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ORIGINAL PAPER

Low Exposure X-ray Transmission Measurements for Contrast Media Detection with Filtered X-rays

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A feasibility study of X-ray transmission measurements for contrast media with less exposure using filtered X-rays and their energy information is described. Based on measurements of the energy spectra of La-filtered X-rays, sensitivity to the contrast media and the relative quantity of dose exposure of filtered X-rays as the ratio to white X-rays are shown. The dose exposure for the measurement of iodine contrast media is reduced by more than 50% with a 400 μm thick La filter. By using a filter with atomic number close to that of iodine such as Cs and Ba, a greater reduction in dose exposure with a smaller increase in the current of the X-ray tube is expected.

KEYWORDS: filtered X-rays, energy information, contrast media, X-ray transmission measurement, energy subtraction method, sensitivity, dose exposure

I. Introduction

X-ray transmission measurement is one of the most useful methods for the detection of tumors in the human body. X-ray computed tomography (CT) is an extension of the X-ray transmission method. Both in X-ray transmission measurements and CT measurements, images are given as the number of transmitted X-rays, which is reduced from the number of incident X-rays due to the absorption coefficients of tissue and bone in the subject. Identifying tumors, such as cancers, by X-rays, however, is not a simple job: the absorption coefficient of tumors is not very different from normal tissue. The resulting black and white X-ray photograph, *i.e.*, radiograph, often does show any significant difference between normal tissue and tumors. To overcome this situation, many attempts to utilize the energy information of incident X-rays, *i.e.*, energy subtraction methods, have been reported.

The energy subtraction method in off-line measurements was carried out by two photographic films with an absorber between them for bone minerals and chest radiography.[1–4) The first film had information for all of the incident X-rays passing through the subject. The second film was sensitive only to those X-rays with sufficient energy to pass through the absorber between the two photographic films. By comparing the first and second radiographs, tissues and bones with different X-ray absorption coefficients could be identified.

These days, digital radiographs can be easily taken. The energy subtraction method is performed on-line with computers. The most idealized energy subtraction method can be carried out with two monochromatic X-rays: with a contrast media such as iodine and two monochromatic X-rays with energies E_1, E_2, with the the K-edge energy of iodine, E_K, between them ($E_1 < E_K < E_2$). Two radiographs of the subject with an iodine contrast media are taken. Tumors, which have a higher concentration of iodine contrast media, absorb the X-rays with energy E_2 more than normal tissue, are clearly detected by subtracting the digital radiographs each other.[5) This ideal method, however, is not a practical one from two points of view that: (1) monochromatic X-rays are available in synchrotron facilities, but not in hospitals, and (2) changing the energies of monochromatic X-rays is time consuming and allows tumors in a subject to change their positions.

One practical example of the digital energy subtraction method is employed for the measurement of bone mineral density.[6,7) In this method, white X-rays are transformed by filters into two quasi-monochromatic X-rays. Using semiconductor detectors with two single channel analyzers, the number of X-rays with higher and lower energies, which were transmitted through a subject, could be measured. The combination of the two digital radiographs gave the density of the bone minerals.

In this paper, we propose a method for measuring the thickness of the contrast media, such as iodine, in a subject with less exposure. Our future goal is to detect tumors and varices in periodic health checks with a smaller amount of contrast media and lower dose exposure: These days the quantity of dose exposure in medical treatment is not counted as the dose exposure of individuals. However, it should be taken into account as the personal record of accumulated dose exposure in the future and should be reduced as much as possible. Effort expended on the reduction of dose exposure both in medical treatment and in medical check-ups will increase in importance in terms of individual and public dose exposure, because the opportunities for X-ray exposure are increasing rapidly.

For the purpose described above, we employ filtered X-rays instead of white X-rays, semiconductor detectors as X-ray detectors, and the energy subtraction method to utilize energy information. Hereafter, we call this method, FIX-ES, for convenience. Dose exposure as a result of X-rays with low energy, which cannot penetrate the human body and which cannot give information about the inside of the body, should be eliminated by X-ray filters. Semiconductor detectors allow measurements of X-rays with higher quantum efficiency than

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conventional scintillator and photo-multiplier systems, and they provide energy information about the X-rays. This energy information is applied to the energy subtraction method.

