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[45] **Dec. 10, 1974**

[54] **COMPENSATION FOR PATIENT THICKNESS VARIATIONS IN DIFFERENTIAL X-RAY TRANSMISSION IMAGING**

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[52] **U.S. Cl.** **250/402,** 250/510

[51] **Int. Cl.** **G03b 41/16**

[58] **Field of Search** 250/510, 402, 320, 321, 250/322, 323

[56] **References Cited**

UNITED STATES PATENTS

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Primary Examiner—James W. Lawrence
Assistant Examiner—C. E. Church
Attorney, Agent, or Firm—Burmeister, Palmatier & Hamby

[57] **ABSTRACT**

Differential x-ray images can be produced by subtracting two different x-ray images which are produced by using two different quasi-monoenergetic x-ray spectra. Two different x-ray filters may be employed alternately to produce such spectra. For example, two different filters containing iodine and cerium may be employed alternately to produce two different x-ray spectra having peaks at different energies. The x-rays to be filtered may be derived from an ordinary x-ray tube which produces a continuous spectrum of x-rays over a wide band of energies. The purpose of producing the differential x-ray images is to subtract or cancel out the portions of the x-ray images which are due to the ordinary tissues of the patient, particularly the soft tissues, so that the presence of certain contrast substances in the patient's body will be emphasized or enhanced. Such contrast substances include iodine, xenon or barium, introduced into the bloodstream, the lungs or the food canal of the patient. The ordinary soft tissues of the patient transmit x-rays of different energies to different extents. While the two different x-ray images can be balanced for any particular thickness of the patient, variations in such thickness over the field of view tend to upset such balance. Thus, without compensation for patient thickness variations, cancellation of the x-ray images due to the ordinary tissues of the patient can be achieved at only one value of patient thickness. In accordance with the present invention, such patient thickness compensation is achieved by adjusting the composition and density of the two x-ray filters, and by adjusting the high voltage supplied to the x-ray tube, so that the ratio of the two different x-ray images produced by soft tissues remains nearly constant over a wide range of patient thickness. Thus, substantial cancellation of the soft tissue images can be achieved over a wide range of variations in the patient thickness. The x-ray filtration is adjusted so that one of the x-ray filters produces a spectrum of transmitted x-rays having two peaks at energies below and above the energy of the peak produced by the other x-ray filter. Thus, the sum of the images produced by the two peaks tends to remain in a constant relationship to the image produced by the other spectrum, despite variations in patient thickness. The supply voltage to the x-ray tube is varied as an inverse function of the average patient thickness. Thus, the voltage is reduced when the average patient thickness increases. The adjustment of the supply voltage to the x-ray tube changes the two different quasi-monoenergetic x-ray spectra so as to optimize the compensation for patient thickness variations.

30 Claims, 12 Drawing Figures

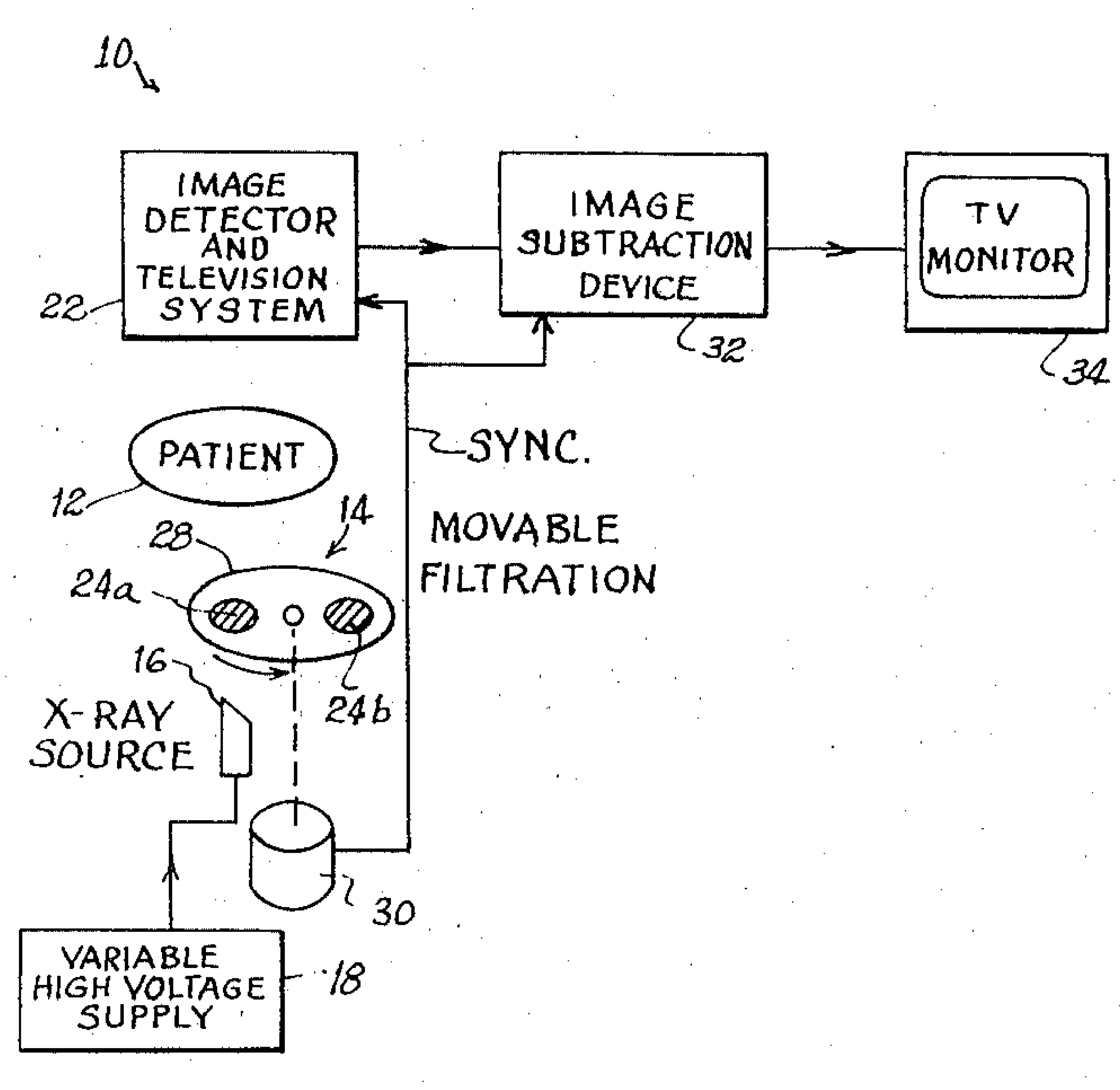


FIG. 1

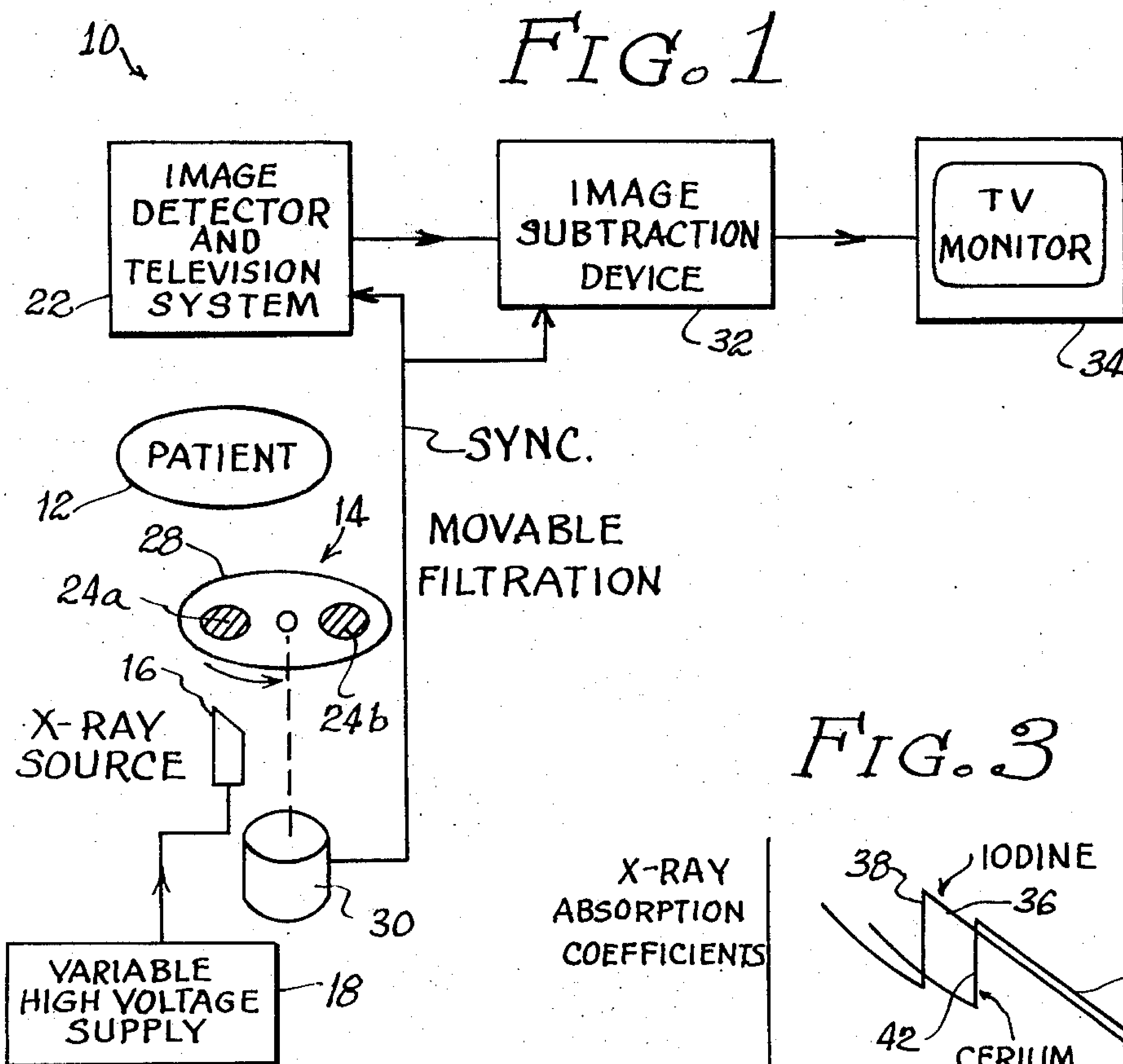
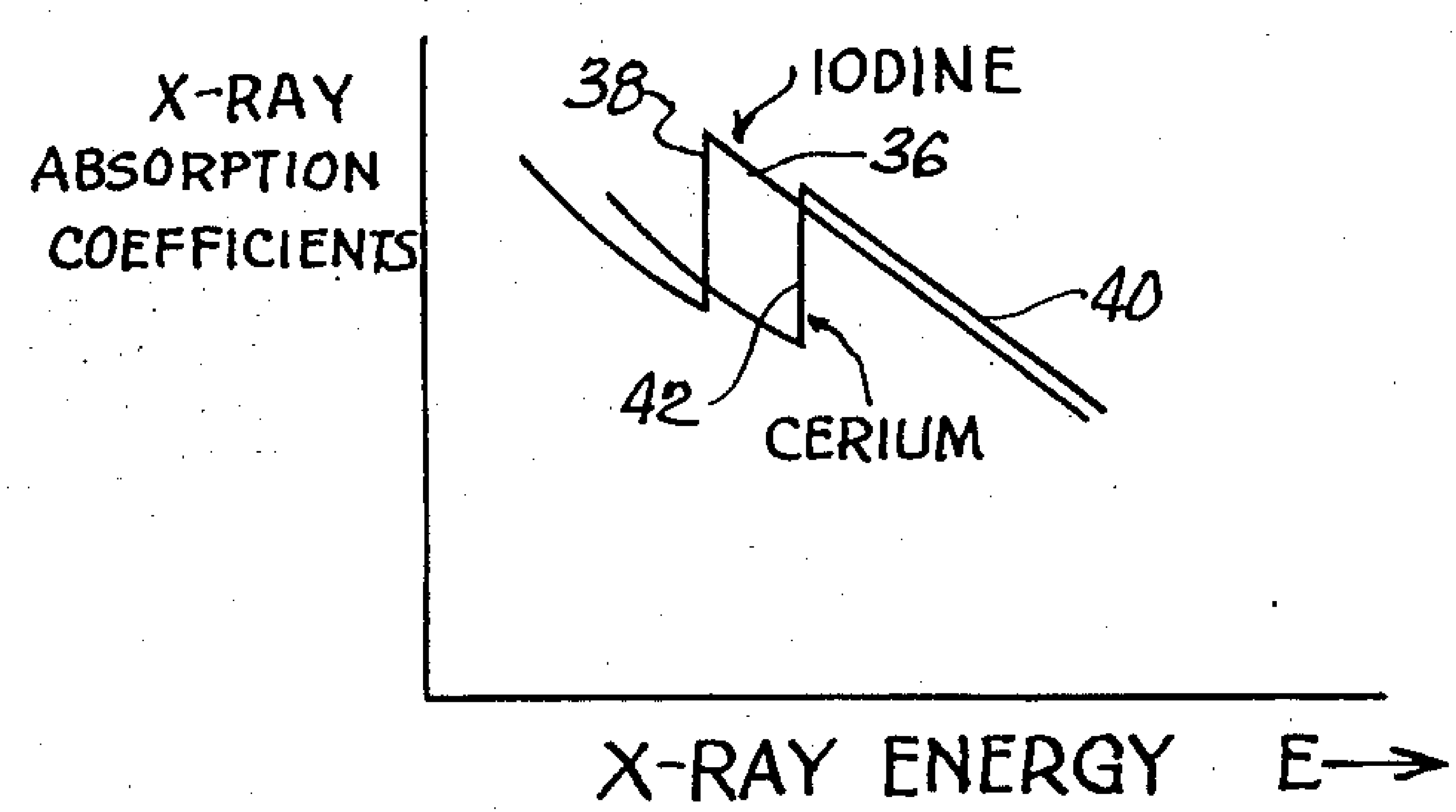


