Computerized image analysis for quantitative measurement of vessel diameter from cineangiograms

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ABSTRACT Subjective estimates of the angiographic severity of coronary artery stenoses show variability and inaccuracy. We therefore tested the accuracy of a newly developed computerized image analysis system for quantitating vessel diameter from cineangiograms. Fourteen cylindrical phantoms of known diameter were filled with contrast medium and filmed over a wide range of clinically relevant radiographic conditions in order to develop regression equations that related computer-derived to anatomic diameters. Computer measurements of vessel diameter were unaffected by vessel size, magnification, focal spot size, thickness of scattering medium, kilovolt peak, or location within the radiographic field, but a correction factor was necessary for a small but significant (p < .01) linear dependence on contrast medium concentration. The accuracy of computerized vessel diameter measurements ranged between ± 59 and ± 137 μ for all conditions except for rapid vessel motion and contrast medium concentrations of 50% or less meglumine diatrizoate (Renografin 76), both of which resulted in reduced accuracy as well as in the inability to locate lumen edges of vessels less than 1 mm in diameter.


SUBJECTIVE VISUAL ESTIMATES of percent stenosis of a coronary artery from cineangiograms have been shown to be characterized by a large interobserver variability and a discrepancy between angiographic and postmortem estimates of lesion severity has been noted in a number of studies. As a result, attempts have recently been made to quantitate luminal dimensions more precisely. For visual measurements calipers, optical scale devices, and computer-assisted edge tracking have recently been used to attain this goal. Semiautomated and automated edge-tracking techniques applied to digitized cineangiographic images have likewise been recently used in attempts to further objectify measurements of luminal dimensions. Although the reproducibility of various objective methods of diameter measurement has been studied, the accuracy of these measurements has received little attention.

The fact that radiographic vessel images have edge gradients rather than sharply demarcated edges is not widely appreciated by angiographers. Accurate diameter measurement is dependent on precise localization of the anatomic vessel edge within the edge gradient, and the magnitude of error in diameter measurements associated with arbitrary assignment of the vessel edge within the edge gradient may be considerable. A new method for computerized edge detection of coronary arteries from digitized 35 mm cineangiographic images with the use of specific algorithms to locate spatially disparate points within the edge gradient has been developed. The hypothesis that each of these points can be mathematically related to the anatomic edge was tested by filming contrast medium-filled cylindrical phantoms over a wide range of cineangiographic conditions and comparing computer-derived measurements of diameter from digitized frames with the known diameters.
Materials and methods

Cylindrical phantom construction. Three plastic blocks of “female” casts of a series of 14 metal cylinders were made from a rapidly polymerizing monomer base (Batson's corrosion compound No. 17, Polysciences, Warrington, PA). Each block measured approximately \(4 \times 2 \times 1\) cm and contained either the lower, middle, or higher ranges of phantom diameters. As measured with a micrometer caliper, the diameters of the metal cylinders ranged from 0.50 to 6.26 mm and agreed with corresponding measurements of internal diameter to within 10 \(\mu\)m. Exposure of a solid 10 cm block of the polymer along with an equal thickness of water resulted in less than a 5% difference in corresponding film optical densities, which were within the linear portion of the characteristic curve of film. The radiographic attenuation by the casting material was similar, therefore, to water.

Exposures. A clinically relevant range of the following radiographic variables, each of which may potentially effect the magnitude and shape of the edge gradient of a radiographic vessel image, was chosen for study:

1. Vessel diameter: 0.50, 0.71, 0.80, 0.88, 1.23, 1.56, 1.97, 2.34, 2.75, 3.14, 3.52, 3.94, 5.46, and 6.26 mm.
2. Concentration of contrast medium (meglumine diatrizoate, Renografin 76): 25%, 50%, 60%, 70%, 80%, 90%, and 100%.
3. Focal spot size, as measured with a 1° Siemens star test pattern: 0.3 and 1.0 mm.
4. Position and orientation within the x,y plane across the radiographic field.
5. Magnification: 1.05, 1.11, 1.25, 1.40, 1.67, and 2.00.
6. Vessel motion: 0, 10, 25, and 50 cm/sec.
7. Kilovoltage: 70, 80, 90, and 100 kV.
8. Thickness of scattering medium: 10, 15, and 20 cm of water.

