# Can the same edge-detection algorithm be applied to on-line and off-line analysis systems? Validation of a new cinefilm-based geometric coronary measurement software

**In the Cardiovascular Measurement System (CMS) the edge-detection algorithm, which was primarily designed for the Philips digital cardiac imaging system (DCI), is applied to cinefilms. Comparative validation of CMS and DCI was performed in vitro and in vivo with intracoronary insertion of stenosis phantoms in anesthetized pigs. The "obstruction diameter" (OD) was measured at the artificial stenoses visualized by angiography with calibration at the isocenter (ISO) and catheter calibration (CATH) and compared with the true phantom diameters. A clinical comparison of OD, reference diameter (RD), and percentage diameter stenosis (DS) was performed on 70 corresponding images from post-PTCA angiograms. In vitro, OD (CMS) yielded an accuracy of 0.18 ± 0.14 mm with 100% (correlation coefficient: $r = 0.97, y = 0.06 + 0.75x$, standard error of estimate [SEE] = 0.09) and 0.19 ± 0.15 mm with 50% contrast ($r = 0.94, y = 0.02 + 0.81x$). OD (DCI) yielded an accuracy of 0.11 ± 0.06 mm with 100% ($r = 0.99, y = -0.03 + 0.91x$, SEE = 0.05) and 0.24 ± 0.13 mm with 50% contrast ($r = 0.94, y = 0.29 + 6.69x$, SEE = 0.12). In vivo, OD (CMS) yielded an accuracy of 0.18 ± 0.23 mm with ISO ($r = 0.89, y = 0.02 + 0.83x$, SEE = 0.22) and 0.26 ± 0.24 mm with CATH ($r = 0.89$, $y = 0.06 + 0.72x$, SEE = 0.19). OD (DCI) yielded an accuracy of 0.08 ± 0.15 mm with ISO ($r = 0.96, y = 0.08 + 0.86x$, SEE = 0.14) and 0.18 ± 0.21 mm with CATH ($r = 0.92, y = 0.09 + 0.76x$, SEE = 0.17). The clinical comparison showed reasonable agreement for OD only ($r = 0.81, y = 0.26 + 0.81x$, SEE = 0.29). Transformation of an edge-detection algorithm from a digital to a cinefilm-based system can lead to impairment of measurement reliability. (Am Heart J 1993;126:312-321.)**

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Cinefilm-based automated geometric measurements still represent the most common approach for the application of quantitative coronary analysis.[1, 2] Advantages of this technology are the accuracy of the calibration technique based on direct measurement of the catheter tip,[3, 4] as well as the opportunity for retrospective analysis in core laboratories where large multicenter trials can be objectively evaluated by independent investigators.[5] Continuous improvement in digital imaging techniques, however, has prompted the development of "filmless" catheterization laboratories with commercially available analytic software packages allowing on-line application of quantitative coronary measurements on digital images during the catheterization procedure.[6] The coexistence of cinefilm-based and digital approaches for quantitative geometric coronary analyses raises the question of whether specific edge-detection algorithms developed for the assessment of coronary dimensions can be applied to both imaging systems without altering the reliability of the measurements.

In the new cinefilm-based Cardiovascular Measurement System (CMS; Medis, Nuenen, The Netherlands), an edge-detection algorithm that was developed primarily for the digital cardiac imaging system (DCI; Philips, Best, The Netherlands) is adapted

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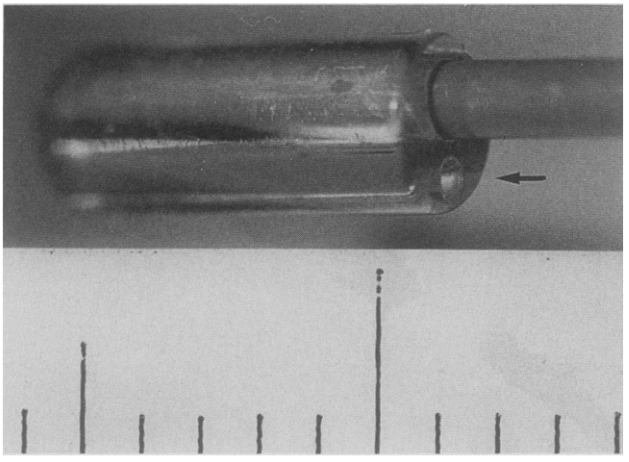

**Fig. 1.** Plexiglass stenosis phantom with eccentric lumen of 0.5 mm (outer diameter 3.0 mm, length 8.4 mm), mounted at tip of 4F Fogarty catheter for percutaneous insertion in a swine coronary artery. Entrance of stenosis channel is marked by *arrow*.

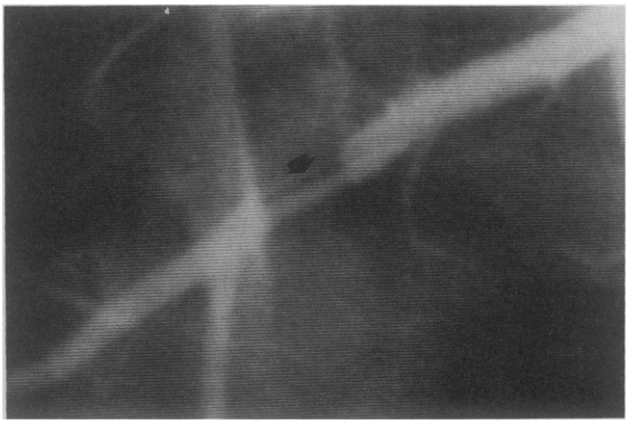

**Fig. 2.** Angiographic visualization of 1.4 mm stenosis phantom *(arrows)* in intracoronary wedge position of left anterior descending artery.

for application in conventional cinefilm imaging.[7] The goal of the present investigation was validation of this new quantitative coronary analysis software both in vitro with a phantom model and in vivo with percutaneous intracoronary insertion of stenosis phantoms in anesthetized pigs. To define the influence of different calibration techniques on the accuracy and variability of in vivo geometric coronary measurements by the new system, analyses with calibration at the radiographic isocenter were compared with those that used the angiographic catheter as a reference. Finally, the CMS and DCI systems were compared during analysis of coronary arteriographic images from patients with coronary artery disease.

