Published in final edited form as:

Radiology. 2006 April ; 239(1): 79–85.

Effect of Blood Vessels on Measurement of Nodule Volume in a Chest Phantom¹

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Abstract

Purpose—To identify, by using a chest phantom, whether vessels that contact lung nodules measuring less than 5 mm in diameter will affect nodule volume assessment.

Materials and Methods—Forty synthetic nodules (20 with ground-glass attenuation and 20 with solid attenuation) that measured less than 5 mm in diameter were placed into a chest phantom either adjacent to ($n = 30$) or isolated from ($n = 10$) synthetic vessels. Nodules were imaged by using low-dose (20 mAs) and diagnostic (120 mAs) multi-detector row computed tomography (CT). Nodules that were known to lie in direct contact with vessels were confirmed by visual inspection. Nontargeted 1.25×1.00 -mm sections were analyzed with a three-dimensional computer-assisted method for measuring nodule volume. A mixed-model analysis of variance was used to examine the influence of several factors (eg, the presence of adjacent vessels; tube current–time product; and nodule attenuation, diameter, and location) on measurement error.

Results—The mean absolute error (MAE) for all nodules adjacent to vessels was 2.3 mm^3 , which was higher than the MAE for isolated nodules (1.9 mm^3) ($P < .001$). This difference proved significant only for diagnostic CT (2.2 mm^3 for nodules adjacent to vessels vs 1.3 mm^3 for nodules isolated from vessels) ($P < .05$). A larger MAE was noted for nodules with ground-glass attenuation (2.3 mm^3) versus those with solid attenuation (2.0 mm^3), for increasing nodule volume (1.66 mm^3 for nodules smaller than 20 mm^3 vs 2.83 mm^3 for nodules larger than 40 mm^3), and for posterior nodule location ($P < .05$).

Conclusion—The presence of a vessel led to a small yet significant increase in volume error on diagnostic-quality images. This represents less than one-third of the overall error, even for nodules larger than 40 mm^3 or approximately 4 mm in diameter. This increase, however, may be more important for smaller nodules with errors of less than 3 mm^3 .

Given the long-standing interest in obtaining precise, reproducible measurements for known and/or potentially malignant nodules at computed tomography (CT) in routine clinical practice, as well as the potential of using low-dose CT for lung cancer screening, considerable research has been devoted to assessing and validating computer-assisted measurements. Of particular concern is the reliable assessment of nodule growth, which is used as an indicator of malignancy for indeterminate nodules and as a method of surveillance for known malignant nodules in patients undergoing therapy. Currently, nodules are typically measured by using electronic calipers. These simple one- or two-dimensional approximations, however, are subject to large

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Author contributions: Guarantors of integrity of entire study, J.P.K., H.R.; study concepts/study design or data acquisition or data analysis/interpretation, all authors; manuscript drafting or manuscript revision for important intellectual content, all authors; manuscript final version approval, all authors; literature research, J.P.K., R.M.; experimental studies, J.P.K., R.M., E.B., C.S., M.K., H.R.; statistical analysis, J.S.B., H.R.; and manuscript editing, J.P.K., J.S.B., D.P.N., H.R.

Authors stated no financial relationship to disclose.

interobserver and intraobserver variability (1–3). This is especially true for pulmonary nodules that are smaller than 5 mm in maximal dimension; in these nodules, small variations in diameter measurements may translate into larger volume errors and therefore substantially limit the assessment of growth.

Computer-assisted methods for measuring and tracking nodule volumes have the potential to improve precision and have therefore been actively investigated in phantoms (4–6) and in vivo (4,6–10). In an early study by Yankelevitz et al (7), researchers demonstrated good accuracy for three-dimensional methods of measuring the volume of spherical and nonspherical synthetic nodules. The results of more recent studies, however, have shown that measurement methods may be affected by a number of factors related to the attenuation and size of a nodule and by technical factors, such as the use of low-dose techniques and the type of reconstruction algorithm used (5).

