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In-Vivo Validation of On-Line and Off-Line Geometric Coronary Measurements Using Insertion of Stenosis Phantoms in Porcine Coronary Arteries

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Geometric coronary artery measurements with the Phillips Digital Cardiac Imaging System (DCI) and the Cardiovascular Angiography Analysis System (CAAS) were validated using percutaneous insertion of radiolucent stenosis phantoms in swine coronary arteries. Angiographic visualization of the stenosis lumens (ϕ 0.5, 0.7, 1.0, 1.4, 1.9 mm) was simultaneously recorded on DCI and cinefilm. The acquisition systems were calibrated by either the diameter of the guiding catheter (catheter CAL) or the isocenter method (isocenter CAL). Minimal luminal diameters (MLD) obtained with CAAS and DCI on 20 corresponding cineframes were compared with the true phantom diameters (PD). The accuracy of MLD measurements with the CAAS using isocenter CAL was −0.07mm, the precision 0.21 mm ($r = 0.91$; $y = 0.30 + 0.79x$; SEE = 0.19), with catheter CAL the accuracy was 0.09 mm, the precision 0.23 mm ($r = 0.89$; $y = 0.19 + 0.74x$; SEE = 0.19). The accuracy of MLD measurements using the DCI with isocenter CAL was 0.08 mm, the precision 0.15 min ($r = 0.96$; $y = 0.08 + 0.86x$; SEE = 0.14), with catheter CAL the accuracy was 0.18 mm, the precision 0.21 mm ($r = 0.92$; $y = 0.09 + 0.76x$; SEE = 0.17). DCI underestimated PD with isocenter CAL ($p < 0.05$) and with catheter CAL ($p < 0.001$). MLD can be measured with high accuracy, both applying on-line digital as well as off-line cineangiographic analysis. The results of digital measurements demonstrate high reliability of the new digital software package.

Key words: Quantitative coronary arteriography, anesthetized pigs, coronary artery disease

INTRODUCTION

The geometric quantification of coronary stenoses plays a deciding role in the evaluation of coronary artery disease. Although the functional significance of an obstructive lesion cannot always be settled from the arteriogram alone [1], quantitative coronary arteriography still remains the most important approach for the assessment of short- and long-term outcome of interventional therapies, as well as for the investigation of progression or regression of coronary heart disease [2].

Measurement of absolute coronary luminal dimensions has been well documented to be more reliable and reproducible than percent diameter stenosis estimations, which rely on the assumption of ''normality'' of a reference contour [3–5]. There is still some uncertainty, however, about the accuracy and precision of computer systems that perform these measurements either from conventional cinefilms or from digitally acquired coronary arteriograms [6–13].

The aim of this study was to compare the new Automated Coronary Analysis analytical software package (ACA) operating on-line on the Philips Digital Cardiac Imaging System (DCI) with the well-established Cardiovascular Angiography Analysis System (CAAS), which is applied to off-line analyses of cinefilms. Geometric coronary luminal measurements obtained by each system were validated in vivo by performing controlled coronary

