The CT's sample volume as an approximate, instrumental measure for density resolution in densitometry of the lung

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Ultimately CT-densitometry of the lung should give comparable results on all scanners. One prerequisite for this is the use of the same density resolution. Unfortunately, density resolution is impractical as a performance specifying parameter because it depends on the cellular material scanned. Therefore, another parameter that can be used for scanner and protocol characterization, and that does not depend on a special phantom, would be highly preferable. We investigated how well the CT's nominal sample volume (V), calculated from section thickness and in-plane spatial resolution as specified by the CT manufacturer, can serve as a simple measure for density resolution. Six CT scanners were studied using foam and lung phantoms. On all scanners we observed for foam an approximately linear relation between density resolution and $V^{-1/2}$. Density resolution on different scanners varied to some extent. These differences can be interpreted as being caused by deviations of the true sample volume from the nominal value: the 95%-confidence interval runs for instance for $V=8$ mm^3 from 4.6 mm^3 to 16.9 mm^3. Acceptability of this spread depends on the consequences for parameters of clinical interest, like percentiles and pixel indexes. To evaluate this we used data from a previous patient study on the dependence of histogram parameters on sample volume. With these data it is found that large interscanner differences in histogram parameters are possible for small values of V, as used in thin-section densitometry. For larger values of V, as required for a more adequate density resolution, the differences are much smaller and probably acceptable when compared to other sources of variability in lung densitometry. In conclusion, for sections of at least 2 mm and smooth reconstruction filters, corresponding to $V \gtrsim 8$ mm^3, the CT's nominal sample volume might be used for interscanner and interprotocol comparison of density resolution. [S0094-2405(97)00410-0]

Key words: densitometry, lung, density, CT technology

I. INTRODUCTION

CT-densitometry of the lungs will not find broad introduction until protocols have been developed that give comparable results on all scanners. On the long term a situation as achieved in standard lung function tests should be realized. One important key to comparability is the use of spirometrical definition of the degree of inspiration. This technique has been introduced a number of years ago and is now commonly used on some scanners.[1] Another important key is density resolution; its importance was only recently pointed out.[2]

In analogy with definitions of resolution in other fields of science, density resolution was defined as the width of the CT number histogram obtained from a sample of uniform average composition. Density resolution determines the way the actual mass distribution in the lung is reflected in a histogram. It directly affects all histogram parameters that depend on the shape of the histogram. For instance, when poor density resolution broadens a histogram, percentiles and pixel indexes will change. The mechanism controlling density resolution is most easily explained using the CT's sample volume, the latter being defined as the volume in a scanned section over which the attenuation value shown in a pixel of an image has effectively been averaged. Poor density resolution, leading to a broadened CT number histogram, is obtained for sample volumes that are similar or smaller in size than the structures in lung and foam. In sample volumes of this small size the relative amounts of air and solid will vary strongly, and since the CT number is proportional to the amount of solid that is present in the sample volume, a large spread in CT numbers will be obtained. Of course, there is also a contribution of quantum noise to density resolution, but for lung and coarse foams this contribution is small for normal mAs-values.[2] Density resolution can be improved by increasing the sample volume, i.e., by using a thicker section and a smoother reconstruction filter that lowers the in-plane spatial resolution. It will be clear from this discussion that two scanners, or scanning protocols, only will produce the same histogram when the density resolution is the same in both cases.

A problem of density resolution as a performance specifying parameter, for some scan protocol on some scanner, is its dependence on the material being scanned, specifically the dimensional characteristics and density of the cellular structure. Thus for specifications to be generally useful it would be required that they were obtained with identical cellular phantoms. This is hard to realize because foam characteristics are difficult to control. Nevertheless, there is a

TABLE I. CT scanners, scan techniques, and spatial resolution of reconstruction filters.

