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Standardization of Computed Tomography Images by Means of a Material-Selective Beam Hardening Correction

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Abstract: Polychromaticity of the X-rays used in computed tomography (CT) has made it difficult to establish an absolute scale for CT values and has made quantitative comparisons between patients unreliable. The spectral shift of the X-rays depends on the material distribution within the structure measured and is significant if substantial amounts of bone, fat, or injected contrast material are present. A material-selective beam hardening correction procedure has been developed that allows the reconstruction of good approximations of linear attenuation coefficients with respect to a reference energy. With the aid of mathematical simulations and measurements on a physical phantom, the feasibility of the procedure and its insensitivity with regard to energy settings and other machine parameters are documented. **Index Terms:** Computed tomography—Beam hardening correction—Absolute CT scale—Normalized CT numbers—Spectral shift artifact correction.

Beam hardening artifacts have been recognized as a serious problem in computed tomography (CT) (1–4). While a linearization of the raw data can eliminate some of these artifacts (5–8), effects due to bone, muscle, and fat cannot be handled simultaneously by linearization. It is therefore necessary to apply a correction for the polychromaticity of X-rays that is based on the material distribution within the measured body cross section. In an earlier work (9), we proposed such a procedure for the elimination of the distinctive beam hardening artifacts in the presence of high concentrations of contrast material. Other postprocessing correction procedures are reported (10–12) with special emphasis on the suppression of bone induced artifacts.

In this paper we present an iterative correction procedure for beam hardening artifacts that is based on the distribution of bone, muscle, and fat. The corrected reconstructions are standardized to a fixed photon energy and are independent from kVp settings, X-ray filters, water bags, or detector type used. Since most of the existing CT systems have effective X-ray energies close to 70 keV, this energy was chosen as a reference or standard. This means that the corrected and standardized CT images are identical with those made with a hypothetical monoenergetic photon source of 70 keV.

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METHOD

The transmission measurements in X-ray CT depend on the spectrum of the photon source used and on the atomic composition of the absorbing medium. The number of photons $I(p,\theta)$ that are transmitted after a path p with projection angle θ may be described by the equation

$$I(p,\theta) = \int I_0(E) \exp\left[-\sum_k \mu_k(E)\, L_k(p,\theta)\right] dE \qquad (1)$$

where $I_0(E)$ is the energy spectrum of the incident beam, $\mu_k(E)$ is the linear attenuation coefficient of the material k at energy E, and $L_k(p,\theta)$ is the thickness of the material k along this ray.

The projections

$$M(p,\theta) = \ln\{I_0/I(p,\theta)\}$$
$$\text{with} \qquad I_0 = \int I_0(E)dE \qquad (2)$$

may be used as a basis for the reconstruction of a CT image, but in general there will be no linear relationship between the CT values in the reconstruction and the linear attenuation coefficients in the measured body cross section.

In the monoenergetic case, Eqs. 1 and 2 can simply be written as

$$M_{E_0}(p,\theta) = \ln\left\{I_0(E_0)/I_{E_0}(p,\theta)\right\} = \sum_k \mu_k(E_0)L_k(p,\theta) \quad (3)$$

where, in our case, E_0 is taken to be 70 keV.

Let us assume for a moment that the X-ray spectrum and the material distribution are already known. Then monoenergetic and polyenergetic projections could be calculated according to Eqs. 2 and 3, and hence also

$$Q(p,\theta) = M_{E_0}(p,\theta)/M(p,\theta) \quad (4)$$

We call $Q(p,\theta)$ beam hardening correction factors. If the measured projections are first multiplied with the pertinent $Q(p,\theta)$ and then used for the reconstruction of a CT image, attenuation coefficients will result as if we had measured with a monoenergetic X-ray beam.

