The effect of fat on the coherent-to-Compton scattering ratio in the calcaneus: A computational analysis

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(Received 3 April 1995; accepted for publication 5 May 1995)

The coherent-to-Compton scattering ratio (CCSR) is a technique that has been proposed for measuring trabecular bone mineral density (TBMD). This paper investigates the effect of fat on the CCSR and its correlation to the error in TBMD measurements. It is a computational study to determine the relationship between the magnitude of fat error and the momentum-transfer variable x, which represents the incident photon energy and the scattering angle. Variation in fat content contributes significantly to the error in CCSR measurements. When employing a typical ^{241}Am source ($E_\gamma = 59.45$ keV), the resulting error decreases with increasing momentum-transfer variable or angle. For example, the error ranges from +14 mg/cc at an angle of 45° ($x = 18.3$) to +3 mg/cc at an angle of 135° ($x = 44.3$) for an osteoporotic trabecular region (100 mg/cc mineral) of a calcaneus that contains 6% less fat than a calibration standard. The error is about 0.3–1.2 mg/cc less for regions containing 2–3× more bone mineral and is reduced and opposite in sign for regions containing about 7% more fat than the calibration standards (e.g., −9 mg/cc at 45° and −1.5 mg/cc at 135°). Others have shown that the intrinsic sensitivity of the CCSR method for measuring TBMD at a given photon energy generally increases with increasing detector angle. Thus large angles are advantageous both for reduced sensitivity to fat variation and increased sensitivity to bone mineral variation. The primary disadvantage is reduced count rates that degrade precision unless long counting times are employed. For experimental studies, a compromise angle must be chosen to insure adequate counting statistics for reasonable precision and examination times while providing moderate mineral sensitivity and moderate fat error.

Key words: scattering, fat, bone mineral density, trabecular, calcaneus

I. INTRODUCTION

Puumalainen *et al.*[1] were the first to describe the method of using the coherent-to-Compton scattering ratio (CCSR) to measure trabecular bone mineral density (TBMD). An advantage of the CCSR method is that the BMD measurements are relatively unaffected by the attenuation effects of surrounding tissues. These attenuation effects are nearly the same for coherent and Compton scattered photons because they have similar energies. Another advantage is, like CT, the ability to separately analyze the trabecular bone. Trabecular bone is approximately 8× more active metabolically than cortical bone and is therefore a better indicator of disease states.[2] Trabecular bone has a different structure than cortical bone but its elemental composition is the same. One of the most commonly used measurement sites for CCSR is the calcaneus, which contains mostly trabecular bone. At this site in adults, the trabecular bone is surrounded by yellow marrow, which is approximately 80% fat.[3]

Compositional effects, such as variations in fat content, may cause an inaccuracy in the measurement of BMD by CCSR. It has been determined that the CCSR is dependent upon the effective atomic number (Z_{eff}) of the scattering material. One might presume that the error due to fat is negligible, since the Z_{eff} of bone mineral is much greater than that of fat. However, Ling *et al.* found, experimentally, that there is approximately a 2.5% decrease in the measured CCSR for every 10% increase in fat content.[4] They used ashed bone samples saturated in increasing concentrations of tertbutyl alcohol ($Z_{\text{eff}} = 6.15$)–water solutions. Their CCSR system employed an ^{241}Am ($E_\gamma = 59.45$ keV) source and a 90° detection angle. Tertbutyl alcohol was used as a fat ($Z_{\text{eff}} = 5.94$) simulator.[4] (The above-listed Z_{eff}'s were calculated using the method described in Johns and Cunningham.[5])

A number of beam energies and numerous detection angles have been employed by TBMD researchers using the CCSR technique. Stalp and Mazess noted that there is an increased Compton energy shift with higher energy beams, which allows the use of small-angle scattering.[6] They also stated the Compton and coherent cross sections decreased with increasing energy. This is advantageous for penetrating tissue surrounding the assay site but has the disadvantage of decreased scatter.[6] Kerr *et al.*, using a ^{153}Sm ($E_\gamma = 103.2$ keV) source, chose a 22.5° detection angle.[7] At 15°, the coherent peak was not detectable while at 28°, the coherent peak was noticeable but had poor shape. An acceptable compromise was 22.5° because the coherent peak was fairly well resolved. Karellas *et al.* stated that theoretically, for 60-keV photons and the range of Z corresponding to trabecular bone, increasing the scatter angle results in a stronger power dependence of the CCSR on Z.[8] This implies that by increasing the scatter angle, smaller changes in the TBMD can be detected and thus one improves the sensitivity of the TBMD measurement. Their experiment supported this hypothesis. This experiment, using K_2HPO_4 as the assay material and ^{241}Am ($E_\gamma = 59.45$ keV) as the source, indicated that an

