

Precise Measurement of Vertebral Mineral Content Using Computed Tomography

Christopher E. Cann and Harry K. Genant

Abstract: A precise method of measuring vertebral mineral content by computed tomography has been developed. Calibration phantoms scanned with patients are used to provide corrections for machine drifts. Advanced software manipulation of the display images is used to reposition precisely to the mid-portion of each vertebral body. A long-term precision of 2.8% for mineral content has been obtained in excised vertebrae in a phantom simulating the human torso. **Index Terms:** Spine, mineral content—Computed tomography—Osteoporosis—Phantoms.

Quantitative bone mass measurements are widely used in the assessment of skeletal status in metabolic bone disease. Considerable effort has been expended to optimize measurements in the peripheral bones, originally using the linear photon absorptiometry method of Cameron and Sorenson (1) and more recently with the use of gamma-ray and X-ray computed tomography (CT) (2-6). The CT methods have been useful in separating cortical from trabecular bone in the radius. This ability has enhanced the sensitivity to metabolic changes by measuring trabecular bone with its high turnover rate independently of low turnover cortical bone. However, the measurement of radial mineral does not correlate well with mineral content in the axial skeleton in health or disease (7,8). Clinically, the most significant problems of bone loss occur in the vertebrae and femoral neck. Thus, the capability of direct measurement of bone mass in these regions would be a significant advance in the clinical management of metabolic bone disease.

The usefulness of CT for measuring bone mineral in the vertebrae lies in its ability to quantitatively image a thin transverse slice through the abdomen. With this technique, one has the ability to spatially separate cortical and cancellous bone in the spine. Thus, one can measure changes in bone mineral content (a) noninvasively, (b) in cancellous bone, where measurement of changes is more sensitive than in cortical bone, and (c) at the site of clinical

involvement, in the vertebrae. The potential widespread use of such a technique with the large number of whole body CT scanners available adds to its clinical importance.

The use of CT for measuring vertebral mineral has been studied by several investigators (5,9-12). There are several major technical problems that must be solved in order to make this a useful technique. The diagnostic capability of vertebral CT in metabolic bone disease depends on the accuracy of measurement of the bone mineral contained within the vertebral body. This accuracy is influenced by two factors: the accuracy of the calculated attenuation coefficients (CT numbers) and the model assumed for vertebral composition. There are significant inaccuracies in the calculated CT numbers for most scanners, primarily due to hardening of the polychromatic X-ray spectrum as it passes through the body. Most scanners incorporate a first order correction for this effect based on hardening in water. Second order (postprocessing) corrections to remove artifacts due to bone may improve image quality but may also lead to other inaccuracies in the calculated CT numbers (13). An alternative method for a beam-hardening correction is the use of dual-energy CT scans (14). With this technique, a system is calibrated using two known standards such as water or Lucite (soft tissue equivalent) and K_2HPO_4 or aluminum (mineral equivalent). Projection data obtained at two different scan energies are combined to produce two new sets of data that correspond to a "soft tissue" image and a "mineral" image. These reconstructed images are free of any beam-hardening artifacts and can be used to obtain accurate CT numbers at any effective scan energy.

From the Department of Radiology (M-398), University of California, San Francisco, California 94143. Address reprint requests to Dr. Cann.

The highly accurate determination of spinal cancellous mineral content is difficult with a single energy CT scan, even if the CT numbers are accurate. The CT number represents an average of the attenuation coefficients of all the materials present in the vertebral body, including mineral, collagen, soft tissue, and fat. For a single energy CT scan, a two component mixture is assumed (i.e., soft tissue and mineral). The substantial error introduced by variable (up to 20%) fat content can only be eliminated by using scans at multiple energies and taking advantage of the energy dependence of μ for bone mineral. While this technique has been demonstrated in phantoms (5), the accuracy *in vivo* must still be determined. As yet this has not been reported.

High measurement precision is necessary to provide sensitivity for small mineral changes in serial studies to assess the efficacy of treatment regimens or the progression of disease processes. The technical problems that reduce the precision of vertebral mineral measurement are in general different from those affecting accuracy. The two main problems limiting precision in our experience are scanner-related changes with time and patient repositioning between studies. Previous studies (15) have shown that there is significant drift in the measured CT number of mineral-equivalent calibration phantoms over both the long term (several months) and short term (minute to minute). These changes have been attributed to several factors including X-ray tube aging and heating, detector drifts, and software and hardware modifications made by the manufacturer. Our approach to this problem has been to develop correction methods that are generally applicable to all CT scanners rather than to a particular machine configuration. This will allow scanner-independent measurements to be made and results from various investigators to be directly compared.