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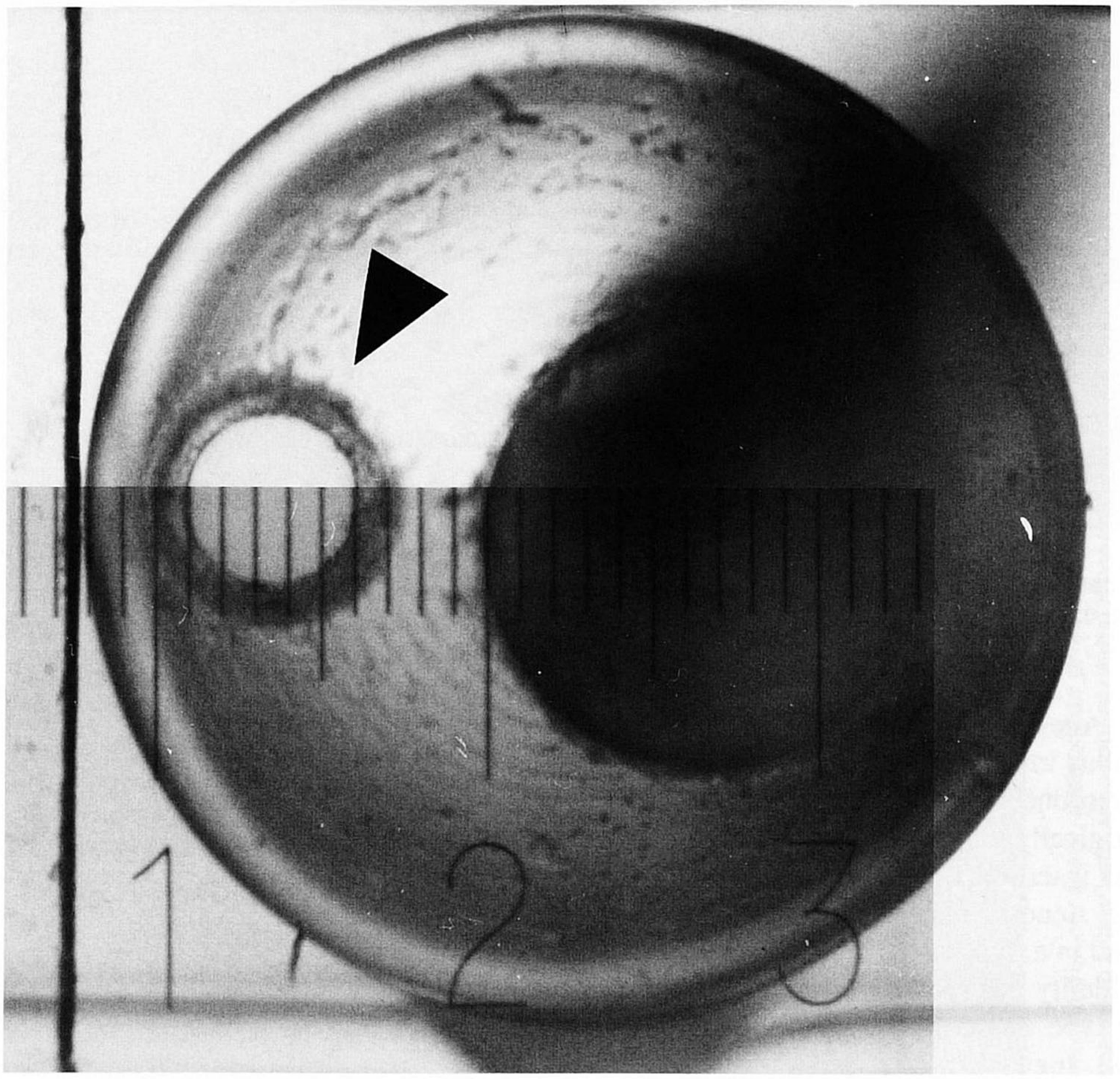

Fig. 1. View at the opening (arrow) of the stenosis channel of a 0.5 mm plexiglass phantom (outer diameter 3.0 mm).

angiography in a domestic swine model with simulated coronary artery stenoses produced by serial percutaneous insertion of graded stenosis "phantoms." In order to investigate the influence of standard calibration techniques on the accuracy and precision of geometric coronary measurements, analyses with calibration carried out at the radiographic isocenter were compared with those using the angiographic catheter for calibration purposes.

METHODS

Stenosis Phantoms

The stenosis phantoms were produced at the Workshop of the Erasmus University Rotterdam and consisted of radiolucent plexiglass (acrylate) or polymide cylinders with precision-drilled eccentric circular lumens (tolerance 0.01 mm) or 0.5, 0.7, 1.0, 1.4, and 1.9 mm in diameter (Fig. 1). The outer diameters of the cylinders were 3.0 or 3.5 mm; the length was 8.4 mm. Acrylate was used to produce the phantoms with small stenosis diameters (0.5, 0.7 mm), whereas the less fragile polyimide was better suited to the drilling of large stenosis diameters (1.0, 1.4, 1.9 mm). Parallel to the stenosis lumen, a second hole of 1.3 mm in diameter was drilled in the cylinders to attach them to the tip of 4 F Fogarty catheters (Vermed, Neuilly en Thelle, France). The central lumens of these catheters contained a removable metal wire, which was used for intracoronary insertion of the phantoms as well as for their positioning in the radiographic isocenter (Fig. 2).