Comparing FIX-ES to the off-line energy subtraction methods described in Refs. 1–4), FIX-ES gives less dose exposure and can be applied to *in-situ* observation. The energy subtraction method with monochromatic X-rays can be performed only in large synchrotron facilities and allows movement of tumor between exposures, whereas FIX-ES diagnosis will be taken place in any hospitals with X-ray tubes and requires only single exposure of X-rays.

Here we discuss the sensitivity of FIX-ES to the contrast media and the possibility of reducing the quantity of dose exposure by simple experiments and simulation calculations based on the experiments. Our next move will be to take digital radiographs of subjects and discuss them.

II. Experiments

1. Experimental Setup

A schematic drawing of our experimental setup is shown in **Fig. 1**. The X-ray tube and X-ray generator employed were H7131 (Hitachi Medical Co.) and RXB-100 (TOREC Co. Ltd.), respectively. The maximum current of this X-ray generator was 5 mA. A lead slit (1 mm×4 mm) was placed at the window of the X-ray tube. After passing the slit, the X-rays were attenuated by an Al filter of 2 mm in thickness and a La filter made of La_2O_3 powder sandwiched between acryl boards (4 mm in total thickness) with various thicknesses over an area of 3 cm×3 cm. The X-rays were collimated by Pb and Al plates with a hole of 0.5 mm in diameter, as shown in Fig. 1. As a detector for X-rays, a Ge detector (IGP105185, Princeton Gamma-Tech) was used with a pulsed optical feedback type preamplifier (PO-15B, Princeton Gamma-Tech). The Ge detector was wrapped with a Pb shield to avoid room-return X-rays. The energy spectra were measured with a multi-channel analyzer with a main amplifier (System 8000, Princeton Gamma-Tech).

To simulate a human body, a water layer 20 cm thick was inserted between the Al, La filters and the Ge detector as a phantom. In this water layer, the layer of iodine contrast media was defined by an acrylic box 2 cm thick, filled with water and a suitable amount of iodine tincture (60 g of iodine, 40 g of KI in 1,000 m*l*) to obtain the required thickness of the iodine layer. The distance between the X-ray tube and the Ge detector was set at 3 m for the measurements of the energy spectra of white and filtered X-rays without the phantom. When the phantom was inserted, the distance was reduced to 1.5 m, because of the attenuated intensity of X-rays.

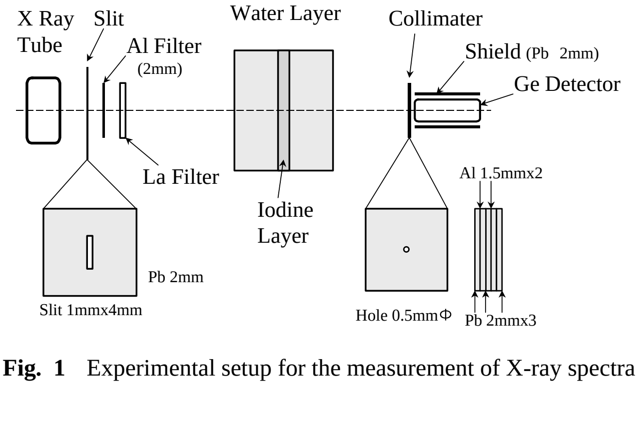

Fig. 1 Experimental setup for the measurement of X-ray spectra

The intensity of X-rays emitted from the target of an X-ray tube is described by the following equation,[8)]

$$I(E)dE = cZ(E_0 - E)dE. \tag{1}$$

Here, $I(E)$ is the intensity of X-rays with energy E, c is a constant, Z is the atomic number of the target material of the X-ray tube and E_0 the tube voltage. When absorbers, such as an Al filter with 2 mm in thickness, as indicated by the Drugs, Cosmetics and Medical Instruments Acts, are inserted, the energy spectrum of X-rays is given by,

$$\varphi(E)dE = I(E)\exp\left\{-\sum_i \mu_i(E)x_i\right\}dE. \tag{2}$$

Here, $\varphi(E)$ is the intensity of the transmitted X-rays with energy E, $\mu_i(E)$ and x_i are the absorption coefficient as a function of energy and the thickness of i-th material, respectively.

Examples of measured X-ray spectra of white and filtered X-rays are shown in **Fig. 2** by solid lines. The dotted lines in Fig. 2 are the results of calculations using Eqs. (1) and (2). The experimental and theoretical results agree fairly well. The higher counts in the low energy parts are due to room-return X-rays and background noise.