The problem of repositioning to the same site in the vertebral body in repeat studies is a considerable one and is primarily responsible for the previous lack of success of spinal mineral measurements by CT. The central cancellous portion of the vertebral body is often inhomogeneous. Therefore, it is preferable to sample as large a volume of the body as possible to make mineral measurements. This central portion is usually on the order of 13 to 15 mm thick, with the outer regions becoming more dense near the endplates. With the use of a thick CT slice of 10 to 13 mm, a positioning error of only 2 to 3 mm will cause measurement errors of 3 to 4% in mineral content on the average, with errors up to 10% possible. A thinner (5 mm) slice can be used to eliminate inclusion of the endplate regions, but then inhomogeneities within the body region may cause errors with the use of this small volume. The ideal case appears to be a slice approximately 10 mm thick positioned within 1 to 1.5 mm of the midplane

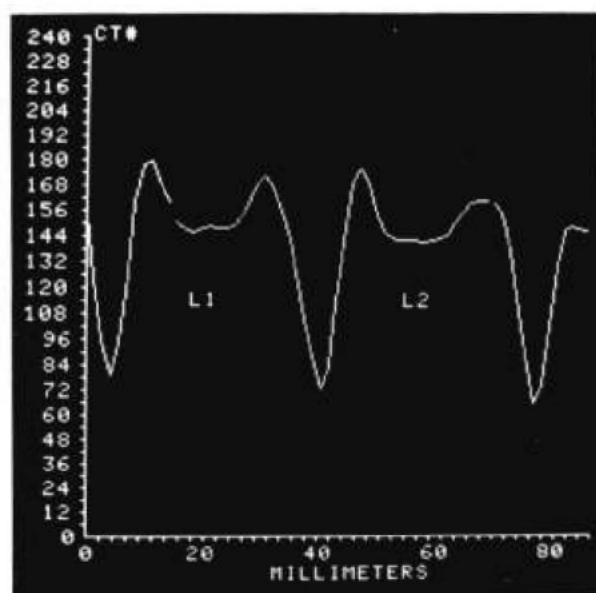


FIG. 1. Distribution of CT numbers (500 scale) through the first and second lumbar vertebrae of a normal adult male. The average CT number of the midportion of 64 consecutive 1.35 mm thick synthesized images is plotted against the longitudinal position. Original data were obtained as 18 contiguous 5 mm thick slices. The vertebral endplates (high numbers), midportion, and disc spaces are clearly visible.

of the vertebral body. Phantom studies (15) have shown that positioning errors of this magnitude cause measurement errors in mineral of only 1 to 2% because of the existence of a relative plateau in the CT number in the midportion of the vertebral body (Fig. 1). Another factor in precise positioning is the angulation of the spine relative to the CT scanner gantry. Average angulation in our experience is 5 to 7°. Differences of angulation between studies average 2 to 3°, although up to 8° of variation has been noted. Therefore, it is necessary to be able to correct for this angulation as well.

We have developed techniques for vertebral mineral determination by CT that have significantly reduced the errors due to machine drift and repositioning. High precision for mineral measurement over long time periods in excised human vertebral specimens is demonstrated, and the clinical applicability of these techniques is discussed.

METHOD

The development of techniques for vertebral mineral determination used in this study was accomplished using excised human vertebrae and phantoms constructed in our laboratory. Vertebral specimens were isolated fresh from cadaveric spines and were placed in a phantom containing calibration materials (Fig. 2). Each specimen was sealed in a cylindrical container that was then inserted into a larger phantom that simulated the

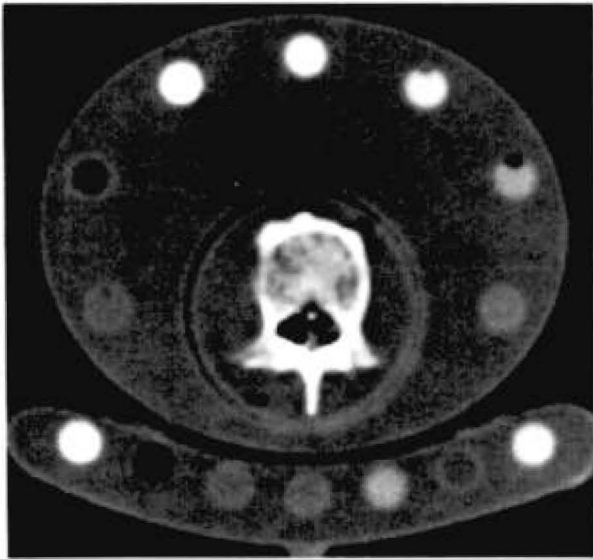


FIG. 2. CT scan of an excised vertebral specimen in a phantom simulating the human torso. Cylindrical tubes contain calibration solutions for mineral, soft tissue, and fat. Crescent shaped phantom underneath torso phantom is used for calibration purposes in all patient studies

human torso. This arrangement allows removal and reinsertion of the specimens for repeat studies with a positioning error of 0.2 to 0.3 mm relative to the phantom.