Animal Preparation

We used 4 Yorkshire pigs of average weight, 45–50 kg, which were kept fasting for 8 hr and sedated using intramuscular ketamine (20 mg/kg) and intravenous metomidate (5 mg/kg). The animals were intubated and connected to a Servo-ventilator (Elema, Schönander, Sweden) for volume-assisted ventilation with a mixture of oxygen and nitrous oxide. Ventilator settings were adjusted during the experiments to maintain normal arterial pH (7.35–7.45), pCO2 (35–45 mmHg) and pO2

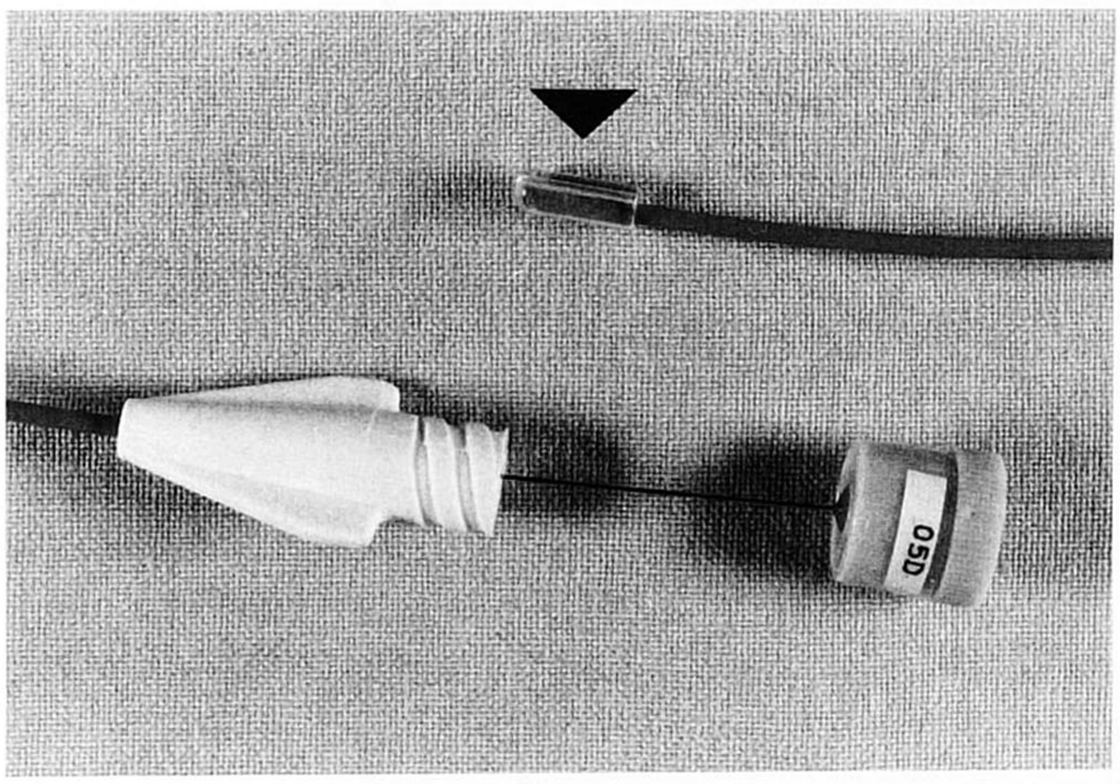

Fig. 2. Phantom catheter with removable metal wire. At the tip of the catheter the 0.5 mm phantom is mounted (arrow).

(>150 mmHg). Anesthesia was maintained with a continuous intravenous infusion of pentobarbital (5–20 mg/kg/h). Valved introducer sheaths (12F: Vygon, Ecouen, France) were surgically placed in both carotid arteries to allow sequential insertion of the angiographic guiding catheter and the stenosis phantoms. An 8F introducer sheath was placed in a femoral artery for the introduction of a 7F high fidelity micromanometer (disposible microtip catheter, type 811/160, Crodis-Sentron, Roden, The Netherlands). Jugular venous access was secured for the administration of medications and fluid. Each animal received an intravenous bolus of acetylsalicylic acid (500 mg) and heparin (10,000 IU) and a continuous infusion of heparin (10,000 IU/h) was maintained throughout the procedure to prevent clot formation.

Calibration of the Quantitative Coronary Analysis Systems

Two different calibration methods were applied to both coronary analysis systems. (1) Calibration at the isocenter: A cylindrical metallic object (drill-bit) of known diameter (3.0 mm) was placed at the isocenter of the X-ray system and recorded both digitally and on cinefilm. For each system the available calibration procedures using automated edge detection were applied to the images obtained, yielding the corresponding calibration factors (mm/pixel). (2) Conventional catheter calibration: The nontapering part of the tip of each 8F polyurethane guiding catheter (El Gamal, Type 4, Schneider, Minneapolis, MN) was measured (diameters of the individual catheters ranging from 2.49 to 2.54 mm) with a precision-micrometer (No. 293-501, Mitutoyo, Tokyo, Japan; accuracy 0.001 mm). The catheter was then introduced into the ascending aorta via the left carotid artery and engaged in the ostium of the left coronary artery. Before injecting contrast medium the catheter tip was flushed with saline and recorded on DCI and cinefilm for subsequent measurement by automated edge detection with each system.