All exposures were made with a Philips cinefluoroscopic unit with a 6 inch cesium iodide image intensifier. The unit had an option for use of a microfocus x-ray tube that was used only for studies of focal spot size. Tube-to-object distance was 110 cm and object-to-image intensifier distance was 10 cm for all studies except for those of variable 5, wherein tube-to-object distance was progressively decreased for a stepwise increase in magnification. Pulse width was 5 msec for all exposures; for studies of variables 1 through 7, 10 cm of water, 75 kV (except for variable 7), and between 100 to 200 mA were used to obtain cineradiographs of clinically acceptable quality. For studies of variable 8, voltages of 75, 82, and 97 kV were selected for the three thicknesses of water, while the number of milliamperes was adjusted to obtain an acceptable exposure. Except for in studies for variable 2, 100% meglumine diatrizoate was used. Films were developed in a Jamieson processor at 30 ft/min, with the developer at 29°C, fixer at 26°C, and water at 16°C. Kodak XX film was used for all exposures.

Each of the three blocks of contrast medium-filled phantoms was exposed separately and was centrally positioned so that all 14 phantoms were studied for variables 1 through 3 and 5 through 8. The length of each block was approximately 50% of the diameter of the radiographic field, which was always collimated in a manner identical to that used in the clinical setting. For studies of variable 4 (field position), a 3.14 mm phantom was positioned vertically and horizontally in each position of a field divided into 3 × 3 array.

FIGURE 1. Angiographic images of a contrast medium-filled phantom vessel. A. Magnified view of segment of radiographic image. Note the indistinct edges. B. Within the operator-defined window, x,y coordinates and gray levels of pixels are digitized. C. Pixels corresponding to computer-located edges (see text) are lit up for display.
After each phantom exposure, a 1.0 cm rectilinear grid was positioned at the same distance from the image intensifier as that used for the phantoms in order to obtain a magnification factor.

During cardiac catheterization in four patients, the phantoms were filmed over the cardiac silhouette, above the anterior chest wall, in order to determine the effect of the inhomogeneity in the thickness of chest tissues across the object plane on the accuracy of the vessel diameter measurements.

**Image analysis.** An operator-interactive method for computerized analysis of vessel images from 35 mm cineangiographic film has been developed and described, in part, previously. An optically magnified portion of a cineradiographic frame was digitized by a Vidicon camera/digitizer interfaced to a Vax 11/780 computer. Within the 640 × 480 pixel array available for digitization, a window over a vessel segment of interest was created on a video display by the operator, and transverse camera scans were obtained with a resolution of approximately 15 μ/pixel (figure 1). Since 8 × magnification of a vessel image was present on 35 mm frames, a resolution of approximately 12 pixels per anatomic millimeter was used for each study. The grey level of each pixel was digitized in an eight-bit logarithmic scale (256 levels). For all analyses, scans were performed four times and averaged.

Two different algorithms were used for locating points within the edge gradient of the bell-shaped densitometric scans of the phantom images. For algorithm 1 a seventh-order polynomial was fit to the densitometric scan, which comprised the mean of 10 adjacent transverse camera scans; an inflection point and a base point were derived from the polynomial (figure 2).

A magnification factor was determined by scanning a cineangiographic frame with a 1 cm rectilinear grid. The distance between two grid lines, which were recognized as points of minimum optical density, was measured at the same position in the radiographic field as each phantom of interest. Vessel diameter was determined as the horizontal distance between computer-located edges. For the short (about 0.8 mm) vessel segment studied, corresponding to the 10 adjacent camera scan lines, in order to reduce statistical fluctuations in the measured diameters from the effects of radiographic noise such as those resulting from quantum mottle and film grain, measurements were always repeated at the same film position on eight sequential frames and averaged.

For algorithm 2 the maximum densitometric gradient was automatically found by identifying the maximum slope of a second-degree polynomial, which was fitted to the gray-scale values of a fixed number of consecutive pixels within a moving window across a transverse densitometric vessel scan (figure 3). A similar edge-detection algorithm has been described by Selzer et al. The horizontal distance between the maximum slope points on either side of the vessel image was converted to a diameter measurement by use of the grid-derived magnification factor. In order to reduce statistical fluctuations in diameter measurement as a result of radiographic noise, edges were tracked in an automated manner over a 5 mm length of each vessel segment for each frame studied and a measurement of mean diameter was obtained.

The number of consecutive pixels used within the moving window was found to affect the location of the maximum slope. Therefore, a comparison was made of the maximum slope-derived diameters obtained with 3, 7, 11, 15, 19, and 23 consecutive pixels.