FIG. 3



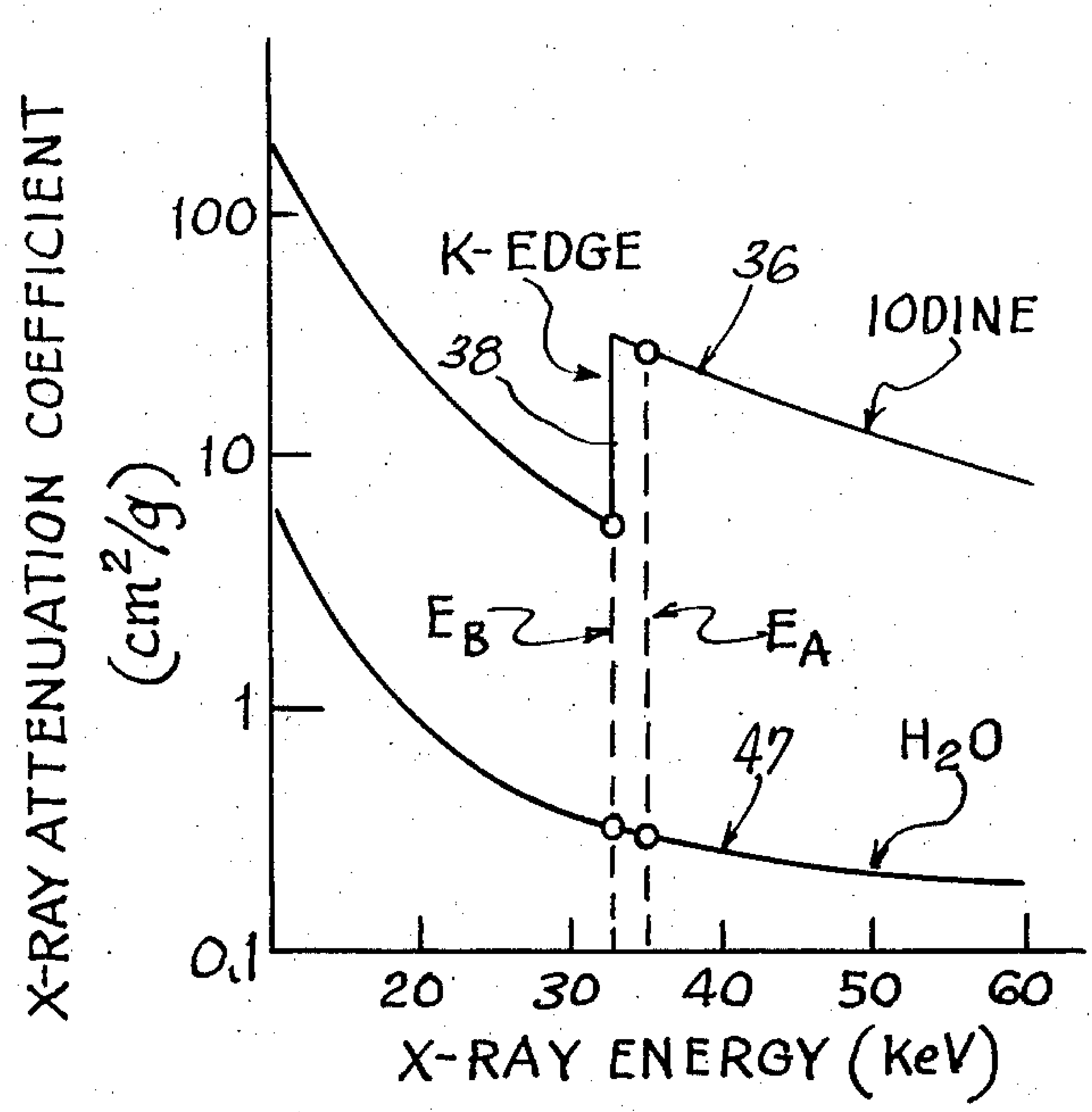


FIG. 2

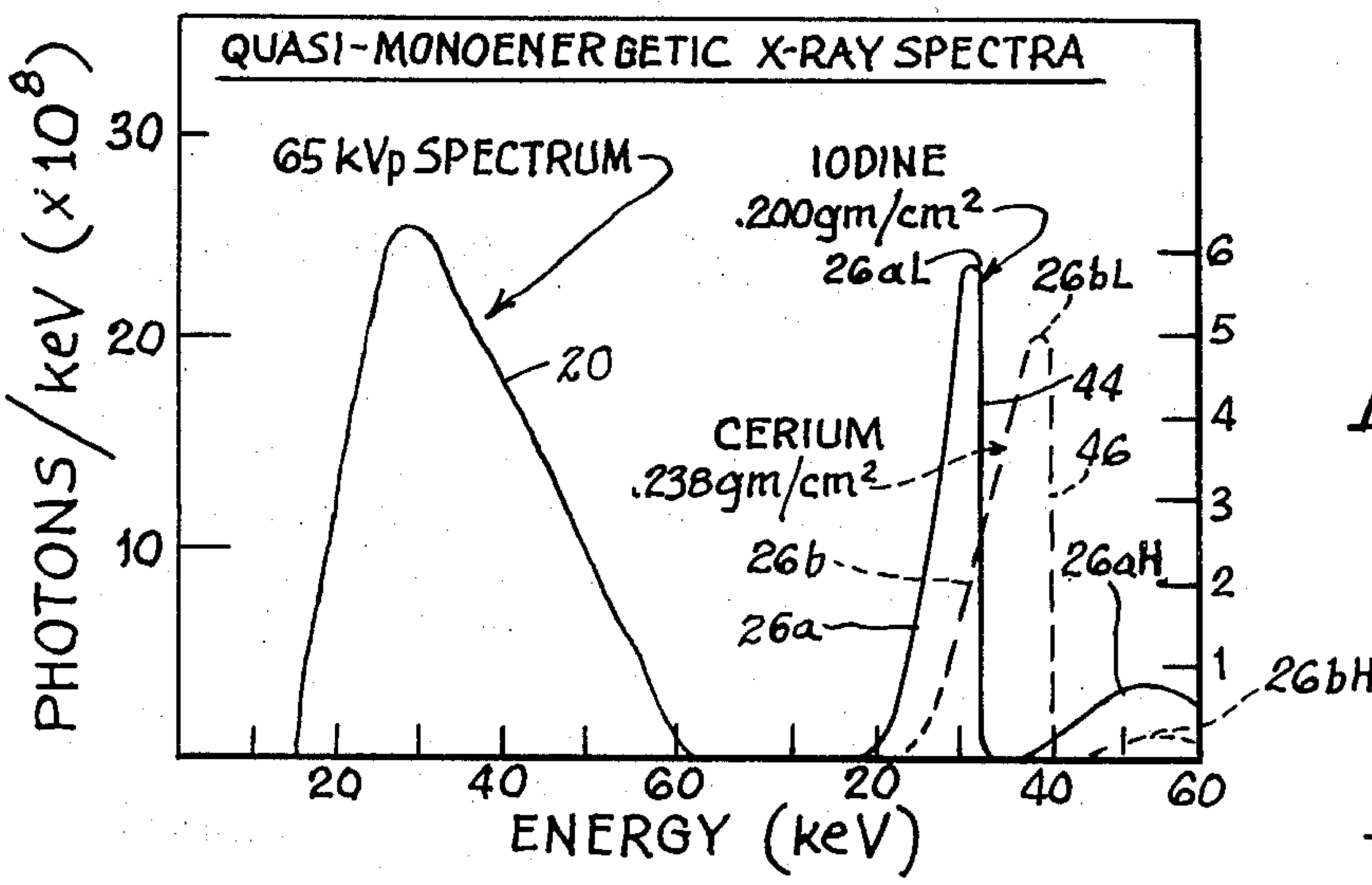