## METHODS

### Experimental validation with stenosis phantoms

*Stenosis phantoms.* For in vitro and in vivo validation we used radiolucent cylindrical plexiglass or polymide stenosis phantoms with precision-drilled eccentric circular lumina (tolerance 0.01 mm), 0.5, 0.7, 1.0, 1.4, and 1.9 mm in diameter (Figs. 1 and 2). The outer diameters of the cylinders were 3.0 or 3.5 mm, and the length was 8.4 mm. Acrylate was used to produce the phantoms with small stenosis diameters (0.5 and 0.7 mm), whereas the less fragile polyimide was better suited to the drilling of large stenosis diameters (1.0, 1.4, and 1.9 mm). Parallel to the stenosis lumen a second hole, 1.3 mm in diameter, was drilled in the cylinders to attach them to the tip of 4F Fogarty catheters (Vermed, Neuilly-en-Thelle, France). The central lumina of these catheters contained a removable radiopaque metal wire that was used for intracoronary insertion of the phantoms, as well as for their positioning in the radiographic isocenter during the in vivo experiments.

*In vitro experiments.* The stenosis phantoms were serially inserted into the center of cylindrical plexiglass models with a concentric channel, 3.0 mm and 3.5 mm in diameter. The plexiglass channel including the artificial stenosis was then filled with contrast medium (Iopamidol 370, Schering, Berlin, Germany; 370 mg iodine/ml) at a concentration of either 100% or 50%. Digital and cinefilm acquisition was performed with an additional thickness of plexiglass blocks (12.5 cm anterior and 5 cm posterior to the models) to approximate the density of water. The addition of the plexiglass blocks results in a more appropriate level of kilovolts (75 kV) and a scatter medium that more closely approximates the radiologic scatter in the humen thorax during angiography. On each phantom filled with contrast medium, the measurement of the obstruction diameter was carried out by the DCI system. The studies were then repeated with the second concentration of contrast medium. Subsequently the cinefilms were processed routinely and analyzed off line with the CMS.

*In vivo experiments.* The experimental approach that uses catheter-mounted stenosis phantoms in normal coronary arteries of anesthetized pigs has already been described elsewhere.[3] Again two different calibration methods were applied to both coronary analysis systems: calibration at the isocenter and conventional catheter calibration. By use of these two approaches to calibration, two series of measurements were obtained for both the digital and cinefilm angiographic acquisition systems.

*Image acquisition and processing.* Simultaneous digital and cineangiography procedures were performed at 25 frames/sec. Particular care was taken to minimize foreshortening of the segment of interest and to avoid overlap with other vessels or structures. The 5-inch field mode of the image intensifier (focal spot 0.8 mm) was selected, and the radiographic system settings were kept constant (kVp, mA, x-ray pulse width) in each projection. All phantoms were imaged isocentrically. The digital angiograms were acquired on the Philips DCI system, which employs a ma-

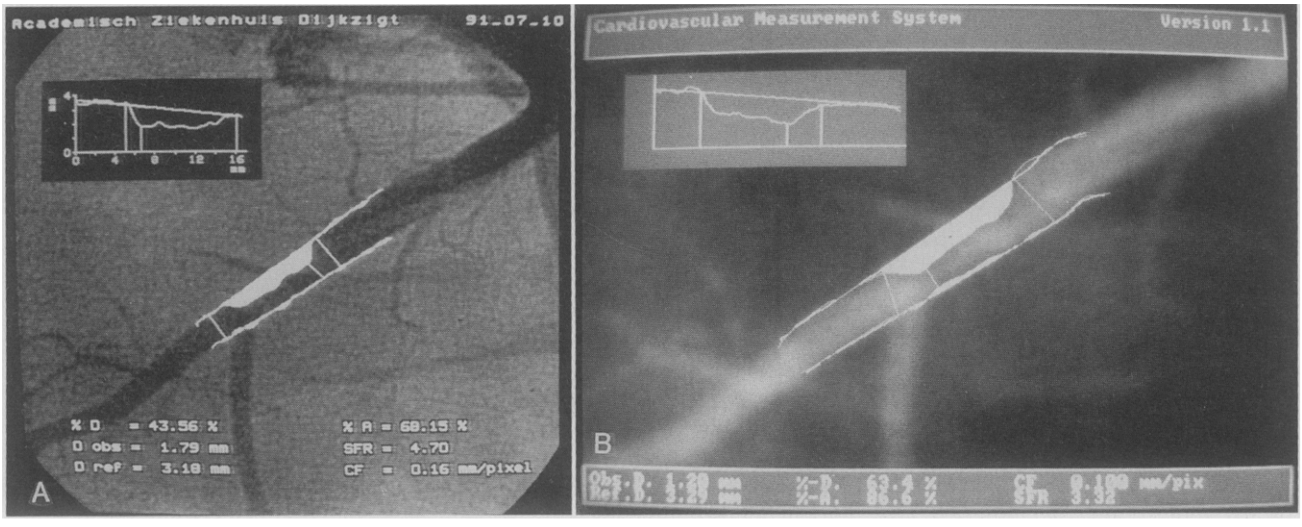


**Fig. 3.** Angiographic image of 1.9 mm stenosis phantom with digital **(A)** and cinefilm **(B)** assessment of obstruction diameter on corresponding end-diastolic frames.

trix size of 512 × 512 pixels. The horizontal pixel size was 200 μm, and the density resolution was 8 bits (256 density levels). The images were stored on a 474 MB Winchester disk, and quantitative analysis of the stenosis phantom was performed on line with the Automated Coronary Analysis (ACA) analytic software package.[6] The corresponding 35 mm cineframes (CFE type 2711, Kodak, Paris, France) were used for off-line analysis with the CMS.[7] This procedure includes recording with a charge coupled device (CCD) camera (pixel matrix 760 horizontal × 576 vertical) with a CAP-35E cinevideo converter (Medis, Nuenen, The Netherlands) and transfer to the analogue-digital converter of the CMS (pixel matrix 512 × 512).