**Statistical analysis.** For each radiographic condition and for each method of locating the vessel edge, a linear regression equation was developed relating known to measured phantom diameters. In order to compare linear regression equations for modifications of a given radiographic variable, analysis of covariance was used to test the null hypothesis that the slopes and intercepts of the equations were the same.

**Results**

**Algorithm 1.** The computer-derived base and inflection point diameters obtained with algorithm 1 were found to be consistently larger and smaller, respectively, than the anatomic diameters for all exposures. A close linear relationship was found between known phantom diameter and each computer-derived diameter (r ≥ .99) for all radiographic conditions. A small deviation from linearity was noted for phantom diameters of less than 1 mm. Second-order polynomial regression equations were therefore developed for vessels less than 1 mm in diameter.

**Algorithm 2.** The computer-derived base and inflection point diameters obtained with algorithm 2 were found to be consistently larger and smaller, respectively, than the anatomic diameters for all exposures. A close linear relationship was found between known phantom diameter and each computer-derived diameter (r ≥ .99) for all radiographic conditions. A small deviation from linearity was noted for phantom diameters of less than 1 mm. Second-order polynomial regression equations were therefore developed for vessels less than 1 mm in diameter.

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*For the data obtained with this algorithm see National Auxiliary Publications Service document No. 04093 for 4 pages of supplementary material. Order from NAPS c/o Microfiche Publications, P.O. Box 515, Grand Central Station, New York, NY 10163. Remit in advance U.S. funds only, $7.75 for photocopies or $4.00 for microfiche. Outside the U.S. and Canada add postage of $4.50 for the first 20 pages and $1.00 for each additional page; $1.50 for microfiche postage.*
For linear regression equations relating computer-derived to known diameters (including diameters less than 1 mm) for each radiographic condition, analysis of covariance demonstrated no change in the slope or in the intercept of each equation (p > .05) when modifications of each of the following radiographic variables were compared: focal spot size, magnification, kilovoltage, and thickness of scattering medium (water).

For the contrast concentration of 25% meglumine diatrizoate and for rapid motion, phantoms with a diameter of less than 1 mm were excluded from analysis because of unacceptable image quality.

Contrast medium concentration had a significant (p < .01) effect on the intercept of both the base and inflection point linear regression equation (figure 4). A linear dependence of the base point intercept (r = .70) and inflection point intercept (r = .91) on concentration was noted, so that base point diameter decreased a mean 270 μm, and inflection point diameter decreased a mean 220 μm, as concentration of contrast medium decreased from 100% to 25% for phantoms greater than 1.0 mm in diameter. No effect of contrast medium concentration on the slope of either the base point or the inflection point regression equation was noted.

Motion resulted in a significant change (p < .01) in the base point regression equation intercept at a maximum velocity of 50 cm/sec, such that the base point at this velocity was found approximately 150 μm outside of its usual location for a 3 mm vessel.

After correction for magnification secondary to pin-cushion distortion, no effect of field position or orientation on either the base point or inflection point diameter of the 3.14 mm phantom could be demonstrated.

Common regression equations relating base and inflection point diameters to known diameters were determined from the data for all radiographic conditions except for those of rapid motion and contrast medium concentration of 25% (figure 5). Overall accuracy of these equations in predicting true diameter from the base and inflection point widths was determined. The mean absolute difference between the computer-derived predicted diameter and the known diameter for all conditions (excluding rapid motion and 25% meglumine diatrizoate concentration) for all phantoms was ±73 and ±74 μm for base and reflection point diameters, respectively, when the background consisted of a homogeneous thickness of water.

The accuracy of the inflection point diameter measurement was reduced to ±103 μm when phantom images were superimposed over the inhomogeneous background of chest tissues. The base point diameter measurement likewise was less accurate for the inhomogeneous background (±137 μm) and was unreliable for the 0.5 mm phantom.

**Algorithm 2.** A linear relationship was found between known diameters and corresponding diameters determined from computer recognition of the maximum slope on either side of the vessel image (r > .99) for each radiographic condition. However, the location of the maximum slope within the edge gradient was markedly affected by the number of pixels included within the moving window across the transverse densitometric vessel scan. Both the slope and the intercept of the linear regression equation relating measured and known diameters were affected by the size of

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the filter, i.e., the number of pixels included in the window. For windows of 9 or more pixels, there was a loss of the ability to discriminate between the size of phantoms less than 1 mm in diameter. Since the use of less than 7 pixels often resulted in spuriously found edges secondary to film grain and quantum mottle, a 7 pixel filter appeared to provide an optimal trade-off between the degree of smoothing of the data and the ability to measure the diameter of vessels of less than 1 mm. Comparison of linear regression equations for various radiographic conditions was, therefore, performed with edges located with a seven-point filter.