The presence of nearby vessels, which have attenuation values similar to those of nodules, can complicate the segmentation of nodules from the surrounding lung, thereby resulting in inaccurate measurements of nodule volume. The semiautomated segmentation of nodules by using helical CT data in vivo has been compared with the manual segmentation of nodules by a radiologist (11). To our knowledge, however, the effect of vessels on volume measurement error has not been subject to detailed study (12).

Therefore, the purpose of this study was to identify, by using a chest phantom, whether vessels that contact lung nodules measuring less than 5 mm in diameter will affect nodule volume assessment.

Materials and Methods

Phantom Construction

A chest phantom (Computerized Image Reference Systems, Norfolk, Va), which was composed of materials that simulated the attenuation of tissues in the thorax, was obtained, and 40 wells were drilled into the lung parenchyma (5). Twenty wells measuring 8 mm wide were drilled into the plastic material of each lung. In each lung, groups of four wells were placed in five locations along the anterior, medial, posterior, and lateral pleural surfaces and in the center of the lung.

The objective was for each well to simulate the microenvironment of a branching vessel and a nodule. Synthetic nodules and vessels were machined from plastic material created by blending epoxy resins and urethanes (Computerized Image Reference Systems). Vessels were constructed by using precision milling machines equipped with micrometer dial indicators. Vessels had a soft-tissue attenuation of approximately 40 HU. The cross-sectional dimensions of the 13th and 14th through 11th and 12th order branches of the pulmonary arteries (13) were chosen to simulate the common sizes of vessels that could abut small nodules. Half of the vessels were 5 mm in diameter and branched into two 3-mm branches, and the remaining vessels were 3 mm in diameter and branched into two 1.5-mm vessels. The originating and two-branch vessels were each 0.5 cm in length.

Forty nodules measuring less than 5 mm in approximate diameter were manufactured with two different attenuations. Twenty were manufactured with ground-glass attenuation (–360 HU), and 20 were manufactured with solid attenuation (40 HU). For each attenuation category, the 20 nodule specimens were divided into equal groups of four sizes, with five nodules in each size category. The four nodule sizes that were selected to approximate spheres measured 4.9, 4.0, 3.2, and 2.4 mm in diameter. The spherical nodules were then filed to roughen their surface

and distort their shape. The true volume of each nodule was then determined by using the specific gravity of the material and nodule weight, with volumes within $\pm 0.5\%$ (5).

Nodules were placed into all 40 wells of the chest phantom; 30 nodules were placed into wells that were adjacent to a vessel and 10 nodules (the control group) were placed into wells that were isolated from vessels. Combinations of vessel sizes, nodule locations, and nodule characteristics (diameter and attenuation) were determined by using randomization. The vessels and nodules within each well were surrounded with an organic mixture to simulate realistic lung attenuation and texture (5).

CT Scanning

CT was performed with a multi-detector row CT scanner (Somatom Volume Zoom Plus 4; Siemens Medical Solutions, Iselin, NJ) by using the standard protocol for chest CT at our institution (140 kVp, 0.5-second rotation time, 1.25-mm-thick transverse sections obtained at 1.0-mm intervals, and 30-cm field of view). Given our interest in evaluating the influence of radiation dose on volume measurement, the phantom was scanned by using two tube current-time products: 20 and 120 mAs. These two tube current-time products are typically used to perform low-dose (20 mAs) and diagnostic (120 mAs) CT (Fig 1). All images were reconstructed with a high-frequency reconstruction algorithm. The phantom was imaged on 5 consecutive days by using the same CT scanner, without an effort to reproduce phantom location. For the wells that contained vessels, a radiologist visually inspected the images to ensure that the nodules appeared to abut a portion of the vessel, with less than 1 voxel of separation.