Specimens were scanned using our research scanner, which is a highly modified version of a GE CT/T 7800. A reconstruction diameter of 35 cm and pixel size of 1.35 mm were used. Tube potential was 80 kVp. Five millimeter thick slices were obtained every 5 mm in a rapid scanning mode. A series of 24 scans covering three contiguous lumbar vertebrae were obtained in 84 sec of scanning time. The torso phantom was placed on top of our standard crescent shaped phantom containing solutions of K_2HPO_4 (mineral equivalent), ethanol (fat), glycerol and

water (soft tissue) for calibration. This was done to simulate the standard clinical situation in which our patients lie on this phantom for calibration purposes.

Analysis for mineral content of the specimens by CT was done using a specially programmed off-line computer. The 24 contiguous reconstructed axial transverse slices obtained in each series of scans were used to form a volume containing the vertebrae. The midplane of each vertebral body was defined on a sagittal or coronal image by using the endplates and intervertebral disc spaces as anatomical landmarks. A 9.5 mm (seven pixels) thick true axial transverse slice angled perpendicular to the axis of the spinal column was synthesized about this midplane (Fig. 3). The plateau region within the vertebral body is about 15 mm thick in most cases. When the axial transverse slices are interpolated, this provides 11 lines of data, each 1.35 mm thick (Fig. 4). Restricting the thickness of a synthesized image to seven lines of data (9.5 mm) minimizes any contribution from an axial slice containing vertebral endplate. Z-axis position of the slice was varied ± 1 pixel and cortical structures in these synthesized images were matched to those in previous studies to provide the most precise repositioning. The average CT number within a 3.7 cm^2 elliptical region of interest (ROI) positioned approximately 2 mm from the cortical shell and 4 mm from the anterior wall of the spinal canal was obtained in each vertebral body. This ROI covered 25 to 35% of the cancellous area of the vertebral body and included a 3.5 cm^3 volume of vertebral cancellous bone. The mean CT number of each body and the CT numbers of the calibration solutions contained in the crescent shaped phantom were used to calculate the mineral equivalent for each vertebra in terms of milligrams of K_2HPO_4 per milliliter.

A series of experiments were done to study the effects of positioning and machine variation on vertebral mineral measurements.

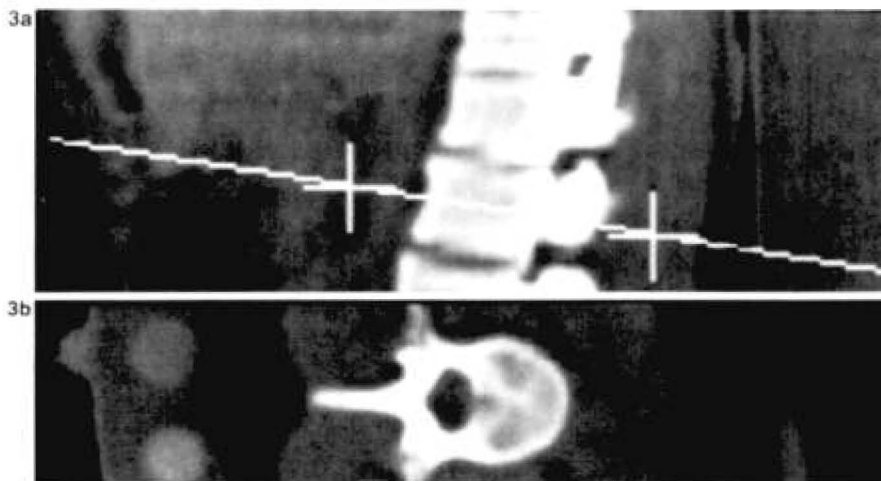


FIG. 3. a: Reformatted sagittal image constructed from 18 axial transverse scans (T12, L1, L2, L3). Cursor line defines midplane of vertebral body b: Synthesized true axial transverse image 9.5 mm (seven pixels) thick centered about cursor line in (a)