*Edge-detection analysis.* Ten in vitro and 19 corresponding in vivo frames were suitable for measurement of the obstruction diameter at the site of insertion of the stenosis phantoms, both digitally and from cinefilms. A sufficiently long segment of the contrast-filled lumen including the stenosis phantom was selected for quantitative analysis on all images; care was taken to define the same segment length on corresponding digital and cinefilm images. On the DCI system and on the CMS, the user is requested to define only a starting and an end point for the vessel segment, and a centerline through the vessel between these two points is subsequently defined automatically. For both the DCI system and the CMS the basic automated edge-detection technique is identical; it is based on the weighted sum of the first and second derivative functions applied to the brightness profiles along scan lines perpendicular to a model with minimal cost criteria. The algorithm primarily developed for the digital system has been tuned for use on cinefilms with the CMS.[6, 7]

The edge detection is carried out in two iterations and two spatial resolutions. In the first iteration the scan model is the initially detected center line and edge detection takes place at the 512 × 512 matrix resolution. Here the contours detected in the first iteration function as scan models. In the second iteration, a region of interest centered around the defined arterial segment is magnified digitally by a factor of two with bilinear interpolation. On the CMS and the DCI the obstruction diameter is determined as the distance between the two vessel contours at the site of maximum percentage diameter stenosis.

During analysis of the smallest stenosis phantom (0.5 mm), the automatically traced center line was occasionally corrected on the DCI and on the CMS. Manual corrections to the automatically detected contours were found to be unnecessary with either the DCI or CMS, with the site of obstruction diameter in the stenosis phantom being defined satisfactorily by the automatic measurement systems. When a degree of obstruction resulting from cellular material or partial thrombosis was obvious within the phantom channel, the site of obstruction diameter assessment was then user defined. An example of digital and cinefilm measurements of obstruction diameter in a stenosis phantom of 1.9 mm is shown in Fig. 3.

*Assessment of reproducibility.* To assess the variability of repeated obstruction diameter measurements carried out with the CMS, one representative cineangiographic frame of each size of the stenosis phantoms (0.5, 0.7, 1.0, 1.4, and 1.9 mm) was analyzed 15 times by the same operator by means of fully automated software without any user interaction on contours of the artificial lesion and on the site of obstruction diameter assessment.

**Clinical comparison of CMS and DCI measurements.** Post-PTCA angiograms from 31 patients were acquired digitally and on cinefilm and were used for a comparison of geometric coronary measurements at the site of the previous dilation. Parameters of comparison were the absolute measurement value of the obstruction diameter, the reference diameter derived from a computed reference contour, and the relative value of percentage diameter stenosis.

*Coronary angiography, image acquisition, and processing.* In a group of 31 patients who underwent successful PTCA, follow-up coronary angiography was performed after 6 months. Seven-French diagnostic polyurethane catheters (Judkins, Cordis Corp., Miami, Fla.) were used, isosorbide dinitrate (1 to 2 mg) was injected intracoronarily 1 minute before injection of medium contrast to control vasomotor tone, and coronary angiography was performed by manual injection of Iopamidol 370 at 37° C. During coronary angiography, simultaneous digital and cineangiographic acquisition was performed in two projections with the 5-inch field mode of the image intensifier.

The digital angiograms were acquired with the Philips DCI system. The views were selected to minimize foreshortening of the involved coronary segments and to separate them from adjacent structures as much as possible. From each digital angiogram that fulfilled the requirements of image quality for automated quantitation (no superimposition of surrounding structures, no foreshortening of the vessel at the site of the lesion), a homogeneously filled end-diastolic coronary image was selected. Thus, 70 frames of 34 coronary segments were available for on-line quantitative analysis during the catheterization procedure with the Automated Coronary Analysis package of the DCI system.[6] Lesions of the left anterior descending artery were involved in 29 of the 70 frames (41%), lesions of the left circumflex artery in 18 (26%), and lesions of the right coronary artery in 23 frames (33%). The corresponding 35 mm cineframes (CFE type 2711) were visually selected and used for off-line analysis with the CMS system.[7]

*Calibration of the quantitative coronary analysis systems.* Both coronary analysis systems were calibrated by means of measurements of the catheter tip by the automated edge-detection technique resulting in corresponding calibration factors (mm/pixel). In the case of the DCI system the catheter size indicated by the manufacturer was introduced for on-line calibration. In the case of the CMS the nontapering part of the tip of each catheter was measured with a precision micromanometer (No. 293-501, Mitutoyo, Tokyo, Japan) before CMS analysis.

*Assessment of obstruction diameter.* On the 70 corresponding end-diastolic images available for quantitative analysis, the obstruction diameter was assessed digitally and from cinefilm (Fig. 3). Anatomic landmarks (side branches) were used to define the same segment length to be analyzed on corresponding digital and cinefilm images. The algorithm for the determination of obstruction diameter used on the DCI and CMS was described earlier in this report.

*Calculation of reference diameter and percentage diameter stenosis.* On both the CMS and the DCI system an estimation of the normal or predisease arterial size and luminal wall location is obtained on the basis of a second-degree polynomial computed through the diameter values of the proximal and distal portions of the arterial segment followed by the so-called iterative linear regression technique.[6, 11] Tapering of the vessel to account for a decrease in arterial caliber associated with branches is taken care of in these two approaches. The reference diameter (RD) is then taken as the value of the RD function at the location of the minimal luminal diameter (MLD). Percentage diameter stenosis (DS) is calculated from RD and MLD as follows: $DS = (1-MLD/RD) \times 100\%$.

*Statistical analysis.* To validate the CMS, individual values for obstruction diameter obtained by CMS and DCI with both calibration techniques were compared with the true phantom diameters by a paired $t$ test. The mean of the signed differences between phantom diameter values and individual obstruction diameters was considered an index of accuracy and the standard deviation of the differences an index of variability. Corresponding variability values were compared by means of Pitman's test.[12] To assess the agreement between the image acquisition systems, individual differences between the obstruction diameter measured by the CMS and that measured by the DCI system were plotted against the individual mean values according to the statistical approach proposed by Bland and Altman.[13] The standard deviation of the mean value from 15 obstruction diameter measurements on the same angiographic phantom was considered a measure of reproducibility. This value was calculated separately for all five stenosis phantoms. The mean reproducibility was defined as the mean value from those five reproducibility values. For clinical comparison of geometric measurements by means of both systems, individual data for obstruction diameter, reference diameter, and percentage diameter stenosis obtained by DCI and CMS were compared by paired $t$ test. Mean values of the signed differences from the parameters obtained with both acquisition systems including the respective standard deviations were calculated. The individual data acquired with the CMS were plotted against those obtained by the DCI system, and linear regression analysis was applied for each parameter.

