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TRANSMISSION SCANNING II*

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ABSTRACT

Work has been continued on the development of techniques for imaging spatial elemental distributions by means of differential absorption about the K-absorption edge of the particular element Z. A method to correct for spurious effects due to differential absorption in overlying absorber is described. Initial clinical studies of thyroid iodine images on patients are presented. Application of absorption edge transmission scanning to thin *in-vitro* samples of bone is also discussed.

Introduction

Work has been continued on the technique of imaging general stable elemental distributions by absorption edge scanning. Absorption edge scanning may be defined as the numerical comparison of the intensity of photon beams transmitted through a sample at energies above and below an absorption edge, the binding energy of an atomic shell, e.g. K, L (K or L-edge). In our previous papers^{1,2}, we have explored the feasibility of using high resolution solid state detectors (Ge and Si) to extend the classical methods of absorption edge transmission scanning. At present, Mistrretta *et al.*³ is extending these non-spectrometric methods through the use of modern image intensification techniques. Our spectrometric method utilized the energy resolution of these solid state detectors to increase the sensitivity and relax the time modulation requirements on the incident beam without the complexity and poor efficiency of a crystal diffractometer. This paper presents the extension of our phantom studies to human patients and includes a discussion of the problems associated with this transition.

Transmission scanning has also been used in the past to scan samples *in vitro*.⁴ We have developed instrumentation which will scan thin slices of bone *in vitro* for calcium. Although we have not quantified the iodine content in thyroid transmission scanning, the capability to do so is inherent in the method. *In-vitro* scanning of bone demands quantitative results and these methods have been developed. We hope, in the future, to extend this method to other elements as the need and application arise.

Part I. Iodine Absorption Edge Scanning

Experimental Apparatus

Although details of the experimental apparatus have been described previously,² (hereafter referred to as I) we will briefly describe it here. Figure 1 is a schematic view of the physical geometry of the

system. The subject to be scanned is held stationary on a horizontal table while the source and detector move in a rectilinear raster pattern. The transmitted beam is detected by a high purity Ge detector, whose efficiency at the 33 keV iodine K-edge is ~ 100%. The x-ray source is a secondary target of mixed salts of barium and cerium (32.06 and 34.56 keV K_{α} lines of each element). The excitation of this source is provided by a Machlett OEG-60 end window tube operated at 60 kVCP. Collimation is provided by a Pb collimator 3 inches long, with 2 mm pinhole apertures.

In I an x-ray source with a broad energy spectrum was used. Since this was primary radiation from a powerful (2 kW) tube, very fine spatial collimation was possible; the inherent spatial resolution of the system in this configuration was less than 1 mm. Since we are interested in comparing transmission at energies immediately above and below the edge, a high count rate at these energies is very desirable. Any intensity far from the absorption edge yields no useful information in determining the distribution of the element under investigation. On the contrary, photons at these energies are detrimental to the operation of the system since they saturate its count rate capacity and cause such undesirable effects as pileup and baseline shift. These effects cause energy spectrum distortion, which, in turn, can cause a spurious quantity of the element under investigation to be detected. In addition to such spectrum distortion effects, a spurious effect can be induced by the variation in the amount of overlying absorber. We will discuss this effect and present an analysis correcting for it in a later section. However, a result of this analysis shows that the narrower the energy acceptance windows are about the K-edge, the less sensitive the system is to overlying absorber variations. In I we compromised on an energy acceptance of 2 keV on each side of the absorption edge of iodine for a total acceptance of 4 keV. This setting was adequate for phantom studies, however the larger variations of absorber thickness in the human body necessitated the reduction of the energy spread about the absorption edge. The secondary target system was implemented in a geometry which improved the rate from Ref. 1, maintaining it at approximately the level quoted in Ref. 2, (I), in between 1 and 2×10^4 /sec in each energy interval.

* This work was performed under the auspices of the U. S. Atomic Energy Commission.

† See Ref. 2 for a more comprehensive survey of the literature. We note that this technique has been described as early as 1925 by Glocker *et al.*, Ann. Physik **76**, 369(1925).

