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Influence of fat on bone measurements with dual-energy absorptiometry

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Summary

In order to investigate the influence of fat on bone in dual-energy absorptiometry measurements, we evaluated a special phantom on the three scanners: Lunar DP3, Lunar DPX and Hologic QDR-1000. The phantom employed hydroxyapatite blocks of various thicknesses to simulate bone, water to simulate muscle and lucite to simulate fat. The lucite plates were arranged in one and two layers in three different configurations: over the whole measurement area, over the hydroxyapatite blocks only and at both sides of the hydroxyapatite blocks.

For all scanners, no influence of fat could be demonstrated if it was homogeneously distributed over the whole measurement area. However, changes in area bone-density were observed if fat was distributed inhomogeneously over the measurement area. Fat over only the bone area reduced the measured bone values by 0.051 g/cm^2 per cm fat layer. Fat over only the soft-tissue area increased the measured bone values by the same amount. These results apply to the Lunar DPX scanner. The results for the Lunar DP-3 scanner are similar; those for the Hologic QDR-1000 show a slightly smaller fat dependence of 0.044 g/cm^2 per cm fat layer. The fat influences are not dependent on the amount of bone and only minimally on the soft-tissue thickness.

A change of 50% in the fat content of the bone marrow will change the measured area bone-density of an averaged sized vertebra by 5–6% depending on scanner model. Inhomogeneous fat distribution in soft tissue, resulting in a difference of 2 cm fat layer between soft-tissue area and bone area, will influence the measured area bone-density by 9–10%.

Key words: Dual-photon absorptiometry; Dual-energy X-ray absorptiometry; Bone measurement; Fat influence; Osteoporosis

Introduction

Since the commercialization of dual-photon absorptiometry, the accuracy of the method under various conditions has been a major topic of discussion. One factor

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influencing the measurement of bone mineral is the unknown composition of soft tissue in the beam path through the bone. An approximate value for this soft tissue parameter is obtained by measuring the attenuation behavior of soft tissue where bone is not present. Specifically, the ratio of the attenuation coefficient of the high- and low-energy beam, which depends on the composition of the soft tissue in the beam path, is measured. The same ratio is assumed to apply for soft tissue above and below the bone as well as for bone marrow. Two different implementations can be found in commercial systems for the calculation of the soft-tissue attenuation-ratio. In one case, the ratio is calculated on a line-by-line basis and applied to the bone measured on the same scan line [1]; in the other case, the ratios are averaged for all scan lines and applied uniformly to the whole bone [2]. Both methods extrapolate the measured soft tissue ratio to the area of the bone, but neither has the ability to take into account the soft tissue ratio of the marrow. However, the shorter pathway of the photon beam through marrow compared to over- and underlying soft tissue minimizes adverse influences of variable marrow compositions.

Several studies have investigated the accuracy of dual-energy absorptiometry by comparing the measured mineral content with the ash weight of bone [3,4]. Although the correlation coefficients are consistently very high (>0.95), the standard error of the estimate is the more relevant parameter, because it provides an indication of the deviation of the measurement from its true value. This error is commonly about 10% and reflects the combined errors of the absorptiometry method as well as the ashing and weighing procedure. Whereas some part of this error is due to the limited precision of the measurement methods used, another part is likely based on the unknown fat content of the bone marrow. Because these studies generally involve the absorptiometric measurement of excised bones, possibly suspended in some soft-tissue-equivalent liquid, the error of soft-tissue inhomogeneity is not taken into account.

For this study, we set out to assess the influence of fat in soft tissue and bone marrow on measured bone-mineral density. Both, homogenous and inhomogeneous distribution of fat were studied on three instruments: Lunar DP3, Lunar DPX and Hologic QDR-1000. In order to keep the measurement conditions reproducible for all three instruments, a phantom was used to simulate the various measurement situations.

