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Optimal CT settings for bone evaluations

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Abstract. In CT densitometry the precision, sensitivity and accuracy of the procedure are limited by photon noise, the degree of tissue differentiation and artefacts caused by the finite size of the X-ray beam, hardening of the X-rays and scattered radiation. These artefacts are highly dependent on the CT settings, such as X-ray spectrum and effective energy, intensity of photon flux and collimation of the beam. To monitor the progression of a disease or the efficacy of a therapy, longitudinal studies are necessary; thus the radiation dose per examination should be kept at a minimum. For bone evaluation of the peripheral skeleton it is shown that there is a narrow optimal energy range of 30-40 keV where a dose below 150 μSv is sufficient to detect a change of 1% in trabecular bone density at the 0.1 confidence level. Concomitantly accuracy might be better than 1% if adequate beam collimation is applied.

1. Introduction

Compared to the use of computed tomography (CT) as an imaging technique, quantitative computed tomography (QCT) is as yet little used (Proc. 3rd Int. Workshop on Bone and Soft Tissue Densitometry Using CT 1983). Major reasons for the restraint in the application of QCT are the limited sensitivity, precision and accuracy of the analysis based on standard CT images. These limitations are dependent on the CT settings, such as the intensity of photon flux, the energy spectrum of the photons and the size of the photon beam.

The assessment of the progress of a disease or the efficacy of a therapeutic procedure requires a temporal series of high precision QCT examinations. These examinations should be made at the lowest possible radiation dose. However, the precision of a QCT evaluation is limited by the photon noise of the measured projection data which causes noise in the reconstructed CT images (Seitz *et al* 1985). Precision is dependent on radiation dose and hence dependent on the energy of the X-rays used. This energy also determines the sensitivity of the QCT parameters to changes in the evaluated tissue compositions. Thus X-ray energy has to be optimised with regard to precision and sensitivity of the procedure.

The CT image reconstruction procedure is based on the assumption that the projection data are line integrals of the linear attenuation coefficient along the X-ray beam. Ideally this requires monoenergetic X-rays and infinitesimally thin photon beams. Deviations from these requirements lead to artefacts in the reconstructed images, which are non-local and dependent on the object and scanner. A finite beam size is associated with both linear (Verly and Bracewell 1979) and non-linear (Glover and Pelc 1980, Joseph and Spital 1981) degradations. Additional degradations are associated with

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beam hardening (Stonestrom *et al* 1981) and scattered radiation (Stonestrom and Macovski 1976).

To describe gradual changes in the peripheral skeleton which might be well below 1% it has been shown (Rüegsegger *et al* 1982) that trabecular bone density is a very useful parameter. Trabecular bone is examined in the end sections of the distal radius or the distal tibia. The special purpose CT system used for these evaluations (Stebler and Rüegsegger 1983) is based on a second generation measuring device with well collimated x-ray beams that ensure a high scatter rejection. Spectral hardening is reduced by heavy filtering (98% intensity loss) of the output of the x-ray tube and by careful calibration of the projection data.

2. Method

The evaluation of optimal CT settings for bone densitometry is based on reconstructions from computer simulated transmission data of mathematically described models. These models were selected to conform in principle with *in vivo* measurements obtained in the distal forearm and the lower leg. Accordingly, they consist of two hollow cylinders of heavily absorbent material representing compact bone, filled with a less dense medium equivalent to a mixture of trabecular bone and bone marrow. The bone simulating materials are enclosed in a material simulating soft tissue (see figure 1). The parameters used for the numerical simulations are given in tables 1 and 2.

