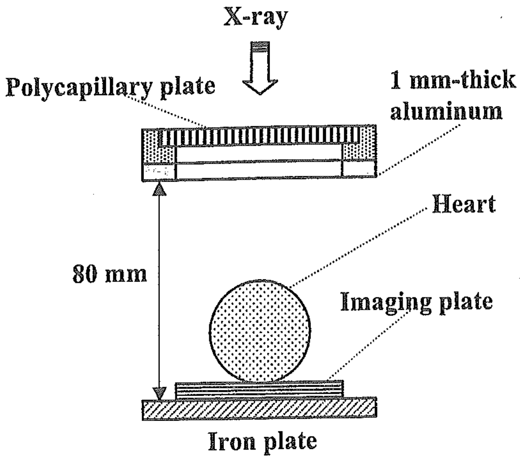

Fig. 16. Angiography of a heart extracted from a rabbit using iodine-based microspheres.

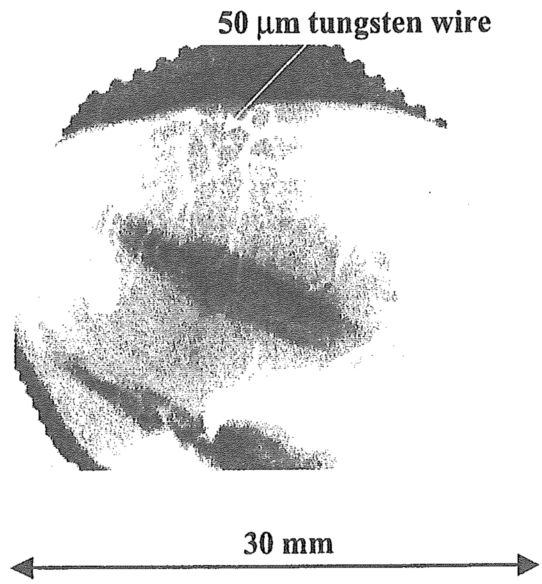

Fig. 17. Angiogram of the heart using the polycapillary.

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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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デジタルX線撮影システムを利用した準単色ファインポリキャピラリーイメージング
―― 医用X線レンズ ――

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要旨：ポリキャピラリープレートと銅対陰極付きX線管を用いた準単色X線撮影に関して記述した。管電圧は12から22 kVの範囲で調整され，管電流はフィラメントの温度により3.0 mA以下に調整された。X線照射時間は撮影に適正なX線強度が得られるように制御され，実効焦点サイズは2.0×1.5 mmであった。ポリキャピラリープレートは浜松ホトニクス社製のJ5022-16でプレート厚は1.0 mmであった。外径，有効径，そして孔径はそれぞれ33 mm，27 mm，10 μmであった。管電圧が17 kVの条件下で銅のK系列特性（準単色）X線は，厚さ10 μmの銅フィルターを透して出力され，これらのX線はポリキャピラリーにより準平行化された。X線像はイメージングプレート付きのデジタル撮影システム（CR）により撮影された。空間分解能はテストチャートとプレート間の距離を増しても変化しなかった。撮影では50 μmのタングステンワイヤーが認識され，造影では100 μm程度の微小血管が観察できた。

キーワード：準平行X線撮影，準単色X線，特性X線，X線レンズ，ポリキャピラリープレート

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Cone-beam K-edge angiography utilizing cerium x-ray generator in conjunction with cerium oxide filter
—— Observation of fine blood vessels ——

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Abstract : The cerium x-ray generator is useful in order to perform cone-beam K-edge angiography because K-series characteristic x rays from the cerium target are absorbed effectively by iodine-based contrast mediums. The x-ray generator consists of a main controller and a unit with a high-voltage circuit and a fixed anode x-ray tube. The tube is a glass-enclosed diode with a cerium target and a 0.5 mm-thick beryllium window. The maximum tube voltage and current were 65 kV and 0.4 mA, respectively, and the focal-spot sizes were 1.2 × 0.8 mm. Cerium K-series characteristic x rays were left using a cerium oxide filter, and the x-ray intensity was 0.50 μC/kg•s at 1.0 m from the source with a tube voltage of 60 kV, a current of 0.40 mA, and an exposure time of 1.0 s. Angiography was performed with a computed radiography system using iodine-based microspheres 15 μm in diameter. In angiography of non-living animals, we observed fine blood vessels of approximately 100 μm with high contrasts.

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

1. INTRODUCTION

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

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for K-edge angiography.

In order 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 to produce quasi-monochromatic x rays without using a K-edge filter. Therefore, we have performed a demonstration of cone-beam K-edge angiography[19] utilizing a cerium plasma generator, since K-series characteristic x rays from the cerium target are absorbed effectively by iodine.

Recently, we have developed a steady state x-ray generator utilizing a cerium-target tube, and have demonstrated enhanced K-edge angiography utilizing a barium sulfate filter.[20] In this research, $K\alpha$ lines (34.6 keV) were left by absorbing $K\beta$ lines (39.2 keV), and bremsstrahlung x rays with photon energies of lower than the barium K-edge (37.4 keV) were also observed. However, because cerium $K\beta$ lines are also absorbed effectively by iodine, both $K\alpha$ and $K\beta$ lines should be selected to perform angiography. In measurements of x-ray spectra, although we usually employed a cadmium tellurium detector with a photon energy resolution of 1.7 keV, the resolution should be improved as much as possible 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 using a cerium oxide K-edge filter.

2. GENERATOR

Figure 1 shows the block diagram of the x-ray generator, which consists of a main controller and an x-ray tube unit with a Cockcroft-Walton circuit and a cerium-target tube. The tube voltage, the current, and the exposure time can be controlled by the controller. The main circuit for producing x rays is illustrated in Fig. 2, and employed the Cockcroft-Walton circuit in order to decrease the dimensions of the tube unit. In the x-ray tube, the negative high voltage is applied to the cathode electrode, and the anode (target) is connected to the tube unit case (ground potential) to cool the anode and the target effectively. The filament heating current is supplied by an AC power supply in the controller in conjunction with an insulation transformer. The x-ray tube is a glass-enclosed diode with a cerium target and a 0.5-mm-thick beryllium window. 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 cerium oxide filter for absorbing bremsstrahlung rays.

In designing the filter, the surface density of the cerium oxide powder is important, since the x rays are absorbed effectively by the powder as compared

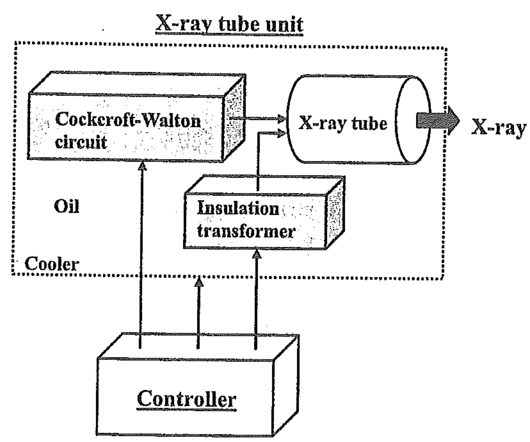

Fig. 1. Block diagram of compact x-ray generator with cerium-target radiation tube, which is used specially for K-edge angiography using iodine-based contrast mediums.

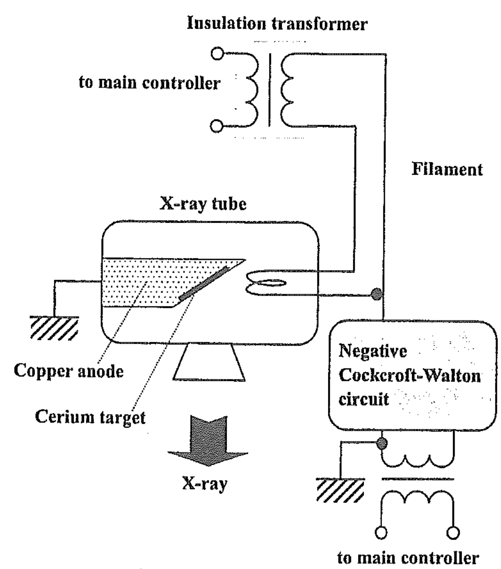

Fig. 2. Main circuit of x-ray generator.

with the PMMA powder. In this case, the density was approximately 10 mg/cm^2, and a K-edge powder filter (Fig. 3), consisting of cerium oxide and PMMA powders, was employed.