Quantification Methods

A thoracic radiologist (J.P.K., with 5 years of experience) created an over-inclusive three-dimensional volume of interest (R_i) around nodule i by using locally developed, interactive, three-dimensional visualization and segmentation software (Fig 2). R_i encompassed the entire nodule not only in each transverse section but also in the craniocaudal dimension; R_i also included some lung background and voxels that were affected by partial volume averaging. R_i consisted of regions of interest (ROIs) that were placed around the nodule on every image that contained a nodule. ROIs were placed to avoid adjacent vessels and the chest wall. Because ROIs were placed around the nodules, the observer had no indication of the volumes being calculated.

A computer program that was written in the computer programming language C was developed to measure the volume of each nodule by using an improved version of the previously described partial volume method (5). The program estimated the pure lung attenuation (L_i) for the region R_i by averaging the attenuation of all voxels located along the edge of R_i that were less than -700 HU. The pure nodule attenuation (N_i) was then estimated by averaging the attenuation of a fixed fraction (F) of the densest voxels in R_i . The optimal value for F , as determined experimentally by using phantoms, was 5%, which means that the densest 5% of all voxels with attenuation values greater than -700 HU were used for N_i .

The program then considered a set of voxels (W_i) in R_i that had an attenuation value that was greater than the variable threshold (T_i), which was halfway between the attenuation of the surrounding lung L_i and the attenuation N_i of the central region of the nodule. T_i was calculated as $T_i = (L_i + N_i)/2$. The algorithm used W_i to create two sets of voxels, which were designated as W_{i+} (a superset of W_i) and W_{i-} (a subset of W_i). W_{i+} was defined as a morphologic increase of W_i by 1.5 mm, and W_{i-} was defined as the erosion of W_i by 1.5 mm (optimized through phantom experiments). The set difference $dW = W_{i+} - W_{i-}$, which represented a 3-mm-thick hollow shell, was then constructed to maximize the likelihood of containing the lung-nodule

interface (partially volumed voxels) while minimizing the containment of both the core of the nodule and the surrounding lung. Finally, the partial volume method (5) was applied to dW , and the resulting volume was added to the interior nodule volume of W_i to obtain the volume of the entire nodule.

To summarize, 40 nodules (20 with ground-glass attenuation and 20 with solid attenuation) were imaged five times with two different tube current–time products, which led to 400 observations for the first observer.

To estimate interobserver reliability, a second observer (M.K., a research assistant who had recently graduated from medical school) was blinded to the results of the first observer and independently measured the 40 nodules on CT images obtained on one of the five imaging days. This observer analyzed images that had been obtained by using both tube current–time products; thus, 80 observations were made by the second observer. The selection of the imaging day was performed randomly. The same volume quantification method was used by both observers.

Statistical Analysis

A mixed-model analysis of variance was used to examine the influence of imaging day, lung side (left lung or right lung), tube current–time product (20 or 120 mAs), nodule attenuation (ground glass or solid), presence or absence of vessels, and nodule location (anterior, medial, central, posterior, or lateral) on the assessment of absolute volume error. The model that was used to predict absolute error included true nodule volume as a fixed numeric factor and nuisance variables (day and side) and all categoric variables of interest (nodule diameter, radiation dose, attenuation, location, and vessel presence) as fixed classification factors; the model also contained all two-factor interactions among all variables of interest (except nuisance variables).

The variance-covariance structure was modeled by assuming that the error terms of the model were exchangeable when obtained from the same ROI and independent when obtained from different ROIs (owing to the acquisition of multiple measurements for each ROI). When a significant effect was detected for a multilevel nominal factor (eg, location), the Tukey honestly significant difference was used to make all pairwise comparisons among each level of a particular factor while maintaining the familywise type error rate for the set of comparisons at or below the nominal 5% level.

Results are summarized in terms of the least-squares adjusted mean absolute errors (MAEs), which are denoted by MAE and represent estimates of MAE at each level of a given factor that has been adjusted for the effects of all other factors, and the standard error of the MAE. The uncorrected nodule volumes obtained by each observer were used for the assessment of interobserver variability. Interobserver correlation was assessed by using the Pearson correlation, and a mixed-model analysis similar to that outlined above was used to evaluate the difference between observers and to assess interactions between observers and factors of interest. A P value of less than .05 was considered to indicate a statistically significant difference. All statistical computations were performed by using a commercially available software program (SAS, version 9.0; SAS Institute, Cary, NC).