Methods and Materials

The phantom representing bone in soft tissue was an epoxy block, 66 × 150 × 175 mm (h × w × l), containing in the center three hydroxyapatite inserts of 35 × 50 mm (w × l) and nominal thicknesses of 15, 26 and 40 mm, respectively (Fig. 1). The area densities of these blocks cover a range of 0.7–1.9 g/cm^2 as measured by the Lunar scanners.

The phantom block was measured under various combinations of fat and soft-tissue layers. In dual-energy absorptiometry, the characterizing parameter for soft tissue or fat is the ratio of the mass attenuation coefficient of the two photon energies.

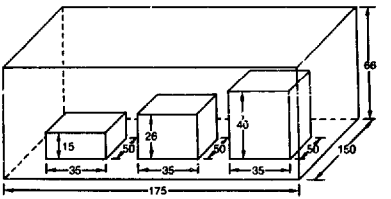

Fig. 1. Phantom representing three thicknesses of bone in soft tissue. The soft tissue block consists of epoxy, the bone inserts of hydroxyapatite.

From the candidate materials available, we chose water for soft tissue and lucite for fat (cf. Appendix).

For the measurements, the phantom was placed at the bottom of a plastic container and submerged in water. The water level for the first measurement was 50 mm above the surface of the epoxy block. The whole block was scanned, covering the hydroxyapatite inserts and sufficient area left and right for determination of the soft-tissue parameters. The scan field was limited to the area of the epoxy block. Additional scans were performed with several configurations of lucite plates added on top of the epoxy block (Table 1), always keeping the water level 50 mm above the epoxy block. The size of the lucite plates was 23 × 50 × 175 mm (t × w × l). Each plate was one third the width of the epoxy block and could be positioned over the hydroxyapatite blocks in the center or over the epoxy/soft-tissue area on either side. Six lucite plates were used, ranging in thickness from 22.70 to 23.60 mm. With the arrangement given in Table 1, we simulated three distinct cases: (i) fat added homogeneously across the whole measurement area, which represents various proportions of fat in the soft tissue of a patient; (ii) fat added only to the area above the bone which is equivalent to adding fat to the bone marrow or above/below the bone

Table 1

Experimental set-up

Measurement no.	Number of lucite plates	
	Over hydroxyapatite	Over epoxy
1	–	–
2	1	1
3	2	2
4	1	–
5	2	–
6	–	1
7	–	2

Positioning of lucite plates for the simulation of fat effects.

(this represents patients with a higher concentration of fat in soft tissue or bone marrow along the beam path through bone as compared with soft tissue on either side of the bone); and (iii) fat added only to the soft tissue on both sides of the bone (where the soft-tissue ratio is measured) simulating, as in the second case, inhomogeneous fat distribution, but now with a higher fat concentration in the soft tissue on both sides of the bone.

In order to study the influence of fat for various patient sizes, the same set of seven measurements as listed in Table 1 was repeated with water levels of 100 and 200 mm above the epoxy block. As three hydroxyapatite thicknesses were measured in each configuration, the influence of fat changes at various bone mass levels could be documented as well.

The evaluation of the measurements was done with the software provided by the manufacturers. In order to exclude effects of edge detection and only to consider the response of the scanners to fat changes, the results were obtained for an area inside each of the hydroxyapatite blocks and expressed in g/cm^2. Hologic provides a software option that allows for the definition of a region of interest for averaging values. For the Lunar DP3, individual profiles had to be evaluated by carefully controlling the soft-tissue level and moving the left and right edge point into the flat area at the top of the curve. The Lunar DPX evaluations were made in a similar way as the Hologic evaluations by defining a region of interest.

Results

The three sets of experiments with water levels of 50, 100 and 200 mm over the epoxy block rendered the same results for each scanner. Apart from the statistical noise which increased with larger absorber thickness, no systematic difference was observed due to increased water levels. For this reason, detailed presentation of the results will be restricted to those measured at 50 mm water level. Figure 2 shows the measured area density of the seven different configurations of fat simulation for the three hydroxyapatite blocks and the three scanners.