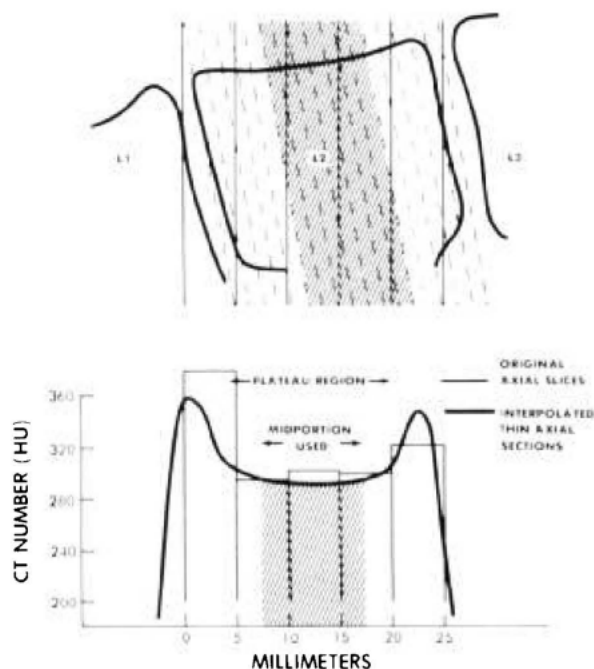


FIG. 4. Diagram of the manner in which synthesized images are obtained. **Top:** Solid lines indicate original 5 mm thick contiguous axial transverse slices; dashed lines are 1.35 mm (one pixel) thick reformatted true axial transverse images; shaded area indicates region of vertebral body used to synthesize 9.5 mm (seven pixels) thick slice. **Bottom:** CT number (HU: 1,000 scale) profile of L2 in a normal adult male indicates original scans and interpolated images. Shaded region corresponds to area in top diagram.

1. An excised spine containing three vertebrae, L2 to L4, was scanned in the phantom three separate times within a 2 hr period. The spine was moved 1.5 mm in the z-axis relative to the scanner beam after each series of scans. Data from this experiment were used to determine the variation in measured mineral content caused by errors in repositioning to the same volume in the vertebral body when using the data analysis techniques outlined.

2. The effect of positioning the elliptical ROI in the x-y plane was studied. The ROI was positioned within the synthesized axial transverse image for several cases, and the mean CT number was determined. The ROI was then moved one pixel and two pixels in each direction (x and y), and the mean CT number was obtained for each new position of the ROI. Sensitivity of the mean CT number to the ROI position in the x-y plane was determined using this set of data.

3. Another experiment was performed to further study the z-axis aspects of positioning reproducibility. Needles were inserted at a sharp angle into vertebrae and into water to provide a high contrast object that moved in the x-y plane as 9.5 mm thick images were synthesized serially at 0.5 mm intervals (i.e., with 9.0 mm overlap) along the z-axis.

The position of the needle in each synthesized slice was determined. These data provided a measure of the minimum number of distinct images that could be produced using these reconstruction techniques, and hence the minimum z-axis positioning capability.

4. Significant variation in the CT number of calibration solutions has been observed previously over the short term (minute to minute). An experiment was done to define the sources of this variation. The calibration phantom was scanned rapidly at the same level, 12 scans in 40 sec at 80 kVp to determine the scan-to-scan variation. A second set of data was then obtained by scanning the calibration phantom with a patient. Seventeen scans were obtained in 75 sec with scan table indexing of 5 mm (z-axis) between scans. These two sets of data were analyzed to determine the separate contributions to the observed variation in calibration CT numbers from (a) inherent machine variation from scan to scan and (b) the effect of varying bone and soft tissue in the patient at different locations along the spine.

5. A spine specimen was scanned in a phantom on two different GE CT/T 7800 scanners in our hospital. Similar scanning procedures were used, and data were analyzed as outlined. This experiment provided some measure of the dependence of mineral measurement on the machine used.

The data obtained in these five studies provided independent information about density and positioning reproducibility and the expected variation in the CT number of calibration materials.

6. To determine the long-term reproducibility of these scanning and data analysis techniques, a spine specimen was scanned several times over a period of months. The mineral equivalent in each vertebral body was determined for each set of scans and the variation calculated. This defined the overall precision of vertebral mineral measurement in serial studies with excised specimens in a human torso phantom.