 0094-2405/95/22(8)/1229/6/$6.00

increasing sensitivity (slope of the CCSR versus bone mineral concentration line) at larger scatter angles more than compensates for the decreased precision caused by the lower counting statistics.[8] This conclusion was also experimentally verified by Gigante and Sciuti, using $Ca_3(PO_4)_2$ in a glycerine mixture as the assay material and ^{241}Am as the source.[9] They also stated that the backscattering mode ($\theta=135°$) produces an increase in relative sensitivity and enables the use of a compact measuring head.[9] These advantages compensate for the decrease in counting rate in the coherent peak. Larger scattering angles result in better separation of the energies of the Compton and coherent peaks. Leichter *et al.* stated that if one keeps the angle constant and increases the photon energy, the improvement in energy separation does not compensate for loss in precision, which is caused by lower counting statistics.[10]

Ndlovu, Farrell, and Weber[11] performed both theoretical and experimental analyses of the minimum detectable bone mineral difference of a coherent scatter technique as a function of angle. Both the intrinsic sensitivity and the variance of the measurement were taken into account. They found that "for an incident photon energy of 60 keV, the minimum detectable bone mineral difference is practically independent of scattering angle while for an incident energy of 100 or 122 keV the scattering angle must be less than 70° to optimize the minimum detectable difference."[11] Furthermore, according to these authors, small scattering angles must be avoided because the shape of the scattering volume is difficult to confine to the scattering region and the energy separation between Compton and coherent peaks is small.

The experimental error encountered in the CCSR technique is ~5%.[12] This error has been attributed to fat and counting statistics.[4] As described above, others have determined the relationship between counting statistics errors and the momentum-transfer variable (i.e., photon energy and scatter angle). To our knowledge, the relationship between the magnitude of the fat error and the momentum-transfer variable has not been determined. In this paper, we describe a computer-simulation study to evaluate this latter relationship.

II. THEORY

Coherent scattering has negligible energy losses and is predominantly forward directed due to the small recoil given by the atom. The scattered photon changes direction and thus the momentum is transferred to the atom or bound electron. The atom is neither ionized nor excited. The differential coherent scattering cross section per atom is the differential Thomson scattering cross section per electron modified by the square of the atomic form factor $F(x,Z)$ and is as follows:

$$\frac{d\sigma_{\text{coh}}}{d\Omega}=\frac{r_e}{2}(1+\cos^2\theta)[F(x,Z)]^2, \tag{1}$$

where r_e is the classical electron radius, which is 2.817×10^{-15} m, θ is the scattering angle, and Z is the atomic number of the scatterer.[13] The variable x represents the momentum transfer that occurs in the scattering process and is as follows:

$$x=\frac{1}{\lambda}\sin(\theta/2), \tag{2}$$

where λ is the photon wavelength in angstroms (Å).[13]

Compton scattering is the scattering of a photon by a free or a loosely bound electron, where the photon imparts some of its energy to the electron and changes direction. The differential cross section for Compton or incoherent scattering is proportional to the mass density of the scattering material and decreases with increasing incident photon energy. It is the differential Klein–Nishina collision cross section per free electron modified by the incoherent scattering function $S(x,Z)$ and is as follows:

$$\frac{d\sigma_{\text{Com}}}{d\Omega}=\frac{r_e}{2}[1+k(1-\cos\theta)]^{-2}\left[1+\cos^2\theta+\frac{k^2(1-\cos\theta)^2}{1+k(1-\cos\theta)}\right]S(x,Z), \tag{3}$$

where k is the photon energy in electron rest-mass energy units [$k=E(\text{eV})/511003.4$] and θ, r_e, x, and Z have been previously defined.[13] $S(x,Z)$ is the product of the number of electrons per atom and the probability that an atomic electron undergoes a transition to an excited state or the continuum. This transition is a result of the momentum transfer. The binding energy of a tightly bound electron may be greater than the free-electron recoil energy and thus may not be involved in the incoherent scattering event.[13] Equation (3) may be referred to as the differential incoherent (bound-electron Compton) scattering cross section per atom. The differential cross section units are $(m^2/\text{atom})/\text{sr}$.