2. Energy Spectrum of Filtered X-rays with Iodine Contrast Media

A typical X-ray spectrum with an iodine contrast media 100 μm thick is shown in **Fig. 3**. The K-edges of La and I are clearly seen at 38.9 keV and 33.2 keV. The dotted line in Fig. 3 is the calculation result using Eqs. (1) and (2). The agreement between the experimental and simulated X-ray spectra is excellent, except that the energy part is higher than the K-edge of La. We do not pay any significant attention to this disagreement, because the energy spectra below the energy of the La K-edge were used in our study. In Chap. III, we employed the simulated X-ray spectra for the estimating the thickness of the iodine contrast media, the sensitivity to

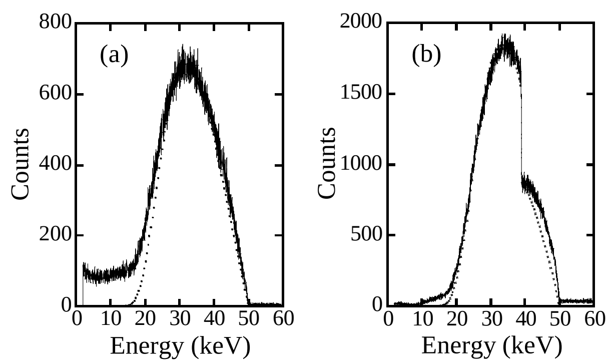

Fig. 2 Experimental (solid lines) and simulated (dotted lines) energy spectra for (a) white and (b) filtered X-rays

The voltage of the X-ray tube is 50 kV. The thickness of the La filter was 40 μm. An Al filter of 2 mm in thickness was used for (a) and (b).

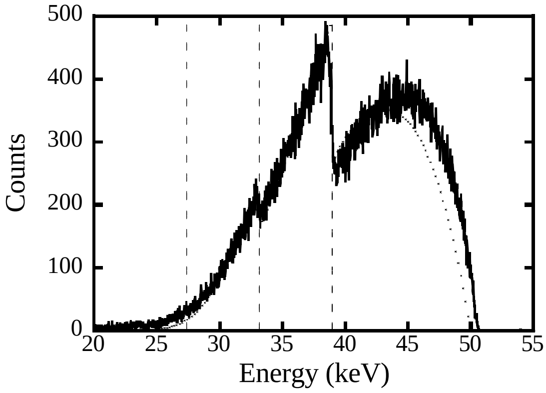

Fig. 3 Experimental (solid lines) and simulated (dotted lines) energy spectra for filtered X-rays with a phantom of a water layer (20 cm) and an iodine layer (100 μm)

Filters employed were Al with 2 mm thickness, and La with 40 μm thickness. The voltage of the X-ray tube was 50 kV. The dashed lines indicate the energies of 27.4 keV, 33.2 keV (K-edge of I), and 38.9 keV (K-edge of La).

the iodine contrast media, and the quantity of dose exposure with FIX-ES.

3. Estimating the Thickness of the Iodine Contrast Media in Tumors

Estimating the actual iodine thickness concentrated in a tumor is difficult due to the varying conditions of the tumors. To obtain a rough standard of the iodine thickness, however, we took the following stand point.[9)]

In medical treatments, typically 100 ml of iodine contrast media is injected into the human body. The amount of iodine in the contrast media is nearly 30 g/100 ml. On the other hand, the amount of blood in a human body is 5,000–6,000 ml. If the injected iodine contrast media is distributed uniformly throughout the blood, we would have 5 mg/ml of iodine in the blood. The other extreme assumption is that the iodine contrast media is diluted by the whole body, for example, with 60 kg in weight. Assuming the density of the human body is 1 g/cm^3, the density of iodine in this case is estimated to be 0.5 mg/ml. With the estimation described above and the density of iodine, 4.9 g/cm^3, the thickness of the iodine layer in a tumor of 1 cm in thickness is estimated to be 1–10 μm.

III. Analysis

1. Estimating the Thickness of Iodine Contrast Media by X-rays

In order to exploit energy information from X-ray spectra, we propose the use of energy integrals of measured X-ray events to reduce statistical errors. This method is similar to the one reported in Refs. 6) and 7), using two transmitted X-ray photon numbers with different energies E_A, E_B. We extend this method to the energy integrals of X-ray spectra.