Manufacturer[a]	Scanner type	HV[b] (kV)	Time[b] (s)	TL[b] (mAs)	Reconstruction filter/FWHM PSF (mm)[c]		
					Standard	Smooth	Very soft
GE	HiSpeed	140	1	210	STD/1.16	soft/1.25	-
Philips	SR7000	140	2	250	5/1.2	3/1.5	-
Philips	Tomoscan AV	140	2	250	5/1.09	3/1.36	-
Picker	PQ-2000	130	1	200	STD/1.02	smooth/1.36	-
Siemens	Somatom Plus	137	1	220	7055/1.13	7057/1.40	7059/1.86
Siemens	Somatom Plus 4	140	1	200	AB50/1.12	AB30/1.34	AB10/1.67

[a]GE Medical Systems, Milwaukee, Wisconsin; Philips Medical Systems, Best, The Netherlands; Picker International, Inc., Cleveland, Ohio; Siemens Aktiengesellschaft, Erlangen, Germany.
[b]HV: High voltage; Time: acquisition time of a single, nonspiral scan; TL: tube load.
[c]All resolution data are for the large focus of the x-ray tube.

clear need for some simple and practical measure of density resolution for interprotocol and interscanner comparison. On physical grounds, as discussed in the previous paragraph, one might expect that the CT's sample volume (V) could be a suitable parameter. This parameter would be particularly attractive when it could be approximated by a value calculated from the nominal specifications for section thickness and in-plane spatial resolution as provided by the manufacturer, obviating special measurements by the densitometrist.

In a previous study it was shown for a very simple model of a cellular solid that one expects in fact a linear relation between density resolution and $V^{-1/2}$, at least when V is somewhat larger than the cell size of the cellular solid.[2] (This inverse square root dependence of density resolution on V may be grasped immediately for the simple model of sampling in a random mixture of identical particles and air: the number of particles in a volume V, or equivalently the density, has a relative standard deviation that is proportional to $V^{-1/2}$; for details see Ref. 2.) Encouraging results for several different foams, and air, were obtained for the particle scanner that was used in that study. In the present study we extended these measurements to six CT scanners from four different manufacturers using foam and lung phantoms. Basically we wanted to answer the question whether, for a given cellular solid, some nominal sample volume always corresponds to the same density resolution. To this purpose we decided: (a) to investigate the relationship between density resolution and nominal sample volume on several scanners; (b) to compare the results from different scanners; and (c) to estimate the limitations of the use of the nominal sample volume as a relative measure for density resolution.

II. MATERIALS AND METHODS

A. CT scanners

Six CT scanners from four different manufacturers were included in this study. These scanners, and the technique parameters used on them, are given in Table I. Information on in-plane spatial resolution, expressed as the full width at half-maximum (FWHM) of the point spread function (PSF), was provided by all manufacturers, except Picker. Picker supplied instructions how to measure the PSF and they concurred with our results. On all scanners a calibration scan was executed before performing the actual measurements. We applied in all cases a thorax/lung protocol, a 360-degree scan (not spiral), a matrix size of 512×512, and a field of view (FOV) of approximately 350 mm.

B. Phantom studies

Due to the radiation dose associated with CT one cannot use volunteers or patients for protocol or scanner characterization. Phantoms were therefore used: One phantom was a 5 cm thick slice of polymethylmethacrylate (PMMA) in the form of a cross section of an average male thorax. In the positions of the lungs were cavities. One of these cavities contained polyethene (PE) foam of a density of 0.096 g/cm^3 (96 kg/m^3). This foam has a cell diameter of approximately 1.0 mm. We will refer to this foam as Foam 96. The other lung cavity was left empty.

On all CT systems scans were performed with section thicknesses of 1 or 1.5 mm, depending on availability, 2 mm if available, 3 mm, 5 mm, and 10 mm. The scan data were reconstructed with several reconstruction filters (also called kernels).

In all images a large elliptical region of about 110 cm^2 was drawn in the area covered by Foam 96. A CT number frequency distribution (histogram) was generated from the data in this region. The average CT number and the standard deviation (sd) of the histogram were determined. None of the CT scanners did provide the width of the histogram, which is, according to the definition, the looked for density resolution. However, density resolution is directly proportional to the standard deviation (for this foam $2.05 * \mathrm{sd}$ according to previously performed experiments).[2] Since the sd is also a more robust parameter than the width of the histogram,[2] we will further use this sd to represent density resolution.