The coherent-to-Compton scattering ratio (CCSR, which will be represented mathematically as R) may be constructed by dividing Eq. (1) by Eq. (3) as follows:

$$R=\frac{d\sigma_{\text{coh}}}{d\Omega}\Big/\frac{d\sigma_{\text{Com}}}{d\Omega}=h(\theta,E)\frac{[F(x,Z)]^2}{S(x,Z)}, \tag{4}$$

where

$$h(\theta,E)=[1+k(1-\cos\theta)]^2(1+\cos^2\theta)\times\left[1+\cos^2\theta+\frac{k^2(1-\cos\theta)^2}{1+k(1-\cos\theta)}\right]^{-} \tag{5}$$

$h(q,E)$ is a function that is dependent upon the scatter angle and the incident photon energy.[10]

The CCSR has a power relationship to Z in the region of elemental interest for trabecular bone and fat. This power dependence is based upon the ratio F^2/S and is independent of $h(\theta,E)$. It may be written as follows:

$$R\alpha Z^b, \tag{6}$$

where b is the power dependence or exponent. Olkkonen *et al.*[14] showed for the atomic number range of 7.5–100, b is approximately 3. The effective Z's of fat, yellow marrow,

TABLE I. The elemental composition and Z_{eff} of selected body tissues.

Body tissue	Elemental composition (% by mass)[a] H	C	N	O	Elements with $Z>8$	Z_{eff}[b]	ρ(g/cc)
Adipose 2	11.4	59.8	0.7	27.8	Na(0.1), S(0.1), Cl(0.1)	6.47	0.950[c]
Adipose 3	11.6	68.1	0.2	19.8	Na(0.1), S(0.1), Cl(0.1)	6.27	0.933[c]
Cortical bone	3.4	15.5	4.2	43.5	Na(0.1), Mg(0.2), P(10.3), S(0.3), Ca(22.5)	13.84	1.92
Yellow marrow	11.5	64.4	0.7	23.1	Na(0.1), S(0.1), Cl(0.1)	6.36	0.945[c]

[a]Elemental composition based on Woodard and White (Ref. 3).
[b]Based on calculation shown in Johns and Cunningham (Ref. 5).
[c]Calculated from water, fat, protein, and mineral compositions given in Woodard and White (Ref. 3).

trabecular bone, and bone mineral (calcium hydroxyapatite) are 5.94, 6.36, 13.84, and 16.24, respectively.[3,4] With the Z_{eff} having a cubed power dependence, one might infer that the CCSR of a mixture of trabecular bone in yellow marrow would be dominated by the bone mineral contribution. Therefore, bone mineral measurements made with this technique would be relatively invariant to the effects of fat. This study investigates the latter supposition for a physiological range of fat content.

III. COMPUTATIONAL METHOD

We chose to model the calcaneus because this is a common site for clinical measurements.[7,15] The average TBMD in the calcaneus for normal healthy subjects is 250 mg/cc.[15] This corresponds to a mass percent bone of 36.2% (assuming the mass fraction of mineral:bone is 0.58, $\rho_{bone}=1920$ mg/cc, and $\rho_{yellow\ marrow}=945$ mg/cc)[2]. TBMDs of 100 (16.7% bone mass fraction), 200 (30.8%) and 300 mg/cc (42.8%) were chosen to simulate patients with disuse osteoporosis (paraplegic patients), low mobility healthy people, and highly healthy subjects, respectively.[15] As previously stated, yellow marrow, which is very similar to adipose tissue in its composition,[3] mostly surrounds trabecular bone in the calcaneus. Woodard and White defined three types of adipose tissues, adipose 1, 2, and 3.[3] Adipose 2 is the mean adipose composition tissue whereas adipose 1 and 3 are the mean minus one standard deviation and mean plus one standard deviation in composition, respectively.[3] Based on the typical variability ($\pm 10\%$) in fat content of yellow marrow,[2,4] we used the following "marrows" in this study: adipose 2 (74.1% fat), yellow marrow (80.4% fat), and adipose 3 (87.3% fat).[3] These form a reasonable range with which to calculate the effect of fat on the CCSR. We eliminated adipose 1 (61.4% fat)[3] and glycerol trioleate (100% fat)[3] from consideration as effective substitutes for yellow marrow, because they contained too little or too much fat. Table I describes the elemental compositions, effective atomic numbers Z_{eff}, and mass densities ρ of the tissues considered in this study.