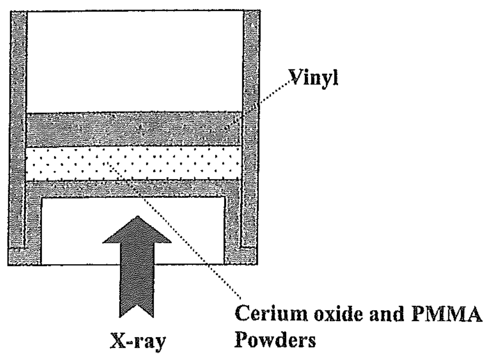

Fig. 3. Schematic drawing of cerium oxide powder filter.

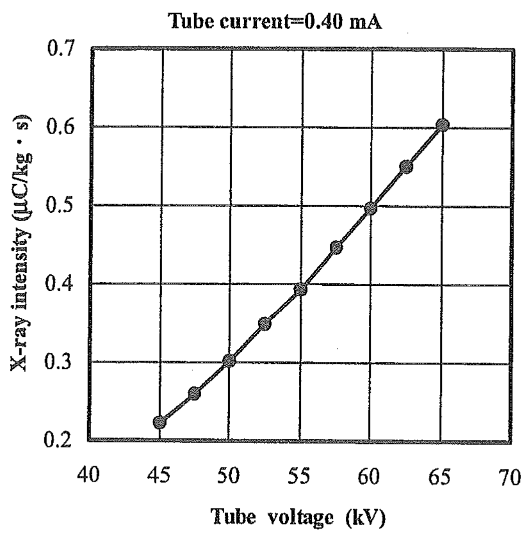

Fig. 4. X-ray intensity measured at 1.0 m from x-ray source according to changes in tube voltage.

3. CHARACTERISTICS

3.1 X-ray Intensity

X-ray intensity was measured by a Victoreen 660 ionization chamber at 1.0 m from the x-ray source using the filter (Fig. 4). At a constant tube current of 0.40 mA, the x-ray intensity increased when the tube voltage was increased. In this measurement, the intensity with a tube voltage of 60 kV and a current of 0.40 mA was 0.50 μC/kg・s at 1.0 m from the source with errors of less than 0.2%.

3.2 Focal Spot

In order to measure images of the x-ray source after the 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 1.2 $\times$ 0.8 mm (Fig. 5).

3.3 X-ray Spectra

In order to measure x-ray spectra, we employed a germanium detector (GLP-10180/07-P, Ortec Inc.) with a photon energy resolution of approximately 0.12 keV (Fig. 6). 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.

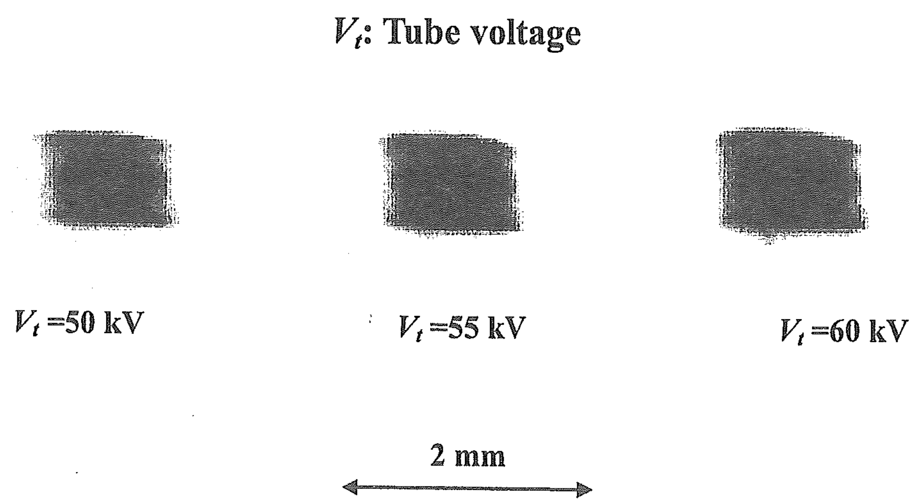

Fig. 5. Effective focal spots with changes in tube voltage.

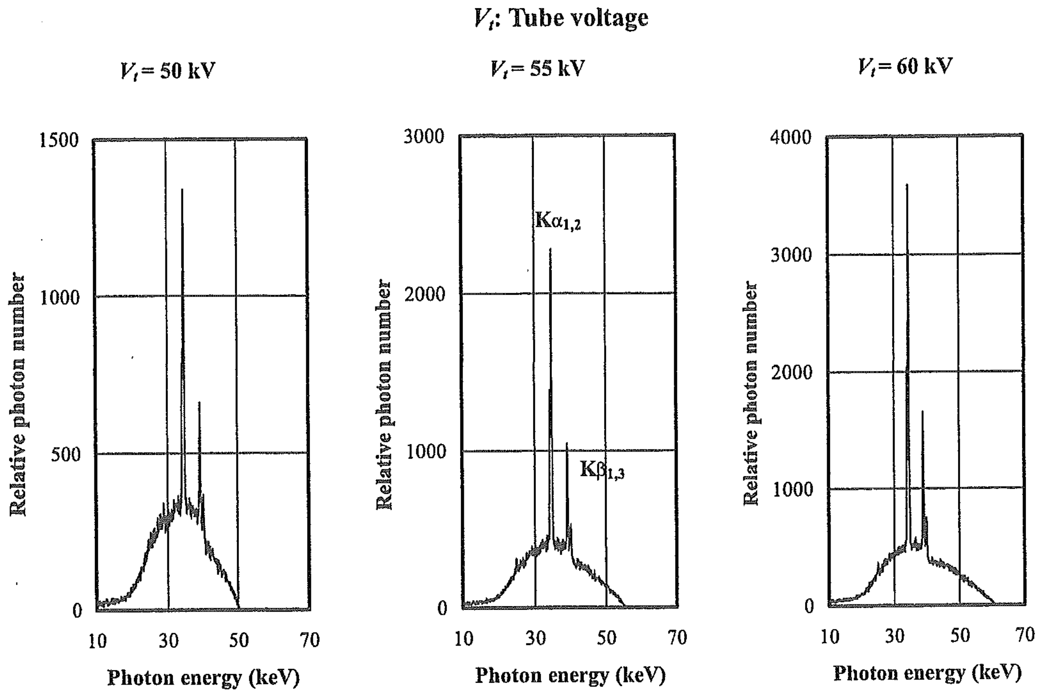

Fig. 6. X-ray spectra measured using germanium detector with changes in tube voltage.

4. K-EDGE ANGIOGRAPHY

Figure 7 shows the mass attenuation coefficients of iodine at the selected energies; the coefficient curve is discontinuous at the iodine K-edge. The average photon energy of the cerium Kα and Kβ lines are shown just above the iodine K-edge. Cerium is a rare earth element and has a high reactivity; however, the average photon energies of Kα and Kβ lines are 34.6 and 39.2 keV, respectively, and iodine contrast mediums with a K-absorption edge of 33.2 keV absorb the lines easily. Therefore, blood vessels were observed with high contrasts.

The angiography was performed by the CR system (Konica Regius 150) using the filter, and the tube voltage and the distance (between the x-ray source and the imaging plate) were 60 kV and 1.5 m, respectively. Firstly, rough measurements of spatial resolution were made using wires. Figure 8 shows radiograms of tungsten wires coiled around a rod made of polymethyl methacrylate. Although the image contrast decreased somewhat with decreases in the wire diameter, due to blurring of the image caused by the sampling pitch of 87.5 μm, a 50 μm-diameter wire could be observed.

Angiograms of rabbit hearts are shown in Fig.

