ANNALS OF BIOMEDICAL ENGINEERING 2, 194–205 (1974)

An Extension of Classical Bone Mineral Measurements

P. RÜEGSEGGER, P. NIEDERER, AND M. ANLIKER

Institut für Biomedizinische Technik
der Universität Zürich und ETH Zürich,
Rämistrasse 100, 8006 Zürich, Switzerland

Received November 26, 1973

Cameron's concept of bone mineral measurement using soft γ-rays has been extended to determine the area and shape of the compacta cross-section and the corresponding mean linear absorption coefficient. By repeating the linear scanning process at the same section for different directions a polygon is defined which circumscribes the outer contour of the bone section. With the aid of a PDP 11-45 computer and an iterative technique the compacta area and the mean linear absorption coefficient are evaluated. In addition the computer synthesizes iteratively an image of the actual compacta cross-section. Thanks to the redundancy of the absorption data the values for the total mineral content, the compacta area and the mean density can be determined as many times as the scanning process has been carried out. The fluctuations of these values from their averages can apparently be interpreted as a measure of the consistency of the data. To validate the method and assess the accuracy and statistical fluctuations of the measurements an aluminum tube with circular cross-section has been used as a model for a bone. The data indicate that the total mineral content, the compacta area and the mean density can be determined with an error on the order of 1%. Data from measurements on excised human femurs have been obtained and the results are equally consistent as those of the aluminum model.

I. INTRODUCTION

The noninvasive determination of bone mineral content as introduced by Cameron *et al.* (1962, 1967) is based on the absorption of soft γ-rays. By scanning a given section of a bone with a collimated beam from a monoenergetic γ-source and computing the area under the corresponding absorption curve one obtains a measure for the total mineral content at that section. In order to account for the effects of the surrounding tissue, the part of the body under examination is either immersed into water or a second independent absorption measurement is carried out using γ-rays with a different energy (Cameron 1963).

In performing long term (longitudinal) investigations of demineralization or studies of the efficacy of therapeutic procedures one is confronted with the problem of examining the very same bone section in each of the successive absorption measurements. Natural variations in the mineral content along the bone can, in principle, mimic larger temporal changes in the total mineral content than those which may be due to the disease or therapy if we do not scan the very same cross-section in each examination. Therefore, the intersection point of the scanned plane with some fixed bone axis as well as the orientation of this plane with respect to the axis have to be kept within close tolerances.

For diagnostic purposes one has to allow for upper and lower limits which enclose a relatively wide interval of "normal" values for the total mineral content in certain sections of a specific bone. If additional quantitative information on

(1) the actual outer contour of the bone section examined,
(2) the distribution of the compacta and its area and
(3) the separate mineral contents of the compacta and the spongiosa

were available, these limits of normalcy could very likely be narrowed and made more incisive.

On the basis of studies on excised bones performed in our laboratory it appears that the Cameron technique can be extended to yield additional data on (1) and (2) as well as on the mean linear absorption coefficient of the compacta. Even though no distinction is yet made between compacta and spongiosa this supplementary information may allow for a better definition of the bone status and range of normalcy. It also eliminates the need to examine the very same cross-section of a bone in a longitudinal study if the mean linear absorption coefficient proves to be insensitive to the variations in the positioning and orientation of the scanning beam.

A technique to determine a comparative image of a given bone section has recently been published by Peters, Smith and Gibson (1973). By taking x-ray pictures from different coplanar directions (in the plane of the section) and employing a Fourier reconstruction method they synthesized the shape of the section on the basis of densitograms of the x-ray images. Stimulated by the success of the computerized transverse scanning system conceived and designed by Hounsfield (1973) we set our goal to develop a simple iterative procedure for quantifying the bone features outlined above as accurately as possible and without the use of x-ray films.

