A GAMMA-RAY COMPUTED TOMOGRAPHY SCANNER FOR THE QUANTITATIVE MEASUREMENT OF BONE DENSITY

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ABSTRACT

A special purpose gamma-ray computed tomography scanner has been developed for the precise measurement of bone density in the distal forearm. Details of the scanner hardware and computer analysis technique are given. Suitable phantoms have been used to test the operation of the scanner, which has been used to measure trabecular and cortical bone density with a precision better than 1%.

Keywords: Bone, computed tomography, osteoporosis, densitometry

INTRODUCTION

Osteoporosis is a bone disease characterized by progressive loss of bone mass often resulting in fracture after minimal trauma. Fractures of the distal forearm, proximal femur and vertebrae are associated with the disease which affects 25% of the female population over the age of 65. Despite intensive study of this painful and debilitating disease over the past half century, there has been no significant improvement in treatment. The goal of the present study is to develop a non-invasive technique which can be used to monitor accurately the course of osteoporosis and the response to treatment.

While the normal ageing process in women results in a bone loss of 1% per year from 45 to 75 years of age, an annual loss of 2% or more may result in skeletal problems[1]. In order to assess the condition within clinically useful time scales it is necessary to measure bone content with a reproducibility of better than 1%. Photon absorptiometry has been widely employed in the measurement of skeletal status since its introduction twenty years ago[2]. In this technique a highly collimated beam of low energy radiation is obtained from a radionuclide source and monitored by a scintillation detector. The source and detector pass across the limb and the changes in emergent beam intensity are proportional to bone mass. The precision is typically 2–4% in routine practice, resulting in only a moderate discrimination between normals and osteoporotics.

All bone undergoes continuous renewal. Bone turnover occurs more rapidly in trabecular bone, the spongy inner structure, than in the outer framework of cortical bone[3]. It is likely that changes will be seen earlier in trabecular bone. Computed tomography (CT) can provide a three-dimensional image of an object, so permitting the separate quantification of trabecular and cortical bone, but commercial CT scanners are not ideally suited for this task. Artefacts caused by the polychromatic nature of X-rays and the relatively large beam width which they use, lead to distortions in the reconstructed CT values, particularly when measuring bone. Corrections for these sources of error might be applied if the original X-ray transmission data could be used, but this is usually not available. Even if they could provide satisfactory results, commercial scanners are expensive and subject the patients to radiation doses which are undesirably high for repeated use in serial studies.

A special purpose CT scanner for bone measurement using a radionuclide source instead of an X-ray tube was first described by Ruegsegger and others in 1976[4], it was used to measure bone density in the distal forearm. The distal radius is an accessible bone and there are readily measurable amounts of trabecular and cortical bone at this site. Previous studies have shown good correlation between the distal radius and the neck of femur[5] and between trabecular bone in the radius and in the spine[6,7].

The special purpose CT scanner described in this report uses the same approach as that of Ruegsegger *et al.* but is of a new design in which a number of innovative features are combined to give improved precision, speed of measurement and patient comfort, at relatively low cost.

EQUIPMENT

The CT scanner in *Figure 1* is a second generation translation–rotation type in which the radiation source and detectors translate across the width of the arm followed by rotation of the gantry by 2.25°. The translation is repeated at the new angle, and the whole process is repeated through a total rotation of 180°.

The mechanical design was aimed at keeping the scanner as compact as possible with backlash as near to zero as practicable. The translation motion

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Figure 1 Special purpose CT scanner for bone measurement. Trabecular and cortical bone can be separately measured for 2 mm slices of the forearm.