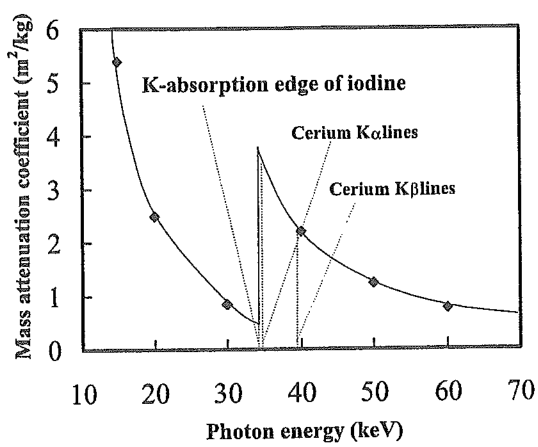

Fig.7. Mass attenuation coefficients of iodine, and average photon energies of cerium Kα and Kβ lines.

9. These two images were obtained using iodine and cerium microspheres of 15 μm in diameter. In the case where the cerium spheres were employed, the coronary arteries were barely visible. Figures 10 and 11 show angiograms of a larger dog heart and a rabbit thigh, respectively, using iodine spheres. In angiography, the coronary arteries were visible, and fine blood vessels of approximately 100 μm could be seen.

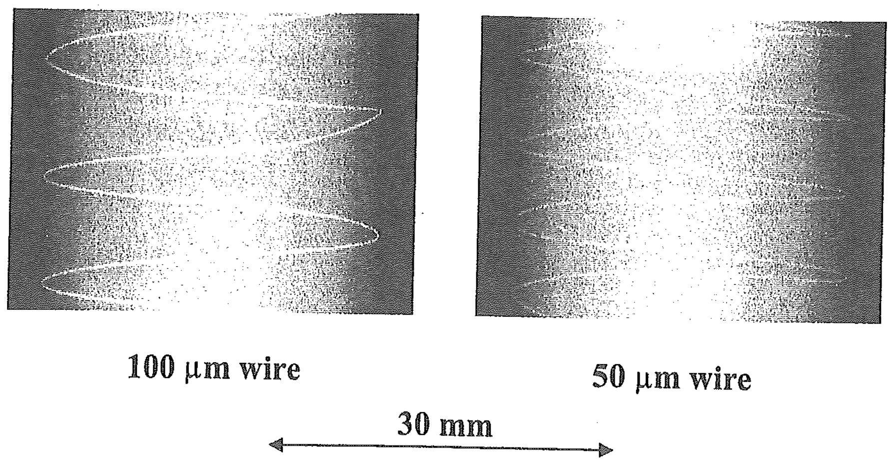

Fig. 8. Radiograms of tungsten wires in PMMA rod with tube voltage of 60 kV.

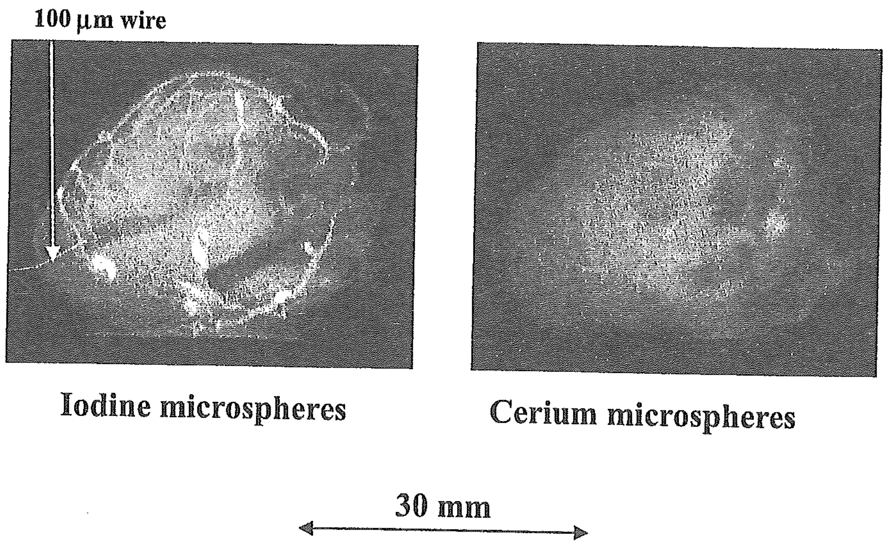

Fig. 9. Angiograms of extracted rabbit hearts using iodine and cerium microspheres with tube voltage of 60 kV.

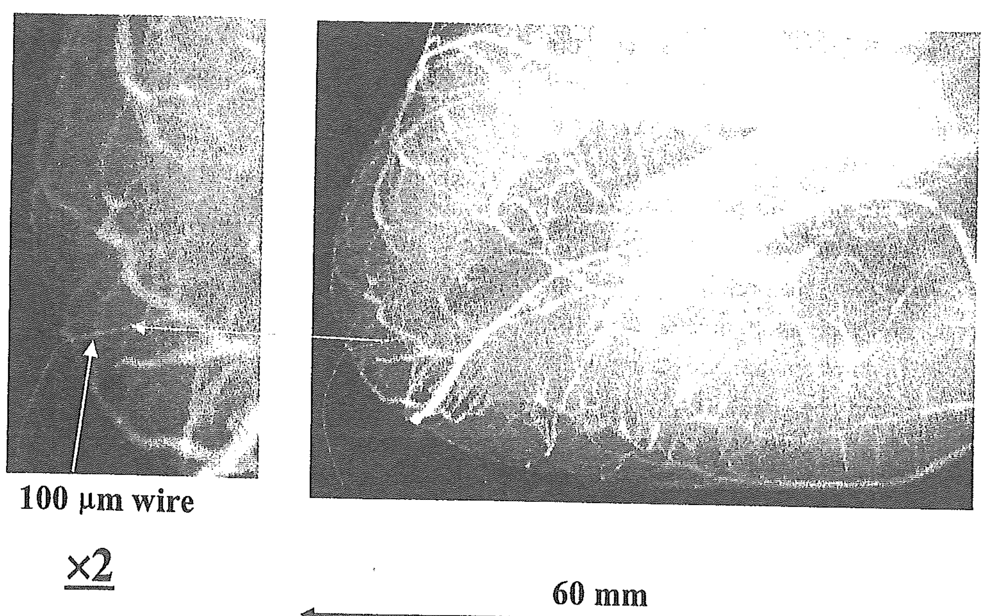

Fig. 10. Angiogram of extracted dog heart using iodine microspheres with tube voltage of 60 kV.

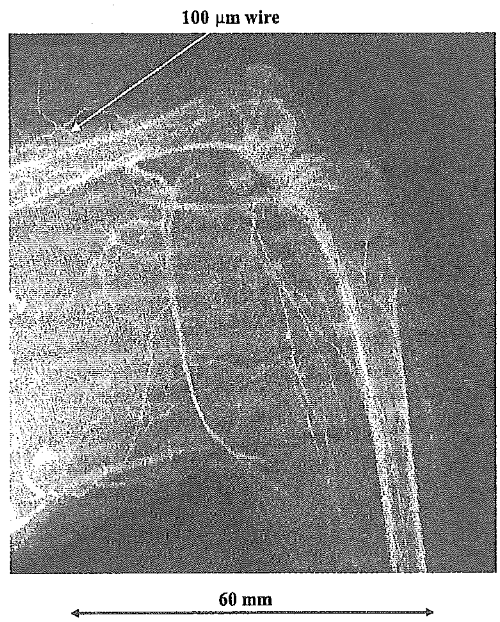

Fig. 11. Angiogram of rabbit thigh with tube voltage of 60 kV.

5. DISCUSSION AND CONCLUSIONS

In summary, we employed an x-ray generator with a cerium-target tube and succeeded in producing cerium characteristic x rays, which can be absorbed easily by iodine-based contrast mediums. The characteristic x-ray intensities increased with increases in the tube voltage, and bremsstrahlung rays were absorbed effectively by the filter.

