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TITLE OF THESIS/TITRE DE LA THÈSE DEVELOPMENT AND ASSESSMENT
OF COMPTON SCATTERED X-RAY
TECHNIQUE FOR BONE DENSITY MEASUREMENT

UNIVERSITY/UNIVERSITÉ CARLETON IN-VIVO

DEGREE FOR WHICH THESIS WAS PRESENTED/
GRADE POUR LEQUEL CETTE THÈSE FUT PRÉSENTÉE Ph.D.

YEAR THIS DEGREE CON. ERRED/ANNÉE D'OBTENTION DE CE GRADE 1981

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DEVELOPMENT AND ASSESSMENT OF COMPTON
SCATTERED X-RAY TECHNIQUE FOR BONE
DENSITY MEASUREMENT IN-VIVO

by

KULVIR KUMAR KAPOOR

A thesis submitted to the Faculty of Graduate Studies
and Research in partial fulfilment of the requirements
for the degree of Doctor of Philosophy

Department of Physics
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March, 1981

The undersigned recommend to the Faculty of Graduate Studies and Research acceptance of the thesis "Development and Assessment of Compton Scattered X-Ray technique for Bone Density Measurement in-vivo", by Kulvir Kumar Kapoor, B.Sc.(Hons.), M.Sc., in partial fulfilment of the requirements for the degree of Doctor of Philosophy.

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ABSTRACT

A Compton Scatter X-ray Bone Densitometer for Bone Density determination in-vivo is described. The system uses a single 90 kVp quasi-continuous X-ray source and three scintillation crystal detectors - two CsI(Na) scintillators for low flux scattered beam and one plastic scintillator for high flux transmission beam. Each density determination involves four intensity measurements using digital electronics - two of transmitted beams and two of compton scattered radiation. Theoretical studies are made on the contribution from multiple scattering phenomenon which limits the precise determination of density. Preliminary studies have been made on beam hardening effects which arise due to the polychromatic nature of X-ray photon beam. A number of spectra for transmitted and scattered X-ray beams have been taken under various configurations to make some qualitative assessment of the complexity of the problem involved. The main emphasis has been on the compromise between the radiation dose to the limited area of the human subject under investigation and the statistical precision of density measurement. In-vitro measurements on non-biological samples have been made to assess the accuracy and precision of the system before performing in-vivo studies on subjects.

A series of normal volunteer human subjects selected at random from among the Carleton University community has been studied to collect density data for normals. The site chosen for density measurement was in the tibia of right leg of the subject. Using linear regression, measurements were normalized to 40 yrs age & 70 kg wt for male subjects and 40 yrs age & 60 kg wt for female subjects. It is found that the bone density values thus obtained had a mean of 1.35 g/cm^3 with FWHM of 0.20 g/cm^3 for males and 0.35 g/cm^3 for females. The frequency distribution of density values showed they were skewed towards higher density from mean value and peaked at 1.48 g/cm^3 both for male and female subjects. In-vivo density measurements on human subjects were obtained with a standard deviation of the order of $\pm 1.5 \%$ from counting statistics with exposure dose to the localized region under study of about 200 mR. Repositioning of bone site under study reduces the precision to about $\pm 5 \%$.

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to my wife

ACKNOWLEDGEMENTS

I wish to express my sincere thanks to Dr. R. L. Clarke for his supervision and encouragement. The valuable guidance provided by Dr. R. D. Barton during the period Dr. Clarke was on sabattical leave is duly acknowledged. I also express my gratitude to Dr. R. G. Baker for many helpful suggestions.

I wish to thank K. Hafner and J. Sliwka for their assistance in the mechanical design and construction of the modified version of Compton X-Ray Bone Densitometer. The technical assistance provided by Gerry Black in designing optically-coupled isolator circuit for noise reduction and assembling the photomultiplier tube base circuits is gratefully acknowledged. I am also indebted to L. Mueller and L. Bird for providing electronic equipment support as and when required during the course of this study. Thanks are due to Roselyn O'Driscoll and D. Jessup for co-ordination in arranging volunteer subjects for final study. Thanks are also due to Dr. D. Johnson for providing facilities and volunteers at the Carleton University Sports Medicine Clinic to do some initial in-vivo measurements on subjects.

I do not have words at my command to express the indebtedness I owe to Dr. M. K. Sundaresan for his most genuine encouragement, help and guidance.

I will be failing in my duty if I do not acknowledge my gratitude to the volunteers among Carleton University community who willingly offered themselves for carrying out the in-vivo studies with the densitometer, and whose whole-hearted co-operation was essential for the successful completion of this work.

Finally, I wish to express my gratitude to my parent laboratory, the Institute of Nuclear Medicine and Allied Sciences, Research & Development Organisation, Ministry of Defence, India, for according permission and granting study leave to carry out Ph.D. studies in Canada.

A part of this work was presented in the poster session of 3rd Conference on Medical Physics organised by the Association of Medical Physicists of India, held at Bombay, India, during March 19-21, 1980.

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CHAPTER 1

INTRODUCTION AND REVIEW OF CURRENT METHODS OF BONE MINERAL MEASUREMENT IN-VIVO

1.1 Introduction

The accurate quantification of bone mineral mass has acquired great importance over the past fifteen years. The large amount of activity stems from the desire to measure accurately the mineral content of bones in routine diagnostic procedures. Precise determination of skeletal demineralization in patients suffering from bone mineral metabolic disorders is a problem both for the practicing physician and for the clinical investigator. The problem becomes most evident when attempts are made to detect early disease states characterised by slightly reduced bone mass or during attempts to measure changes in mineralization due to therapeutic manoeuvres or the natural course of disease.

Although much information can be obtained using instrumentation and techniques available today, there still exists the necessity and desire for greater precision and more versatility in the selection of bone location.

In studying the factors which affect the bone mineral content, one must consider the activity and stress, lactation, nutrition, medication, disease, age, sex and race of the subject. There have been a number of clinical studies examining the bone

mineral content in people with osteoporosis (22, 35, 40, 102), with osteodystrophy (74), with recent fractures (102) , with primary hyperparathyroidism (38, 79), and patients on hemodialysis and following renal transplantation (7).

There have also been studies on measurement of decalcification in astronauts after prolonged weightlessness periods (67). In addition, many other studies have been made to collect normative data (17, 24, 61, 71). Studies have also been made on effect of physical activity on bone mineral density (104, 110).

These clinical studies have made use of a number of techniques as well as a number of different bone sites. Thus, it would be beneficial to summarize the different techniques employed in bone mineral determination with emphasis on the assumptions made in addition to the advantages or limitations of each. It should be emphasized that a high degree of precision and versatility in bone sites are possibly the two most important criteria in any technique.

1.2 Current Methods of Bone Mineral Measurement in-vivo:

1.2.1. X-Ray Photodensitometry:

The earliest method of estimating bone density has for some years involved standardised radiographic techniques. In these, the bone site of interest and a reference standard (for example, an aluminium step wedge) are radiographed together (6, 10, 21, 42, 73, 75, 84, 95,

98, 114). For certain parameters reproducibility may be high, but there are limitations in the use of heterogeneous X-rays and of photographic film procedures. The photographic film is nonlinear with respect to both the amount and the energy of the X-rays. Developing the film introduces additional variations. The net result of these problems is that a large change in bone mineral mass (30 to 50 %) must occur between the taking of two X-rays of the same patient before a radiologist can be sure that there has been a change (9, 63). Each of the problems can be reduced by special methods, but the determination of bone mineral mass by this technique has been limited to only a few laboratories in the world since a much higher level of accuracy is required. Some of the limitations of this technique may be avoided by the use of a scintillation counter and radioisotope source emitting homogeneous radiation. The main trouble is estimation of overlying tissue and beam hardening of polychromatic X-ray beam.

1.2.2. Photon Absorptiometry:

The most popular commercially available technique is the one developed by Cameron and Sorenson (8, 9, 10, 11, 12, 13, 99, 101), and utilized by various workers (39, 50, 55, 58, 84a).

The fundamental principle on which all in-vivo measurements of bone density are dependent involves the difference between the attenuation coefficients of bone and

of soft tissue for low energy X- or gamma-radiation.

The maximum discrimination between bone and soft tissue in thin specimens is obtained with photon energies in the region of 30 kev. Cameron and Sorenson (10) have described measurements on the metacarpals using 27.3 kev (tin filtered) radiation from I-125. Several small bone sites have been studied by these and by other authors with this radionuclide (48, 99).

In the case of large bones, however, or for body sites which involve a long total path length for photon beam, the attenuation is considerable and the transmitted count rate correspondingly small. In such cases, a higher photon energy will give better detection statistics in spite of the poorer discrimination between bone and soft tissue. Am-241 radiation, photon energy 59.6 kev, was found to be suitable for bone density measurements in those sites involving a total tissue thickness of 5 to 30 cms., (84a, 114). Hanson (41) has made an analysis of different radionuclides, namely, Gd-153, I-125 & Am-241, for use in absorptiometry studies.

Cameron and Sorenson Photon Absorptiometry technique uses I-125 or Am-241 monoenergetic photon source which is collimated and rigidly coupled with a scintillation detector. The transmission of the well collimated photon beam is recorded as the beam scans across the limb in question at right angles to the long axis of the bone. The absorption in the compact bone gives a measure of its bone

mineral content. The scan measures the count rate relative to the rate with only soft tissue in the beam path. The reference count rate is measured before the scan starts when the beam is positioned over soft tissue next to the bone. The total absorption is then obtained from the area under the curve of counts above the soft-tissue level as a function of position across the bone. By assuming a composition for the bone and determining the volume of the bone from the width of the bone as indicated by the scan, a mineral content per unit length is determined. In order to minimize artifacts produced by variation in soft tissue, the limb being studied is submerged in water which has a photon absorptive coefficient very close to that of soft tissue. Sorenson & Mazess (1960) have made studies on the effect of fat on bone mineral determination using this technique.

For a monoenergetic photon beam of initial intensity I_0 , the intensity after the passage through a thickness T of tissue equivalent material is I :

$$I = I_0 \exp(-\mu_m \rho_m T)$$

where μ_m is the mass absorption coefficient of tissue and ρ_m is the mass density. The intensity I after passage through an equal thickness of bone mineral T_b and tissue T_m , i.e., $T = T_b + T_m$, is given by:

$$I = I_0 \exp(-\mu_m \rho_m T_m - \mu_b \rho_b T_b)$$

$$\text{or } I = I_0^* \exp (- \mu_b \rho_b T_b + \mu_m \rho_m T_b) \quad (1.2)$$

This can be rewritten in terms of the equivalent thickness of compact bone mineral as :

$$T_b = \ln (I_0^* / I) / (\mu_b \rho_b - \mu_m \rho_m) \quad (1.3)$$

By measuring I at closely spaced intervals across the bone, one obtains a series of equivalent thicknesses which by summing gives the equivalent cross-sectional area of compact bone mineral in the bone. From the observed cross-section and known density of apatite one can then determine the mass of bone mineral in a unit length of bone.

This method requires accurate knowledge of the composition of the bone mineral to determine the mass absorption coefficient μ_b . It is further assumed that all non-bone mineral substances absorb radiation to the same degree. In eq.(1.2), the substitution into I_0^* makes use of the assumption that the absorption coefficients for the tissue and tissue-equivalent substance are the same.

A reproducibility of about 2 % is achievable with modern photon absorptiometry bone scanning systems. For longitudinal studies of bone diseases, reproducibility of 2 % is pretty good. And for spot diagnosis, accuracy of about 2 % is normally needed. However, it would seem desirable to have a technique with greater accuracy and more versatility in the bone site location. The method has mostly been used for measurements done on the radius about 10 cm from the distal end (10, 12, 21).

Since photon absorptiometry technique uses low energy radionuclides, scintillation detector has been replaced by a multiwire proportional chamber by some workers in their photon absorptiometry system (43).

The photon absorptiometry technique measures the bone mineral mass as a linear density (81). The unit is gms per cm. To be more specific, the measurement is mass (grams) of mineral ash residue per unit length (cm.) of bone. The bone is taken to be a cylinder (not necessarily circular) which is scanned in a path perpendicular to its axis. The value of bone mineral given by the bone mineral analyzer, based on Photon Absorptiometry, is the value which would be obtained if a 1 cm thick slice were cut out of a bone which is not cylindrical, i.e., the cross-section is not uniform over its length; the measurement value is simply the average linear density over the width of the scan path.

It may be noted that the photon absorption technique provides no information at all about the depth dimension of the bone, i.e., if the bone could be crushed flat, while maintaining the same width dimension, the bone mineral distribution would look identical to the photon beam. Thus there is no way a volume density in terms of g/cm^3 could be determined by this technique alone. However, it is possible to estimate the volume density by combining photon absorptiometric measurements with auxiliary measurements of bone size, such as cortical thickness or total bone width. These may be determined from radiographs

or from measurements of the width of an absorptiometric scan across the bone (48,101).

1.2.3. Dichromatic Photon Absorptiometry:

As an extension of the monochromatic absorptiometry method of Cameron, a dual energy technique has been developed. This method was developed mainly to correct for errors in the Cameron technique due to the presence of adipose tissue (28, 37, 51, 52, 72, 93). It can be shown that this technique gives an expression for bone mineral content, BMC (g/cm) in the following form:

$$\text{BMC (g/cm)} = \frac{\rho_b}{a} L_1^* \left(1 - \frac{c b}{a d} \right)^{-1} \left(1 - \frac{b L_2}{d L_1} \right) \quad (1.4)$$

where $a = \mu_{b1} - \mu_{s1}$, $b = \mu_{f1} - \mu_{s1}$,

$c = \mu_{b2} - \mu_{s2}$, $d = \mu_{f2} - \mu_{s2}$,

and where the subscripts are defined by: b represents bone mineral, s represents soft tissue, f represents fat, and subscripts 1 and 2 refer to photon energies 1 and 2. L is defined as the experimental quantity $\delta \sum \ln(I_0^*/I)$ for the respective energies.

Although eq (1.4) requires the determination of only two constants, c/a and b/d ; these ratios are subject to a certain amount of error. For example, an abrupt change in thickness of the sample can cause a resultant error in the ratios of up to 5% (88).

This method has the following substantial error sources when used trying to correct for tissue: scattering, geometry, composition, and Poisson statistical error (88).

Dichromatic methods have been used for studies on humans to determine the bone-mineral content of the spine (69, 96) and also to determine the fat content of soft tissues (70). Ellis et al (28) have described a dual photon absorptiometry system utilizing a HpGe detector (high purity germanium) and Microprocessor Controller. The controller portion of the microprocessor allows for variation in total scan length, step size, and counting time per step. These options allow for multiple measurements without changes in the detector, source, or collimator detector. Errors in BMC due to multiple small angle scatters are reduced as a consequence of the excellent energy resolution of the HpGe detector which is of the order of 530 ev at 60 kev. The system uses a dual source-holder using 200 mCi I-125 and 100 mCi Am-241.

Mathematically, extending the dichromatic method to the analysis of systems with greater numbers of components by using measurements at greater numbers of photon energies is a simple matter. If there are n components, measurements must be made at n photon energies and a system of n simultaneous equations must be solved. A trichromatic method for determining the amounts of bone mineral, iodine, and soft tissue in the neck was developed by Jacobsen

(47a). His system uses a rotating fluorescence target in an X-ray beam to generate secondary emissions of three different energies and a complex system of moving step wedges. The complexity of this system has precluded its general use.

1.2.4. Coherent & Compton Scattering Technique:

Olkkonen and co-workers (83) introduced a simplified photon scattering method based on the measurement of coherently and compton scattered photons. Recently Kerr et al (60) have also made a detailed study of this method for application in bone mineral determination. It can be shown that the coherently scattered cross-section is given by (40a):

$$\sigma_{\text{coh}} = \text{const.} \cdot |F(q, Z)|^2 \quad (1.5)$$

where $F(q, Z)$ is the form factor. For elastic scattering this is found to be proportional to Z^2 (97). Thus the coherent cross-section is proportional to Z^2 . The number of photons from the incident beam that emerges after coherent scattering is proportional to the number of photons in the incident beam and the number of electrons/cm³ the target (which is proportional to Z & ρ). Thus, the peak area of coherently scattered photons is given by

$$A_{\text{coh}} \propto Z^3 \quad (1.6)$$

The number of compton scattered photons depends upon the number of photons in the incident beam, the compton cross-section and the electron density (which is proportional to Z/A). Thus the peak area of the compton scattered photons is given by :

$$A_{\text{comp}} \propto Z / A \quad (1.7)$$

For the ratio of the peak areas, the following expression is obtained:

$$\frac{A_{\text{coh}}}{A_{\text{comp}}} = \text{Const.} \frac{\bar{Z}^3}{\bar{Z}} \bar{A} \quad (1.8)$$

where $\bar{Z}$ and $\bar{A}$ are the average charge and mass numbers of the target atoms. $\bar{Z}$ and $\bar{A}$ are considerably higher for bone (hydroxyapatite) than for organic components (collagen, fat). Therefore, the $A_{\text{coh}} / A_{\text{comp}}$ ratio is strongly related to the mineral content of bone. The constant in eq.(1.8) is dependent upon attenuation coefficients of the coherent and compton scattered photons.

The measuring procedure consists of the detection of the scattered photons in a narrow beam geometry using a semiconductor detector and a multi-channel analyzer. In the scattering spectrum the peaks due to coherently and compton scattered photons are well separated by the high energy resolution of the solid-state detector.

This technique does not account for any change in attenuation coefficients resulting in a change in thickness or material. It would seem important that an accurate.

determination of attenuation coefficients (both for coherent and compton scattered photons) be made before it is possible to use this technique confidently irrespective of the amount of tissue overlying the measuring site.

1.2.5. In-vivo Neutron Activation Technique:

Another physical technique for measuring bone mineral in-vivo takes advantage of the fact that nearly all of the calcium in the body is in the bones. This technique is called in-vivo activation. The whole body is irradiated with energetic neutrons that convert a small amount of the calcium and some other elements into radioactive forms that give off energetic gamma rays, and the emitted gamma rays are then detected and counted. The gamma rays from radioactive calcium can be identified by their unique energy (3.08 & 4.07 mev), and the number of them indicates the amount of calcium in the body. The amount of bone mineral is then obtained by multiplying by a constant (14, 18, 19, 41a, 72a, 105). Both the source of neutrons and the special whole-body detecting equipment are expensive and not practical for routine clinical measurements. In addition, because the relatively large radiation exposure is a hazard, it is undesirable to use the technique on healthy subjects to obtain normal data.

1.2.6. In-vivo Proton Activation Technique:

Eilbert (27) has described the method of Proton Activation Analysis for in-vivo calcium measurement. Proton activation of Ca-40 in bone produces 2.17 Mev gamma ray emitting K-38 with a half-life of 7.7 minutes. This in turn is counted in a low background whole body counting assembly. The technique makes use of a 160 Mev proton cyclotron and its main advantages include direct measurement of calcium in the lumbar vertebrae and the restriction of the dose to the immediate vertebral region due to finite range of protons and their low transverse scattering because of large mass.

1.2.7. Radionuclide Uptake Technique:

Tc-99^m labelled phosphate/diphosphate compounds as well as F-18 widely used in Radio-isotope scanning for bone diseases, have suitable properties for use in studies of bone metabolism (34, 77). A shadow-shield whole-body counter has been used to measure whole-body retention of μ Ci activities of Tc-99^m HEDP/EHDP in normal males up to 6-hours after injection (4, 3). F-18 scintigraphy has been performed using dual headed scanner with 5" x 5" Sodium iodide crystals (77). The short half-life of F-18 results in a lower radiation dose per scan per patient permitting frequent measurements of the affected areas.

1.2.8. Trans-imaging Technique using Rectilinear Scanner/ Gamma Camera :

Trans-imaging as a technique for determining bone mineral content (BMC) has been used by a few workers (90, 108). The area of bone to be studied by trans-imaging is placed between a source of gamma rays or X-rays and the large surface scintillation detector of a gamma camera (79a). The entire bone region of interest is illuminated by a broad beam of radiation and a shadow image of this region is displayed by the gamma camera system. With the large area of the gamma camera detector, relatively small radiation sources could be used. In another system, the two photon attenuation technique has been utilized with a modified dual-headed whole-body clinical rectilinear scanner to make rectilinear scans of the vertebral column (90). Here the source/detector system consists of a 2-Ci Gd-153 source and 2 x 0.5 inch NaI(Tl) crystal detectors. The 44- and 100-keV lines of Gd-153 have been shown to be optimum for the measurement of the vertebral bone mineral content.

Rectilinear scans allow visualization of the vertebrae, thus allowing use of internally recognizable and relocateable fiducial structures for localization. With this technique the linear density (g/cm), which presents some difficulty in interpretation when applied to irregular objects, should be replaced by an estimate of the total BMC in the sampled region. Precision of the order of $\pm 5\%$ is achievable with these trans-imaging techniques.

1.2.9. Computed Tomography for Bone Mineral Determination:

Computed Tomography, (CT) provides a new modality for bone mineral determination in-vivo. In principle, Computed Tomography determines the spatial distribution of linear attenuation coefficients in a thin slice of the body. The attenuation values (CT numbers) represent the average attenuation of all substances contained in each pixel. The usual CT methods cannot reliably distinguish between pixels that contain more than one substance because of volume averaging. This difficulty is overcome by taking CT scans at two different kVp settings e.g., 140 kVp (28 mA) and 100 kVp (40 mA) (116, 117). The mass of calcium in each pixel can be related to the difference in the CT numbers measured at the two kVp settings with the assumption that the CT numbers for soft-tissue are the same. The high cost and large radiation dose of CT procedures make alternate methods attractive because similar information is provided at approximately one-hundredth to one-thousandth of the radiation dose. Some workers (3, 29) have also reported in-vivo bone density measurement using single energy gamma-ray Computed Tomography using Radionuclides.

In Computed Tomography, the subject is scanned trans-axially at a selected site with a narrow beam of X-rays or gamma-rays from a radionuclide. The intensity of the transmitted beam is measured at intervals, and the linear translation movement of source and detector (s) is repeated at different angles. The distribution of the

attenuation coefficients is calculated by the dedicated computer from the measured transmission data. - The resulting matrix of over 16,000 local linear attenuation coefficients is visualized in the form of a cross-sectional (density) image (46).

1.2.10. Photon Scattering Technique :

In the techniques described above there is the disadvantage that either some assumption has to be made about the shape of the bone and its chemical composition or the measuring system may be highly expensive.