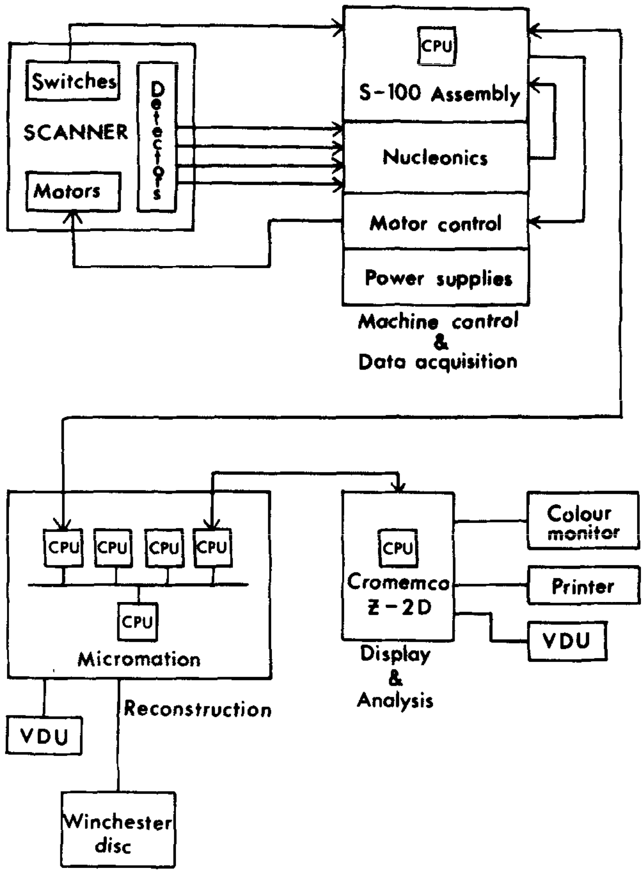


Figure 2 Block diagram of the CT bone scanner.

was achieved using a stepper motor, toothed belt drive and precision screw. Preloaded ball races were used in the screw drives to ensure zero backlash and minimal wear. The rotary motion was driven by a second stepper motor through a zero backlash gearbox unit connected to a pinion driving a large ring gear. Translational positioning accuracy of ± 0.015 mm was achieved with this arrangement, the angular setting error being less than 0.1°. A unique feature of the scanner is the mounting of the whole scanner head on linear bearings which enable it to be driven by a third stepper motor through 50 mm axially along the length of the arm.

The radiation source is a 500mCi (18 GBq) 1 mm diameter bead of Iodine 125 (Amersham International, IMC 51133) changed at three-monthly intervals. The source is collimated to produce four beams of radiation which have a narrow spectrum of energies around 29 keV. Each beam is detected by a highly collimated 12 mm thick NaI (Tl) scintillation crystal. The four photomultipliers operate in count mode and the energy selected pulses are interfaced to the computer by means of purpose built electronic counters.

Three computers are employed, one for machine control and data acquisition, a second, which incorporates five processors, for reconstruction calculations, and a third for display and analysis (*Figure 2*). Commercially available computers and sub-units, all based upon the Z80 microprocessor and S–100 bus were used throughout, supplemented in the machine controller by counter, timer and interface boards of our own design. Programmes for the machine controller are stored in internal ROM. The other computers have disc storage.

In preparing for a scan, details such as scan length and scanning speed are passed from the reconstruction computer to the machine controller, which then directs the mechanical movement of the machine, collects data from the nuclear electronics, and passes it after each translation to the reconstruction computer. This then carries out all the numerical processing before passing the results to the display computer.

The reconstruction computer is a Micromation 'Mariner'. Its use for numerical processing is one of the principal features of the scanner. This computer is designed for use on a number of simultaneous tasks; the tasks are processed by separate 'satellite' central processing units (CPU) each of which has its own memory and facilities for external communication. Shared facilities, printing, mass storage and task intercommunication, are managed by a further 'master' CPU. In this application each satellite CPU processes a separate part of the reconstruction routine, data being passed from one CPU to the next as each completes its operation on a given block of data. Use of the filtered back projection method of Shepp and Logan[8] allows the computation to be divided into separate sequential tasks and permits it to begin as soon as the first segment of data has been collected. The first processor corrects the data for count losses due to deadtime and calculates the projection, the relative count-rate as a function of position, for each translation of the radiation beam across the arm. A further correction is applied to take account of the beam hardening artefact resulting from the polychromatic nature of the radiation source[9]. Two satellite processors, acting in parallel, perform convolution filtering on each projection, and pass the modified data to the fourth processor to reconstruct the cross-sectional image by backprojection. This then passes its data to the third microcomputer, a Comemco Z–2D, for display on a colour monitor. The computers, programmed in Z80 assembly language, produce an image within a minute of the scan finishing. If artefacts are produced by movement of the patient's arm they are detected immediately and the scan can be repeated. The display computer is used to store the images on flexible discs and to perform the analysis; the results and images are both reproduced on a printer.