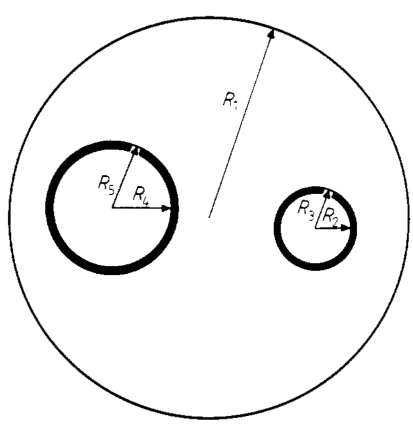

Figure 1. Mathematical models of the human forearm (radius) and lower leg (tibia). The selection of optimal CT settings is object dependent. CT bone evaluations of the peripheral skeleton are numerically simulated for cylindrically shaped trabecular bone, cortical bone and soft tissue compartments. Linear attenuation coefficients used are given in table 1. Radii used are given in table 2.

Let us assume that bone trabeculae have the same composition as compact bone. Then trabecular bone density ρ_t is given by

$$\rho_t = A\rho_c \quad (1)$$

where ρ_c is the density of compact bone ($g\,cm^{-3}$) and A is the fraction by volume of trabecular bone.

Table 1. Linear attenuation coefficients μ_s, μ_c and μ of the mathematical models for soft tissue surrounding the bones, cortical bone and the trabecular bone compartment respectively. A is the fraction by volume of trabecular bone.

Energy (keV)	μ_s (cm^{-1})	μ_c (cm^{-1})	μ (cm^{-1}) ($A=0.2$)	μ (cm^{-1}) ($A=0.05$)
25	0.523	4.472	1.188	0.572
30	0.388	2.760	0.787	0.418
35	0.319	1.886	0.582	0.338
40	0.279	1.393	0.465	0.291
45	0.253	1.092	0.393	0.262
50	0.236	0.898	0.346	0.243
55	0.223	0.767	0.314	0.229
60	0.213	0.674	0.290	0.218

Table 2. Radii used in the mathematical models of the human forearm (radius) and lower leg (tibia).

	Radius (mm)	
R	Forearm	Lower leg
R_1	30.0	50.0
R_2	5.0	6.0
R_3	6.2	8.0
R_4	7.7	15.8
R_5	8.9	17.8

The attenuation of the x-ray beam passing through the trabecular bone region is described by the linear attenuation coefficient μ of the mixture of trabecular bone and bone marrow

$$\mu = A\mu_c + (1-A)\mu_m \tag{2}$$

where μ_c and μ_m are the linear attenuation coefficients of compact bone and bone marrow respectively.

The sensitivity of detecting bone changes is dependent on the amount of bone material present in the trabecular bone region. Numerical simulations were therefore made for two extreme cases, $A = 0.2$ (healthy subject) and $A = 0.05$ (diseased subject). Most patients have a trabecular bone density between these two extremes.

The numerical simulations were made for a translate-rotate CT scanner, taking 128 views and 256 samples per view. For each sample of the projection data the integral

$$P = \frac{1}{HW} \int_H \int_W \int_L \mu(x, y, z)\, dl\, dw\, dh \tag{3}$$

was calculated along path L for a non-divergent monoenergetic x-ray beam of width W and height H (slice thickness). The mean number of transmitted photons is given by $\bar{N} = \bar{N}_0 \exp(-P)$ where $\bar{N}_0$ is the mean number of detected photons without any attenuation of the beam. Photon noise from a pseudorandom generator is added to the mean value $\bar{N}$ to get the actual number N of transmitted photons. The projection data are then calculated as $\ln(\bar{N}_0/N)$, assuming a detection system working in the single photon counting mode.

A filtered backprojection technique was used to reconstruct CT images from the projection data (Shepp and Logan 1974). The reconstruction matrix contains 256×256 pixels with a pixel size of 0.102 mm^2 for the simulated radius and 0.230 mm^2 for the simulated tibia, resulting in field sizes of 81.2 and 122.8 mm respectively. The CT images are evaluated with regard to trabecular bone changes. For that purpose we define a 'bone parameter' μ' as the average linear attenuation coefficient characterising the material in the trabecular bone region. To avoid any contribution from cortical bone a central disc limited to 50% of the total bone area is used for the evaluation.