FIG. 4

FIG. 5

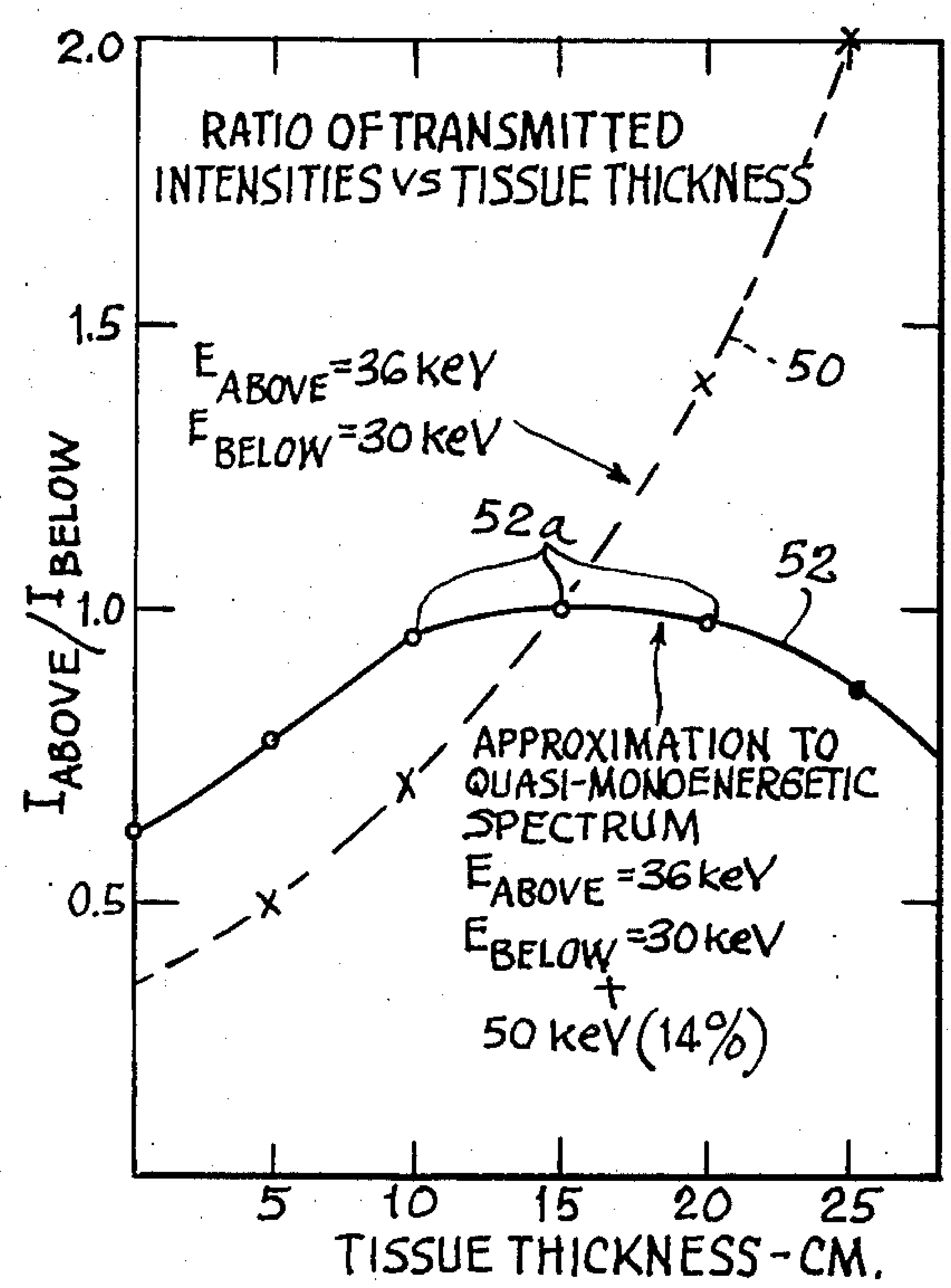
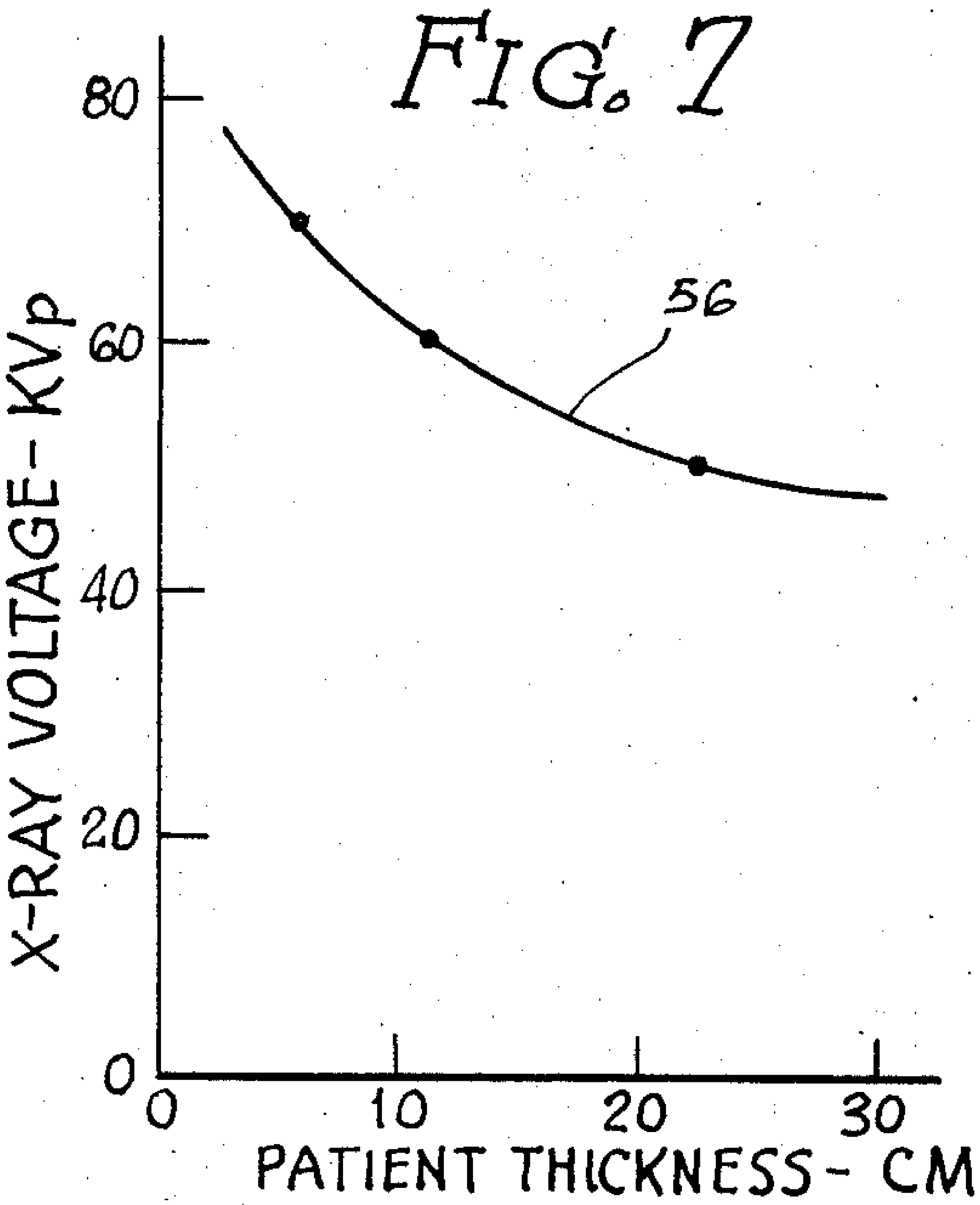