The sd was plotted as a function of $V^{-1/2}$, with V the CT's nominal sample volume, calculated as $V=S*L^2$, with Ssection thickness, and L the effective in-plane sample size. For a Gaussian PSF it is readily shown that

$$L=\frac{\sqrt{2\pi}}{2\sqrt{2\ln 2}}\,\mathrm{FWHM}=1.064\ \mathrm{FWHM},$$

with FWHM the full width at half-maximum of the PSF. (The effective sample area A is

$$A=\int_0^\infty e^{-r^2/2\sigma^2}2\pi r\,dr=2\pi\sigma^2.$$

Since

$$\sigma=\frac{\mathrm{FWHM}}{2\sqrt{2\ln 2}},\quad A=\frac{\pi\ \mathrm{FWHM}^2\ S}{4\ln 2}.$$

Although the PSF is rotationally symmetric, we defined for convenience L as the width of an equally sized square, thus $L=\sqrt{A}$.) More or less Gaussian PSF's have been reported for not too sharp reconstruction filters.[3] Air in the empty lung cavity was analyzed in the same way. In a previous study a linear relationship was found between density resolution and $V^{-1/2}$ for foam, and also for air.[2] For each scanner the sd versus $V^{-1/2}$ scatter diagrams were therefore fitted with a linear function.

In order to quantify the spread in nominal sample volume for a given density resolution 95%-confidence interval predictions were calculated. Data from all scanners were put together, and after elimination of the y-intercept and performing a logarithmic transformation, linear regression was performed, 95%-confidence intervals were calculated, and then the data were backtransformed. The logarithmic data transformation is applied to allow for the fact that the spread in density resolution, just like density resolution itself, goes to zero with $V^{-1/2}$. The present approach even assumes that both depend linearly on $V^{-1/2}$.

On all scanners we also investigated for Foam 96 the effect of FOV on the relation between density resolution and nominal sample volume. For a FOV of 350 mm and a 512 $\times$512 matrix the pixel size is 0.68 mm. Although this value is always smaller than the FWHM of the PSF (see Table I), it is not so much smaller that changing it slightly might not affect the PSF and sample volume to some degree.

The second phantom was a humanoid thorax containing dog lung.[4] One lung in the images was analyzed using a circular region with an area of 20 cm^2. Great care was taken in positioning of phantom and region to limit differences between subsequent scans caused by analyzing different tissue.

All images were analyzed on the various acquisition systems, except in our own hospital, where the data were transferred to an image processing station (ICON Power PC, Siemens Gammasonics, Hoffman Estates, IL).

III. RESULTS

A. Phantom studies: Foam

Figure 1 shows by way of example the CT number histograms for Foam 96 obtained on the Somatom Plus of our hospital. In Fig. 2 we show for all scanners the relation between density resolution, here represented by the standard deviation of the histogram, and $V^{-1/2}$. Except for some scatter in the data the relation appears to be linear. Table II presents results for the separate scanners in the form of slopes and intercepts of lines fitted to the data. When the data from all scanners together are fitted the slope is 31.1 H mm$^{3/2}$, the intercept -2.1 H. (H stands for Hounsfield. It is the commonly used unit of attenuation in CT.) Note that data obtained with thin sections (1 and 1.5 mm) were not included in fitting,[2] and are not shown in Fig. 2. The CT's sample volume was calculated from nominal section thickness and in-plane spatial resolution, both as specified by the manufacturer. For the Somatom Plus in our hospital we compared the