Although the cerium x-ray generator used in this research produces both the characteristic and the bremsstrahlung x rays, bremsstrahlung intensity can be decreased effectively by considering the angle dependence without using the filter, since bremsstrahlung rays are not emitted in the opposite direction to that of electron acceleration. Subsequently, the generator produced maximum number of characteristic photons was approximately 3×10^7 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.

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セリウムX線装置と酸化セリウムフィルターを利用したコーンビームKエッジ造影
── 微小血管の観察 ──

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要旨：セリウム対陰極から発生するK系列特性X線はヨウ素系造影剤に効率良く吸収されるので，コーンビームによるKエッジ造影に有用である。X線装置はメインコントローラー，そして高電圧回路とX線管のユニットなどからなる。X線管はガラス封じ込み二極管で，セリウム対陰極と0.5 mm厚のベリリウム窓を有する。管電圧と電流の最大値はそれぞれ65 kVと0.4 mAで，実効焦点サイズは1.3×0.9 mmであった。セリウムのK系列特性X線は酸化セリウムのフィルターを用いて制動X線を吸収することにより得られ，X線強度は線源から1.0 mの位置で，管電圧60 kV，そして管電流0.4 mAの条件下で，0.5 μC/kg•sであった。血管には直径15 μmのヨウ素プラスチック微小球が充填され，デジタル撮影装置（CR）で造影された。動物ファントムの造影では，100 μm程度の血管が高コントラストで観察できた。

キーワード：X線発生装置，セリウムターゲット，準単色X線，特性X線，Kエッジ造影

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Variations in Cerium X-ray Spectra and Enhanced K-Edge Angiography

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A cerium-target X-ray tube is useful in 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 1.0-mm-focus diode with a cerium target and a 0.5-mm-thick beryllium window. The maximum tube voltage and current were 65 kV and 0.4 mA, respectively. Cerium Kα rays were selected out using a barium sulfate filter, and the X-ray intensities without filtering and with a barium sulfate filter were 209 and 16.8 μGy/s, respectively, at 1.0 m from the source with a tube voltage of 60 kV and a current of 0.40 mA. Angiography was performed with an X-ray film using the filter and iodine-based microspheres 15 μm in diameter. In the angiography of nonliving animals, we observed fine blood vessels approximately 100 μm in diameter with high contrasts.
[DOI: 10.1143/JJAP.44.8204]

KEYWORDS: X-ray tube, cerium target, monochromatic X-rays, Kα rays, K-edge angiography

1. Introduction

Monochromatic parallel X-ray beams have been used to perform enhanced K-edge angiography[1–4)] using iodine-based contrast media because the X-rays with photon energies of approximately 35 keV are absorbed easily by iodine with a K-edge of 33.2 keV. In conjunction with a high-resolution camera, fine blood vessels of approximately 50 μm can be observed.[3)] Although the parallel beams have also been employed to perform phase-contrast radiography,[5–7)] weakly absorbing materials have been observed with high contrasts.

Flash radiography of biomedical tissues has been investigated for a number of years, and several different flash X-ray generators have been developed corresponding to specific radiographic objectives.[8–11)] The advantages of flash radiography include the use of K-series characteristic X-rays and their relatively good imaging contrast. However, monochromatic flash radiography[12–15)] has encountered difficulties in increasing X-ray duration, and in performing X-ray computed tomography (CT).

Recently, a steady-state X-ray generator utilizing a cerium-target tube[16)] has been developed, and has been employed to perform enhanced K-edge angiography achieved with cerium Kα rays and iodine-based contrast media, since Kα rays (34.6 keV) are absorbed effectively by iodine. In this case, because the sampling pitch of a computed radiography system (Konica Minolta Regius 150)[17)] is 87.5 μm, a spatial resolution of approximately 100 μm has been obtained. Therefore, the resolution should be minimized by using a film or decreasing the pitch.

In the above-mentioned preliminary experiment,[16)] we employed a cadmium tellurium detector with a photon energy resolution of 1.7 keV to measure X-ray spectra from the cerium target. However, the resolution should be minimized to measure the characteristic X-ray intensity and to confirm the K-edge effect of a barium sulfate filter for absorbing Kβ and bremsstrahlung X-rays, because the photon energy width of the K-series lines is approximately 1 keV.

In the present research, we measured the X-ray spectra from a cerium target tube using a germanium detector with a photon energy resolution of 0.12 keV and performed a preliminary study on enhanced K-edge angiography achieved with cerium Kα rays.

2. Generator

Figure 1 shows a block diagram of the X-ray generator, which consists of a main controller and an X-ray tube unit

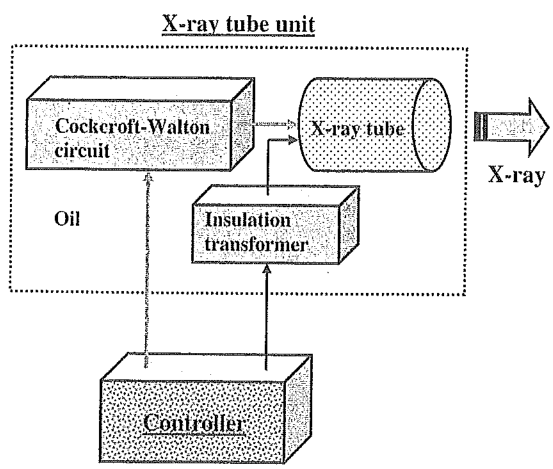


Fig. 1. Block diagram of compact X-ray generator with cerium-target radiation tube, which is used particularly for K-edge angiography using iodine-based contrast media.

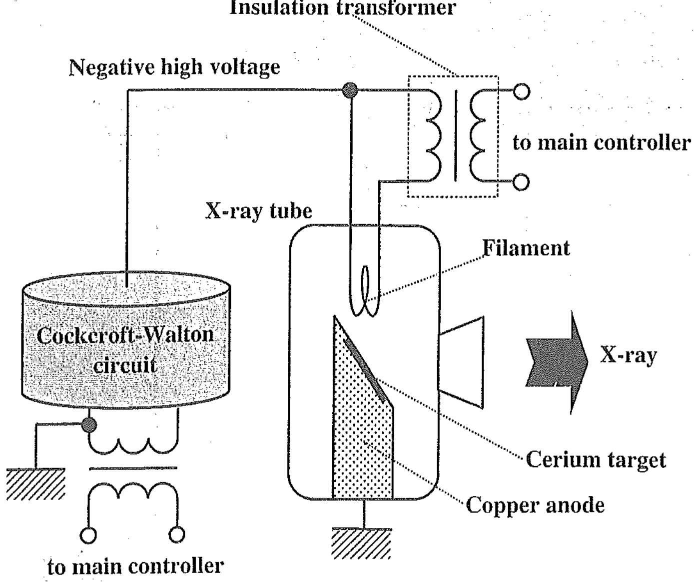


Fig. 2. Main circuit of X-ray generator.

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 it employed the Cockcroft–Walton circuit in order to decrease the dimensions of the tube unit. In the X-ray tube, a high negative 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. The tube is a conventional diode with a plate cerium target, a 1.0 mm focus, a take-off angle of 22°, and a 0.5-mm-thick beryllium window. In this experiment, the tube voltage 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 rays are selected out using a barium sulfate filter for absorbing bremsstrahlung and Kβ rays. 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 poly(methyl methacrylate) (PMMA) resin. In this case, the density was approximately 30 mg/cm^2.

3. Characteristics

3.1 X-ray intensity

The X-ray intensity rate was measured by a Victoreen 660 ionization chamber at 1.0 m from the X-ray source (Fig. 3). At a constant tube current of 0.40 mA, the X-ray intensity increased when the tube voltage was increased. At a tube voltage of 60 kV and a current of 0.40 mA, the intensities without filtering and with the filter were 208 and 16.8 μGy/s, respectively, with errors of less than 0.2%. The X-ray intensity was limited because the thermal contact between the target and the anode was not good.