FIG. 7



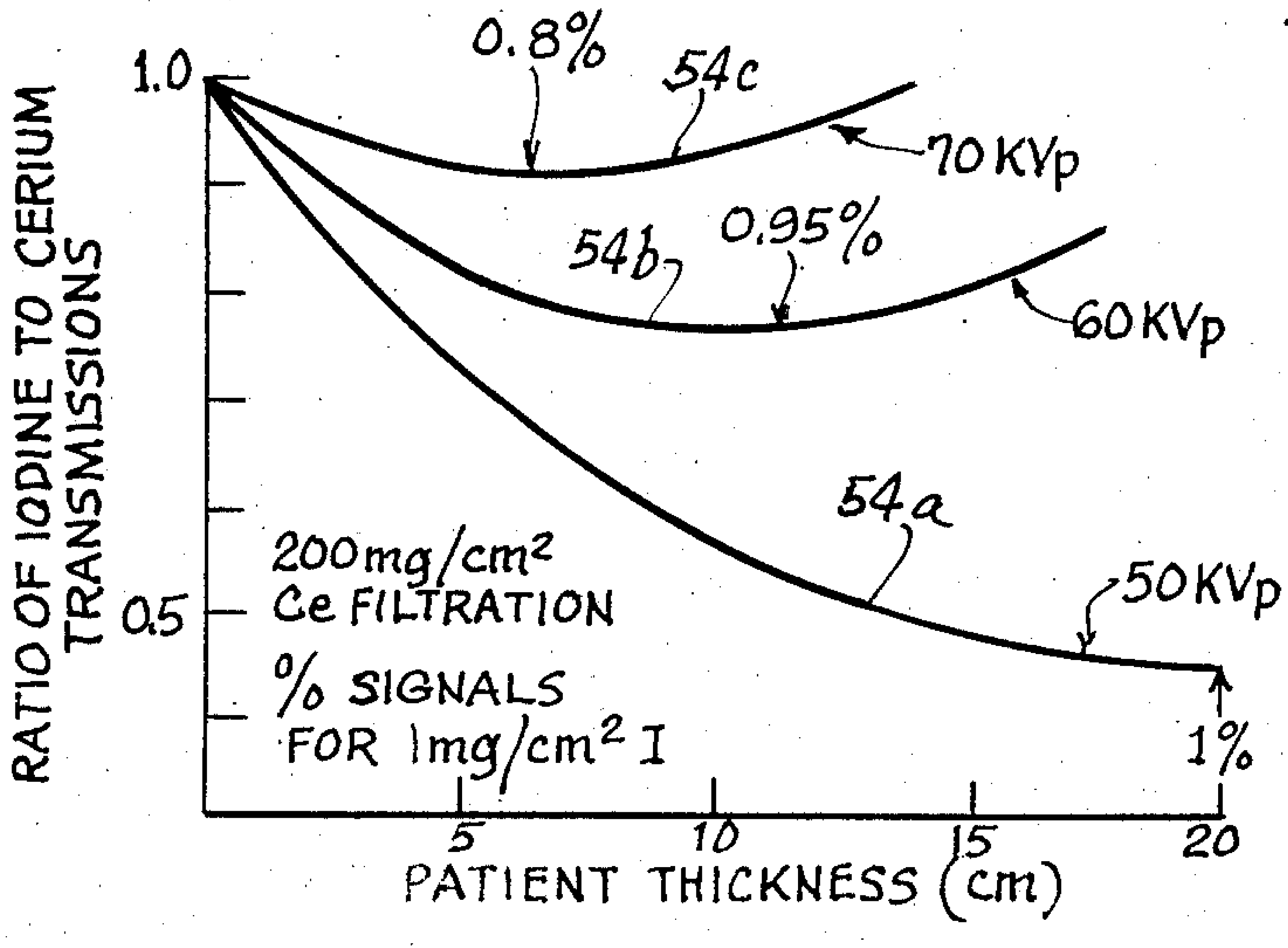


FIG. 6

FIG. 8

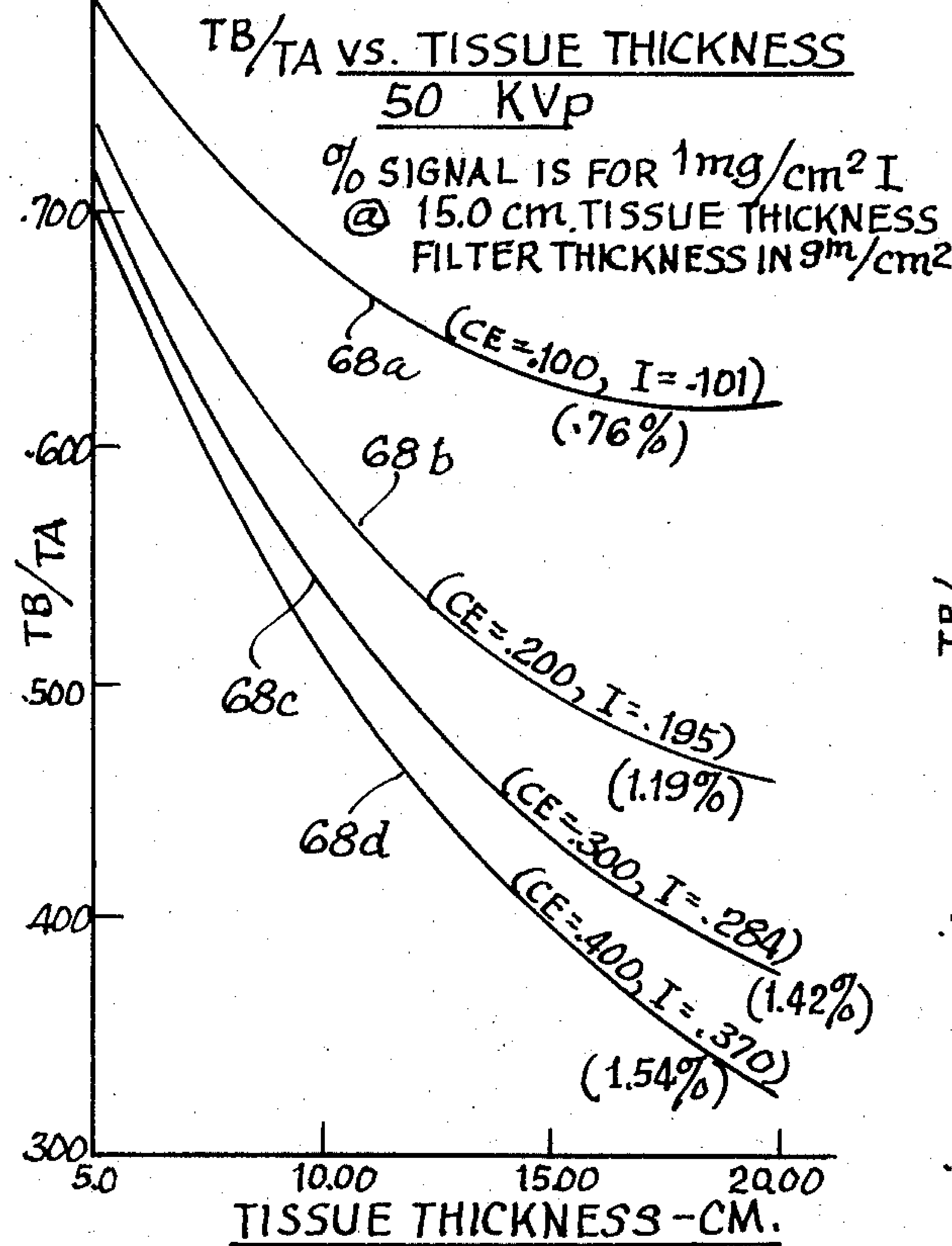


FIG. 9

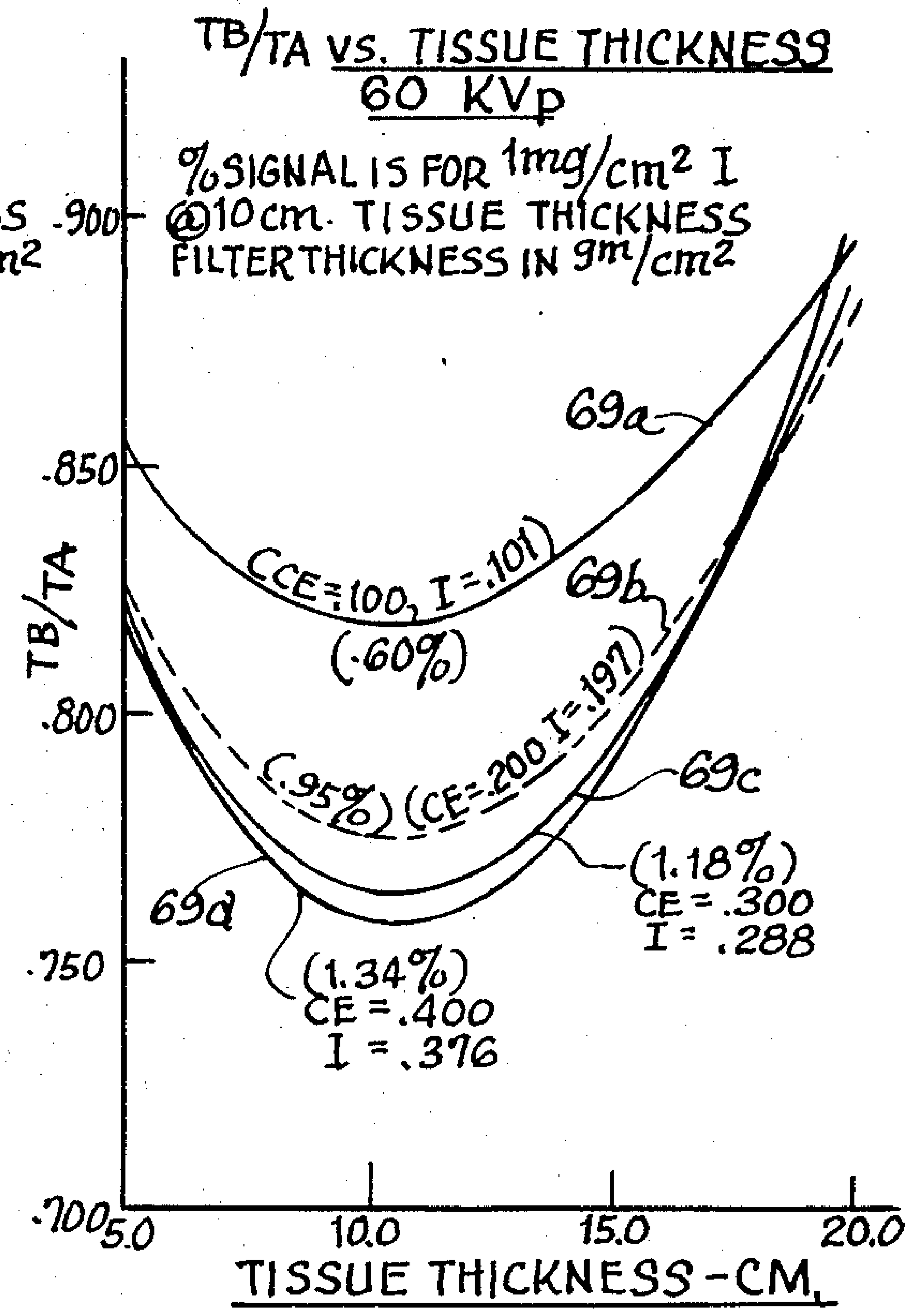
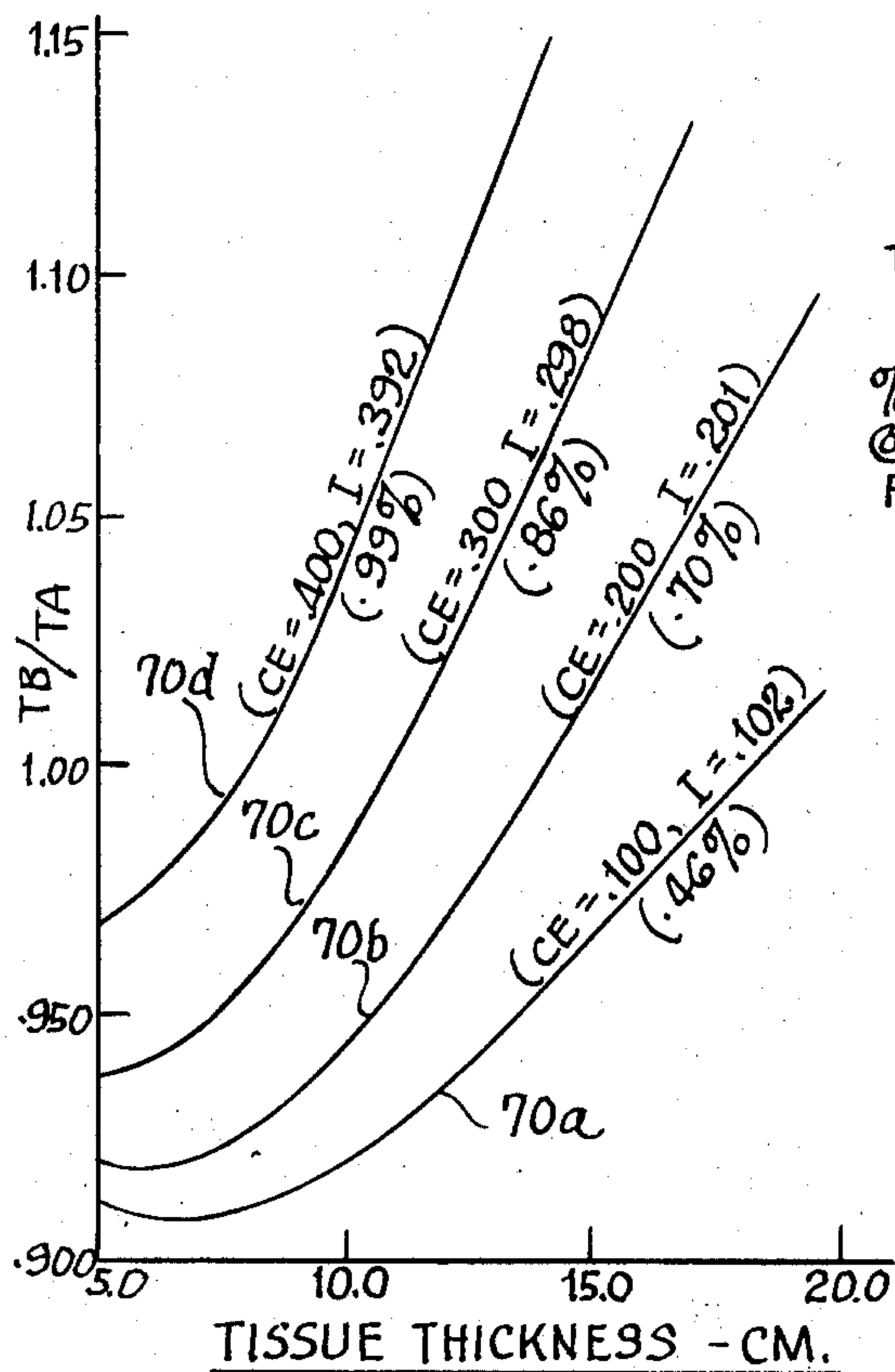


