A few additional exercises with a pocket calculator will demonstrate that the discrepancy for symmetric count distributions is not large except at relatively low count values. However, the situation becomes even worse at low count values, because the Poisson distribution is not symmetric. Specifically, the probability of obtaining a result smaller than the mean is greater than the probability of obtaining a result larger than the mean. This further skews the mean of the logarithm of observed counts, <ln(N)>, to smaller values. Figure 1 demonstrates this for a Poisson distribution having a mean of 8.

Figure 2 shows the discrepancy between the mean of the logarithm of observed counts versus the logarithm of the mean for Poisson distributions having different mean values, m. To compute the values for Figure 2, I assumed that observations with N = 0 would be set to N = 1, since ln(0) is undefined. Errors are obtained even for m = 100, but the discrepancy becomes large for m < 20.

Note also that all discrepancies, except for m = 1 (not shown), are negative. In Equations 1–3, the smallest count values generally are those for the low-energy bone counts, N_{lb}. Although all of the count values are affected by this statistical artifact, and to some extent cause offsetting errors in Equations 1–3, the largest effect occurs for low-channel bone counts. This would cause an *increase* in calculated BMD.

Figure 3 shows the effect of this statistical artifact versus low-channel baseline counts per pixel for BMD = 1 g/cm^2 in 25-cm muscle, including the effect of the statistical artifact on all of the count measurements in Equations 1–3. The computations for this figure used the following input data derived from the simulation model described in Ref. *2*: $N_{hs}/N_{ls} = 6$, $u_{lb} = 0.858$ cm^2/g, $u_{hb} = 0.201$ cm^2/g, R = 1.56, CF = 1.84.

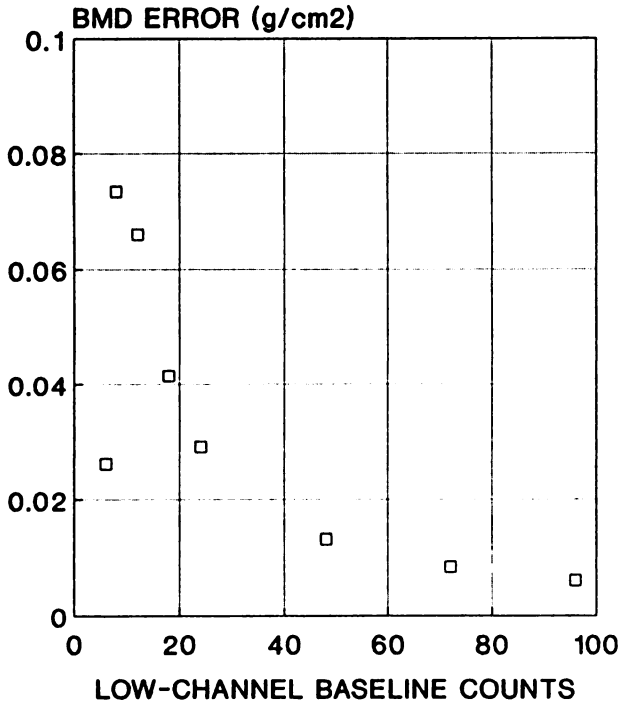

FIGURE 3
Error in computed BMD vs. low-channel soft-tissue baseline counts, calculated from simulation data (see text).

Figure 3 suggests that a marked increase in calculated BMD should be observed when low-channel baseline counts become smaller than ~50 counts per pixel. The effect actually observed may vary on different systems, depending on corrections that may be employed at low count rates. It also will vary with patient thickness and BMD, which affect the actual count values in Equations 1–3. An "exact" correction for the statistical artifact could be made using the data in Figure 2 to correct individual count data. Without such corrections, it would be advisable to avoid DPA measurements with counts per pixel less than ~50.

Note that the effects described here depend on counts per pixel, not source activity or attenuation. Furthermore, they occur in any measurement that includes logarithmic averaging of low-count data.

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James A. Sorenson
University of Wisconsin
Madison, Wisconsin

Dual-Photon Absorptiometry: Depleted Sources Inappropriate in Obese Patients with Narrow Collimation

TO THE EDITOR: DaCosta et al. (*1*) recently demonstrated differences of bone mineral density (BMD) results associated with marked differences of source strength and attenuation thickness using gadolinium-153 (^{153}Gd) dual-photon absorptiometry. A small increase of BMD at low source activities (<0.3 Ci) does occur in thick patients (>20 cm) using older spine software on the Lunar DP3 scanner. The well-documented increase of ~0.02 to 0.04 g/cm^2 amounts to a 2%–4% increase at the typical BMD for elderly women of 1.0 g/cm^2 (*2–6*). This shift of spine results, due to a software bug, was corrected in later software versions (*4, 5*); it is not evident in femur scans. The unusual findings of DaCosta et al. using recent software may be due to their use of (a) 8-mm detector collimation, (b) a very depleted source, (c) two different sources for high- and low-activity determinations, and (d) a large thickness of a nonphysiologic attenuator.