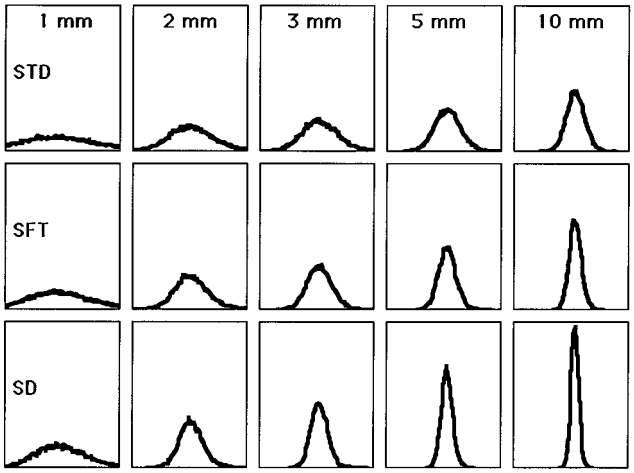

FIG. 1. CT number histograms from Foam 96 as a function of section thickness (columns) and reconstruction filter (rows). Horizontal axis: CT number from -960 H to -860 H. Vertical axis: CT number frequency, all scales identical. Scanner Somatom Plus. The nominal sample volume changes from 1.4 mm^3 (left top) to 39 mm^3 (right bottom).

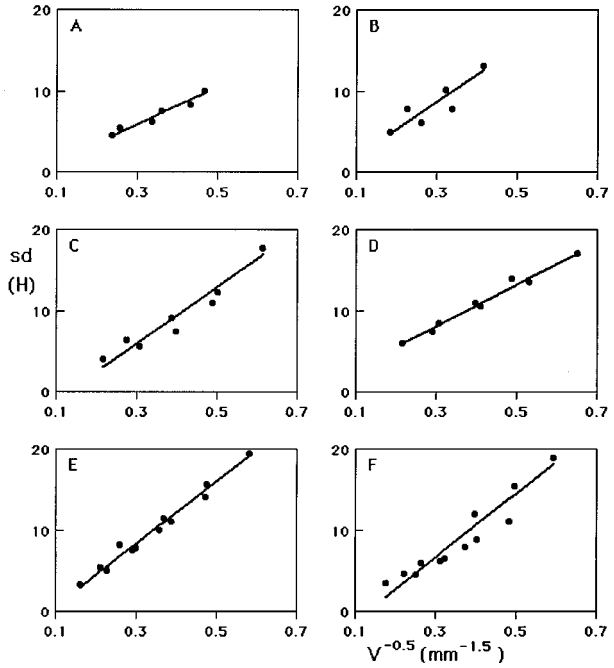

FIG. 2. Standard deviation of the histogram from Foam 96 in the PMMA thorax phantom as a function of $V^{-1/2}$, with V the CT's nominal sample volume. A fit with a line is also shown. (A) GE-HiSpeed, (B) Philips SR 7000, (C) Philips Tomoscan AV, (D) Picker PQ-2000, (E) Siemens Somatom Plus and (F) Siemens Somatom Plus 4.

TABLE II. PMMA thorax phantom: Slope and offset of linear fit to histogram's sd versus $V^{-1/2}$ curves. Note: density resolution is approximately 2.05×sd for foam, 2.36×sd for air (Ref. 2).

	Foam 96[a]		Air	
Scanner	Slope (H mm$^{3/2}$)	Intercept (H)	Slope (H mm$^{3/2}$)	Intercept (H)
HiSpeed	21.5	−0.6	7.8	1.1
SR 7000	31.1	−0.9	13.9	2.0
Tomoscan AV	32.2	−3.8	13.4	−0.7
PQ-2000	26.0	+0.2	11.8	1.8
Somatom Plus	37.5	−2.9	19.6	−0.8
Somatom Plus 4	36.3	−4.3	10.5	0.3

[a]Slopes and intercepts are from the fitted lines in Fig. 2.

specification on resolution with our PSF measurements:[2] All values agreed within 0.02 mm, which is quite satisfactory.

The scatter that is present in the data points is largely systematic of nature, not stochastic, as was shown by repeated measurements performed on all scanners after 6 months. The results of the two sets of experiments were very similar: The mean of the pairwise calculated differences between the two measurements of the sd was for the Hispeed -0.1 ± 0.3 H, for the SR 7000 -0.3 ± 0.2 H, for the Tomoscan AV 0.7 ± 1.0 H, for the PQ 2000 2.2 ± 0.8 H, for the Somatom Plus 0.2 ± 0.3 H and for the Somatom Plus 4 it was 1.0 ± 0.8 H.