Results

The MAE of volume measurement was significantly affected by tube current–time product, nodule attenuation, nodule location, nodule diameter, and the presence of a vessel; the only statistically significant interaction occurred between vessel presence and radiation dose (Table

1). With respect to MAE, there were no significant differences in the two (left vs right) lungs ($P = .138$) or among the different days of acquisition ($P = .324$).

Volume error was greater ($P = .004$) for nodules with ground-glass attenuation ($2.37 \text{ mm}^3 \pm 0.12$ [standard error]) (Fig 3) than for those with solid attenuation ($1.97 \text{ mm}^3 \pm 0.11$). In terms of MAE, the 95% confidence interval for the difference between nodules with solid attenuation and those with ground-glass attenuation extended from 0.07 to 0.73 mm^3 .

A significant increase in MAE ($P = .028$) was associated with an increase in true nodule diameter. The MAE was $1.66 \text{ mm}^3 \pm 0.16$ for nodules smaller than 20 mm^3 ; $2.25 \text{ mm}^3 \pm 0.15$ for nodules $20\text{--}40 \text{ mm}^3$; and $2.83 \text{ mm}^3 \pm 0.17$ for nodules larger than 40 mm^3 .

Different locations of the nodules within the lungs were associated with significant differences in measurement error. Specifically, nodules in the posterior aspect of each lung (Fig 3) were associated with larger MAEs than were those in the anterior, medial, central, or lateral aspect of each lung ($P < .001$, $P = .012$, $P < .001$, and $P = .002$, respectively [corrected with Tukey honestly significant difference]) (Table 2). There were no other significant differences among nodule locations in terms of MAE.

There was a significant interaction ($P < .001$) between the presence of vessels and radiation dose (Table 3) in terms of the effect on MAE. When the diagnostic CT technique (120 mAs) rather than the low-dose CT technique (20 mAs) was used, the presence of vessels led to an average increase in MAE of 0.9 mm^3 (95% confidence interval: 0.4, 1.4). MAE was greater with the low-dose CT technique ($2.36 \text{ mm}^3 \pm 0.15$) than with the diagnostic CT technique ($1.25 \text{ mm}^3 \pm 0.12$) in the absence ($P = .003$) but not in the presence ($P = .459$) of adjacent vessels. The 95% confidence interval for the increase in MAE associated with a change in radiation dose for nodules that were not adjacent to vessels extended from 0.49 to 1.73 mm^3 .

While the interobserver correlation was high ($r = 0.982$), the mixed-model analysis indicated that the observers provided significantly different ($P = .013$) volume assessments for the same set of 40 nodules imaged with the two different tube current–time products. Specifically, the observer who had less experience provided volume measurements that were a least-squares adjusted average of 0.84 mm^3 higher than those provided by the observer who had more experience (Table 4), with 95% of these differences between -5.4 and 2.4 mm^3 . No significant interaction, however, was identified between the observer and any of the factors of interest (attenuation, location, radiation dose, or vessel presence). This suggests that the difference between the observers tended to remain constant across the levels of each factor (eg, the difference between the mean volume assessments of the two observers was 0.84 mm^3 for nodules with solid attenuation and 0.86 mm^3 for nodules with ground-glass attenuation). As a result, the data from the two observers did not produce substantially different estimates for the effect of each factor on MAE.

Discussion

Measurement of nodule volume in vivo may be hindered by the presence of adjacent vessels. This makes segmentation difficult, both for the radiologist and for computer-assisted programs. Automated segmentation of nodules adjacent to blood vessels on helical CT images has been addressed primarily in vivo (10,11,14). By using a two-dimensional method, Zhao et al (11) demonstrated no significant differences between automated segmentation of nodules and manual segmentation of nodules by a radiologist (mean difference, 0.97 pixels; standard error, 7.98 pixels). In a subsequent study on three-dimensional automated segmentation of nodules from adjacent blood vessels, the visual inspection of two experienced chest radiologists confirmed excellent segmentation results in only 77% of 31 cases (14). To date, the effect of

vessels on measured volume by using the reference standard of known nodule volume has not been studied.