Similar to the finding with algorithm 1, the linear regression equation of algorithm 2 was unaffected by focal spot size, magnification, kilovoltage, or thickness of scattering medium. Likewise, no effect of radiographic field position or orientation of maximum slope-derived diameter measurements of the 3.14 mm phantom was found, once magnification was corrected for pincushion distortion. In addition, no effect of motion, including rapid motion, on the regression equation was found.

A linear dependence of the intercept of the regression equation on the concentration of contrast medium was found (r = .98), an observation similar to that for algorithm 1 (figure 4). As the concentration of meglumine diatrizoate was changed from 100% to 25%, the maximum slope diameter decreased a mean 300 μ for phantoms greater than 1.0 mm in diameter.

A common linear regression equation for algorithm 2 was calculated as for algorithm 1 and its accuracy in predicting anatomic diameter calculated for all radiographic conditions except for 25% contrast medium concentration and rapid motion (figure 5). The mean absolute difference between computer-derived and known diameters was ± 59 μ.

The effect of an inhomogeneous background on the accuracy of the diameter measurements, which was decreased ± 108 μ, was similar to that for the inflection point diameter measurements.

Programs for automated edge tracking of a vessel image with any one of the three edge-detection algorithms were developed (figure 6). The center line of the vessel was determined, in an automated fashion, to be the midpoint between vessel edges. A fourth-order polynomial was fit to points along the center line, and
FIGURE 6. Video display of magnified view of a cineangiographic image of a right coronary artery, right anterior oblique view, of a patient during cineangiography. With any one of the three edge-detection algorithms (the maximum slope algorithm in this case), edge tracking of a coronary vessel image may be performed within an operator-defined window (A & B). Vessel diameter is displayed graphically as a function of distance along the center line (C).

Discussion

Cineangiography provides potentially precise information regarding the in vivo luminal dimensions of coronary arteries in man. As a result, angiographic studies have been advocated as a means to evaluate the effects of short- and long-term interventions on the caliber of the atherosclerotic coronary artery lumen. A frequent lack of reproducibility of subjective estimates of percent stenosis has been noted in a number of studies, however, and has provided an impetus for the development of objective methods for quantitating luminal dimensions.

The reliability of these objective methods depends not only on the reproducibility, but also on the accuracy of vessel diameter measurements. The latter has been difficult to study, however, because of the complexity of the radiographic imaging process. The accuracy of diameter measurements from cineangiograms depends on localization of the vessel edge within the edge gradient of the radiographic image. All potential causes of unsharpness in the radiographic vessel image may affect the geometry of the edge gradient and the distance between the vessel edges, perpendicular to each point along the center line, was then displayed graphically (figure 6, C).
hence may affect the localization of the vessel edge within the gradient. Knowledge of the modulation transfer function (MTF) associated with each such cause could theoretically be useful in precisely relating the spatial coordinates of the vessel image to corresponding points within the object plane, and image reconstruction techniques might be applied to recover a sharp image of the vessel edge. However, measurement of the MTF of each of the multiple causes of unsharpness would first be necessary. Measurement of the MTF of the focal spot alone is a major task; not only is the size of the spot potentially important, but its shape and intensity distribution may also affect a vessel image. Moreover, "blooming" of the focal spot may occur when current is increased. Measurement of the MTF of an image intensifier is likewise difficult and MTF differs when measured in the radial and tangential directions. Although the MTF of the cineradiographic film used can be measured, variation in the measurement might result from variation in film processing and developing. In addition, Compton scattering and off-focus radiation as well as kilovoltage, vessel size, and contrast medium concentration may affect radiographic contrast, and, therefore, may potentially affect the size and shape of the edge gradient.

Because of the complexity of the radiographic imaging process, we used empirical methods to locate the vessel edge within the edge gradient. In our automated image analysis system for tracking vessel edges from radiographs, different algorithms were used to locate three spatially disparate points within the edge gradient. The hypothesis that each of the corresponding computer-derived diameters could be used to calculate anatomic diameters was tested by analysis of films of cylindrical phantoms of known dimensions over a wide range of clinically relevant cineradiographic conditions.

Linear regression equations relating computer-derived to known diameters were unchanged for modification in kilovolt peak, magnification, thickness of scattering medium, position and orientation within the radiographic field, or focal spot size. A close linear relationship between vessel size and computer-derived diameter was found for all conditions, although a small deviation from linearity for diameters of less than 1 mm for the base and inflection point diameters was noted; second-degree polynomials were, therefore, developed for these computer-derived diameters.