RESULTS

The studies outlined above provide information about errors in repositioning and those due to scanner drift and give a measure of overall precision for the techniques being used. For the elliptical ROI in each vertebra, a change in position of the ROI of one pixel in any direction in the x-y plane produces an average change of 0.8 Hounsfield units (HU: 1,000 scale) in the mean CT number about an average of 240 HU for the spine specimen. Positioning of the ROI in serial studies is done with all synthesized images on the video display at the same time to minimize errors (Fig. 5). The error in ROI positioning could contribute up to 15% of the overall error in the determination of vertebral mineral.

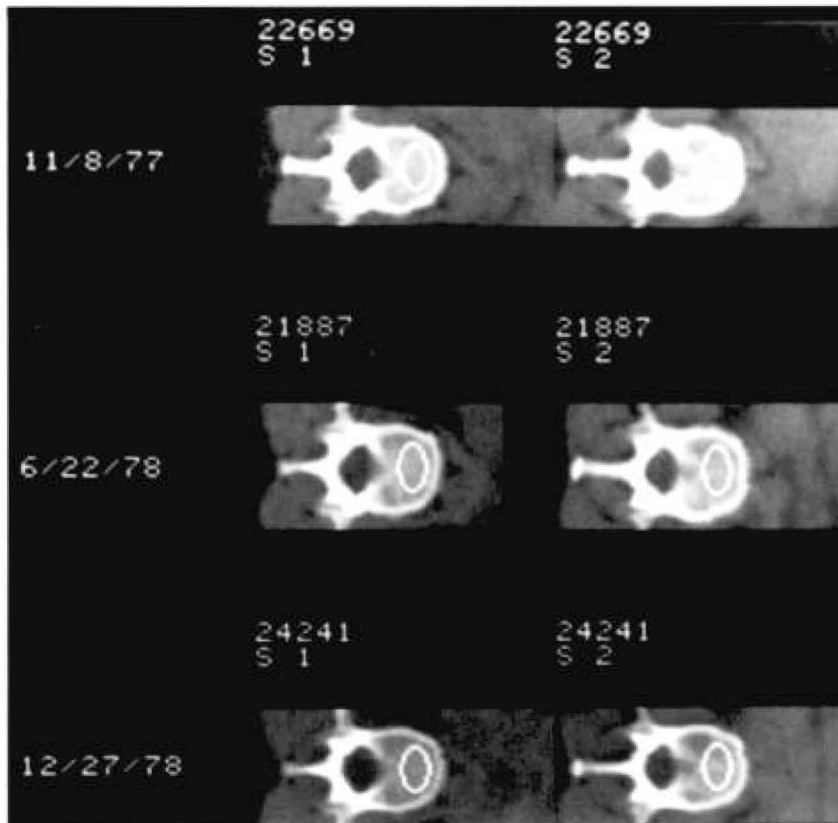


FIG. 5. Display of synthesized images of L1 (left) and L2 in one subject scanned three times over a 1 year period. Precise positioning is obtained using cortical structures in the synthesized images. Elliptical ROI is used to determine mean CT number of vertebral cancellous bone in each image.

Longitudinal axis positioning reproducibility for high contrast objects is determined using the needle study. An analysis of the number of distinct images versus z-axis position provides a resolvable separation of 1.35 mm, or approximately 1 pixel thickness. A similar longitudinal resolution is obtained with the excised vertebrae and with vertebrae *in vivo* by using high contrast cortical structures present in the synthesized image as positioning landmarks.

Short-term variability in the calculated mineral content is obtained from scanning the excised spine repeatedly and analyzing the data using the techniques outlined. The 9.5 mm synthesized images are obtained for each vertebra in each of several scans, and mean CT numbers from the vertebrae and calibration solutions are used to calculate the mineral content. The average deviation (coefficient of variation) in vertebral mineral content for the three vertebrae used in this experiment is 1.5%.

The short-term variation in CT number of the calibration solutions scanned rapidly at the same table index is 1.2 to 2.4 HU (SD from 12 scans) for solutions including alcohol, glycerol, water, and various concentrations of K_2HPO_4 . There is no dependence of variation in CT number on the material being scanned. Results from rapid sequential scans at 5 mm intervals through the lumbar spine are shown (Fig. 6). The variation obtained in standard

solutions when scanning through different levels in the body, up to 30 HU, is significantly larger than the stationary scan-to-scan machine drift of 1.2 to 2.4 HU. This appears to be due primarily to the different amounts of spectral hardening that occur when passing through either dense bone or less dense intervertebral disc spaces. To reduce the error from this source, calibration data are now obtained only from axial transverse slices containing the midportion of the vertebral body.