In order to estimate the thickness of the iodine contrast media, it is necessary to have a relationship between the integrated energy spectra of the transmitted X-rays and the thickness of the contrast media. We estimated the partial integrals of the transmitted X-rays for the two energy ranges: ϕ_1 for 33.2–38.9 keV, and ϕ_2 for 27.4–33.2 keV. Each energy range has an energy of 5.7 keV, which corresponds to the energy difference of the K-edges of La and I. An estimation was carried out for the La filter 0 μm thick (white X-ray) to 700 μm as a function of the thickness of iodine contrast media. An example is shown for ϕ_1 in **Fig. 4**. The estimated results indicate that the integrated events fit excellently by the relationship,

$$\begin{pmatrix} a_{11} & a_{12} \\ a_{21} & a_{22} \end{pmatrix} \begin{pmatrix} x_1 \\ x_2 \end{pmatrix} = \begin{pmatrix} \ln(\phi_{10}/\phi_1) - C_1 \\ \ln(\phi_{20}/\phi_2) - C_2 \end{pmatrix}. \tag{3}$$

Here, a_{ij} $(i,\ j{=}1,\ 2)$ are the constants determined by the i-th integration range in an X-ray energy spectrum and the j-th material (for example, iodine for $j{=}1$, and water for $j{=}2$), x_j the thickness of the j-th unknown material in the subject, C_i the contribution of known materials such as the La filter in the i-th integration range, ϕ_i the integrated results of the energy spectrum of filtered X-rays in the i-th integration range, and ϕ_{i0} the integrated results of the X-rays without a phantom. For the measurement of ϕ_{i0}, either white or filtered X-rays can be used.

By solving Eq. (3), we obtain the thickness of the iodine contrast media, x_I, for the phantom consisting of water and iodine as,

$$x_I = a\left\{\ln\left(\frac{\phi_2}{\phi_{20}}\right) - \frac{\ln\dfrac{\phi_{2W}}{\phi_{20}}}{\ln\dfrac{\phi_{1W}}{\phi_{10}}}\ln\left(\frac{\phi_1}{\phi_{10}}\right) - C\right\}. \tag{4}$$

In Eq. (4), a is a constant due to the iodine and C is the contribution by materials other than the water and iodine. The values of ϕ_{1W} and ϕ_{2W} are the integrated events for the same energy ranges of ϕ_1 and ϕ_2 but for a water layer with a thickness t. The fraction $\ln(\phi_{2W}/\phi_{20})/\ln(\phi_{1W}/\phi_{10})$ has a value of nearly $4/3$, canceling the thickness of the water layer. The estimated thickness of the iodine contrast media is shown in **Fig. 5** as a function of the thickness of the iodine contrast media in the phantom for filtered X-rays of La 100 μm thick. Although the estimated thickness of the iodine contrast media is shifted nearly 4 μm, the inclination agrees excellently with

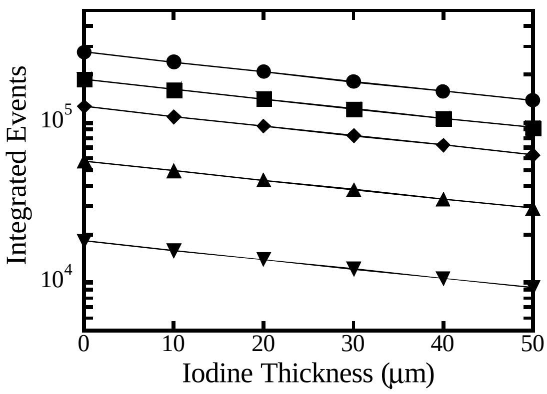

Fig. 4 Example of integrated events ϕ_1 for the energy range of 33.2–38.9 keV of filtered X-ray spectra as a function of the thickness of iodine layer

The thicknesses of La filters are 0 μm, *i.e.*, white X-rays (●), 100 μm (■), 200 μm (◆), 400 μm (▲), and 700 μm (▼).