TECHNIQUE

The length of the patient's ulna, defined as the distance between the elbow and the palpable tip of the styloid process with the arm in extreme flexion, is measured using a specially designed height gauge. The arm of the patient is secured in the arm holder of the scanner and positioned approximately by means of a light beam. An accurate determination of the site to be used for measurement is then made by means of a scanning radiograph, or scout scan. For this the scanner translates 24 times across the arm moving by 2 mm between each scan to produce an image which covers 48 mm of the distal forearm (*Figure 3*). The ends of the radius and ulna can be clearly defined and the scanner automatically positioned at a site 3% from the end of the ulna. A set of three juxtaposed 2 mm thick slice measurements are made starting at the 3% position and a separate measurement is made at 25% of the ulna length, a site at which there is only cortical bone.

Each scan is completed in about two minutes, the computation occupying a further minute; the whole procedure requires not more than twenty minutes, during which the patient is comfortably seated. Once seated, no movement is required of the patient and no movement is imposed on the arm by the machine. The whole procedure is designed to minimize the occurrence of movement artefacts.

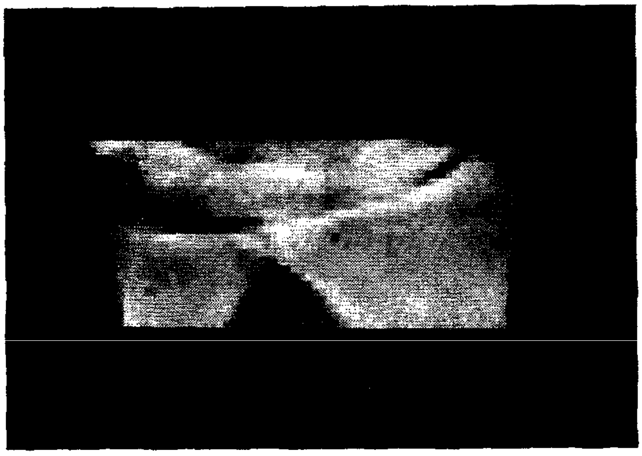
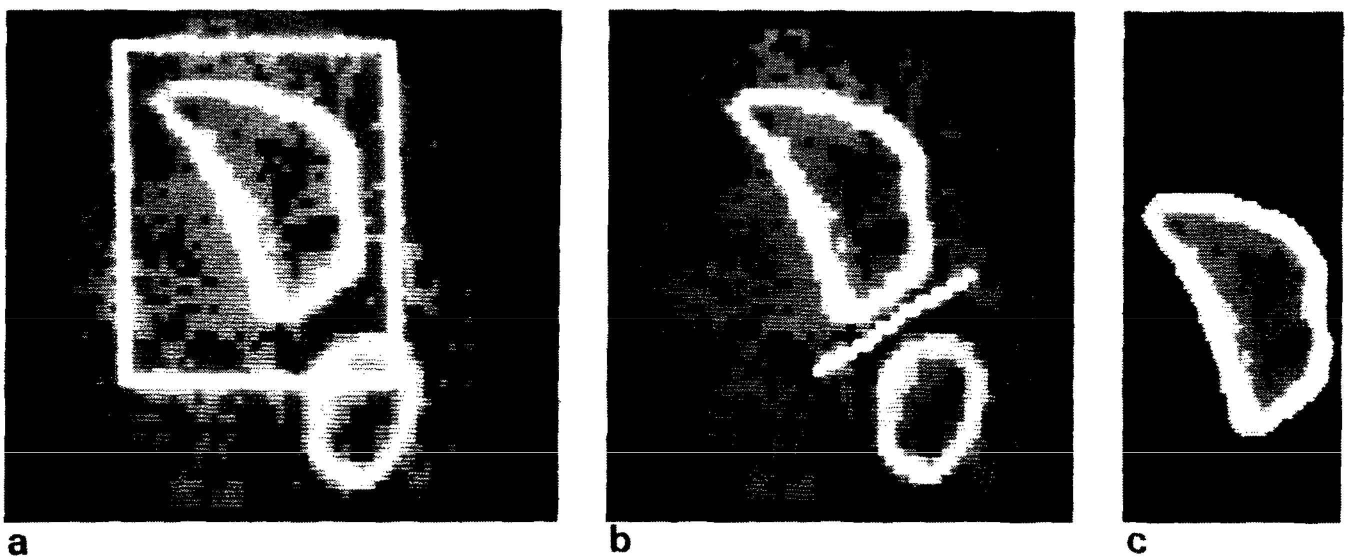


Figure 3 Scanning radiograph used to position the scanner accurately. The distal end of the radius (right) and ulna can be defined.

