Accuracy and precision of CT angiography in a model of carotid artery bifurcation stenosis.

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Accuracy and Precision of CT Angiography in a Model of Carotid Artery Bifurcation Stenosis

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PURPOSE: To determine optimal acquisition parameters and measurement techniques for CT angiography of the carotid bifurcation. **METHODS:** Anatomic phantoms were created in which the diameter of the carotid artery stenoses ranged from 15% to 95%. Initially, we compared the accuracy of stenosis determination obtained by using various values of section collimation and table pitch. Subsequently, applying the combination of collimation and pitch that yielded the greatest longitudinal coverage without degradation in accuracy, we compared the accuracy of measurements performed with various display algorithms, including axial, magnified axial, maximum intensity projection (MIP), and shaded surface display (SSD) images. Last, we determined the effect on accuracy of varying both window and level settings. The standard of reference for all measurements was considered to be caliper measurements made of the models at the time of their construction. **RESULTS:** CT angiography was highly accurate for determining the percentage of stenosis; the average difference between CT angiographic measurements and the standard of reference was less than 1% for all parameter combinations and measurement techniques. Precision varied among the measurement techniques. Magnified axial images provided more precise measurements than either the MIP or SSD images. Although there was a trend toward improved precision with the use of magnified versus unmagnified axial images and MIP versus SSD images, neither of these comparisons reached statistical significance. Systematic error was produced by changing the level setting from that halfway between the luminal density and vessel wall density. Random error was introduced by using window settings greater than zero. **CONCLUSION:** CT angiography was highly accurate and precise for determining percentage of stenosis. The highest precision was attained by using magnified axial images with the level halfway between luminal density and vessel wall density and with the window set to zero.

Index terms: Arteries, computed tomography; Arteries, stenosis and occlusion; Computed tomography, three-dimensional

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The North American Symptomatic Carotid Endarterectomy Trial (NASCET) (1) and Asymptomatic Carotid Atherosclerosis Study (ACAS) (2) demonstrated the benefit of carotid endarterectomy in patients with high-grade carotid stenosis. Their results have prompted increased interest in the optimal method of imaging patients before carotid endarterectomy. Duplex Doppler sonography is the most common screening test for patients with suspected carotid stenosis. Many practitioners require confirmatory, conventional contrast angiography after carotid Doppler sonography and before carotid endarterectomy. Computed tomographic (CT) angiography has been proposed as a substitute for conventional angiography in the evaluation of carotid stenosis (3, 4). However, most studies of carotid CT angiography published to date consist of relatively small clinical series (5–10).

The accuracy and precision of CT angiography in determinations of carotid stenosis may be affected by the acquisition parameters used

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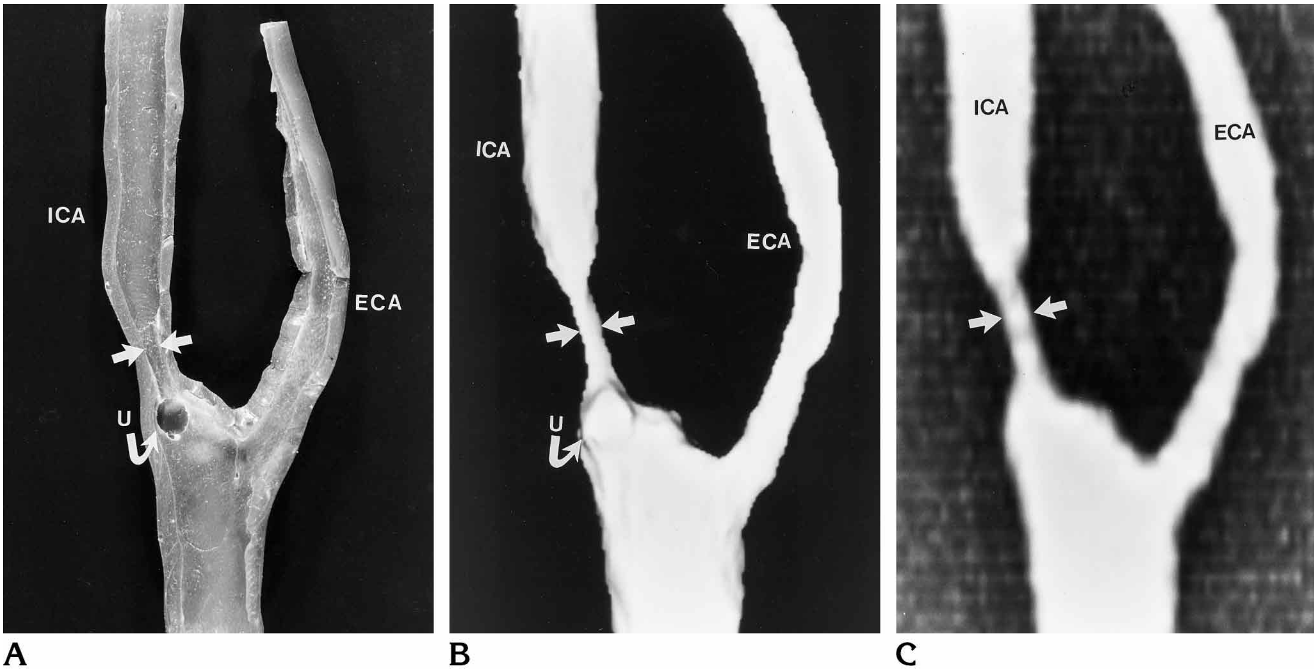