A method for determining the electron and physical density in bone tissue which does not require a prior knowledge of the shape of the bone and the surrounding tissue has been developed by Clarke & Van Dyk (15, 16) and also by Kennett & Webber (56, 59, 111, 112, 113). Their approach for measuring density with the use of compton scattered photons allowed the study of irregular shaped bones without requiring any detailed knowledge of the surrounding material. The method requires the measurement of the ratio of compton scattered to unscattered gamma rays, as a beam of collimated gamma rays passes through the region under investigation. The use of a quasi-continuous X-ray spectrum as the source of photons was introduced by Mortimer (76) and was further developed by White (115). These workers concluded that by using X-rays rather than a radioactive source, it would be possible to reduce the

exposure time due to the increased source strength, study smaller bones since smaller collimators may be used, and use the X-rays to locate the area under study.

Clarke & Van Dyk (15) have described two scattering methods for bone density measurement which are referred to as the high-energy method and the low-energy method. In the high-energy method use is made of two monoenergetic sources namely Co-60 (mean energy 1.25 kev) and Cs-137 (0.662 mev) with 50° scattering angle. Kennett, Garnett & Webber (56, 59) used Au-198 (0.412 mev) and Hg-203 (0.279 mev) as photon sources. Due to high dosages involved, the high-energy method was not used, however, for in-vivo bone density measurements. In the low-energy method, the photon sources used have energy much less than 0.51 Mev so that the Compton energy shift is negligible for any desired scattering angle θ . Kennett et al (56) used Sm-153 (103 kev) and Tm-170 (84 kev) as the two photon sources. Mortimer (76) and White (115) both used a single 90 kVp X-Ray machine as the photon source in their studies. From a theoretical analysis of the procedure, it has been found by Garnett et al (36) that for clinical use the optimum energy of the primary photon beam is close to 90 kev and the system performance drops off by a factor of two at about 200 kev. Piper et al (86, 87) have concluded that the combinations Gd-153 (100 kev) Tm-170 (84 kev) and Ce-144 (134 kev) Co-57 (122 kev) have the highest potential in terms of system performance and dose reduction to the specimen under study.

In-vitro measurements of density using photon scattering techniques have shown an excellent correlation with density measured by Archimedes principle and the precision of density measurements is high, usually of the order of about 1.5 to 2 % in most cases (15, 36, 44, 56, 76, 87, 91, 92, 112, 113, 115).

Dyck (25) has described some of the limitations of single photon compton scattering method for in-vivo monitoring of skeletal mineralization. Farmer et al (32) have also described limitations of the technique.

Luther (65) has used the compton scattering technique to find correlation between os calcis & spinal bone densities. Leichter (66) et al have also utilized the technique for bone density measurements.

The present study is a continuation of the preliminary work carried out by Mortimer and White using X-rays. The apparatus has been modified to overcome some inherent problems in their design. The main problem encountered in their design was to prevent photomultiplier tube saturation of the detector looking at the high flux X-ray transmission beam. About 1 mm thick lead attenuator was used in front of the detector directly facing the X-ray source. In second source position, the roles of the two detectors are reversed, and in this case, the same attenuator was removed from the first detector and put in front of the second. The presence of lead attenuator leads to undesirable beam hardening effects and the density

measurements become critically dependent on the lead material thickness chosen. In the present system a third detector using plastic scintillator and photomultiplier tube base circuit with higher current rating has been incorporated and the high flux transmission beam is detected by this detector in either source position. The other two detectors, using CsI(Na) crystals, are only used for detecting the low flux scattered beam. No lead attenuator is needed in this design.

It will be worthwhile to outline the two methods described by Clarke and Van Dyk for density measurements using Compton Scattering technique.

1.2.10.1. The High-Energy Method:

The principle of high-energy method as outlined by Clarke & Van Dyk (15) can be understood from Fig.1.1. Two collimated sources of photons (S_1 and S_2) are used of intensities I_1 and I_2 . Transmission measurements are made by collimated detectors : I_{T1} along path 1X3, for source S_1 and I_{T2} along path 2X4 for source S_2 . These are given by

$$I_{T1} = I_1 A_{1X} A_{X3} D_1 \quad (1.9)$$

$$I_{T2} = I_2 A_{2X} A_{X4} D_2 \quad (1.10)$$

D_1 & D_2 are the detection efficiencies of D_1 and D_2 detectors respectively. The two beam paths, 1X3 and 2X4, intersect at the point X, with an angle θ between them. A_{1X}

FIG. 1.1. CONFIGURATION OF SOURCES AND DETECTORS FOR ELECTRON DENSITY MEASUREMENT USING THE HIGH-ENERGY TECHNIQUE. (a) INITIAL CONFIGURATION; (b) SAMPLE ROTATED THROUGH 180° .

and A_{x3} are the transmission factors corresponding to beam energy E_1 , and A_{2x} and A_{x4} correspond to E_2 .

The detector D_2 now measures scattered intensity at angle θ from source S_1 . This is given by

$$I_{S_1} = I_1 A_{1x} A_{x4} D_2 G \rho \quad (1.11)$$

G is a factor made up of the Compton cross-section for E_1 scattered through angle θ , and the number of electrons per cm^3 of the selected volume. The energy shift of E_1 for scattering at angle θ is given by the well known Compton Equation (30,64) and the angle θ is so chosen that this shift corresponds to energy E_2 of the second source S_2 .

The test body is then rotated through 180° to give a second scattering measurement with intensity

$$I_{S_2} = I_1 A_{3x} A_{x2} D_2 G \rho \quad (1.12)$$

We assume that $A_{2x} = A_{x2}$ and $A_{3x} = A_{x3}$

From eqns (1.9) to (1.12) we obtain

$$\rho = \left[(I_2 D_1 / I_1 D_2) \times (I_{S_1} I_{S_2} / I_{T_1} I_{T_2}) \right]^{1/2} \quad (1.13)$$

1.2.10.2. The Low-Energy Method:

In the low-energy method, the Compton energy shift being negligible, a single source can be used and it is moved from a first to a second source location to obtain the desired readings. The subject is no longer rotated. This can be seen in Fig.1.2. Eqns (1.9) and (1.10) apply with I_1 and I_2 replaced by a single source strength I ie

FIG. 1.2. CONFIGURATION OF SOURCES AND DETECTORS FOR ELECTRON DENSITY MEASUREMENT USING THE LOW-ENERGY TECHNIQUE. (a) INITIAL CONFIGURATION; (b) SOURCE MOVED TO SECOND LOCATION.

$$I_{T1} = I A_{1x} A_{x3} D_1 \quad (1.14)$$

$$I_{T2} = I A_{2x} A_{x4} D_2 \quad (1.15)$$

The scattering equations are now given by

$$I_{S1} = I A_{1x} A_{x4} D_2 G \quad (1.16)$$

$$I_{S2} = I A_{2x} A_{x3} D_1 G \quad (1.17)$$

Thus from eqns (1.14) to (1.17), we obtain

$$G = [I_{S1} I_{S2} / I_{T1} I_{T2}]^{1/2} \quad (1.18)$$

The assumption made is that A_{x3} and A_{x4} appearing in eqns. (1.16) and (1.17) are identical with the corresponding quantities appearing in eqns (1.14) and (1.15).

The same measurements are repeated on a calibration standard such as water or lucite. Combining these measurements with those on sample, one then gets the following expression for electron density* of the sensitive volume in the sample:

$$d_e = \frac{G(\text{Sample})}{G(\text{Water})} = \left[\frac{I_{S1} I_{S2}}{I_{T1} I_{T2}} \right]_{\text{Sample}}^{1/2} \times \left[\frac{I_{T1} I_{T2}}{I_{S1} I_{S2}} \right]_{\text{Water}}^{1/2} \quad (1.19)$$

The relation between d_e and Archimedes mass density, d , is given by

$$d_e = \frac{\sum_i Z_i a_i}{\sum_i A_i a_i} \times \frac{d}{0.55} \quad (1.20)$$

where Z_i and A_i are the atomic numbers and mass numbers of the constituent elements, and the a_i 's are the numbers of atoms per molecule of each element. The coefficient 0.55 comes from the value of $\sum Z_i a_i / \sum A_i a_i$

* implies specific electron density

for the water standard.

1.3. Compton Scattering Methods vs Transmission

Techniques:

It has been pointed out in section 1.2.2 that a transmission measurement, such as obtained using Cameron photon absorptiometry, yields the mineral mass within a cylinder of bone and expressed in units of g/cm. A scattering measurement, however, yields the average density within a selected volume which will include both the inorganic and organic phases of bone (111). The Compton scattering method offers the advantage of measuring absolute density expressed in g/cm³, independent of sample size and thickness of surrounding tissue. Most importantly, the selected volume can be restricted to trabecular bone by proper collimation and positioning, and it should be possible to obtain accurate and precise measurement of changes in density within the trabecular bone.

CHAPTER 2

PRINCIPLES OF COMPTON SCATTER X-RAY BONE DENSITOMETER

2.1. Introduction:

Gamma rays or X-rays of energy less than about 1 Mev will most likely interact with matter either photoelectrically or by Compton scattering. In Compton scattering an incident photon is scattered by an orbital electron. The angle through which such a photon is scattered determines the energy of the scattered photon. Thus, photons from a polyenergetic incident beam scattered through a given angle will also be polyenergetic.

The electron density of a material is related to the microscopic Compton scattering cross-section which may be determined by measurement of a photon beam scattered from a well defined scattering volume. Electron densities are directly related to mass densities through the use of the effective ratio of the atomic number to the atomic weight of the material.

2.2. Compton Scattering:

In the Compton scattering process, the electrons are treated as free (i.e. unbound) and therefore scatter photons independently of one another (i.e. incoherently); therefore

there are random phase relations between waves scattered by different electrons (33). The intensity of incoherently scattered radiation at any point in space is the sum of the intensities of radiation scattered from independent scattering elements (electrons in this case).

The energetics of compton scattering by a free electron may be viewed as the elastic collision between a photon and an unbound electron which is at rest in the laboratory frame, and is thus governed by the energy & momentum conservation laws. Fig. 2.1 is the schematic diagram of the elastic collision process between a photon and a free electron which is at rest (30). Let the energy of the incident photon be $E = h\nu$ and the momentum $p = h\nu/c$. The scattered photon is emitted at an angle θ with an energy $E' = h\nu'$ and momentum $p' = h\nu'/c$, while the electron recoils at an angle ϕ with a kinetic energy $T = (mc^2 - m_0 c^2)$ and momentum $p_e = m\beta c$, where $\beta = v/c$, and $m = m_0 / \sqrt{1 - \beta^2}$, m_0 being electron rest mass.

The mathematical statements of the conservation of energy and linear momentum yield the following equations:

$$h\nu = h\nu' + mc^2 - m_0 c^2 \quad (2.1)$$

$$h\nu/c = \frac{h\nu'}{c} \cos\theta + m\beta c \cos\phi \quad (2.2)$$

$$0 = \frac{h\nu'}{c} \sin\theta - m\beta c \sin\phi \quad (2.3)$$

If different variables are eliminated in turn, the following expressions can be obtained from these equations:

FIG. 2.1. (a) SCHEMATIC DIAGRAM OF COMPTON SCATTERING OF A PHOTON OF ENERGY $h\nu$ BY A FREE ELECTRON.
(b) SCHEMATIC DIAGRAM TO SHOW DIFFERENTIAL CROSS SECTION PER UNIT SOLID ANGLE.

(a) Wavelength shift:

$$\lambda' - \lambda = \Delta\lambda = \frac{h}{m_0 c} (1 - \cos\theta) \quad (2.4)$$

(b) Scattered photon energy:

$$h\nu' = \frac{h\nu}{1 + \frac{h\nu}{m_0 c^2} (1 - \cos\theta)} \quad (2.5)$$

$$\text{or } h\nu' = \frac{h\nu}{1 + \alpha (1 - \cos\theta)} \quad (2.5a)$$

(c) Kinetic energy of the recoil electron:

$$T = h\nu - h\nu' = h\nu \frac{\alpha (1 - \cos\theta)}{1 + \alpha (1 - \cos\theta)} \quad (2.6)$$

(d) Relation between scattering angles:

$$\cot\phi = (1 + \alpha) \tan \frac{\theta}{2} \quad (2.7)$$

It can be noted from these equations that as the incident photon energy increases a small span of scattered-electron angles corresponds to a large span of scattered-photon angles. Further, as the incident photon energy increases, a large fraction of the total energy is given to the recoil electron for a given photon-scattering angle.

The differential cross-section for the scattering of the radiation with free electrons is given classically by the Thomson cross-section (30):

$$\frac{d\sigma_T}{d\Omega} = \frac{1}{2} r_0^2 (1 + \cos^2\theta) \quad (2.8)$$

where $r_0 = e^2/m_0c^2$ is the classical electron radius.

The correct expression for the differential electronic cross-section for the probability of a photon being scattered into a unit solid angle about angle θ has been calculated from quantum mechanics by Klein and Nishina (57). It can be written in a number of ways. Two of the more common ways are (33):

$$\frac{d\sigma_t}{d\Omega} = \frac{1}{2} r_0^2 \left[\frac{1}{1 + \alpha(1 - \cos\theta)} \right]^2 \left[1 + \cos^2\theta + \frac{\alpha^2(1 - \cos\theta)^2}{1 + \alpha(1 - \cos\theta)} \right] \quad (2.9)$$

$$\frac{d\sigma_t}{d\Omega} = \frac{3}{16\pi} \frac{\lambda^2}{(\lambda')^2} \left(\frac{\lambda}{\lambda'} + \frac{\lambda'}{\lambda} - \sin^2\theta \right) \quad (2.10)$$

In Eq.(2.10), λ is in units of (h/m_0c) , which is often called the Compton wavelength of the electron.

The scattering component of the total differential cross-section can be obtained by weighting the total cross-section with the fractional energy carried off by the scattered photon. Using eq.(2.5a), the expression for the scattering component of the total differential cross-section is given by:

$$\frac{d\sigma_s}{d\Omega} = \frac{1}{2} r_0^2 \left[\frac{1}{1 + \alpha(1 - \cos\theta)} \right]^3 \left[1 + \cos^2\theta + \frac{\alpha^2(1 - \cos\theta)^2}{1 + \alpha(1 - \cos\theta)} \right] \quad \dots\dots\dots(2.11)$$

If eq.(2.9) is regarded as a differential number scattering and absorption probability, eq.(2.11) will be recognized as a differential energy scattering probability.

To obtain the total (nondifferential) Compton electronic

cross-section, integrate eq.(2.9) over all scattering angles $(0 \rightarrow \pi)$.

$$e^{\sigma_t} = 2\pi r_0^2 \left\{ \frac{1+\alpha}{\alpha^2} \left[\frac{2(1+\alpha)}{1+2\alpha} - \frac{\ln(1+2\alpha)}{\alpha} \right] + \left[\frac{\ln(1+2\alpha)}{2\alpha} - \frac{1+3\alpha}{(1+2\alpha)^2} \right] \right\} \quad (2.12)$$

The scattering component can be written

$$e^{\sigma_s} = 2\pi r_0^2 \left[\frac{4\alpha^2}{3(1+2\alpha)^3} - \frac{1+\alpha}{\alpha^2(1+2\alpha)^2} (1+2\alpha - 2\alpha^2) + \frac{1}{2\alpha} \ln(1+2\alpha) \right] \quad (2.13)$$

The difference between these two is the absorption cross-section.

At low energies a Compton scattering interaction involves very little energy loss by the photon. In the limit, as the photon energy goes to zero, i.e., for $\alpha \ll 1$, the differential cross-section reduces to

$$\frac{d\sigma_s}{d\Omega} = \frac{1}{2} r_0^2 (1 + \cos^2\theta) \quad (2.14)$$

This limiting case is the well known Thomson differential cross-section. The integral of this differential cross-section over all solid angles yields

$$e^{\sigma_s} = \frac{8\pi}{3} r_0^2 = \frac{8\pi}{3} (e^2 / m_0 c^2)^2 = 6.65 \times 10^{-25} \text{ cm}^2/\text{electron} \quad \dots(2.15)$$

This value is often called the Thomson classical scattering cross section for a free electron. It is seen to be independent of photon energy.

For small values of α , eq.(2.12) reduces to

$$\sigma_t = \frac{8\pi}{3} r_0^2 (1 - 2\alpha + 5.2\alpha^2 - 13.3\alpha^3 + \dots) \text{ cm}^2/\text{electron} \quad (2.16)$$

The average (or total) collision cross-section is the probability of removal of the photon from a collimated beam while passing through an absorber containing one electron/cm².

In the present study, the maximum energy that a photon may have is about 90 kev. In this case, $\alpha \approx .18$, which justifies the use of the classical cross-section.

2.3. Other Interactions:

In addition to Compton scattering, photons may interact with matter through the photoelectric effect, coherent scattering and pair production. Pair production requires a minimum photon energy of 1.02 Mev and thus cannot occur in the present energy range. For soft tissue, (H,C,N, and O) there is no contribution from photoelectric effect or coherent scattering larger than about 1% of the total interaction cross-section in the present energy range (5). For bone, coherent scattering accounts for less than 10% of the scattering, and photoelectric effect about 20% of the total cross-section (5). In both cases the remainder is Compton scattering.

2.4. Determination of Bone Density:

The method used in this study was developed by Clarke and Van Dyk (15) and described as the photon

scattering technique. Both High-Energy and Low-Energy methods have been examined by various workers (15, 45, 56, 59, 112).

The technique used in this study makes use of the low-energy method and employs a single polychromatic photon source (an X-ray Machine) and three detectors, one to measure the count rate of transmitted photons and the other two to measure the count rate of scattered photons. The configuration of the source and detectors is shown in Fig.2.2(a). As can be seen in this diagram, the projection of the collimator holes defines a small volume within the sample under study. This volume is referred to as the sensitive volume. It is the volume which is common to both the scattered and transmitted beams at the point where the scattered and transmitted beams intersect.

If the beams are manipulated in such a way that for two separate exposures the scattered and transmitted beams follow the same path, then the equations involving attenuations and electron densities become solveable.

Considering Fig.2.2(a), the intensity of the incident beam as it impinges on the sensitive volume is

$$I_1 e^{-\mu_1 t}$$

where

I_1 is the intensity of Source 1
 μ_1 is the total linear attenuation coefficient for the incident beam

$$P_e = \left\{ \sqrt{\frac{S_1 S_2}{T_1 T_2}} \right\}_{\text{SAMPLE}} \div \left\{ \sqrt{\frac{S_1 S_2}{T_1 T_2}} \right\}_{\text{STANDARD}}$$

FIG. 2.2. CONFIGURATION OF SOURCES & DETECTORS FOR ELECTRON DENSITY MEASUREMENT
(a) INITIAL CONFIGURATION; (b) SOURCE MOVED TO SECOND LOCATION.

t is the distance indicated in Fig.2.2 (a).

The sensitive volume acts as a source of intensity $I_1 e^{-\mu_1 t}$. The transmitted photons will be attenuated over a distance 'x' and will be detected at D3 with a counting rate given by

$$T_1 = D_3 I_1 e^{-\mu_1 (t + x)} \quad (2.17)$$

where D_3 also denotes the detection efficiency of detector D3.

The number of scattered photons for a given angle θ from the sensitive volume depends on:

- (a) the Compton cross-section, $\sigma(\theta) \approx \sigma(\text{Thomson})$
- (b) the intensity at the sensitive volume, $I_1 e^{-\mu_1 t}$
- (c) the electron density of the sample, $\rho_e = \frac{N_0 \bar{Z}}{\bar{A}} \rho_m$

where

$\bar{Z}$ is effective atomic number

N_0 is Avogadro's number

ρ_m is mass density

$\bar{A}$ is effective atomic weight.

Then the number of scattered photons is

$$n = k I_1 e^{-\mu_1 t} e^{\sigma} \frac{N_0 \bar{Z}}{\bar{A}} \rho_m \quad (2.18)$$

k being constant; or

$$n = K \frac{\bar{Z}}{\bar{A}} I_1 e^{-\mu_1 t} \rho_m \quad (2.19)$$

K being another constant.

There will be attenuation of these photons over the distance y , resulting in counting rate at detector D1 of:

$$S_1 = D_1 I_1 e^{-\mu_1 z} \frac{z}{\bar{A}} K \rho_m e^{-\mu_2 y} \quad (2.20)$$

where D_1 is the detection efficiency of detector D_1 and μ_2 is the total linear attenuation coefficient for the scattered beam.

The source is then moved to the second location S as shown in Fig. 2.2(b). Detector D_3 takes the position of D_1 , D_1 moves to original source position S_1 , and detector D_2 takes the previous position of detector D_3 . The intensity of the source is the same as in the above consideration. In this configuration, the transmitted counting rate is:

$$T_2 = D_3 I_1 e^{-\mu_1 (z+y)} \quad (2.21)$$

The scattered counting rate is found as in eq. (2.20) and becomes:

$$S_2 = D_2 I_1 e^{-\mu_1 z} K \frac{z}{\bar{A}} e^{-\mu_2 x} \rho_m \quad (2.22)$$

From eqs. (2.17, 2.20, 2.21, and 2.22), one obtains the ratio:

$$\frac{S_1 S_2}{T_1 T_2} = \frac{D_1 D_2}{(D_3)^2} K^2 \rho_m^2 \left(\frac{z}{\bar{A}}\right)^2 \frac{e^{-\mu_2 (x+y)}}{e^{-\mu_1 (x+y)}} \quad (2.23)$$

To simplify the above ratio, it is necessary to have $\mu_1 = \mu_2$. Since total linear attenuation coefficients are energy dependent (10) it is necessary that the transmitted and scattered beams have

the same energy distribution. In this situation, the ratio gives the relation:

$$\left[\frac{S_1 S_2}{T_1 T_2} \right]^{1/2} = \left[\frac{(D_1 D_2)^{1/2}}{D_3} \right] \cdot \frac{Z}{\bar{A}} \cdot K \cdot \rho_m \quad (2.24)$$

The constant in eq.(2.24) is measured using a sample of known density, usually water or lucite. With water as standard, one obtains the ratio

$$\left[\frac{S_1 S_2}{T_1 T_2} \right]_{\text{water}}^{1/2} = \left[\frac{(D_1 D_2)^{1/2}}{D_3} \right] \cdot \left[\frac{Z}{\bar{A}} \right]_{\text{water}} \cdot K \quad (2.25)$$

Dividing eq(2.24) for sample by eq.(2.25) for water standard, the following relation is obtained for the mass density of sensitive volume of the sample:

$$\rho_m = \frac{\left(S_1 S_2 / T_1 T_2 \right)_{\text{Sample}}^{1/2}}{\left(S_1 S_2 / T_1 T_2 \right)_{\text{Water}}^{1/2}} \times \frac{\left(Z / \bar{A} \right)_{\text{Water}}}{\left(Z / \bar{A} \right)_{\text{Sample}}} \quad (2.26)$$

If, however, lucite is chosen as the standard (this is what has been taken as standard in our study here) its mass density is 1.185, and in this case, the following relation holds:

$$\rho_m = \frac{\left(S_1 S_2 / T_1 T_2 \right)_{\text{Sample}}^{1/2}}{\left(S_1 S_2 / T_1 T_2 \right)_{\text{Lucite}}^{1/2}} \times \frac{\left(Z / \bar{A} \right)_{\text{Lucite}}}{\left(Z / \bar{A} \right)_{\text{Sample}}} \times 1.185 \quad (2.27)$$

Thus the four measured counting rates on the sample

result in the desired quantity.