The radiation dose to the patient is very low, typically about 40 mRad and is confined to a small region of the distal forearm.

The rate of change of cross-sectional area of the radius is approximately 7% per mm at the distal site. As the scanner is able to measure cross-sectional area to better than 1% it is possible to use the area, and linear interpolation between slices to identify precisely during repeat examinations, sites within the bone.

In order to obtain quantitative information from the images the bone boundaries have to be defined. The operator selects a region which includes the radius and if necessary eliminates the ulna from the image (*Figure 4*). A surface detection algorithm defines the edge of the bone separating it from the surrounding soft tissue. Trabecular bone occupies more than 50% of the cross-sectional area of the bone at the distal site[10]. Shells of pixels (picture elements) are stripped from the bone image, starting at the outer contour and removing



a b c

Figure 4 Reconstructed image of the forearm showing the analytical procedure. The radius is defined (a) and the ulna eliminated (b). A surface detection algorithm separates the bone from the surrounding soft tissue (c). The irregular white outlines represent bone cross-sections and the white rectangle that part of the image accepted by the computer for analysis.

the ring of cortical bone, until 50% of the pixels remain. This region represents trabecular bone. Individual pixel values represent linear attenuation coefficients (CT value in cm^{-1}) and therefore the average CT value for the total bone section and for the trabecular bone component can be determined. To enable comparisons to be made with other techniques it is possible to estimate the trabecular bone density ($g\ cm^{-3}$). This can be done by dividing the tissue into a bone component and a soft tissue component and making the assumption that the attenuation properties of the soft tissue component are constant for all measurements[11].

The results of a measurement, which can be produced immediately after the patient's visit, are stored in a data base on the hard disc of the main computer. They are readily available for serial studies on individual patients and on patient groups.

PERFORMANCE

Repeated measurements of a solution of K_2HPO_4 in a phantom gave a coefficient of variation of 0.04%. This error includes noise due to the reconstruction algorithm and photon statistics. A linear relationship was obtained for various solutions of K_2HPO_4 covering the full range of trabecular bone density (*Figure 5*).

The largest single error in using the scanner in clinical practice is caused by the irreducible variation in repositioning the arm, changes of density of several percent per millimetre being typical for the distal radius. Bone CT values were measured for six sets of scans on a normal subject using the technique previously described; measurements were made at approximately weekly intervals over a period of 46 days. The results for

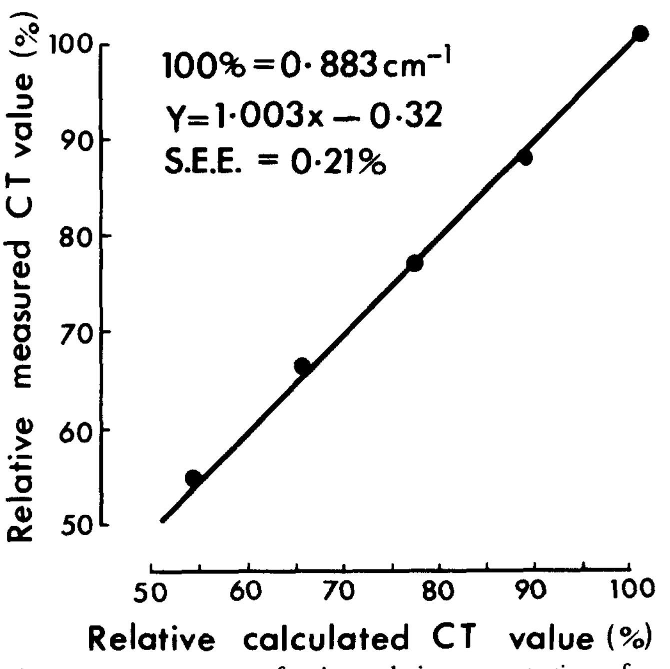


Figure 5 Measurement of various solution concentrations of K_2HPO_4 showing the scanner precision for a range of bone densities.