3.2 X-ray spectra

In order to measure X-ray spectra, we employed a germanium detector (GLP-10180/07-P, Ortec Inc.) (Fig. 4 Without filtering, when the tube voltage was increased, th X-ray intensities of cerium K-series characteristic line increased, and both the maximum photon energy and th bremsstrahlung X-ray intensity increased. Using the filte both the Kβ lines and the bremsstrahlung X-rays with photo energies higher than the barium K-edge of 37.4 keV wer absorbed effectively, and sharp Kα lines were left. Wit increases in the tube voltage, the Kα intensity substantiall increased, and the maximum photon energy increased.

In order to perform K-edge angiography, the Kα rays ar useful, and the high-energy bremsstrahlung X-rays decreas the image contrast. Using the filter, because bremsstrahlun X-rays with energies higher than 60 keV were not absorbe easily, the tube voltage for angiography was determined t be 60 kV. Subsequently, low-energy bremsstrahlung ray

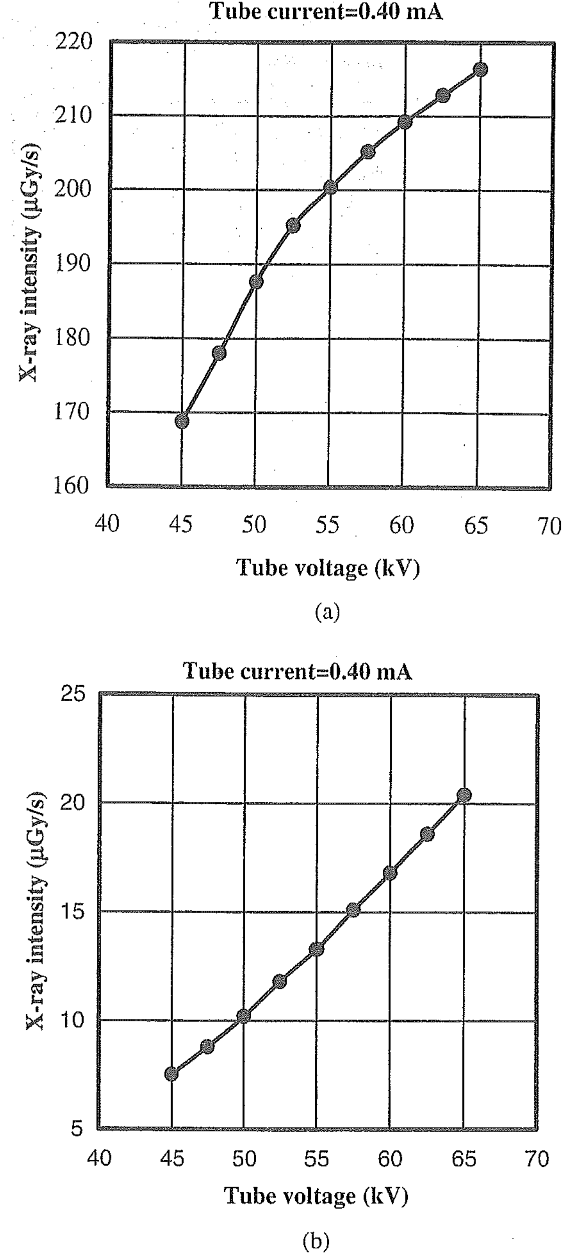
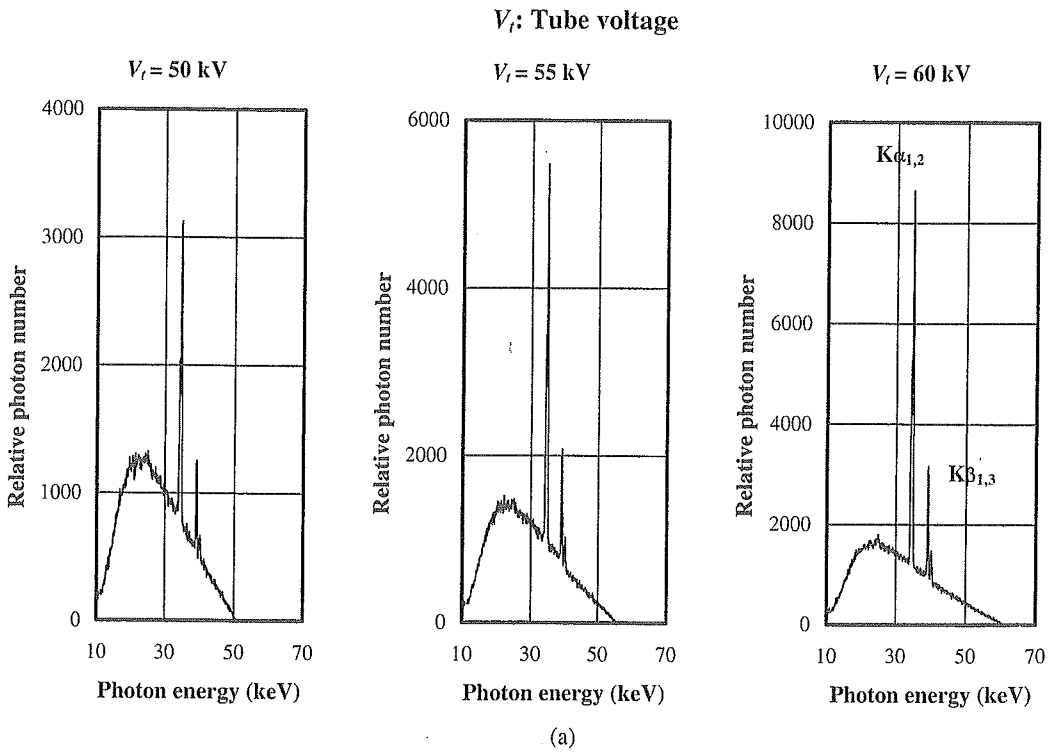


Fig. 3. X-ray intensity measured at 1.0 m from X-ray source according t changes in tube voltage (a) without filtering and (b) using barium-sulfa filter.



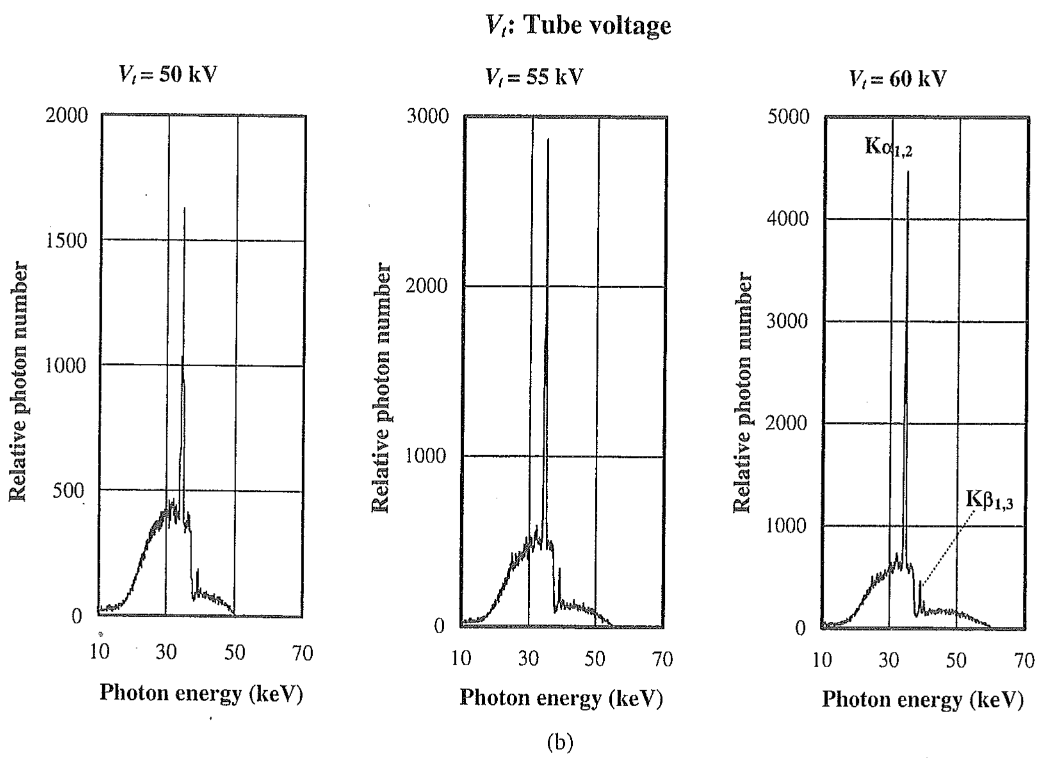


Fig. 4. X-ray spectra measured using germanium detector with changes in tube voltage (a) without filtering and (b) using barium sulfate filter.

with energies lower than the K-edge should be minimized using the filter or an aluminum filter to increase the blood-vessel contrast, since the iodine contrast media transmit the rays easily.