Figure 3 shows the results from the calculation of 95%-confidence interval predictions of $V^{-1/2}$ for given density resolution. Only data from section thicknesses between 2 and 10 mm have been used in this estimation. The full 95%-confidence interval varies from 45% of the predicted $V^{-1/2}$ value ($\langle V^{-1/2}\rangle$) at the lowest standard deviation, where $\langle V^{-1/2}\rangle = 0.19$ mm$^{-3/2}$, to 75% at the highest, where $\langle V^{-1/2}\rangle = 0.65$ mm$^{-3/2}$. For comparison the data from thin sections (1 and 1.5 mm) are also shown in Fig. 3 (open squares). When these data are included in the fit the 95%-confidence interval increases slightly: For instance, at a $\langle V^{-1/2}\rangle = 0.35$ mm$^{-3/2}$, corresponding to $\langle V\rangle = 8$ mm^3, from 62% to 68% of $\langle V^{-1/2}\rangle$.

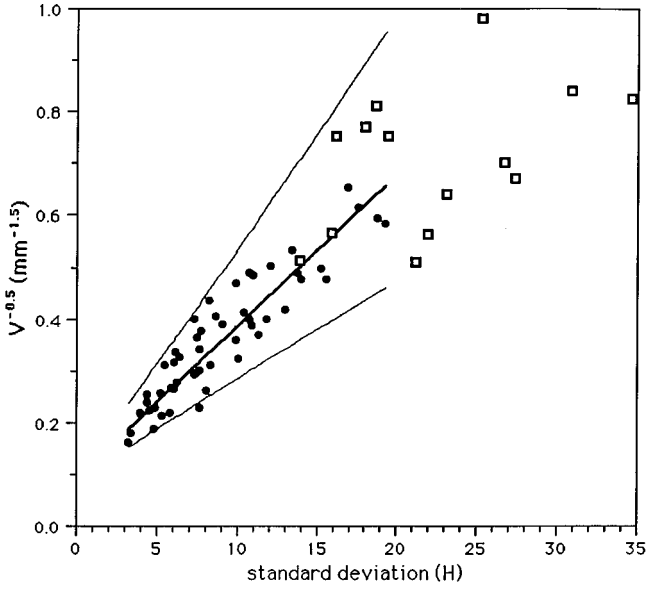

FIG. 3. Foam 96 in the PMMA thorax phantom: 95%-confidence interval for $V^{-1/2}$ (thin lines) as a function of the standard deviation of the histograms, based on sections of 2 mm and thicker. Predicted values are shown also (fat line). Scanners: HiSpeed, SR 7000, Tomoscan AV, PQ-2000, Somatom Plus, Somatom Plus 4. Data from 1 mm and 1.5 mm sections (open squares) are shown for comparison.