II. METHOD

To determine size and shape of the compacta and its mean linear absorption coefficient at a given cross-section of a bone, we are repeating the conventional linear scanning process at that section 13 times by rotating the collimated beam of γ-rays through 15° after each scan (Fig. 1). The range of the linear scan is divided into $N = 1000$ equal intervals. For scan direction n and interval k the mean pathlength of the γ-beam through the bone section is $\delta(n, k)$. The corresponding transmission rate of photons is denoted by $I(n, k)$. If the beam width is small compared to the variations of the cross-section and the γ-ray source is monoenergetic, the transmission rate of the photons is given by

$$I(n, k) = I_0 \exp(-\rho\mu\delta(n, k)) \qquad (1)$$
$$n = 1, 2, \ldots, 13$$
$$k = 1, 2, \ldots, N,$$

were I_0 is the transmission rate for $\delta = 0$, ρ the density of the medium and μ the absorption coefficient.

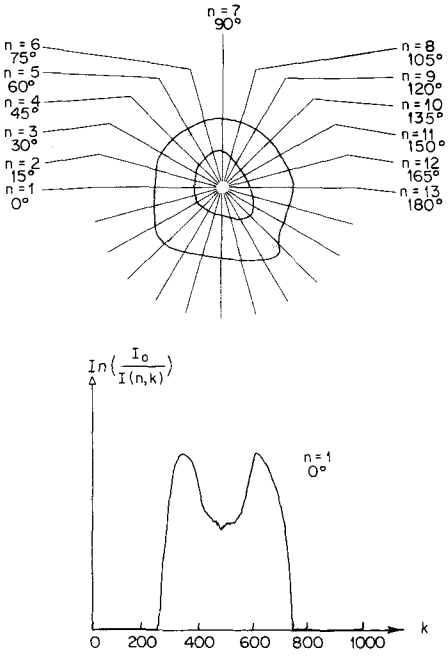

FIG. 1. Illustration of bone cross-section with the scan directions n and the absorption curve for $n = 1$ or 0°.

The transmission rate corresponding to each interval is stored "on line" in a PDP 11-20 computer. At the end of the 13 linear scans we arrange the total of the 13000 values $I(n, k)$ as a matrix

$$G(n, k) = ln(I_0/I(n, k)) = D\delta(n, k), \tag{2}$$

where $D = \mu\rho$ is the mean linear absorption coefficient of the bone tissue.

From the data given by the matrix $G(n, k)$ and the associated beam directions and interval lengths we first determine the geometric parameters for a 24 sided polygon which circumscribes the outer contour of the bone section. To this end the absorption curves for the first and the 13th scan are used to compute the center of rotation. The 24 tangents to the circumference (Fig. 2) are determined from the directions of the γ-ray beam and the corresponding segments of the linear scan range for which the transmission rate is less than $I_0 - 4\sqrt{I_0}$.

For beam direction n the mean thickness of the polygon for interval k is $F_a(n, k)$ (Fig. 3). The difference between $F_a(n, k)$ and the pathlength $\delta(n, k)$ represents that part of the polygon thickness which does not contain any bone tissue. We therefore call this difference

$$F_i(n, k) = F_a(n, k) - \delta(n, k), \tag{3}$$

the "Loch-distance".[1] This "Loch-distance" includes both the medullary canal or cavity in the bone section and the difference between the envelopping polygon

[1] "Loch" means hole in German.

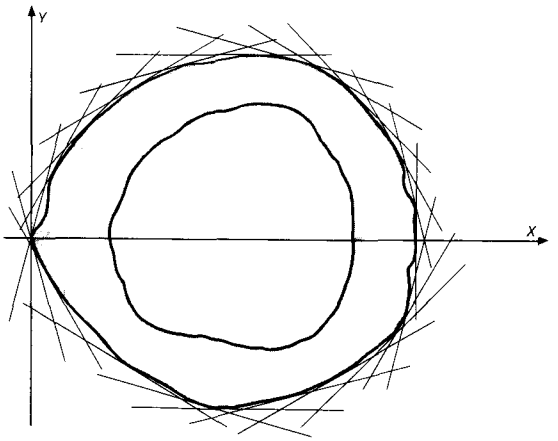

FIG. 2. Actual bone cross-section with circumscribed 24-sided polygon.

and the true outer contour of the bone (Fig. 3). If we arrange the values for $F_a(n, k)$ corresponding to each n as arrays, Eq. 3 defines a "Loch-matrix".