Using these two approaches to calibration, two series of measurements were obtained for both the digital and cinefilm angiographic acquisition system.

Coronary Angiography and Placement of Stenosis Phantoms

After engaging the guiding catheter in the left main coronary artery, isosorbide-dinitrate (1 mg) was administered intracoronarily to control the coronary vasomotor tone prior to the insertion of the phantoms, then a first angiogram was carried out, for orientation purposes. Coronary angiography was performed by ECG-triggered injection of 10 ml iopamidol 370 (Schering, Berlin, Germany; 370 mg iodine/ml) at 37°C with an injection rate of 10 ml/second (rise time = 0) using a pressure injector (Mark V, Medrad, Pittsburgh, PA). To minimize the effect of ventilation on angiographic acquisition, the respirator was disconnected during contrast injection.

The stenosis phantoms were serially wedged in the left anterior descending or left circumflex artery and positioned in the X-ray isocenter using the tip of the metal wire as a marker, which was removed prior to angiography.

Image Acquisition and Processing

Simultaneous digital and cine-angiography was performed at 25 frames per second. Particular care was taken to minimize foreshortening of the segment of interest and to avoid overlap with other vessels or structures.

The 5″-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 matrix 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. From each digital angiogram that fulfilled the requirements of image quality for automated quantitation (no superimposition of surrounding structures, no major vessel branching at the site of the phantom position), a homogenously filled end diastolic coronary image was selected and quantitative analysis of the stenosis phantom was performed on-line (Fig. 3) with the new Automated Coronary Analysis (ACA) analytical software package [14].

The corresponding 35-mm cineframes (CFE Type 2711, Kodak, Paris, France) were used for off-line anal-