## RESULTS

**Assessment of obstruction diameter in vitro.** With the CMS, an accuracy of 0.18 mm and a variability of ±0.14 mm was obtained with 100% contrast medium (Fig. 4, *A*). Linear regression analysis demonstrated a high correlation obstruction diameter and phantom diameter values ($r = 0.97$, $y = 0.06 + 0.75x$, SEE = 0.09). However, the true phantom diameters were significantly underestimated by measurement of obstruction diameter ($p < 0.01$). The corresponding analyses with 50% contrast yielded an accuracy of 0.19 mm and a variability of ±0.15 mm ($r = 0.94$, $y = 0.02 + 0.81x$, SEE = 0.14) but also underestimated the true phantom diameters ($p < 0.01$). The corresponding digital measurements with 100% contrast medium yielded an accuracy of 0.11 mm and a variability of ±0.06 mm with an excellent correlation ($r = 0.99$, $y = -0.03 + 0.91x$, SEE = 0.05), as shown in Fig. 4, *B*. The difference in variability for digital and cinefilm-based measurements was significant ($p < 0.05$). With 50% contrast medium, the accuracy of the digital system was 0.24 mm and the variability

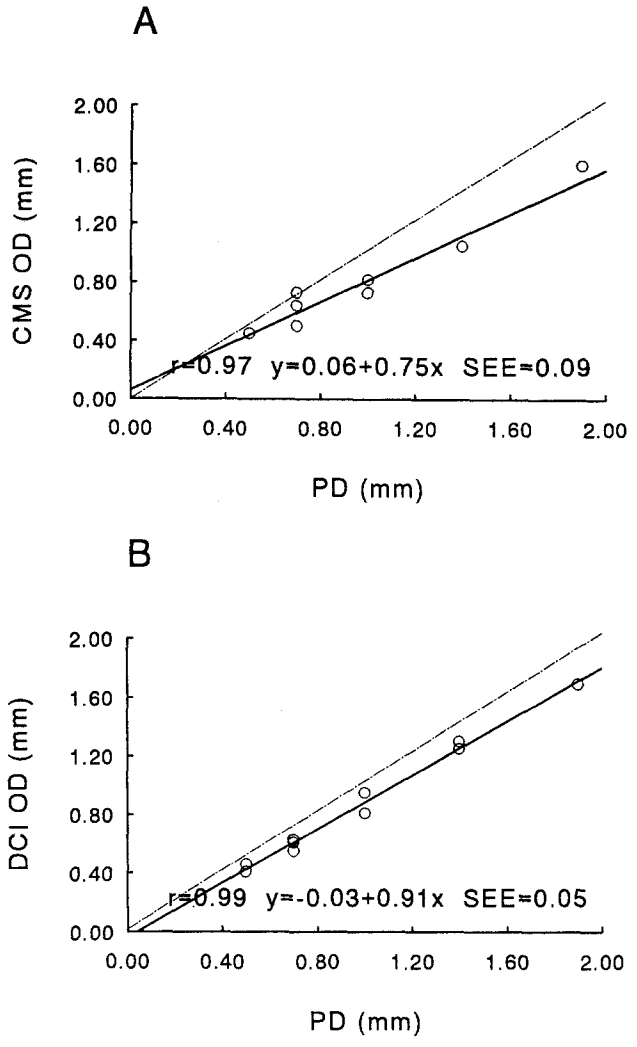


**Fig. 4.** Results of validation with in vitro experiments with 100% contrast medium. **A,** Obstruction diameters *(OD)* obtained by Cardiovascular Measurement System *(CMS)* are plotted against true phantom diameters *(PD)*. **B,** OD values acquired with the Digital Cardiac Imaging System *(DCI)* are plotted against phantom diameters. Graphs include lines of idendity and results of linear regression analyses.

was ±0.13 mm ($r$ = 0.94,y = 0.29 + 0.69x, SEE = 0.12).

**Assessment of obstruction diameter in vivo.** By use of calibration at the isocenter (Fig. 5,*A*), an accuracy of 0.18 mm and a variability of ±0.23 mm was obtained with the CMS. Obstruction diameters and true phantom diameters correlated well ($r$ = 0.89, y = 0.02 + 0.83x, SEE = 0.22), although most of the obstruction diameter values lay below the line of idendity except for the smallest phantom diameter. The underestimation of the true phantom diameter by the CMS measurement was statistically significant ($p < 0.01$). When the calibration was performed on the angiographic catheter, the obstruction diameter measurements by CMS yielded an accuracy of 0.26 mm and a variability of ±0.24 mm. As shown in Fig. 5, *B*, there was good correlation between obstruction diameter measurements and phantom diameter values ($r$ = 0.89, y = 0.06 + 0.72x, SEE = 0.19); however, the degree of underestimation was more pronounced ($p < 0.001$).

The digital measurements of obstruction diameter obtained with calibration at the isocenter yielded an accuracy of 0.08 mm and a variability of ±0.15 mm. Obstruction diameter and phantom diameter values correlated well ($r$ = 0.96, y = 0.08 + 0.86x, SEE = 0.14). Similar to the CMS, an underestimation of the true phantom lumen diameter with the digital approach ($p < 0.05$) was observed. Again this underestimation was more pronounced for the large stenosis phantoms (Fig. 5, *C*). There was significantly less ($p < 0.05$) variability with digital measurements, however, compared with CMS measurements. The corresponding measurements with catheter calibration (Fig. 5, *D*) yielded an accuracy of 0.18 mm and a variability of ±0.21 mm. Although there was good correlation between obstruction diameter measurements and phantom diameter values ($r$ = 0.92, y = 0.09 + 0.76x, SEE = 0.17), a similar degree of underestimation ($p < 0.001$) was demonstrated.