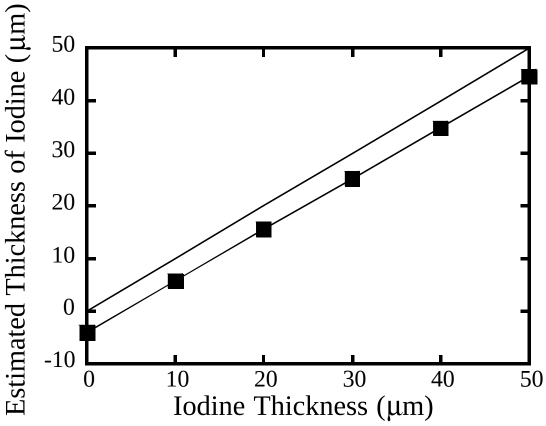

Fig. 5 Estimated thickness of iodine layer by FIX-ES as a function of iodine thickness

The line with squares is the estimation. The line without symbols is the reference.

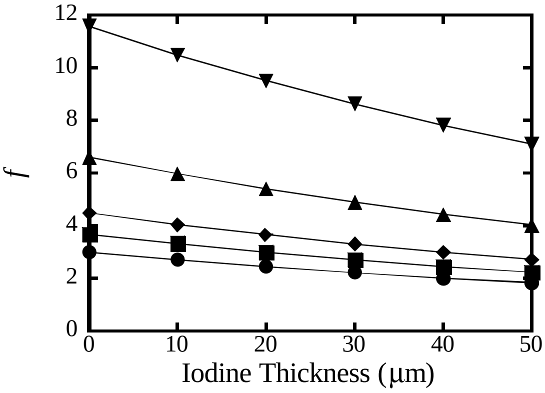

Fig. 6 Sensitivity of FIX-ES to the thickness of iodine

The thicknesses of the La filters are $0\,\mu$m, *i.e.*, white X-rays (●), $100\,\mu$m (■), $200\,\mu$m (◆), $400\,\mu$m (▲), and $700\,\mu$m (▼).

the one of the actual iodine thickness. The thickness of the La filter has very little influence on the proportionality of the result of FIX-ES, but on the offset of the result.

In the case where there are more than two different materials in a subject, such as tissue, bone and contrast media, we simply have to define three integrating ranges: by solving an equation similar to Eq. (3), but with a 3×3 matrix, the thickness of the contrast media is obtained in the same way.

2. Sensitivity to the Iodine Contrast Media Thickness

In order to reduce the charge to patients, the quantity of iodine contrast media injected into them should be decreased as much as possible. In this section, the sensitivity of FIX-ES to the thickness of the iodine contrast media is estimated. In the case of X-ray radiographs, the iodine contrast media is shown as a smaller number of transmitted X-rays. Here, we employ the estimation formula f such that,

$$f \equiv \frac{\phi_1}{\phi_2}. \tag{5}$$

The function f is only an example of the sensitivity estimation, and many other estimating functions could be proposed, such as, $(\phi_1-\phi_2)/\phi_2$. Equation (5) is chosen for its simplicity.

In this formula, f becomes smaller with thicker contrast media. The change of f, *i.e.*, the sensitivity of FIX-ES to the thickness of the contrast media, as a function of the iodine thickness and the thickness of the La filter is shown in **Fig. 6**. Compared to the result with white X-rays (La thickness, $0\,\mu$m), f has a higher value as the thickness of the La filter increases. Changes due to the iodine thickness also become steeper with thicker La filters, especially in the thin iodine region.

3. Estimation of X-ray Dose Exposure

In this section, we would like to compare the relative quantity of X-ray dose exposure of filtered X-rays to white X-rays. As a condition for the estimation, we define the statistical error of the function f as g where,

$$g = \left|\frac{df}{f}\right| = \sqrt{\left(\frac{d\phi_1}{\phi_1}\right)^2 + \left(\frac{d\phi_2}{\phi_2}\right)^2}. \tag{6}$$

The errors, $d\phi_1$ and $d\phi_2$, are given as the statistical errors. They are expressed as the square root of ϕ_1 and ϕ_2, respectively. We have a simple equation for estimating the condition of relative dose exposure as,

$$g = \sqrt{\frac{1}{\phi_1} + \frac{1}{\phi_2}}. \tag{7}$$

The function g means the quality of the radiograph images, which corresponds to the statistical error of the number of measured electric charges, such as electrons and holes in semiconductor detectors and electrons in scintillator and photo-multiplier systems.

The comparison of dose exposures is performed with the condition that,

$$g|_{\text{filtered}} = g|_{\text{white}}, \tag{8}$$

where $g|_{\text{filtered}}$ and $g|_{\text{white}}$ stand for the errors of f in cases of filtered and white X-rays. To achieve this condition, the tube current of the filtered X-rays was adjusted.