For the calculations and measurements described in the *Results* section, the continuous spectrum $I_0(E)$ was replaced by a set of discrete energies. We found that three effective energies are appropriate if the respective intensities I_{0i} are carefully selected. These intensities are given by the spectrum of the X-ray tube, the filters used, the detector response, and the effective energies selected. For reliable results, an experimental determination of I_{0i} is therefore advised. To this end, two wedges of materials with precisely known linear attenuation coefficients (we used H_2O and CCl_4) and geometries were scanned. By comparing pairs of measurements with the same detector output we obtain

$$\sum_{i=1}^{3} I_{0i}\exp(-\mu_{i1}L_1) = \sum_{i=1}^{3} I_{0i}\exp(-\mu_{i2}L_2) \quad (5)$$

The linear attenuation coefficients μ and the wedge thicknesses L are known, and the spectral intensities I_{0i} may be determined with the help of a least-squares procedure.

The material distribution $L_k(p,\theta)$ needed to calculate the beam hardening correction factors may be estimated from a reconstruction that is based on linearized data. Each pixel in this reconstruction is assumed to contain bone, a mixture of bone and fat, muscle, a mixture of muscle and fat, fat, or a mixture of muscle and air. The linear attenuation coefficients (or CT values) of the mixtures are considered to be linear compositions of those of the "pure" materials. One might argue that it would be more appropriate to use mixtures of all three materials. Yet with the help of numerical simulations we found that the beam hardening correction factors $Q(p,\theta)$ are only slightly altered (on the order of 0.1%) if such improved tissue compositions are used. On the other hand, the reconstruction from linearized data may have beam hardening artifacts that influence the estimated material distribution significantly. In this case, the reconstruction made after each projection value was modified with its $Q(p,\theta)$ may be used in a second iteration for the calculation of an improved estimate of the material distribution. In all practical cases studied, only two iterations were needed to achieve an accuracy compatible with that of the values reported in the literature used for the attenuation coefficients of the material components.

RESULTS

The feasibility of the proposed procedure was first studied with numerically simulated projection data. Four cases are compared: monoenergetic data, polyenergetic data (no correction at all in the

1a,b

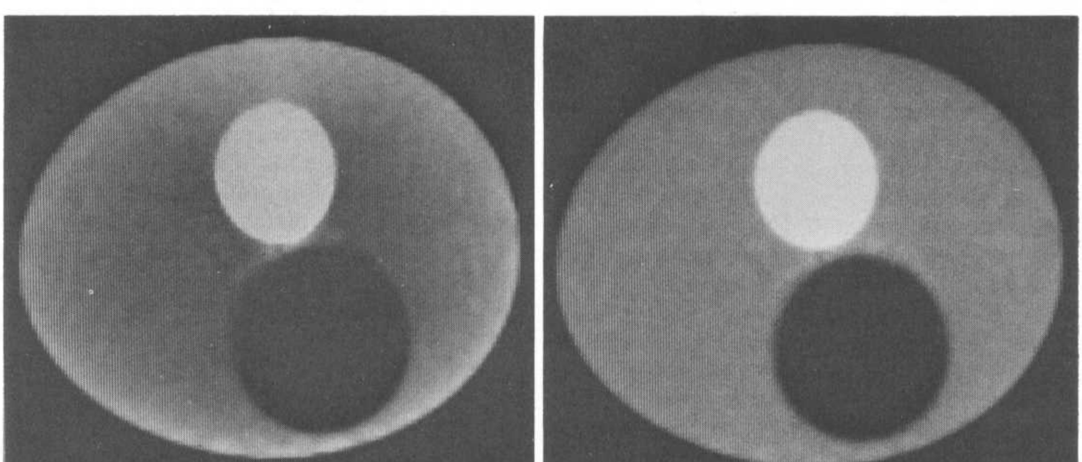

FIG. 1. Beam hardening artifact in the presence of extended bone and fat masses. In this numerical phantom study, a body cross section was simulated by a water ellipse with a larger diameter of 35 cm, an elliptical vertebra (30% bone mineral), and a circular fat mass. The beam hardening artifact (**a**) is completely eliminated after material-selective beam hardening correction of the projection data (**b**).