Fig 1. *A*, Cut section shows the endoluminal surface of a representative phantom with a 75% proximal internal carotid artery (*ICA*) stenosis. Minimum residual diameter within the stenotic ICA is denoted by *straight arrows*. The external carotid artery (*ECA*) is also depicted. The common carotid artery is not labeled. A small ulcer (*U*) is present immediately proximal to the ICA stenosis (*curved arrow*).

Shaded surface display (*B*) and maximum intensity projection (*C*) of the same model shown in *A*.

for CT angiography and by the measurement techniques. Optimization of multiple acquisition parameters is difficult in the clinical setting, because either a large number of patients who had been imaged with each parameter would have to be compared or each patient would have to receive unacceptable doses of radiation from multiple scans. In addition, in the clinical setting, the standard of reference is conventional angiography, which is subject to significant interobserver variability (11, 12).

We present the results of a CT angiographic parameter optimization protocol performed with carotid bifurcation phantoms. The use of phantoms precludes the difficulties encountered in clinical CT angiography by allowing multiple scans without exposing patients to radiation and by providing a standard of reference for the percentage of stenosis independent of conventional angiography.

Materials and Methods

Twenty-five anatomic models of carotid stenosis were created with stenoses ranging from 15% to 95%. Because both the severity and length of stenoses are important in the performance of CT angiography, the spectrum of stenosis depicted was critical. Since 70% stenosis was the cutoff for a significant stenosis in the NASCET trials, 10 of the phantoms were created with 60% to 80% stenosis. Nine phantoms had a greater than 70% stenosis, with the remainder in the mild to moderate categories. The approximate length of stenosis in the models ranged from 5 to 10 mm, the typical length seen in our clinical practice. Although shorter, 1- to 2-mm stenoses (webs) are seen, they are not the typical appearance of atherosclerotic disease. In some phantoms, small ulcers or "pits" were created in the internal carotid artery (ICA) adjacent to the stenosis.

Phantoms were constructed as hollow latex tubes depicting the typical geometry of the carotid bifurcation (Fig 1). The luminal diameters in two orthogonal projections were measured with precision calipers at the time of construction at the site of maximum stenosis and in the normal distal ICA. Stenoses were made oval, with the degree of eccentricity determined by the ratio of the maximum diameter divided by the minimum diameter at the point of maximum stenosis. The contrast difference between the vessel lumen and the vessel wall was made equal to 200 Hounsfield units (HU) by filling the phantoms with dilute contrast material to simulate the contrast difference obtained in clinical practice. In our institution, iodinated contrast material (300 mg/mL) is injected at rates of 3 mL/s, so the contrast difference is normally greater than 200 HU.