Several methods for estimating the quantity of dose exposure have been proposed. In this stage of our study, we estimate the quantity of dose exposure as the ratio of dose exposure of filtered X-rays to that of white X-rays. The employed estimating formula for dose exposure is given by,

$$X = \int_0^{E_0} \varphi(E) \frac{\mu_w}{\rho} dE. \tag{9}$$

Here, $\varphi(E)$ is the intensity of incident X-ray, which is transmitted through the Al and La filters, as given in Eq. (2), to the subject, and μ_w/ρ is the mass energy-absorption coefficient of water. In **Fig. 7**, the energy spectra, (a) before and (b) after passing through the phantom of 20 cm of water layer, of white and filtered X-rays are shown. The estimation function g has the condition written in Eq. (8) in Fig. 7(b). By changing the thickness of the La filter, the tube current which satisfies the condition of Eq. (8) is determined, and the amount of dose exposure X is calculated from Eq. (9) with the obtained tube current. In **Fig. 8**, relative dose exposures are shown as func-

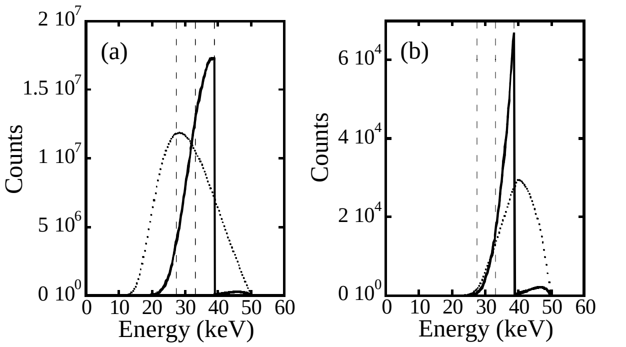

Fig. 7 Calculated energy spectra for white (dotted lines) and filtered (solid lines) X-rays (a) before and (b) after passing through the water phantom without an iodine layer

The thickness of the La filter was 400 μm. The values of the function g of white and filtered X-rays in (b) are the same. Dashed lines indicate energies of 27.4 keV, 33.2 keV (K-edge of I), and 38.9 keV (K-edge of La).

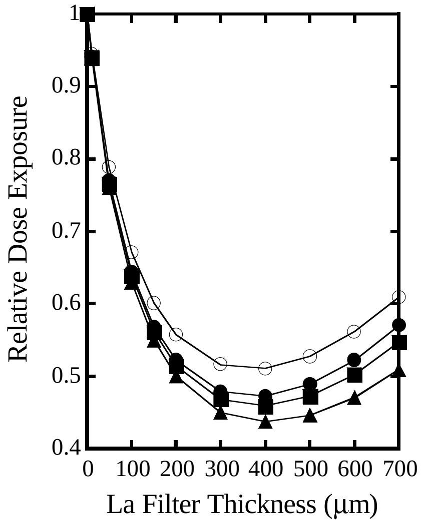

Fig. 8 Estimated quantity of dose exposure as a function of La filter thickness

The thickness of the iodine layers are 0 μm (●), 30 μm (■), and 50 μm (▲) with a tube voltage of 50 kV. Open circles show the case of no iodine layer with a tube voltage of 100 kV. The dose exposures are normalized to the one of white X-rays (no La filter), in each case.

tions of thicknesses of La filter for tube voltages of 50 kV (●) and 100 kV (◯) without the iodine contrast media. The same calculation procedures were repeated for a tube voltage of 50 kV with the thickness of iodine contrast media of 30 μm (■) and 50 μm (▲). The amount of dose exposure in each thickness of iodine contrast media were normalized to the one with a La filter thickness of 0 μm, *i.e.*, white X-rays.

The relative dose exposure has the minimum value at a La filter thickness of nearly 400 μm, and is less than 50% of that of white X-rays in case of no iodine contrast media and with a tube voltage of 50 kV. The minimum value decreases as the thickness of iodine contrast media increases.

For filters other than La, the estimated relative exposure doses are shown in **Fig. 9** in the case of no iodine contrast media. As the atomic number decreases, the minimum relative dose exposure has less value. In **Fig. 10**, tube currents for the cases of filter thickness of materials in Fig. 9 are shown.