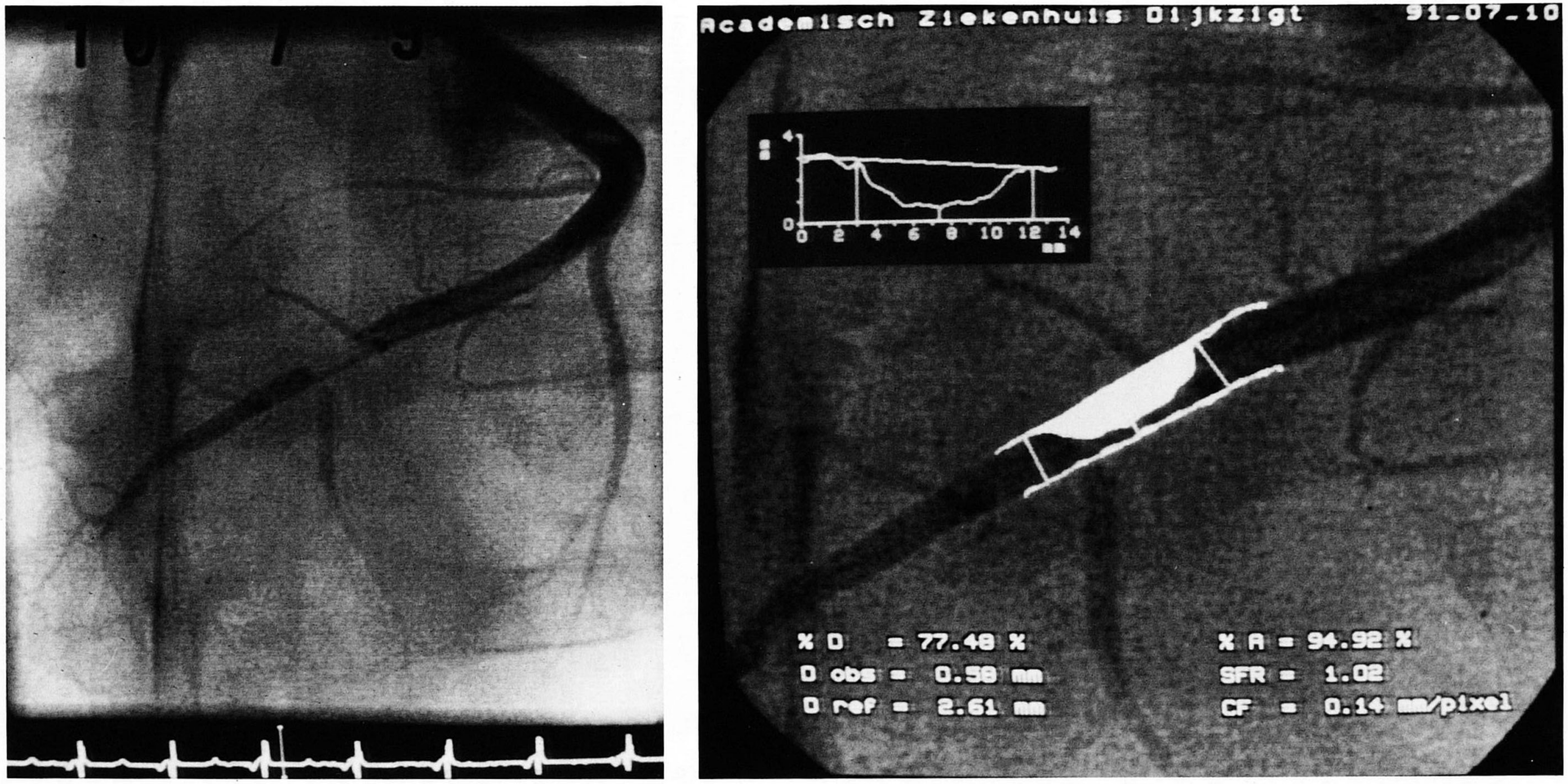

Fig. 3. Angiographic visualization of the artificial coronary obstruction produced by a 0.7 mm stenosis phantom in the left anterior descending artery with subsequent digital measurement of MLD.

ysis with the CAAS system [15]. This procedure allows the digital selection of a 6.9 × 6.9 mm region-of-interest (ROI) out of the 18 × 24 mm cineframe for digitization into a 512 × 512 pixel matrix using a CCD camera (8 bits = 256 density levels). Effectively, this means that the entire cineframe of size 18 × 24 mm can be digitized at a resolution of 1,329 × 1,772 pixels. A correction for pincushion distortion was applied in the CAAS system.

Measurement of Minimal Luminal Diameter

Twenty corresponding end diastolic frames were suitable for measurement of the minimal luminal diameter of the stenosis phantoms both digitally and from cinefilm. A sufficiently long segment of the artery 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 CAAS system the user defines a number of centerline points within the arterial segment, which are subsequently connected by straight lines, serving as a first approximation of the vessel centerline. On the DCI system the user is requested to define only a start and an end point of the vessel segment, and a centerline through the vessel between these two points is subsequently defined automatically. On both the DCI system and CAAS the basic automated edge detection techniques are similar; they are based on the first and second derivative functions applied to the brightness profiles along scanlines perpendicular to a model using minimal cost criteria [14, 15].

With CAAS, the edge detection algorithm is carried out in two iterations. First, the model is the initially defined centerline and, second, the model is a recomputed centerline, determined automatically as the midline of the contour positions, which were detected in the first iteration.

With DCI, the edge detection algorithm is also carried out in two iterations and two spatial resolutions. In the first iteration the scan model is the initially detected centerline 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 ROI centered around the defined arterial segment is digitally magnified by a factor of two with bilinear interpolation. Furthermore, the edge detection algorithm is modified to correct for the limited resolution of the entire X-ray imaging chain [14]. This allows a more accurate determination of vessel sizes less than 1.2 mm diameter.

We took occasional advantage of the opportunity to correct the automatically traced centerline on the DCI during the analysis of the smallest stenosis phantom (0.5 mm). Manual corrections to the automatically detected contours were found, in general, to be unnecessary, either with DCI, or CAAS, with the site of minimal luminal diameter in the stenosis phantom being defined satisfactorily by the automatic measurement systems. When a degree of obstruction due to cellular material or partial thrombosis was obvious within the phantom channel the site of MLD-assessment was then user-defined. An example of digital and cinefilm measurements of minimal luminal diameter in a stenosis phantom of 1.9 mm is shown in Figure 4.