To study the non-linear effect of the finite beam size on the bone parameter μ' equation (3) was expanded to

$$P = -\ln \frac{1}{HW} \int_H \int_W \exp\left(-\int_L \mu(x, y, z)\, \mathrm{d}l\right) \mathrm{d}w\, \mathrm{d}h \tag{4}$$

taking into account that measured projection data represent the logarithm of a spatial average of x-ray intensities.

3. Results

The differentiation of equation (2) gives

$$\frac{\mathrm{d}\mu}{\mu} = \left(1 - \frac{\mu_m}{\mu}\right) \frac{\mathrm{d}A}{A}$$

where $\mathrm{d}A/A = \mathrm{d}\rho_t/\rho_t$ from equation (1). This yields the sensitivity of μ to small changes of the trabecular bone density ρ_t

$$\frac{\mathrm{d}\mu/\mu}{\mathrm{d}\rho_t/\rho_t} = 1 - \frac{\mu_m}{\mu} \tag{5}$$

Figure 2 shows a plot of the sensitivity for the two extreme cases $A = 0.2$ and $A = 0.05$ in the energy range 25–60 keV. Sensitivity diminishes gradually as the energy of the x-rays increases or the density of trabecular bone decreases.

The precision of the bone parameter μ' is described by the ratio $\sigma/\overline{\mu'}$. The mean $\overline{\mu'}$ and the standard deviation σ are to be calculated from a set of CT images. This means that precision refers to an ensemble of reconstructions; it should not be evaluated from the fluctuation of pixel values of one CT image (Seitz *et al* 1985). Precision is dose dependent

$$\sigma^2/\overline{\mu'}^2 = K/D \tag{6}$$

(Brooks and Di Chiro 1976). The dose D is the average energy per unit mass absorbed by the irradiated tissues. Therefore, dose depends on the mass distribution of the model, the densities of the tissues and the slice thickness (set to 2 mm). It is calculated by multiplying the number of absorbed photons with their energy. (Scattering is disregarded.)

Simulations were used to calculate the factor K for the models at different x-ray energy levels. This is the basis of figure 3 where the energy and object dependence of the precision of the bone parameter is plotted for a fixed radiation dose.

Trabecular bone density might be decreased or increased by disease or therapy. Let us assume that a change $\Delta\rho_t/\rho_t$ in trabecular bone density occurred between two

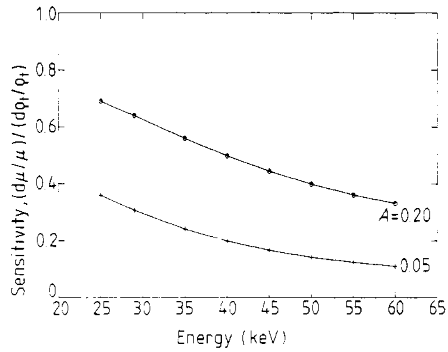

Figure 2. Sensitivity of μ to changes of the trabecular bone density ρ_t. The energy dependence of the sensitivity is given for two extreme cases: ○, $A = 0.2$ (high trabecular bone density, normal subject); and +, $A = 0.05$ (low trabecular bone density, osteoporotic patient).

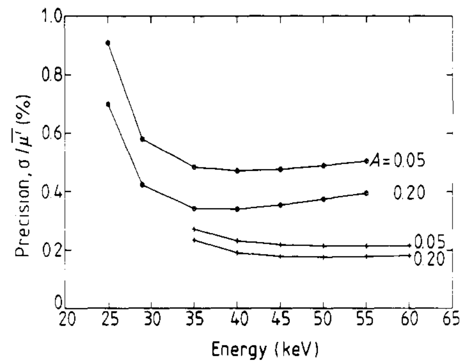

Figure 3. Precision of the bone parameter μ'. The energy dependence of the precision is given for: ○, the forearm (radius) and +, the lower leg (tibia) for high and low density. Radiation dose was 20 μSv (2 mrem).

measurements resulting in the bone parameters μ_1' and μ_2'. What radiation dose (i.e. precision σ/μ') is required for the difference $\Delta\mu' = \mu_1' - \mu_2'$ to be a significant indicator for the bone density change? The measured difference $\Delta\mu'$ is normally distributed about $\overline{\Delta\mu'}$ with standard deviation $2^{1/2}\sigma$ (μ_1' and μ_2' are statistically independent and normally distributed with standard deviation σ; the standard deviations for the two bone parameters are equal for small changes in the trabecular bone density).

