 kV.

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

3.4 *K-Edge Angiography*

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

The angiography was performed by a CR system (Konica Minolta Regius 150) using the filter, and the distance (between the x-ray source and the imaging plate) was 1.5 m. First, rough measurements of spatial resolution were made using wires. Figure 7 shows radiograms of tungsten wires coiled around a rod made of polymethyl methacrylate. Although the image contrast decreased somewhat with decreases in the wire diameter, due to blurring of the image caused by the sampling pitch of 87.5 μm, a 50-μm-diam wire could be observed.

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

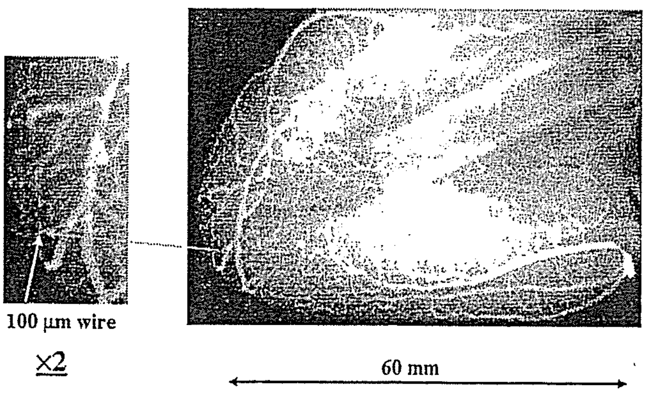

Fig. 9 Angiograms of an extracted dog heart using iodine microspheres with a tube voltage of 60 kV.

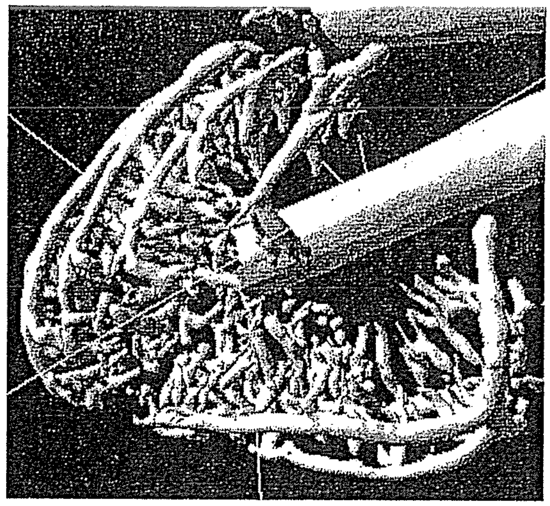

Fig. 10 3-D image of coronary arteries constructed from x-ray CT images by Pascal.

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

4 Conclusion and Outlook

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

Although the cerium x-ray generator used in this research produces both the characteristic and the bremsstrahlung x-rays, bremsstrahlung intensity can be decreased effectively by considering the angle dependence without using the filter, since bremsstrahlung rays are not emitted in the opposite direction to that of electron trajectory in Sommerfeld's theory.[22] Subsequently, the generator-produced maximum number of characteristic photons is approximately 3.5×10^7 photons/cm$^2\cdot$s at 1.0 m from the source, and the photon count rate can be increased easily by improving the target.

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

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

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

Acknowledgment

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 2003).

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