The results from the series of scans taken over a several month period are given in Table 1. The coefficient of variation (CV) for the CT number (attenuation coefficient) normalized to water (0 HU) at 80 kVp is 4.8 HU, or 0.3%. The CV for calculated mineral content normalized to calibration solutions is 2.8%. This estimate of precision includes both repositioning and machine drift errors.

Variation in the mineral content when determined on two different GE CT/T 7800 scanners is less than 2%, or less than the longer-term variation on one scanner.

The results from this series of experiments with excised vertebrae in a phantom simulating the human torso show that vertebral mineral content can now be measured precisely over the long term using CT. Approximately 1.5% of the 2.8% variation seen can be attributed to errors in repositioning

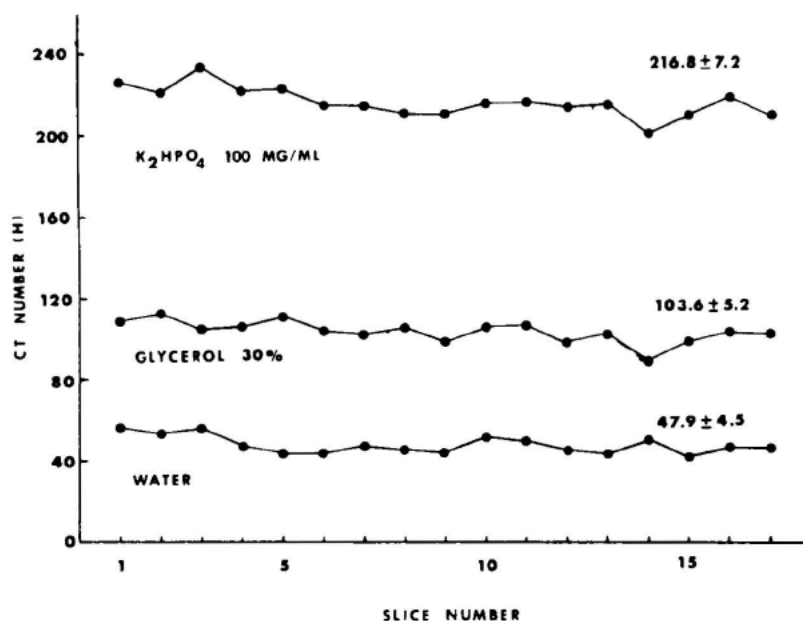


FIG. 6. Variation in CT number (HU) of calibration solutions in a phantom scanned with a patient. Seventeen scans were taken in 75 sec with 5 mm table indexing between scans.

to the same volume in the vertebral body. The remaining 1.3% error appears to be due to longer-term variations in the scanner.

DISCUSSION

The development of CT for vertebral mineral determination has been made difficult by technical problems affecting the precision and the accuracy of measurement. The major problems affecting precision have been machine drift and patient repositioning between studies. Experience with our scanners has demonstrated significant variation in the CT number of calibration solutions and water even from scan to scan. Detector responses are ex-

tremely uniform in this scanner and reference detectors outside the reconstruction circle are used to normalize for X-ray output variations. It appears that the observed short-term CT number variations are due to changes in beam quality as it passes through varying thicknesses of bone and soft tissue. Thus, scanning of a phantom before or after a patient does not correct for the short-term variability. The effects of this variation are minimized by using calibration data present in the same scans as the patient data used.

Long-term changes in the scanner appear to be related to such phenomena as X-ray tube aging and self-filtration, which produce changes in the effective energy of the scanner beam. Most CT scanners

TABLE 1. Variation in CT number and calculated mineral content with time in an excised spine (L2-L4)^a

Vertebra	CT number (HU)				Mineral content (mg K ₂ HPO ₄ /ml)			
	Scan 1	Scan 2	Scan 3	$\bar{X}$	Scan 1	Scan 2	Scan 3	$\bar{X}$
L2	200.6	196.9	192.8	196.8	138.3	129.5	134.8	134.2
L3	181.9	174.6	176.0	177.5	127.4	115.3	127.0	123.2
L4	191.0	196.4	182.0	189.8	133.4	129.0	134.6	132.3

Average deviation in CT number:

$$\frac{\sum_{i,j} (x_{i,j} - \bar{x}_i)}{9} = 3.6 \text{ HU.}$$

Average deviation in mineral content:

$$\sum_{i,j} \frac{(x_{i,j} - \bar{x}_i)}{\bar{x}_i} \times 100 = 2.8\%.$$

^a Three scans obtained over 2 months at 80 kVp.