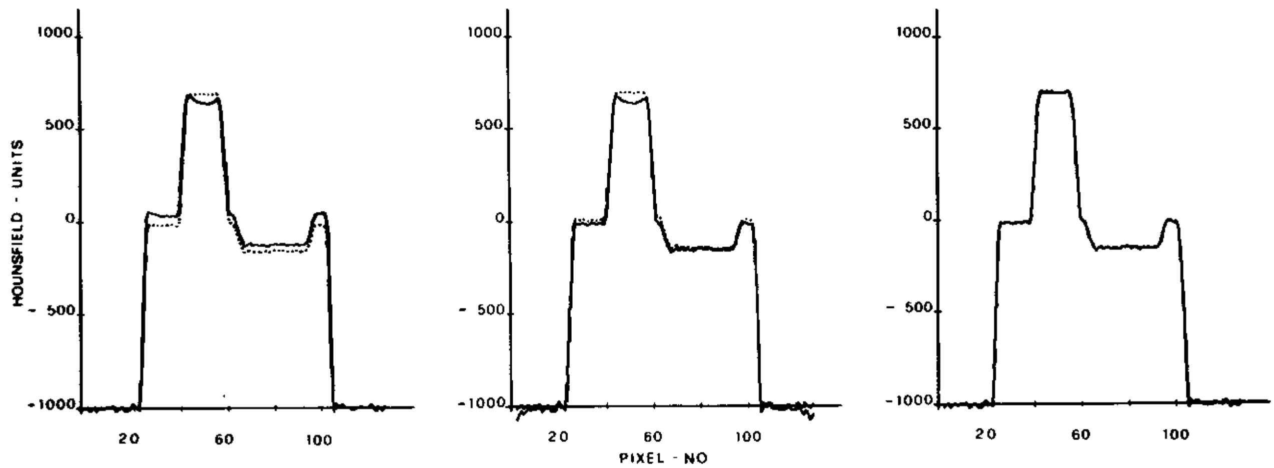

FIG. 2. Deviations of CT values reconstructed on the basis of polyenergetic data from those reconstructed on the basis of monoenergetic data. The CT values plotted are those of the central column of the reconstruction matrix in Fig. 1. The uncorrected polyenergetic reconstruction, the polyenergetic reconstruction after linearization, and the polyenergetic reconstruction after material-selective beam hardening correction are given from left to right. For comparison purposes, the profile of CT values reconstructed from monoenergetic data (dashed line) is also given in each case. The linearization of the projection data leads to reasonably good corrections of the soft tissue values but may even worsen the CT values in the bone region. The material-selective beam hardening correction allows the reconstruction of virtually monoenergetic pictures.

reconstruction), polyenergetic data with linearization of the simulated detector output, and polyenergetic data with the proposed material-selective correction procedure. Then measurements of a physical model were made with a translational-rotational type whole body scanner (parallel beam).

The mathematical model used represents a water-containing elliptical body whose large and small diameters are 35 and 25 cm, respectively. It contains a circular region of fat and a simulated vertebra consisting of a material with the properties of spongy bone (fat with 30% bone mineral). The monoenergetic data were created assuming a 70 keV photon beam. For the polyenergetic case, a 120 kVp X-ray setting and a 0.2 mm copper filter were assumed. Figure 1 illustrates typical beam hardening artifacts in the reconstructed cross sections of the described mathematical phantom. The magnitude of these errors can best be appreciated by plotting the CT values along the middle column of the reconstruction matrix (Fig. 2). The deviation of the CT values for water reconstructed on the basis of the polyenergetic data from those reconstructed on the basis of the monoenergetic data can be as much as 10% (Fig. 1a). Linearization reduces this error to 3% but does not lead to improvement in the case of the simulated vertebra. After material-selective beam hardening correction, the CT image is practically indistinguishable from that reconstructed from monoenergetic data (Fig. 1b).