Statistical Analysis

Using both calibration methods (calibration at the isocenter, catheter calibration), the individual data for minimal luminal diameter obtained by CAAS and DCI were compared with the true phantom diameters by a t-test for paired values. The mean of the signed differences between individual minimal luminal diameter and phantom diameter values was considered an index of accuracy and the standard deviation of the differences an index of precision. The minimal luminal diameter values acquired with both systems (CAAS, DCI) and both calibration methods were plotted against the phantom diameter values and a linear regression analysis was applied. Minimal luminal diameter values obtained by CAAS and DCI with both calibration methods were similarly compared using a linear regression analysis. To assess the agreement between the image acquisition systems the individual differences between the minimal luminal diameter measured by CAAS and the minimal luminal diameter measured by DCI were plotted against the individual mean values according to the statistical approach proposed by Bland and Altmann [16]. The precision of the minimal luminal diameter measurements obtained by the two different calibration methods were compared, for both CAAS and DCI, using Pitman's test [17].

RESULTS

The individual minimal luminal diameter measurements obtained by a CAAS and DCI using the calibration at the isocenter are listed in Table IA. The mean phantom diameter was 1.12 mm; the mean minimal luminal diameter measured by CAAS was 1.19 mm and by DCI 1.04 mm.

The measurements of minimal luminal diameter (MLD) obtained with each system using catheter calibration are listed in Table IB. The mean minimal luminal diameter was 1.05 mm for the CAAS and 0.96 mm for the DCI system.

Cinefilm Assessment of Minimal Luminal Diameter with Calibration at the Isocenter

The accuracy of minimal luminal diameter measurements using the CAAS system with calibration at the isocenter was −0.07 mm, the precision 0.21 mm. The