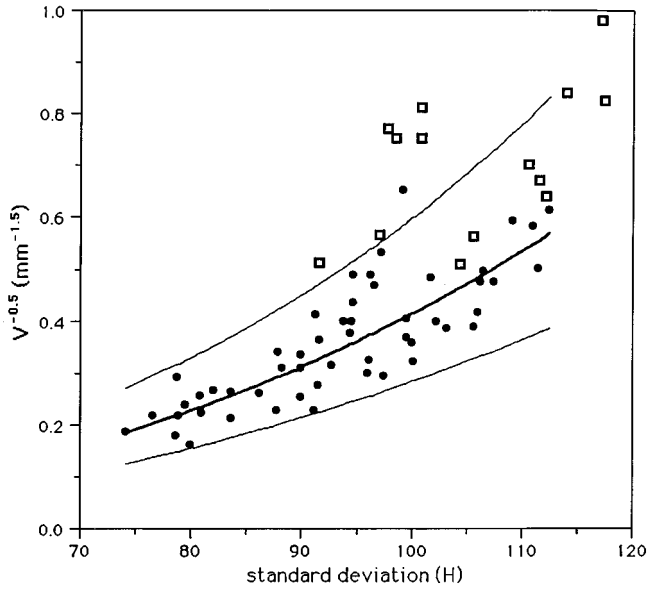

FIG. 4. Dog lung in the humanoid thorax phantom: 95%-confidence interval for $V^{-1/2}$ (thin lines) as a function of the standard deviation of the histograms, based on sections of 2 mm and thicker. Predicted values are shown also (fat lines). Scanners: HiSpeed, SR 7000, Tomoscan AV, PQ-2000, Somatom Plus, Somatom Plus 4. Data from 1 mm and 1.5 mm sections (open squares) are shown for comparison.

The dependence of density resolution on field of view, was generally very small on all scanners: changing the FOV from approximately 300 mm to 480 mm or 500 mm, for 1 mm or 1.5 mm sections and a standard reconstruction filter, resulted on all scanners in changes in standard deviation that were below 1 H. Only on the GE HiSpeed the density resolution improved for the largest FOV's: The standard deviations being 18.7 H at a FOV of 200 mm, 18.6 H at 280 mm, 18.7 H at 330 mm, 18.7 H at 380 mm, 17.6 H at 420 mm and 15.7 H at 480 mm.

The conformity of all scanners regarding mean density estimation was quite satisfactory: For Foam 96 the mean difference between the CT number for Foam 96 and the CT number for air was 91.2 H$\pm$1.1 H, where we averaged over all scanners and all section thicknesses. The small spread is in agreement with previous findings.[5]

B. Phantom studies: Dog lung

Figure 4 shows the results from the estimation of the 95%-confidence interval of $V^{-1/2}$ for dog lung, using the data from all scanners, but again excluding results for 1 and 1.5 mm sections. The full 95%-confidence interval is ap-

TABLE III. Difference in histogram parameters corresponding to a $V^{-1/2}$change equal to 95% confidence interval.

V[a] (mm^3)	95%-predict. interval (% of $\langle V^{-1/2}\rangle$)	P(10)[b] (H)	P(90)[b] (H)	PI(−950)[c] (%)	PI(−905)[c] (%)	sd[d] (H)
1.4	87	33	27	21	18	58
8	62	10	8	6	5	17
27	46	4	3	2	2	7
Slope[e]		−45 H mm$^{3/2}$	37 H mm$^{3/2}$	28% H mm$^{3/2}$	25% mm$^{3/2}$	79 H mm$^{3/2}$

[a]The values correspond with $\langle V^{-1/2}\rangle = 0.845$ mm$^{-3/2}$, 0.354 mm$^{-3/2}$, and 0.192 mm$^{-3/2}$, respectively.
[b]Percentile P(x) is the CT number below which x% of the histogram area extends.
[c]Pixel index PI(N) is the percentage of the histogram area below CT number N.
[d]Standard deviation of the histogram.
[e]Reference 6, Table I. Maximum of mean values from various groups.

proximately 77% (±2%) of the predicted $V^{-1/2}$ value over the whole fitted interval. For comparison data from 1 and 1.5 mm sections are included in Fig. 4 (open squares). When these thin sections are included in the fit the 95%-confidence interval increases to about 100% of $\langle V^{-1/2}\rangle$.

IV. DISCUSSION

For all scanners we found for Foam 95 an approximately linear relationship between density resolution and $V^{-1/2}$, with V the CT's nominal sample volume. Density resolution was here expressed by the standard deviation (sd) of the histogram. Intrascanner deviations from linearity were for some scanners very small, for others in a few cases similar in magnitude to the interscanner spread. Possible explanations for the observed intrascanner deviations from linearity are: (a) The full width at half maximum (FWHM) of the point spread function (PSF) does not accurately characterize the effective sample width if the PSF is not truly Gaussian. Stated otherwise, modulation transfer functions may differ, even when the FWHM of the PSF is identical. (b) Small differences between nominal and effective section thickness may exist.