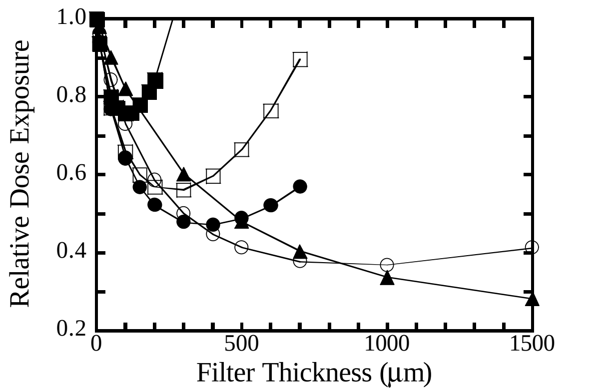

Fig. 9 Estimated quantity of exposure dose as a function of thickness for Sm (■), Ce (□), La (●), Ba (◯) and Cs (▲) filters

These relative dose exposures are estimated for the cases of no iodine layer and tube voltage 50 kV.

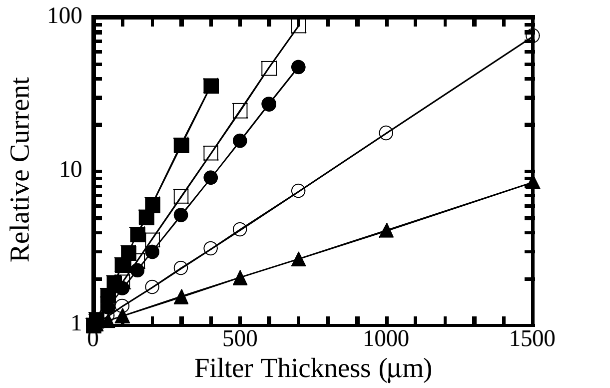

Fig. 10 The X-ray tube currents to achieve the results of Fig. 9 for the filters of Sm (■), Ce (□), La (●), Ba (◯) and Cs (▲)

The tube currents are normalized to that of white X-rays.

IV. Discussion

1. Estimation Methods on FIX-ES

The thickness of the water layer of the phantom employed in the experiments was 20 cm. The usual thickness of the water layer for simulating the human body is 25 cm to 30 cm. Our phantom thickness was chosen taking into account the small current of our X-ray tube. Because experimental results were used to prove the validity of the simulation calculation, the thinner water layer did not affect our results.

In determining the thickness of the contrast media, we are able to measure both ϕ_{10} and ϕ_{20} before exposing the subject to X-rays, *i.e.*, the subject was exposed only once. This enabled us to reduce the quantity of dose exposure and to avoid the movement of tumors which causes unclear observed radiographs.

In actual diagnosis, medical doctors judge tumors from the change in the density of black dots in X-ray radiographs. In this paper, we have carried out an analytical estimation of the sensitivity and we have not been able to make black and white images at this stage. In the estimation using the function f, the thickness of the contrast media reflected in the numerator: the thicker the contrast media the smaller the value of f. Visual images of f and its error g are shown in **Fig. 11**. In the case of paying attention to a certain point in a radiograph, a large error g makes the blackness of the point obscure, even in the case of high blackness, as shown in Fig. 11(a), and sometimes makes high blackness look like low blackness, as shown in Fig. 11(c). Even if the value of f, *i.e.*, the blackness of the point, is small, the smaller error makes the tumor image clearer. This small error corresponds to radiographs taken with a greater quantity of X-rays. In light of the discussion above, we think that estimating the quality of images with the function g is very similar to judging tumors by the density of black dots in radiographs.

In estimating the quantity of X-ray dose exposure, the current of the filtered X-rays was adjusted to have the relationship of Eq. (8) on g. The current of the filtered X-rays for this condition is determined uniquely: the energy spectra of the transmitted filtered X-rays with the same thickness of filter and the same contrast media, and the number of events is proportional to the current of the X-ray tube. As a result, the tube current of the filtered X-rays is determined uniquely to make g have the same value as with the white X-rays.

The reason that the relative dose exposure has a minimum value is explained as follows. The error g is taken as a kind of statistical error and is proportional to the inverse square root of the tube current. On the other hand, the quantity of dose exposure is proportional to the tube current. These two components work in opposite ways in the evaluation of the relative dose with the same quality of radiograph images as white X-rays and give a minimum point.