**Comparison between cinefilm and digital measurements in vivo.** A direct comparison between obstruction diameter (OD) CMS and DCI measurements is shown in Fig. 6. The plot of differences from CMS-OD and DCI-OD values versus the mean values from both shows agreement between digital and cinefilm measurements over the whole range of phantom sizes. This holds for calibration at the isocenter (Fig. 6, *A*) and for catheter calibration (Fig. 6, *B*).

**Reproducibility of CMS measurements.** The results of 15 repeated analyses of obstruction diameter for each stenosis phantom are shown in Fig. 7. The variability of measurements was ±0.06 mm for the 0.5 mm and 1.4 mm phantom, ±0.07 mm for the 0.7 mm and 1.9 mm phantom, and ±0.12 mm for the 1.0 mm phantom. Thus the mean reproducibility for all phantom sizes was ±0.08 mm.

**Clinical comparison.** The comparative assessments of obstruction diameter, reference diameter, and percentage diameter stenosis obtained with the CMS and DCI are shown in Fig. 8. Plotted against the digital measurements the majority of data points for obstruction diameters from 70 measurements obtained by the CMS lay below the line of idendity (Fig. 8, *A*). The mean difference and standard deviation from the DCI and CMS were 0.07 mm and 0.31 mm,

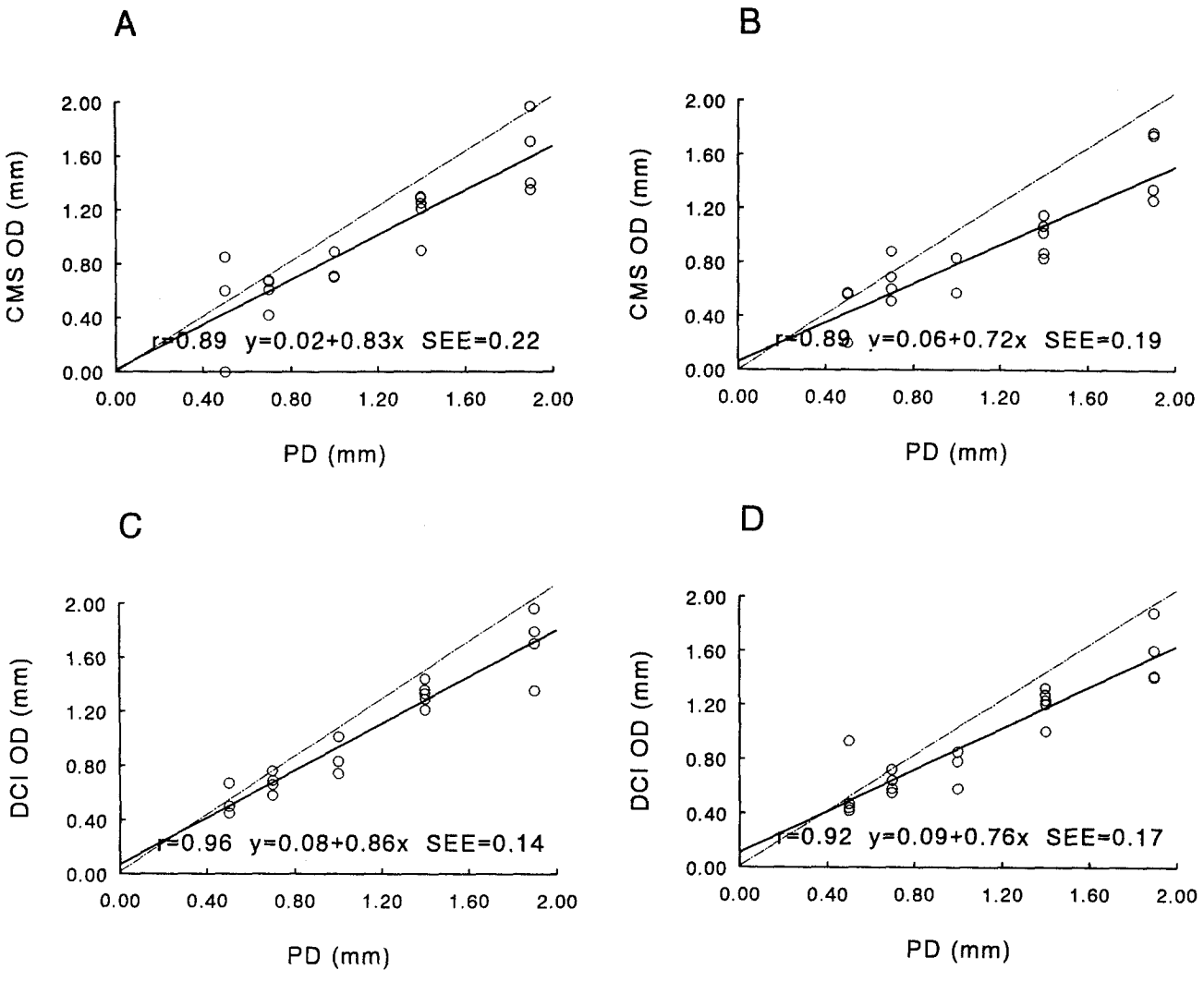


**Fig. 5.** Results of validation with animal experiments. Obstruction diameter values *(OD)* assessed with the Cardiovascular Measurement System *(CMS)* by means of calibration at isocenter **(A)** and catheter calibration **(B)** are plotted against true phantom diameters *(PD)*; corresponding measurement points from Digital Cardiac Imaging System *(DCI)* are plotted in **C** and **D**. Graphs include lines of idendity and the results of linear regression analyses.

respectively. The correlation between both series of measurements was reasonable ($r = 0.81$, $y = 0.26 + 0.81x$, SEE = 0.29), and there was no statistically significant difference. The individual values for reference diameter obtained by the CMS show a higher degree of scatter along the line of idendity when plotted against those obtained by the digital system (Fig. 8, *B*). The mean difference between the DCI system and the CMS was $-0.18 \pm 0.65$ mm. There was a statistically significant overestimation of the reference diameter by the CMS ($p < 0.05$). The correlation between both series of measurements was poor for this parameter ($r = 0.52$, $y = 1.13 + 0.66x$, SEE = 0.62).