FIG. 10



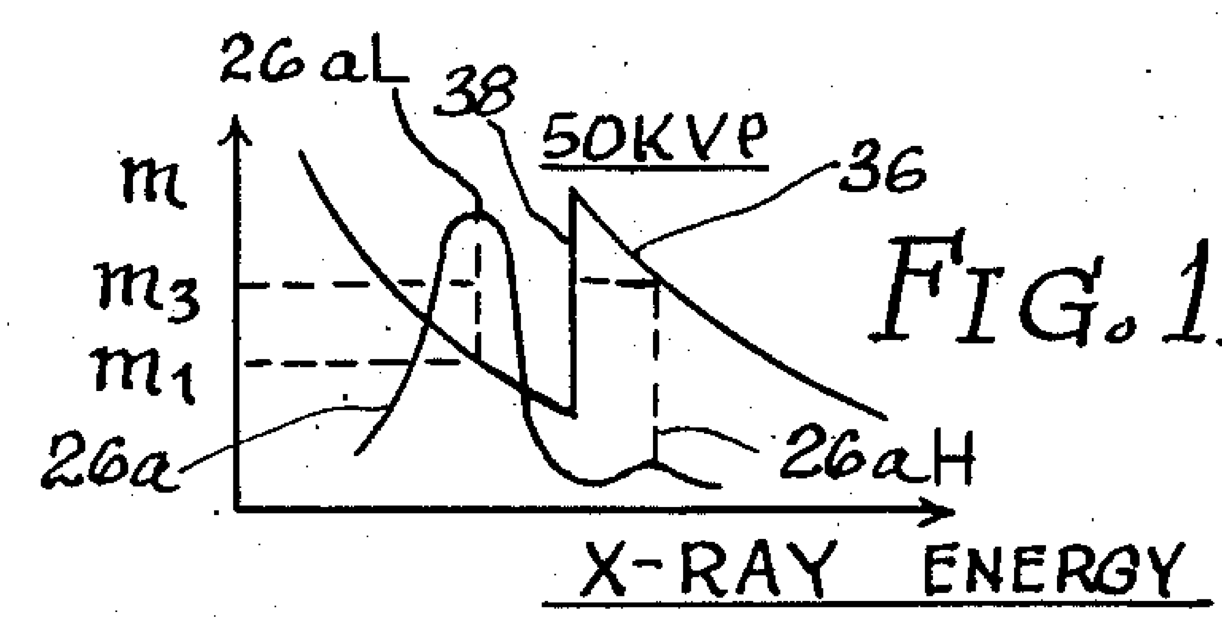


FIG. 11

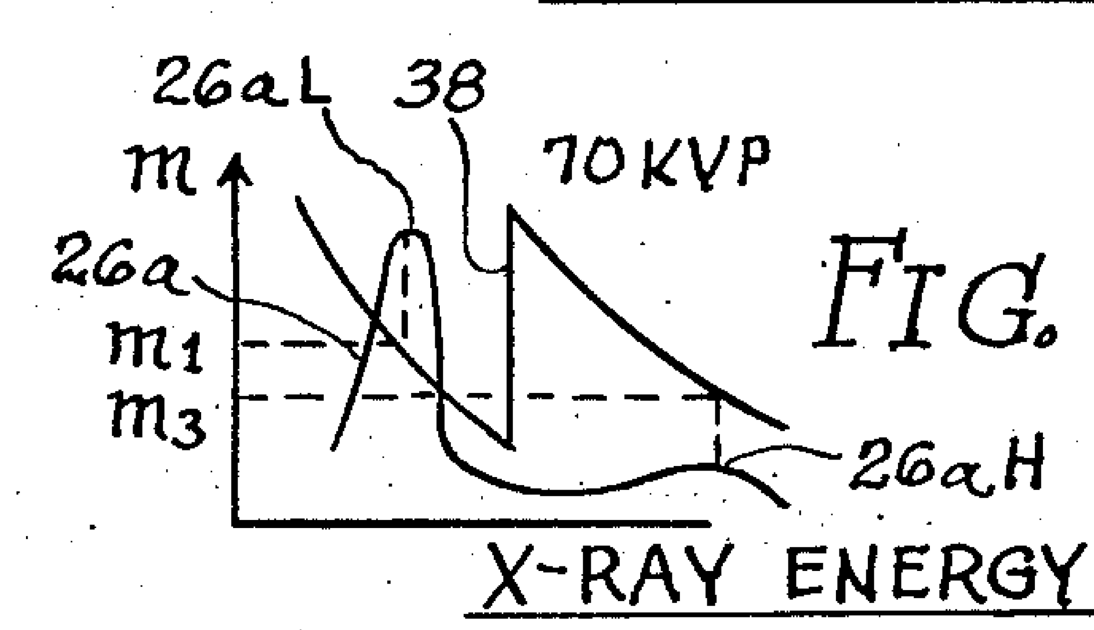


FIG. 12

COMPENSATION FOR PATIENT THICKNESS VARIATIONS IN DIFFERENTIAL X-RAY TRANSMISSION IMAGING

This invention relates to improvements in differential x-ray imaging techniques. A differential x-ray image can be produced by deriving the difference between two different x-ray images, produced under different conditions as to x-ray energy, or as to other factors. For example, the two different x-ray images can be produced by using two different monoenergetic or quasi-monoenergetic x-ray spectra. Two different monoenergetic x-ray beams at different energies will produce different x-ray images, because the two different beams will be transmitted and absorbed differently by the tissues and other substances in the patient's body. By subtracting the two images, a differential x-ray image can be produced which will often emphasize or enhance the visibility of particular tissues or substances in the patient's body. Thus, for example, the visibility of x-ray contrast materials can be enhanced. Examples of such materials are iodine, xenon and barium, which can be introduced into the bloodstream, the lungs or the food canal of the patient.

Instead of using two different monoenergetic x-ray beams, it is generally more convenient to employ two different quasi-monoenergetic x-ray spectra, which may have peaks at different energy levels. The two different x-ray spectra will produce different x-ray images which can be subtracted to produce a differential x-ray image. By using this technique, the visibility of contrast media or particular tissues can often be enhanced.

The two different quasi-monoenergetic x-ray spectra can be produced by using two different x-ray filters alternately. The x-rays to be filtered may be derived from an ordinary x-ray tube which produces a continuous spectrum of x-rays over a wide band of energies. The filters produce selective absorption of the x-rays so as to modify the continuous spectrum to produce peaks at different energy levels.

One important purpose of producing the differential x-ray images is to subtract or cancel out the portions of the x-ray images which are due to the ordinary tissues of the patient, particularly the soft tissues, so that the visibility of certain distinctive tissues or contrast substances in the patient's body will be enhanced. To produce good cancellation, the portions of the two different images which are due to the soft tissues must be balanced. However, even the ordinary soft tissues of the patient transmit x-rays of different energies to different extents. While the two different x-ray images can be balanced for any particular thickness of the patient, variations in such thickness over the field of view tend to upset such balance. Thus, without compensation for patient thickness variations, cancellation of the x-ray images due to the ordinary tissues of the patient can be achieved at only one value of patient thickness.

One object of the present invention is to provide effective compensation for variations in the thickness of the patient over the field of view, so that cancellation of the image portions due to ordinary soft tissue can be accomplished to a high degree over a range of patient thicknesses.

A further object is to provide effective compensation for patient thickness variations without resorting to a complex electronic compensation system.

In accordance with the present invention, compensation for patient thickness variations is achieved by adjusting the nature of the monoenergetic or quasi-monoenergetic x-ray spectra. It has been found that the quasi-monoenergetic spectra can be adjusted by adjusting the x-ray filtration and the supply voltage of the x-ray source. The adjustment of the voltage changes the x-ray spectrum produced by the x-ray tube. It is prefered to adjust the x-ray filtration so that one of the x-ray filters transmits a spectrum having two x-ray peaks at energies below and above the energy of the peak produced by the other x-ray filter. Thus, the image produced by the first x-ray filter consitutes the sum of the image components produced by the two peaks. Such sum tends to remain in a constant relationship to the image produced by the second spectrum, despite variations in patient thickness. The peaks of the x-ray spectra can be adjusted by varying the composition and density of the x-ray filters. It is prefered to vary the supply voltage to the x-ray tube as an inverse function of the average patient thickness. Thus, the voltage is reduced when the average patient thickness increases. The adjustment of the supply voltage changes the x-ray spectrum produced by the x-ray tube, and thereby modifies the two different quasi-monoenergetic x-ray spectra transmitted by the two x-ray filters. By adjusting the supply voltage, it is possible to optimize the compensation for patient thickness variations.

Many different materials may be employed in the x-ray filters. For example, one filter may contain iodine, while the other contains cerium. This combination of filters is particularly valuable for enhancing the visibility of iodine in the patient's body. The iodine filter produces a spectrum having first and second peaks at low and high energy levels. The cerium filter transmits a spectrum having a peak at an intermediate energy, higher than the low energy peak but lower than the high energy peak produced by the iodine filter. By adjusting the supply voltage, the relative magnitudes of the high and low energy peaks can be adjusted.

Further objects, advantages and features of the present invention will appear from the following description, taken with the accompanying drawings, in which:

FIG. 1 is a diagrammatic illustration of an x-ray system to be described as an illustrative embodiment of the present invention.

FIG. 2 is a set of graphs illustrating the variation in the x-ray attenuation coefficients, as a function of x-ray energy, for iodine and for water, which accounts for most of the attenuation produced by soft tissue.

FIG. 3 is a set of graphs illustrating the variation in the x-ray attenuation coefficients for iodine and for cerium.

FIG. 4 is a set of graphs illustrating the wide band x-ray spectrum produced by an x-ray tube, and also the quasi-monoenergetic x-ray spectra which may be produced by using iodine and cerium filters in conjunction with the x-ray tube.

FIG. 5 is a set of graphs illustrating the effects of variations in the patient tissue thickness, with and without compensation.

FIG. 6 is a set of graphs illustrating the manner in which compensation can be achieved for different ranges of patient thickness.

FIG. 7 is a graph illustrating the manner in which the x-ray supply voltage may be varied to achieve compensation for different values of patient thickness.

FIGS. 8–12 are additional graphs illustrating the effects of changing the filtration densities and the supply voltage.

As just indicated, FIG. 1 illustrates a system **10** for producing differential x-ray images, with compensation for variations in the thickness of the patient or subject **12** to be x-rayed. In general, the thickness of the portion of the patient to be x-rayed may vary in a more or less irregular manner over the field of view. While the cross sectional shape of the patient **12** is shown as a simple oval, it will be understood that the actual shape is more or less irregular.

The x-ray system **10** comprises an x-ray source assembly **14** for producing monoenergetic or quasi-monoenergetic x-ray spectra. In this case, the source assembly **14** is adapted to produce quasi-monoenergetic x-ray spectra. Thus, the x-ray source assembly **14** comprises an ordinary x-ray tube or source **16** which may be energized by a variable high voltage supply **18.** An x-ray source of this type produces a continuous x-ray spectrum over a wide band of energies, as represented by a graph **20** in FIG. **4.** The maximum energy of the band of x-rays is determined by the maximum voltage applied to the x-ray tube **16,** in this case sixty-five kilovolts peak (KVp).