Given the opportunity provided by a phantom model to compare measured nodule volume with true nodule volume directly, it was our objective to analyze the effect of vessels on the measurement error associated with a semiautomated computer-assisted method for the assessment of nodule volume. The volume errors in this study averaged 2 mm^3 . To our knowledge, no other researchers have reported on the effect of the presence of adjacent vessels on nodules with known volumes. Results of error analyses for nodules without adjacent vessels are consistent with our values (5). Smaller errors, however, were reported by Yankelevitz et al (7), who used diagnostic CT to image 3–6-mm synthetic solid spheres that were surrounded by air and had no adjacent vessels. In their study, Yankelevitz et al reported better than 3% accuracy. For a typical 3-mm nodule with a volume of 16 mm^3 , this corresponds to an error of 0.5 mm^3 , which is a factor of four smaller than the values observed in our study. The high level of accuracy that was achieved in the study by Yankelevitz et al may be the result of three factors: (a) the use of a targeted, smaller field of view, (b) the high level of contrast between solid-attenuation synthetic spheres and the surrounding air rather than simulated lung parenchyma and vascular structures, and (c) the use of smaller reconstruction intervals on the order of 0.5 mm.

Our analysis demonstrated a significant difference in measurement error between nodules that “touched” vessels and those that were isolated from vessels. This effect, however, was restricted to images obtained by using diagnostic CT (120 mAs). Also, the magnitude of the effect, as measured by an increase in MAE, was only 0.9 mm^3 . These findings should be interpreted in view of previously reported MAE values for nodules without nearby vessels, which ranged from 1.5 mm^3 for 8-mm^3 nodules to 3.8 mm^3 for 61-mm^3 nodules (5). Results should also be viewed in light of our current technique, which results in a MAE ranging from 1.7 mm^3 for nodules smaller than 20 mm^3 to 3.0 mm^3 for nodules larger than 40 mm^3 .

Thus, the effect of vessels may be negligible when measuring nodules larger than 40 mm^3 in volume, or about 4 mm in diameter. In distinction, the effect of vessels cannot be ignored, particularly for nodules that are smaller than 20 mm^3 in volume or approximately 3.3 mm in diameter (assuming a spherical shape). The presence of a vessel did not have a significant effect on measurement error for images obtained with the low-dose technique, which suggests that image noise was most likely the dominant source of error on low-dose CT scans. In this study and as reported previously, the measurement error was found to be greater for the low-dose CT than for diagnostic CT (5).

Our analysis also addresses the influence of nodule positioning, nodule attenuation, and differing tube current–time products on measurement error. MAEs for nodules in the posterior aspect of the lungs were greater than those for nodules in other locations in the thorax. McCullough and Morin (15) demonstrated the effect of thoracic geometry on nodule attenuation measurements in a chest phantom. They showed that nodule attenuation varied with location and was also dependent on the CT scanner. When assessing nodule volume, it is important to be aware of the reduced accuracy associated with posteriorly located nodules. Nodules with ground-glass attenuation were, as expected, associated with a higher measurement error, which was likely related to the lower contrast of these lower attenuation nodules relative to the surrounding lung parenchyma.

We found no significant difference between the volume errors obtained from phantom nodules imaged on different days, which is reassuring given that such methods would ultimately be most useful in assessing nodule volume at different times. However, a number of factors that are present in vivo can introduce variation in measuring nodule volume in patients; these factors

include respiratory variations that can lead to different background attenuation, changes in the position of the nodules relative to adjacent structures, changes in nodule shape, and motion artifacts. Therefore, measurement errors obtained in phantoms may be smaller than those obtained in vivo.