A similar low correlation is found for the relative parameter of percentage diameter stenosis, as shown in Figure 8, *C*. The mean difference between the values obtained by the DCI system and the CMS was $-5.14 \pm 14.04\%$. The overestimation of percentage diameter stenosis by the cinefilm-based analysis system was statistically significant ($p < 0.01$). An example of fully automated geometric measurements in both systems after successful PTCA of a stenosis in the proximal right coronary artery is shown in Fig. 9. This example demonstrates that application of the same edge-detection algorithm in corresponding frames from two different imaging systems can lead to different results.

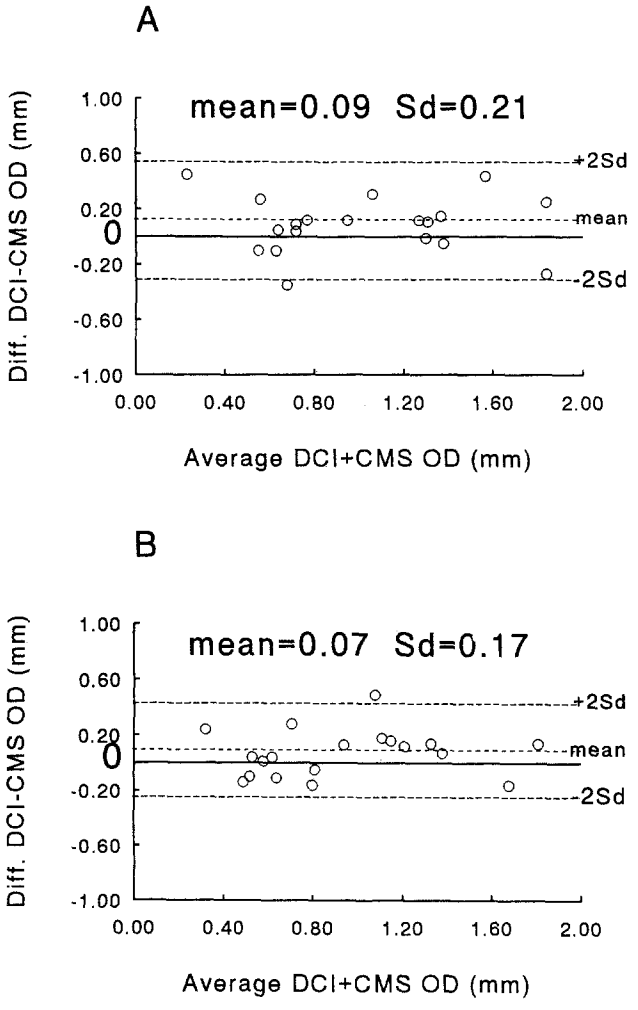


**Fig. 6.** Comparison of digital and cinefilm-based measurements. Plot of differences between digital *(DCI)* and cinefilm *(CMS)* measurements versus mean values from both calibration at isocenter **(A)** and catheter calibration **(B)** with mean difference and twofold standard deviation displayed.

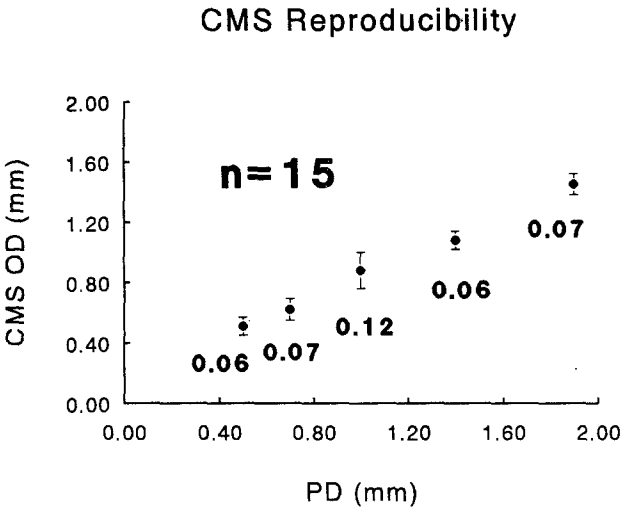


**Fig. 7.** Reproducibility of Cardiovascular Measurement System *(CMS)*. Mean values from 15 measurements of obstruction diameter *(OD)* obtained with cardiovascular measurement system on one representative frame of each size of the stenosis phantoms (0.5, 0.7, 1.0, 1.4, 1.9 mm) are plotted with respective standard deviation as measure of reproducibility.

## DISCUSSION

The development of "filmless" catheterization laboratories is creating a transitional stage during which cinefilm-based systems will coexist with completely digitized facilities. Quantitative geometric measurements, however, will be carried out in both types of catheterization laboratories and will thus be applied to different imaging systems. The present validation compares the same quantitative coronary analysis software but applied to different types of imaging systems with respect to accuracy, variability, and reproducibility both in vitro and in vivo. The software of the new CMS is based on an edge-detection algorithm that was developed for the Automated Coronary Analysis package of the Philips DCI system and was subsequently tuned for application to cinefilms.[6,7] Geometric measurements by the Automated Coronary Analysis package of the DCI system have been validated in a recent study at the Thoraxcenter by means of intracoronary insertion of angiographic stenosis phantoms into an anesthetized swine model.[3] The same experimental approach was used in the present investigation to compare the new cinefilm-based CMS with the DCI system.

**In vitro measurements of stenosis phantoms.** Measurement of obstruction diameter with 100% contrast medium revealed a change of accuracy values from 0.11 to 0.18 mm when the edge-detection algorithm designed for digital images is applied to conventional cineframes. This loss of accuracy is combined with a significant underestimation of true phantom diameters ($p < 0.01$), which is particularly evident with large phantom diameters as illustrated by a decrease in the slope of the regression line from 0.91 to 0.75 in Fig. 4, *B* and *A*, respectively. We also observed an increase in variability from $\pm 0.06$ mm to $\pm 0.14$ mm ($p < 0.05$). With 50% contrast medium, accuracy and variability were similar with both systems, probably because of a higher degree of scatter with both measurement systems. Nevertheless, underestimation of phantom diameters by means of assessments on the cinefilm-based system was again significant ($p < 0.01$).