It is obvious from the data, that the noise increases with increasing thickness of hydroxyapatite blocks. Also the larger noise overall for the Lunar DP3 scanner suggests a lower photon flux compared to the X-ray scanners.

The experiments with 0, 1 and 2 lucite layers over the whole measurement area reveal no significant changes due to the lucite layers added. The average coefficient of variation for the three measurements at each block thickness is 0.6% for the Hologic QDR-1000, 1.3% for the Lunar DPX and 3.0% for the Lunar DP3. Taking into account the difference in evaluation area (7 cm^2 for DPX versus 9 cm^2 for QDR-1000) and assuming the pixel noise to be caused solely by photon noise, the Hologic QDR-1000 appears to detect 3.8-times more photons than the Lunar DPX. This does not mean that the radiation dose to the patient is higher by the same factor in the Hologic scanner, because the differences in dual energy beams and detection systems play a major role in dose efficiency. This analysis suggests, however, that in cases of obese patients with large bones the Hologic QDR-1000 will be bet-

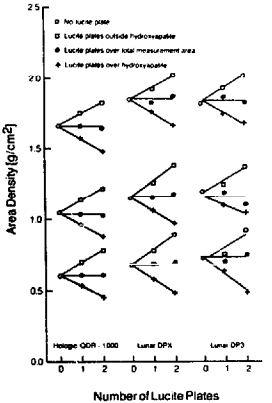

Fig. 2. Measured area density of hydroxyapatite blocks for the three tested scanners. For the points representing 0, 1 and 2 lucite plates over the whole measurement area, the horizontal line indicates the average value for these points. The sloped lines are fitted to three points: to the points representing one and two lucite layers in each of the other configurations and a point representing the previously obtained average assumed to pertain to 0 layers of lucite plates.

ter suited to provide data that do not suffer from photon starvation.

Lucite plates over only the hydroxyapatite blocks reduce the measured density values, whereas lucite blocks only beside the hydroxyapatite blocks increase them. In order to quantify the influence of the lucite plates, we calculated the slopes of sets of three points as indicated in Fig. 2. For the density without lucite plates, we averaged the three values of 0, 1 and 2 plates over the whole measurement area, in order to obtain better precision for the baseline point. This was acceptable as the lucite layers over the whole measurement area did not invoke systematic changes in any of the scanners. No systematic influence of hydroxyapatite thickness on slopes could be observed for any set of measurements. Similarly, there was no difference

Table 2

Results of lucite measurements

Scanner	Change of measured area density (g/cm² per lucite plate)		
	Lucite over center	Lucite over sides	Average magnitude
Lunar DP3	−0.093 ± 0.038	0.091 ± 0.026	0.092 ± 0.031
Lunar DPX	−0.096 ± 0.012	0.097 ± 0.015	0.096 ± 0.013
Hologic QDR-1000	−0.082 ± 0.008	0.085 ± 0.005	0.083 ± 0.006

Influence of lucite plates on measured area density of hydroxyapatite blocks.

due to the various water levels. The average slopes for all block sizes and water levels are listed in Table 2. Analysis by t-test reveals no significant differences between the magnitude of the positive and negative slopes ($P > 0.3$) for any scanner. Therefore, the average magnitude of the slopes indicating the change of measured hydroxyapatite density per lucite plate is also given. Analysis of the magnitude of change by t-test indicates a significant ($P < 0.05$) difference between Lunar DPX and Hologic QDR-1000. The larger error associated with the Lunar DP3 does not provide a significant difference between this scanner and any of the others.

Conclusions

The experiments performed allow for the following statements applicable to all three scanners investigated.

(i) Complete layers of lucite over the total measurement area have no influence on the measured area density of the hydroxyapatite blocks. This confirms that all scanners are measuring correctly the soft-tissue composition as a ratio of attenuation coefficients between the high- and low-energy beam. Consequently, variable fat content in the soft-tissue compartment of a patient will not influence the measured bone values as long as the average soft-tissue/fat ratio above, within and below the bone is the same as outside the bone where the attenuation ratio for soft tissue is measured.