By combining (2) and (3) we have for every scan direction n a total of N equations

$$F_i(n, k) = F_a(n, k) - G(n, k)/D \quad k = 1, \ldots, N \tag{4}$$

for $N + 1$ unknowns F_i $(n, 1)$, F_i $(n, 2)$, . . . , F_i (n, N) and D. An additional independent relation for these unknowns can be formulated by stipulating that the array of "Loch-distances" assume the value zero for certain intervals k. This relation should be met with sufficient accuracy when the contour of the bone section is approximated by a 24 sided polygon. To find the intervals of the linear scan range

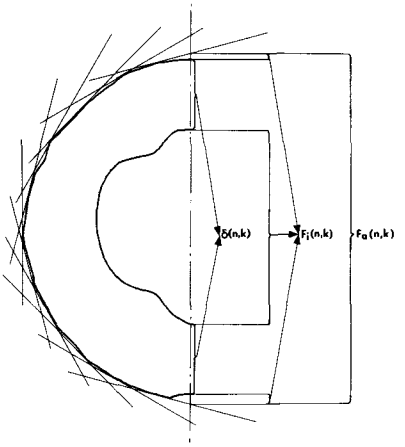

FIG. 3. Definition of polygon thickness F_a, "Loch-distance" F_i and path length δ (n, k) of γ-beam through compact bone.

for which the array F_i (n, k), $[k = 1, 2, \ldots, N]$, is zero or approaches zero, we make use of the following iterative procedure.

The area circumscribed by the polygon is first assumed to be filled with material whose density D' is chosen such that it is significantly smaller than the maximum of any value of $G(n, k)/F_a(n, k)$. Such a material would produce an absorption curve as sketched in Fig. 4 by the dashed line. From the fact that the array of corresponding "Loch-distances" $F_i'(n, k) = F_a(n, k) - G(n, k)/D'$ is negative for certain scan intervals, we infer that the chosen value D' was too small. We increase D' until the array $F_i'(n, k)$ contains no longer negative elements except for statistical fluctuations. The associated value for D' is considered the actual mean linear absorption coefficient D for beam direction n. An example of a representative array of "Loch-distances" is graphically illustrated in Fig. 5. The cavity and the effects of spaces between the polygon and the outer contour of the bone can readily be identified. To evaluate the area of the compacta cross-section we integrate the difference $F_a(n, k) - F_i(n, k)$ along the linear scan direction.

By repeating this iterative procedure for each of the n linear scans we arrive at 13 values for D and for the area of the compacta cross-section. The fluctuations of these values could be interpreted as a measure of the consistency of the final results and the accuracy of the method.

A picture of the bone cross-section can be synthesized to true scale with the aid of an iterative computer program. The cross-section is imbedded in a raster of 100×100 fields. In a first iteration we identify all fields whose centers are outside the polygon with a zero, all others with a one. The fields thus labeled with a "one" are assumed to be filled with a material whose density is identical with the average density determined from the 13 scans. For the bone section defined in this manner we compute the corresponding hypothetical absorption matrix $G_0^*(n, k)$. We then assume that one of the fields within the polygon is actually a cavity by

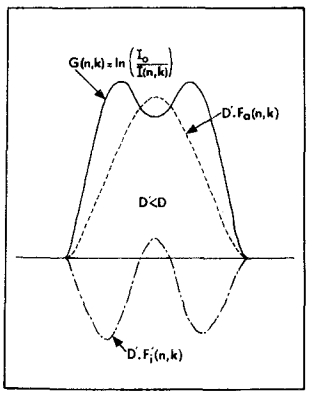

FIG. 4. Graphic representation of measured absorption curve defined by $G(n, k)$ and of hypothetical absorption curve $D' \cdot F_a(n, k)$ corresponding to the polygon being filled with a material whose mean linear absorption coefficient is D'. The curve $D' \cdot F_i'(n, k)$ illustrates the difference between the hypothetical and the measured absorption curve; $D' \cdot F_i'(n, k)$ no longer assumes negative values when D' is equal to the true value D, for which $F_i'(n, k)$ becomes $F_i(n, k)$.