**In vivo measurements of stenosis phantoms.** The results of these in vitro studies are confirmed by the outcome of our animal experiments in which we serially implanted the same stenosis phantoms into porcine coronary arteries. Calibrated at the radiographic isocenter (corresponding to the in vitro trial), we found a change in accuracy values for obstruction diameter from 0.08 mm to 0.18 mm when the algorithm was applied in cinefilm images and an increase of variability from ±0.15 mm to ±0.23 mm ($p < 0.05$). The underestimation of true phantom diameter values that has already occurred with digital measurements ($p < 0.05$) was more pronounced when the edge-detection algorithm was applied to the corresponding cineframes ($p < 0.01$). When the imaging systems were calibrated on the angiographic catheter, we found a change of accuracy values from 0.18 mm (digital measurements) to 0.26 mm (cinefilm-based measurements), whereas the variability increased from ±0.21 mm to ±0.24 mm. It appears from Fig. 5 that these differences are explained by a higher degree of scatter and a more pronounced underestimation of large phantom diameters.

**Stenosis phantom geometry.** The variable shape of human coronary artery stenoses[14] has prompted the use of noncircular stenosis phantoms for the validation of quantitative coronary angiographic analysis systems.[15] This approach seems to be particularly relevant in measurements of minimal cross-sectional area by densitometry.[16] Cylindrical phantoms, which have been used in our experiments, however, fulfill the requirements for the application of two-dimensional geometric measurements and therefore are eminently satisfactory as surrogates of coronary obstructions.

**Calibration at the isocenter versus catheter calibration.** To be able to compare in vivo results with those obtained from in vitro assessments, we performed geometric measurements with two calibration methods: calibration at the radiographic isocenter, which is used for in vitro settings, and catheter calibration, which represents the calibration technique conventionally used for clinical studies.[17] The use of angiographic catheters for the calibration of quantitative coronary analysis systems may influence the outcome of luminal diameter measurements, because varying catheter composition may result in varying x-ray attenuation[18] and therefore in differences in the automated detection of the contour points. In our in vivo study only one type of catheter was used for calibration, and therefore the influence of different materials on calibration was excluded. Another geometric error is introduced if the planes of calibration and measurement are not identical.[19] This error can be

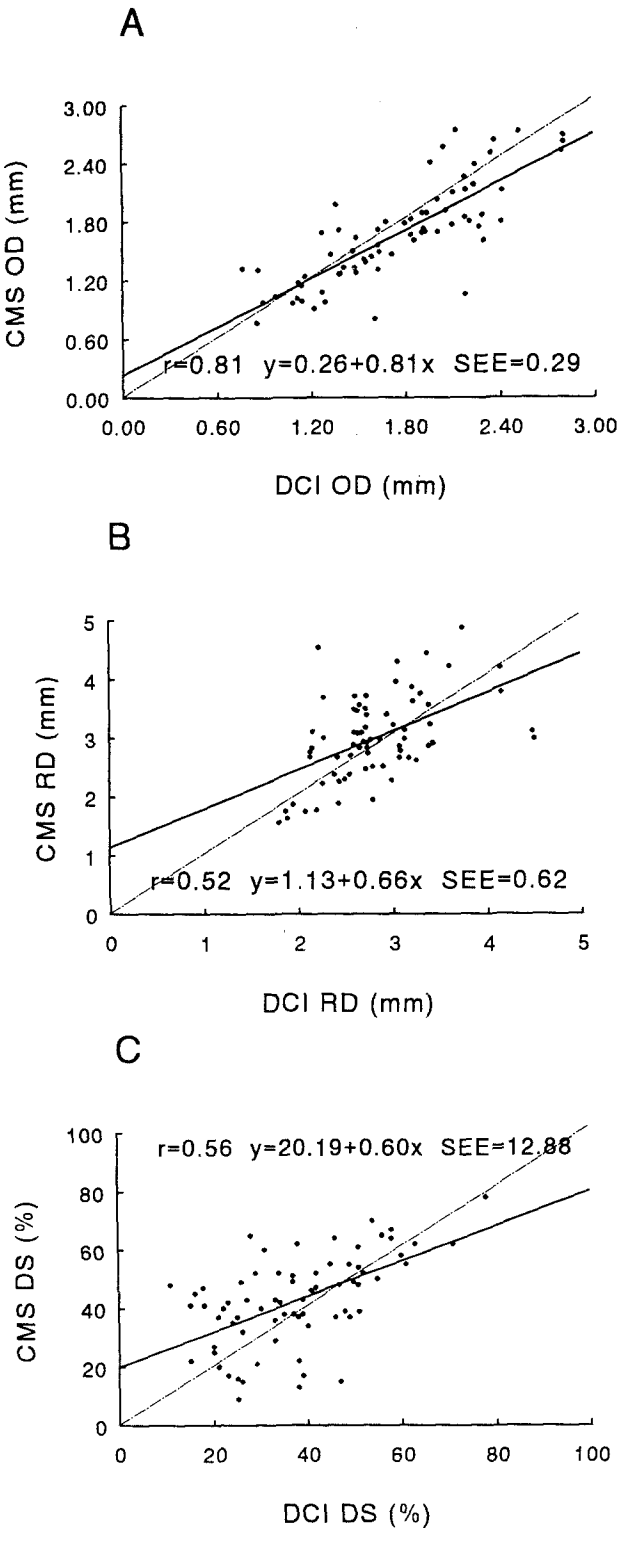


**Fig. 8.** Clinical comparison of digital and cinefilm-based measurements. Digitally acquired values of obstruction diameter *(OD)* **(A)**, reference diameter *(RD)* **(B)**, and percentage diameter stenosis *(DS)* **(C)** are plotted against corresponding values obtained by the cinefilm-based system. Plots include lines of idendity and results of linear regression analyses.