The hypothesis (H_0) that μ_1' and μ_2' are the same can be correctly rejected with a probability of 90% (two-tailed test) if $\Delta\mu'$ exceeds $\Delta\mu'_{\text{limit}} = 1.645(2)^{1/2}\sigma$ (figure 4*a*). In order to allow only a 10% probability (one-tailed test) of incorrectly accepting H_0, $\overline{\Delta\mu'}$ has to be equal to $\Delta\mu'_{\text{limit}} + 1.282(2)^{1/2}\sigma$ (figure 4*b*).

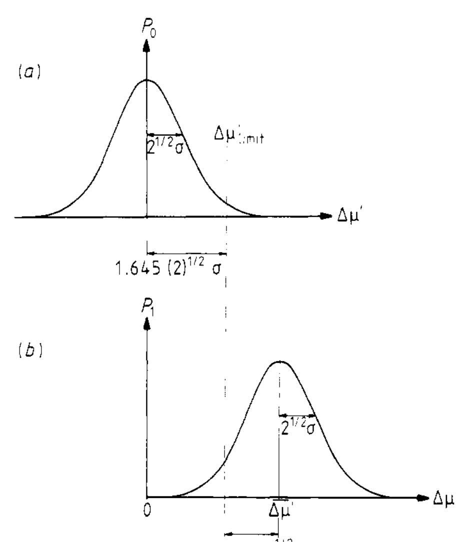

Figure 4. Frequency distribution of the difference $\Delta\mu'$ between two measurements. (*a*) Zero centred distribution P_0 for $\Delta\mu'_{\text{limit}}$; (*b*) shifted distribution P_1 for $\overline{\Delta\mu'}$.

Using equations (5) and (6) it follows

$$D = K\frac{[1.645(2)^{1/2}+1.282(2)^{1/2}]^2}{(\Delta\rho_t/\rho_t)^2[1-(\mu_m/\mu)]^2}$$

This dose D is the minimal dose per examination that is required to detect the relative change $\Delta\rho_t/\rho_t$ in the trabecular bone density from two examinations. It should be noted that the requirement for setting up an experiment and establishing σ by defining the minimally required radiation dose is different from deciding if $\Delta\mu'$ between two measurements is statistically significant. In this latter case $\Delta\mu' = 1.645(2)^{1/2}\sigma$ is sufficient to be different from $\Delta\mu' = 0$ at a 90% significance level.

In figures 5(a) and 5(b) the minimal dose is plotted for the two extreme cases $A = 0.2$ and $A = 0.05$. Note the sharp optimum in the range between 30 and 40 keV. The lower dose required for tibia examinations is partially a consequence of the larger evaluation volume available at the lower extremity.