4. K-edge Angiography

Because the average photon energy of $K\alpha$ is 34.6 keV, iodine contrast media with a K-absorption edge of 33.2 keV absorb the $K\alpha$ lines easily. Therefore, blood vessels were observed with high contrasts. In order to observe fine blood vessels approximately 50 µm in diameter, the angiography was performed using an X-ray film (Fuji IX 100), iodine microspheres 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. First, rough measurements of spatial resolution were made using wires. Figure 5 shows radiograms of tungsten wires coiled around rods made of PMMA with an X-ray exposure time of 300 s. Although the

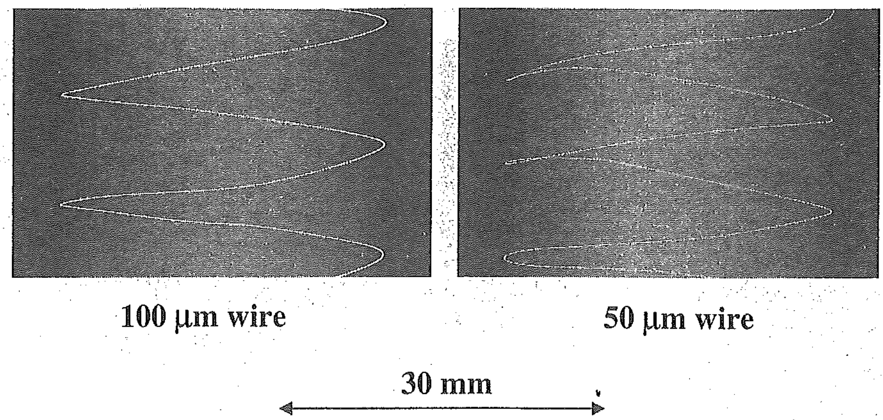


Fig. 5. Radiograms of tungsten wires coiled around PMMA rods.

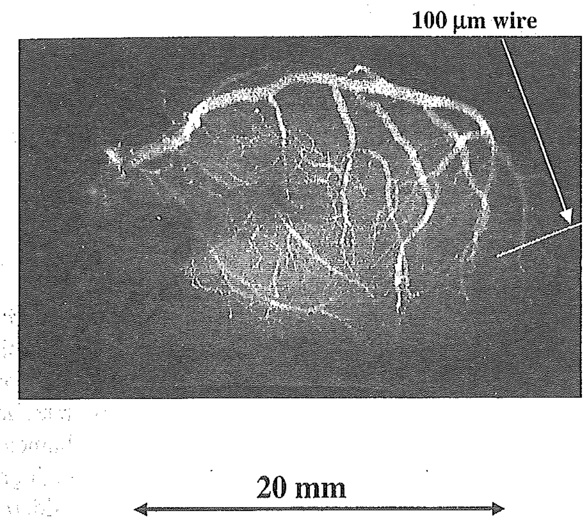


Fig. 6. Angiograms of extracted rabbit heart using iodine microspheres.

image contrast hardly varied with decreases in the wire diameter, a 50-μm-diameter wire could be observed clearly.

Figures 6 and 7 show angiograms of a rabbit heart and a thigh, respectively, with an exposure time of 300 s. The coronary arteries in the heart and fine blood vessels in the thigh with diameters of approximately 100 μm were visible. Figure 8 shows an angiogram of a dog heart in a 100-mm-thick water phantom with an exposure time of 1,500 s. Because the size of the dog heart is almost the same as that of a human heart, human coronary arteries can be observed. For comparison, we show a three-dimensional (3D) image of the coronary arteries constructed from X-ray CT images taken by Pascal (Digital Culture Tech. Corp.) with a tungsten X-ray tube (Fig. 9). This heart was the same as that used in K-edge angiography and was observed from the same direction by rotating the three-dimensional (3D) image; CT angiography was performed without using the water phantom. Using this 3D angiography achieved with a multislice helical CT, fine blood vessels were not observed at all.

5. Discussion

In the present research, we employed an X-ray generator with a cerium-target tube and succeeded in producing cerium characteristic X-rays, which can be absorbed easily by iodine-based contrast media. Both the characteristic and bremsstrahlung X-ray intensities increased with tube voltage

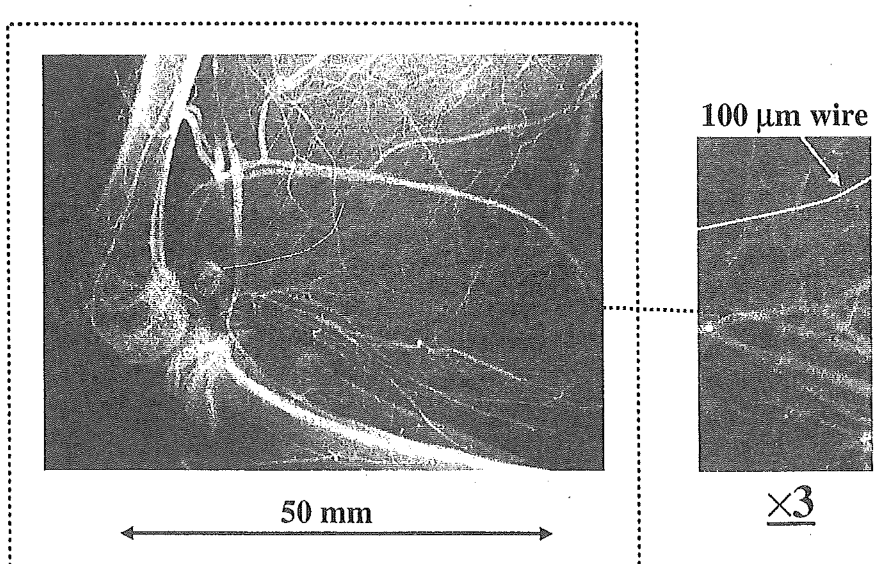


Fig. 7. Angiogram of rabbit thigh using iodine microspheres.

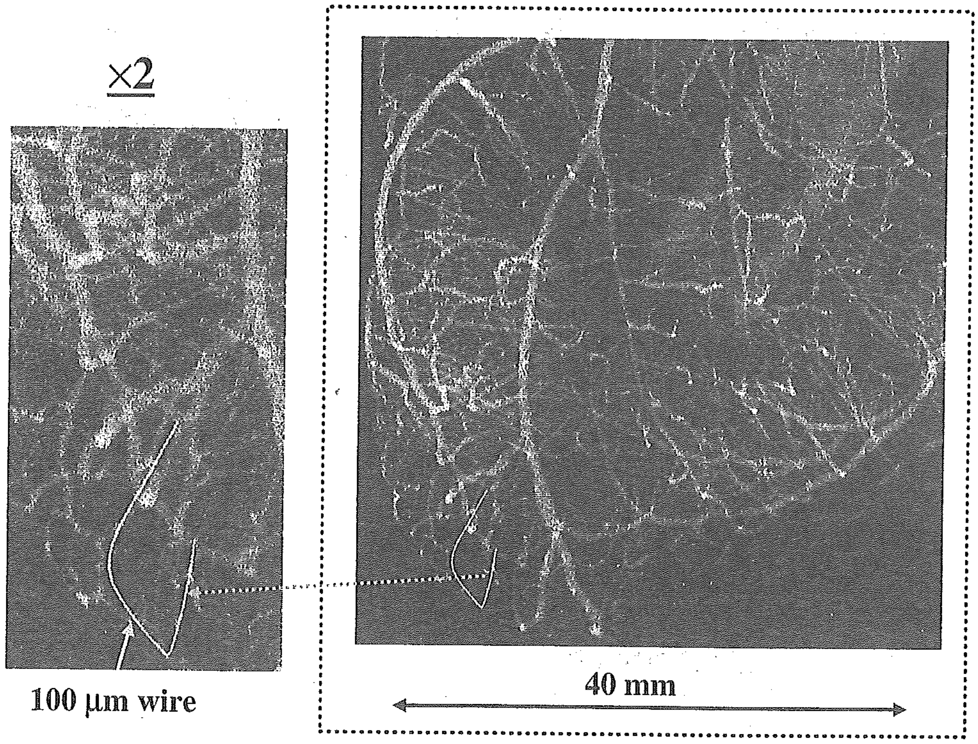


Fig. 8. Angiogram of extracted dog heart in 100-mm-thick water phantom using iodine microspheres.