Concentration of contrast medium had a small but significant (p < .01) effect on all computer-derived diameters. For each of these diameters a linear relationship was found between concentration of contrast medium and measured diameter and this was used to develop a linear correction factor to apply to the regression equations for deviations from 100% concentration of contrast medium. Cinedensitometric methods are being developed by us12, 28 to determine concentration of contrast medium within vessels so that an appropriate linear correction factor may be applied.

Motion blurring resulted in a significant increase in base point diameter only for the maximal velocity of 50 cm/sec. Inflection point and maximum slope diameters were unaffected by motion. Since human coronary arteries may occasionally achieve a velocity of 27 cm/sec for a short period during the cardiac cycle, motion blurring would not significantly affect computer measurements of diameter.

The accuracy of our computer-derived diameter measurements was determined as the mean absolute difference between measured and known values. Despite the potential influence of many radiographic variables on computerized diameter measurements, the overall accuracy of the base point—, inflection point—, and maximum slope—derived diameter measurements was found to be ±73, ±74, and ±59 μ, respectively, over a wide range of clinically relevant radiographic conditions, when the radiographic background was homogeneous.

As anticipated, the inhomogeneity of chest tissues resulted in a slight reduction in the accuracy of the maximum slope— and inflection point—derived diameter measurements (to ±108 and ±103 μ, respectively), and most greatly affected the accuracy of the base point—derived diameter measurements (±137 μ).

It is anticipated that the accuracy of computer measurements of vessel diameter from coronary cineangiograms in the clinical setting will be less than that from the present phantom studies. Instead of a carefully positioned 1 cm rectilinear grid, as in our study, a catheter image is often relied on to obtain a radiographic magnification factor. Diameter measurements of the catheter image are less precise than measurement of a 1 cm grid spacing, and out-of-plane magnification differences between the catheter tip and vessel should be corrected by use of two orthogonal radiographic views when available. A further reduction in accuracy may be expected to arise from inhomogeneous mixing of blood and contrast medium, particularly when coronary flow rates are elevated, such as is commonly noted in patients with aortic valve disease.

Diameter measurement of a coronary vessel segment of interest should be made, of course, perpendicular to the center line at the level of the segment. In
the present image analysis system, a series of mid-
points between vessel edges is found automatically and
a fourth-degree polynomial is fit to the midpoints to
obtain a center line. A scan that is perpendicular to
each point along the center line is then used to compute
diameter as a function of distance along the center line
(figure 6, C). Short (approximately 0.8 mm) vessel
segments were scanned in the present study by averag-
ing 10 adjacent camera scan lines. Should a very short
segment of stenosis, e.g., a shelflike lesion, require
scanning, flexibility within the image analysis pro-
gram allows for averaging any smaller number of adja-
cent scan lines. When a smaller number of adjacent
scans is used, however, the diameter information may
have to be averaged over a greater number of frames in
order to achieve the accuracy achieved with 10 scan
lines. It is possible that variations in luminal geometry,
as a result of lumen encroachment by atherosclerotic plaques, may affect the size and the shape of the radi-
ographic edge gradient and, hence, measurements of
coronary artery diameter from angiograms. Replicate
arterial phantoms of nonaxisymmetric human ather-
sclerotic lumina are currently being used in our labora-
tory to study such potential effects.30 As the accuracy
of measurement of vessel diameter by objective meth-
ods improves, the ability to use the coronary arterio-
gram as a means for evaluating potentially useful inter-
ventions on the atherosclerotic coronary lumen will be
enhanced. In addition, studies of the hemodynamic
significance of coronary lesions may be facilitated. For
example, all epicardial segments of a coronary artery
may be commonly involved in the atherosclerotic pro-
cess, so that a "normal" reference segment may not be
available for measuring percent stenosis. Measure-
ment of the absolute luminal dimension of a seg-
mental narrowing may then become useful in assessing
lesion severity. Such measurements should be highly
accurate because of the potentially marked effect of
subtle changes in diameter on coronary flow.31-33

In this study we have applied a method of computer-
ized image analysis to digitized cineangiographic im-
ages of phantom vessels of precisely known dimen-
sions in order to relate spatially disparate points within
the radiographic edge gradient to the anatomic bound-
ary of the vessels. An objective technique for measur-
ing vessel diameter in an automated fashion, irrespec-
tive of radiographic conditions, has thereby been
developed.

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