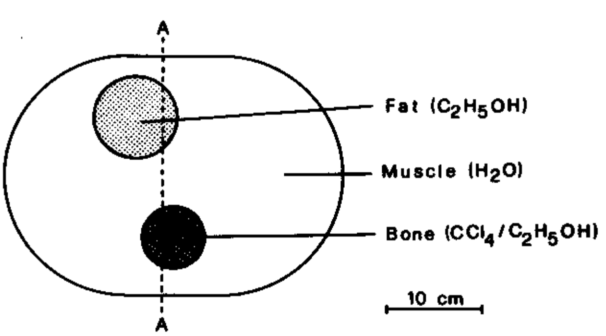

FIG. 3. Sketch of physical phantom. The liquids simulating fat, muscle, and bone are filled in Lucite containers with walls of 1 mm thickness. This model was measured with a translational-rotational type scanner. Profiles of the reconstructed CT values along line A-A are shown in Fig. 4.

In the physical model shown in Fig. 3, muscle is again simulated by water, whereas fat is replaced by C_2H_5OH and bone by a mixture of CCl_4 and C_2H_5OH. Figure 4 plots the reconstructed CT values along line A-A for successive corrections. The tabulated values (13) of the linear attenuation coefficients at 70 keV are indicated by dashed lines. It is obvious that not only the image quality is improved, as can be seen from the flat response within the homogeneous media, but that also the CT values gradually approach those given in the literature. With no correction at all, the CT values for water deviate as much as +15%. After linearization, this error is reduced to −5%. With the material-selective beam hardening correction, the residual error is approximately −1.5%. A similar drastic error reduction is achieved in the "fat" and "bone" regions. Whereas the deviations of the "fat" values are reduced from +9.4% to +0.1%, those for "bone" improve from −13.5% to −1.4%. The physical model was also measured with X-ray settings at 100, 120, and 140 kVp. Figure 5 depicts the CT values of the central column for all three energies before and after the material-selective

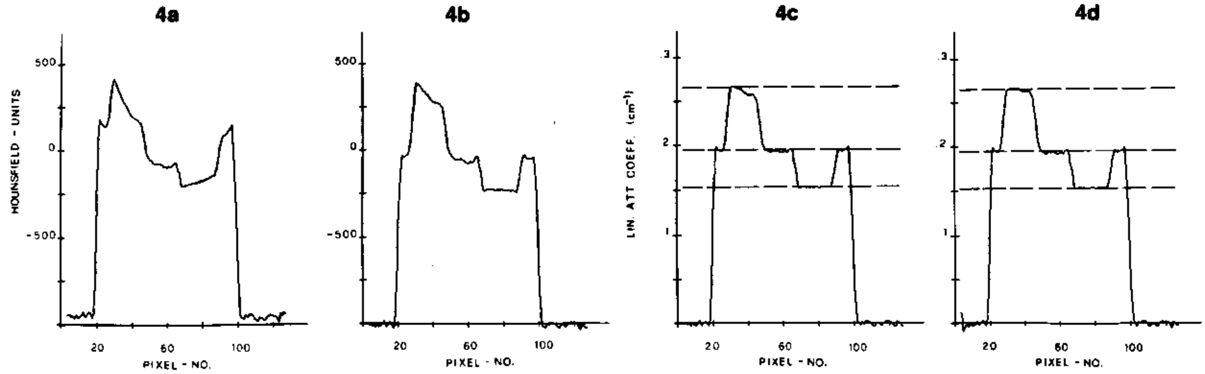

FIG. 4. Reconstructed CT values from measurements at 100 kVp along Line A-A of Fig. 3 for successive corrections. **a:** Uncorrected. **b:** After linearization of the measured projections. **c:** After first iteration of the material-selective beam hardening correction. **d:** After second iteration. Linearization (b) results in reliable muscle values, but the correction is too high for fat and too low for bone. After the material-selective beam hardening correction, the CT numbers may be changed to linear attenuation coefficients (cm^{-1}). The dashed lines indicate the tabulated values at 70 keV for the materials used.

beam hardening correction. According to Table 1, the maximum deviation of the reconstructed linear attenuation coefficients does not exceed 1.6%. Since this is true for all three energies, it appears that the proposed correcting procedure is insensitive to machine parameters.