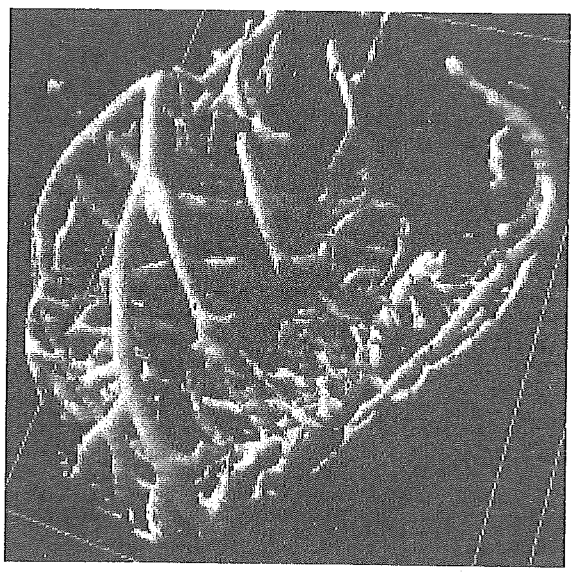


Fig. 9. Three-dimensional image of coronary arteries constructed from X-ray CT images taken by Pascal.

without filtering. Using the filter to absorb $K\beta$ and bremsstrahlung X-rays, $K\alpha$ rays were left, and we performed K-edge angiography using the filter with a tube voltage of 60 kV. To produce clean characteristic X-rays without using the filter, the angle dependence of the bremsstrahlung intensity should be considered, since bremsstrahlung rays are not emitted in the direction opposite that of the electron trajectory in Sommerfeld's theory.[18)]

Currently, angiography is performed using both the bremsstrahlung and characteristic X-rays produced from a tungsten X-ray tube. However, enhanced K-edge angiography in this work was primarily performed using cerium $K\alpha$ rays. Using the filter, the maximum number of $K\alpha$ 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. For example, a new rotation anode tube has been designed to increase the X-ray dose rate, and the rate can be increased by increasing the anode diameter.

In energy-selective imaging including K-edge angiography, the filtering effect of the absorber should be considered, and the X-ray spectra using the filter at a tube voltage of 60 kV hardly varies with changes in the thickness of the water phantom according to the spectrum estimation. Due to the absorption coefficient, $K\beta$ rays are also useful for angiography, and both the $K\alpha$ and $K\beta$ rays can be left using a cerium oxide filter with a surface density of approximately 10 mg/cm^2. In addition, an aluminum filter with a thickness of approximately 3.0 mm is useful in absorbing unnecessary bremsstrahlung X-rays with energies lower than the K-absorption edge.

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Enhanced K-edge Angiography Utilizing Tantalum Plasma X-ray Generator in Conjunction with Gadolinium-Based Contrast Media

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The tantalum plasma flash X-ray generator is useful for performing high-speed enhanced K-edge angiography using cone beams because K-series characteristic X-rays from the tantalum 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 cold-cathode 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 K-series characteristic X-ray intensities of cerium increased. The K lines were clean and intense, and hardly any bremsstrahlung rays were detected. The X-ray pulse widths were approximately 100 ns, and the time-integrated X-ray intensity had a value of approximately 300 μGy at 1.0 m from the X-ray source with a charging voltage of 80 kV. Angiography was performed using a filmless computed radiography (CR) system and gadolinium-based contrast media. In the angiography of nonliving animals, we observed fine blood vessels of approximately 100 μm with high contrasts. [DOI: 10.1143/JJAP.44.8716]

KEYWORDS: angiography, gadolinium-based contrast media, characteristic X-rays, quasi-monochromatic X-rays, tantalum K lines

1. Introduction

Enhanced K-edge angiography[1–4] has been performed utilizing monochromatic parallel X-ray beams produced from synchrotron orbital radiation using a monochrocollimator. The photon energies of the beams are approximately 35 keV, and are absorbed effectively by iodine-based contrast media with a K-absorption edge of 33.2 keV. Nowadays, an X-ray generator with a cerium-target tube[5] can be used in order to perform the K-edge angiography because K-series characteristic X-rays with photon energies just beyond the K-edge are absorbed effectively by iodine.

To perform high-speed biomedical radiography, we have developed several different high-dose-rate X-ray generators corresponding to specific objectives. For example, flash X-ray generators[6–9] with cold-cathode tubes produce extremely short X-ray pulses with durations of less than 1 μs, and the X-ray duration can be controlled accurately from 10 μs to 1.0 ms in cases where stroboscopic X-ray generators[10,11] utilizing hot-cathode triodes are employed.

Recently, although clean K-series characteristic X-rays of copper[12] and nickel[13] have been produced using plasma flash X-ray generators, low-intensity bremsstrahlung X-rays have been observed using a molybdenum target.[14] Therefore, we have performed preliminary experiments for producing clean high-photon-energy characteristic X-rays from molybdenum, silver and cerium targets using a compact flash X-ray generator with a disk-cathode tube,[15] and have succeeded in producing clean characteristic X-rays using the angle dependence of bremsstrahlung X-ray distributions. However, the X-ray intensity should be increased to a sufficient level for iodine angiography by increasing the electrostatic energies in the generator.

Since K-series characteristic X-rays from ytterbium, tantalum, and tungsten targets are absorbed effectively by gadolinium-based contrast media used in MRA, these X-rays are very useful for performing enhanced K-edge angiography. As compared with K-edge angiography using an iodine medium with an X-ray photon energy of 35 keV, the absorbed dose can be decreased easily in cases where the gadolinium medium is employed.

In the present research, we developed an intense quasi-monochromatic plasma flash X-ray generator with a tantalum target tube, and used it to perform a preliminary study on angiography achieved with tantalum K-series characteristic X-rays.

2. Principle of 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 tantalum $K\alpha$ lines is shown above the gadolinium K-edge. The average photon energy of tantalum $K\alpha$ lines is 57.1 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.

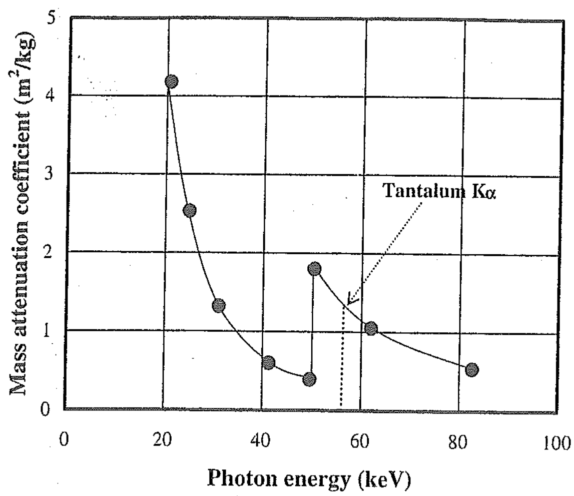

Fig. 1. Mass attenuation coefficient of gadolinium. The average photon energy of tantalum Kα lines is shown above the gadolinium K-edge.