2.5. Error Estimation due to Poisson statistics on Density Measurement with Compton Bone Densitometer;

The measured density with Compton X-ray Bone Densitometer is determined from eq.(2.26) or eq.(2.27). To estimate the precision of measurement let us consider eq.(2.24) which can be written as:

$$P_X = \left(\frac{S_1 S_2}{T_1 T_2} \right)_X = S_X^2 \left[\frac{D_1 D_2}{D_3^2} \right] \quad (2.28)$$

for the sample, and

$$P_S = \left(\frac{S_1 S_2}{T_1 T_2} \right)_S = S_S^2 \left[\frac{D_1 D_2}{D_3^2} \right] \quad (2.29)$$

for the water standard,

where

$$S_X = (\bar{Z} / \bar{A})_X \cdot K \cdot \rho_m \quad (2.30)$$

$$S_S = (\bar{Z} / \bar{A})_S \cdot K \cdot 1 \quad (2.31)$$

From eqs.(2.28) & (2.29), one may write

$$S_X = S_S \sqrt{\frac{P_X}{P_S}} \quad (2.32)$$

Assuming S_S for the standard can be determined perfectly, one has for the relative error in S_X , the following expression:

$$\left(\frac{\sigma_{S_X}}{S_X} \right)^2 = \frac{1}{4} \left[\left(\frac{\sigma_{P_X}}{P_X} \right)^2 + \left(\frac{\sigma_{P_S}}{P_S} \right)^2 \right] \quad (2.33)$$

Again using eq.(2.28), one has

$$\begin{aligned} \left(\frac{\sigma_{P_X}}{P_X} \right)^2 &= \left(\frac{\sigma_{S_{1X}}}{S_{1X}} \right)^2 + \left(\frac{\sigma_{S_{2X}}}{S_{2X}} \right)^2 + \left(\frac{\sigma_{T_{1X}}}{T_{1X}} \right)^2 + \left(\frac{\sigma_{T_{2X}}}{T_{2X}} \right)^2 \\ &= \left[\frac{1}{S_{1X}} + \frac{1}{S_{2X}} + \frac{1}{T_{1X}} + \frac{1}{T_{2X}} \right] \end{aligned} \quad (2.34)$$

and a similar expression holds for relative error in P_S .

We have made use of the well known relation $\sigma = \sqrt{N}$

for the standard deviation in count rate N due to

Poisson statistics.

S_{1X} , S_{2X} and T_{1X} , T_{2X} stand for the scattered and transmission count rates respectively, for the sample in the measuring position. Similarly we will have count rates for the standard in the measuring position and these are denoted by S_{1S} , S_{2S} , T_{1S} , T_{2S} .

Substituting eq.(2.34) and similar expression for the standard, in eq.(2.33), we have

$$\left(\frac{\sigma_{S_X}}{S_X} \right)^2 = \frac{1}{4} \left[\frac{1}{S_{1X}} + \frac{1}{S_{2X}} + \frac{1}{T_{1X}} + \frac{1}{T_{2X}} + \frac{1}{S_{1S}} + \frac{1}{S_{2S}} + \frac{1}{T_{1S}} + \frac{1}{T_{2S}} + \frac{L}{L} \right] \quad (2.35)$$

Since different scatter count rates observed are of the same order of magnitude, to a first approximation, we can assume that

$$S_{1X} = S_{2X} = S_{1S} = S_{2S} = S \quad (\text{say})$$

Similarly for the transmission count rates, assume

$$T_{1X} = T_{2X} = T_{1S} = T_{2S} = T \quad (\text{say})$$

We then have, as a first approximation,

$$\frac{\sigma_{S_X}}{S_X} \approx \sqrt{\frac{1}{S} + \frac{1}{T}} \quad (2.35a)$$

Further, the scattered count rates being orders of magnitude

smaller than the transmission count rates, one may ignore the second term on right hand side of eq.(2.34) and one then gets

$$\frac{\sigma_{S_X}}{S_X} \approx \sqrt{\frac{1}{S}} \quad (2.35b)$$

It therefore follows that the relative error in determination of the density of the sample (ρ_m being contained in the expression for S_X) is mainly governed by the scattered count rates observed in detectors D1 & D2.

In deriving eq.(2.35), we have not taken account of the background count rates in the detectors. Assuming again that the background count rate in transmission detector D3 is very small compared to true transmission rate, and if S_{Total} denotes the observed scattered count rate including the background, B.G., one may write

$$\frac{\sigma_{S_X}}{S_X} = \sqrt{\frac{\sigma_{Total}^2 + \sigma_{B.G.}^2}{S^2}} = \sqrt{\frac{S_{Total} + B.G.}{S}} \quad (2.36)$$

In the in-vitro experiments described in Chapter 6, we observe typically the following count rates:

Total Scatter counts (including B.G. : 5000

B.G. : 100

In this situation, one gets

$$\sigma_{S_X} / S_X = \frac{\sqrt{5000 + 100}}{4900} \approx 1.5 \%$$

The scatter count rates for samples with comparatively high ρ and Z values may be quite low,

and in such cases it is essential to keep the background count rate as low as possible. Various methods for background noise reduction have been described in chapter 3.

2.6. Beam Hardening:

When a polyenergetic X-ray beam passes through a body, the exit intensity may be expressed as:

$$I = \int_0^{E_0} I_0(E) \exp \left[- \mu(E) X \right] dE \quad (2.37)$$

where $I_0(E)$ is the energy dependent initial photon intensity and $\mu(E)$ is the linear attenuation coefficient of the material along the path. X indicates the length of the material and E_0 is the maximum energy in the X-ray spectrum. The projection along the beam path is then given by

$$P_{poly}(X) = \ln(I_0 / I) \quad (2.38)$$

where

$$I_0 = \int_0^{E_0} I_0(E) dE \quad (2.39)$$

is the incident intensity. In the monoenergetic case, $I_0(E)$ is equal to:

$$I_0 \delta(E_1 - E),$$

and upon substituting this into eq.(2.37) and using eqs.(2.38) & (2.39), one obtains the monoenergetic projection P_{mono} which is given by

$$P_{mono} = \mu(E) X \quad (2.40)$$

Let us assume that the object of interest consists of only bone (or any other sample under study) and tissue (or lucite material surrounding the sample under study). We will assume that the beam enters into a slab of bone (or any other sample) of thickness x_B and after passing through bone (or sample) enters into a slab of tissue (or lucite) of thickness x_T . The beam then passes through a lead (Pb) attenuator covering the front end of transmission detector after it comes out of tissue.

The energy weighted attenuation coefficient of bone (or sample) at a distance x'' from the entrance point of the beam is given by

$$\mu_B(x'') = \frac{\int_0^{E_0} \mu_B(E) I_0(E) \exp(-\mu_B(E) x'') dE}{\int_0^{E_0} I_0(E) \exp(-\mu_B(E) x'') dE} \quad (2.41)$$

If the beam enters into tissue after passing through an x_B thickness of bone (or sample), the energy weighted tissue attenuation coefficient at a point x' from the bone-tissue (or sample-lucite) interface is given by

$$\mu_T(x_B, x') = \frac{\int_0^{E_0} \mu_T(E) I_0(E) \exp(-\mu_B(E) x_B - \mu_T(E) x') dE}{\int_0^{E_0} I_0(E) \exp(-\mu_B(E) x_B - \mu_T(E) x') dE} \quad (2.42)$$

Now if the beam enters into lead attenuator after having passed through an x_T thickness of tissue (or lucite), the energy weighted lead attenuation coefficient at a point x from the tissue-lead (or lucite-lead) interface is given by:

$$\tilde{\mu}_L(X_T, x) = \frac{\int_0^{E_0} \mu_L(E) I_0(E) \exp(-\mu_B(E) X_B - \mu_T(E) X_T - \mu_L(E) x) dE}{\int_0^{E_0} I_0(E) \exp(-\mu_B(E) X_B - \mu_T(E) X_T - \mu_L(E) x) dE} \quad (2.43)$$

Integrating and adding eqns.(2.41), (2.42) & (2.43) yields

$$P(X_B, X_T, X_L) = \int_0^{X_B} \tilde{\mu}_B(x'') dx'' + \int_0^{X_T} \tilde{\mu}_T(X_B, x') dx' + \int_0^{X_L} \tilde{\mu}_L(X_T, x) dx \quad (2.44)$$

If one substitutes eqns.(2.41), (2.42) & (2.43) into eq.(2.44) and performs the integration, it can be seen that $P(X_B, X_T, X_L)$ is identical to P_{poly} as given in eq.(2.38) when there is only bone (or sample), tissue (or lucite) and lead present along the beam. The distance integration in eq.(2.44) does not imply any linearity since the energy weighted attenuation coefficients are strong functions of distance even for three homogeneous materials.

For a monoenergetic beam

$$I_0(E) = I_0 \delta(\tilde{E} - E)$$

and eqs.(2.41), (2.42) & (2.43) become

$$\tilde{\mu}_B(X_B) = \mu_B(\tilde{E}) \quad (2.45)$$

$$\tilde{\mu}_T(X_B, X_T) = \mu_T(\tilde{E}) \quad (2.46)$$

$$\tilde{\mu}_L(X_B, X_T, X_L) = \mu_L(\tilde{E}) \quad (2.47)$$

From eqs.(2.45), (2.46) & (2.47), it is seen that the monoenergetic coefficients do not depend on the thickness of the material along the path. The polyenergetic coefficients $\tilde{\mu}_B$, $\tilde{\mu}_T$ and $\tilde{\mu}_L$ (eqs.2.41,2.42,2.43), on the other hand, do depend on the amount of material

traversed.

Let us assume that the thicknesses of bone (or sample) and tissue (or lucite) traversed by the X-ray beam are such that the initial 40 kev average beam energy shifts to 60 kev average energy. In that situation, the transmission factor increases by approximately 12 % through the tissue (or lucite). However the differential scattering cross-section also changes by about 12 % at 90° scattering angle for a change of energy from 40 kev to 60 kev as can be seen from the following:

$$\begin{aligned} \frac{d\sigma}{d\Omega} &= \frac{r_0^2}{2} p^2 (1 - p + p^2) = \frac{r_0^2}{2} \frac{1}{(1 + \alpha)^2} \left(1 - \frac{1}{1 + \alpha} + \frac{1}{(1 + \alpha)^2} \right) \\ &= \frac{r_0^2}{2} \frac{1}{(1 + \alpha)^4} (1 + \alpha + \alpha^2) \end{aligned}$$

where

$$p = \frac{h\nu'}{h\nu} = \frac{1}{1 + \alpha(1 - \cos\theta)}$$

α being equal to $h\nu / m_0 c^2$, and $\theta = 90^\circ$

At 40 kev, $\alpha = 40 \text{ kev} / 510 \text{ kev} = .078$

and at 60 kev $\alpha = .12$

Therefore, $\frac{d\sigma}{d\Omega} = \frac{r_0^2}{2} (.80)$ at 40 kev

& $\frac{d\sigma}{d\Omega} = \frac{r_0^2}{2} (.72)$ at 60 kev

ie about 12 % change in cross-section from 40 to 60 kev change in average energy.

The effects due to changes in differential scattering cross-section and transmission factor tend to cancel each other and therefore the beam hardening effects due to

the first two integrals in eq.(2.44) are more or less cancelled out due to the shift in differential scattering cross-section. The only term remaining unaccounted for in eq.(2.44) is then the third integral which is due to exit beam passing through the lead attenuator before arrival at the detector. It is this unaccounted third integral which leads to complications in density measurements with the two detector X-ray Bone Densitometer as described by White (115).

It therefore becomes important to get rid of the lead attenuator as a means of limiting the count rate in the transmission detector looking at the transmitted beam which is of very high intensity compared to scattered beam. In the present Compton X-ray Bone Densitometer, this has been achieved by incorporating a third detector into the system which is a fast plastic detector suitably collimated and capable of high count rates without much saturation effects and without needing lead attenuator in its front end.

CHAPTER 3

DESCRIPTION OF COMPTON X-RAY BONE DENSITOMETER APPARATUS

3.1. Introduction

The apparatus used in this experiment is a modification of the systems developed by Mortimer (76) and White (115). Both workers used two detectors in fixed positions at right angles to one another and an X-ray source which could rotate from position 1 to position 2 through 90° . With the source in position 1, one detector measured the high flux transmitted intensity while the other measured the scattered intensity which was orders of magnitude smaller than the transmitted intensity. In position 2 of the X-ray source, their roles were reversed. To cope up with the high flux transmitted intensity, the detector measuring it was shielded in front with a 1 mm thick lead disc. The use of such a lead disc placed a limitation on accuracy of measurements due to the severe beam hardening effects caused in this high density, high Z material. To overcome this difficulty, a single source-three detector system has been developed, the third detector being a fast plastic scintillator coupled to fast digital counting electronics.

It may be pointed out that the Compton Bone Densitometers based on Clarke's low energy method (15) as developed and utilized by other workers are all based on two detectors (44, 56, 65, 66, 83, 112). The beam hardening problem does not exist in their systems since they make use of monochromatic photon sources and placing an absorber in front of the detector looking at the high flux transmitted beam does not lead to any complications. The necessity for the three detector system has arisen only due to the use of a polychromatic X-ray source for bone density measurement using Compton scattering method. The advantage of availability of very high flux of photons from an X-ray source even at low filament current is offset by the beam hardening effects in the sample under study and further reduced by the absorber placed in front of the detector looking at the transmitted beam. The three detector design partially overcomes this difficulty by eliminating the lead absorber which is always needed for the two detector system using polychromatic X-ray source.

3.2. Description of Experimental Apparatus:

The system designed and assembled for this study can be divided under two main sections: the bone density mechanical apparatus and the detection electronics. The bone density mechanical apparatus includes collimators, sample positioning and all other mechanical components. The detection electronics includes the photon source as well as

all detection and gating circuitry. Each part is described below separately.

3.2.1. X-Ray Source:

The X-ray source used was a Picker mobile unit capable of producing 90 kVp at a maximum current of 15 ma. The X-ray spectrum was observed to be similar to that described by Johns & Cunningham (49), and is shown in Fig.3.1(a). The spectrum was taken using a 1.65 mm thick aluminium filter and a 3" NaI(Tl) detector placed 90 cms from the source. A 3.86 cm thick lead brick with 6 mm dia. hole was placed in front of the detector to prevent saturation due to excessive counting rate. The X-ray source filament current was kept at the lowest setting of about 0.2 mA. The spectrum shown in Fig.3.1(a) shows an X-ray cut-off energy of about 95 kVp. The high energy tail is due to the background counting rate of the detector which has not been subtracted. Fig.3.1(b) shows the energy calibration curve used to identify the X-ray peak intensity location for the spectrum obtained using a hybrid filter composed of (1.65 mm Al + .8 mm Cu + 1 mm Al). In this case the X-ray peak intensity position has shifted to a higher energy region of 51 kev due to the additional filtration introduced.

The head of the X-ray set was mounted on a rotating gantry so that it could be rotated through 90° from position 1 to position 2 together with the three detectors which also

FIG. 3.1(c).

**X-RAY INTENSITY SPECTRUM WITH NaI (TL)
 DETECTOR WITHOUT LEAD ABSORBER.
 (BACKGROUND NOT SUBTRACTED)**

FIG. 3.1 ENERGY CALIBRATION CURVE FOR 3" NaI(Tl) DETECTOR TO
(b) IDENTIFY X-RAY SOURCE PEAK INTENSITY OF TRANSMITTED BEAM

were rigidly mounted to the same gantry, as described in section 3.2.9 and shown in Fig.3.2.

3.2.2. CsI(Na) Detectors, D1 & D2 for scattered beam:

All scattered X-ray intensity information was detected using two CsI(Na) crystals (3/4" x 1") which were optically coupled to RCA 4802 photomultiplier tubes. The PM tube is 0.75 " in diameter and 3.8 " in length. It is a twelve stage tube with S-11 spectral response. The high sensitivity and small size of this tube make it highly suitable for this application.

The CsI(Na) scintillation crystal from Harshaw has been chosen since it is non-hygroscopic and has higher density than NaI(Tl). Its decay constant is 1.0 microsecond and it has a light output of 0.85 relative to NaI(Tl). Fig.3.3(a) gives the schematic for its photomultiplier tube base circuit. The tube is operated under maximum gain conditions making it unnecessary to use the preamplifier circuit.

The energy spectra shown in Figs.3.4(a) & (b) give the energy resolution (FWHM) of the two detectors at two different energies, namely, 59.6 kev and 122 kev. These were obtained using Am-241 and Co-57 radionuclides.

TOP VIEW

SIDE VIEW

COLLIMATED DETECTOR DESIGN

Scale: 1 : 3

**FIG. 3.2. SCHEMATIC SKETCH OF COMPTON
X-RAY BONE DENSITOMETER.**

3.2.3. Plastic Detector, D3, for transmitted beam:

The detector, D3, is required to be mounted directly opposite the X-ray source on the gantry and is used to detect the transmitted beam in both positions of the X-ray source. Consequently the detector was designed using NATON 136 plastic scintillator ($3/4 \times 2$) coupled to an RCA 4802 photomultiplier tube. This particular scintillator was chosen because of its fast decay time of 1.6×10^{-9} sec. which gives it the capability of counting a very high X-ray intensity without pile-up. The detector is found suitable for detecting low energy X-rays which are encountered in the transmitted beam in the present study. To make effective use of this plastic scintillator, the output of this detector was coupled to the fast electronic counting system described later in this chapter. Fig.3.5 shows a typical curve obtained by observing PM tube noise with variation of EHT. Fig.3.3(b) shows the PM base circuitry for the plastic detector.

3.2.4. Digital Electronics for CsI(Na) Detectors Outputs:

The negative pulses from the PM tube anode are fed to an Ortec Multimode Amplifier, Model 410, followed by an Ortec Timing Single Channel Analyzer, Model 420. The latter was then fed to an Ortec Timer-Scaler, Model 431, which is gated by the pulse from the gating logic circuitry. The output of either of the two CsI(Na) detectors, as appropriate, is passed on to the counting electronics

through a two way switch. The block diagram is shown in Fig.3.6(a).

3.2.5. Digital Electronics for Plastic Detector Output:

The negative fast rise time pulses from the output of the Plastic Detector, D3, are fed to an EG&G Quad Amplifier, Model AN201/N, an EG&G Dual Discriminator, Model T105/N,, and then to a Canberra Nuclear Counter, Model 1491, through an EG&G inverting transformer, Model IT100. The scaler is gated by the pulse from the gating logic circuitry. The block diagram is shown in Fig.3.6(b).

Figs. 3.7 (a), (b), & (c) show the Detector-X-ray mounting gantry and the counting electronics to which the outputs of the three detectors are coupled.

3.2.6. Logic Gating Circuit:

The Logic Gating Circuit is essentially that described by White (115). The purpose of the logic circuit is two-fold: to gate all digital data going to the counting electronics and to act as an accurate exposure timer.

Since it is known that the X-ray pulses occur during the negative portion of the transformer voltage, this voltage can be used to monitor the X-ray pulse and thus only collect the desired data. The X-ray Monitor is shown in the block diagram in Fig.3.8. It is the input of a 741

Operational Amplifier (Series SN72, Texas Instruments), whose saturated output provides a square wave which is used for the logic functions.

This output provides the basis of the logic input pulse for both the variable pulse shaping and the various scalers in the logic circuitry. The first scaler is a divide by six scaler, which produces an output pulse for every six input pulses. Since there are sixty pulses a second, this allows an output pulse every tenth of a second, which is a convenient choice for timing. The purpose of the skip scaler is to have the first twelve X-ray pulses not included in either the data collection or the collection time. This decision resulted from studies (109) which showed that the first few X-ray pulses were both different in energy spectrum and inconsistent when compared with the pulses obtained after the X-ray unit settled down from any initial shock. All timing was done by comparing (at the toggle switches) with the actual number being counted by the timer scalers. When these are equal, a pulse is produced from the comparator which resets the flip flops to prevent any other pulse from being included and also shuts off the X-ray unit.

It was found desirable to have control over the shaping of the eight msec X-ray pulses. In this way it is possible to have more uniform pulses which approach a square pulse by eliminating any low energy tails. This was made possible by using 555 timers (Series SN72, Texas

Instruments) as a variable pulse delay and a variable pulse width. In this study, the eight millisecond pulse which had a two millisecond delay, was generated from the back edge of the pulse delay. It was set for a four millisecond width. Thus, all collectable data occurred during this pulse and only this data was counted by the scalers in the counting electronics which were externally gated on during this logic pulse duration. The net result of this gating system was the counting of a four millisecond X-ray pulse every sixteen milliseconds. This also allowed only four msec of background to be subtracted.

As well as control over each individual pulse, there is a fixed delay of a fraction of a second after pressing the start button. This delay is to prevent any noise from the push button from reaching the logic circuit which could result in false counts. Finally, there is about a one second delay before the logic circuit is initiated which allows the resetting of all flip flops and scalers. The single start push button starts both the electronics and the X-ray unit. Also a light bulb was included as an exposure indicator. The detailed circuitry can be found in the work described by White (115).