To calculate the CCSR, one must obtain the momentum-transfer variable x value. Table III lists the momentum-transfer variables, which are determined by the particular incident energies and scattering angles chosen. They have been used by researchers in previous experiments. The values x and Z are necessary to find the atomic form factor $F(x,Z)$ and the incoherent scattering function $S(x,Z)$ from the tables of Hubbell *et al.*[13] A method described by Tartari and Casnati to determine the molecular weight specific atomic form factors and scattering form factors was used and they are[16]

$$F_m^2=\sum_i \left[\left(\frac{w_i}{A_i}\right)F^2(x,Z_i)\right] \tag{7}$$

and

$$S_m=\sum_i \left[\left(\frac{w_i}{A_i}\right)S(x,Z_i)\right]. \tag{8}$$

To demonstrate the validity of the CCSR calculation, the value of b, which is the power dependence of the CCSR on Z (for $Z=8-11$), was computed and compared to values obtained by previous researchers. A theoretical calibration curve for the CCSR at 71° using calcium hydroxyapatite in petrolatum as described experimentally by Leichter *et al.* was also computed.[17]

The magnitude of the error in the CCSR determined TBMD due to fat content variation was estimated by the following procedure: First, using Eq. (4), the CCSR was computed as a function of mass percent bone for mixtures of bone in adipose 2, adipose 3, and yellow marrow. This computation was performed at various momentum-transfer values (angle and energy combinations). In each circumstance (momentum-transfer value), a plot of CCSR versus mass percent bone in yellow marrow was used as the calibration line. The CCSRs for given amounts of bone (e.g., those corresponding with 100, 200, and 300 mg/cc mineral) in the other marrows (adipose 2 and adipose 3) were computed. The cor-

TABLE II. Computed mass percentages of bone for osteoporotic (100 mg/cc mineral), healthy (200 mg/cc mineral), and very healthy (300 mg/cc mineral) bone mixed in the various marrows. The mass percentages were computed using Eq. (10).

TBMD (mg/cc)	Marrow type	Mass percentage bone (%)
100	adipose 2	16.62
	yellow	16.70
	adipose 3	16.88
200	adipose 2	30.67
	yellow	30.78
	adipose 3	31.06
300	adipose 2	42.70
	yellow	42.83
	adipose 3	43.14