are calibrated with air and water, and these calibrations are done frequently. The precision for a water or soft tissue phantom on a day-to-day basis is quite high. Precision for measurement of a material such as concentrated K_2HPO_4 simulating cortical bone is also high on any particular day. This material has a significantly different μ than water, however, and the energy dependence of μ for bone is much greater. Thus, while a shift in the energy spectrum of the X-ray tube will be compensated in soft tissue by the water calibration, its effect on bone mineral measurement will be quite marked. The average day-to-day variation in our scanners is 8 to 10 HU at 80 kVp for a vertebral bone standard, with longer-term drift of 30 to 40 HU. Extreme variations can occur with major changes such as a new X-ray tube or new reconstruction software provided by the manufacturer.

Perhaps of major significance is the fact that the long-term drifts we have observed are definitely a function of the mineral content. Standard solutions ranging from 50 to 625 mg/ml K_2HPO_4 , simulating the range from vertebral to cortical bone, exhibit drifts at both 80 kVp and 120 kVp which are dependent on K_2HPO_4 concentration. These changes are approximately the same, on a percentage basis, for all solutions and at both energies. Recently, Weissberger et al. (9) made the point that the precision in the CT number H is independent of the absolute value of H . This is the case for scans taken within a short time period, and our data indicate a precision of 1.5 to 2.5 HU for scans obtained within a 1 hr period. However, for clinical studies, the time period between scans is on the order of 6 to 12 months, and the large long-term drift contributes most to imprecision in these cases. Because of these drifts, we have concluded that all clinical cases for mineral determination must be scanned with a calibration phantom in the field.

The studies using phantoms indicate that repositioning to a specific location using special reconstruction software and visual cues can be obtained within 1.3 to 1.5 mm. Images are obtained by linear interpolation on a pixel-by-pixel basis to provide 3.5 lines of data for each original 5 mm thick axial slice. The scanner used for these studies generates an image that equally weights all volume elements within the 5 mm slice because of the square-wave shape of the beam profile at the level of the patient. Thus, each line of interpolated data is a weighted image of the attenuation values at each point as long as the true change in composition within the vertebrae is relatively monotonic.

A single thin (~ 5 mm) scan obtained through the midportion of a vertebral body is not sufficient to determine mineral content. In many cases of metabolic bone disease, the vertebrae may contain multiple foci of trabecular loss. Mineral determination using a thin slice that contains a focal lesion can

misrepresent the average vertebral mineral content. Conversely, a single 10 mm thick original axial transverse slice does not provide the same density resolution as two 5 mm thick slices. The longitudinal resolution for a 10 mm thick scan is half a slice thickness, or 5 mm. Only if a 10 mm thick scan can be positioned precisely to within 2 mm by external means should its use be considered.

A significant consideration for any patient study is the ease of performance and data analysis. Data are obtained from each subject in a short period of time: a single energy (80 kVp) study with localization takes approximately 15 to 20 min from the time the patient arrives to the end of data collection. Off-line computer processing of the data takes 1 hr. This relatively small investment of time for each patient study makes this technique suitable for clinical studies on a routine basis.

The techniques we have developed provide a precise method for measuring vertebral mineral in serial studies. A previous study by Weissberger et al. (9) addressed the sensitivity of mineral measurement in trabecular versus cortical bone and calculated that a change in cortical mineral content would provide a change in μ 8.6 times that of an equal percentage mineral change in the vertebral bone mixture due to the higher volumetric concentration of mineral in compact bone. The authors concluded that a cortical measurement should provide greater sensitivity to bone mineral changes than a measurement of purely trabecular bone. While this may be true based on purely physical parameters, the physiological differences between cortical and trabecular or axial and peripheral sites may override any physical disadvantages. Trabecular bone has been estimated to turn over at a rate eight times higher than cortical bone (16). This factor in itself would balance the decrease in physical sensitivity. The measuring site may have a significant effect on the overall sensitivity for mineral change as well. The weight-bearing axial skeleton may lose bone more rapidly than trabecular bone in the radius (17,18), indicating that a measurement of trabecular bone in the axial skeleton may provide better sensitivity for metabolic change (17,19).