Using 8-mm collimation rather than the standard 13-mm collimation reduces the count flux by over twofold. When using 8-mm collimation, sources can be used only to ~0.5 Ci, so the 0.3 Ci source used by DaCosta et al. was 6 mo beyond its useful life. Moreover, two different sources were used in testing. Nilas et al. (*6*) have shown that shifts of several percent can accompany use of different depleted ^{153}Gd sources, possibly as a consequence of contamination.

DaCosta et al. tested at a water thickness of 24.5 cm, which is equivalent to the attenuation seen in a patient 26-cm thick (15 cm of lean tissue and 11 cm of fat, which has 20% less linear attenuation than water). In our examination of several

hundred women, we found that an effective thickness of 24.5 cm occurs at a body weight of 90 kg. The regression was: THICKNESS (cm) = 0.182 + WEIGHT + 8 (r = 0.85; s.e.e. = 1.5 cm). The effective attenuation needed to achieve the low count rates seen by DaCosta et al. occurs at a water thickness of 28 cm with 13-mm collimation that is standard for the DP3 scanner. This would be seen in a patient weighing ~110 kg with an anatomical thickness of 31 cm (15 cm lean + 16 cm fat). It is inappropriate to use depleted sources on such obese subjects. On the other hand, studies by Dawson-Hughes et al. (*5*) have shown that accurate but not precise results can be obtained even with depleted sources on subjects of 28-cm thickness. Phantoms designed to test thickness response of DPA or DEXA scanners *cannot* consist of either water alone or plastics. Dual-energy systems begin to vary in response to soft-tissue composition at thicknesses >20 cm. Typically, scanners are calibrated at normal composition (25% fat) from 15–20 cm and produce accurate data at 25 cm only if the soft tissue is ~40% fat (15 cm of water + 10 cm of oil).

DaCosta et al. (*1*) imply that source activity is critical for precise determinations using DPA. The precision of DPA on the spine in many studies using the Lunar DP3 averaged 1.8% even when older software was used (*7*). The precision error reported by Dawson-Hughes et al. (*5*) was within 2%, and that reported by the researchers at Mt. Sinai was 2.5% (*8*). Correction for the small influence of source activity on typical patient results under usual conditions could reduce the precision error slightly. However, the major uncertainties in spinal determinations are (a) confusion of the L1-L3 sequence with the L2-L4 and (b) misplacement of edges and baselines. In a reanalysis of thousands of spine scans from many institutions, the above operator errors were several times greater than the uncertainty associated with source activity effects.

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Richard B. Mazess
Lunar Radiation Corp.
Madison, Wisconsin

REPLY TO DR. SORENSON: Dr. Sorenson's proposal to account for the increase in bone mineral density observed at low counting rates is appreciated. Although the 'statistical artifact' described is a potential explanation for the error observed by us, we are unable to evaluate the hypothesis further, since the actual count rate information is unavailable to DP3 users. Lunar DP3 software alters the raw count information during acquisition and stores calculated bone mineral values that cannot be converted back to raw data (i.e., high- and low-energy photon counts measured through bone and soft tissue). The software algorithms used are considered proprietary information and have not been available to users for review. This further highlights the problem of having to rely on software in which raw data are not retained and in which the basis of calculations is concealed from the users.

Dr. Sorenson's proposed correction for the statistical artifact on individual count data (Fig. 2) cannot be applied. We hope that the industry will evolve toward a standard which makes documentation of the algorithms, as well as raw count data, available to users.

Dr. Sorenson describes effects that he attributes to counts per pixel, and not source activity or attenuation; however, counts per pixel are indeed a function of source activity and attenuation.

Maria DaCosta
Stanley J. Goldsmith
Diane Meier
Marjorie Luckey
Mt. Sinai Medical Center
New York, New York

REPLY TO DR. MAZESS: We are pleased that Dr. Mazess, the manufacturer of the device used in our study, essentially acknowledges the validity of our observations (*1*). He incorrectly, however, implies that our findings are due to an 'unusual' scanning configuration.

Dr. Mazess suggests that our study used outmoded acquisition parameters. In fact, the manufacturer's original technical guidelines for collimation and scan speed specified the 'high resolution' parameters used in our study (8 mm and 2.5 mm/sec, respectively). In a February 11, 1985 correspondence to customers, Lunar announced an *optional* configuration of 13 mm collimation and 5 mm/sec speed as a mechanism "to allow shorter scan times and longer source life." The original configuration, which has been referred to as the 'high resolution scan,' or 'slow scan,' was still recommended "to achieve the best precision with older sources or with low bone values." Since our primary concern in the conduct of a longitudinal research study was precision (not economic considerations) and since we recognized that we might indeed be studying patients with low bone mineral values, we elected the more rigorous methodology.