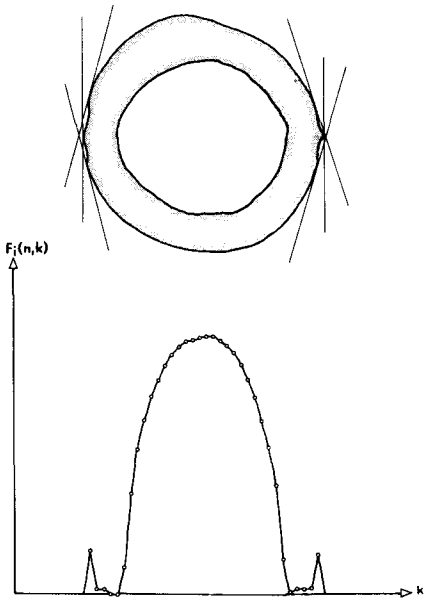

FIG. 5. Illustrations of the "Loch-distance" $F_i(n, k)$ and associated bone section. Note that $F_i(n, k)$ assumes only positive values except for statistical fluctuation.

changing its label from a "one" to a "zero" and we compute for the cross-section altered in this manner the associated hypothetical absorption matrix $G_1^*(n, k)$. Both of these hypothetical absorption matrices $G_0^*(n, k)$ and $G_1^*(n, k)$ will differ from the measurement matrix $G(n, k)$. This difference can be expressed in terms of a quadratic deviation for each of the twelve independent scan directions. If now $G_1^*(n, k)$ has quadratic deviations which are larger for at least two scan directions than the corresponding deviations of $G_0^*(n, k)$ then the element is reset to "one", otherwise it stays a "zero" and the matrix $G_1^*(n, k)$ becomes the new $G_0^*(n, k)$. This procedure is repeated for every one of the field elements within the polygon.

In a second iteration the fields containing a "one" are again successivly set to "zero." If the new hypothetical absorption matrix $G_1^*(n, k)$ has quadratic deviations which are larger for at least three scan directions than the corresponding deviations of $G_0^*(n, k)$ then the element retains the "old" label, otherwise it stays changed and the matrix $G_1^*(n, k)$ becomes $G_0^*(n, k)$.

The iteration process for the synthesis of the cross-section is terminated when either the compacta area of the picture agrees within certain limits with that computed from the difference $F_a(n, k) - F_i(n, k)$ or when the eleventh iteration has been completed.

III. EXPERIMENTAL PROCEDURE

A schematic sketch of the experimental arrangement used to establish the feasibility of the method is given in Fig. 6. The mechanical part of the densitometer is

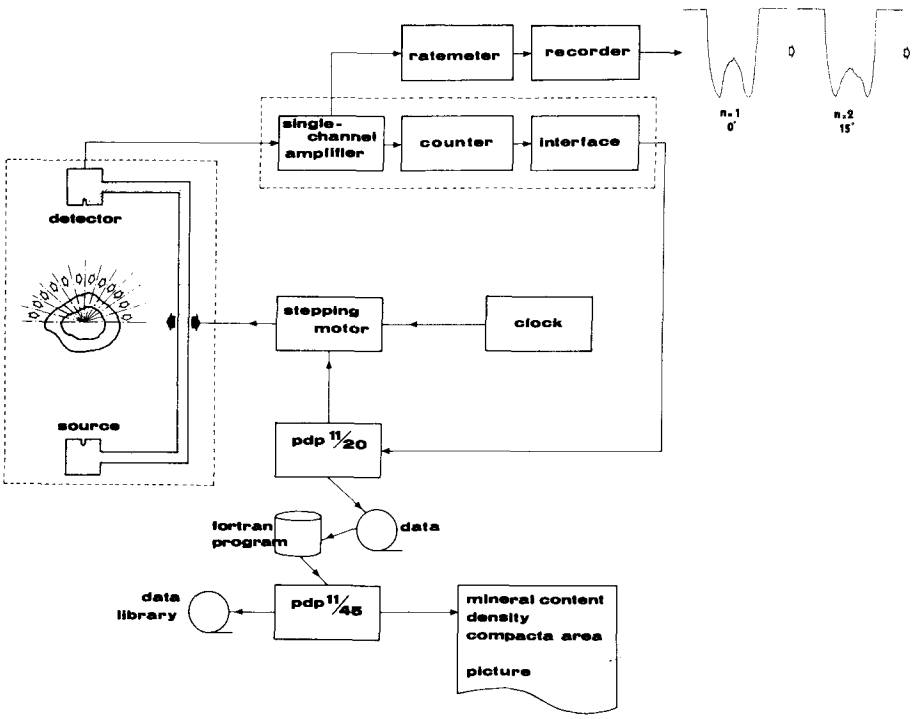

FIG. 6. Schematic illustration of experimental arrangement.