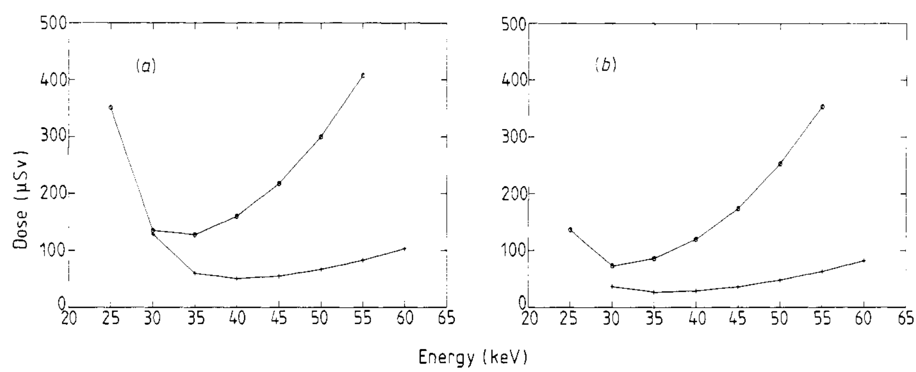

Figure 5. Radiation dose to detect trabecular bone changes. The energy dependence of radiation dose is given for: ○, the forearm (radius) and +, the lower leg (tibia). The plots are given for: (a) $A = 0.2$ (healthy subject) and a change $\Delta\rho_t/\rho_t = 1\%$ in the trabecular bone density; and (b) $A = 0.05$ (osteoporotic patient) and a change $\Delta\rho_t/\rho_t = 4\%$ in the trabecular bone density. (10 μSv = 1 mrem.)

The finite beam size leads to both linear and non-linear degradations of the projection data. The linear degradation causes a blurring of the CT image. The non-linear degradation stems from the fact that while each measured sample of the projection data represents the logarithm of a spatial average of x-ray intensities, the reconstruction procedure requires a spatial average of line integrals of the linear attenuation coefficient.

Figure 6 shows the radial dependence of the average linear attenuation coefficient $\mu(F)$ in a circle coaxial to the outer bone contour. The area F is normalised with the total bone area, so that $\mu(F = 0.5) = \mu'$. To study the influence of both finite beam width and finite beam height, the model depicted in figure 1 had to be extended in the axial direction. The tapering of the bones in the axial direction was approximated with cones (tapering angle 36°). The computer simulations used a non-divergent beam (width 1 mm, height 2 mm) of monoenergetic x-rays. The increase in the CT values in the region of trabecular bone due to the finite beam size is energy dependent. At 40 keV the bone parameter μ' is elevated by 0.6% for the tibia and 0.92% for the radius. For lower energies these shifts are larger (1.55 and 2.4% at 30 keV); for higher energies they gradually vanish (0.15 and 0.24% at 60 keV).

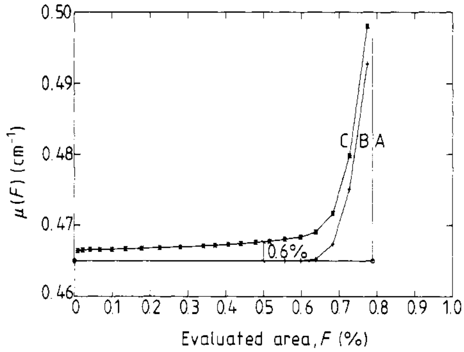

Figure 6. Degradation of $\mu(F)$ due to the finite beam size. $\mu(F)$ is the average linear attenuation coefficient calculated in a circle (area F, normalised to the total bone area) coaxial to the outer bone contour of the tibia using an energy of 40 keV. The plots are given for: A, an infinitesimally thin X-ray beam; B, a beam width of 1 mm and a beam height of 2 mm simulating only the linear degradation; and C, a beam width of 1 mm and a beam height of 2 mm including the non-linear degradation.

4. Discussion

The specifications of a QCT based bone densitometer can be described by the four parameters sensitivity, precision, accuracy and radiation dose. These parameters are highly energy dependent and are also very much dependent on the object to be examined. Therefore, it is not recommended that the same CT settings are used for all measuring sites of the human body. We focused our attention on the evaluation of peripheral bones, but the method described to get an optimal energy with respect to minimum dose can be used for any QCT parameter at any measuring site.

In a publication by Keller *et al* (1980) the optimum energy selection for CT iodinated contrast studies was reported. Measuring a head phantom with a CT-1010 in the range 80–140 kV$_p$ they found a monotonically increasing detectability with decreasing energy when exposure was held constant. This is in agreement with our theoretical findings and our computer simulations (figure 3).

With regard to minimum dose per CT examination peripheral bone measurements should be performed in the energy range 30–40 keV (figures 5*a* and 5*b*). For energies below 30 keV the sensitivity of the linear attenuation coefficient on changes of the trabecular bone density is higher (figure 2), but the precision of the measured bone parameter μ' is much smaller (figure 3). For energies above 40 keV precision is higher but the sensitivity is diminished. Energies higher than 40 keV are not recommended for an additional reason. The soft tissue compartment in the trabecular bone region is assumed to have a constant composition during a course of bone evaluations. However, bone marrow might change its composition or density, thus simulating a variation in the bone density. To minimise this effect full use should be made of the high contrast between bone and soft tissue at low energies.