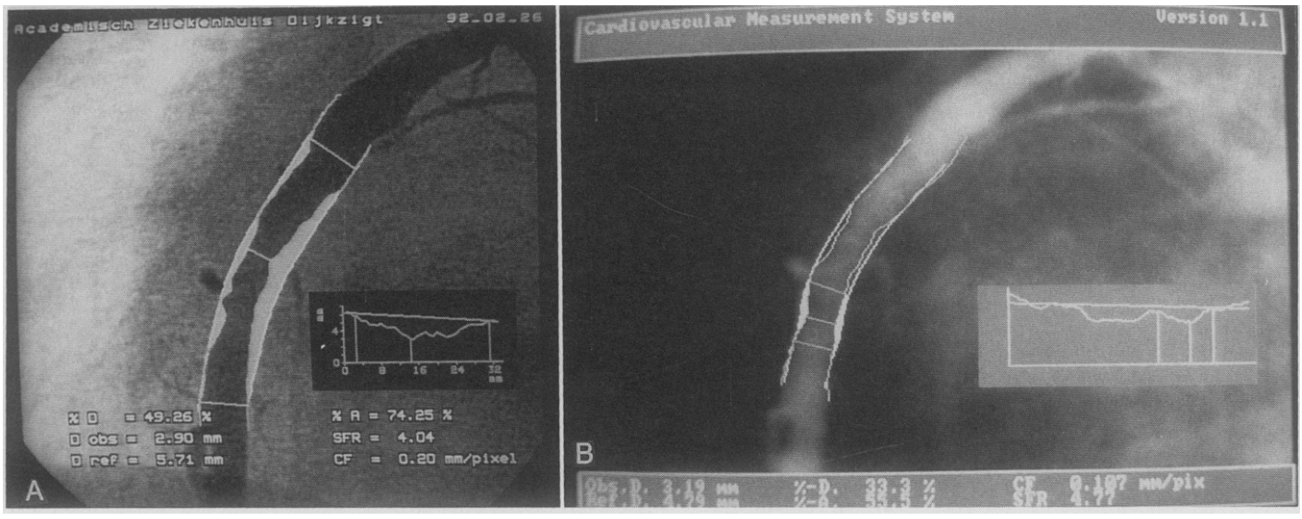


**Fig. 9.** Geometric coronary measurements 6 months after successful PTCA of stenosis in proximal right human coronary artery obtained with digital **(A)** and cinefilm-based **(B)** quantitative measurement system on corresponding end-diastolic images.

circumvented by out-of-plane correction as proposed by Wollschläger et al.[20] or by calibration at the isocenter of the x-ray system. The results of the present study show that in general both digital and cinefilm measurements obtained by catheter calibration are smaller than those obtained by calibration at the isocenter. Theoretically a greater distance between the image intensifier and the catheter tip than between the image intensifier and the isocenter would result in out-of-plane magnification producing smaller calibration factors. A similar effect might have been produced by pincushion distortion for which no correction is made in either system. Both factors could explain the smaller measurements when catheter calibration was applied.

**Gray scale representation and matrix mismatch.** The loss of accuracy and the increase in variability occuring when an edge-detection algorithm is transferred from a digital to a cinefilm-based analysis system may at least in part be explained by differences in the gray scale representation on digital and cinefilm images. If the tuning of an algorithm is guided by simultaneous in vitro and in vivo validation studies, a correction for those differences should be possible. In the case of the CMS, the mismatch between the matrix of the CCD camera (760 H × 576 V) and the analogue-digital converter (512 × 512) might have an additional impact on the outcome of corresponding geometric measurements.

Although the adaptation of an edge-detection algorithm to various imaging systems may impair the accuracy of geometric measurements, direct comparison of DCI and CMS assessments of phantom "obstruction diameters" yielded an acceptable agreement over the range of phantom sizes (Fig. 6). This comparison, however, does not take into account that both systems underestimate true stenosis diameters. In spite of the above-mentioned disadvantages, the adaptation of the edge-detection algorithm from digital to cinefilm-based assessments did not affect the high reproducibility of automated geometric coronary measurements. The reproducibility of obstruction diameter measurements with the CMS ranged from ±0.06 mm to ±0.12 mm, which corresponds to the reproducibility of the digital system.[21, 22]

**Hemorrheologic factors influencing measurements on stenosis phantoms.** In principle, the use of obstruction diameter as the parameter of choice for comparison with true phantom diameters can be criticized. The size of the stenosis channel theoretically could be underestimated if measurements of the automatic edge-detection algorithm are influenced by the presence of cellular debris collected in the phantom lumen during insertion, by the development of microthrombosis, or by the presence of "noise" from the acquisition system. These occurrences may also explain the frequency of underestimation of the true lumen by all techniques.[3, 23] In our experimental study the obstruction diameter was selected for the comparative assessment of the cinefilm and digital system because it represents a nonarbitrary measurement obtained by fully automated analysis of the entire coronary segment and because it is available on both systems.

**Clinical comparison.** Results of our clinical study demonstrate that absolute measurements of obstruc-

tion diameter show the highest correlation when digital and cinefilm-based analyses are compared (Fig. 8, *A*). The extremely low correlation of reference diameters (Fig. 8, *B*), based on a computed reference contour, could theoretically be explained by the same reasons that may be the cause of a loss of measurement accuracy and an increase in measurement variability. Relatively large diameters (reference diameter) should be affected more than relatively small diameters (obstruction diameter) by differences in gray scale representation on digital and cinefilm images and by a mismatch in pixel matrixes between the cinevideo converter and the CMS. Fig. 8 illustrates that the slope of the regression line decreases progressively from *A* to *C*, where assessments of percentage diameter stenosis are plotted. This phenomenon is not surprising because the random error of obstruction diameter and reference diameter measurements is magnified in the assessment of percentage diameter stenosis.

**Conclusions.** The transformation of an edge-detection algorithm from a fully digital to a cinefilm-based system can lead to impairment of measurement accuracy, which is independent of calibration techniques. A significant increase in measurement variability was observed when the acquisition systems were calibrated at the radiographic isocenter. We recommend proper matching of the pixel matrix at the level of cinevideo conversion whenever a system is adapted for quantitative analysis on cinefilms. Tuning of an algorithm for the application on another imaging system should be guided by the result of simultaneous in vitro and in vivo validation studies to guarantee high reliability of automated coronary measurements.

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