(ii) Layers of lucite over the hydroxyapatite blocks, but not over the remaining measurement area, lower the measured area density of these blocks. The lucite layers over the hydroxyapatite blocks not only simulate inhomogeneous fat distribution in soft tissue, but also fat increases in bone marrow. Wherever a higher proportion of fat is encountered in soft tissue and marrow along the beam path through the bone, as compared to the beam path outside the bone, bone will be quantified too low.

(iii) Layers of lucite over the area on both sides of the hydroxyapatite blocks, but not over the blocks, increase the measured area density of the hydroxyapatite blocks. This set-up simulates inhomogeneous soft-tissue composition by assuming a higher fat proportion in the soft tissue outside the bone than in the soft tissue encountered along the beam path through bone. As the attenuation ratio between high- and low-energy beam is defined outside the bone, but assumed to be the same for the soft tissue above, below and within the bone, the error of the soft-tissue parameter yields a bone density which is too high. This error influences the measured bone density in the opposite direction compared to the error described under (ii).

(iv) In a first approximation, the error induced by lucite plates is proportional to the number of plates added. This applies for both experiments (ii) and (iii). The theoretical difference in linearity between one and two plates added is about 4% (cf. Appendix); however, the precision of the measurements does not allow this error to be noticed.

(v) The absolute amount of increase in area density of the hydroxyapatite blocks is the same as the amount of decrease for an individual scanner. Differences be-

Table 3
Fat influence on bone

Scanner	Change of bone density (g/cm²)	
	Per 1 cm fat	Per 10% fat
Lunar DP3	0.049	0.011
Lunar DPX	0.051	0.012
Hologic QDR-1000	0.044	0.010

Change of measured area density of bone due to the replacement of 1 cm muscle or marrow tissue with fat. Also, the changes are given based on a 10% change of fat content in the marrow of a medium-sized vertebra.

tween scanners are small and may be without practical consequences in cross-sectional studies.

(vi) The increase or decrease in area density of the hydroxyapatite blocks is independent of the thickness of the blocks. Analysis of error propagation (cf. Appendix) confirms that the measured changes in bone density due to soft-tissue composition are independent from the level of bone density. As bone-density results are often quoted with a percentage error, one has to keep in mind that the measurement of a person with low bone density will be subject to a larger relative uncertainty than a measurement of a person with high bone density.

The water-lucite combination does not exactly reflect the muscle-fat situation. It is possible, however, to translate the data obtained with the lucite experiments into fat-equivalent values (cf. Appendix). As the influence of fat is not dependent on the amount of bone present, the expected change in measured bone density is best expressed as change per cm fat layer. If we assume that all errors are due to fat differences in marrow and not due to inhomogeneities of soft-tissue composition, we also can calculate the influence of the relative fat content on a medium sized vertebra with an average path length through marrow of 25 mm (Table 3).

Discussion

Some published theoretical calculations of fat influence on measured area density of bone [5] are based on fat changes of 1 g/cm². With a fat density of 0.916 g/cm³, Sorenson's numbers can be converted to fat changes of 1 cm thickness. For the three scanners in the order listed in Table 3, the fat effect is 0.055, 0.050 and 0.061 g/cm² bone density per 1 cm fat replacement. Although these numbers are somewhat higher than the results from our experiments, the differences are small. Measurements with a commercial phantom that allows the simulation of various marrow-fat compositions yielded 1.2% change in bone density with a 10% change in fat content of the marrow [6]. Relative to a bone density of 1.0 g/cm², this change is equivalent to 0.012 g/cm². Although this is in good agreement with our own results, an exact comparison is not possible without knowledge of the phantom dimensions used in that study.