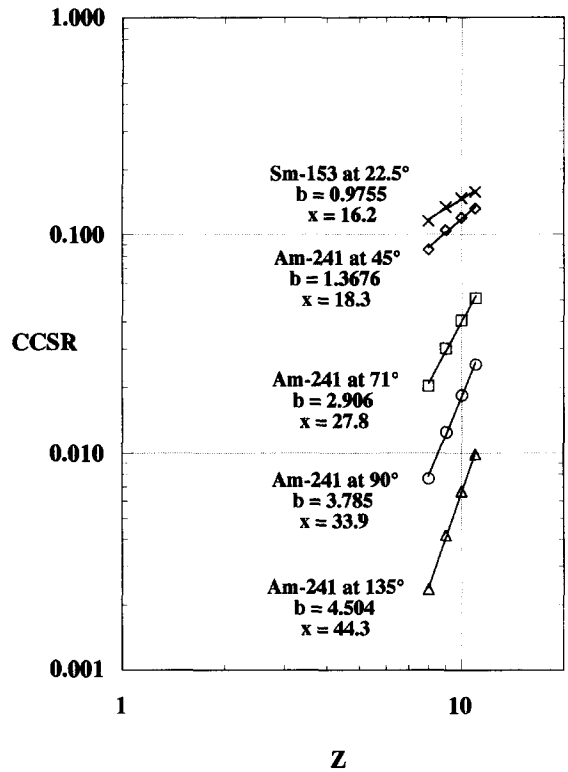

FIG. 1. The coherent-to-Compton scattering ratio calculated for integer atomic numbers in the range ($8 \leq Z \leq 11$). The lines and the value b, which is the power dependence, were determined by a power curve fit. x is the momentum-transfer variable.

responding mass percentages of bone were determined from the calibration line. The resultant mass percentages were converted to mineral density, and the errors were computed as the differences between the true mineral densities and the values determined from the calibration line. The equations employed to convert mass percent bone (B) to mineral density (c_{mineral}) and vice versa are the following:

$$c_{\text{mineral}} = 0.58 \times \frac{B}{(B/1000 \times \rho_{\text{bone}}) + (100 - B/1000 \times \rho_{\text{marrow}})} \tag{9}$$

and

$$B = 100 \times \frac{c_{\text{mineral}}}{0.58} \times \left[\left(1 - \frac{c_{\text{mineral}}}{0.58} \times \frac{1}{1000 \times \rho_{\text{bone}}} \right) \times \left(1000 \times \rho_{\text{marrow}} + \frac{c_{\text{mineral}}}{0.58} \right) \right]^{-1}, \tag{10}$$

where c_{mineral} is in units of mg/cc and 0.58 is a factor that represents the mass fraction of mineral in bone. ρ_{bone} and ρ_{marrow} are units of g/cc.

IV. RESULTS

In Fig. 1, the CCSR is plotted against $Z(=8$ to 11) for the noted energies and angles. A power curve fit was applied and the power dependence b was determined. When the momentum-transfer variable x is small (<20), b is less than two ($b<2$). This implies that there might be less sensitivity to the change in TBMD. At $x>20$, the slopes on the log–log plot are steeper and the implication is that the CCSR technique is much more sensitive to TBMD changes. Using an ^{241}Am source ($E_\gamma = 59.45$ keV), the slope is significantly greater at 135° ($b=4.5$) than at 90° ($b=3.785$). This agrees with Leichter *et al.*[10]

TABLE III. Linear regression fits to the computed CCSR vs mass percentage bone data for all the situations studied in this paper.

E_γ (keV)	θ (deg)	Momentum-transfer variable x	Marrow type	Slope	Intercept	r value
103.2	22.5	16.2	adipose 2	0.001062	0.06131	0.99992
			yellow	0.001084	0.05893	0.99991
			adipose 3	0.001100	0.05712	0.99991
59.45	45	18.3	adipose 2	0.000786	0.04266	0.99991
			yellow	0.000805	0.04060	0.99992
			adipose 3	0.000818	0.03910	0.99992
59.45	71	27.8	adipose 2	0.000318	0.00798	0.99997
			yellow	0.000324	0.00732	0.99996
			adipose 3	0.000327	0.00688	0.99993
59.45	90	33.9	adipose 2	0.000222	0.00283	0.99983
			yellow	0.000224	0.00257	0.99987
			adipose 3	0.000225	0.00240	0.99992
59.45	135	44.3	adipose 2	0.000136	0.00079	0.99996
			yellow	0.000136	0.00071	0.99995
			adipose 3	0.000137	0.00064	0.99995

Table II lists the computed mass percentages of bone for various concentrations of mineral in different marrows.

The equations of the linear regression fits to the CCSR versus mass percent bone data for all situations studied are listed in Table III. Notice that all correlation coefficients r are approximately equal to 1.0, which indicates the high degree of linearity for these functions.

Table IV summarizes the errors in the CCSR determined TBMD due to variations in fat content for the situations studied.