A limitation of our study was the use of only one method for volume measurement. Therefore, the errors associated with other segmentation techniques may be affected differently by the presence of vessels that are in close proximity to a nodule. Our method was semiautomated and required a human observer to manually draw an ROI around the nodule while excluding structures that were not considered part of the nodule. The volume assessment algorithm used the ROI as an initial approximation and was not sensitive to the exact placement of the ROI. While there are reports of automated algorithms for separating nodules from nearby vessels, the effect of vessels on measurement errors associated with other methods remains to be established.

Another limitation was that we did not refine the study to investigate the variable represented by the distance between the nodule and the nearest vessel. Moreover, this investigation did not evaluate the effect of more complex relationships between nodules and vessels (eg, a vessel that passes through a nodule or a nodule that has a broad base of contact with an adjacent vessel). For these more complicated associations that occur more frequently in nodules with ground-glass attenuation (which are less common and more problematic), the magnitude of error is likely to be greater than that reported in this study. While the simulation of other complicating factors, such as the presence adjacent pathologic abnormalities in the lung, was not performed, the heterogeneous mixture that was used to fill the wells around the nodules and vessels afforded additional texture to the simulated parenchyma.

Practical applications

With our method, small nodules can be measured with an accuracy on the order of 2–3 mm³. Thus, a volume increase over time that is significantly greater than 2–3 mm³ (eg, 10 mm³) suggests growth rather than measurement error. The presence of a vessel led to a small yet significant increase in nodule volume error (from 1.3 to 2.2 mm³, or by 0.9 mm³) on diagnostic-quality images. This represents less than one-third of the overall error, even for nodules with a volume of more than 40 mm³, or a diameter of approximately 4 mm. This increase, however, may be more important for smaller nodules with errors of less than 3 mm³.

Moreover, posterior positioning in the thorax, low-dose CT, and nodules with ground-glass attenuation were associated with higher volume errors, potentially limiting the predictive value of volume measurements for nodules with these characteristics.

Acknowledgements

Supported in part by a Scholars Award from the RSNA Research and Education Foundation, the Society of Thoracic Radiology, and NIH 1 K23 CA096604-01A1

The authors acknowledge the contributions of Emilio Vega, RT, and Bernard Assadourian, RT, in the Department of Radiology at New York University Medical Center for their technical assistance with CT.