3.2.7. Electronic Circuit to eliminate noise from Logic Circuit mixing with useful data counting rate:

Fig.3.9 is a schematic of the circuit that was assembled and incorporated in the logic box to overcome the problem of undesirable noise entering from the Relay Circuit in the Logic Box into the data collection electronics. The circuit basically uses an optically-coupled isolator FCD 810 (Fairchild) which is a gallium arsenide infra-red emitting diode and a silicon npn phototransistor in close proximity. Optical intercoupling provides a high degree of ac and dc isolation. The device provides an isolation resistance of as high as 10^{11} ohms between input and its output. The noise introduced in the data channel was significantly reduced on incorporating this circuit in the logic circuitry designed by White. (115).

3.2.8. Different Counting Electronic Systems to maximize

Signal to Noise Ratio for Scattered X-ray beam:

Since Scattered beam intensity is orders of magnitude smaller than transmitted beam intensity and statistical precision of density measurement is mainly governed by scattered counting rate, it was found desirable to consider different counting electronic systems to optimize the signal to noise ratio. Due to the low photon energy of the X-rays, a large part of the signal data also merges in photomultiplier tube noise and therefore it

was necessary to extract the useful information from the PM tube noise.

The extraction of useful data from noise becomes possible owing to the fact that CsI(Na) crystal pulses have a decay time of approximately 1.25 microseconds (106) whereas the single electron photocathode noise of the PM tube is about 5 nanosecond wide. The pulse rise times for data signal and noise are thus widely different and one can set up the logic counting electronics to discriminate one from the other.

Four different methods were studied and these are shown in block diagrams in Fig.3.6(a), Fig.3.10(a), Fig.3.10(b), and Fig.3.10(c). These were compared against the conventional counting electronics system utilizing a preamplifier coupled between the detector output and the main amplifier employing single integration of the pulse output from the detector. The results of this study are shown in Fig.3.10(d). It was decided to resort to the double integration technique as an optimum choice for counting scattered beam since it uses a minimum of instrumentation and gives reasonably good performance towards reduction in PM tube noise.

3.2.9. Bone Density Mechanical Apparatus:

The Bone Density Apparatus was designed to be functional for a variety of conditions, such as a change in the scattering angle or for different bones under examination.

The main mechanical device consists of an annular steel ring on which the X-ray unit and the three detectors are mounted in fixed configuration mutually at right angles to one another. The whole assembly is fixed on a table with 16 cm dia. hole at its centre. Roller bearings were placed below the steel ring so that it could be rotated by 90° from position 1 to position 2.

Four brass collimators were designed and were fixed, one each, in front of the X-ray source and the three detectors. The Plastic detector was fixed in position directly opposite the X-ray source. The brass collimators are all 8 cm long and 3 cm in diameter, with wall thickness of 0.5 cm. Each brass collimator contains two lead plugs (which are interchangeable) of thickness 0.8 cm and aperture diameter 0.5 cm (other apertures of 0.35 cm and 0.65 cm can also be used). The lead plugs of various diameters have been used in experimental studies carried out by White (115) and Mortimer (76). With this design there is a large cavity between lead plugs which limits multiple scattering within the collimator.

Acrylic templates made of 1 mm thick plastic sheet were used for positioning the sensitive volume of the

subject under study at the isocentre of the assembly. The procedure is shown in Fig.7.1(e) of Chapter 7.

FIG. 3.3(a). RCA 4802 - 0.5mA PM BASE
CIRCUIT FOR CsI(Na) DETECTORS
D1 AND D2.. MAX. EHT: -2000 V.

FIG. 3.4. ENERGY SPECTRA OF CsI(Na)
DETECTORS USING Am-241 & Co-57.
(a) DET D1: 70%(59.6 KEV), 46%(122 KEV)
(b) DET. D2 67%(59.6 KEV), 45%(122 KEV)
RESOLUTION (FWHM).

FIG. 3.5 EHT Vs PMT (RCA 4002, D3) NOISE

Fig. 3.6 ELECTRONICS TO MAXIMIZE S/N RATIO OF OUTPUT SIGNAL FROM
(a) CsI(Na) DETECTORS SEEING SCATTERED X-RAY BEAM. (METHOD V)

(b)

Fig 3.6(b) ELECTRONICS FOR DETECTING OUTPUT SIGNAL FROM PLASTIC DETECTOR SEEING TRANSMITTED X-RAY BEAM

(a)

(b)

(c)

FIG. 3.7. X-RAY BONE DENSITOMETER SYSTEM.

(a) X-RAY SOURCE — DETECTOR MOUNTING GANT.RY.

(b) LOGIC CIRCUIT BOX TO CONTROL X-RAY EXPOSURE & TIMING.

(c) DIGITAL COUNTING ELECTRONICS.

(MCA, PULSE GEN, OSC. ONLY FOR SPECTRA STUDY)

FIG. 3.9. SECTION OF LOGIC BOX CIRCUIT SHOWING INCORPORATION OF OPTICALLY-COUPLED ISOLATOR FCD 810 CIRCUIT AND ISOLATION TRANSFORMER HAMMOND 166J6.

(a)

FIG. 3.10(a). ELECTRONICS TO MAXIMIZE S/N RATIO OF OUTPUT SIGNAL FROM CsI(Na) DETECTORS SEEING SCATTERED X-RAY BEAM (METHOD I)

(b)

FIG. 3, 10(b). ELECTRONICS TO MAXIMIZE S/N RATIO OF OUTPUT SIGNAL FROM CsI(Na) DETECTORS SEEING SCATTERED X-RAY BEAM (METHOD II)

FIG. ELECTRONICS TO MAXIMIZE S/N RATIO OF OUTPUT SIGNAL
3.10(C). FROM CsI(Na) DETECTORS SEEING SCATTERED X-RAY BEAM
(METHOD III)

FIG. SIGNAL TO NOISE RATIO OF OUTPUT SIGNAL FROM CSI 3.10(4) (Nd) DETECTORS, SEEING SCATTERED X-RAY BEAM, TO OBTAIN 2% PRECISION IN SCATTER COUNTING RATE, COMPARED FOR FIVE DIFFERENT METHODS OF ELECTRONICS SET-UPS

CHAPTER 4

MULTIPLE SCATTERING - PRELIMINARY

THEORETICAL STUDIES

4.1. Introduction:

The density as measured using eq.(2.26) depends upon the number of photons which have undergone single scattering interactions within the sensitive volume of the sample. Thus, any detectable multiple scattered photon would result in a falsely high scattered count rate. The origin of this multiple scattered photon may be outside the region of interest which is the sensitive volume defined by the intersection of incident photon beam and the field of view of the collimated detector looking at the scattered beam.

Qualitatively, one can visualize that as the sample size is increased, the amount of multiple scattering will increase, since the probability of scattering is dependent on the path length the photon travels. Thus, for different sized objects, corrections for multiple scattered photons may be necessary.

For samples with large dimensions compared to the dimensions of the detector collimator, the physics of the system may discriminate against triple and higher order scattering since the path length traversed by photons

increases with the order of scattering and consequently attenuation takes its toll (33). The above statement may also approximately hold for samples whose dimensions are small or comparable to that of the collimator diameter since the probability of escape of the photon from the sensitive volume will increase with the order of scattering. We shall therefore consider only double scattering which is the simplest case of multiple scattering.

Mathematical evaluation of multiple scattering effect is a complex, highly geometry dependent effect. The contribution due to this effect can, of course, be obtained through Monte Carlo simulation in which the wide range of conditions and boundary values can be readily included. Kennett and Weber (59), however, have pointed out that the multi-dimensional nature of the problem, in relation to anatomical and density variations, virtually precludes the use of the Monte Carlo technique. These workers, therefore, devised a simplified preliminary approach to the problem of multiple scattering and made use of it for corrections to density measurements with their Compton Bone Densitometer system. The same approach has also been adopted by White (115) in his preliminary investigations on non-biological samples studied in-vitro with X-ray Compton Bone Densitometer. Ebifegha (26) has adopted a mathematical approach to the problem of double scattering and has found some success in applying corrections to in-vitro density measurements to a limited extent.

4.2. Expression for Single Scattering:

It may be recalled from Chapter 2 that the incident and scattered photon energies ($h\nu$ and $h\nu'$, respectively) in a Compton interaction are related to the angle of scattering θ (where θ is chosen such that $\theta = \pi$ refers to backscatter) by the following relation:

$$P = h\nu / h\nu' = \frac{1}{1 + \alpha (1 - \cos\theta)} \quad (4.1)$$

where $\alpha = h\nu / m_0 c^2$.

The probability of photon energy being scattered through any given angle θ is described by the differential energy scattering cross-section (per electron) per unit solid angle about θ by the Klein-Nishina formula quoted in Chapter 2. This may be written in the form:

$$\frac{d\sigma_s}{d\Omega} = \frac{1}{2} r_0^2 P^2 (1 - P \sin^2\theta + P^2) \quad \text{cm}^2/\text{electron/steradian} \quad (4.2)$$

where P is as defined in eq. (4.1) and $r_0 = e^2 / m_0 c^2$ is the classical electron radius.

Physically this means that a single electron located in a photon beam of unit intensity will scatter an intensity of

$$[d\sigma_s / d\Omega]$$

into a unit solid angle about θ (33). In effect the scattering electron will look like a point source of photons with emitted flux varying with scattering angle θ .

If the incident mono-directional flux is I_0 , the flux scattered by a single electron into unit solid angle about θ is obviously

$$I = I_0 \frac{d\sigma_s}{d\Omega} \quad (4.3)$$

Hence at a distance R from the scattering electron, neglecting attenuation through this distance, the scattered flux will be

$$I = \frac{I_0}{R^2} \frac{d\sigma_s}{d\Omega} \quad (4.4)$$

By substituting eq.(4.2) for $\frac{d\sigma_s}{d\Omega}$, one can therefore write

$$\frac{I}{I_0} = \frac{1}{2} r_0^2 \frac{p^2}{R^2} [1 - P \sin^2 \theta + p^2] \quad (4.5)$$

Denoting $\frac{1}{2} p^2 [1 - P \sin^2 \theta + p^2]$

by $F_1(E_0, \theta)$, where E_0 equals the incident photon energy, $h\nu$, we can rewrite eq.(4.5) as

$$\frac{I}{I_0} = \frac{r_0^2}{R^2} F_1(E_0, \theta) \quad (4.6)$$

Utilizing the above expression, we can immediately write down the energy flux reaching the CsI(Na) scintillation detector, D1 or D2, attributable to radiation scattered within volume dV_1 in sensitive volume, through angle θ (which is equal to 90° in the present densitometer set up):

$$I_s = I_0 r_0^2 F_1(E_0, \theta) \left(\frac{N_0 Z}{A} \rho \right) \Delta V_1 \frac{e^{-\mu(E_1) r_1}}{r_1^2} \quad (4.7)$$

where $(N_0 Z/A)$ is the number of electrons per cubic cm of the material in the scattering sensitive volume and r_1 is the dist. between the detector and sensitive volume. The exponential factor takes into account the attenuation of the scattered beam along its path to the detector, $\mu(E_1)$ being linear attenuation coefficient for scattered photon of reduced energy $E_1 (= h\nu')$.

The total photon flux reaching the detector by a single scattering is then obtained by integrating over all volume elements within the sensitive volume from which single scattering can occur.

4.3. Mathematical Expression for Double Scattering:

We wish to consider the radiation from the X-ray source located at Q which is first scattered by the electrons in the infinitesimal volume dV_1 , within the cylindrical zone defined by intersection of the incident collimated X-ray beam with the cylindrical sample placed in the sensitive volume region, to a second volume element dV_2 within the conical region defined by intersection of cone formed with collimated point detector, D1 (or D2) located at O, and the cylindrical sample, and then subsequently scattered to the detector located at O as shown in Fig. 4.1. The geometry shown corresponds to the experimental set up of the Compton Bone Densitometer used in the present study for density measurement.

FIG. 4.1. SCHEMATIC REPRESENTATION OF DOUBLE SCATTERING PROBLEM USING SPHERICAL COORDINATES WITH ORIGIN AT POINT DETECTOR.

Referring to Fig.4.1, obviously the expression for doubly scattered photon flux is of the following form:

$$I_{ss} = \int_{V_1} \int_{V_2} I_0 \left(\frac{e^2}{m_0 c^2} \right)^2 \frac{F_1(E_0, \theta_1)}{e^{-\mu(E_1) |\vec{r}_1 - \vec{r}_2|}} \left(\frac{N_0 Z \rho}{A} \right) \Delta V_1 e^{-\mu(E_1) \left(\frac{t}{2} - r_1 \sin \theta_1 \right)} \\ \times \frac{F_1(E_1, \theta_2)}{(|\vec{r}_1 - \vec{r}_2 |)^2} \left(\frac{e^2}{m_0 c^2} \right)^2 \frac{F_1(E_2, \theta_2)}{e^{-\mu(E_2) (|\vec{r}_2 - a \sec \theta_2 |)}} \left(\frac{N_0 Z \rho}{A} \right) \Delta V_2 \frac{1}{r_2^2} \quad (4.8)$$

where we have neglected the curvature of the surface of the sample, and have assumed that a point at the centre of the detector behaves in the same way as an appropriate average of all points on the detector located at at the origin.

For the energies of interest (30 - 90 kev for the X-ray source used in this study), we make the simplifying assumptions that scattering is isotropic and that the energy change on scattering does not result in a significant change in attenuation. With these assumptions we may replace $\mu(E_1)$ and $\mu(E_2)$ simply by the symbol μ . The expression for double scattering, then, takes the following form:

$$I_{ss} = \int_{V_1} \int_{V_2} I_0 F_1 F_2 \left(\frac{e^2}{m_0 c^2} \right)^4 \left(\frac{N_0 Z \rho}{A} \right)^2 \Delta V_1 e^{-\mu \left(\frac{t}{2} - r_1 \sin \theta_1 \right)} \\ \times \frac{e^{-\mu |\vec{r}_1 - \vec{r}_2|}}{(|\vec{r}_1 - \vec{r}_2 |)^2} \frac{e^{-\mu (|\vec{r}_2 - a \sec \theta_2 |)}}{r_2^2} \Delta V_2 \quad \dots (4.9)$$

2

where F_1 and F_2 are for simplicity taken nearly equal for the range of energies E and E and they refer to scattering at 90° from the incident direction of photon beam.

In the figure shown, we have chosen a right hand spherical co-ordinate system with the origin at the location of the detector and the x-axis is shown out of the plane of the figure.

The symbols in eq. (4.9) have the following meaning:

- I_0 : intensity of the incident photon beam
- N_0 : Avogadro's number
- Z & A : effective atomic number and mass number of the sample
- $|\vec{r}_1 - \vec{r}_2|$: distance traversed by photon on scattering from the first volume element dV_1 to the second volume element dV_2
- a : distance between point detector D_1 (or D_2) located at the origin of co-ordinates to the nearest edge of the cylindrical sample of diameter, t .
- F_1 & F_2 : normalized differential scattering cross-sections at 90° scattering angle and energy E_0 (or E_1 which are nearly the same for low energy X-ray photons).
- μ : linear attenuation coefficient of the sample material through which the photons traverse, taken as roughly the same for incident and scattered beams.

The limits of integration are dependent on the size of the sample and they are given by:

- θ_1 : 0 to β where β is the polar angle subtended by the line joining the edge of the sample nearest to X-ray source with the origin.

θ_2 : 0 to α , α being the half angle of the circular cone defined by the detector collimator geometry.

ϕ_1 : $(\frac{3\pi}{2} \pm \delta)$. and $(\frac{\pi}{2} \pm \delta)$, where δ is the small angular range defined by the width of the incident photon beam.

ϕ_2 : 0 to 2π

r_1 : $(a + \frac{t}{2} - \epsilon) \sec\theta_1 \operatorname{Cosec}\phi_1$ to $(a + \frac{t}{2} + \epsilon) \sec\theta_1 \operatorname{Cosec}\phi_1$,

ϵ being the half width of the incident photon beam.

r_2 : $a \sec\theta_2$ to $(a + t) \sec\theta_2$.

The six dimensional integral in eq.(4.9) becomes difficult to evaluate analytically due to the fact that the greatest contribution to double scattering is anticipated to be from the volume elements in V_1 (cylindrical region) and V_2 (conical region) which are very close to the sensitive volume under study (ie the region at the intersection of cylindrical and conical volumes) and for such volume elements r_1 becomes nearly equal to r_2 . This leads to the complexity since singularities will appear where $r_1 = r_2$. Moreover, the sophistication of the geometry and the limits of integration involved in the problem make the analytical solution prohibitive. The only approach, therefore, left at one's disposal is to use the numerical integration techniques, such as sixteen point gaussian quadrature (1, 62).

The numerical integration was, however, not carried out, since Ebifegha (26) has tackled the problem using cartesian co-ordinates and a slightly different approach.

He has shown that a reduction of about 50 % in the observed discrepancy between the compton density and the Archimedes density can be achieved. He has also concluded that the discrepancy between the compton density and the Archimedes density will be substantially reduced by using samples and calibration standards of equal dimension. It may be pointed out that all experimental studies on in-vitro and in-vivo compton mass density measurements described in Chapters 6 & 7 have been carried out using the lucite calibration standard of nearly the same shape and diameter as the specimen under investigation.

It is worthwhile to note that the correction scheme as outlined by Ebifegha (26) is useful while dealing with high density materials such as aluminium and teflon, with densities more than double that of lucite standard, where the compton density measurements could be 10 % or more higher than the true values due to multiple scattering contribution. The discrepancy between compton mass density and Archimedes density is reduced if the density and linear attenuation contrast between the sample under investigation and the calibration standard employed are nearly the same (26). The range of compton mass densities measured in the present study lies between about 0.8 to 1.8 g/cm³ and in this range the discrepancy between compton mass density and Archimedes density is less than 5 % as will be evident from the experimental studies carried out in Chapter 6. Since

the precision attainable with the present system is about 5%, it may not be necessary to apply multiple scattering corrections to the measurements made with the present system. All that is essential is to choose the lucite calibration standard of nearly the same dimensions as the specimen under investigation.

CHAPTER 5

STUDY OF TRANSMITTED AND SCATTERED X-RAY BEAM SPECTRA IN VARIOUS CONFIGURATIONS

5.1. Introduction:

The output of the X-ray tube used in the Compton X-Ray Bone Densitometer is distributed across a range of energies up to a maximum of 90 kVp. Different parts of the X-ray spectrum are attenuated by different amounts when passed through the same material. The low energy (softer) X-rays are preferentially absorbed and the peak of the spectral distribution shifts towards the higher energies or hardens. As a result, the value of the attenuation within a particular sensitive volume region will depend on the length of the path traversed by the X-ray beam through the material surrounding the sensitive volume.

A series of experiments was performed to study the spectral shapes of scattered and transmitted X-ray beams after the incident beam has traversed through the material under investigation. Different filter combinations have been used to alter the spectral shape of the emerging beam from the X-ray source and the effects studied on their passage through the material. Some experimental investigations have also been done on multiple scattering effects.

The Compton Bone Densitometer as developed by White (115) made use of two detectors and a 1 mm thick lead absorber was placed in front of the detector seeing the high flux transmitted beam. In the second position of the X-ray source, the role of the two detectors was reversed and the lead absorber was placed in front of the second detector now looking at the transmitted beam. Additional beam hardening caused in the lead absorber was undesirable and was likely to give erroneous results in density measurements. The nature of this effect was also studied to some extent from the spectra obtained from the present system by simulating the two detector system conditions for density measurement.

The series of spectra recorded in the present study elucidates many other effects such as PM-tube saturation and pulse pile up in the NaI(Tl) detector. This was placed facing the incident beam at 90 cm. distance replacing the present detector D3 (plastic detector) to simulate White's two detector system. This study confirmed the necessity of replacing CsI(Na) detector with a fast plastic scintillation detector and with PM tube circuitry for the higher counting rate capability required for recording the transmitted beam.

5.2. Recording of Spectra:

The series of spectra shown in this chapter were photographed and the data printed on the Teletype by feeding the detector output to a Kicksort 4096 Channel Pulse Height Analyzer Model 711, the output of which was coupled to a

Serial Readout Teletype Model TT/33. A Polaroid camera Model CR-9 was attached to the analyzer display scope screen for taking pictures after storing the spectra in the memory of the analyzer. An Ortec preamplifier Model 113 was used when taking the output from detector D1 looking at the X-ray beam scattered at 90 degrees from incident direction on the sample under study.

5.3. Beam Hardening Effects in Transmission Beam:

The beam hardening effects within the samples for the transmitted beam become evident from the spectra shown in Fig.5.1. These spectra were obtained using a 3" NaI(Tl) detector placed directly opposite the X-ray source at 90 cm distance. A 3.86 cm lead brick with centred 6 mm dia. hole was kept in front of the detector to prevent saturation as far as possible.

As is evident from Fig.5.1 and Fig. 5.2, there is considerable shift in the spectral peak region for aluminium and carbon-tetrachloride samples, in particular, when compared with the 6.35 cm dia. lucite standard. For Al, the Z value (=13) is about twice that for lucite ($Z=6.47$) whereas for carbon-tetrachloride ($Z=16.67$) it is about two and half times. The gravimetric densities of these materials are:

Al : 2.70 g/cm^3 , CCl_4 : 1.59 g/cm^3 , Lucite : 1.185 g/cm^3 .

The large spectral shifts observed for these cases are caused by high Z and ρ values since photoelectric absorption

is strongly dependent on the fourth power of Z (49), and the photon attenuation follows the exponential absorption behaviour dependent on mass density and total mass attenuation coefficient; namely

$$I = I_0 e^{-(\mu/\rho)(\rho x)}$$

For light materials or others with Z and ρ values not too different compared with lucite standard, the spectral shifts are not very significant.

The beam hardening situation is therefore likely to lead to an inaccurate determination in density measurement of samples such as Aluminium and Carbon-tetrachloride. That such is the case will be evident from the density measurements shown in the next chapter. The approach to the problem here is merely qualitative.

Fig.5.4 and Figs. 5.1(G) & (H) demonstrate the distortion caused in the transmission spectrum by using 1 mm thick lead disc absorber in front of the NaI(Tl) detector. There is a severe distortion introduced in the spectra due to additional beam hardening taking place in the high density, high Z lead absorber. The beam hardening in Pb is so severe that it is likely to affect the density measurement results. It was owing to this distortion, that the two detector system had to be abandoned where it was essential to make use of Pb absorber for detecting high flux transmission beam. This led to the present design of the three detector system where the transmission beam is always

seen by the same fast plastic detector capable of handling high photon flux without pile-up or saturation effects.