V. DISCUSSION

The results in Table IV demonstrate that 6%–7% variations in the fat content of the marrow within the calcaneus can lead to significant errors in estimating the TBMD when using the CCSR. The absolute values of the errors are essentially independent of the magnitude of the true TBMD. The values are about the same (within ±1.5 mg/cc) for the "osteoporotic" (100 mg/cc), "healthy" (200 mg/cc), and "very healthy" (300 mg/cc) individuals. Consequently, as a percentage, the errors are much greater for the osteoporotic individuals, who would likely make up the majority of a study population. For these individuals, the errors are as large as 13.8%. When a typical ^{241}Am source is employed ($E_\gamma = 59.45$ keV), the error due to fat content variability decreases with an increase in momentum-transfer variable or scatter angle. For example, a reduction in fat content by 6% (e.g., yellow marrow→adipose 2) results in errors in the estimated TBMD of an osteoporotic individual of +13.8%, 10.8%, 6.2%, and 2.9% at angles of 45°, 71°, 90°, and 135°, respectively. These errors follow the same trend, but are

TABLE IV. Errors in computed mineral contents due to variations in fat content. Plots of CCSR vs mass percent bone for bone in yellow marrow (80.4% fat) were used as calibration lines for computing the mineral concentrations. Errors are listed for true values of 100, 200, and 300 mg/cc mineral in adipose 2 (74% fat) and adipose 3 (87.3% fat). [Error=computed TBMD-true TBMD, (mg/cc mineral).]

Momentum-transfer variable x	100 mg/cc in adipose 2	100 mg/cc in adipose 3	200 mg/cc in adipose 2	200 mg/cc in adipose 3	300 mg/cc in adipose 2	300 mg/cc in adipose 3
16.2 ($E=103.2$ keV, $\vartheta=22.5°$)	+11.8	−8.1	+11.4	−7.3	+10.9	−6.6
18.3 ($E=59.45$ keV, $\vartheta=45°$)	+13.8	−9.1	+13.3	−8.2	+12.6	−7.4
27.8 ($E=59.45$ keV, $\vartheta=71°$)	+10.8	−6.9	+10.3	−6.5	+9.6	−6.3
33.9 ($E=59.45$ keV, $\vartheta=90°$)	+6.2	−3.3	+5.8	−2.7	+5.5	−2.3
44.3 ($E=59.45$ keV, $\vartheta=135°$)	+2.9	−1.5	+2.6	−0.7	+2.3	−0.2

smaller in magnitude and opposite in sign for a comparable increase in fat content [e.g., −9.1%, −6.9%, −3.3%, and −1.5% at angles of 45°, 71°, 90°, and 135°, respectively, for a 6.9% increase in fat content (yellow marrow→adipose 3)].

Errors for the small momentum-transfer variable associated with the CCSR technique suggested by Kerr *et al.*[7] ($x=16.2$, $E_\gamma=103.2$ keV, $\theta=22.5°$) are fairly large and comparable to those for ^{241}Am at 45°.

Our computed errors for the situation studied by Shukla *et al.*[12] ($x=27.8$, $E_\gamma=59.45$ keV, $\theta=71°$) range from +5.2% to −3.2% for a healthy individual. These values are in good agreement with their experimentally determined accuracy of ±5%.

Our results indicate that the error due to fat content variability can be reduced significantly by employing a large scatter angle or large momentum-transfer variable. However, there are two disadvantages associated with large scatter angles. The first is a decrease in counting statistics, which is caused by less scatter. The second is increased differential attenuation by the surrounding tissues due to a greater difference between the energies of the Compton and coherent scattered photons. In terms of the uncertainty in TBMD, the former problem is counteracted somewhat by the increase in bone sensitivity with an increase in scatter angle.[8,10] The degree to which this occurs is subject to debate.[11] The latter problem is minimized by selecting a measurement site, such as the calcaneus, which has little overlying tissue.

In general, for experimental CCSR studies, a compromise scatter angle must be chosen which insures (1) adequate counting statistics for reasonable precision, without requiring excessively long examination times or high radiation dose, and immunity to motion artifacts, (2) a well-defined scatter volume, and (3) adequate intrinsic sensitivity. Our study indicates that another factor must be considered. This is the scatter angle dependence of the error in TBMD estimates due to variations in fat content. This error can be as large as 14% at small angles such as 45°, and it is less than half this value (~6% or less) at larger angles, such as 90°.

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