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DETERMINATION OF VERTEBRAL BONE MINERAL MASS BY
TRANSMISSION MEASUREMENTS*

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ABSTRACT

A dichromatic photon attenuation technique for the precise determination of lumbar vertebral bone mineral mass in vivo has been developed. A 100 mCi source of ^{153}Gd was used as a source of 44 keV and 100 keV photons. The collimation of the NaI(Tl) detection defined a 2.5 cm by 1 cm beam at the vertebra. The average of three scans at 2.5 cm intervals along the vertebral column was determined. The lower edge of the beam of the first scan was positioned coincident with the transtuberular plane and the position of the scanner relative to the top of the patient's head was recorded for repositioning purposes. No radiographs were taken. The total radiation exposure was measured with TLD for the three scans and was 2 mR to the dorsal surface of the trunk. The measurement took 25 minutes.

The precision of 7 measurements in vivo over a period of 3 months was 5%. The measurement of 8 excised vertebral columns under 15 cm of water was compared with their ashed mass per unit length (400°C for 48 hours). The ashed mass could be estimated from the scan measurements with a standard error of 3%. These experiments suggest, if the measurement system is properly calibrated, the dichromatic photon attenuation technique provides an accurate and precise measurement of vertebral bone mineral mass.

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INTRODUCTION

A dichromatic photon attenuation technique for the precise and accurate determination of lumbar vertebral bone mineral mass in vivo was developed because changes in the vertebrae have been assumed to larger and happen sooner than in the rest of the skeleton. Roos¹ described a method using the radiations from ¹³⁷Cs and ²⁴¹Am for which the gamma-ray beam at the vertebra was circular with a diameter of at least 2 cm. The technique developed utilizes 100 Ci of ¹⁵³Gd^{9,2}, a newly developed source,* which emits two essentially monoenergetic photon beams of 44 keV and 100 keV. The NaI(Tl) scintillation detector separates the radiation into two photopeaks. The 44 keV photopeak is the Eu K x-rays and the 100 keV photopeak is two gamma-rays of 97 keV and 103 keV. The collimation defined a beam of 2.5 cm by 1 cm at the vertebra. The length of the slot was aligned with the axis of the vertebral column. In order to reduce the day-to-day variation caused by repositioning errors, the average of three scans along the lumbar vertebral column at 2.5 cm intervals was determined.

METHOD

The bone mineral mass along any path in the body can be estimated from the transmission of two monoenergetic photon beams along that path using Equation (1).

$$\text{Mass}_{\text{BM}} = C (A - \ln I(44 \text{ keV}) + R \ln I(100 \text{ keV})) \quad (1)$$

$$C = \frac{\mu_{\text{ST}}(100 \text{ keV})}{\mu_{\text{BM}}(43 \text{ keV}) \mu_{\text{ST}}(100 \text{ keV}) - \mu_{\text{BM}}(100 \text{ keV}) \mu_{\text{ST}}(43 \text{ keV})}$$

$$A = \log I_0(44 \text{ keV}) - R \log I_0(100 \text{ keV})$$

$$R = \mu_{\text{ST}}(44 \text{ keV}) / \mu_{\text{ST}}(100 \text{ keV})$$

the derivation of Equation (1) is described in detail in the appendix. This particular formula was chosen to permit easy and accurate extraction of the bone mineral mass information from the data. The use of a baseline, A, eliminates the need to determine the unattenuated intensities of the two x-ray beams, which is required in some dichromatic attenuation techniques.³ The ratio of mass attenuation coefficients for soft tissue, R, can be accurately and easily determined from the measurement of the mass attenuation for each beam in water. For the ¹⁵³Gd beams, R determined from compilations⁴ for water is 1.536 compared with the value for lean muscle which is 1.541. The value of K was determined from an ash study in the same manner single monoenergetic photon technique was calibrated.⁵ The ash study also estimated the magnitude of various errors, particularly the error caused by the large photon beam.

* Obtained from Isotopes Development Section, Oak Ridge National Laboratory, Oak Ridge, Tennessee

For the measurement the patient layed on a table. The ^{153}Gd source was 38 cm below the table top rigidly coupled to a 5 cm x 1.3 cm NaI(Tl) scintillation detector, 34 cm above the table top. In normal operation the apparatus was moved across the torso by a screw drive.