The beam hardening effects seen in the scattered beam are shown in Fig.5.7. The spectra shown were taken with CsI(Na) detector, D1, mounted on the present Bone Densitometer at right angles to the X-ray source beam direction. The beam hardening effect is much less pronounced here due to the degradation in incident beam energies because of Compton scattering which masks the photoelectric absorption effects. Fig.5.6 shows a series of spectra taken for scattered beam with D1 for different X-ray filament current settings with 6.35 cm dia. lucite cylinder placed in the sensitive volume position..

5.4. X-Ray Beam Filtration:

The spectra shown in Fig.5.3 depict the effect of filtering the incident X-ray beam with different filters.

The filters used were:

- (a) 1.65 mm thick Aluminium prefilter,
- (b) 1.65 mm Al prefilter + .8 mm Cu + 1 mm Al,
- and (c) 1.65 mm Al + .5 mm Sn + .8 mm Cu + 1 mm Al.

As is evident the X-ray beam becomes harder as more and more filtration is introduced at the X-ray Source output. Owing to the radiation dose considerations, it is advantageous to cut off the lower energy X-ray flux as much as possible and for this reason it would be better to make use of (Al + Sn + Cu + Al) combination filter. In this

situation, however, there was considerable reduction in scattered beam count rate and a comparatively large time is required for each exposure to achieve the desired statistical precision. It was therefore decided to use the hybrid filter, namely (1.65 mm Al prefilter + .8 mm Cu + 1 mm Al), as a compromise between radiation dose and statistical precision.

5.5. Some Other Effects:

Fig. 5.3(A) is just to show the saturation and pile-up effect that has occurred in the NaI(Tl) detector due to very high X-ray flux reaching the detector in the transmission beam.

Fig. 5.4 shows spectra observed with the 3" NaI(Tl) detector with Lead disc absorber in front of the detector and various samples placed in the sensitive volume region. Spectra obtained without lead absorber disc are also shown in some of the figures for comparison.

Fig. 5.8 shows the effect of filtration on X-ray beam by taking spectra of the scattered beam from 6.35 cm dia. lucite in sensitive volume region. The detector, D1, mounted at right angles to incident beam direction was used to take these spectra.

5.6. Multiple Scattering Effects:

The spectra of scattered beam shown in Fig. 5.9 were taken using 6.35 cm/ 5.7 cm dia. lucite cylinders with

centred holes of different diameters. CsI(Na) detector, D1, at right angles position was used to record the scattered intensity. It will be seen from these spectra that there is a significant contribution due to multiple scattering effects occurring between 0.5 cm and 1.5 cm radial region round the centre of the cylinders. Beyond 1.5 cm, the contribution is negligible. The problem of multiple scattering effects leading to erroneous density measurement results for high Z, high density materials, has been studied by Ebifegha (26). This problem was not studied in any greater detail in the present study since we chose the lucite standard diameter as close to the size of the sample under study as feasible. Thus the contribution from multiple scattering will be of the same order both for the sample under study and the lucite standard. The density effect on multiple scattering, however, could not be taken into account by this method of comparing with the lucite standard having the same diameter as the sample under investigation.

(A) Filament current: .2 mA.
a. 6.35 cm dia lucite,
b. CCl_4 in lucite, c. 5.7cm Al.

(B) Filament current: 0.2mA.
a. 6.35 cm lucite, b. Teflon in
lucite, c. H_2SO_4 in lucite.

(C) Filament current: .2 mA.
a. 6.35 cm lucite, b. 2.5 cm Al
in lucite, c. 5.7 cm dia Al.

(D) Filament current: .2mA.
a. 6.35 cm dia lucite,
b. Methanol in lucite.

(E) Filament current: .2mA.
a. 6.35cm lucite, b. HCl in lucite.

(F) Filament current: .2mA.
a. 6.35 cm lucite, b. 5.0 cm
lucite, c. 2.0 cm lucite.
(H)a. 6.35 cm lucite, .2mA, 6mm hole,
b. 6.35 cm lucite, 2mA, 1mm Pb,
c. 2.5 cm lucite, 2mA, 1mm Pb
covering 6mm dia hole in brick.

(G)a. 6.35 cm lucite, .2mA, 6mm hole in
Pb brick, b. 2.0cm lucite, 2mA, 1mm
Pb covering 6mm hole, c. 6.35 cm
lucite, 2mA, 1mm Pb covering hole.

Fig.5.1 - 3" NaI(Tl) detector, 90cm from X-Ray source, 3.86 cm thick
lead brick with 6mm hole in its front, looking at
transmission beam for various samples. In (G) & (H), hole
covered with 1mm thick Pb disc. Filtration used:
(1.65mm Al + .5mm Sn + .8mm Cu + 1mm Al).

(A) a. Methanol in lucite,
b. $\text{CaH}_4(\text{PO}_4)_2 \cdot \text{H}_2\text{O}$ in lucite

(B) a. Benzene in lucite,
b. 5.7 cm dia Al.

(C) a. 6.35 cm lucite, b. 1.5 cm Al in lucite,
c. CCl_4 in lucite, d. 2.5 cm Al in lucite.

(D) a. 6.35 cm lucite,
b. CCl_4 in lucite.

(E) Filament current: 2 mA
a. 6.35 cm lucite, b. CCl_4 in lucite.

(F) a. 2.0 cm lucite, b. 6.35 cm lucite,
c. 8.6 cm lucite.
Filament current: 2 mA

(G) a. 6.35 cm dia lucite,
b. 2.5 cm dia Al in lucite,
c. 5.7 cm dia Al.
Filament current: 2 mA.

Fig. 5:2-3" NaI(Tl) detector, 90 cm from Source, 3.86 cm thick Pb Brick with 6 mm hole in its front, looking at transmission beam for various samples in position. Filament current set at 0.2 mA unless otherwise stated.

Ch 16-64
1.54 kev/ch
a. 35825(35)
b. 31475(32)

Ch 16-64
1.54 kev/ch
a. 31913(36)
b. 27244(40)

(A) No sample in sensitive volume.
a. Filament current: 1 mA,
b. Filament current: 2 mA.

(B) No sample in sensitive volume,
a. Filament current: 0.5 mA,
b. Filament current: 0.2 mA.

Ch 16-64
1.54 kev/ch
a. 31355(41)
b. 33410(32)
c. 32967(24)

Ch 16-72
1.54 kev/ch
a. 31152(40)
b. 12930(48)
c. 12767(50)
d. 9973(51)

(C) 6.35 cm Lucite in position.
Filter: a. (Al+Sn+Cu+Al),
b. (Al+Cu+Al), c. 1.65mm
Al prefilter.

(D) 6.35 cm Lucite in position.
a. 2mA, (Al+Sn+Cu+Al) filter, b. 2mA,
(1.65mm Al) prefilter, 1mm thick Pb disc
covering 6mm hole in Pb brick, c. 2mA,
(Al+Cu+Al), 1mm absorber(Pb), d. 2mA,
(Al+Sn+Cu+Al) filter,
1mm Pb absorber.

(E) 0.2mA, 6.35cm
Lucite.
a. 1.65mm Al,
b. (Al+Cu+Al),
c. (Al+Sn+Cu
+Al) filter.

Ch 16-72
a. 15510(37)
b. 7506(45)
c. 3566(47)
1.1 kev/ch

(F) 2mA, 6.35 cm Lucite,
a. 1.65mm Al filter,
b. (Al+Cu+Al) filter,
c. (Al+Sn+Cu+Al) filter.

(G) 1.65 mm Al filter.
Filament current:
a. 2mA, b. 1mA,
c. 0.5 mA,
d. 0.2 mA.

Ch 16-96
1.1 kev/ch
a. 37977(35)
b. 23131(42)
c. 13577(46)

Ch 8-72
1.1 kev/ch
a. 41796(34)
b. 34396(33)
c. 22727(34)
d. 14341(34)

Fig. 5.3- 3" NaI(Tl) detector, 90 cm from X-Ray source, seeing
transmission beam under various configurations. 3.86 cm
thick lead brick with 6mm hole in its front.

Ch 8-72
1.1 kev/ch
a. 31824(37)
b. 20448(38)
c. 13948(41)

- (A) No Pb absorber, Filament: 1mA,
Filter: 1.65 mm Al.
a. 6.35 cm lucite, b. H_2SO_4 in
lucite, c. CCl_4 in lucite.

Ch 16-64
1.54 kev/ch
a. 33842(40)
b. 9567(50)
c. 9907(54)
d. 7632(52)

- (C) Filter: (Al+Sn+Cu+Al).
a. 6.35 cm lucite, .2mA, Pb brick
with 6mm hole, b. 2cm lucite, 2mA,
1mm Pb disc covering hole, c. 6.35 cm
lucite, 2mA, 1mm Pb disc, d. 8.6 cm
lucite, 2mA, 1mm Pb disc.

a. 30248(44) Ch 16-64
b. 11132(55) 1.54 kev/ch

- (E) 6.35 cm lucite in position.
Filter: (Al+Sn+Cu+Al), .2mA.
a. Pb brick hole covered with 1mm Pb,
b. Pb brick hole covered with 1.4mm
thick Pb disc; 20.4 sec exposure.

Ch 16-64
1.54 kev/ch
a. 29784(40)
b. 9017(51)

- (G) 6.35cm lucite, (Al/Sn/Cu/Al)
a. .2mA, Pb brick; b. 2mA, 1mm Pb.

Ch 8-80
1.1 kev/ch
a. 5451(35)
b. 4103(37)

- (B) 1mm thick Pb absorber covering
6mm dia hole in lead brick, 1mA.
a. 6.35 cm lucite, b. CCl_4 in lucite.

Ch 8-80
1.1 kev/ch
a. 5451(32)
b. 4692(32)

- (D) 1mm Pb disc covering 6mm dia hole
in 3.86 cm thick Pb brick,
Filter: 1.65mm Al. a. 6.35 cm
lucite, b. H_2SO_4 in lucite.

a. 29784(41) Ch 16-64
b. 13700(52) 1.54 kev/ch
c. 9017(51)
d. 7071(50)

- (F) Filter: (Al+Sn+Cu+Al).
a. 6.35cm lucite, Pb brick with hole
.2mA; b. 2.0cm lucite, hole covered
with 1mm Pb; 2mA; c. 6.35cm lucite
Pb brick hole covered with 1mm Pb,
2mA; d. 8.6cm lucite, 1mm Pb, 2mA.

Ch 16-64
1.54 kev/ch
a. 30248(44)
b. 11065(51)
c. 4368(30)

- (H) .2mA, (Al/Sn/Cu/Al), 6.35cm lucite.
a. Pb brick with hole; b. 1mm Pb,
20.4sec exp.; c. 2mm Pb disc, 20.4 sec.

Fig5.4 - 3" NaI(Tl) detector, 90cm from source, Pb brick with 6mm hole
in front, seeing transmission beam to study Pb absorber effect.

(A) Filament current: 2 mA
a. 6.35 cm lucite, b. H_2SO_4 in lucite, c. CCl_4 in lucite

(B) Filament current 1 mA
a. 6.35 cm lucite, b. H_2SO_4 in lucite.

(C) Filament current: 1 mA
a. 6.35 cm lucite, b. Teflon in lucite, c. Al in lucite.

(D) Filament current: 1 mA
a. 6.35 cm lucite, b. CCl_4 in lucite.

(E) Filament current: 1 mA
a. 6.35 cm lucite, b. Acetone in lucite.

(F) Filament current: 1 mA
a. 6.35 cm lucite, b. Glycerine in lucite.

Fig. 5.5- CsI(Na) detector, D1, seeing scattered beam for various samples placed in sensitive volume to study beam hardening effects within the sample. Detector EHT: -1950V. X-Ray beam filtered with 1.65 mm Al in (B), (C), (D) & (E); and filtered with (1.65mm Al + .8 mm Cu + 1 mm Al) in (A).

Ch 16-64
1.2 kev/ch
a. 8178(40)
b. 2891(43)

Ch 16-64
1.2 kev/ch
a. 14262(41)
b. 2891(42)

(A) Exposure: 20.4 sec
a. Filament current: 1 mA,
b. Filament current: 0.2 mA

(B) Exposure: 20.4 sec
a. Filament current: 2 mA,
b. Filament current: 0.2 mA

Ch 16-64
1.2 kev/ch
a. 4865(40)
b. 2891(40)

(C) Exposure: 20.4 sec
a. Filament current: 0.5 mA,
b. Filament current: 0.2 mA.

Ch 16-64
1.2 kev/ch
a. 14262(43)
b. 8178(40)
c. 2891(40)

Ch 12-72
0.9 kev/ch
a. 14807(44)
b. 7989(46)
c. 4883(48)
d. 2725(48)

(D) Exposure: 25.5 sec.
a. Filament current: 2 mA,
b. Fil. Curr. 1 mA, c. Fil. Curr.
.2 mA.

(E) a. Fil. Curr. 2 mA, b. Fil. Curr.
1 mA, c. Fil. Curr. 0.5 mA,
d. Fil. Curr. 0.2 mA.

Fig.5.6 -CsI(Na) detector; D1, looking at scattered beam under various configurations. X-Ray beam filtered with (Al+Sn+Cu+Al) combination in (A), (B), (C) & (D). 1.65 mm Al prefilter used in (E). 6.35 cm dia lucite placed in sensitive volume region.

(A) a. 2 cm dia lucite,
b. 5.7 cm Al, 15.3 sec exposure.

(B) a. 2.0 cm lucite, b. 6.35 cm lucite,
c. 8.6 cm lucite, d. 5.7 cm Al,
15.3 sec exposure

(C) a. Teflon in Scintillation vial,
b. HCl in scintillation vial.

(D) a. 1.5 cm dia Al,
b. 2.5 cm dia Al.

(E) a. Glycerine in scintillation vial,
b. Water in scintillation vial,
c. Sucrose in scin vial,
d. Acetone in scin. vial.

Fig.5.7 - CsI(Na) detector, D1, seeing scattered beam from various samples placed in sensitive volume region. X-Ray filament current: 2 mA. 1.65 mm Al filter used. 5.1 sec exposure unless stated.

(A) a. 1.65 mm Al prefilter.
b. (1.65 mm Al + .8 mm Cu + 1 mm Al) filter, c. (1.65 mm Al + .5 mm Sn + .8 mm Cu + 1 mm Al) filter.

(B) 20.4 sec exposure.
a. (1.65 mm Al + .8 mm Cu + 1 mm Al) filter, b. (1.65 mm Al + .5 mm Sn + .8 mm Cu + 1 mm Al) filter.

(C) a. 1.65 mm Al prefilter.
b. (1.65 mm Al + .8 mm Cu + 1 mm Al) filter.

Fig. 5.8 - CsI(Na) detector, D1, looking at scattered beam under different X-Ray beam filtrations to study effect of using filtered incident X-Ray beam. 6.35 cm dia lucite placed in sensitive volume position. X-Ray filament current set at 2 mA.

Ch 18-80
1.2 kev/ch
a. 16099(40)
b. 13803(41)
c. 909(39)
d. 384(37)

Ch 16-80
1.2 kev/ch
a. 14872(39)
b. 2540(41)

- (A) a. 5.7 cm dia lucite,
b. 5.7 cm lucite with 0.5 cm hole,
c. 5.7 cm lucite with 3 cm hole,
d. 5.7 cm lucite with 4.5 cm hole.

- (B) a. 6.35 cm dia lucite,
b. 6.35 cm lucite with 1.5 cm
dia hole.

Ch 16-80
1.2 kev/ch
a. 16689(40)
b. 8027(39)
c. 2879(40)
d. 1214(39)

- (C) a. 5.7 cm dia lucite, b. 5.7 cm lucite
with 1 cm hole, c. 5.7 cm lucite
with 1.5 cm hole, d. 5.7 cm lucite
with 2.5 cm hole.

Fig. 5.9- CsI(Na) detector, D1, seeing scattered beam from lucite cylinder having holes of varying diameters in centre, to study multiple scattering effects from medium surrounding sensitive volume region. Source filament current set at 2 mA; 20.4 sec exposure each.

Ch 1-64
a. 10350
b. 6345
c. 5018
d. 1688

(A) Filament current: 0.2 mA
a. 6.35 cm lucite, b. H_2SO_4 in lucite, c. CCl_4 in lucite, d. 5.7 cm dia Al.

Ch 1-64
a. 15550
b. 10350
c. 9698
d. 4445

(B) Filament current: 0.2 mA
a. Acetone in lucite, b. 6.35 cm lucite, c. Teflon in lucite, d. 2.5 cm Al in lucite.

Ch 1-64
a. 27280
b. 21531
c. 17202
d. 4766

(C) a. 6.35 cm lucite, b. H_2SO_4 in lucite, c. CCl_4 in lucite, d. 5.7 cm Al
Filament current: 2 mA

Ch 1-55
a. 9126
b. 13032
c. 16272
d. 24133

(D) 6.35 cm lucite in position.
Filament current:
a. 0.2 mA, b. 0.5 mA, c. 1 mA, d. 2 mA.

2 μ sec/cm
1 volt/div

(E) Transmission beam output as seen on Tektronix 454 Oscilloscope screen.

Fig. 5.10-Plastic detector, D3, output to Multichannel Analyzer/ Oscilloscope for transmission X-Ray beam via Ortec Preamp., and 410 Multimode Amplifier. Detector EHT: -1400 V. Different samples placed in sensitive volume. Filtration used: (0.8 mm Cu + 1 mm Al + 1.65 mm Al prefilter).

CHAPTER 5

DENSITY MEASUREMENTS ON NON-BIOLOGICAL MATERIALS

6.1. Introduction:

To assess the applicability of the Compton X-ray Bone Densitometer for in-vivo bone density measurements on human subjects, a number of experiments were performed on non-biological materials in various configurations to ensure that the system could be effectively utilized for in-vivo studies with minimal radiation exposure dose and reasonable statistical precision. To achieve this ultimate objective, a series of experiments were performed using various filter combinations to attenuate the lower energy X-ray photons since these are preferentially absorbed photoelectrically in the specimen; hence the Compton scatter contribution is small. This leads to undesired radiation hazard without improving statistical precision. The different filters used were: (i) Al, (ii) (Al+Cu+Al) and (iii) (Al+Sn+Cu+Al) combinations. For each filter used, Compton density measurements were made on various materials and assessment was made on the statistical precision based on the scatter count data and repeated measurements, and the exposure radiation dose was measured using a pen dosimeter. The intention was to find a filter combination which would

give the highest value for the ratio of scatter count to radiation dose and at the same time give fairly accurate values for compton mass density. The experimental results obtained answer this question very well and establish that a hybrid filter, such as (Al+Cu+Al), would be the optimum choice for the Bone Densitometer.

Experiments were also performed in which the discriminator bias (used for processing the D3 transmitted output data) was set at low and high cut-off values to see whether this had any significant effect on density measurements. The base line setting for the single channel analyzer (used to process the scatter output data from D1 or D2 detector) was kept at just above the PM tube single electron noise level so as to gather maximum scatter counts. The results of these experiments show that the density measurements may be affected in some situations by using a high cut-off setting.

The effects of variations in a.c. line voltage and changes in sample size were also studied. This lead to the necessity of using a stabilized a.c. supply for the X-ray source. It also lead to the requirement of using the calibration constant for the lucite standard of nearly the same diameter as the sample under investigation.

6.2. Measurement Procedure :

Repeated measurements were made on electron density determination of various organic and inorganic liquids as

well as solid materials like aluminium, teflon and $\text{CaH}_4(\text{PO}_4)_2 \cdot \text{H}_2\text{O}$, spread over a period of six months. The last sample mentioned was selected since its chemical form is similar to the accepted chemical formula for mature bone i.e. $\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2$ (Calcium Hydroxyapatite) and the compact bone is known to be composed of about 75% hydroxyapatite crystals and 25% organic bone matrix (76).

To measure the electron density of an object, it was placed centrally at the sensitive volume region. Since the ultimate aim was to apply the method at the tibia site for in-vivo determination, the liquid samples were contained in polythene scintillation vials, 2.54 cm. in diameter, and the vials in turn were placed in a smooth fitting central hole in 6.35 cm dia. lucite cylinder. A solid 6.35 cm dia. lucite cylinder was used as the standard for obtaining the calibration constant. Figs.6.1(a) & (b) depict the positioning of the sample under study in the centre of the Densitometer.

The collimated filtered X-ray beam (0.5 cm collimation), was directed at the sample for 5.1 sec duration in position 1 of the source. Digital counts of the scattered and transmission beams were recorded as detected by detectors D1 and D3 respectively. The two CsI(Na) detectors D1 and D2, meant for the scattered beam, were coupled to the counting system via a two way switch. In Pos. 1 of the source, the switch was set to count the scattered intensity as measured by detector D1. The

FIG. 6.1. SET UP FOR DENSITY MEASUREMENT.

- (a) 6.35 cm LUCITE CYLINDER STANDARD IN POSITION FOR DETERMINATION OF CALIBRATION CONSTANT.
- (b) 6.35 cm DIAMETER LUCITE CYLINDER WITH 2.54 cm DIA. CENTRAL HOLE FOR PLACING SCINTILLATION VIALS FILLED WITH SAMPLES, POSITIONED FOR DENSITY MEASUREMENT OF SAMPLE.

COLOURED PICTURE

detected counts in this position are the S1 and T1 counts. Then the entire movable circular gantry was rotated through 90° so that the X-ray source takes Position 2, detector D3 replaces the earlier location of D1 and detector D2 replaces earlier location of D3. The switch was set in the other direction so that now scattered intensity is measured by detector D2 and transmitted intensity is still counted by detector D3. The D3 detector is always directly opposite the X-ray source in both positions of the X-ray source. A 5.1 sec. exposure is again given and the counts recorded are the S2 and T2 counts. The counts obtained in the measurement are corrected for background counts in each detector. The electron density of the material within the sensitive volume is then obtained from :

$$\rho_e = G_w \left[\frac{S_1}{T_1} \frac{S_2}{T_2} \right]_{\text{Sample}}^{1/2} \quad (6.1)$$

The constant G_w is evaluated by doing the same set of measurements for the 6.35 cm dia. solid lucite cylinder. The electron density obtained in this method is relative to that of lucite. The physical density is determined by multiplying the relative electron density by the ratio of the effective (Z/A) for lucite to that of the sample and then multiplying by the physical density of lucite ($= 1.185 \text{ g/cm}^3$) measured for this particular sample of lucite.

$$\rho_m = \rho_e^{\text{sample}} \times \left[\frac{Z}{A} \right]_{\text{lucite}}^{\text{Sample}} \times 1.185 \quad (6.2)$$

The value of the calibration constant is known from the counts S_1 , S_2 , T_1 , T_2 obtained for the lucite standard. Having obtained these counts, the physical density of the material under study is finally obtained from the relation:

$$\rho_m = \left[\frac{Z}{A} \right]_{\text{Lucite}} \cdot \left[\frac{S_1 S_2}{T_1 T_2} \right]_{\text{Sample}}^{1/2} \times 1.185 \quad \text{..... (6.3)}$$

Sample
lucite

which was derived in Chapter 2.