2. Practicability

This paper described a feasibility study of filtered X-rays as a method for achieving less contrast media and X-ray dose exposure: we have neglected the performance of conventional X-ray tubes and semiconductor detectors. For example, the X-ray tubes in practical transmission diagnosis are operated at nearly 800–900 mA, a current which is almost the same as with the maximum performance of the X-ray tubes. On the other hand, as shown in Fig. 10, tube currents for filtered X-rays to have the relationship of Eq. (8) range some to some tens of that for white X-rays. A tube current of several amperes is not practical. In **Fig. 12**, a summary of Figs. 9 and 10 is shown. For the La filter, a minimum dose exposure of nearly 50% of white X-rays is obtained with a thickness of 400 μm and a corresponding current nine times as high as white X-rays. When we choose Cs as the filter material, we have 40% and 34% of dose exposure with a thickness of 700 and 1,000 μm, and a corresponding current of 2.7 and 4.2 times as high as white X-rays. These are the best parameters in our study, if we could have enough tube current.

We think, however, that taking informative X-ray radiographs with filtered X-rays is possible. The amount of X-rays equivalent to nearly one ampere of tube current is too much for identifying tumors. Reducing the current of X-ray tubes to a quarter of an ampere may not change much the quality of the resulting radiograph. Based on this condition for white X-rays, a Cs filter with 700 μm in thickness brings us the same quality of radiograph, and a 60% reduction in dose exposure. Studies of X-ray radiographs with less X-ray tube current will be performed as an extended theme of this research.

Another field of medical treatment with X-ray exposure, called interventional radiology (IVR), in which interventions are carried out by a catheter and observations by radiographs, has the problem of skin ulcers caused by the X-ray exposure. With the invention of a catheter with an iodine compound coating or catheters made of materials with high atomic numbers, our proposed method would be applicable for the observation of a catheter with less dose exposure.

V. Conclusion

To decrease the quantity of X-ray exposure in X-ray transmission measurements, it is important to reduce X-rays with low energy, which only contribute to dose exposure but not to information about tumors in a subject. To detect the contrast media, such as iodine, in tumors, the energy subtraction method is quite useful. With the use of filtered X-rays and the energy subtraction method, higher tube currents and energy information about the X-rays are indispensable. To

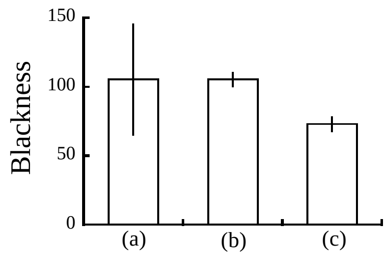

Fig. 11 Schematic drawings of the meanings of the function f and its error g

The function f corresponds to the “Blackness” as shown by squares. The “Blackness” is high when the contrast media is thin and the number of transmitted X-rays is large. (a) high blackness with large error, (b) high blackness with small error, and (c) low blackness with small error. Case (a) corresponds to the measurement with less X-rays. The measured point with image quality (a) is sometimes regarded as the image (c).

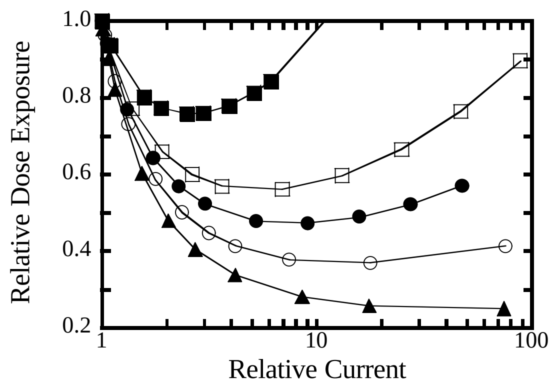

Fig. 12 Summary of Figs. 9 and 10

The symbols show the filters of Sm (■), Ce (□), La (●), Ba (◯) and Cs (▲).

meet the requirements described above, we have proposed the energy subtraction method by semiconductor detectors with filtered X-rays, based on the measurements of the energy spectra of La filtered X-rays. In the measurements, we do not always have to measure energy spectra with a multichannel analyzer, but to count X-rays in two energy ranges, such as 33.2–38.9 keV and 27.4–33.2 keV with two single-channel-analyzers and counters. With energy information about the filtered X-rays, the thickness of the contrast media is sensitively measured with less quantity of dose exposure, compared to white X-rays. Although a higher tube current is required, radiographs with the same quality will be taken by filtered X-rays with 50% or less dose exposure than with white X-rays.

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