For the first scan the position of the lower edge of beam was coincident with the transtubercular plane. Three scans cephalic along the lumbar vertebral column at 2.5 cm intervals were performed. During the first set of measurements the position of the scanner relative to the top of the patient's head was recorded to permit repositioning of the patient when the measurements were repeated. No radiographs were used to reposition the patient.

The electronic components of the equipment are shown in Figure 1. The system was assembled from modules built to the Nuclear Instrument Module (NIM) standards. The two photopeaks from ^{153}Gd were separated by a dual channel analyzer. The analyzer settings for the lower energy photopeak (44 keV) were a lower level of 28 keV with a window of 32 keV, and for the higher energy photopeak (100 keV) a lower level of 91 keV with a window of 31 keV. Both of these settings included essentially all of their respective photopeaks of the pulse height spectrum shown in Figure 2 while avoiding the small photopeak due to the 69 keV gamma-ray.

The mass attenuation coefficients of aluminum for the 100 keV photopeak were independent of window size. The mass attenuation coefficients for the 44 keV decreased 0.06% per 1 keV increase in window for windows centered on the photopeak. This dependence of the measured attenuation coefficient on window was probably caused by the poly-energetic character of the Eu K x-rays ($E_K = 41$ keV and $E_K = 48$ keV).

The number of count from each analyzer were counted for a fixed time, typically 10 seconds, determined by the timer. The data was then recorded on a magnetic tape. Other digital information was recorded on the magnetic tape by setting the thumb switches of the parametric recorder. The magnetic tape was transferred to the UNIVAC 1108 computer; for which computer programs were written to estimate the bone mineral mass.

The entrance exposure during a lumbar vertebral mass measurement was measured with TLD and was less than the sensitivity of the method (3 mR). The exposure rate from a 50 mCi source of ^{153}Gd at the table top was 3.5 mR/hr. The measurements took typically less than one half hour.

RESULTS

The method was calibrated and the accuracy was evaluated from the measurement of 8 excised vertebral columns under 15 cm of water. The columns were ashed at 400°C for 48 hours. The majority of the processes were removed for the vertebrae. The ash mass per unit length is plotted as a function of the absorptiometric measurement in Figure 3. The standard error of estimate from a linear regression was 3% indicating the absorptiometric measurement was a good estimator of vertebral bone mineral mass. The large value of the intercept in the regression equation was the effect of a large beam.

Figure 1
Electronic components of the vertebral bone
mineral mass analyzer.

Figure 2
Pulse height spectrum for ^{153}Gd . Window = 2.6 keV

Figure 3
Regression of the absorptiometric estimate of bone mineral mass and the actual ashed vertebral bone mineral mass.

The variation of the vertebral bone mineral mass along a lumbar spine that was in a torso phantom of Mix D⁸ was estimated by scanning the phantom at 0.5 cm intervals from L3 to L5. The results are illustrated in Figure 4. All values are within 3% on the mean value.

A precision of 5% for measurements in vivo was determined for seven measurements on one subject made at 0.5 cm scan intervals (the distance between point measurement). Each measurement took about 50 minutes. The precision on excised columns under water was unchanged if the scan intervals were 1 cm. This permitted a reduction of the measurement time to 25 minutes. With this scan interval the precision for ten measurements each on five subjects averaged 13%. An inspection of the results suggested the major source of error was the determination of the baseline, A, because some scans were started or ended too close to the vertebrae. The precision of one subject was reduced from 17% to 7% by correction of this error. A prospective study is underway to evaluate the precision at 0.5 cm scan intervals with particular consideration given to the determination of the baseline.

ANALYSIS OF ERROR

The largest error of this absorptiometric measurement of vertebral bone mineral mass was caused by the variation of the determination of the baseline, A. One can expect, if finer intervals and more measurements outside of the bone are made, this error can be reduced to values determined by Poisson statistics of counting photons. In principle this type of variation can be reduced to any value by increasing the source activity or increasing the measurement time. For the subject measurements reported this variation was less than 5%. The ultimate limitations on the precision are the ability to measure the same portion of the skeleton and patient movement.

The accuracy of the method was determined by:

- (1) the size of the photon beam
- (2) the variation of the soft tissue composition across the vertebrae
- (3) the monochromaticity of the beams
- (4) the amount of scatter detected

Contrary to one's expectation the large beam (0.5 cm by 2.5 cm) cause a rather small error of less than 3% for technique calibrated by an ash study.