In one set of measurements, shown in Fig.6.7, we have taken water as the standard to obtain the calibration constant. This was done to ensure that the accuracy of density results is not much affected by replacing lucite with water.

6.3. Results of Compton Mass Density Measurements:

The compton mass densities of various specimens were measured under various configurations with the intention ultimately to find the optimum set of conditions for utilizing the method for in-vivo density measurements at the tibia site on human subjects with suitable compromise between statistical precision and a minimal radiation dose to the exposed site.

The results obtained are plotted against Archimedes gravimetric densities measured on a chemical balance. Typical graphs are shown from Fig.6.2 to Fig.6.13.

From eqs.(6.1) and (6.3), it is evident that the measured compton mass density is independent of the

attenuation of the material surrounding the sensitive volume for a very narrow photon beam. However, for a beam of finite cross-section, the transmitted and scattered photons do not traverse equal path lengths and hence the observed density is not totally independent of the attenuation effect. Details of the theoretical considerations and the assumptions used with Compton photon densitometers are explained in detail elsewhere (112, 115).

As can be seen in Figs. 6.2 to 6.13, the Compton mass density was measured for twelve samples of known gravimetrically measured density. The Compton values agree closely with the true (Archimedes) values for most of the substances. The small deviations from the actual densities are within the 3 % uncertainty presently achievable with this Compton X-Ray Bone Densitometer. Thus any bias for correction of measured Compton mass densities is not significant, at least for the density range of 0.77 to about 1.8 g/cm³. It may be observed from the lines of identity drawn in each graph that the mass density as obtained using the Compton Densitometer increases relative to Archimedes values over the range of densities used in this study. The excess in observed density is significant for samples of density above approximately 1.8 g/cm³, and it is in this range where corrections may be necessary to the measured Compton mass densities.

Fig.6.2 shows the results obtained using (1.65 mm Al + 1mm Al) filter and a filament current of 1 mA. The statistical precision of measurement is better than 1% due to the large scatter count rate for most of the samples. This filter was, however, not acceptable in in-vivo studies due to the large radiation dose to the subject as compared with other filter combinations. This matter has been discussed in section 6.5 of this chapter.

Fig.6.3 shows the results using the (Al+Cu+Al) filter combination, 1 mA filament current and 2.1 sec per exposure. The statistical precision in this situation is about 3% and for this set of conditions the exposure dose at the sensitive volume location is of the order of 100 mR per density measurement.

Fig.6.4 shows the results for the same filter combination as above but the filament current and the time for each exposure have been increased to 2 mA and 5.1 Sec respectively in order to attain better statistical precision. In this situation, one obtains better than 2% precision and an exposure dose of the order of 300 mR per density measurement. Tests were also performed by placing the samples in eccentric position in a hole drilled off-centre in a 6.35 cm dia. lucite cylinder. This geometry corresponds to the in-vivo measurements at the tibia site. The results in this case were almost identical to those shown in Fig.6.4. The results for eccentric

positioning are shown in Fig.6.13. Further, the results shown in Fig.6.4 were also unaffected by changing the EG&G discriminator bias from -50 mV to -300 mV for processing the data from the plastic detector, D3. This suggests that (Al+Cu+Al) filter removes a significant amount of unwanted low energy photons from the X-ray beam and a large number of D3 output pulses lie in the region from -300 mV to -500 mV, -500 mV being the maximum height of the negative fast pulses from the D3 detector after amplification by an EG&G fast amplifier.

Figs.6.5 to 6.12 show the results of density measurements using a (Al+Sn+Cu+Al) filter combination and different filament current settings. Although the results obtained are fairly accurate for most of the samples studied, this filter combination was not adopted for in-vivo measurements, due to the reduced statistical precision as compared to (Al+Cu+Al) filter. It may be seen in Fig.6.12 that the density results are significantly higher than the true values for a large number of samples studied when the fast discriminator (used to process signal data from the plastic detector, D3) bias was set to a high level of approx. -350 mV or more. This was possibly due to the underestimation of transmitted counts since a large number of D3 output pulses corresponding to a definite range of X-ray photon energies having the corresponding counterpart in the scattered beam and detected by D1 (or D2) have been cut off due to high discriminator bias

setting. The fraction of D3 output pulses is large in this range for lucite and comparatively small for high density materials such as CCl_4 , H_2SO_4 and teflon. The ratio

$$\left[\frac{S_1 S_2}{T_1 T_2} \right]_{\text{sample}}^{1/2} \div \left[\frac{S_1 S_2}{T_1 T_2} \right]_{\text{lucite}}^{1/2}$$

is therefore overestimated in this situation yielding higher density values. From Fig. 6.11, however, it is apparent that the density results are close to the true values when the fast discriminator bias was set at a low value of about -50 mV.

The tibia density measurements on human subjects discussed in the next Chapter were all obtained using X-ray filament current in the range 1.5 to 2 mA, and 5.1 sec exposure for each X-ray source position, and the source beam was filtered using (1.65 mm Al + 0.8 mm Cu + 1 mm Al) hybrid filter. This situation corresponds to the in-vitro sample measurements depicted in Fig. (6.4). The results of the Compton mass density measurements under these conditions are tabulated in Table 6.1.

Fig. 6.18 & Fig. 6.19 are included here to illustrate the variation in counting rates recorded by the three detectors for different samples in the measuring position. The curves show the order of counts observed in each case. It may be noted that the precision of density measurement is mainly governed by the scattered count rates since the transmitted count rates in D3 are many times higher than the scattered rates.

TABLE 6.1. Compton Mass Density Results using
 (0.8mm Cu + 1mm Al + 1.7 mm Al Prefilter)
 Source filtration and 2.0 mA fil. current
 (collimators: 0.5 cm diameter)

SAMPLE	EFFECTIVE A/Z	ACTUAL DENSITY (g cm ⁻³)	MEASURED COMPTON MASS DENSITY (g cm ⁻³)
Acetone	1.812	0.79	0.76 ± 0.02
Benzene	1.857	0.88	0.85 ± 0.01
Water	1.82	1.00	0.97 ± 0.02
HCl	2.026	1.18	1.21 ± 0.02
Glycerine	1.84	1.26	1.25 ± 0.03
Dextrose	1.875	1.22	1.23 ± 0.02
CCl ₄	2.08	1.58	1.51 ± 0.03
H ₂ SO ₄	1.96	1.84	1.89 ± 0.04
Teflon	2.083	2.14	2.34 ± 0.04
Aluminium	2.077	2.70	3.11 ± 0.09

It was thought worthwhile to make some in-vitro measurements using slit collimators and for this purpose 1 cm x 0.5 cm rectangular lead plugs were inserted to replace the original 0.5 cm dia. circular lead plugs. The results are shown in Fig.6.21. This collimation system, however, was not adopted since it gave a relatively higher radiation dose to the specimen under study without much gain in the statistical precision.

In all these figures shown on compton mass density measurement, the bold lines drawn in each figure at 45° inclination correspond to the line of identity, i.e., the line joining points of equal Archimedes and compton mass densities.

6.4. Variation of Calibration Constant with

Line Voltage changes:

Since the calibration constant is anticipated to change with sample diameter due to multiple scattering or beam hardening effects and with line voltage variation due to X-ray energy spectral shift, a number of experiments were performed to study the effect of these parameters on the measurement of the calibration constant. A lucite cylinder, 6.35 cm in diameter, was once again chosen as the standard.

A variac was used to vary the line voltage supplied to the X-ray machine. The line voltage was varied in the range 110 V to 120 V and monitored with a Weston Model 1240 digital multimeter. Measurements of S1, S2, T1, T2 counts

were made using the lucite cylinder in the sensitive volume location for the different mains voltages chosen. For each mains voltage reading, the ratio $(S_1 S_2 / T_1 T_2)^{1/2}$ was evaluated after appropriate correction for background counts in the three detectors. Fig. 6.14 is a typical plot depicting the calibration constant, G_w , variation with A.C. mains voltage.

6.5. Variation of Calibration Constant with Sample Size:

The calibration constant, G_w , was also evaluated by taking measurements on solid lucite cylinders of different diameters ranging from 1.5 cm to 8.65 cm. This was done for different filtrations applied to the X-ray beam from the source. The results are shown in Figs. 6.15 to 6.17. The variation in calibration constant seen in these curves is mainly due to multiple scattering effects.

It was due to this variation in calibration constant with diameter that the calibration standard used for density measurements was chosen to be of nearly the same diameter as the specimen under consideration. Since the samples in this study were all contained in 6.35 cm dia. lucite cylinder, the calibration standard was a solid lucite cylinder, 6.35 cm in diameter.

6.6 Radiation Exposure Doses:

Fig. 6.20 is a sketch of the room where the Compton X-ray Bone Densitometer is located. A radiation survey was carried out in the room to assess the exposure doses to the

operator and to the subject in various positions when the X-ray machine is switched on. The dose rates shown in the figure were observed with the X-ray source operated at 2 mA, 90 kVp. The measurements were made with a Victoreen Model 444 Exposure measuring system. A 6.35 cm dia. lucite cylinder was placed in the centre of the Densitometer to simulate the density measuring procedure. It may be noted that the dose rate at the operator's seating position behind the concrete brick wall is of the order of 0.05 mR/hr (the same as the background reading), and the dose rate by the side of the writing desk is approx. 0.1 mR/hr. In fact nobody sits by the side of the writing desk while the bone density measurements are being carried out. The operator wears a film badge to keep record of integrated doses received during working with the Densitometer to ensure that the maximum permissible levels as recommended by ICRP are not exceeded.

A radiation survey was also carried out in the immediate vicinity of the Densitometer at different planes above the X-ray set mounting platform to make an assessment of the radiation dose to various limbs of the human subject under study. The density measuring procedure on human subjects is outlined in Chapter 7 and shown in Fig 7.1. Table 6.2 gives approximate exposure dose rates in the proximity of right and left legs of the subject. From the data given in this table and knowing that each density measuring procedure involves two X-ray shots, each 5.1 sec.

TABLE 6.2. DOSE RATES IN THE VICINITY OF X-RAY SET AT VARIOUS PLANES ABOVE X-RAY MOUNTING PLATFORM.*

PLANE HEIGHT ABOVE X-RAY PLATFORM (cm)	DOSE RATE IN THE PROXIMITY OF RIGHT LEG (mR/hr)	DOSE RATE IN THE PROXIMITY OF LEFT LEG (mR/hr)
0	6.5	1.8
4	60.0	3.0
9	40.0	5.8
14	10.0	5.5
18	5.0	4.0
27	2.0	3.0
30	1.8	2.2
35	1.5	1.2
40	1.4	1.0

- * i) X-ray set filament current 2mA at 90 kVp.
 ii) Subject sitting at 25 cms. horizontal distance from X-ray set and at 30 cms. height from X-ray set platform.
 iii) 6.35 cm dia. lucite cylinder, 6 cm high, in centre of Densitometer Table to simulate scattered radiation.

duration, one can calculate the integrated doses to various portions of the human subject. These values range from 0.09 mR to the leg portion immediately above the sensitive volume selected on the right tibia to about 0.002 mR to the gonads. The exposure doses are extremely small and the hazard to the subject is felt to be acceptable.

The exposure doses mentioned in the above paragraphs do not take into account the radiation dose to the region directly in the path of the X-ray beam. The collimated X-ray beam being 0.5 cm. in diameter, a cylindrical region 0.5 to 0.6 cm in diameter across the width of the right leg around the sensitive volume will be exposed to the high flux X-ray beam. An estimate of this dose was made by using a Bendix pen dosimeter Model 886 S.No. 000215 placed in the central 2.5 cm dia. hole in 6.35 cm dia. lucite cylinder, which, in turn, was positioned in the centre of the Densitometer. Two X-ray exposures, each of 5.1 sec duration, were given and the integrated dose as recorded by the pen dosimeter was read. The sensitive region of the pen dosimeter is about 2.5 cm long and a region about 0.6 cm long is intercepted by the X-ray beam. The exposure dose to the specimen is, therefore, nearly four times the reading observed on the dosimeter. All in-vivo measurements described in Chapter 7 were made using a hybrid filter composed of (1.65mm Al+0.8mm Cu+1mm Al) and a filament current of approx. 1.75

mA at 90 kVp, and two exposures, each of 5.1 sec duration. Under these conditions the total exposure dose as estimated from the dosimeter reading was found to be approx. 200 mR.

Radiation doses using the pen dosimeter were recorded for various other filter combinations. For each filter combination, the scattered count rate was also measured with detector D1 (or D2) for the same length of time as the pen dosimeter was exposed. The ratio:

Scattered count rate / Exposure Dose, which may be called the figure of merit for the system, and denoted by the symbol, η , was thereby calculated for each filter combination. The approximate values of η obtained for different filter combinations as applicable to the Densitometer are given below:

(a) 1.65 mm Al filter	100
(b) (1.65 mm Al + .8 mm Cu + 1 mm Al)	220
(c) (1.65mm Al + .5mm Sn + .8mm Cu + 1mm Al)	200

The choice of (1.65 mm Al + .8 mm Cu + 1 mm Al) hybrid filter for in-vivo studies thus becomes obvious.

FIG.62 MEASURED DENSITY VS TRUE DENSITY

FIG. 6.3. COMPTON DENSITY VS ARCHIMEDES DENSITY

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SAMPLES PLACED IN 2.5 cm DIA HOLE IN 6.35 cm DIA LUCITE CYLINDER
 FILTER:- (8 mm Cu + 1 mm Al) + 1.7 mm Al

FILAMENT CONTROL = 30 (± 2 mA) TIME PER EXPOSURE:- 5.1 Sec.

BACKGROUND :- D1:- 60, D2:- 120, D3:- 650

NOTE:- THESE RESULTS NOT AFFECTED
 BY CHANGING EGG DISC BIAS (USED FOR
 PROCESSING D3 OUTPUT) FROM -50 mV
 TO -300 mV

(ii) ALSO NOT DEPENDENT ON ECCENTRICITY
 OF HOLE IN LUCITE CYLINDER

(iii) D1, D2:- CsI(Na) DETECTORS
 D3:- PLASTIC DETECTOR

FIG. 6. 4.

COMPTON MASS DENSITY Vs ARCHIMEDES DENSITY

FIG. 6.5. MEASURED DENSITY VS TRUE DENSITY

FIG. 6.6. MEASURED DENSITY VS TRUE DENSITY

FIG.6.7. MEASURED DENSITY VS TRUE DENSITY

FIG. 6.8 MEASURED DENSITY VS TRUE DENSITY

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FIG. 69 MEASURED DENSITY VS TRUE DENSITY

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FIG. 6.10. MEASURED DENSITY VS TRUE DENSITY

FIG 6 MEASURED DENSITY VS TRUE DENSITY

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FIG. 6.12 MEASURED DENSITY VS TRUE DENSITY

SAMPLES PLACED IN 2.5cm DIA. ECCENTRIC HOLE IN 6.35cm DIAM.
LUCITE CYLINDER

FILTER :- (.8mm Cu + 1mm Al) + 1.7mm AL

FILAMENT CONTROL :- 30 (# 2mA)

TIME PER EXPOSURE :- 5.1 Sec.

BACKGROUND :- D/1:- 60, D2 :- 120, D3 :- 620

FIG. 6.3 COMPTON DENSITY VS ARCHIMEDES DENSITY

CALIBRATION STANDARD: 6.35 cm dia. Lucite
 FILTER: .8 mm Cu + 1 mm Al + 1.7 mm Al Prefilter

110 V = 84 kVp X-Rays
 112.5 V = 86 kVp "
 115 V = 88 kVp "
 117.5 V = 92 kVp "
 120 V = 94 kVp "

S.D. approx 1% for smaller values of Constant
 (Points 1 & 2)
 & approx 1.5% for larger values of
 constant (Points 3 to 6).
 For (115 ± 5) V, Calibration Constant
 = .0416 ± 2.5%

FIGG14 VARIATION OF CALIBRATION CONSTANT WITH MAINS
 A. C VOLTAGE

FIG. 6.15. VARIATION OF CALIBRATION FACTOR WITH LUCITE STANDARD DIAMETER (FOR STUDYING MULTIPLE SCATTERING EFFECT)

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FIG. 6 VARIATION OF CALIBRATION CONST.
WITH LUCITE CYLINDER DIAMETER

FIG 6.17. VARIATION OF CALIBRATION CONSTANT
 WITH LUCITE CYLINDER DIA USING 1cm X
 0.5 cm. SLIT COLLIMATORS

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FIG. 6. B VARIATION OF S1, T1 (POS 1) AND S2, T2 (POS 2) FROM LIQUID TO LIQUID

FIG. 9 VARIATION OF S1, T1 (POS 1) AND S2, T2 (POS 2) COUNTS FROM LIQUID TO LIQUID

FIG. G.20 SKETCH OF ROOM WHERE X-RAY BONE DENSITOMETER IS LOCATED. APPROX. RADIATION DOSE RATES ARE SHOWN AT VARIOUS POSITIONS.

FIG. 62 COMPTON MASS DENSITY VS
ARCHIMEDES USING 1cm. X 0.5cm.
SLIT COLLIMATORS.

CHAPTER 7

IN-VIVO BONE DENSITY MEASUREMENTS ON HUMAN SUBJECTS

7.1. Introduction:

The Compton X-ray Bone Densitometer has been utilized to investigate the trend of normative in-vivo bone density data for normal human subjects. A total of 95 subjects were included in the study (53,54). To get this population of volunteer subjects, an appeal for volunteers was made by the Department of Physics to the Carleton University community and the response was most encouraging indeed. A specimen of this appeal appears in appendix A.

Measurements on thirty three subjects were made during the period mid Aug 78 to mid Oct 78 at the Carleton Sports Medicine Clinic where the two detector X-Ray Bone Densitometer of White (115) was positioned. Study on the rest of the sixty two subjects was carried out at a later period, from Nov 79 to Dec 79 using the later three detector system. The apparatus was located at the Physics Department building for this second set of measurements.

Table 7.1 gives the description of the pertinent data of the subjects and the bone density values as evaluated from the specific electron density of the sensitive volume measured with the densitometer relative to

the lucite standard. Appendix B gives a specimen of the bone density record of each individual together with the declaration which each individual was required to sign before carrying out the test. This was done since the test involves exposing a small section, namely, about 0.2 sq. cm. area at the tibia site, to a beam of X-rays used in the apparatus.

7.2. Measuring Site and Procedure:

The site chosen for density determination was in the sharp ventral edge of the right tibia about 1 cm below the surface and above the ankle at $1/4$ th of tibia length measured from distal end on the outer side of tibia, with the subject in sitting position and right foot resting on a plastic foot rest adjustable both vertically and horizontally. Plastic templates were used to position the sensitive volume region at the centre of the source-detector geometry. Figs. 7.1 (a) to 7.1 (d) depict the position of measuring site and the seating position for density determination.

Measurements were made by giving two 5.1 sec exposures at the chosen site from positions 1 and 2 of the X-ray source. The X-ray beam was collimated to a diameter of 0.5 cm, and the X-ray filament current was set at about 1.75 mA. The X-ray beam was filtered using (1.65 mm Al + 0.8 mm Cu + 1 mm Al) hybrid filter. Under these settings, the exposure dose to the localized region under study was of

FIG. 7.1. DENSITY MEASUREMENT PROCEDURE ON HUMAN SUBJECTS.

(a) ANKLE TO KNEE LENGTH, (b) CIRCUMFERENCE AT CHOSEN TIBIA SITE, (c & d) DIAMETERS IN TWO MUTUALLY AT RIGHT ANGLES POSITIONS AT CHOSEN TIBIA SITE.

COLOURED PICTURE

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(e)

(f)
POS. 1

(g)
POS. 2

FIG. 7.1. (e) POSITIONING OF CHOSEN TIBIA SITE USING PLASTIC TEMPLATE, (f) X-RAY EXPOSURE IN POS. 1 OF SOURCE, (g) X-RAY EXPOS. IN POS. 2.

the order of 200 mR per density determination. As described in the previous chapter, the dose estimate was made by placing a pen dosimeter in a smooth fitting central hole in a 6.35 cm diameter lucite cylinder which in turn was placed in the middle of the Bone Densitometer so that the X-ray beam passes through the dosimeter during each exposure.

The calibration standards used were 5.7 cm dia, 6.35 cm dia. and 7.5 cm dia solid lucite cylinders depending on the diameter of the leg at chosen site.

7.3. Experimental Results:

To examine the correlation, if any, between the physical density of tibia at chosen site vs age or weight, several graphs have been drawn as shown in Figs. 7.3 to 7.11. In the case of males, the data was normalized to 70 kg., 40 yrs old whereas for females normalization was done for 60 kg., 40 yrs. old. Regression lines, drawn in Figs. 7.7 and 7.9, were utilized for normalization in case of males, and Figs. 7.6 and 7.8 for females.

Fig. 7.10 is the histogram of frequency distribution of normalized densities for males and Fig. 7.11 is the histogram for females. From these plots it is seen that both for males and females the curves are slightly skewed towards a density value greater than the mean value. The mean value in both cases is about 1.35 g/cm^3 with FWHM of 0.2 g/cm^3 for males and 0.35 g/cm^3 for females. The peak

value for the skewed distribution occurs at about 1.48 g/cm^3

The mean value of 1.35 g/cm^3 as found in this study, more or less, agrees with the results quoted by Webber et al (112), and Kennett et al (56).

7.4. Conclusions:

For the population studied in this work, it may be seen from Figs. 7.6 to 7.9 that correlation between density vs age and density vs weight for males and females was somewhat poor. From the regression lines drawn in these figures, the approximate relationships obtained are given below:

Males :	Density	$\propto$	(weight) ^{1/3}
	Density	$\propto$	(Age) ^{- 1/20}
Females :	Density	$\propto$	(Weight) ^{1/2.5}
	Density	$\propto$	(Age) ^{- 1/5}

One may try to infer from this data that in case of females, the decrease in density occurs faster than for males with respect to age. Also both for males and females, density varies directly as approximately cube root of weight of subject.