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Above this rate, energy spectrum distortion creates a spurious effect. These lines lie within 1.25 keV of the edge, and we are less sensitive to the effects of overlying absorber. In order to achieve this rate, some spatial resolution has been sacrificed. Spatial resolution is now ≈ 3 mm with the collimator described above. Figure 2 shows a graph of the incident energy spectrum. Notice that the Ce and Ba K α lines do not interfere with the absorption edge energy of 33.17 keV.

Figure 3 shows a block diagram of the electronics. This again has been described in detail in I. This system calculates C_1 , the amount of iodine in grams/cm² in a picture resolution area as a function of T_A and T_B , the number of counts above and below the absorption edge in each resolution area. In I, C_1 was calculated as T_A/T_{B-1} . Now, however, to take into account corrections for overlying absorber, $C_1 = f(T_A, T_B)$ where the form of this function is described in the following section. We have replaced the analogue divider with a more general analogue function module necessary to compute $f(T_A, T_B)$.

Calculation of C_1

Let us assume a general case of N unique x-ray energies and N unique substances to be detected. We need not specify whether such a system is in general physically realizable. Later we shall go to the specific cases under study. Each incident line of energy E_i shall have an intensity of T_i^0 . Thus the intensities of the N transmitted beams are:

$$T_i = T_i^0 e^{-\sum_{j=1}^N \mu_{ij} \sigma_j}, \quad i = 1, N \quad (1)$$

where μ_{ij} is the mass absorption coefficient for the j th substance at the i th energy in cm²/gm. σ_j is the amount of substance j in g/cm² in the transmitted beam. Clearly, we may rewrite the above as

$$\ln T_i = -\ln T_i^0 - \sum_{j=1}^N \mu_{ij} \sigma_j, \quad i = 1, N \quad (2)$$

The formal solution of this set of equations can be written

$$\sigma_i = \frac{N-1}{\sum_{j=1}^{N-1} \mu_{ij}^2} \quad i = 1, N \quad (3)$$

We may make a simple observation here: the set of equations has no solution if any two columns of the μ matrix are linear multiples of each other. Physically this means that if any two substances in the sample have mass absorption coefficients with the same energy dependence, then the amounts of the two different substances present cannot be measured. Thus, schemes where one seeks to determine N different substances using N x-ray lines of different energies are difficult to realize since most substances tend to have mass absorption coefficients of very similar dependence as a function of energy.⁵ Notice that this objection is not true where one is dealing with two substances one of which has an absorption edge in the region of interest. Then the functional dependences of the mass absorption coefficients are quite different.

We may also calculate the standard deviation expected in terms of the number of photons measured. Define

$\Delta \sigma_i$ as the standard deviation of σ_i , $i = 1, N$

$\Delta \sigma_i$ as the measured standard deviation of the transmitted beam; then by the well-known methods of error propagation.

$$(\Delta \sigma_i)^2 = \sum_{j=1}^N (\mu_{ij}^{-1})^2 (\Delta \sigma_j)^2, \quad i = 1, N \quad (4)$$

$$\text{But } \Delta \sigma_j = \frac{1}{\sqrt{T_j}}$$

$$\text{Thus } (\Delta \sigma_i)^2 = \sum_{j=1}^N (\mu_{ij}^{-1})^2 / T_j \quad (5)$$

Finally, the expression for T_j in terms of the incident beam intensity, may be substituted thus

$$(\Delta \sigma_i)^2 = \sum_{j=1}^N (\mu_{ij}^{-1})^2 \frac{e^{-\sum_{k=1}^N \mu_{jk} \sigma_k}}{T_j^0}, \quad i = 1, N \quad (6)$$

If we assume $T_j^0 = T^0$ for all j , the above reduces to

$$(\Delta \sigma_i)^2 = \frac{1}{T^0} \sum_{j=1}^N (\mu_{ij}^{-1})^2 \exp \left\{ \sum_{k=1}^N \mu_{jk} \sigma_k \right\} \quad (7)$$

Equation (7) is useful in optimizing parameters of the system as will be shown later in the section on bone scanning in vitro.