The influence of unknown fat content in bone marrow as measured in our experiments appears to be low. However, they are by a factor of 3–4 higher than the 0.3% estimated in some earlier publication [7]. Due to the larger pathway through soft tissue as compared with bone marrow, the unknown proportion of fat in the soft tissue along the beam path through the bone may influence the accuracy of the results to a larger extent than the uncertainty of the fat content of bone marrow. An investigation by Farrell and Webber [8] revealed that 25% of the investigated patients had a fat inhomogeneity of more than 8.2%, i.e., there was 8.2% more fat beside the vertebrae than above and below them. For an average sized patient of 20 cm thickness, this results in 1.6 cm fat which in turn will underestimate the area bone-density by 7–8% depending on scanner model used.

Acknowledgements

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Appendix

Selection of materials for soft tissue and fat

The critical parameter for eliminating the contribution of soft tissue in dual-energy absorptiometry of bone is the ratio of the mass attenuation coefficients at the two photon energies. The selection of materials most closely reflecting soft tissue or fat

Table A1

Mass attenuation coefficients (μ)

Compound	μ (cm^2/g)			
	40 keV	50 keV	80 keV	100 keV
Fat	0.2353	0.2102	0.1794	0.1684
Lucite	0.2310	0.2055	0.1748	0.1639
Polystyrene	0.2151	0.1970	0.1722	0.1623
Muscle	0.2635	0.2240	0.1819	0.1692
Water	0.2629	0.2245	0.1833	0.1706

The mass attenuation coefficients are listed for various compounds at energies of interest for dual-energy absorptiometry[9].

must be based on their similarity in the ratio of mass attenuation coefficients. Table A1 lists the mass attenuation coefficients of fat and muscle as well as some possible phantom materials at various energies. The three scanners to be used for the experiments employ different photon energies. ^{153}Gd (Lunar DP3) can be approximated with two energies at 42 and 100 keV [5]; these energies are corrected for self-attenuation in the source pellet and filtration by the aluminum exit-window. The X-ray sources of the two other scanners work on different principles: k-edge filtration for Lunar DPX and dual-energy pulsing for Hologic QDR-1000. In addition, the spectrum of the X-ray tubes is more subject to beam-hardening effects and consequently dependent on object thickness. For our purpose of defining the ratio of attenuation coefficients, we assumed the Lunar DPX scanner to work with 40 and 70 keV, the Hologic QDR-1000 scanner with 50 and 100 keV [5]. A list of several ratios of mass attenuation coefficients is given in Table A2 for the materials listed in Table A1. It is apparent that water is the best substitute for muscle. Polystyrene and lucite are candidates for fat, lucite being somewhat closer to fat than polystyrene. With water and lucite chosen to mimic soft tissue and fat, the difference in the ratios of the mass attenuation coefficients is 20% smaller than desired. However, it is easy to take this into account when the results of these experiments are interpreted as ac-

Table A2

Ratio (r) of mass attenuation coefficients

Compound	r			
	40/80 keV	40/100 keV	50/80 keV	50/100 keV
Fat	1.312	1.397	1.172	1.248
Lucite	1.322	1.409	1.176	1.254
Polystyrene	1.249	1.325	1.144	1.214
Muscle	1.449	1.557	1.231	1.324
Water	1.434	1.541	1.225	1.316

The ratio r of mass attenuation coefficients is given at various pairs of energies for the compounds listed in Table A1.

tual changes in the proportion of soft tissue and fat, because the effects turned out to be approximately linear with respect to the thickness of the lucite layer for the range investigated.