3. Generator

3.1 High-voltage circuit

Figure 2 shows a block diagram including the electric circuit 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 tantalum 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 tantalum ions and electrons, around the target. Because bremsstrahlung rays are not emitted in the opposite direction to that of the electron trajectory in Sommerfeld's theory[16] (Fig. 4), tantalum K-series characteristic X-rays can be produced without using a filter.

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 (two times the charging voltage) and 40 kA, respectively.

4.2 X-ray output

The 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 charging

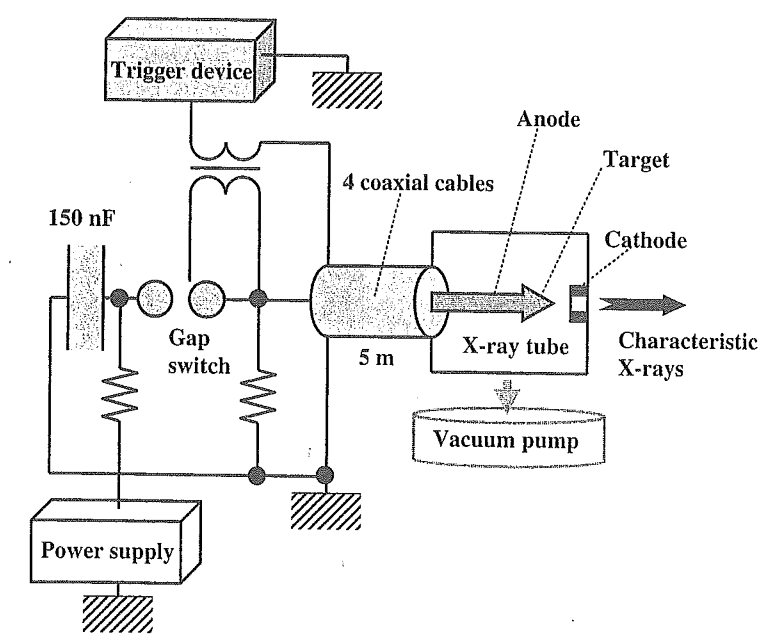

Fig. 2. Block diagram including high-voltage circuit of intense quasi-monochromatic plasma flash X-ray generator with tantalum-target tube.

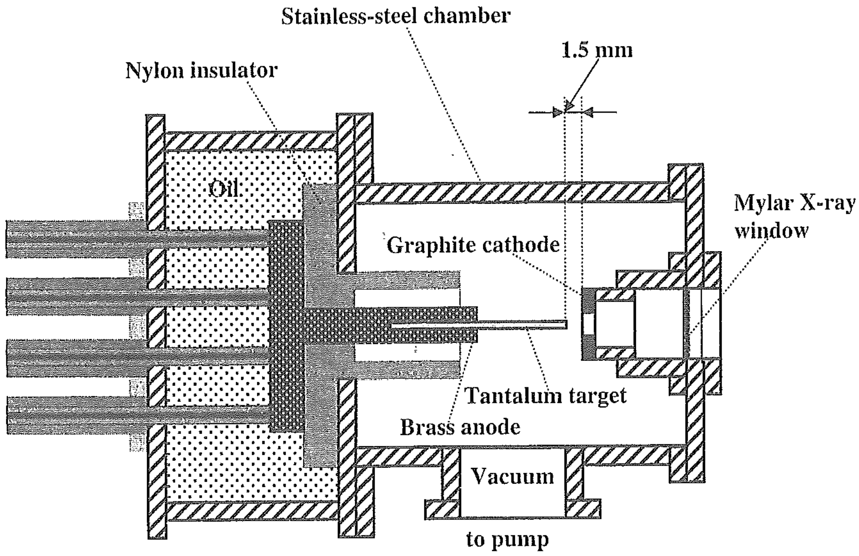

Fig. 3. Schematic drawing of flash X-ray tube with rod-shaped tantalum target.

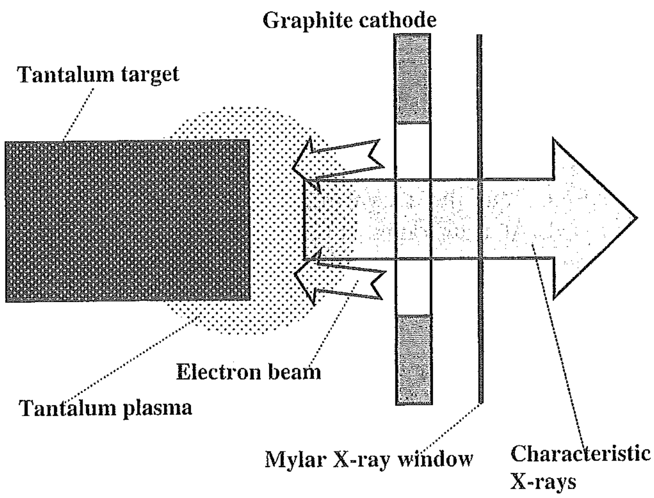

Fig. 4. Irradiation of K-series characteristic X-rays of tantalum.

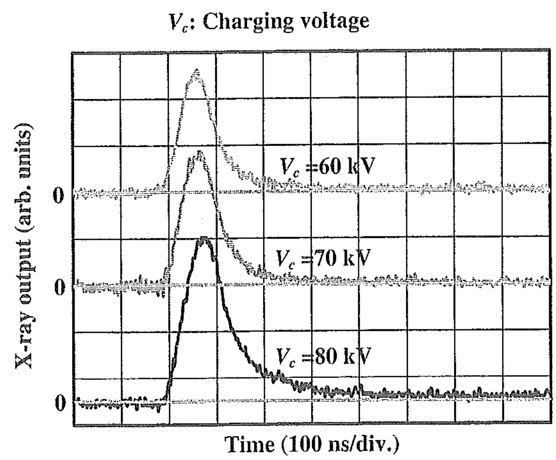

Fig. 5. X-ray outputs detected using combination of plastic scintillator and photomultiplier.

voltage. The X-ray pulse widths were approximately 100 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 300 μGy per pulse at 1.0 m from the X-ray source with a charging voltage of 80 kV.

4.3 X-ray source

In order to observe the characteristic X-ray source, we employed a 100-μm-diameter pinhole camera and an X-ray film (Polaroid XR-7) (Fig. 6). When the charging voltage was increased, the plasma X-ray source grew, and both spot dimension and intensity increased. Because the X-ray intensity is the highest at the center of the spot, both the dimension and intensity decreased as the thickness of a filter for absorbing X-rays increased and as the pinhole diameter decreased.

4.4 X-ray spectra

X-ray spectra were measured using a transmission-type spectrometer[14] with a lithium fluoride curved crystal 0.5 mm in thickness. The X-ray intensities of the spectra were detected by an imaging plate of a CR system[17] (Konica Regius 150) with a wide dynamic range, and relative X-ray intensity was calculated from Dicom original digital data corresponding to X-ray intensity; the data was scanned by a

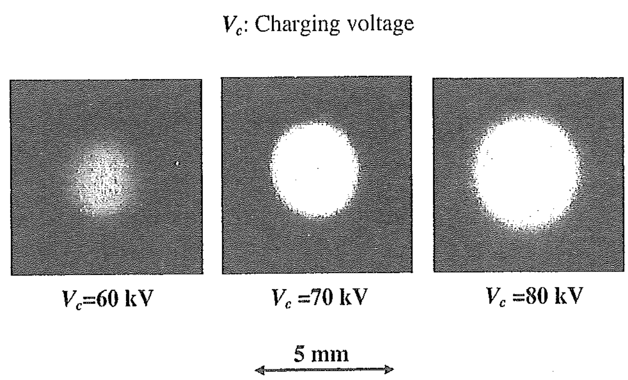

Fig. 6. Images of characteristic X-ray source obtained using pinhole camera with changes in charging voltage.