We now return to the specific case of iodine absorption edge scanning. Figure 4 is a graph of the absorption coefficient for iodine, the substance to be detected, and compact bone, an overlying absorber.⁶ If one makes the identification for the mass absorption coefficients

$$\mu_{11} = \mu_- \text{ (iodine below K-edge),}$$

$$\mu_{12} = \mu_1 \text{ (bone below K-edge),}$$

$$\mu_{21} = \mu_+ \text{ (iodine above K-edge),}$$

$$\mu_{22} = \mu_2 \text{ (bone above K-edge),}$$

then we may specifically solve Eq. (3) for the case $N = 2$:

$$C_1 = \frac{a_B - (\mu_1/\mu_2)a_A}{\mu_- - \mu_+ \left(\frac{\mu_1}{\mu_2} \right)} \quad (8)$$

where C_1 is the amount of iodine in grams/cm²
 a_B is the log of the counts at energy B (BaK α)
 a_A is the log of the counts at energy A (CeK α).

A similar expression may be obtained for σ_B , the quantity of bone in grams/cm². We are, however, only interested, in this case, in the correction to σ_I , so the measurement of σ_B is not made.

Define $\Delta u = u_- - u_+ \left(\frac{\mu_1}{\mu_2} \right)$. Then we may explicitly form

the function $\exp \{ \sigma_I \Delta u \} = R$, where

$$R = \exp \{ \sigma_I \Delta u \} = \frac{T_B^0}{T_B} \cdot \left(\frac{T_A}{T_B^0} \right)^{\mu_1/\mu_2} = \text{const.} \times \frac{T_A^{\mu_1/\mu_2}}{T_B} \quad (9)$$

It is this function of R which is generated by the analogue function generator shown in Fig. 3. For suitably small $\sigma_I \Delta u$, R is proportional to $1 + \sigma_I \Delta u$. As the "1" is subtracted in the analogue circuitry, as shown in Fig. 2, we have a quantity at the output proportional to the amount of iodine present in the sample and insensitive to the amount of overlying absorber.

Although the above analysis treats the case of bone, any overlying absorber may be corrected in the same manner. In thyroid scans shown below, we have used this technique to correct for variations in quantity of soft tissue when bone is not present in the beam.

Figures 5a and b show the output voltage from the electronics in Fig. 3, which should be proportional to iodine content alone and independent of variation in overlying absorber thickness. In Fig. 5a no iodine was placed in the beam, but various amounts of CaCl_2 were. There are large changes in output, V_{out} , which cannot be distinguished from those caused by the presence of iodine. The solid line is a theoretical fit to the data using the calculated response of the system. In Fig. 5b output voltage, V_{out} , is plotted as a function of the amount of iodine in the beam, σ_I . Although the variations in density due to CaCl_2 are as large as in Fig. 5a, the effect of these is compensated by calculating the amount of iodine from Eq. (9). The quantity μ is μ_1/μ_2 . This value is in close agreement with the value calculated from the tables of mass absorption coefficients.⁶ The solid line is again a theoretical fit calculated from the anticipated response of the system. Figures 6a, b, c show the response of the system to a Picker thyroid phantom filled with 2 mg/cm² iodine in the thick lobe. Placed in the upper right hand corner is a square of aluminum .250" thick. Three values of $\mu_1/\mu_2 = 1.00, 1.15$ and 1.16 are shown in Figs. 6a, b, c. Figure 6c shows we have restored the image by compensating for the overlying absorber. Figure 6a shows the full effect of the uncompensated absorber on the image.

Thus we have demonstrated a correction for the effect of overlying absorber shown in Fig. 5a, and Fig. 5 of I. There are some conditions which must be put on the composition of the subject to be scanned for this compensator to operate at peak efficiency. The ratio of μ_1/μ_2 of the absorber must not change as a function of spatial coordinates over the sample. In practice this means we cannot correct variations in bone and soft tissue at the same time since μ_1/μ_2 is different for each of these substances. It is for this reason that we have used this compensator ultimately as a device to correct for variations in soft tissue in transverse scans of the thyroid, examples of which are shown in the next section.

Application to Human Thyroids

We have attempted thyroid scans in both lateral and anterior patient positions. In each case the patients chosen were euthyroid according to conventional medical screening procedures at the BNL Medical Department. Patients were chosen at random from among the general hospital population. In each case, the conditions of the scan were kept as close as possible to realistic clinical practice; the time spent on each scan was approximately twenty minutes.