Multiple spiral CT angiographic scans were made of each phantom. Spiral CT angiography was performed with collimation of 1.5, 2.0, and 3.0 mm. For each collimation, a separate scan was made with a pitch (ratio of table speed to section thickness) of 1:1, 1.25:1, and 1.5:1. The recon-

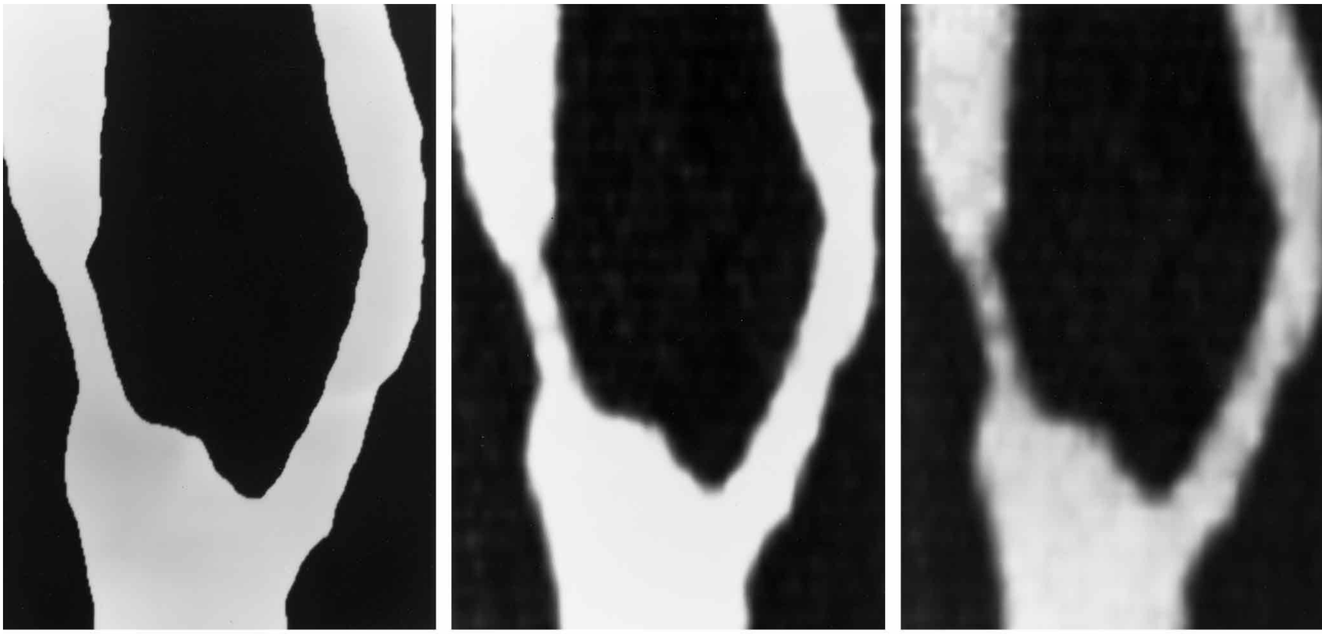

Fig 2. Maximum intensity projection images with various window settings but with level setting held constant.

A, Optimal window setting, with window setting equal to zero. Note distinct vessel wall, which minimizes subjective edge determination.

Window settings equal to 100 HU (*B*) and 200 HU (*C*) show the increase in subjective determination of vessel edge with higher window settings.

struction interval for all scans was 1 mm. All scans were obtained on a Picker PQ 2000 (Cleveland, Ohio) with 140 kV, 200 mAs, 260-mm field of view, and 180° linear interpolation. These are the routine parameters for neck CT at our institution.

All measurements were made by two radiologists who were blinded to the percentage of stenosis introduced into the model at the time of construction. The percentage of stenosis was determined by using NASCET criteria (1). Measurements were made on the computer console and on film, with and without ×8 magnification. The vessel edge was determined subjectively by the readers.

Axial images were analyzed by using orthogonal measurements of the stenosis and of the normal distal ICA. Maximum intensity projection (MIP) and shaded surface display (SSD) images in three projections (anteroposterior, lateral, and 45° oblique) were created and measurements of the percentage of stenosis were obtained. For the projection images, MIP and SSD, the surrounding vessels were removed with a computer-generated cutting tool to prevent overlap with the vessel of interest similar to the clinical situation in which adjacent tissue must be removed before a vessel can be displayed. The window and level settings were varied and the percentage of stenosis remeasured (Figs 2 and 3). For all images, the level initially was set halfway between the density within the vessel wall and vessel lumen. The level was then adjusted above and below this value by increments of 10 HU and the percentage of stenosis remeasured. For axial and MIP images, the window was set to zero to create a binary black and white image that produces a distinct edge to the vessel. The window was then increased in increments of 10 up to 200 HU and the percentage of stenosis remeasured.