The x-ray beam from the x-ray source **16** is directed through the patient **12** to an image detector **22.** To produce quasi-monoenergetic x-ray spectra, it is prefered to provide a plurality of selectively usable x-ray filters, two such filters **24***a* and **24***b* being shown. The x-ray filters **24***a* and *b* may contain various materials capable of absorbing x-rays in a selective manner, so that the continuous band of x-rays produced by the x-ray tube **16** will be converted into x-ray spectra having one or more peaks at various energy levels. A wide variety of filtering materials may be employed. For example, the filters **24***a* and *b* may utilize iodine and cerium. FIG. **4** includes graphs **26***a* and *b* representing quasi-monoenergetic spectra which may be produced by the use of the iodine and cerium filters. It will be seen that the spectrum **26***a* produced by the iodine filter **24***a* comprises a low energy peak **26***a*L and a high energy peak **26***a*H. The magnitude of the low energy peak **26***a*L is greater than the magnitude of the high energy peak **26***a*H. Similarly, the spectrum **26***b* produced by the cerium filter includes a low energy peak **26***b*L and a high energy peak **26***b*H. In this case, the high energy peak **26***b*H is so small as to be insignificant. The magnitude of the low energy peak **26***b*L for the cerium filter is comparable to the magnitude of the low energy peak **26***a*L for the iodine filter. It will be seen that the low energy peak **26***a*L for the iodine filter occurs at a lower energy than the low energy peak **26***b*L for the cerium filter. On the other hand, the high energy peak **26***a*H for the iodine filter occurs at a higher energy than that of the low energy peak **26***b*L for the cerium filter.

The x-ray filters **24***a* and *b* are arranged to be used alternately, so as to produce two different x-ray images. As shown in FIG. **1,** the filters **24***a* and *b* are movable alternately into the beam of x-rays from the x-ray tube **16.** Thus, the filters **24***a* and *b* may be mounted on a movable indexing member, such as the illustrated rotatable disc **28.** A mechanical device, such as the illustrated motor **30,** may be employed to rotate the disc **28.**

The two different x-ray images produced by the use of the filters **24***a* and *b* are detected by the image detector **22** and are subtracted by an image subtraction device **32.** The differential images thus produced may be displayed in some suitable fashion, in this case by a television monitor **34.**

Any known or suitable system may be employed for subtracting the x-ray images. For example, the image detector **10** may be arranged to produce positive and negative optical images corresponding to the two different x-ray images. The positive and negative images are then combined to produce the desired subtraction.

The prefered system is to convert the two different x-ray images into television images which are subtracted electronically to produce a differential television image, to be displayed on the television monitor **34.** Thus, the image detector **22** preferably comprises a television system for converting the x-ray images into electronic television signals. The image subtraction device **32** comprises means for subtracting the successive television images. Such means may utilize one or more image storage tubes, or other storage devices. For example, a silicon target storage tube may be employed, so that one of the two images can be written negatively, while the other image is written positively. In this way, any identical portions of the two images will be cancelled out. A differential image will develop on the silicon screen of the storage tube. This differential image can be displayed on the television monitor **34.**

While various image subtraction systems may be employed, it is prefered to utilize the highly advantageous image subtraction system disclosed in the co-pending patent application of Charles A. Mistretta and Michael G. Ort, Ser. No. 369,824, filed June 14, 1973. Such system uses two stages of subtraction. In such system, the x-ray images are converted into video signals, which are then processed through two successive stages of video subtraction, preferably utilizing two different types of video storage devices. The second stage of video subtraction also involves integration of the differential features, so that the contrast and visibility of such features can be built up over a multiplicity of cycles. In the prefered form of such subtraction system, an intensification screen and a television camera are employed to convert the x-ray images into first and second video image signals, which are successively supplied to a video difference detector. Such detector preferably utilizes a video storage tube capable of storing video images in the form of electrical charges distributed over a dielectric layer on a conductive target back plate.

The first and second video images are supplied sequentially to the storage tube, which produces a first differential video signal corresponding to the difference between the first and second video signals. The first differential video signal is then supplied to a second integrating and subtracting storage device, preferably utilizing a second storage tube capable of storing video images in the form of electrical charges on a mosaic of dielectric islands on a conductive back plate. The first differential video signal is written positively on the target of the second storage tube. The first storage tube is then employed to develop a second differential video signal corresponding to the difference between the second and first video image signals. The second differential video signal is written negatively on the target of the second storage tube. In this way, the differential features of the first and second differential video signals are integrated and enhanced, while the identical or non-differential features of the first and second differential video signals are combined subtractively so as to cancel them from the target of the second storage tube.

To obtain the maximum enhancement of the differential features, this cycle of subtraction and integration is repeated so that a multiplicity of cycles are completed. The enhanced image on the target of the second storage tube can be read, as desired, and reproduced as a visible display on a television monitor. By this system, differential features amounting to only a fraction of one percent of the full contrast range of the successive x-ray images can be enhanced to full contrast so that such differential features will be clearly visible.

Reference may be had to such co-pending application for a more detailed disclosure of such differential imaging system. Any other known or suitable differential imaging system may be employed.

The iodine and cerium filters **24***a* and *b* are especially valuable for visualizing small quantities of iodine in the patient's body. Iodine is present naturally, particularly in the thyroid gland. Moreover, iodine may be introduced into the patient's body through the bloodstream or the food canal, to serve as a contrast medium or substance.

FIG. 2 comprises a graph **36** representing the variation of the x-ray attenuation coefficient for iodine, as a function of the x-ray energy, expressed in kiloelectron volts (KEV). It will be seen that the x-ray attenuation coefficient generally decreases with increasing x-ray energy. However, the graph **36** has an abrupt discontinuity, usually refered to as the k-edge **38**, at which the x-ray attenuation coefficient increases sharply. Thus, at an x-ray energy E_A above the k-edge energy, the attenuation coefficient is abruptly and substantially greater than at an x-ray energy E_B, below the k-edge. The iodine and cerium filters make it possible to utilize the k-edge discontinuity **38** to produce an enhanced differential feature between the two x-ray images, so that the visibility of such feature will be increased in the differential x-ray image.

Thus, it will be seen from FIG. **4** that the low energy peak **26***a* L for the iodine filter is at approximately 30,000 electron volts, below the k-edge energy of about 33,000 electron volts. The low energy peak **26***b*L for the cerium filter is at an energy of about 40,000 electron volts, above the k-edge energy. Thus, the x-ray spectrum produced with the cerium filter will be attenuated much more by iodine than will the spectrum produced with the iodine filter.

FIG. **3** comprises graphs which compare the x-ray absorption coefficients for iodine and cerium. It will be seen that FIG. **3** again includes the graph **36** for iodine, showing the k-edge discontinuity **38.** FIG. **3** also includes a graph **40** for cerium. Here again, the x-ray absorption coefficient generally decreases with the increasing x-ray energy. However, there is a sharp discontinuity or k-edge **42** at a particular energy level, at which the x-ray absorption coefficient increases sharply. The k-edge **42** for cerium is at a substantially higher energy than the k-edge **38** for iodine.

The spectrum **26***a* for the iodine filter in FIG. **4** includes a sharp drop **44** which corresponds to the k-edge **38** for iodine. Similarly, the spectrum **26***b* for the cerium filter in FIG. **4** includes a sharp drop **46** which corresponds to the k-edge **42** for cerium.

FIG. 2 also includes a graph **47** which represents the variation of the x-ray attenuation coefficient for water, as a function of x-ray energy. It will be seen that attenuation coefficient decreases gradually with increasing x-ray energy. The absorption coefficient for water corresponds generally to the absorption coefficient for soft tissue. It will be seen that the absorption coefficient for the energy E_A above the k-edge for iodine is slightly less than the absorption coefficient for the energy E_B, below the k-edge **38.** It is possible to compensate for this slight difference for any particular value of patient thickness. This can be done, for example, by adjusting the relative densities of the iodine and cerium filters **24***a* and *b* so that the intensity of the x-rays transmitted through the iodine filter **24***a* and the pertinent portion of the patient's body is the same as the intensity of the x-rays transmitted through the cerium filter at the pertinent portion of the patient's body.

However, in accordance with the present invention, it has been found that effective compensation can be provided for variations in the thickness of the patient over a wide range, by adjusting the filters and the supply voltage to the x-ray tube, so as to produce particularly advantageous quasi-monoenergetic x-ray spectra.

As previously indicated, one of the filters, in this case the iodine filter **24***a*, is adjusted to produce an x-ray spectrum having two quasi-monoenergetic peaks, while the other filter, in this case the cerium filter **26***b*, is adjusted to produce only one significant quasi-monoenergetic peak. The two peaks produced by the first filter are below and above the energy level of the peak produced by the second filter. Thus, as to x-rays in the high energy peak for the first filter, the absorption coefficient for soft body tissues is less than for x-rays in the peak produced by the second filter.

As previously indicated, spectra of this type are illustrated in FIG. **4**, in which the spectrum **26***a* for the iodine filter has a low energy peak **26***a* L and high energy peak or bump **26***a* H. At an intermediate energy, the spectrum **26***b* produced by the cerium filter has a single significant peak **26***b*L. There may be a small high energy peak or bump **26***b*H for the cerium filter, but this small bump is so insignificant as to be negligible.

Quasi-monoenergetic spectra of this type produce compensation for variations in the thickness of the patient, because the high energy peak **26***a*H. of the first filter gradually predominates over the intermediate energy peak **26***b*L of the second filter as the patient thickness increases, while the low energy peak **26***a*L of the first filter gradually predominates over the peak **26***b*L for the second filter, with decreasing patient thickness. Due to this action, it is possible to achieve a close balance between the x-ray intensities transmitted through the two filters and the patient's body for a considerable range of patient thicknesses.

It will be evident from FIG. **4** that as the patient thickness increases, the low energy iodine peak **26***a*L is reduced with respect to the main cerium peak **26***b*L, but the high energy peak or bump **26***a*H of the iodine spectrum grows relative to the cerium peak **26***b*H. The sum of the high and low energy iodine portions **26***a*L and **26***a*H of the spectrum **26***a* thus remains approximately equal to the transmitted intensity of the cerium spectrum. The ratio of the transmitted intensities can be tuned to unity over a fairly broad range of patient thicknesses, as illustrated in FIG. 5, which compares the ratios of the transmitted intensities for the case of monoenergetic x-ray lines situated at the positions of the main iodine and cerium bumps 26aL and 26bL and for the case of three monoenergetic x-ray lines approximating the spectra provided by the iodine and cerium filters.

Thus, FIG. 5 comprises a first graph 50, shown in a broken line, which plots the ratio of I_{ABOVE} to I_{BELOW}, as a function of tissue thickness, expressed in centimeters. I_{ABOVE} is the intensity of the x-rays transmitted through the tissue at an x-ray energy of 36,000 electron volts (keV). This energy is above the k-edge for iodine. I_{BELOW} is the transmitted x-ray intensity through the body tissue for an x-ray energy of 30 keV, below the k-edge for iodine.

It will be seen that the graph 50 rises steadily with increasing tissue thickness, and passes through unity for only one value of tissue thickness, approximately 15 centimeters.

FIG. 5 includes a second graph 52, shown in a full line, in which I_{BELOW} is produced by two x-ray spectral lines, at 30 keV and at 50 keV. The latter line, at 50 keV, corresponds to the high energy peak 26aH of the spectrum 26a produced by the iodine filter 24a. It will be evident that the graph 52 has a broad plateau 52a, centered at an average value of patient thickness which can be chosen by adjusting the filters and the supply voltage. In this case, the plateau 52a is centered at an average patient thickness of about 15 centimeters. This plateau provides a broad region of relative insensitivity to patient thickness. Thus, the patient thickness can vary over the field of view without materially affecting the ability of the differential x-ray system to cancel out the portions of the x-ray images due to ordinary soft tissue.