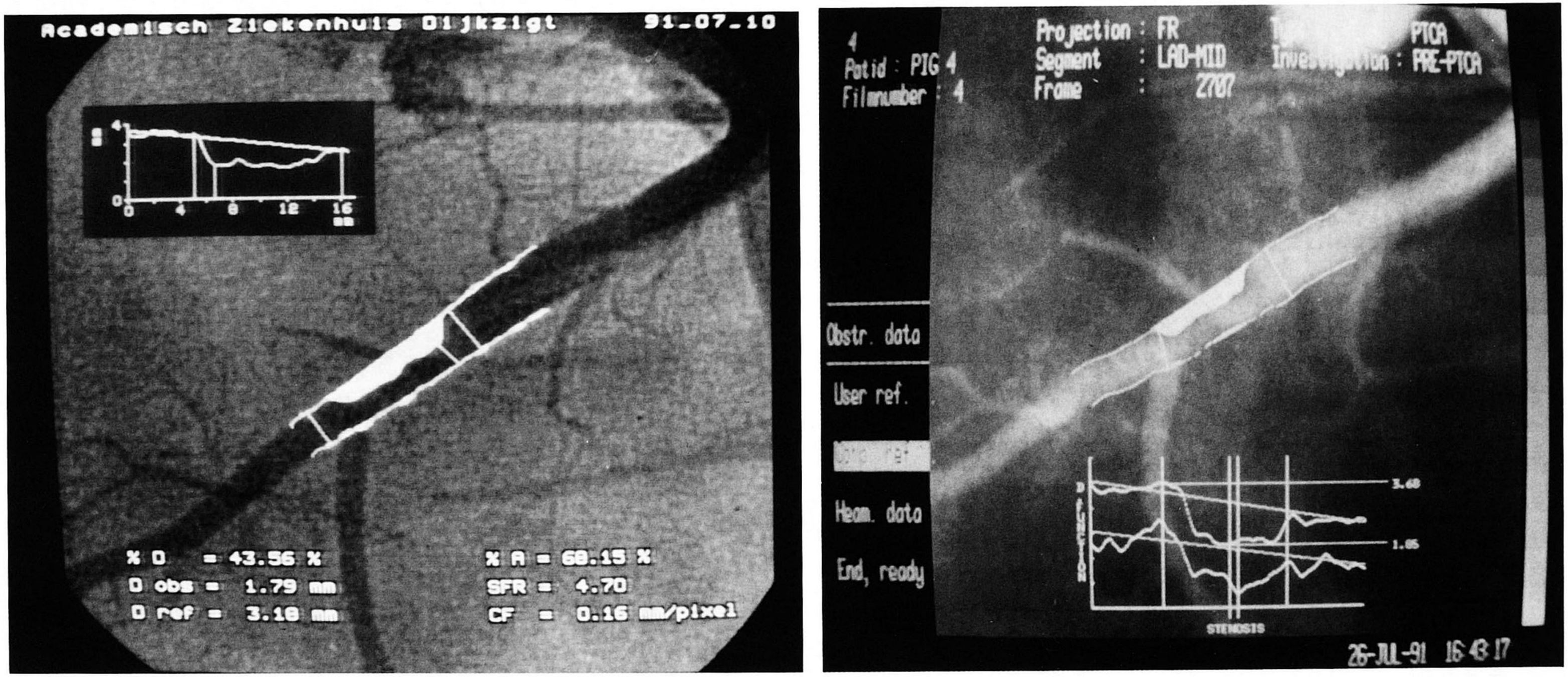

Fig. 4. Angiographic image of a 1.9 mm stenosis phantom with digital (left) and cinefilm (right) assessment of MLD on corresponding enddiastolic frames.

TABLE I. True Phantom Diameters (PD) Listed with Minimal Luminal Diameters (MLD) Obtained by the Cardiovascular Angiography Analysis System (CAAS)—CAAS MLD—and the MLDs Assessed by the Digital Cardiac Imaging System (DCI)—DCI MLD—Including Differences Between True Diameters and Measurement Values

NB	PD (mm)	CAAS MLD (mm)	Difference PD—CAAS MLD (mm)	DCI MLD (mm)	Difference PC—DCI MLD (mm)
A. The phantom diameter (PD) measured with the CAAS- and DCI-system					calibration at the isocenter
1	1.4	1.14	0.26	1.21	0.19
2	0.7	0.70	0.00	0.76	−0.06
3	0.5	0.94	−0.44	0.67	−0.17
4	1.9	2.03	−0.13	1.96	−0.06
5	1.9	1.82	0.08	1.70	0.20
6	1.4	1.36	0.04	1.33	0.07
7	1.4	1.31	0.09	1.36	0.04
8	1.0	1.05	−0.05	1.01	−0.01
9	1.0	0.92	0.08	0.83	0.17
10	0.7	0.81	−0.11	0.66	0.04
11	0.7	0.79	−0.09	0.58	0.12
12	0.5	0.65	−0.15	0.45	0.05
13	0.5	0.69	−0.19	0.50	0.00
14	1.9	1.85	0.05	1.79	0.11
15	1.4	1.66	−0.26	1.44	−0.04
16	1.0	0.88	0.12	0.74	0.26
17	0.7	0.75	−0.05	0.69	0.01
18	0.5	1.20	−0.70	0.50	0.00
19	1.9	1.90	−0.00	1.35	0.55
20	1.4	1.42	−0.02	1.29	0.11
		n.s.			
			$p<0.05$		
Mean	1.12	1.19	−0.07	1.04	0.08
Sd			0.21		0.15
B. The phantom diameter (PD) measured with the CAAS- and DCI-system					catheter calibration
1	1.4	1.18	0.22	1.00	0.4
2	0.7	0.57	0.13	0.72	−0.02
3	0.5	0.67	−0.17	0.93	−0.43
4	1.9	1.95	−0.05	1.60	0.3
5	1.9	1.86	0.04	1.88	0.02
6	1.4	1.16	0.24	1.27	0.13
7	1.4	1.17	0.23	1.20	0.2
8	1.0	0.93	0.07	0.85	0.15
9	1.0	0.79	0.21	0.78	0.22
10	0.7	0.70	0.00	0.55	0.15
11	0.7	0.79	−0.09	0.58	0.12
12	0.5	0.45	0.05	0.44	0.06
13	0.5	0.57	−0.07	0.47	0.03
14	1.9	1.51	0.39	1.41	0.49
15	1.4	1.42	−0.02	1.32	0.08
16	1.0	0.79	0.21	0.58	0.42
17	0.7	0.63	0.07	0.64	0.06
18	0.5	1.16	−0.66	0.42	0.08
19	1.9	1.45	0.45	1.40	0.5
20	1.4	1.33	0.07	1.23	0.17
		n.s.			
			$p<0.001$		
Mean	1.12	1.05	0.09	0.96	0.18
Sd			0.23		0.21