The accuracy of the bone parameter μ' depends slightly on the size of the X-ray beam (figure 6). Using a beam of 1 mm × 2 mm and 40 keV, accuracy of the bone parameter is better than 1% at peripheral measuring sites. To achieve, however, a high reproducibility the same region of interest has to be evaluated in all tomograms.

Since bone density is non-homogeneously distributed it is necessary to establish a method for the evaluation of identical bone samples in longitudinal studies. This is achieved by measuring stacks of tomograms whereby the stacks are positioned with the help of a scanning radiograph. The cross sectional area of the distal radius or distal tibia varies monotonously in the axial direction of the bone. Matching the stacks according to the cross sectional area of the bone allows calculation of the bone parameter in a volume common to all examinations. This procedure (Rüegsegger *et al* 1982) is highly reproducible so that full use of the optimal energy selection can be made. Thus measurements of the radius and the tibia permit a very precise and accurate evaluation of the trabecular bone at low radiation dose.

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Résumé

Réglage optimum des paramètres du scanneur pour l'évaluation des densités d l'os.

Dans la mesure des densités à l'aide du scanneur, la reproductibilité, la sensibilité et l'exactitude du procédé sont limités par le bruit, le degré de différenciation des tissus et les artéfacts induits par la largeur du faisceau de RX, le durcissement du faisceau et le rayonnement diffusé. Ces artéfacts dépendent de façon importante des paramètres de réglage du scanneur, tels que le spectre de Rayons X, l'énergie effective, la fluence des photons et la collimation du faisceau. Comme la surveillance de l'évolution d'une maladie, ou de l'efficacité d'un traitement, nécessitent plusieurs examens étalés dans le temps, la dose délivrée par examen doit être maintenue à une valeur minimale. Pour l'évaluation de la densité de l'os du squelette, les auteurs montrent qu'il existe un intervalle d'énergie optimal étroit de 30 à 40 keV, pour lequel une dose inférieure à 150 μSv suffit pour détecter une modification de 1% dans la densité de l'os trabéculaire, avec un niveau de confiance à 0,1. De même, l'utilisation d'une collimation adaptée du faisceau pourrait amener la précision à mieux que 1%.

Zusammenfassung

Optimale CT-Einstellungen für Knochendensitometrie.

Bei computertomographischen Dichtebestimmungen wird die Genauigkeit und die Sensitivität des Verfahrens begrenzt durch das Quantenrauschen, den Grad der Gewebedifferenzierung und Artefakte als Folge der endlichen Breite und Höhe des Photonenstrahls, der Strahlaufhärtung und der Streustrahlung. Das Quantenrauschen, der Grad der Gewebedifferenzierung und die Artefakte sind abhängig von den CT-Einstellungen wie mittlere Energie und Spektrum der Röntgenstrahlung, Strahlintensität und Strahlkollimierung. Um den Krankheitsverlauf oder den Therapieerfolg zu kontrollieren, sind wiederholte Messungen erforderlich; die Strahlendosis pro Messung sollte deshalb möglichst niedrig ausfallen. Es wird gezeigt, dass für Knochendichtemessungen am peripheren Skelett ein optimaler Energiebereich zwischen 30 und 40 keV existiert, indem eine Dosis von weniger als 150 μSv genügt, um eine Dichteveränderung des trabekulären Knochens von 1% mit 90% Sicherheit zu erkennen. Nichtlineare Effekte, die ihren Ursprung in den endlichen Dimensionen des Photonenstrahls haben, erhöhen den trabekulären Knochenparameter. Diese systematische Erhöhung kann bei einer Energie von 40 keV und entsprechender Strahlkollimierung kleiner als 1% gehalten werden.

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