The material-selective beam hardening correction procedure is now in use for bone mineralization studies and radiation treatment planning. The linear attenuation coefficients applied in these cases of pure bone, muscle, and fat are taken to be 0.480, 0.205, and 0.178 cm^{-1} (14, 15), respectively, at 70 keV, corresponding to approximately 1,500, 50, and −90 Hounsfield units (1,000 scale), respectively. The mixtures of tissue materials are, of course, not uniquely defined by the CT values. However, this ambiguity appears to be irrelevant in view of the facts that the beam hardening correction factor is defined as the ratio of polyenergetic to monoenergetic projections along the same path through the

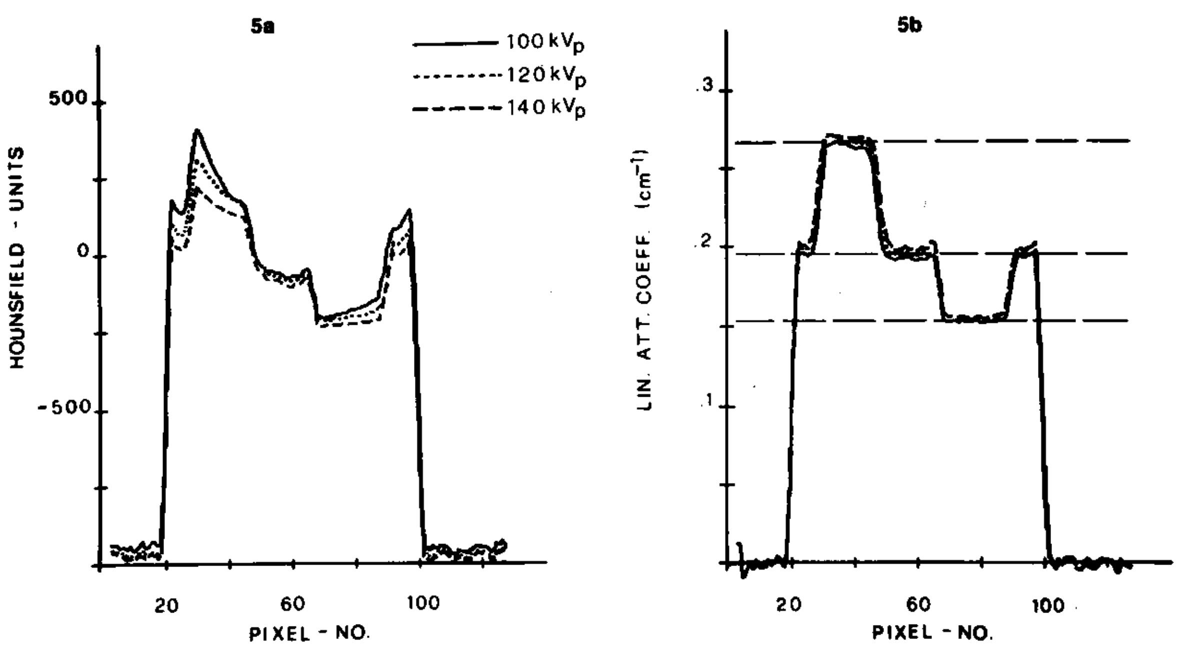

FIG. 5. Insensitivity of the material-selective beam hardening correction from machine parameters. **a:** The CT values along line A-A (Fig. 3) are given for the uncorrected reconstructions from measurements at 100 kVp, 120 kVp, and 140 kVp. **b:** The respective linear attenuation coefficients at 70 keV after two iterations are given. The dashed lines indicate the tabulated values for the linear attenuation coefficients of the materials used. Maximum deviation from these values is 1.6% for all three voltage settings.