Based on data collected from random Carleton University normal population, we may write the following approximate relationships between the variables : Tibia density, Age, Weight and Sex :

$$D = \alpha (W)^{1/3} (A)^{-1/20} \quad \text{Males}$$

$$D = \beta (W)^{1/2.5} (A)^{-1/5} \quad \text{Females}$$

where D = Physical density obtained from electron density determination,

W = Weight (kg),

A = Age (yrs.)

α, β are constants.

7.5. Discussion:

The question may arise whether the spread of data is real or is due to some systematic errors inherent in the measuring apparatus. So far as the measuring apparatus is concerned, it has been ascertained from a long term repeated in-vitro measurements on various non-biological materials, that a reproducibility of within 5% is achieved with this system. Therefore, the only other factors to account for the spread in normative data collected in this study must be due to the combined effect of the following parameters:

- (a) uncertainty in relocatibility of measuring site,
- (b) variability in ~~density~~ of the subjects investigated.

It has not been possible to establish the extent of error in relocatibility of the measuring site. This would necessitate taking a radiograph of the tibia site to ensure that the same region is under investigation for each

individual. Although, the X-ray source used in the present system could be utilized for this purpose, it was not used for taking tibia radiographs, since the study involved normals and it was undesirable to add additional radiation hazard to the subject.

Our alignment system for relocating the tibia site made use of acrylic templates. They were positioned against the sharp ventral edge about the chosen tibia site. This alignment technique is far from perfect in the sense that in many cases there may be considerable thickness of soft tissue above the ventral edge thereby making it difficult to ensure the same region of tibia is within the sensitive volume.

In the absence of this confirmation, we relied on results previously obtained by White (115) by means of repeated density measurements on the distal radius of five volunteers. The positioning error as estimated by White was of the order of 5%. Keeping this in view, we may attribute about 10% (to be on the high side) contribution to the spread of data on account of positioning error. Then the other 90% contribution to the spread of data comes from the actual variability of density in the subject under investigation.

The second question that arises is whether the variations in density values observed for the group of normals studied with respect to age, sex and weight are real. It is well known that bone density increases with

weight and after attaining a maximum at a certain age starts declining in the 40's. Whether the increase in ~~compton~~ mass density with respect to body weight follows a linear relation is not well established. In the absence of availability of such information, we may assume that the trend shown in this study may be real. The decrease in compton mass density at tibia site with age as predicted by this data seems a bit abnormal. In fact, one expects a faster decline in density with age than observed in this study. Further study is needed to arrive at any conclusion regarding variation of compton mass density with age and sex.

The data obtained for the normals in this study can be interpreted as indicating that: (a) the loss of bone mineral is more rapid in females than males, a fact, which has already been well established. (b) the compton mass density values at tibia site for males normalized to 70 kg, 40 yrs old are approx. the same as for females normalized to 60 kg, 40 yrs old. With normalization done as above, it is found that the average value of compton mass density is equal for both males and females. Rassow (92) has observed a similar trend in his studies comprising 120 subjects.

The results obtained in this study show a range of values of about 1.35 ± 0.30 (FWHM) g/cm^3 for the normals, indicating a spread of about 20 % about the mean value of 1.35 g/cm^3 . It may be noted that this value of FWHM

(which is 2.35 times the standard deviation for a normal distribution) is much greater than the standard deviation associated with the density measurement with the Bone Densitometer (which is of the order of 5 %). This leads to the conclusion that the variability in density values at the tibia site must be about ± 20 %. Thus for normal subjects, both males and females, density values ranging from about 1.1 g/cm^3 to about 1.6 g/cm^3 should be taken as normal values. This spread being relatively large for normal subjects, the Bone Densitometer would be best utilized specially in case of serial studies in clinical cases, where the same subject with bone disease is followed over a length of time from months to years during treatment.

TABLE 7.1. MASTER CHART

(CARLETON UNIVERSITY VOLUNTEER SUBJECTS STUDY GROUP)

S.No.	SUBJECT	SEX/ AGE	WEIGHT (kg)	HEIGHT (cm)	PHY. ACTIVITY	HEALTH/ MED.HIS.	AGE AT MENOPAUSE	LEG CIR. (cm)	MEASURED DENSITY (g/cm ³)
1	A.B.	M/24	77	175	H	N	N/A	25	1.28
2	K.B.	M/21	75	168	A	N	N/A	24	1.27
3	J.H.	M/19	84	180	A	N	N/A	26	1.14
4	L.P.	M/48	73	168	A	N	N/A	24	1.49
5	G.G.	F/62	68	171	H	N	-	38	1.41
6	D.P.	F/64	54	164	A	N	-	19	1.34
7	J.F.	F/54	57	163	A	N	-	22	1.31
8	B.F.	F/38	58	166	A	N	-	24	1.49
9	M.V.	F/29	61	161	A	N	-	34	1.38
10	C.I.	M/27	73	177	A	N	N/A	25	1.45
11	L.E.	M/55	65	165	A	N	N/A	22	1.24
12	G.J.	M/22	66	175	A	N	N/A	23	1.49
13	M.Z.	F/20	61	170	A	N	-	29	1.58
14	K.K.	M/28	73	183	A	N	N/A	22	1.35
15	R.C.	F/34	51	168	A	N	-	21	1.34

TABLE 7.1 - (Contd.)

S.No.	SUBJECT	SEX/ AGE	WEIGHT (kg)	HEIGHT (cm)	PHY. ACTIVITY	HEALTH/ MED.HIS.	AGE AT MENOPAUSE	LEG CIR. (cm)	MEASURED DENSITY (g/cm ³)
16	P.M.	F/24	57	163	A	N	-	23	1.35
17	C.J.	F/36	54	165	A	N	-	24	1.31
18	M.W.	M/31	79	183	H	N	N/A	23	1.46
19	H.V.	F/14	50	150	A	N	-	26	1.18
20	E.C.	M/57	67	163	A	N	N/A	22	1.36
21	C.S.	M/55	79	196	A	N	N/A	23	1.53
22	G.P.	M/20	88	179	A	N	N/A	27	1.66
23	L.D.	M/22	63	172	A	N	N/A	24	1.33
24	M.T.	F/39	57	168	A	N	-	26	1.26
25	P.D.	F/35	55	160	A	N	-	23	1.39
26	A.W.	F/38	64	170	A	N (μ)	-	28	1.73
27	S.A.	M/24	75	180	A	N	N/A	27	1.17
28	V.P.	F/62	68	163	A	N	-	28	1.51
29	L.A.	M/20	72	180	A	N	N/A	25	1.21
30	A.W.	F/19	57	161	H (v)	N	-	23	1.63
31	H.D.	M/31	64	177	H	N	N/A	23	1.40
32	J.M.	F/19	59	170	A	N	-	25	1.36
33		M/38	83	180	A	N	N/A	23	1.29
34	S.D.	F/51	70	163	A	N	-	26	1.51

(μ) Fracture on right leg

(v) Marathon Runner

TABLE 7.1 - (Contd.)

S.No.	SUBJECT	SEX/ AGE	WEIGHT (kg)	HEIGHT (cm)	PHY. ACTIVITY	HEALTH/ MED.HIS.	AGE AT MENOPAUSE	LEG CIR. (cm)	MEASURED DENSITY (g/cm ³)
35	O.J.	M/59	83	175	A	N	N/A	24	1.72
36	K.J.	M/50	77	180	A	N	N/A	24	1.44
37	D.R.	M/53	82	185	A	N	N/A	22	1.43
38	R.S.	F/40	48	155	A	N	-	22	1.19
39	J.P.	F/32	57	170	A	N (τ)	-	22	1.63
40	M.P.	F/37	66	171	A	N	-	24	1.55
41	R.K.	F/30	64	170	A	N	-	26	1.59
42	F.P.	F/32	52	154	A	N	-	24	1.55
43	C.J.	M/31	61	173	A	N (α)	N/A	25	1.28
44	M.B.	F/32	55	157	A	N	-	24	1.38
45	R.H.	M37	70	175	A	N	N/A	23	1.53
46	T.K.	F/50	64	169	A	N (θ)	-	25	1.60
47	S.D.	E/30	44	152	A	N	-	20	1.24
48	M.H.	M/45	72	180	A	N	N/A	22	1.22
49	S.J.	M/43	69	175	A	N	N/A	22	1.29
50	M.G.	M/46	82	178	A	N	N/A	26	1.60
51	B.J.	M/45	73	179	A	N (δ)	N/A	23	1.39
52	N.L.	M/45	73	179	A	N	N/A	24	1.22
53	R.R.	F/33	50	165	A	N	-	21	1.52

(τ) Runner's knee

(α) Left foot crushed in 1977

(θ) Taking estrogen hormone

(δ) Hair line fracture (left toe)
one week before study

TABLE 7.1. - (Contd.)

S.No.	SUBJECT	SEX/ AGE	WEIGHT (kg)	HEIGHT (cm)	PHY. ACTIVITY	HEALTH/ MED.HIS.	AGE AT MENOPAUSE	LEG CIR.	MEASURED DENSITY (g/cm ³)
54	C.I.	F/32	57	173	A	N	-	24	1.55
55	E.I.	F/46	70	170	A	N (ρ)	-	26	1.67
56	C.T.	M/59	64	173	A	N	N/A	23	1.00
57	B.T.	F/30	51	152	A	N	-	22	0.99
58	H.J.	M/60	109	185	A	N (ω)	N/A	26	1.65
59	J.P.	M/41	66	182	A	N	N/A	24	1.47
60	B.I.	F/50	76	163	A	N	41	27	1.35
61	B.B.	M/58	67	173	A	N	N/A	23	1.15
62	P.E.	F/34	68	175	A	N	-	22	1.58
63	M.M.	F/47	64	165	A	N	46	21	1.17
64	F.J.	F/35	60	173	A	N	-	22	1.50
65	M.M.	M/51	69	173	A	N (θ)	N/A	22	0.98
66	S.B.	M/21	60	175	A	N (η)	N/A	23	0.95
67	B.L.	M/38	77	183	A	N	N/A	26	1.03
68	A.S.	M/52	81	185	A	N	N/A	25	0.95
69	M.R.	M/40	68	173	A	N	N/A	21	1.39
70	D.M.	M/37	89	185	A	N	N/A	22	1.00
71	T.C.	M/65	83	185	A	N	N/A	23	1.54
72	W.D.	M/54	76	173	A	N	N/A	24	1.29

(ρ) Hypothyroid; on thyroxine

(ω) Arthritis (since past six years)

(θ) Hammer Toes (both) since age of 25

(η) Slight Achilles (right foot) Rupture

TABLE 7.1. - (Contd.)

S.No.	SUBJECT	SEX/ AGE	WEIGHT (kg)	HEIGHT (cm)	PHY. ACTIVITY	HEALTH/ MED.HIS.	AGE AT MENOPAUSE	LEG CIR. (cm)	MEASURED DENSITY (g/cm ³)
73	D.V.	F/20	59	170	A	N	-	23	1.29
74	M.P.	M/48	75	183	A	N (β)	N/A	23	1.09
75	G.N.	F/45	68	155	A	N	-	26	1.10
76	D.J.	F/46	49	155	A	N	-	20	1.20
77	G.G.	F/63	57	171	A	N	45	24	0.97
78	R.W.	F/56	52	160	A	N (γ)	50	23	1.34
79	G.S.	F/48	50	151	A	N	48	21	1.04
80	C.V.	F/57	64	168	A	N	54	22	1.18
81	H.E.	F/58	60	173	A	N (ε)	-	23	0.97
82	D.A.	M/56	82	175	A	N	N/A	26	1.15
83	K.K.	M/39	84	180	A	N	N/A	26	1.38
84	C.R.	M/57	71	175	A	N	N/A	23	1.52
85	M.K.	M/50	59	173	A	N	N/A	21	1.18
86	R.D.	M/43	95	178	A	N (σ)	N/A	25	1.68
87	N.H.	M/66	77	183	A	N	N/A	23	1.00
88	B.D.	M/49	86	183	A	N (†)	N/A	23	0.94
89	U.P.	F/59	67	175	A	N	45	20	1.00
90	J.C.F	32	64	161	A	N (λ)	-	24	1.54
91	B.J.	F/30	66	165	A	N	-	24	1.32

(β) Bone Spur (Right) toe

(γ) Splay feet

(ε) Osteo-Arthritis; extra bone growth at
back near 5th lumbar

(σ) Diabetic

(†) Paget's disease; Right arm was broken
a year before study(λ) Stress factor in left foot ankle;
Hypothyroid

TABLE 7.1. - (Contd.)

S.No.	SUBJECT	SEX/ AGE	WEIGHT (kg)	HEIGHT (cm)	PHY. ACTIVITY	HEALTH/ MED.HIS.	AGE AT MENOPAUSE	LEG CIR. (cm)	MEASURED DENSITY (g/cm ³)
92	W.I.	M/34	64	182	A	N	N/A	24	1.18
93	I.W.	M/58	73	184	A	N	N/A	24	1.47
94	K.E.	F/40	58	170	A	N	-	24	1.16
95	C.B.	M/45	70	178	A	N	N/A	23	1.09

ABBREVIATIONS USED

M	-	Male
F	-	Female
A	-	Average
N	-	Normal
H	-	Heavy
N/A	-	Not applicable
PHY.	-	Physical
MED.HIS.	-	Medical History

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FIG. 7.2 MEASURED PHYSICAL DENSITY VS WT. IN MALE SUBJECTS

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FIG. 7.3. MEASURED PHYSICAL DENSITY VS BODY WEIGHT IN HUMAN SUBJECTS (FEMALES)

FIG.7.4. MEASURED PHYSICAL DENSITY VS AGE IN HUMAN SUBJECTS(MALES)

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FIG. 7.5.

MEASURED PHYSICA DENSITY VS AGE IN HUMAN SUBJECTS (FEMALE)

FIG.76.LOG (PHYSICAL DENSITY) VS LOG (WEIGHT)
 IN HUMAN SUBJECTS (MALES) NORMALISED
 TO 40 YRS. OLD MALE

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FIG. 7.7.
LOG (PHYSICAL DENSITY) vs LOG (WEIGHT) IN
HUMAN SUBJECTS (FEMALES) NORMALIZED
TO 40 YRS OLD FEMALE.

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FIG. 7.8.
 LOG (PHYSICAL DENSITY) VS LOG (AGE) IN HUMAN
 SUBJECTS (MALES) NORMALIZED TO 70Kg MALE.

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FIG. 7.9.
LOG (PHYSICAL DENSITY) VS LOG (AGE) IN HUMAN
SUBJECTS (FEMALE) NORMALIZED TO 60 Kg FEMALE.

FIG. 7.10.
FREQUENCY DISTRIBUTION OF LOG (DENSITY) IN
HUMAN SUBJECTS (MALES). DENSITY NORMALIZED
TO 70 Kg. MALE 40 YRS OLD.

PEAK CORRESPONDS TO
 PHYSICAL DENSITY OF 1.51 g/cm^3
 $\overline{\log D} = .133 \quad \bar{D} = 1.36 \text{ g/cm}^3$
 FWHM = $.33 \text{ g/cm}^3$

FIG. 7.11.

FREQUENCY DISTRIBUTION OF LOG (DENSITY)
 IN HUMAN SUBJECTS (FEMALES), DENSITY
 NORMALIZED TO 66% FEMALE 40 YRS OLD.

CHAPTER 8

DISCUSSION AND CONCLUSIONS

8.1. Reproducibility and Radiation dose:

The Compton X-ray Bone Densitometer provides a fairly precise method for determining the mineral content of bone through the measurement of electron density of bone at a specific site. Typically, in-vivo measurements on bones of the extremities, such as the radius and the tibia, are reproducible to within about 5% (S.E.). Measurements on non-biological inert samples like organic and inorganic liquids with densities ranging from 0.7 to 1.8 g/cm³ are reproducible to within about ± 1.5 %. The measurements on liquids have been performed under conditions simulating tibia of an adult with surrounding tissue.

The difference between the standard errors in the results for in-vivo measurements and on inert samples may be attributed to the non-reproducibility in relocating the measuring site in in-vivo determinations. The precision of measurements may be improved by repositioning and remeasuring the bone several times and taking the average. This of course will be at the expense of excessive radiation dose to the subject. In the study of inert samples, precision can be improved to high levels by improving the scattered photon statistics since the question of radiation

dose is irrelevant in this situation. The ideal approach to improving the precision in in-vivo measurements is to reduce the repositioning error itself. This will require taking an X-ray picture of the measuring site before carrying out the density measurement. Reproducibility in relocating the measuring site rather than photon counting statistics will still remain the problem of most important concern. There is no point in improving photon statistics either by means of increased counting time or higher X-ray filament current since this will be at the expense of increased radiation, hazard to the subject and the positioning error far outweighs any advantages that one could attempt to obtain from statistics.

8.2. Choice of X-ray Machine:

The present system has made use of a 90 kVp X-ray machine with appropriate filtration. It would be better from radiation dose consideration to use a source with 150 kVp or thereabouts output, since this will shift the peak intensity towards a much higher energy value closer to 90 keV. Here the absorbed dose is a minimum (56), since the photoelectric absorption has a much lower value in this energy region and the Compton scattering cross-section shows an increased value, a point of considerable importance for improving photon statistics which is mainly limited by the scattered beam which is orders of magnitude smaller than the transmitted beam. It will be worthwhile to remember that

even a small increase in kilovoltage of the X-ray output greatly increases the number of high energy photons thereby considerably reducing the radiation dose. Beyond 150 kVp, although the photoelectric absorption is further reduced, the radiation dose increases due to increased energy of the photon beam, since the absorbed dose is determined by the product of photon energy and linear absorption coefficient.

8.3. Filtration :

The use of aluminium-copper hybrid filter has been found to be adequate for cutting out the unwanted lower energy X-ray flux and thereby shifting the spectral energy curve towards the higher energy value. Unfortunately, however, some of the higher energy photons are also removed so that this type of purification of the beam cannot be continued indefinitely with profit (107a). A point is ultimately reached where any further filtration reduces the intensity of the desired high energy components to such an extent that further filtration is useless, and to obtain good photon statistics one has to resort to increasing X-ray filament current to obtain greater incident flux which is not desirable.

The present Densitometer has made use of self-rectified X-ray machine using pulsating A.C. Voltage. It may be pointed that for the same peak voltage, the effective photon energy using pulsating voltage is considerably less than that using constant D.C. potential

(107a) . This is true in both diagnostic and therapeutic X-ray beams. For this reason it would be theoretically advantageous to use a constant potential X-ray machine for the Bone Densitometer; however, such a generator would be more costly and complex and may not be practical for a simple and portable Bone Densitometer of the type developed here.

8.4. Choice of Scintillators for D1 & D2 Detectors:

In the present study, CsI(Na) scintillators have been used for the two detectors meant to detect Compton scattered X-ray beam. It may seem advantageous to replace them with NaI(Tl) scintillators since their decay time is about four times shorter and light yield about 15% larger than CsI(Na); this in turn will result in a better signal to noise ratio keeping in view the fact that single electron photomultiplier tube noise is only a few nanoseconds wide.

8.5. Choice of measuring site:

Various sites have been advocated from time to time for bone density determination and questions are asked as to why one or the other site is not suitable. The two most popular locations have been the distal radius and the os calcis. The os calcis is said to be too sensitive to changes in metabolism and the mobility of the patient (18, 82). Its sensitivity to activity levels and its irregular cross-section make it a somewhat poor choice. The radius is easy to locate and the bone shape does not change too

rapidly at the distal end. This makes the bone suitable for measurement. The humerus can be measured in small children but, because many adults have considerable tissue cover over the humerus and because the bone is significantly larger than the radius and hence cannot be measured in some adults, it does not represent a universally usable measuring site.

The metacarpals and phalanges have also been studied (94). The smaller size of these bones makes them promising as the amount of multiple scattering will be small. The bones of the finger are quite easy to immobilize for study. There is, in addition, only a small amount of cover tissue.

In the present study, we have chosen the tibia site. This is chiefly because of ease of repositioning. It is true, the scatter of the data is higher for this bone as compared to radius and to achieve same precision as for radius, much larger radiation dose is delivered to the subject at the selected site. The system developed is, of course, flexible enough to make measurements on any of the sites mentioned above. In clinical cases of lower limb dysfunction, the proximal end of tibia has been recommended as a sensitive site for bone mineral measurement (2).

8.6. Conclusion :

The Compton X-ray Bone Densitometer described in the present study has great potential for absolute and relative density measurements in-vivo of peripheral bones such as radius, ulna, tibia and os calcis in a clinical

study. The method provides precise and accurate density measurements in-vivo and makes it possible to study irregular shaped bones. A detailed knowledge of the surrounding material is not required. It is expected that the system as described will be useful for the accurate detection of small serial changes in bone density. The use of X-ray source in place of radioactive sources provides high photon flux thereby considerably reducing exposure time, and allowing finer collimation. It also makes possible to assist in the location of area under study. Standard deviation of the order of $\pm 1.5\%$ from counting statistics is achieved with exposure dose to the localized region under study less than 200 mR. Repositioning of bone site under study reduces the precision to about $\pm 5\%$. Multiple scattering and beam hardening effects limit the ultimate accuracy that can be attained.

Finally the system has been used to collect bone density data at the tibia site for normal human subjects. It has been shown that both for males and females, on normalization with respect to age and height, the average Compton mass density obtained is about 1.35 g/cm^3 , and the trend for bone mineral loss with age is more rapid in females than males. These results broadly agree with those in published literature (56, 92, 102) .

APPENDIX A

Carleton University
Ottawa, Canada K1S 5B6

TO THE CARLETON COMMUNITY, AN APPEAL FOR VOLUNTEERS

The undersigned are engaged, with a graduate student, Mr. K. K. Kapoor, in the development of a device for measurement of the state of health of human bone, in this instance the amount of mineral present. We now need to establish the normal variation of bone mineral with age, sex and physical activity. For this we need volunteers who are willing to submit their tibia (the shin bone) for measurement. The actual measurement involves no pain, no undressing or other indignities, a truly negligible amount of X-rays on about 1 cm of the leg and about 10 minutes of your time. We have done some work on the student population. Although more young people are needed, we now particularly need females and males of the 30 to 65 years range. Can you help us? We would be most grateful. For an appointment, please phone Mrs. Donna Jessup or Mrs. Roselyn

O'Driscoll at 6630. The hours during which we can accept appointments are 10-12 A.M. and 2-5 P.M. Mondays to Thursdays, from now until December 20th.