The percentage of stenosis obtained for each set of acquisition parameters and each measurement technique was compared with that obtained at the time of construction and the average difference and standard deviation (SD) obtained. Accuracy (the difference in the measured CT angiographic stenosis as compared with the phantom) of the different techniques was compared for statistical significance by use of Student's *t* test. Precision (the variance in the measured CT angiographic stenosis as compared with the phantom) was compared by the F test of variance.

Results

Validation of the Models

The normal distal ICA in the phantoms averaged 6.2 mm ± 0.2. This diameter was not significantly different from that found in a review of 85 patients in whom the normal distal ICA averaged 6.0 mm ± 0.9 ($P > .5$). In these 85 patients the degree of eccentricity of stenosis was 1.40 ± 0.4, while that of the phantoms was 1.43 ± 0.2 ($P > .5$). Noise, as reflected by the SD in HU in a region of interest in an area of uniform density, was similar in the phantoms

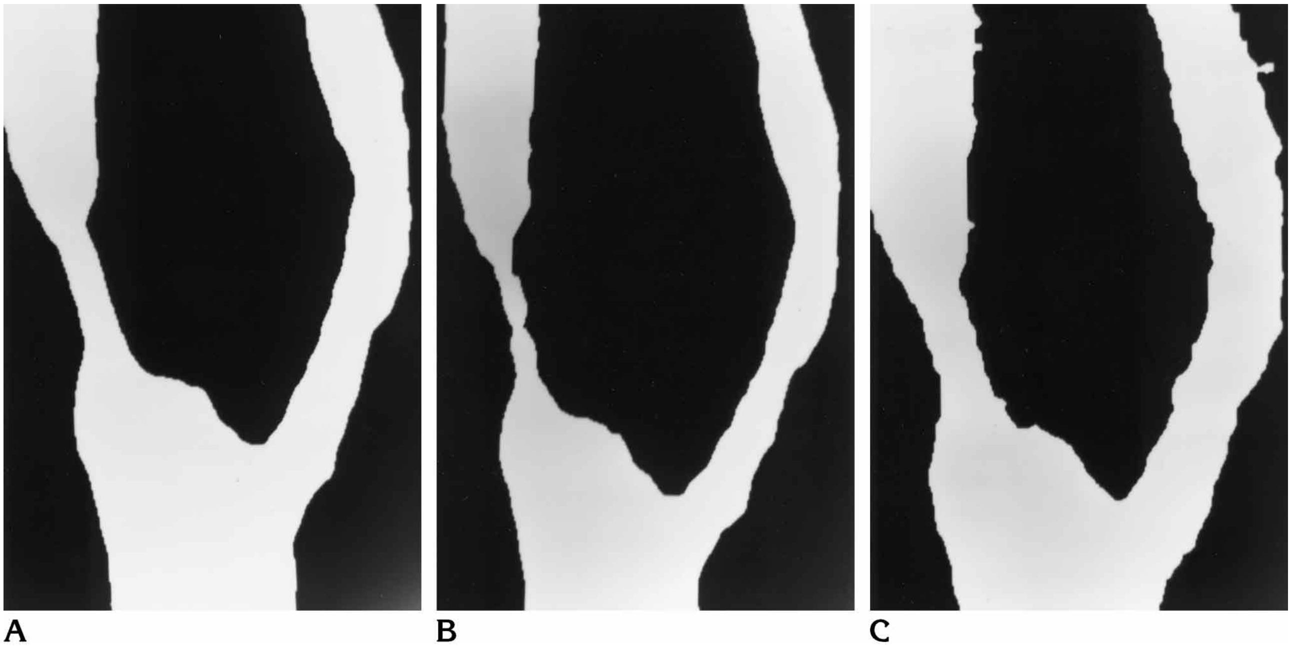

Fig 3. Maximum intensity projection images with various level settings but with window setting held constant.
A, Optimal level setting, halfway between luminal and vessel wall densities.
B, Level setting of 40 HU above that of *A* shows artifactual decrease in residual ICA luminal diameter.
C, Level setting 40 HU below that of *A* shows artifactual increase in residual ICA diameter.