Figure 7a shows the geometrical orientation of the lateral thyroid scans. The beam passes through the patient, as indicated, orthogonal to the plane of the paper and in the area outlined in dotted lines. To help orient the viewer, we have delineated the anatomical region in which the thyroid lies with plastic capillary tubing filled with sodium diatrizoate. These are labeled "iodine filled tubes" in Fig. 7a. As a further aid to the viewer and to guard against spurious responses, we have recorded the intensity of the transmitted beam whose energy is below the iodine absorption edge on a separate film simultaneously with the absorption edge scan. These scans are shown in Fig. 7c. Here we note the absence of any indicators of either the thyroid or the iodine filled tubes. Anatomical features, such as the spine or trachea, may be used as landmarks to further check the position of the thyroid.

Figure 7b shows the results of an iodine absorption-edge transmission scan on a typical patient. The lower scan has the iodine filled capillary tubes in place while the upper scan is a repeat with the tubes removed. The tubes are the dark bands in the scan running obliquely from top to bottom on the right hand side and from left to right in an elliptical fashion. In each case, the thyroid appears clear and well-defined. For these scans, the mechanism of correction for changes in overlying absorber was set to compensate for varying thicknesses of soft tissue since there was no bone lying over the thyroid. A vague outline of the spinal column can be seen to the left of the thyroid since the compensation setting for soft tissue is different from that of bone.

Figure 7d is a scan of the thyroid taken with the patient in the supine position. The upper scan is a film made from the beam whose energy is below the iodine K-edge, as discussed previously, while the lower is the absorption edge transmission scan. The compensation for overlying absorber was set mid-way between the value used for bone and soft tissue. While not compensating either perfectly, this setting compromises the spurious effect caused by density changes in each material. Clearly evident is the anatomical structure lying posterior to the thyroid. However, the dark bands on either side of the spinal column are in the correct position and appear with the correct intensity. Due to the underlying structure, this scan is more difficult to interpret than the lateral view; we include it here for completeness.

The above results do not represent a comprehensive and quantitative clinical trial of absorption edge scanning in thyroid imaging. Rather, by overcoming experimental difficulties, we have tried to present a technique with possible value in a clinical setting. Indeed, we have recognized the need to pursue a technique beyond phantom measurements into preliminary clinical application. Further tests of this technique may now be carried out by an interested clinician. Whether or not the

iodine absorption-edge scanning is ultimately useful, the general technique is one which should be exploited wherever rapid quantitative measurements of minor elemental constituents are desired.

Part II. In Vitro Bone Scanning Using an Absorption Edge Technique

Absorption edge scanning may be applied to almost any element provided the thickness of the sample is such that x-rays of energies in the region of the absorption edge can penetrate. Clearly this rules out applications *in vivo* of relatively light elements, as demonstrated in I. However, it is possible to measure light element distribution in thin samples *in vitro*. We have investigated one such application, i.e. calcium distributions in thin sections of rat femur. Although the elemental distribution can also be elucidated by fluorescent techniques, the transmission geometry allows quantitative determinations.

Experimental Apparatus

Figure 8 is a schematic diagram of apparatus used in the measurement. A beam of x-rays from a Machlett EG-50 tube are collimated by interchangeable pinholes of 5, 10, or 20 thousandths of an inch diameter. A rotating wheel causes the sample to be passed through the beam. The transmitted beam is detected by a high resolution Si(Li) detector which has an energy resolution of 180 eV at 5.9 keV. The energy spectrum is recorded on the pulse-height analyzer and processed off-line to yield the elemental distribution.

Figure 9 shows the energy spectrum of x-rays incident on the sample. The large peak on the left is due to characteristic L x-rays from the rhodium target in the EG-50 tube. Except for this peak the spectrum is quite uniform and thus suitable for transmission scanning.

Figure 10 shows the energy spectrum of the beam transmitted through a typical slice of rat bone .004" thick. Here, the abrupt change in absorption coefficient at 4.037 keV due to the K-edge of calcium is quite evident. Notice the much reduced intensity of the rhodium L x-ray at 2.69 keV.