R. L. Clarke R. D. Barton Department of Physics

Department of Physics. Herzberg Laboratories.

(613)231-6630

APPENDIX B
PROFORMA FOR BONE DENSITY RECORD

BONE DENSITY RECORD

No. _____

DATE _____

TIME _____

1. SUBJECT (NAME)

2. AGE

3. SEX

4. HEIGHT

5. WEIGHT

6. PHYSICAL ACTIVITY (Sports, Work)

7. MEDICAL HISTORY

7a. Normal legs

(Hip, Knee, Foot, Ankle, Length)

7b. Kidney Disease

7c. Bone Disease

7d. Recent Fractures

7e. Other Metabolic

Disorders

8. FOR FEMALES ONLY

IF YOU ARE PREGNANT, OR THINK THERE IS ANY REASONABLE
POSSIBILITY THAT YOU ARE PREGNANT, WE CAN'T PROCEED
WITH THE TEST. THANK YOU ANYWAY

8a. Children (full or Near Full Term - Carried to 7th Month)

Yes _____ No _____

8b. Nursing _____ 8c. Menopause Yes _____ No _____ Age _____

9. FOR OFFICE USE ONLY

9a. Knee to Ankle Length _____

9d. Measured electron
density9b. One fourth above length
from ankle

9e. Physical density

9c. Leg circumference at
above position

1. I hereby agree to have an experimental x-ray test made on my leg. It has been explained to me that the x-ray will be directed through approximately one square-centimetre of the surface of my skin.
2. I understand that this experiment is to assist the development of a device for measuring the health of human bone, and that it is being conducted by researchers in the Department of Physics at Carleton University. It is not a test made by physicians, surgeons, or radiologists, nor is it part of a more general assessment of my health.
3. I have freely and willingly given information about myself on the attached sheet; and to the best of my belief that information is correct, and there is no other information which is relevant. The information which I have given is strictly confidential and may be used for this test only. It may be used by the researchers only in a way which does

not reveal my identity.

4. I have volunteered for the test; there has been no coercion or inducement whatsoever. I hereby release Carleton University, its employees and officers, from all consequences, and waive every and all claims which may at any time now or in the future have for compensation or damages, direct or indirect.

Date:

(Witness)

(Signature)

REFERENCES

1. Abramowitz M and IA Stegun (eds.): "Handbook of Mathematical Functions, Nat. Bur. Std. Appl. Math. Ser. 55, Washington, 1968.
2. Andersson S and BE Nilsson: "The Proximal End of Tibia, a Sensitive Site for Measurement of Bone Mineral Content in Lower Limb Dysfunction"; Proceedings of Third Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
3. Anliker M and W Baumgartner: "Quantification of Bone Mineralization in Growing rats by γ -ray computed Tomography"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
4. Arnold JS, WE Barnes, N Khedkar and M Nelson: "Computerized Kinetic Analysis of Two Tc-99m-Sn-Diphosphonates, demonstrating different binding characteristics"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
5. Attix F, W Roesch and E Tochilin: "Radiation Dosimetry"; Academic Press, New York, 1969.
6. Barnett E and BEC Nordin; Br. J. Radiol., 34, 683, 1961.

- 7.. Bailey RR and A Stokes: "Bone Mineral Content of Patients on Hemodialysis and Following Renal Transplantation"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
8. Cameron EC, RM Boyd, D Luk, HW McIntosh and VR Walker: "Cortical thickness measurements and photon absorptiometry for determination of bone quantity"; CMA Journal, Vol. 116, 145-147, 1977.
9. Cameron JR and JG Skofronick: "Medical Physics"; John Wiley & Sons, N.Y., Chichester, Brisbane, Toronto, 1978.
10. Cameron JR and J Sorenson: "Measurement of Bone Mineral In-vivo: An improved method"; Science, 142, 230-232, 1963.
11. Cameron JR and JA Sorenson: "A Reliable Measurement of Bone Mineral Content In Vivo"; J. Nucl. Med., 8, 268, 1967.
12. Cameron JR, RB Mazess and JA Sorenson: "Precision and accuracy of bone mineral determination by direct photon absorptiometry"; Invest. Radiol., 3, 141-150, 1968.
13. Cameron JR: "Physical Methods of Monitoring Bone Status"; Phys. Med. Biol., 15, 184, 1970.

3 3
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14. Chestnut CH et al: "Correlation of Total Body Calcium (Bone Mass) as Determined by Neutron Activation Analysis with Regional Bone Mass as Determined by Photon Absorption"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
15. Clarke RL and G Van Dyk: "A New Method for Measurement of Bone Mineral Content using both Transmitted and Scattered Beams"; Phys. Med. Biol., 18,4, 532-539, 1973.
16. Clarke RL, ENC Milne, MB ChB and G Van Dyk: "The use of Compton Scattered Gamma Rays for Tomography"; Investigative Radiology, 11, 3, 1976.
17. Clarke RL, D Johnson, ME King and MW White; "X-Ray Bone Densitometer"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June 1978.
18. Cohn SH, MS Roginsky, JF Aloia, KJ Ellis and KK Shukla: "Alternations in skeletal Calcium and Phosphorus in Dysfunction of the Parathyroid"; The Journal of Clinical Endocrinology & Metabolism, 36, 4, 1973.
19. Cohn SH, KK Shukla & KJ Ellis: "A Multivariate Predictor for Total Body Calcium in Man Based on Activation Analysis"; Journal of Nuclear Medicine and Biology, 1, 131-134, 1974.

20. Colbert C, RB Mazess and PB Schmidt; Invest. Radiol.,
5, 336, 1970.

21. Colbert C, PB Schmidt and RB Mazess: "Bone Mineral Determination:
A Comparison of Radiographic Photodensitometry and
Direct Photon Absorptiometry"; Phys. Med. Biol.,
15, 372, 1970.

22. Colbert C and RS Bachtell: "Radiographic Absorptiometry:
System for Measuring Mineral Changes in Patients with
Bone Disease"; Radiological Research Laboratory,
Greene Memorial Hospital, Ohio, 1979 (to be published).

23. Dalen N: "Choice of Measuring Sites"; Proceedings of Int.
Conf. on Bone Mineral Measurement held at Chicago from
12-13 Oct., 1973.

24. Donath A, P Indermuhle and R Baud: "Influence of the Natural
Calcium and Fluoride Supply and of a Calcium
Supplementation on Bone Mineral Content of
Healthy Population in Switzerland"; Proceedings of Int.
Conf. on Bone Mineral Measurement held at Chicago
from 12-13 Oct., 1973.

25. Dyck GE: "Evaluation of a Single-Photon Compton Scattering
Method for the in-vivo Monitoring of Skeletal
Mineralization"; M.Sc. Thesis, Univ. of Manitoba, 1980.

26. Ebifegha ME: "Theoretical and Experimental Studies of
Compton Scatter Densitometry"; M.Sc. Thesis,

Carleton University, 1980.

27. Eilbert R: "In-Vivo Calcium Determination by Proton Activation Analysis"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
28. Ellis KJ, D Vartsky, TB Pearlstein, JL Alberi and SH Cohn: "A Dual Photon Absorptiometry System Utilizing a HpGe Detector and Microprocessor Controller"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
29. Elsasser U, GU Exner, A Prader and P Ruegsegger: "Gamma-Ray CT for Selective Analysis of Trabecular and Compact Bone: Effects of Treatment with Estrogen, Testosterone and Growth Hormone in Children"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
30. Evans RD: "The Atomic Nucleus"; Int. Series in Pure and Applied Physics, McGraw-Hill, New York, 1955.
31. Evens RG et al: "Clinical Use of an In Vivo Measurement of Bone Mineral Content"; Proceedings of "Progress in Methods of Bone Mineral Measurement" held at Bethesda, Md from 15-17 Feb., 1968.
32. Farmer FT and MP Collins: "A Further Appraisal of the Compton Scattering Method for Determining Anatomical

Cross-sections of the Body"; Phys. Med. Biol.,
19, 6, 808-818, 1974.

33. Fitzgerald JJ, GL Brownell and FJ Mahoney: "Mathematical Theory of Radiation Dosimetry"; Gordon and Breach Science Publishers, Inc., New York, London, Paris, 1967.
34. Fogelman I, DL Citrin, RG Bessent and R Mazess: "The Development and Validation of Measurement of 24-hour Bone Uptake of Tc-99 HEDP as an Index of Metabolic Bone Disease"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
35. Fuss N; Ann. Rev. Rheum. Dis., 32, 588-9, 1973.
36. Garnett ES, TJ Kennett, DB Kenyon and CE Webber; Radiology, 106, 209, 1973.
37. Gebhardt M and H Zwicker: "A New Apparatus for Bone Mineral Measurement in vivo"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
38. Genant HK et al: "Skeletal Demineralization in Primary Hyperparathyroidism"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
39. Gilson AJ, MJ Cohen and F Day: "Bone Density Measurement

- by means of Radioisotopic Source Photon Attenuation"; J. Nucl. Med., 6, 345 (abstract), 1965.
40. Griffiths HJ and RE Zimmerman: "The Clinical Application of Bone Mineral Analysis: An Overview"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
- 40a. Grodstein GW: "X-Ray Attenuation Coefficients from 10 kev to 100 Mev"; National Bureau of Standards Circular, 583, 1957.
41. Hanson J: "Analysis of Gd-153 and I-125/Am-241 Sources"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
- 41a. Harrison JE and KG McNeill: "Partial Body Calcium Measurements by In Vivo Neutron Activation Analysis"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
42. Heuck F; Progress in Methods of Bone Mineral Measurement, 1968.
43. Horsman A: "Measurements of Bone Mineral Content In Vivo using a Multiwire Chamber as Detector"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
44. Huddleston AL and D Bhaduri: "Compton Scatter Densitometry in Cancellous Bone"; Phys. Med. Biol., 24, 2, 310-318, 1979.
45. Huddleston AL and D Bhaduri: "Measurement of Bone Density

Using Single Source Compton Scattering Methods";
 Proceedings of 4th Int. Conf. on Bone Measurement
 held at Univ. of Toronto from 1-3 June, 1978.

46. Isotom: Bone Density System, Deptt. of Biomedical Eng.,
 S.F.I.T. and Univ. of Zurich, Switzerland, 1979.

47. Jackson JD: "Classical Electrodynamics"; John Wiley & Sons,
 Inc., New York, London, 1962.

47a. Jacobsen B: "X-ray Spectrometry in vivo"; Amer. J.
 Roentgenol., Radium Ther. Nucl. Med., 91,
 202-210, 1964.

48. Johnston CC Jr., DM Smith, P Yu and WP Deiss Jr.: "In vivo
 measurement of bone mass in the radius"; Metab.,
 Clin. Exp., 17, 1140-1153, 1968.

49. Johns HE and JR Cunningham: "The Physics of Radiology";
 Charles C Thomas, Springfield, Illinois, USA, 1969.

50. Judy PF: "Physical Aspects of I-125 Bone Absorptiometry";
 Proceedings of Int. Conf. on Bone Mineral Measurement
 held at Chicago from 12-13 Oct., 1973.

51. Judy PF and JR Cameron: "A Dichromatic Absorption Method
 for the Measurement of Bone Mineral Content in vivo";
 Phys. Med. Biol., 15, 372, 1970.

52. Kan WC, CR Wilson, RM Witt and RB Mazess: "Direct Readout
 of Bone Mineral Content with Dichromatic Absorptio-
 metry"; Proceedings of Int. Conf. on Bone Mineral

Measurement held at Chicago from 12-13 Oct., 1973.

53. Kapoor KK and RL Clarke: "Development of Compton Scatter X-Ray Technique for Bone Density Measurement In-Vivo"; AMPI Medical Physics Bulletin (abstr.); 5, 2, April, 1980.
54. Kapoor KK, RL Clarke and RD Barton: "Development of Compton Scatter X-Ray Technique for Bone Density Measurement in-vivo"; AMPI Medical Physics Bulletin (extended abstr. accepted for publication), Oct., 1980.
55. Karjalainen P: "A method for determination of the mineral content and mineral density of the distal radius using gamma ray attenuation"; Annals of Clinical Research, 5, 231-237, 1973.
56. Kennett TJ and Webber CE: "Bone Density Measured by Photon Scattering. II. Inherent Sources of Error"; Phys. Med. Biol., 21, 5, 770-780, 1976.
57. Klein O and Y Nishina; Physik, 52, 853, 1929.
58. Klemm T, DH Banzer and U Schneider: " Bone Mineral Content of the Growing Skeleton"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
59. Kennett TJ, ES Garnett and CE Webber: "An in vivo measurement of absolute bone density"; Journal De L'association Canadienne des Radiologistes, 23, 1972.

60. Kerr SA, K Kouris, CE Webber and TJ Kennett: "Coherent Scattering and the assessment of mineral concentration in trabecular bone"; Phys. Med. Biol., 25, 6, 1037-47, 1980.
61. Kuhlencordt F, JD Ringe, HP Kruse and AV Roth: "Bone Mineral Determination of Radius, Ulna, and Finger bones by 1125 Photon Absorptiometry on Healthy Persons"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
62. Kuo SS: "Numerical Methods and Computers"; Addison-Wesley Pub. Co., Mass., Palo Alto, London, Ont., 1965.
63. Lachmann E and M Whelan; Radiology, 26, 165, 1936.
64. Lapp RE and HL Andrews: "Nuclear Radiation Physics"; Prentice-Hall, New Jersey, 1972.
65. Luther R: "Correlation of Os Calcis and Spinal Bone by Compton Scattering"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
66. Leichter I, G Hazan, E Loewinger, A Weinreb and GC Robin: "Measurement of Bone Density by Compton Scattering"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
67. Mack B and F Voft; Am. J. Roentgen. Rad. Therapy & Nuc. Med., 113, 4, 621, 1971.

68. Mazess RB, JR. Cameron, R O'Connor and D Knutzen: "Accuracy of Bone Mineral Measurement"; Science, 145, 388, 1964.

69. Mazess RB, M Ort, PF Judy and W Mather: "Absorptiometric bone-mineral determination using Gd 153"; In "Bone Measurement Conference", USAEC CONF 700515, pp. 308-312, U.S. Dept. of Commerce, 1970.

70. Mazess RB, JR Cameron and JA Sorenson: "Body Composition by Radiation Absorption at two energies"; Phys. Med. Biol., 15, 372, 1970.

71. Mazess RB and JR Cameron: "Bone Mineral Content in Normal U.S. Whites"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.

72. McDonald JM and L Zeitz: "Dual Energy Absorptiometry Technique for Bone Mineral Content Measurement"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.

72a. McNeill KG, JE Harrison, TA Bayley, WC Sturtridge and C Williams: "The Value of Partial Body Calcium Measurements by IVNAA in Clinical Long-Term (5 Year) Longitudinal Studies"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.

73. Meema HE, CK Havus and RE Porrett; Radiology, -82, 986, 1964.

74. Meema H, et al; Radiology, 110, 513-22, 1974.
75. Morgan DB, FW Spiers, CN Pulvertaft and P Fourman; Clin. Radiol., 18, 101, 1966.
76. Mortimer A: "The Measurement of Bone Mineral Content using Scattered and Transmitted X-rays"; M.Sc. Thesis, Carleton University, Canada, 1974.
77. Naftchi NE, AT Viau, GH Sell and EW Lowman: "Effect of Thyrocalcitonin on F-18 uptake and various Biochemical variables in spinal-cord injured subjects with Periarticular Bone formation"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
78. Nalcioğlu O and RY Lou: "Post-reconstruction Method for Beam Hardening in Computerised Tomography"; Phys. Med. Biol., 24, 2, 330-340, 1979.
79. Neer RM, JE Mahaffey, M Daly, D Lans and JT Potts: "The detection of Bone Disease in Primary Hyperparathyroidism"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.
80. Nord RH: "Extracorporeal Bone Scanning"; Norland-Cameron Bone Mineral Analyzer Applications Note No. 3.
81. Norland-Cameron: "Questions and Answers about Bone Mineral Analysis"; Norland-Cameron Bone Mineral Analyzer

Applications Note No. 7.

82. Norland-Cameron; Circular, Norland Instrument Company, 1971.
83. Olkkonen H, et al: "Measurement of Bone Mineral Density using Coherent and Compton Scattering"; Proceedings of Int. Conf. on Bone Mineral Measurement held in New Orleans from 26-28 Jan., 1976.
84. Omnell KA: "Quantitative roentgenologic studies on changes in mineral content in bone in vivo"; Acta Radiol., Suppl. 148, 1957.
- 84a. Overton TR, DS Silverberg, WM Rigal and L. Friedenberg:
"University of Alberta Bone Mineral Analysis System - Performance and Clinical Application"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
85. Phelps ME, EJ Hoffman and M Ter-pogossian Michel: "Attenuation Coefficients of Various Body Tissues, Fluids, and Lesions at Photon Energies of 18 to 136 kev"; Radiology, 117, 573-583, 1975.
86. Piper DG, LE Preuss and FP Bolin: "An Evaluation of Several nuclides for Bone Density Determinations by Compton Scattering"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.
87. Piper DG and Preuss LE: "Absolute Bone Density Measurement

- using Compton Scattered Radiation"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
88. Preuss LE, et al: "Investigation of Certain Errors in Dual Photon Tissue Analysis"; Proceedings of 4th Int. Conf. on Med. Phys. held in Ottawa from 25-30 July, 1976.
89. Price WJ: "Nuclear Radiation Detection (2nd ed.)"; McGraw Hill Book Co., N. Y., San Fran., Toronto, London, 1964.
90. Price RR, J Wagner, K Larsen, J Patton and AB Brill: "Regional and Whole-Body Bone Mineral Content Measurement with a Rectilinear Scanner"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
91. Puumalainen P, A Uimarihuhta, E Alhava and H Olkkonen: "A New Photon Scattering Method for Bone Mineral Density Measurements"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
92. Rassow J: "First Results of a Two-Energy Densitometry Method for Measuring Bone Mineral Concentrations and Bone Densities in vitro and in vivo"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.
93. Reiss KH, K Killig and W Schuster: "Dual Photon X-Ray

Beam Applications"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.

94. Regal R, M Rossi and L Lamarque; J. Radiol.; Electrol., Med. Nucl., 53, 452, 1972.

95. Rokoff SD; Progress in Development of Methods in Bone Densitometry, 1965.

96. Roos B, B Rosengren and H Skoldborn: "Determination of bone mineral content in lumbar vertebrae by a double gamma ray technique"; In "Bone Measurement Conference", USAEC CONF 700515, pp. 243-253. U.S. Dept. of Commerce 1970.

97. Roy RR and BP Nigam: "Nuclear Physics"; John Wiley and Sons, Inc., New York, 1967.

98. Ruegsegger P, P Niederer and M Anliker: "A Method for the Determination of the Compacta Area and the Mean Absorption Density of Human Bones"; Proceedings of Int. Conf. on Bone Mineral Measurement held at Chicago from 12-13 Oct., 1973.

99. Sorenson JA and JR Cameron: "A reliable in vivo measurement of bone mineral content"; J. Bone Joint Surg., 49, 481-497, 1967.

100. Sorenson JA and RB Mazess: "Effects of fat on bone mineral measurement"; In "Bone Measurement Conference",

USAEC CONF 700515, pp. 255-262. U.S. Dept. of
Commerce, 1970.

101. Sorenson JA and JR Cameron: "Transmission Scanning Techniques
In 'Instrumentation in Nuclear Medicine' (GJ Hine and
JA Sorenson, eds.), Vol. 2, pp. 349-384, Academic Press,
N.Y., London, 1974.
102. Smith DM, et al; J. Am. Med. Ass., 219, 3, 1972.
103. Smith E and JR Cameron: "Interpretation of Fracture Index
Charts"; from 'Bone Mineral Analysis'; distributed
by Norland Instruments, 1972.
104. Smith EL: "The Effects of Physical Activity on Bone
in the Aged"; Proceedings of Int. Conf. on Bone
Mineral Measurement held at Chicago from 12-13 Oct., 1973.
105. Smith MA, P Tothill and JA Strong: "In-vivo Neutron
Activation Analysis of Partial Body Calcium using
Californium-252"; Proceedings of 4th Int. Conf. on
Bone Measurement held at Univ. of Toronto from
1-3 June, 1978.
106. Snell AH (ed): "Nuclear Instruments & their uses, (Vol 1),
John Wiley & Sons, N.Y., London, 1962.
107. Spiers FW: "Effective Atomic Number and Energy Absorption
in tissues"; Proceedings of Hospital Physicists
Association held at Birmingham in April, 1944.

107a. Stanton L: "Basic Medical Radiation Physics"; Appleton-Century-Crofts, Education Div., Meridith Corp., New York, U.S.A., 1969.

108. Stash AM and RW Bright: "Recent Advances in Skeletal Trans-imaging"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.

109. Taylor KE: "The significance of Errors in X-ray Images"; from a talk presented at CAP Congress, York Univ., Toronto, 9-12 June, 1975.

110. Watson RC: "Bone Growth & Physical Activity in Young Males"; Proceedings of Int./Conf. on Bone Mineral measurement held at Chicago from 12-13 Oct., 1973.

111. Webber CE: "Experience with Photon Scattering Measurements of Bone Density"; Proceedings of 3rd Int. Conf. on Bone Mineral Measurement held at New Orleans from 26-28 Jan., 1976.

112. Webber CE and Kennett TJ: "Bone Density Measured by Photon Scattering. I. A System for Clinical Use"; Phys. Med. Biol., 21, 5, 760-769, 1976.

113. Webber CE: "Progress in Compton Scattering Measurements of Bone Density"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from

1-3 June, 1978.

114. West RR and GW Reed: "The measurement of bone mineral in vivo by photon beam scanning"; Br. J. Radiol., 43, 886-893, 1970.

115. White MW: "A Study of Bone Density Measurements using the Photon Scattering Technique"; M.Sc. Thesis, Carleton University, Canada, 1977.

116. Wilson CR, C James and DJ Volz: "Quantitative Computed Tomography"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.

117. Witt RM and PR Moran: "Calcium Content Analysis in-vivo with Computed Tomography"; Proceedings of 4th Int. Conf. on Bone Measurement held at Univ. of Toronto from 1-3 June, 1978.

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