The exact location of the plateau 52a of FIG. 5 can be changed by varying the filters and the voltage supplied to the x-ray tube. Suprisingly, it has been discovered that the center of the plateau moves to greater patient thicknesses as the supply voltage to the x-ray tube is lowered. Thus, the supply voltage needs to be adjusted as an inverse function of the average patient thickness. The supply voltage needs to be decreased as the average patient thickness is increased, and vice versa. Thus, for example, it has been found that for average patient thicknesses of 20 centimeters or more, it is highly desirable to use a supply voltage of 50,000 volts peak (kVp), with high tube current to produce the desired intensity of the x-rays, rather than raising the tube voltage.

Thus the supply voltage to the x-ray tube is an important factor in achieving effective compensation for patient thickness variations. The effect of changing the supply voltage is illustrated in FIG. 6, which comprises a series of graphs representing the ratio of cerium to iodine transmissions, as a function of patient thickness, for different supply voltages. The density of the cerium filtration was 200 milligrams per square centimeter.

Specifically, FIG. 6 comprises three graphs 54a, 54b and 54c, plotted for supply voltages of 50 kVp, 60 kVp and 70kVp. Each graph shows a plateau, which is inverted, in this case, because the ratio being plotted is the inverse of the ratio plotted in FIG. 5. The plateau of the curve 54a, plotted for 50 kVp, is centered at a patient thickness of about 22 centimeters, while the curves 54b and 54c, for supply voltages of 60 and 70 kVp, are centered at patient thickness values of about 11 centimeters and 6 centimeters, respectively. The transmission ratio at the center of each plateau is not unity, but compensation for this factor can be made by adjusting the electronic gain in the differential x-ray imaging system. Thus, different values of electronic gain can be used in the television system for the images produced with the use of the iodine and cerium filters, to bring about the optimum cancellation of the portions of the images representing ordinary soft tissue. The plateaus of the curves shown in FIG. 6 then provide wide regions in which the effectiveness of the cancellation is insensitive to variations in patient thickness.

FIG. 7 comprises a graph 56 in which the desirable x-ray supply voltage is plotted against patient thickness. This graph highlights the desirability of reducing the supply voltage to the x-ray tube as the average patient thickness is increased. By thus reducing the supply voltage, it is possible to optimize the compensation for variations in the patient thickness.

If the average patient thickness is decreased, the supply voltage should be increased to optimize the compensation for patient thickness variations. Thus, the supply voltage should be adjusted as an inverse function of the average patient thickness.

It is possible to change the composition and density of the x-ray filters so as to change the quasi-monoenergetic spectra produced by the use of the filters. For a particular average patient thickness, it is possible to adjust the filters so as to optimize the compensation for variations in the patient thickness.

As indicated by the legends in FIG. 4, the curve 26a represents a quasi-monoenergetic spectrum produced by the use of an iodine filter having a density or concentration of 0.200 grams of iodine per square centimeter. The quasi-monoenergetic spectrum represented by the curve 26b was produced by a cerium filter having a density or concentration of 0.238 grams of cerium per square centimeter.

It is readily possible to vary the concentration of the iodine, cerium or other material in the filter. For example, this can be done by constructing each filter so as to include a tank or other receptacle which can be filled with a liquid solution containing the iodine, cerium or other material in a dissolved state. The concentration of the solution can readily be varied. The tank or receptacle may be made of a plastic material or some other material which produces very little attenuation of x-rays.

The curves of FIG. 6 were produced with cerium and iodine filters. The cerium filter has a density or concentration of 200 milligrams of cerium per square centimeter. The iodine concentration of iodine filter was comparable and was adjusted to produce a ratio of unity between the iodine and cerium transmissions for zero patient thickness. The patient thickness was then varied for several different values of supply voltage to the x-ray tube, so as to produce the three curves 54a, 54b and 54c.

In FIG. 6, a percentage figure is indicated by a legend for each curve, representing the percentage of the differential x-ray signal for an iodine concentration of 1 milligram per square centimeter in the patient. The percentage signal applies to the broad minimum or inverted peak in the curve. Each percentage figure represents the maximum signal that can be obtained for that particular supply voltage to the x-ray tube. For a supply voltage of 50 kVp, the maximum differential signal was approximately 1 percent. For a supply voltage of 60 kVp, the maximum differential signal was approximately 0.95 percent. For a supply voltage of 70 kVp, the maximum differential signal was approximately 0.8 percent. It will thus be evident that the percentage value of the maximum differential signal decreases with increasing supply voltage to the x-ray tube. The graphs of FIG. 5 were obtained for monoenergetic spectra which approximate the quasi-monoenergetic spectra produced by iodine and cerium filters, as represented by FIg. 4. The curves **50** and **52** represent the results of calculations based on the use of monoenergetic x-rays. Thus, the curve **50** represents the ratio of I_{ABOVE} to I_{BELOW} as a function of tissue thickness, when the x-ray energy E_{ABOVE} is 36 keV, while the x-ray energy E_{BELOW} is 30 keV. The x-ray energy E_{ABOVE} of 36 keV is above the k-edge for iodine, while the x-ray energy E_{ABOVE} of 30 keV is below the k-edge for iodine. The x-ray intensities I_{ABOVE} and I_{BELOW} represent the transmitted x-ray intensities for the x-ray energies E_{ABOVE} and E_{BELOW}.

The curve **52** of FIg. **5** represents the ratio of I_{ABOVE} to I_{BELOW} for the same x-ray spectra, except that the x-ray component E_{BELOW} at 30 keV is supplemented by an x-ray component at 50 keV having an intensity of approximately 14 percent of the intensity of the 30 keV component.

The x-ray component at 36 keV corresponds generally to the cerium peak **26***b*L of FIG. **4**, which is above the k-edge of iodine, while the x-ray component of 30 keV corresponds generally to the iodine peak **26***a*L, which is below the k-edge for iodine. The x-ray component at 50 keV corresponds generally to the high energy bump **26***a*H for iodine, as represented in FIG. **4**.

FIGS. **8**, **9** and **10** represent the results which are obtained by using different values of filter density or thickness, for different values of the supply voltage to the x-ray tube. In each graph, the ratio of TB to TA is plotted as a function of variations in the tissue thickness of the patient. TB is the transmission of x-rays through the patient with the use of the iodine filter, which has its major spectral peak below the k-edge for iodine. Such peak is designated **26***a*L in FIG. **4**. TA is the transmission of the x-rays through the patient with the use of the cerium filter, having its major peak **26***b*L above the k-edge of iodine.

FIG. **8** comprises four curves **68***a*, **68***b*, **68***c* and **68***d*, representing the results obtained by using four different sets of filters at an x-ray supply voltage of 50 KVp. For the curve **68***a*, the cerium filter had a density of 0.100 grams of cerium per square centimeter, while the iodine filter had a density of 0.101 grams of iodine per square centimeter. For an iodine concentration in the patient of one milligram per square centimeter, the differential signal has a percentage value of 0.76 percent at a patient thickness of 15 centimeters.

For the curve **68***b*, the corresponding values are as follows: cerium filter density 0.200; iodine filter density 0.195; percentage of differential signal, 1.19 percent.

For the curve **68***c*, the corresponding values are as follows: cerium filter density 0.300; iodine filter density 0.284; percentage of differential signal, 1.42 percent.

For the curve **68***d*, the corresponding values are as follows: cerium filter density 0.400; iodine filter density 0.370; percentage of differential signal, 1.54 percent.

It will be understood that the symbol TA in FIGS. **8**, **9** and **10** represents the total number of transmitted photons in the cerium spectrum, having its major peak above the k-edge for iodine. The symbol TB represents the total number of transmitted photons in the iodine spectrum, having its major peak below the K-edge for iodine. For each curve, the relative densities of the cerium and iodine filters were adjusted to produce a ratio of unity between TB and TA, for a patient tissue thickness of zero.

In each of the curves **68***a–d* of FIG. **8**, the ratio of TB to TA decreases with increasing tissue thickness toward a minimum which occurs at approximately 20 centimeters or greater, depending upon the density of filtration. These curves represent the situation for an x-ray tube supply voltage of 50 KVp. For this relatively low supply voltage, and for even lower voltages, any increase in the density of filtration tends to increase the tissue thickness at which the minimum occurs.

FIG. **9** comprises four curves **69***a*, **69***b*, **69***c* and **69***d*, which represent the ratio of TB to TA as a function of patient tissue thickness, for increasing values of filtration, at a supply voltage of 60 KVp. For the curve **69***a*, the cerium filter had a density of 0.100 grams per square centimeter, while the iodine filter had a density of 0.101 grams per square centimeter. The differential signal was 0.60 percent at a tissue thickness of 10 centimeters for an iodine concentration in the patient of one milligram per square centimeter.

The corresponding values for the curve **69***b* are as follows: cerium filtration 0.200; iodine filtration 0.197; differential signal 0.95 percent.

For the curve **69***c*, the corresponding values are as follows: cerium filtration 0.300; iodine filtration 0.288; differential signal 1.18 percent.

For the curve **69***d*, the corresponding values are as follows: cerium filtration 0.400; iodine filtration 0.376; differential signal 1.34 percent.

It will be observed that for the supply voltage of 60 KVp, represented by FIG. **9**, the ratio of TB to TA decreases with increasing tissue thickness until a minimum value is reached, whereupon the ratio increases with further increases in the tissue thickness. The increases in the filtration values do not increase the tissue thickness at which the minimum occurs, to any great extent. Thus, the position of the minimum remains at about 10.5 centimeters, despite the changes in the concentrations of cerium and iodine in the filters. The curves of FIG. **9** indicate that a supply voltage of 60 KVp is appropriate for an average tissue thickness of about 10.5 centimeters, because this value of supply voltage provides effective compensation for variations in patient tissue thickness. Such compensation is optimized by a supply voltage in this general neighborhood.

FIG. **10** comprises four curves **70***a*, **70***b*, **70***c* and **70***d*, representing the ratio of TB to TA as a function of patient tissue thickness for a supply voltage of 70 KVp, at four different filter concentrations or densities. For the curve **70***a*, the cerium concentration was 0.100 grams per square centimeter. The iodine filter concentration was 0.102 grams per square centimeter. The differential signal was 0.46 percent for an iodine concentration of one milligram per square centimeter in the patient.

The corresponding values for the curve **70***b* are as follows: cerium concentration 0.200; iodine concentration 0.201; differential signal 0.70 percent.

The corresponding values for the curve **70***c* are as follows: cerium concentration 0.200; iodine concentration 0.298; differential signal 0.86 percent.

For the curve **70***d*, the corresponding values are as follows: cerium concentration 0.400; iodine concentration 0.392; differential signal 0.99 percent.

It will be observed that the curve **70***a* has a minimum at a tissue thickness of about seven centimeters. The ratio of TB to TA increases for smaller or larger values of tissue thickness.

With increasing filtration, the minimum value of the ratio tends to occur at smaller values of tissue thickness. Thus, the minimum value of the curve **70***b* is at a tissue thickness of about 6 centimeters. Thus, in general, at a supply voltage of 70 KVp or higher, increasing the filtration tends to decrease the tissue thickness at which the minimum occurs. Moreover, the value of the ratio of TB to TA at the minimum point tends to increase with increasing filtration. This is the opposite of the situation for supply voltages of 50 or 60 KVp, at which the value of the ratio at the minimum tends to decrease with increasing filtration.