These two sources of variation, mentioned under (a) and (b), may also partly account for the differences between the various scanners. An additional cause of interscanner variation is a difference in the contribution of quantum noise to density resolution. According to the slopes of the sd versus $V^{-1/2}$ curves of air in the PMMA thorax phantom this is indeed the case (Table II). It is noted that the slope of the sd versus $V^{-1/2}$ curve due to sampling effects only is given by $\sqrt{(\text{slope}^2_{\text{foam 96}} - \text{slope}^2_{\text{air}})}$. For example, for the Somatom Plus this correction reduces the slope from 37.5 H mm$^{3/2}$ to 32.0 H mm$^{3/2}$, a value which is still considerably larger than the value for air that is determined by quantum noise (19.6 H mm$^{3/2}$). Clearly, for common mAs-values the magnitude of the contribution of quantum noise to density resolution is relatively small. Moreover, it are in fact only the (much smaller) interscanner or interprotocol differences that count. In patient studies of lung the contribution of quantum noise to density resolution was likewise small, except at very low density.[2] When the contribution of quantum noise to density resolution becomes substantial, the sample volume can be no longer a good single measure for density resolution, because also the mAs-value will matter in this case.

One can characterize the interscanner differences in nominal sample volume with a 95%-confidence interval: The interval extended from approximately 45% of the predicted $V^{-1/2}$ at $\langle V^{-1/2}\rangle = 0.19$ mm$^{-3/2}$ to 75% at $\langle V^{-1/2}\rangle = 0.65$ mm$^{-3/2}$ (Fig. 3). The large change in relative confidence interval is caused by the non-zero positive intercept in Fig. 3. These 95%-confidence intervals correspond to large changes in V: For instance, for $\langle V^{-1/2}\rangle = 0.35$ mm$^{-3/2}$, corresponding to $\langle V\rangle = 8$ mm^3, the interval runs from 4.6 mm^3 to 16.9 mm^3.

Density resolution is dependent not only on CT parameters but also on the structural properties of the cellular material. Previously we found on one scanner the same linear dependence of density resolution on $V^{-1/2}$ for five different foams and air.[2] Nevertheless, it might be questioned whether a comparison with PE foam and air is sufficient to characterize the various scanners. In order to address this question lung in a humanoid phantom[4] was studied as a more realistic, but also a more difficult model. The 95%-confidence interval was estimated as approximately 77% of $\langle V^{-1/2}\rangle$ over the whole range. This value is larger than the values obtained from the study on Foam 96. To give an example, at $\langle V^{-1/2}\rangle = 0.35$ mm$^{-3/2}$, i.e., at $\langle V\rangle = 8$ mm^3, we have 77% of $\langle V^{-1/2}\rangle$ for lung versus 62% for Foam 96. That the confidence interval obtained from the lung study is somewhat larger than that from the foam study is certainly partly due to problems associated with nonuniformity of the lung. Different section thicknesses sample different tissue, and besides that one is liable to errors in positioning of the phantom and in drawing the region of interest.

Whether this variability in effective sample volume is compatible with a use of the nominal sample volume as a measure for density resolution depends on the consequences for parameters of clinical interest. This can be evaluated using results from another study, on patients, in which the dependence of several histogram related parameters on sample volume was investigated.[6] We showed that it was possible to calculate the change in these parameters per unit change of $V^{-1/2}$, i.e., as the slope of a linear fit to the data. Average

slopes from this study are used to estimate differences in the histogram parameters that correspond to a change in $V^{-1/2}$ equal to the 95%-confidence interval. The results are considered to be also a reasonable estimate for possible histogram parameter changes when using the same nominal sample volume on different scanners. We performed these calculations for three magnitudes of the nominal sample volume: 1.4 mm^3, being a typical sample volume for the combination of a 1 mm section thickness and a standard reconstruction filter, 8 mm^3, the volume corresponding to a 2 mm section and a smooth filter having an in-plane spatial resolution of 1.86 mm, and a still larger sample volume of 27 (3^3) mm^3. The values in Table III were calculated as ''average slope $*$ $\langle V^{-1/2} \rangle$ $*$ relative 95%-confidence interval for Foam 96,'' with ''average slope'' according to the largest mean values in Table I in Ref. 6.