Analysis

Equation (8) may be used to reduce the data from energy spectra of the type shown in Fig. 13. Suitable mass absorption coefficients of calcium must be substituted for the ones used previously. However, the basic analysis remains the same. Thus

$$\sigma_{Ca} = \frac{\alpha_B - (\mu_1/\mu_2)\alpha_A}{\mu_- - \mu_+ \left(\frac{\mu_1}{\mu_2} \right)} \quad (10)$$

where σ_{Ca} is the amount of Ca in the beam in g/cm^2 ,

$\mu_+(-)$ is the mass absorption coefficient of Ca above (below) the K-edge,

$\mu_1(2)$ is the mass absorption coefficient of the overlying absorber above (below) the edge,

$\alpha_{A(B)}$ is the logarithmic transmission in specified energy intervals above (below) the edge.

Since we are not dealing with discrete lines as in the previous case, we must substitute an effective μ chosen at some suitable average energy taken in the energy intervals above and below the edge. The intervals we have chosen are symmetrical about the edge and extend from 160 to 240 eV on either side of it. Thus at the average energy of 200 eV from the edge $\mu_+ = 925 \text{ cm}^2/\text{gm}$ and $\mu_- = 140 \text{ cm}^2/\text{gm}$, μ_1/μ_2 for the overlying absorber, which is the non-calcium portion of the bone, was taken to have the same functional dependence on energy as does H_2O . Hence, $\mu_1/\mu_2 = 1.31$.

Sample Preparation

The primary parameter which one may vary in preparation of a sample is its thickness. As we have shown in I, there is an optimum thickness of the sample which will minimize the relative statistical error on σ . We may, in general, write the expression for the square of the fractional standard deviation, δ_1^2 , as:

$$\delta_1^2 = \left(\frac{\Delta\sigma_1}{\sigma_1} \right)^2 = \frac{1}{\sigma_1^2} \sum_{j=1}^N (\mu_{1j}^{-1})^2 \exp \left\{ \sum_{k=1}^N \mu_{jk} \sigma_{jk} \right\}, \quad i = 1, N \quad (11)$$

For the general solution to the problem, we must find the minimum of δ_1^2 of interest with respect to the thickness, L , of the sample. Sample thickness is proportional to the sum of all σ_i in its composition. Furthermore, $\sigma_i/\sigma_j = \eta_{ij}$ since relative composition of the sample does not vary as a function of thickness. Hence, the equation $d\delta_1^2/dL = 0$ must be solved.

Because of the constraints on composition, the equation

$$\frac{d\delta_1^2}{d\sigma_1} = 0 \quad (12)$$

will have the same solution, $\sigma_{i \text{ opt}}$.

$$\begin{aligned} \frac{d\delta_1^2}{d\sigma_1} = & \frac{-2}{\sigma_1^3} \sum_{j=1}^N (\mu_{1j}^{-1})^2 \exp \left\{ \sum_{k=1}^N \mu_{jk} \sigma_{jk} \right\} \\ & + \frac{1}{\sigma_1^2} \sum_{j=1}^N (\mu_{1j}^{-1})^2 \exp \left\{ \sum_{k=1}^N \mu_{jk} \sigma_{jk} \right\} \sum_{m=1}^N \mu_{jm} \eta_{m1} = 0 \end{aligned} \quad (13)$$

Solving this for $\sigma_{i \text{ opt}}$ yields

$$\sigma_{i \text{ opt}} = \frac{2 \sum_{j=1}^N (\mu_{1j}^{-1})^2 \exp \left\{ \sum_{k=1}^N \mu_{jk} \sigma_{jk} \right\}}{\sum_{j=1}^N \sum_{m=1}^N (\mu_{1j}^{-1})^2 \mu_{jm} \eta_{m1} \exp \left\{ \sum_{k=1}^N \mu_{jk} \sigma_{jk} \right\}} \quad (14)$$

The specific case of absorption-edge scanning of thin slices of bone can be evaluated very simply using Eq. (14). First, however, there are simplifying assumptions which can be made which do not materially affect the resulting calculation $\sigma_{i \text{ opt}}$ in an important specific case.