TABLE 1: Effect of collimation and pitch on accuracy and precision of CT angiographic measurement

Section Collimation, mm	Pitch	Average Difference, %	Standard Deviation, %
1.5	1:1	−0.8	3.3
1.5	1.25:1	−1.1	4.0
1.5	1.5:1	−0.6	4.2
2	1:1	−0.9	3.8
2	1.25:1	−0.3	3.9
2	1.5:1	−0.6	4.2
3	1:1	−0.2	3.9
3	1.25:1	−0.8	4.1
3	1.5:1	−0.6	4.4

Note.—Average difference indicates difference in percentage of stenosis between caliper measurements of phantom and CT angiographic measurements.

(SD, 9.0 HU) and in our routine neck CT scans (SD 8.0 HU). No significant difference was found between measurements of the model made at the time of construction and those made after sacrificing the phantoms at the conclusion of the experiment.

Comparison of CT Angiographic Acquisition Parameters

Table 1 presents the average difference and SD of the difference between the measured percentage of stenosis at the time of construction and the measured stenosis on magnified axial images with varied collimation and pitch. No statistical difference was found between the mean (*t* test) and SD (F test) for any of the combinations of pitch or collimation. For the remainder of the study, a comparison of the measurement techniques was made with the data from the scans using a collimation of 3.0 mm and a pitch of 1.5:1, which represented the parameter combination closest to that used in our routine clinical practice.

Comparison of CT Angiographic Measurement Techniques

Table 2 presents the results of the different measurement techniques comparing axial, magnified axial, MIP, and SSD images. The average difference in measurements of the percentage of stenosis for the model was zero, since this was the standard of reference against which the other methods were compared. Repeat measurements of the model gave the SD for this reference standard, an indicator of the variability of measurements obtained with the precision calipers.

Accuracy.—There was no significant difference in accuracy among the various measurement techniques. All measurement techniques

TABLE 2: Comparison of CT angiographic measurements on axial and projection images

Measurement Source	Average Difference, %	Standard Deviation, %
Model	. . .	2.8
Axial ×8 magnification		
Computer console	−0.6	4.4
Film	−0.5	4.9
Unmagnified axial	−0.9	5.4
MIP	0.2	5.6
SSD	0.4	6.8

Note.—Average difference indicates difference in percentage of stenosis between caliper measurements of phantom and CT angiographic measurements; MIP, maximum intensity projection; and SSD, shaded surface display.

yielded an average difference in measured stenosis of less than 1% compared with the reference standard. Our data suggest that projection images tend to underestimate the degree of stenosis, whereas measurements made on axial images tend to overestimate the degree of stenosis.

Precision.—Magnified axial images were more precise than both MIP and SSD images ($P = .045$ and .001, respectively). The magnified axial images on the computer console yielded the highest precision overall, although this did not reach statistical significance ($P = .08$). The difference between MIP and SSD precision was not statistically significant ($P = .09$).

Effect of Window and Level Settings.—The effect of altering the window and level settings is shown in Figures 4 and 5. Increasing the window above zero introduced a random error, manifested as an increase in the SD in measurement without affecting accuracy (Fig 4). This occurred because of the increase in subjective assessment of the edge of the vessel lumen, as the edges are "grayed out" with increasing window. Maintaining the window at zero but altering the level from that halfway between vessel lumen and vessel wall created a systematic error. The vessel was artifactually decreased in size by using higher levels and increased in size with the use of lower levels (Fig 5).

Interobserver and Intraobserver Variability.—Interobserver variability, as reflected by the SD in the two readers' measurements of percentage of stenosis, was 1.2% for narrow window measurements and increased to 8.2% with wide window measurements. Intraobserver variability was similar to interobserver variability, with SDs of 1.1% and 6.7% for narrow and wide windows, respectively.