TABLE 1. *Percentage deviation of reconstructed linear attenuation coefficients after material-selective beam hardening correction from tabulated values*

	Energy		
Material	100 kVp	120 kVp	140 kVp
CCl_4/C_2H_5OH	−1.4	−0.1	+0.5
H_2O	−1.6	−0.8	+1.1
C_2H_5OH	+0.1	+1.6	+1.3

body, and that beam hardening influences the projections only on the order of 10%.

CONCLUSIONS

The material-selective iterative procedure for beam hardening correction described here appears to provide an absolute CT scale with a linearity on the order of 1%. The procedure is entirely numerical and does not require increases in scan time and radiation dose. The only additional information needed are two wedge measurements for the determination of the effective energy spectrum and the machine parameters. Computation time is increased by a factor of 2.5 to 4 depending on the type of back projection and number of iterations used. In the example to which the correction procedure was applied the body structures were assumed to consist of mixtures of bone, muscle, fat, and air. However, there is no difficulty in adding more substances, such as iodine, if they can be identified in the CT image (9). For routine diagnostic measurements, the beam hardening correction procedures currently applied may be sufficient. The proposed procedure handles the heretofore unsolved problems that arise when CT images from different scanners are to be compared or when absolute values of the attenuation coefficients are needed for tissue identification, radiation treatment planning, or quantitative diagnostic studies.

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REFERENCES

1. Cho ZH: General views on 3-d image reconstruction and computerized transverse axial tomography. IEEE Trans Nucl Sci NS-21:44–71, 1974
2. McCullough EC, Baker HL Jr, Houser OW, Reese DF: An evaluation of the quantitative and radiation features of a scanning X-ray transverse axial tomograph: The "EMI scanner." Radiology 111:709–715, 1974
3. Shepp LA, Logan BF: The Fourier reconstruction of a head section. IEEE Trans Nucl Sci NS-21(3):21–43, 1974
4. Zatz LM, Alvarez RE: An inaccuracy in computed tomography: The energy dependence of CT values. Radiology 124:91–97, 1977
5. Brooks RA, Di Chiro G: Beam hardening in X-ray reconstructive tomography. Phys Med Biol 21:390–398, 1976
6. Macovski A, Alvarez RE, Chan LH, Stonestrom JP: Correction for spectral shift artifacts in X-ray computerized tomography. (In) Proceedings of the Conference on Image Processing, Stanford, California, MPI-5, 1975
7. Herman GT: Correction for beam hardening in computed tomography. Invest Radiol (*in press*).
8. McDavid WD, Waggener RG, Payne WH, Dennis MJ: Correction for spectral artifacts in cross-sectional reconstruction from X-rays. Med Phys 4:54–57, 1977
9. Rüegsegger PE, Ritman EL, Wood EH: Performance of an all electronic cylindrical CT scanning system for dynamic studies of the heart and lungs. (In) Proceedings of the San Diego Biomedical Symposium, 16:143–157, 1977
10. Brooks RA: Bone corrections in computer assisted tomography. (Abstr.) J Comput Assist Tomogr 1:267, 1977
11. Stonestrom P, Alvarez R: Iterative inference: A suppression technique for beam hardening effects. (Abstr.) J Comput Assist Tomogr 1:372, 1977
12. Joseph PM, Spital RD: A method for correcting bone induced artifacts in CT scanners. J Comput Assist Tomogr 2:100–108, 1978
13. Veigele WMJ: Photon cross sections from 0.1 keV to 1 MeV for elements Z = 1 to Z = 94. Atomic Data Tables 5, 1973
14. Phelps ME, Hoffman EJ, Ter Pogossian MM: Attenuation coefficients of various body tissues, fluids, and lesions at photon energies of 18 to 136 keV. Radiology 117:573–583, 1975
15. International Commission on Radiation Units and Measurements (ICRU) Report 17, 1970