The following assumptions are made: a) The absorption coefficient of the overlying absorber (non-calcium portion of bone) is the same in the energy segments above and below the calcium absorption edge, b) the amount of calcium in the bone approaches zero. This second assumption is necessary since this is the operating condition at which we wish to be most sensitive to calcium. Under these conditions, Eq. (14) reduces to

$$\sigma_{\text{opt}} \approx \frac{2}{\mu_2} \quad (15)$$

where σ_{opt} is the optimum amount of non-calcium portion of bone in g/cm^2 to detect small amounts of calcium in bone, and μ_2 is the absorption coefficient of the non-calcium portion of bone. We may take the μ_2 equivalent to that of H_2O .⁸ Thus, $\sigma_{\text{opt}} = .036 \text{ g/cm}^2$ (.014") at the energy of the K absorption edge of calcium, 4.04 keV, for maximum detectability of small amounts of calcium in bone.

We have, in practice, made thinner sections than this since the amount of calcium in bone causes appreciable absorption. To prepare a sample, dry bone was sectioned in slices of approximately .050" thick. It was then lapped flat and mounted in wax on a flat bed. The section was then ground to specified thickness on a surface grinder. After grinding, the bone was transferred to .0005" Kapton foil and inserted in the apparatus shown in Fig. 8. In more routine use, it is anticipated that specialized microtome procedures can be developed.

Standardization

Transmission absorption edge scanning in "good geometry" can be easily standardized provided the mass absorption coefficients are well known. Since the beam is tightly collimated before and after transmission through the sample and since we are working entirely within an evacuated test chamber, the correction for detection of scattered radiation is negligible compared to the statistical accuracy of the measured result. Inspection of Eq. (10), however, shows that the systematic error in determination of the elemental distribution under study is directly proportional to the error in the mass absorption coefficients in the denominator. Hence, these parameters contribute to the systematic error in this determination. Measurements of mass absorption coefficients are few and calculations suggest an accuracy of approximately 10%.⁷ The direct measurement of calcium, Eq. (10), would therefore have a total precision of only 10%. A comparative measurement using samples or standards of known calcium quantities may thus be preferred.

Standardized samples were fabricated by absorbing known amounts of solutions of CaCl_2 on filter paper in regions of known area and then drying the spot formed. The uniformity of distribution of the calcium was checked by making transmission measurements in several places on the sample. Uniformity of distribution was within statistical accuracy of the measurements recorded in table I. The estimated accuracy in the preparation of these standards was $\pm 10\%$. At this time, the accuracy of the method is limited by the lack of precision of the method used to fabricate standards. We are continuing the search for better standard materials and ultimately hope to lessen the systematic error of the technique.

Table I shows the results of the standardization measurements. Column I contains the calcium concentrations of the standards as measured by physical methods

described above. Column II is the $\Delta\mu/\sigma_{\text{Ca}}$, i.e. the numerator of Eq. (10). A linear regression analysis was performed on this data with the origin as a constrained point. The result of this analysis yields $\Delta\mu = 812 \pm 43$ with a χ^2 of .75 per degree of freedom. We have calculated $\Delta\mu$ from the values of mass absorption coefficients given in Storm and Israel.⁷ Corrections were made for the position of the energy windows above and below the edge. Because of the finite energy resolution of the Si(Li) detector, we cannot position these windows exactly at the energy of the K-edge. Hence, the value of the mass absorption coefficient changes from that which one observes in close proximity to the K-edge discontinuity. From this analysis, we have calculated $\Delta\mu = 785 \text{ cm}^2/\text{g}$. This is within 3.4% of the value observed in the calibration procedure.

Calibration may also be achieved with elements near those of interest. Potassium is close to both calcium and phosphorous and has been dried onto filter paper from a salt solution. This weight may be easily standardized using neutron activation and detection of the 1.52 MeV γ -ray from ^{42}K . An initial trial of this method is encouraging.

Bone Scan

Figure 11 shows the result of a typical scan of the elemental calcium distribution in a thin section of rat femur. The diagram in the center of the graph shows schematically the bone slice and indicates the position at which the scan is made. The abscissa is calibrated in thousandths of an inch and the ordinate is in $\sigma_{\text{Ca}}\Delta\mu$. The thickness of the slice is .004". The cortex and marrow regions of the bone are clearly defined. Typical statistical errors on the measurement are approximately 3%. The systematic errors in this situation were discussed in the previous section.