The soft tissues, which for this technique include the organic component of bone, vary in composition. The most significant factor is the larger lipid content of adipose tissue. If the composition of soft tissue has changed, the value of the ratio of mass attenuation coefficients for soft tissue, R, is incorrect for that path of the body. The magnitude of this error was estimated by comparing the baseline values on each side of the vertebrae and the variation in the baseline values. The analysis suggested the variation of soft tissue composition outside the vertebrae should cause an insignificant error, less than 3%. The magnitude of the error caused by adipose tissue inside the vertebral bodies can be as large

Figure 4
The measurement of vertebral bone mineral mass along
the lumbar spine of a phantom.

as 8%.⁷ This error would be reduced, if the calibration formula were obtained from an ash study made on fresh bones. The effects on the monochromaticity of the beams and the scatter also will have little effect, if the measurement is compared with ashed mass of vertebral column.

CONCLUSIONS

The absorptiometric determination of vertebral bone mineral mass using ¹⁵³Gd provided an accurate technique, if the measurement was calibrated by an ash study. The precision of the technique was 5% to 7%. The small variation (3%) of vertebral bone mineral mass along the lumbar spine suggest the variation caused by the processes and gaps from intervertebral discs are defocused with the large photon beam. The maximum change of vertebral bone mineral mass between L3 and L5 was 5% per cm, which implies if the patient can be repositioned with an accuracy of 0.5 cm, the error of repositioning can be reduced to less than 3%.

The purpose of this research was to develop a precise method to evaluate changes of bone mineral mass where each individual could be his own control. For this type of application the concern would be to control the factors that affect accuracy. The presence of adipose tissue inside the vertebral body causes the largest error with a maximum of 8%. During a study this error should not change and the precision of the measurement of vertebral bone mineral mass would be unaffected. The effects caused by crushed vertebrae being examined, the more mineral mass might enter the scan path resulting in apparent increase of bone mineral mass.

Appendix

The bone mineral mass along a path in the body can be estimated from the transmission of two mono-energetic photon beams along the line using the following equation.

$$\text{mass}_{\text{BM}} = \frac{\mu_{\text{ST}}(100 \text{ KeV}) \log \frac{I_0(44 \text{ KeV})}{I(44 \text{ KeV})} - \mu_{\text{ST}}(44 \text{ KeV}) \log \frac{I_0(100 \text{ KeV})}{I(100 \text{ KeV})}}{\mu_{\text{BM}}(44 \text{ KeV}) \mu_{\text{ST}}(100 \text{ KeV}) - \mu_{\text{ST}}(44 \text{ KeV}) \mu_{\text{BM}}(100 \text{ KeV})} \quad (\text{A1})$$

μ = mass attenuation coefficient

m_{BM} = mass of bone mineral along the path in the bone having units of g/cm^2 .

ST = subscript referring to soft tissue

BM = subscript referring to bone mineral

I = transmitted intensity

I_0 = unattenuated intensity

The technique developed scans the ^{153}Gd beam across the vertebral column integrating Equation (A1) across the vertebrae to obtain the mass in section of the column.

Equation (A1) has been rearranged to permit easily and accurate extraction of the information from the data.

$$\text{mass}_{\text{BM}} = C \left[\log I_0(44 \text{ KeV}) - R \log I_0(100 \text{ KeV}) - \log I(44 \text{ KeV}) + R \log I(100 \text{ KeV}) \right] \quad (\text{A2})$$

where

$$C = \frac{\mu_{\text{ST}}(44 \text{ KeV})}{\mu_{\text{BM}}(44 \text{ KeV}) \mu_{\text{ST}}(100 \text{ KeV}) - \mu_{\text{BM}}(100 \text{ KeV}) \mu_{\text{ST}}(44 \text{ KeV})}$$

$$R = \mu_{\text{ST}}(44 \text{ KeV}) / \mu_{\text{ST}}(100 \text{ KeV})$$

In the situation with no bone mineral mass in the photon beam path the terms of Equation (A2) that are functions of the unattenuated intensity can be determined from Equation (A3) which was derived from Equation (A2) by setting m_{BM} to zero.

$$A = \log I_0(44 \text{ KeV}) - R \log I_0(100 \text{ KeV}) = (\log I(44 \text{ KeV}) - R \log I(100 \text{ KeV})) \Big|_{\text{mass}_{\text{BM}}=0} \quad (\text{A3})$$

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