From Table III we conclude that interscanner variability potentially leads to very large variations in histogram related parameters when the sample volume is small, as it is in thin section densitometry. This is another reason, in addition to the very poor density resolution one obtains in this case,[2] not to recommend this protocol. Larger sample volumes, that give a more adequate density resolution allowing ''density spectroscopy,'' have a concommitant lower sensitivity to interscanner variability in effective sample volume. From these examples it is obvious that the nominal sample volume, as calculated from easily accessible scanner specifications, certainly has its limitations as a measure for density resolution. However, for not too small sample volumes, e.g., larger than 8 mm^3, the sensitivity to scanner variations might be acceptable, at least when compared with other sources of variability in densitometry of the lung. We think therefore that the CT's sample volume can be used as a practical, approximate measure of density resolution for the purpose of data comparison. We strongly recommend to specify this sample volume, including section thickness and in-plane resolution from which it was calculated, in all future densitometry studies of the lung. Until now no attention was paid to density resolution, or any other parameter that could be used in deciding on the comparability of lung densitometry results.

The question might be raised whether simple void free solids cannot be used for the characterization of a scanner's density resolution considering that image noise also depends on sample volume. They cannot, however, because noise can differ in a nearly unaccountable way between different scanners, even for the same mAs value or entrance dose, due to differences in x-ray beam quality, detection efficiency or system noise. Two scanners with the same effective sample volume, but different in any of these other aspects will show a different density resolution for a full solid. But for lung and coarse foams, where sampling effects dominate noise, it is likely that the two scanners will behave very similarly.

Interscanner variability might probably significantly be reduced when on all scanners similar filters were introduced for densitometry. Today most scanners do not even have reconstruction filters that are sufficiently smooth to realize isotropic sample volumes of sufficient size. Fortunately, several manufacturers are presently developing lung densitometry options for their scanners. We hope that they will provide tools for tailoring density resolution as well as the required data on sample volumes.

Finally, a word of caution when calculating sample volumes from the manufacturer's specifications. On a few scanners reconstruction filters have been implemented that consist of a reconstruction kernel and an image filter, probably to enlarge for a limited number of reconstruction kernels the number of possible reconstructions. The effect of the image filter cannot be included in the resolution specifications because its effect depends on the field of view. We think it unwise to use such combinations of reconstruction kernel and image filter because the sample volume is difficult to estimate.

V. CONCLUSION

Progress in the field of lung densitometry requires that effects of density resolution are taken into account. We investigated how well the CT's nominal sample volume can serve as a simple measure for density resolution in interscanner and interprotocol comparison. We found that it is probably quite useful as an approximate measure when the sample volumes are not too small.

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[1] W. A. Kalender, R. Rienmüller, W. Seissler, J. Behr, M. Welke, and H. Fichte, ''Measurement of pulmonary parenchymal attenuation: use of spirometric gating with quantitative CT,'' Radiology **175**, 265–268 (1990).

[2] G. J. Kemerink, H. H. Kruize, R. J. S. Lamers, and J. M. A. van Engelshoven, ''Density resolution in quantitative computed tomography of foam and lung,'' Med. Phys. **23**, 1697–1708 (1996).

[3] E. L. Nickoloff and R. Riley, ''A simplified approach for modulation transfer function determinations in computed tomography,'' Med. Phys. **12**, 437–442 (1985).

[4] J. G. Pearce, E. N. C. Milne, G. D. Gillan, and W. W. Roeck, ''Development of a radiographic chest phantom with disease simulation,'' Invest. Radiol. **14**, 181–184 (1979).

[5] G. J. Kemerink, R. J. S. Lamers, G. R. P. Thelissen, and J. M. A. van Engelshoven, ''Scanner conformity in CT densitometry of the lungs,'' Radiology **197**, 749–752 (1995).

[6] G. J. Kemerink, H. H. Kruize, R. J. S. Lamers, and J. M. A. van Engelshoven, ''Lung densitometry: The dependence of CT number histograms on sample volume, and consequences for scan-protocol comparability,'' JCAT (in press).