basically the same as that of conventional systems except that the object under examination can be rotated about a fixed axis. The γ-rays from a 500 mCi ^{125}I source are collimated to a beam with a diam of about 1 mm and the transmitted photons are detected with a photomultiplier (Nuclear Enterprises Model DM 1-1) covered by a NaI cristal of 0.5 mm thickness. With an additional collimator of 1 mm diam placed in front of the detector, the pulse rate is between 10^4/sec and 10^5/sec. The pulses from the photomultiplier are amplified by a Wenzel linear amplifier N-LVA-211 and limited in energy with a Wenzel single channel discriminator N-ED-10 so that only pulses between 15 and 45 keV are admitted. The output pulses from the discriminator have a width of 200 nsec and are recorded by a 16 bit scaler (SEN 4S2003) and a rate meter (Wenzel N-R-107). For visual inspection of the absorption the logarithm of the output from the rate meter is displayed with a pen recorder.

The number of pulses counted by the scaler within fixed time intervals is transferred to the core memory of a PDP 11-20 computer which also gives the start and stop commands to the control unit of the stepping motor driving our linear scanner. Both, the duration of the time intervals and their total number N can be selected. Considering that the linear scanning proceeds at constant speed once the start command has been given, we have for each of the fixed time intervals a corresponding fixed scan interval. For the measurements described here the time in-

terval is 90 msec and $N = 1000$. The total scan distance is 7 cm. After 1000 steps the computer stops the scanner, transfers the data from the core memory to a DEC-Tape and waits until the scan direction has been changed. When all of the 13 scans have been performed the computer asks for the identification number of the next experiment. A small error is introduced into the absorption data by the transfer of the count number from the scaler into the core memory. However this error is negligible because this transfer-process is executed by a single instruction lasting 2.3 μsec.

All computations required for the evaluation of the total mineral content, the mean density, the compacta size and shape are performed on a PDP 11-45 computer equipped with disc, magnetic tape and line printer. However before these computations are executed, the densitometry data $I(n, k)$ is corrected for the "dead time" ($\tau = 1.5$ μsec) of the γ-ray detection and counting system and for the "beam hardening." Considering that the γ-rays from the ^{125}I source are not monoenergetic as assumed for Eq. 1 and that the absorption coefficient μ is a function of the energy of the γ-rays, we have to account for "beam hardening," that is, for the change in the beam energy distribution during the transmission of the rays through the absorbing medium (Sandrik & Judy, 1973). We have therefore normalized the absorption data with the help of the calibration curve from an aluminum wedge given in Fig. 7. Because of the finite dimensions of the collimated γ-ray beam the polygon defined in Chapter II circumscribes a bone section which is slightly scaled up from the true contour. This error calls for a third correction which is empirically determined from absorption measurements of objects with known dimensions.

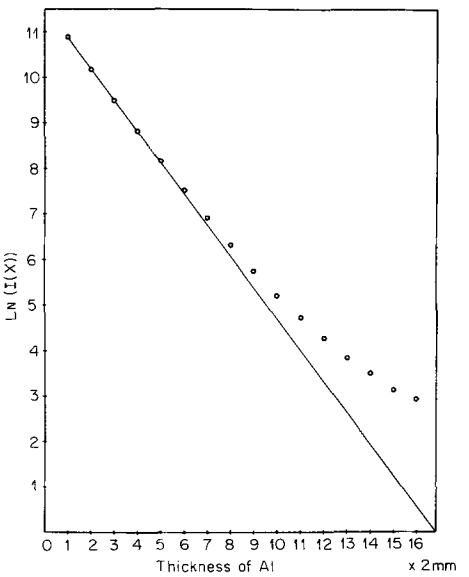

FIG. 7. Effect of beam hardening, background radiation and electronic noise.