In conclusion, precise measurement of vertebral mineral in serial studies has been obtained. Methods have been developed to minimize the errors due to machine drift and patient repositioning, and a positioning capability of ± 1.5 mm has been demonstrated using special software. The ability to synthesize a thick true axial transverse slice from multiple thin contiguous slices taken at any angle relative to the spine allows the use of a large ROI for mineral determination. This reduces the errors from inhomogeneities within the vertebral body and increases the precision for mineral measurement. Long-term precision with excised vertebral specimens in a phantom simulating a human torso is 2.8%

(CV). In serial studies in patients, an observed change of ± 2 CV, or 5.6%, would be interpreted as a significant change in vertebral mineral content.

ACKNOWLEDGMENT

The special image processing software used to do part of this work was provided by Dr. W. H. Berninger and W. M. Leue, General Electric Corporate Research and Development Center, Schenectady, New York. We thank Dr. Douglas P. Boyd for his contributions and Denise Nakano for her help in preparing the manuscript. Dr. Cann would like to acknowledge support from the Radiology Research and Education Foundation.

REFERENCES

1. Cameron JR, Sorenson J: Measurement of bone mineral *in vivo*: An improved method. *Science* 142:230-232, 1963
2. Rueggsegger P, Niederer P, Anliker M: An extension of classical bone mineral measurements. *Ann Biomed Eng* 2:194-205, 1974
3. Reich NE, Seidemann FE, Tubbs RR, MacIntyre WJ, Meaney TF, Alfidi RJ, Pepe RG: Determination of bone mineral content using CT scanning. *Am J Roentgenol* 127:593-594, 1976
4. Posner I, Griffiths HJ: Comparison of CT scanning with photon absorptiometric measurement of bone mineral content in the appendicular skeleton. *Invest Radiol* 12:542-544, 1977
5. Genant HK, Boyd D: Quantitative bone mineral analysis using dual energy computed tomography. *Invest Radiol* 12:545-551, 1977
6. Orphanoudakis SC, Jensen PS, Rauschkolb EN, Lang R, Rasmussen H: Bone mineral analysis using single energy computed tomography. *Invest Radiol* 14:122-130, 1979
7. Dalen N, Jacobson B: Bone mineral assay: choice of measuring sites. *Invest Radiol* 3:174-185, 1974
8. Wilson CR: Prediction of femoral neck and spine bone mineral content from the BMC of the radius or ulna and the relationship between bone strength and BMC. In: *Proceedings of the International Conference on Bone Mineral Measurement*, ed. by RB Mazess, Washington D.C., U.S. Department of Health, Education and Welfare Publication NIH-75-683, 1974, pp 51-59
9. Weissberger MA, Zamenhof RG, Aronow S, Neer RM: Computed tomography for the measurement of bone mineral in the human spine. *J Comput Assist Tomogr* 2:253-262, 1978
10. Bradley JG, Huang HK, Ledley RS: Evaluation of calcium concentration in bones from CT scans. *Radiology* 128:103-107, 1978
11. Laval-Jeantet M, Laval-Jeantet AM, Lamarque JL, Demoulin B: Evaluation de la mineralisation osseuse vertebrale par tomographie computerisee. *J Radiol* 60:87-93, 1979
12. Cann CE, Genant HK, Young DR: Comparison of vertebral and peripheral mineral losses in disuse osteoporosis in monkeys. *Radiology* 134:525-529, 1980
13. Joseph PM, Spital RD: A method for correcting bone induced artifacts in computed tomography scanners. *J Comput Assist Tomogr* 2:100-108, 1978
14. Alvarez RE, Macovski A: Energy selective reconstructions in X-ray computerized tomography. *Phys Med Biol* 21:733-744, 1976
15. Abols Y, Genant HK, Rosenfeld D, Boyd DP, Ettinger B, Gordan GS: Spinal bone mineral determination using computerized tomography in patients, controls and phantoms. In: *Proceedings of the Fourth International Conference on Bone Measurement*, Washington D.C., U.S. Government Printing Office, 1979
16. Report of the Task Group on Reference Man, ICRP Publ. 23, Oxford, Pergamon, 1975
17. Cann CE, Genant HK, Boyd DP: Precise measurement of vertebral density changes in serial studies using computed tomography. (Abstr) *Invest Radiol* 14:372, 1979
18. Schneider VS, Hinderling T, Vogel JM, Rambaut P: CT ¹²⁵I gamma-ray densitometry (CT scanner): Noninvasive measurement of trabecular bone mineral. *J Comput Assist Tomogr* 3:847-848, 1979
19. Cann CE, Genant HK, Boyd DP: Vertebral mineral determination by CT: Clinical results and specimen studies. Presented at 65th Scientific Assembly, Radiological Society of North America, Atlanta, Georgia, 1979