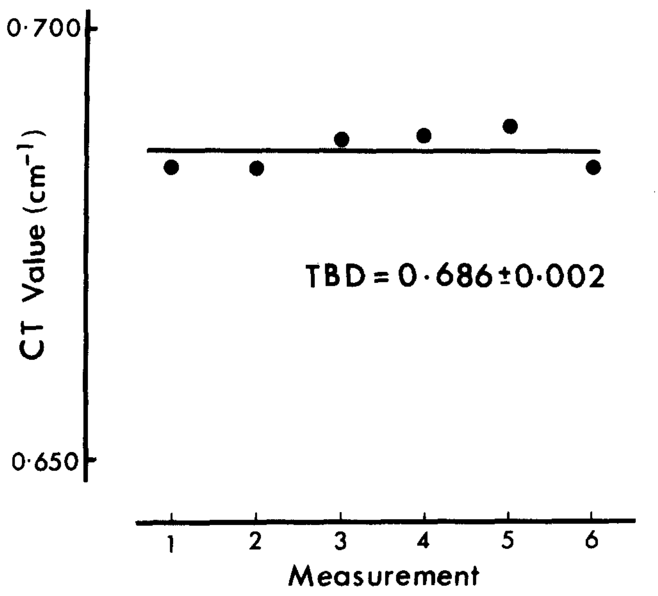


Figure 6 Measurement of trabecular bone CT value for the radius showing the reproducibility for repeat scans over a period of seven weeks.

trabecular bone (TBD) are shown in *Figure 6*; the coefficients of variation were 0.31% for trabecular bone CT value, 0.72% for total bone at the distal site, and 0.87% for the cortical bone measurement. The long term precision of the scanner is being assessed using a perspex phantom containing two aluminium tubes to represent the radius and ulna. Over a period of two months, which has included a source change, the coefficient of variation has been better than 0.3%.

The scanner is being used clinically and over 100 patients have been scanned. The procedure is easily applied and is readily accepted by patients.

CONCLUSION

For a method to be suitable for monitoring accurately the time course of osteoporosis and its response to treatment, serial measurement must be possible over periods of many months with a reproducibility of the order of 1%. CT is the only method by means of which it is possible to quantify trabecular and cortical bone separately during a longitudinal study of individual patients. A CT scanner has been produced which can measure bone mineralization with high precision. The whole measurement procedure takes less than 20 minutes and is performed with minimal discomfort to the patient and with low radiation dose. This self-contained scanner should be a valuable tool both for the assessment of bone loss and for monitoring its response to treatment.

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REFERENCES

1 Mazess, R.B., Non-invasive measurement of bone. *Osteoporosies II*, (Ed. U.S. Barzel) Grune & Stratton, New York, 1979, 5–26.

2 Cameron, J.R., Sorensen, J.A., Measurement of bone mineral in vivo: An improved method. *Science*, 1963, **143**, 230–232.

3 Parfitt, A.M., The physiologic and clinical significance of bone histomorphometric data. *Bone Histomorphometry: Techniques and Interpretation*, (ed. R.R. Recker) CRC Press, Boca Raton, Florida, 1983, 143–223.

4 Ruegsegger, P., Elsasser, U., Anliker, M., Grehm, H., Kind, H., Prader, A., Quantification of bone mineralisation using computed tomography. *Radiology*, 1976, **121**, 93–97.

5 Wilson, C.R., Bone mineral content of the femoral neck and spine versus the radius or ulna. *J. Bone Jt. Surg.*, 1977, **59-A**, 665–669.

6 Ruegsegger, P., Anliker, M., Dambacher, M., Quantification of trabecular bone with low dose computed tomography. *J. Comput. Assist. Tomogr.*, 1981, **5**, 384–390.

7 Hangartner, T.N., Overton, T.R., Recent developments in peripheral CT measurements in Edmonton. *Proc. 3rd Int. Workshop on Bone and Soft Tissue Densitometry using Computed Tomography*, University of Alberta, 1982.

8 Shepp, L.A., Logan, B.F., The fourier reconstruction of a head section. *IEEE Trans. Nuc. Sci.*, 1974, **NS-21**, 21–39.

9 Sandrik, J.M., Judy, P.F., Effects of the polyenergetic character of the spectrum of I–125 on the measurement of bone mineral content. *Invest. Radiology*, 1973, **8**, 143–149.

10 Schlenker, R.A., Von Segger, W.W., The distribution of cortical and trabecular bone mass along the lengths of the radius and ulna and the implications for in vivo bone mass measurements. *Calcif. Tiss. Res.*, 1976, **20**, 41–52.

11 Hangartner, T.N., Overton, T.R., Quantitative measurement of bone density using gamma-ray computed tomography. *J. Comput. Assist. Tomogr.*, 1982, **6**, 1156–1162.