IV. RESULTS

In order to validate the method and assess the accuracy of the results an aluminum tube with circular cross-section was used as a model for a bone. The values for the total linear mineral content, the "compacta" cross-section and the density in g/cm^3 determined from the densitometry data are summarized in Table 1 together with the corresponding reference values derived from the dimensions and weight of the aluminum cylinder and from the absorption coefficient of the aluminum. Executing a single measurement procedure we find that all three quantities agree within 1% with the reference values. The errors given for the values determined from the densitometry data are standard errors of the respective means obtained from the 12 independent scans. The fluctuations of the results from the 12 scans justify the assumption of a normal distribution. Repetitions of the measurement procedure showed that the densitometry results are reproducable within 1%.

Systematic studies on excised human femurs are now in progress. The results from two such bones are summarized in Fig. 8. Densitometry measurements were carried out at 10 different sections beginning near the femoral head. Scan number 10 is situated 6 cm from the fossa intercondylaris. The distance between two individual sections is 2.5 cm. In these examples the total mineral-content shows a change along the femur of about 30%, the mean linear absorption coefficients D however are nearly the same for all sections. The absolute difference in the total mineral content between the two femurs is also on the order of 30%. Within the middle third of the femurs we find that D generally varies not more than $\pm 1\%$.

The consistency of the densitometry data obtained from excised femurs compares favorably with that of the aluminum model. The standard errors of the means for the compacta area and for D are on the order of 1% and as such slightly greater than in the case of the aluminum cylinder. In the central regions of the femurs the reproducability was better than 1%.

A computer print-out of a synthesized compacta cross-section of a femur is illustrated in Fig. 9 together with a photograph of the bone section. Even though the computer picture is compressed by 20% in the vertical direction the similarity is still evident. An actual quantitative comparison was made in the case of the aluminum cylinder but has yet to be carried out for the bone sections.

V. DISCUSSION AND CONCLUSIONS

The potential clinical usefulness of the method described here has been documented in the form of results for an aluminum cylinder and for excised human femurs. For clinical applications involving the femur, the humerus or a finger bone, no basic modifications are needed, but the procedure has to be extended for body parts containing more than one long bone or a purely trabecular bone. In its present form the apparatus is designed to study excised bones. *In vivo* measurements will require a mechanical scanning system which can be rotated around the body part of interest. Such a system is now being assembled.

So far no attempts have been made to optimize accuracy, measuring time and radiation dose. With a 500 mCi source of ^{125}I and a measuring time of 20 min our

TABLE 1
AL-MODEL, COMPARISON BETWEEN REFERENCE VALUES AND DENSITOMETRY RESULTS FROM A SINGLE MEASUREMENT PROCEDURE

	Reference values	Densitometry results
Al-mass [g/cm]	6.56 ± 0.01	6.52 ± 0.04
Compacta area [mm^2]	243.6 ± 0.5	241.3 ± 1.6
Density [g/cm^3]	2.693 ± 0.005	2.677 ± 0.017

present system would utilize a radiation dose of at most 100 mR. In view of the fact that only a body section of about 1 mm thickness would be exposed the associated radiation hazard is a small fraction of that connected with a conventional X-ray picture of the body part of interest. A theoretical evaluation of the newly designed system for clinical applications indicates that the radiation dose and measuring time can be substantially reduced without sacrificing accuracy thanks