Influence of soft-tissue error on bone parameter

Inhomogeneous fat distribution over the measurement cross-section results in errors concerning the estimate of soft-tissue attenuation along the beam path through bone. Considering the basic mathematical equation for area density of bone at a certain measurement location [1],

$$m_b = \frac{(R \ln I' - \ln I) - (R \ln I_0' - \ln I_0)}{\mu_b - \mu_b' R} \qquad R = \frac{\mu_s}{\mu_s'}$$

(m_b, bone density (g/cm^2); μ_s, μ_s', mass attenuation coefficient for soft tissue at the two photon energies; μ_b, μ_b', mass attenuation coefficient for bone at the two photon energies; I, I', attenuated photon count rate at the two photon energies; I_0, I_0', unattenuated photon count rate at the two photon energies)

the influence of an error in the attenuation ratio of soft tissue on bone can be estimated by taking the derivative

$$\frac{\partial m_b}{\partial R} = \frac{(\ln I' - \ln I_0')(\mu_b - \mu_b' R) + [(R \ln I' - \ln I) - (R \ln I_0' - \ln I_0)]\mu_b'}{(\mu_b - \mu_b' R)^2}$$

This can be simplified to:

$$\frac{\partial m_b}{\partial R} = \frac{\ln I' - \ln I_0'}{\mu_b - \mu_b' R} + \frac{m_b \mu_b'}{\mu_b - \mu_b' R}$$

With the substitution:

$$\ln I' - \ln I_0' = \ln \frac{I'}{I_0'} = -\mu_s' m_s - \mu_b' m_b \qquad (m_s, \text{soft-tissue density (g/cm}^2))$$

We obtain:

$$\frac{\partial m_b}{\partial R} = -\frac{\mu_s' m_s}{\mu_b - \mu_b' R}$$

The derivative is not a constant, but dependent on the soft-tissue composition, because this influences μ_s', m_s (through ϱ; $m_s = \varrho_s d_s$ (with ϱ_s = physical density of soft tissue; d_s = attenuation path-length of soft tissue)) and R. The extreme values can be obtained by assuming soft tissue to be all fat or all muscle. With values for soft tissue from Tables A1 and A2 and for bone from the same reference [9], the numerical evaluation of the derivative for an energy pair of 40 and 100 keV yields −12.06

Table A3
Proportion of muscle and fat tissue

Scanner	Energies	Water			Lucite			Fat equiv. per lucite plate (mm)
		r	Muscle (%)	Fat (%)	r	Muscle (%)	Fat (%)	
Lunar DP3	40/100	1.541	88.5	11.5	1.409	6.5	93.5	18.86
Lunar DPX	40/70	1.335	88.9	11.1	1.248	6.3	93.7	19.00
Hologic QDR-1000	50/100	1.316	88.3	11.7	1.254	6.8	93.2	18.75

Proportion of muscle and fat tissue necessary to produce the same attenuation ratio r as water and lucite, respectively. The 70 keV attenuation coefficients were approximated by linear interpolation from the values given in Table 1. The physical density assumed for these calculations were 1.030 g/cm^3 for muscle and 0.916 g/cm^3 for fat[9]. Also given is the fat-equivalent thickness to produce the same effect as 1 lucite plate of 23 mm thickness.

for 100% fat and −14.83 for 100% muscle. The difference between the two results is about 20%. A similar difference would be expected in the slope of bone density versus number of lucite plates if the experiments covered the whole range of extremes. In our case, we added up to two lucite layers, replacing only 8–20% of water thickness per lucite layer, depending on water level used. This should influence the slopes between 0–1 and 1–2 lucite plates by less than 4% (20%·20%) in the case of 50 mm water above the epoxy block. The uncertainty of the individual measurement points did not allow differences this small to be noticed.

Translation of lucite effects to fat-equivalent values

The attenuation ratio for lucite is somewhat higher than that for fat, the attenuation ratio for water is slightly lower than that for muscle. Consequently, the effect simulated with the water-lucite combination is smaller than that to be expected from a muscle-fat combination of same thickness. In order to calculate the thickness of fat necessary to achieve the same effect as with one lucite plate, one can mix fat and muscle at a proportion to produce the same attenuation ratio as lucite. In a similar way, one can calculate the proportion of muscle and fat to produce the attenuation ratio for water. By replacing the water-equivalent muscle-fat mixture with the lucite equivalent fat-muscle mixture, we can compute the actual thickness of fat added to produce the effect of 1 lucite plate. The results are presented for each scanner separately in Table A3.