Abbreviations

MAE	mean absolute error
ROI	region of interest

References

1. Schwartz LH, Ginsberg MS, DeCorato D, et al. Evaluation of tumor measurements in oncology: use of film-based and electronic techniques. *J Clin Oncol* 2000;18:2179–2184. [PubMed: 10811683]
2. Revel MP, Bissery A, Bienvenu M, Aycard L, Lefort C, Frija G. Are two-dimensional CT measurements of small noncalcified pulmonary nodules reliable? *Radiology* 2004;231:453–458. [PubMed: 15128990]
3. Erasmus JJ, Gladish GW, Broemeling L, et al. Interobserver and intraobserver variability in measurement of non-small-cell carcinoma lung lesions: implications for assessment of tumor response. *J Clin Oncol* 2003;21:2574–2582. [PubMed: 12829678]
4. Yankelevitz DF, Gupta R, Zhao B, Henschke CI. Small pulmonary nodules: evaluation with repeat CT—preliminary experience. *Radiology* 1999;212:561–566. [PubMed: 10429718]
5. Ko JP, Rusinek H, Jacobs E, et al. Volume measurement of small pulmonary nodules on chest CT: a phantom study. *Radiology* 2003;228:864–870. [PubMed: 12954901]
6. Jennings SG, Winer-Muram HT, Tarver RD, Farber MO. Lung tumor growth: assessment with CT—comparison of diameter and cross-sectional area with volume measurements. *Radiology* 2004;231:866–871. [PubMed: 15163822]
7. Yankelevitz DF, Reeves AP, Kostis WJ, Zhao B, Henschke CI. Small pulmonary nodules: volumetrically determined growth rates based on CT evaluation. *Radiology* 2000;217:251–256. [PubMed: 11012453]
8. Revel MP, Lefort C, Bissery A, et al. Pulmonary nodules: preliminary experience with three-dimensional evaluation. *Radiology* 2004;231:459–466. [PubMed: 15128991]
9. Kostis WJ, Yankelevitz DF, Reeves AP, Fluture SC, Henschke CI. Small pulmonary nodules: reproducibility of three-dimensional volumetric measurement and estimation of time to follow-up CT. *Radiology* 2004;231:446–452. [PubMed: 15128989]
10. Kostis WJ, Reeves AP, Yankelevitz DF, Henschke CI. Three-dimensional segmentation and growth-rate estimation of small pulmonary nodules in helical CT images. *IEEE Trans Med Imaging* 2003;22:1259–1274. [PubMed: 14552580]
11. Zhao B, Yankelevitz D, Reeves A, Henschke C. Two-dimensional multi-criterion segmentation of pulmonary nodules on helical CT images. *Med Phys* 1999;26:889–895. [PubMed: 10436889]
12. Beer AJ, Kohl G, Schaller S, Rummeny EJ. Automated volumetry of small pulmonary nodules with MSCT: influence of nodule-shape, scan- and reconstruction-parameters on reproducibility [abstr]. *Radiology* 2002;225(P):255–256.
13. Fraser, RS.; Muller, NL.; Colman, N.; Pare, PD. Fraser and Pare's diagnosis of diseases of the chest. Philadelphia, Pa: Saunders; 1999.
14. Zhao B, Reeves AP, Yankelevitz DF, Henschke CI. Three-dimensional multicriterion automatic segmentation of pulmonary nodules of helical computed tomography images. *Opt Eng* 1999;38:1340–1347.
15. McCullough EC, Morin RL. CT-number variability in thoracic geometry. *AJR Am J Roentgenol* 1983;141:135–140. [PubMed: 6602510]

Figure 1. Transverse CT scan of the left side of the chest phantom. Vessels and nodules can be seen imbedded in the simulated lung parenchyma.

Figure 2.
CT scan of 1.25-mm section obtained through a nodule with ground-glass attenuation demonstrates ROI placed around nodule.

Figure 3. Transverse CT scans of 1.25-mm sections obtained through a nodule of solid attenuation located adjacent to a vessel in the posterior region of the left lung. **(a–f)** Diagnostic CT scans demonstrate a nodule (black arrow) adjacent to vessel branches (white arrows). Note the streak artifact in the vicinity of this nodule. **(g)** Low-dose CT scan demonstrates increased image noise and corresponds to **f**.

Table 1
Significance of Factors and Interactions between Factors on Measured Nodule Volume

Parameter	<i>P</i> Value
Nodule location	<.001
Nodule attenuation	.004
Nodule diameter	.028
Vessel presence and radiation dose	<.001
Vessel presence	
120 mAs	<.001
20 mAs	.288*
Radiation dose	
Vessel present	.002
Vessel absent	.459*

* *P* value was not statistically significant.

Table 2

MAE according to Nodule Location

Location	MAE (mm ³)
Anterior	1.95 ± 0.17
Medial	2.01 ± 0.20
Central	1.81 ± 0.19
Posterior	2.84 ± 0.20
Lateral	2.08 ± 0.18

Note.—All data are the MAE ± the standard error of the MAE.

Table 3
Effect of Tube Current–Time Product on MAE in the Presence or Absence of Vessels

Vessel	120 mAs	20 mAs
Present	2.15 ± 0.21	2.30 ± 0.26
Absent	1.25 ± 0.12	2.36 ± 0.15

Note.—Data are the MAE $\pm$ the standard error of the MAE.

Table 4
Difference between Volume Measurements of the Same Nodule for Two Observers

Nodule Volume (mm ³)	Mean	Median	Standard Deviation
0–10	–0.63	–0.30	2.39
10–20	–1.36	–0.70	2.97
20–40	1.51	–0.85	3.43
>40	–0.70	–0.10	3.32

Note.—Data are the volume difference in cubic millimeters.