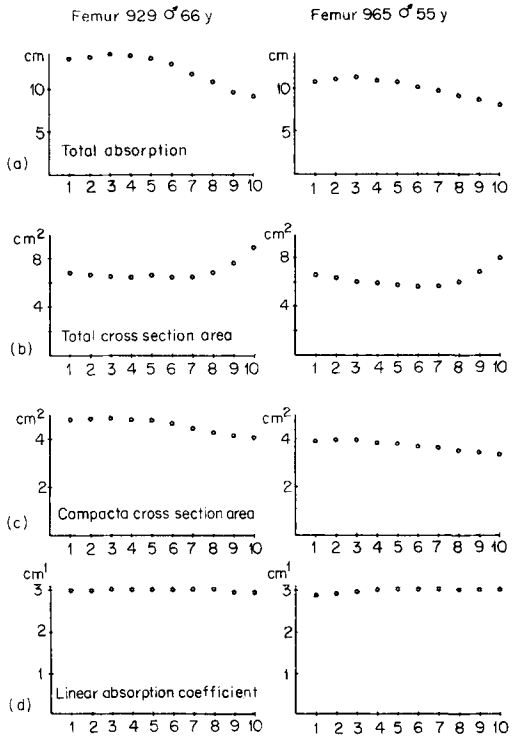

FIG. 8. Densitometry results from two human femurs. From top to bottom are given: (a) The total absorption, i.e., the integral of the absorption curve, as a measure for the total mineral content, (b) the total area defined by the envelope of the cross-section, (c) the compacta area and (d) the mean linear absorption coefficient for different stations along the femur shaft. Station 1 is below the femur neck and Station 10 is 6 cm above the fossa intercondylaris. The distance between successive stations is 2.5 cm. Taking into account the energy spectrum of the radiation we find that the linear absorption coefficient of about 3 cm^{-1} agrees with the data given by Spiers (1946).