Although we have concentrated on calcium, the prospect of extending this technique to phosphorous is very bright. The energy of the K-edge of phosphorous is sufficiently high (2.142 keV) that window absorption does not make the technique impractical. Hence, the measurement of calcium to phosphorous ratios in small bone samples becomes feasible to a probable statistical accuracy of 3%. If the need for a larger scale program arises, we hope to construct an automated system similar to the one described in Part I of this paper, which will obviate the need of a full-time multichannel analyzer and eliminate the time-consuming off-line analysis of the data.

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Table I

Calibration of Calcium Absorption Edge Measurement

Sample number	<u>I</u>	<u>II</u>
	Weight calcium in micrograms per cm ² from physical measurement	$\sigma\Delta u$, uncalibrated x-ray measurement
1A	765 $\pm$ 80	.564 $\pm$.028
1B	752 $\pm$ 80	.658 $\pm$.026
2A	428 $\pm$ 40	.341 $\pm$.024
2B	462 $\pm$ 50	.411 $\pm$.023
3A	187 $\pm$ 20	.113 $\pm$.021
3B	177 $\pm$ 20	.126 $\pm$.021

Calculated $\Delta u = 785 \text{ cm}^2/\text{gm}$

Regression analysis $\Delta u = 812 \pm 43 \text{ cm}^2/\text{gm}$

χ^2 per degree of freedom from regression analysis = .75

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Figure Captions

- Fig. 1 Physical geometry of the scanning system. Note the change to a secondary fluorescence system from the primary target shown in Ref. 2.
- Fig. 2 Incident x-ray energy spectrum. The peaks identified are the Ba K_{α} (32.06 keV), Ba K_{β} (36.54 keV), Ce K_{α} (34.56 keV), and Ce K_{β} (39.43 keV).
- Fig. 3 Block diagram of the electronics necessary to calculate σ_I . This has been modified from Ref. 2 by changing the function of the divider to a more general analogue function generator described in the text.
- Fig. 4 Mass absorption coefficients of iodine and compact bone. The values at the Ba and Ce K_{α} lines are indicated by the tic-marks on each curve.
- Fig. 5 (a) Variation of output voltage from electronics as a function of absorber CaCl_2 present in beam. $m = \mu_2/\mu_1 = 1.00$ (uncorrected case).
- Fig. 5 (b) Variation of output voltage from electronics as a function of iodine present in beam for several amounts of overlying absorber CaCl_2 . $m = \mu_2/\mu_1 = .81$ (corrected case).
- Fig. 6 Response of the system to a Picker thyroid phantom filled with 2 mg/cm² iodine in the thick lobe. In the upper right-hand corner is a square of aluminum .250" thick. Three values of μ_1/μ_2 are shown: a) $\mu_1/\mu_2 = 1.00$, b) $\mu_1/\mu_2 = 1.15$, c) $\mu_1/\mu_2 = 1.16$.
- Fig. 7 Lateral iodine absorption edge transmission scan of the human thyroid. (a) shows the area scanned, (b) shows the iodine distribution in the thyroid, (c) shows simple transmission scan taken at x-ray energies below the iodine K-edge, (d) absorption edge transmission scan of patient in supine position (bottom) and simple transmission for x-rays below the iodine K-edge (top).
- Fig. 8 Schematic diagram of physical apparatus used for scanning thin slices of rat femur in vitro.
- Fig. 9 Energy spectrum of x-ray incident on the sample. The large peak at left is due to characteristic L x-rays from the rhodium target in the EG-50 tube.
- Fig. 10 Energy spectrum of the x-ray beam transmitted through a typical slice of rat bone .004" thick.
- Fig. 11 Calcium absorption edge scan of a thin (.004") slice of rat femur. The abscissa is distance in thousandths of an inch and the ordinate is $\sigma_{\text{Ca}}\Delta\mu$, dimensionless.

0 6" 12"

APPROXIMATE SCALE

FIGURE 1

FIGURE 2

BLOCK DIAGRAM OF ELECTRONICS

FIGURE 3

FIGURE 4

FIGURE 5 a

FIGURE 5b

FIGURE 6

FIGURE 7

FIGURE 7a

FIGURE 8

FIGURE 9

FIGURE 10

FIGURE 11