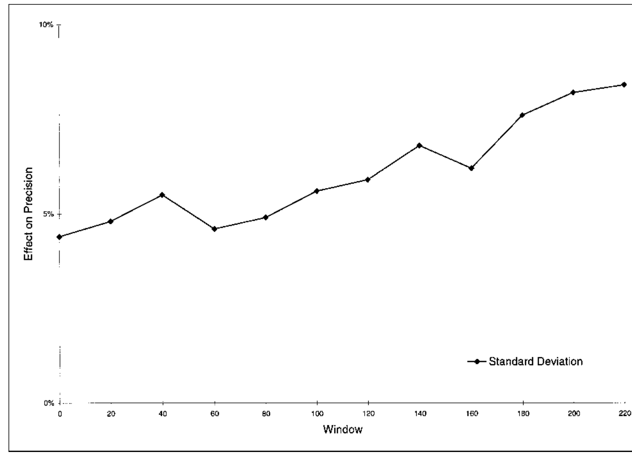

Fig 4. Graph shows the impact of altering window setting on precision. Random error, as reflected in the SD of the average measurement, is introduced with larger window settings.

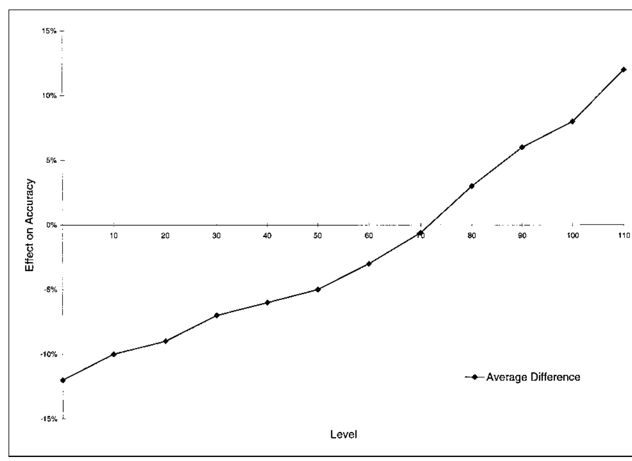

Fig 5. Graph shows the impact of altering level setting on accuracy. Mean error was approximately zero at the level halfway between contrast density of lumen and vessel wall.

The small number of phantoms with an ulcer or "pit" prevented evaluation of the various imaging parameters for the detection of ulcers.

Discussion

Although clinical studies are vital for evaluation of a new technique, it is important for the technique to be optimized before conclusions can be drawn regarding its clinical use (13). Our study attempted to depict as accurately as possible the clinical scenario using an in vitro model. Contrast densities, values of pitch and collimation, and vessel sizes typically encountered in clinical CT angiography were simulated in our phantoms.

We found no statistically significant change in the error of stenosis determination across the range of pitch and collimation in our study. Acquisition parameters were increased from a minimum collimation/pitch combination of 1.5 mm/1:1 to a maximum combination of 3.0 mm/1.5:1 without producing statistically significant decreases in either accuracy or precision. As expected, volume averaging from increased section thickness and pitch caused a small decrease in precision, although this did not reach statistical significance. Therefore, the larger collimation and pitch should be used because they yield greater coverage and/or less radiation exposure without significantly decreasing accuracy. Because our phantoms only modeled typical atherosclerotic plaque, narrower collimation and pitch would be expected to provide significantly better depiction of shorter stenoses, such as the webs typically seen in fibromuscular dysplasia.

In our model, measurements performed on magnified axial images were superior to those made on projection techniques, such as MIP and SSD, primarily because of diminished precision encountered with the latter techniques. We noted slight differences in precision when comparing unmagnified with magnified axial images and when comparing MIP with SSD images, but neither of these differences reached statistical significance. There was no difference in accuracy among the measurement techniques evaluated.

Regardless of measurement technique, alterations in level settings yielded significant changes in systematic error. The effects of variation in level setting can be understood by considering partial volume averaging. If a voxel included portions of both the vessel wall and the vessel lumen, the resultant pixel density would have been a weighted average of these two densities. Our data suggest that, on average, use of a level halfway between the intraluminal density and mural density yields the most accurate determination of vessel diameter. Use of a higher level artifactually decreases the apparent size of the vessel, while use of a lower level artifactually increases the apparent size of the vessel.