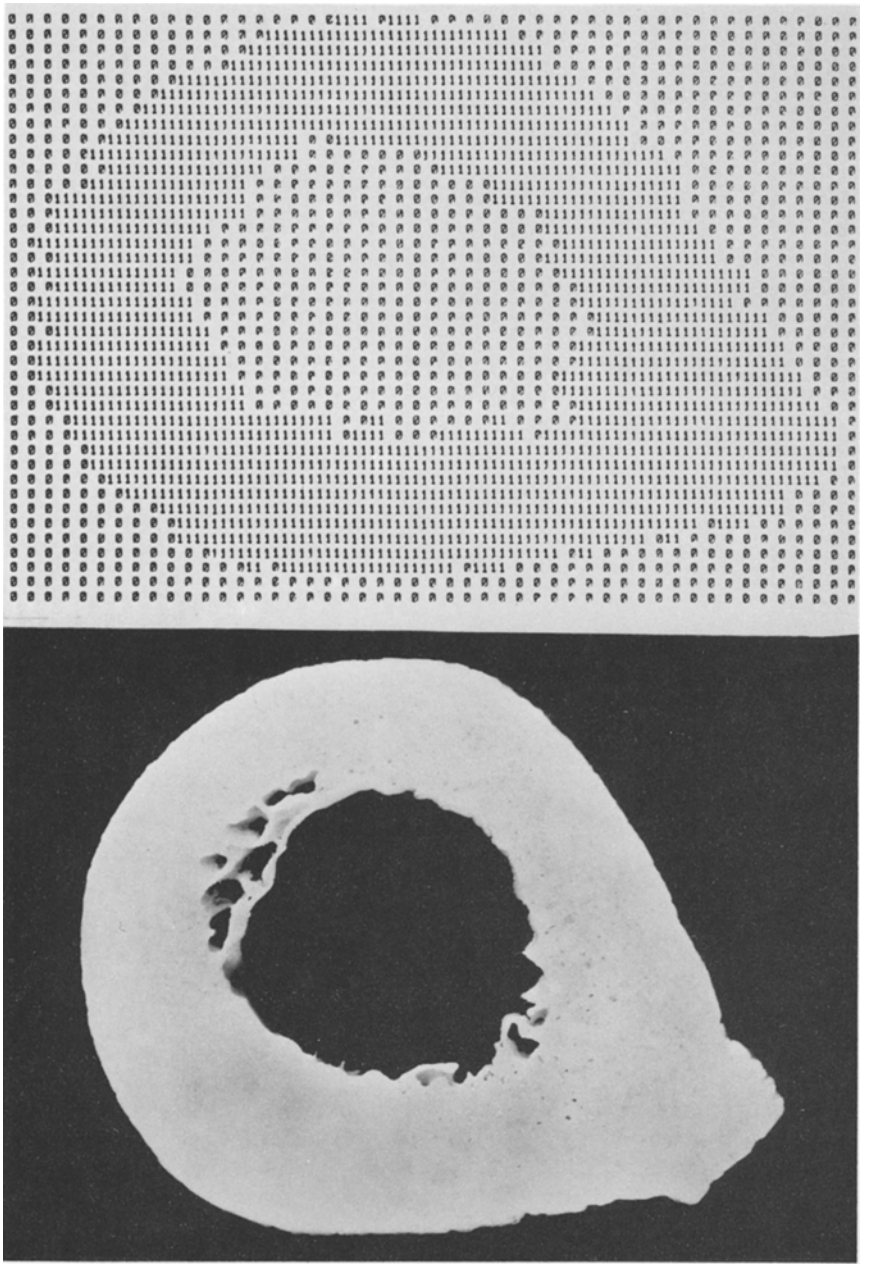

FIG. 9. Above: Computer print of compacta cross-section derived from the densitometry data, 20% compressed in vertical direction. All spongiosa is added to the compacta. Below: Photograph of the same bone cross-section.

to improvements on the collimation and by minimizing the scanning distance. While the accuracy of the data can be increased with a refined mechanical design, it is difficult to predict the error due to the presence of the soft tissue.

The time required for the evaluation of the compacta area and the linear absorption coefficient with our PDP 11-45 computer system is approximately 2 min. If the shape of the compacta cross-section is desired as additional information the computer time is prolonged by another 2 min.

Aspects which govern the applicability of the method are the following. The outer contour of the bone section must be sufficiently regular, that is, it should not exhibit several indentations of a magnitude which is comparable with the compacta thickness. Fortunately all of the long bones in man which lend themselves to a bone mineral analysis meet this criterion in general. For an accurate determination of the compacta area and mean linear absorption coefficient, the γ-beam diameter should be less than half of the compacta thickness. This means that the collimation of the beam may have to be adapted to the type of bone to be examined. If the results show a systematic variation in D for the various scan directions, this may either reflect a true circumferential variation of the compacta density, a coarse approximation of the contour by the 24 sided polygon, or too large a beam diameter. In such cases the average values for the compacta area and D would have to be interpreted with caution, in view of the fact that the procedure is based on a reasonably uniform linear absorption coefficient. Finally, the redundancy of the data in the form of 12 values for the total mineral content, the size of the compacta area and its values for D corresponding to the 12 independent scan directions allows for an assessment of the consistency and reliability of the results.

ACKNOWLEDGMENT

The work described in this paper has been supported in part by the Swiss National Science Foundation under Grant No. 4.0600.72.

REFERENCES

CAMERON, J. R., AND GRANT, RODERICK. An Improved Technic for the Measurement of Bone Mineral Content *in Vivo*, *Radiology* 1962, **78**, 117.

CAMERON, J. R., AND SORENSEN, J. A. Measurement of Bone Mineral *in Vivo:* An Improved Method. *Science* 1963, **142**, 230.

HOUNSFIELD, G. N. Computerized Transverse Axial Scanning. *British Journal of Radiology* 1973, **46**, 1016.

PETERS, T. M., SMITH, P. R., AND GIBSON, R. D. Computer Aided Transverse Body-Section Radiography. *British Journal of Radiology* 1973, **46**, 314.

SANDRIK, J. M., AND JUDY, P. F. Effects of the Polyenergetic Character of the Spectrum of ^{125}I on the Measurement of Bone Mineral Content. *Investigative Radiology* 1973, **8**, 143.

SORENSEN, J. A., AND CAMERON, J. R. A Reliable *In Vivo* Measurement of Bone Mineral Content. *Journal of Bone and Joint Surgery* 1967, **49A**, 481.

SPIERS, F. W. Effective Atomic Number and Energy Absorption in Tissues. *British Journal of Radiology* 1946, **19**, 52.

WOOTEN, W. W., JUDY, P. F., AND GREENFIELD, M. A. Analysis of the Effect of Adipose Tissue on the Absorptiometric Measurement of Bone Mineral Mass. *Investigative Radiology* 1973, **8**, 84.