In summary, the effect of changing the densities or concentrations of the filters depends upon the supply voltage to the x-ray tube. At 70 KVp or higher, increasing the filtration tends to decrease the tissue thickness at which the minimum value of the ratio of TB to TA occurs. At 60 KVp, changing the concentrations of the filters does not affect the position of the minimum point to any substantial extent. At 50 KVp or lower, increasing the filtration tends to increase the tissue thickness at which the minimum ratio occurs.

For an average tissue thickness of about 10.5 centimeters, 60 KVp is an appropriate supply voltage, because the compensation for variations in the patient tissue thickness will be optimized. For average values of patient tissue thickness ranging down to 5 centimeters or less, the voltage should be increased progressively to 70 KVp or higher, to optimize such compensation. As the average tissue thickness increases to 20 centimeters or higher, the voltage should preferably be reduced progressively to 50 KVp or lower, so as to optimize the compensation for variations in the tissue thickness.

FIG. **11** and **12** will be helpful in explaining the effects represented by FIGS. **8–10.** It will be seen that FIGS. **11** and **12** reproduce the graph **36** of FIG. **3**, representing the x-ray absorption coefficient for iodine, plotted as a function of x-ray energy. The x-ray absorption coefficient is designated m in FIGS. **11** and **12.** As before, the graph **36** includes the abrupt k-edge **38.**

FIGS. **11** and **12** also include the graph **26***a* of FIG. **4**, representing the quasi-monoenergetic spectrum produced by the use of the iodine filter **24***a*. The graph **26***a* is superimposed upon the graph **36** in each case.

In FIG. **11**, the spectrum graph **26***a* is drawn for an x-ray supply voltage of 50 KVp, but in FIG. **12**, the graph **26***a* is drawn for a supply voltage of 70 KVp. In FIG. **11**, m_1 represents the value of the x-ray attenuation coefficient for iodine at the x-ray energy value corresponding to the low energy peak **26***a*L of the iodine filter spectrum **26***a*. On the other hand, m_3 represents the value of the x-ray attenuation coefficient at the energy level corresponding to the high energy peak or bump **26***a*H of the iodine filter spectrum. It will be seen that m_3 is substantially greater than m_1. As a result, at a supply voltage of 50 KVp, the high energy bump **26***a*H loses to the low energy bump **26***a*L as the filters are made thicker or denser. As a result of this decrease in the relative strength of the high energy bump or peak **26***a*H, a greater tissue thickness is required before the transmitted x-rays due to the high energy bump **26***a*H catch up with the x-rays transmitted by the cerium filter. The energy level of the x-rays transmitted by the cerium filter is substantially lower than the energy level represented by the high energy bump **26***a*H, so that a greater percentage of the x-rays represented by the high energy bump are able to penetrate the increased tissue thickness.

The effect is just the reverse at 70 KVp, as will be evident from FIG. **12.** It will be seen that the x-ray attenuation coefficient m_1 at the energy of the low energy bump or peak **26***a*L is greater than the x-ray attenuation coefficient m_3 at the x-ray energy corresponding to the high energy bump **26***a*H. Thus, as the filters are made thicker or more dense, the x-rays due to the high energy bump **26***a*H actually gain in magnitude relative to the x-rays due to the low energy bump **26***a*L. Thus, less tissue thickness is required for the more pentrating x-rays due to the high energy bump **26***a*H to equal or exceed the x-rays transmitted through the patient due to the cerium spectrum **26***b*, shown in FIG. **4.** Thus, the minimum along each curve in FIG. **10** occurs at a relatively small tissue thickness. As the filters are made more dense, the minimum occurs at a decreased tissue thickness. This represents the situation at 70 KVp.

In FIG. **8**, representing the situation at 50 KVp, the minimum along each curve occurs at a relatively great tissue thickness. As the filters are made more dense, the minimum occurs at a greater tissue thickness.

FIG. **9** represents an intermediate situation at 60 KVp. In this case, the minimum along each curve occurs at an intermediate tissue thickness. Changes in the tissue thickness do not affect the location of the minimum to any great extent.

From the graphs of FIGS. **8, 9** and **10**, it will be possible for those skilled in the art to select an appropriate supply voltage and appropriate filter densities, according to the average tissue thickness involved, in order to achieve effective compensation for variations in the tissue thickness. The supply voltage and the amount of filtration should be selected to produce a curve having a minimum at or near the average tissue thickness.

It will be evident that iodine and cerium filters may be employed very advantageously for producing differential x-ray images due to iodine in the patient's body. Such iodine may be present naturally, as in the thyroid, or may be introduced into the patient's body as a contrast agent. The iodine and cerium filters are also valuable for producing differential x-ray images due to xenon gas, which may be inhaled into the lungs of the patient. Ordinary xenon gas can be employed, because the xenon does not have to be radioactive. The xenon gas produces a k-edge which is at a somewhat higher energy level than the k-edge for iodine, but at a lower energy level than the k-edge for cerium. Thus, the iodine and cerium filters can be employed to produce a differential x-ray image due to xenon gas in the lungs of the patient. By this technique, the lungs can be visualized with a greater degree of clarity.

Various other contrast media may be employed to produce differential x-ray images. Barium is an example of another contrast substance. For each contrast substance, the composition and density of the x-ray filters **24***a* and **24***b* are selected so as to produce quasi-monoenergetic x-ray spectra having peaks above and below the k-edge for the particular contrast substance. By proper selection of the x-ray supply voltage and the filter densities, effective compensation can be achieved for variations in the tissue thickness of the patient.

We claim:

1. A method of producing differential x-ray images to visualize a contrast material in a patient,

comprising the steps of producing a first x-ray image using an x-ray source and first filter,

producing a second x-ray image using said x-ray source and a second filter,

said filters producing substantially different x-ray spectra,

producing a differential image corresponding to the difference between said first and second x-ray images,

said contrast material being visualized in said differential image,

and adjusting the voltage of said x-ray source as in inverse function of the average patient thickness to minimize the effect of variations in patient thickness upon said differential image.

2. A method according to claim **1**,

in which the voltage of said x-ray source is adjusted to a range which produces a flat peak of the ratio between said first and second x-ray images when plotted against patient thickness.

3. A method of producing differential x-ray images,

comprising the steps of producing a first x-ray image using a first x-ray spectrum,

producing a second x-ray image using a second x-ray spectrum,

and producing a differential image corresponding to the difference between said first and second x-ray images,

said second spectrum having a spectral element of a predetermined energy,

said first x-ray spectrum comprising a low energy spectral element and a high energy spectral element,

said low energy spectral element having an energy below said predetermined energy,

said high energy spectral element having an energy above said predetermined energy.

4. A method according to claim **3**,

in which said first and second x-ray spectra are balanced to provide substantially the same image elements in said first and second x-ray images for soft tissue whereby said image elements will cancel out in said differential image.

5. A method according to claim **4**,

in which said spectral elements are adjusted to produce substantial balance between said first and second spectra over a substantial range of tissue thickness.

6. A method according to claim **3**,

in which said first and second spectra are produced by using first and second x-ray filters in conjunction with an x-ray source.

7. A method according to claim **6**,

in which said first and second filters contain iodine and cerium, respectively.

8. A method according to claim **6**,

in which said x-ray source includes a high voltage x-ray tube,

the supply voltage to said x-ray tube being varied as an inverse function of the average tissue thinkness to maintain a substantial balance between said first and second spectra for soft tissue over a substantial range of tissue thickness.

9. A method according to claim **6**,

including the additional step of changing the density of said filters as a direct function of the average tissue thickness of the patient.

10. A method according to claim **6**,

including the steps of adjusting the density of said filters as a direct function of the average tissue thickness of the patient while adjusting the x-ray energy of said source as an inverse function of said average tissue thickness to achieve effective compensation for variations in said tissue thickness.

11. A method according to claim **6**,

including the steps of adjusting the density of said filters while adjusting the x-ray energy of said source to achieve effective compensation for variations in the tissue thickness of the patient.

12. A method according to claim **6**,

including the steps of adjusting the density of said filters while adjusting the x-ray energy of said x-ray source to produce a flat spot in the characteristic curve of TB/TA plotted as a function of patient tissue thickness at a desired average value of patient tissue thickness,

where TB is the total of the x-ray photons transmitted through the first filter and the patient while TA is the total of the x-ray photons transmitted through the second filter and the patient,

whereby effective compensation is achieved for variations in the patient tissue thickness.

13. A method according to claim **12**,

in which said first and second filters contain iodine and cerium, respectively.

14. A method according to claim **13**,

in which said filters have iodine and cerium concentrations on the order of 100 to 400 milligrams per square centimeter.

15. A method according to claim **12**,

in which said flat spot is in the form of a minimum along said characteristic curve.

16. A method according to claim **12**,

in which said x-ray source includes an x-ray tube supplied with a high voltage,

said voltage being adjusted as an inverse function of the patient tissue thickness.

17. A method according to claim **16**,

in which said first and second filters contain iodine and cerium, respectively.

18. A method according to claim **17**,

in which the iodine and cerium concentrations of the filters are on the order of 100 to 400 milligrams per square centimeter.

19. A method according to claim **18**, in which said high voltage is on the order of 50 to 70 kilovolts peak.

20. A method according to claim **17,**
in which said high voltage is on the order of 50 to 70 kilovolts peak.

21. Apparatus for producing differential x-ray images in which image elements due to the soft tissue of a patient are at least partially cancelled out,
comprising x-ray source means for selectively producing first and second x-ray spectra,
means for producing first and second x-ray images of the patient using said first and second x-ray spectra,
and means for producing a differential image corresponding to the difference between said first and second x-ray images,
said second x-ray spectrum having an x-ray spectral element of predetermined energy,
said first x-ray spectrum having low and high energy spectral elements,
said low energy element having an energy less than said predetermined energy,
said high energy element having an energy greater than said predetermined energy.

22. Apparatus according to claim **21,**
in which said spectral elements have magnitudes such as to produce substantial balance between said first and second x-ray images for soft patient tissue.

23. A method according to claim **22,**
in which said spectral elements are of magnitudes to maintain the balance between said first and second x-ray images for soft tissue over a wide range of tissue thickness variations.

24. Apparatus according to claim **21,**

in which said x-ray source means comprises an x-ray source and first and second selectively usable filter means for producing said first and second x-ray spectra.

25. Apparatus according to claim **24,**
in which said first and second filter means are movable selectively into operative relation with said x-ray source.

26. Apparatus according to claim **24,**
in which said first and second filter means contain iodine and cerium, respectively.

27. Apparatus according to claim **24,**
in which said first and second filter means contain iodine and cerium, respectively, in concentrations on the order of 100 to 400 milligrams per square centimeter.

28. Apparatus according to claim **27,**
in which said x-ray source includes a high voltage x-ray tube and means for supplying said tube with a high voltage on the order of 50 to 70 kilovolts peak.

29. An apparatus according to claim **24,**
in which said first and second filter means contain iodine and cerium, respectively,
said x-ray source including a high voltage x-ray tube and means for supplying said tube with a high voltage on the order of 50 to 70 kilovolts peak.

30. Apparatus according to claim **24,**
in which said x-ray source comprises a high voltage x-ray tube,
and means for varying the high voltage supplied to said tube.

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