Widening the window setting results in an increase in random error, regardless of measurement technique. When the window is set to zero, a binary (black and white) image is created, which reduces variation in the subjective choice of the vessel's edge position. Use of a window setting of greater than zero results in a more subjective edge determination by "graying out" the vessel interfaces.

Radiographic measurement of anatomic structures is dependent on both contrast and resolution. In CT angiography performed with an adequate injection rate and volume, the contrast differences will be high, thereby making resolution the limiting factor. The theoretical vessel size at which accurate measurement begins to fail is approximately twice the pixel size. For carotid CT angiography with a 260-cm field of view and a 512 matrix, the pixel size is 0.51 mm, yielding a lower limit of 1.0 mm at which measurement of the maximum stenosis fails (equal to an 80% stenosis for a 5.0-mm ICA). Measurement of vessel stenosis smaller than this will underestimate residual lumen and overestimate the degree of stenosis. Overestimation of stenoses greater than 80% should not create a clinical problem if the NASCET (>70%) or ACAS (>60%) cut points are used. CT angiographic scans should not falsely suggest occlusion even with high-grade stenosis if contrast is visible in the ICA distal to the stenosis in the cervical carotid artery.

An automated computer measurement of the diameter and area of the vessel would be preferable to the subjective method used, but the software was not available to us. This would remove intraobserver and interobserver variability and reduce total examination time. The level chosen as the threshold for measurement would remain important but the loss of precision associated with wide window settings would be prevented.

Improved precision with magnification is based on pixel size and volume averaging at the vessel edge. If a single pixel covers the vessel edge, half in and half out of the vessel, the pixel density is the mean of the density of the vessel wall and lumen contrast. If two pixels touch at the vessel edge, the correct density is ascribed to each pixel and the measured vessel size is accurate. All combinations of volume averaging between these two extremes produce the variation in the measured vessel size (loss of precision) and an average measurement difference of zero (high accuracy). Magnification without interpolation of the intervening pixels does not change precision. Magnification with linear interpolation produces an intervening pixel with an average density between the two pixels at the vessel edge, reducing the precision error by the magnification factor. The theoretical precision error caused by a pixel size of 0.5 mm in a 5-mm vessel with a 70% stenosis is approximately 5%. This error is reduced to below 1% with ×4 magnification.

Our phantoms did not incorporate the effects of motion. As such, the absolute accuracy data reported here may represent an overestimation compared with that achievable in clinical CT angiography. However, our aim was to compare relative differences among the measurement techniques rather than to determine absolute accuracy. If the effects of motion are relatively similar among the various measurement techniques, then our data are valid.

The effect of intraluminal streaming should not affect the accuracy of CT angiography, as the density of contrast material within the vessel should be uniformly distributed after it is mixed in the heart during the transit from the venous to the arterial system.

Studies have implicated calcification of atherosclerotic plaque as a cause of error in CT angiography (8, 9). High-density tissue, such as calcium adjacent to the vessel lumen, will be volume averaged, causing an apparent increase in vessel size if a level is used that is halfway between the lumen contrast and the lower-density adjacent musculature. We intend to construct additional phantoms in the future that incorporate mural calcification to evaluate this effect more systematically. A possible partial solution involves the use of a different level for determining the vessel edge adjacent to calcification by using a level halfway between the vessel lumen contrast and the density of the calcification. The result is imperfect because atherosclerotic plaque does not calcify uniformly.

Our phantoms were created to simulate the typical carotid bifurcation geometry, in which the carotid artery is only 10° to 20 ° off perpendicular to the CT angiographic section orientation. In patients in whom the carotid artery is more oblique or completely horizontal in the neck, volume averaging would decrease accuracy, systematically overestimating the percentage of stenosis. This effect should be kept in mind when the carotid is more oblique than expected on the CT angiographic study.

Conclusions

CT angiography was highly accurate and precise in our model of carotid stenosis. Parameters currently in use for carotid CT angiography produce highly accurate and precise measurements of the percentage of stenosis. Appropriate choices of window and level settings are critical for accurate and precise measurement of the percentage of stenosis. At least some variability in the published accuracy of carotid CT angiography can be attributed to variability in measurement techniques among studies. Reliance on conventional angiography as the standard of reference in clinical series may also contribute to the variability in the reported accuracy of carotid CT angiography if conventional angiography also has significant error in the measurement of stenosis.

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