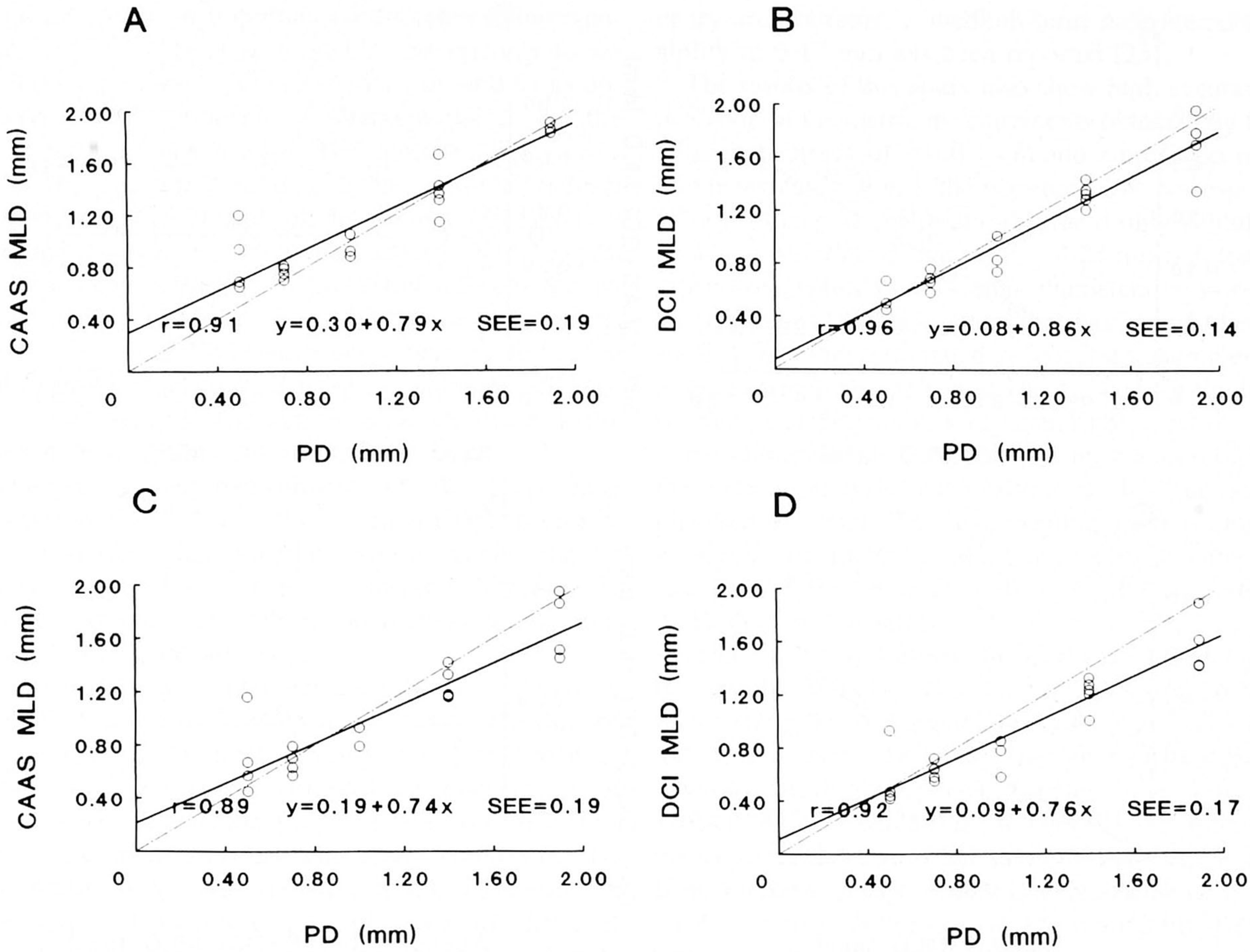

Fig. 5. Cinefilm (A) and digital (B) assessment of MLD with calibration at the isocenter in comparison to cinefilm (C) and digital (D) assessment of MLD using catheter calibration with linear regression analyses and lines of identity.

results of a linear regression analysis are depicted in Figure 5A (correlation coefficient: $r = 0.91$, $y = 0.30 + 0.79x$, standard error of estimate: $SEE = 0.19$). Plotted against the true phantom diameters, the minimal luminal diameter values obtained by CAAS lay close to the line of identity except for the smallest phantom diameter, where a nonsignificant trend towards overestimation was observed.

Digital Assessment of Minimal Luminal Diameter with Calibration at the Isocenter

The digital measurements of minimal luminal diameter obtained with calibration at the isocenter yielded an accuracy of 0.08 mm and a precision of 0.15 mm. The values of minimal luminal diameter and phantom diameter correlated well as illustrated by Figure 5B ($r = 0.96$; $y = 0.08 + 0.86x$, $SEE = 0.14$). However, a paired t-test revealed significant underestimation of the true phantom lumen diameter using the digital assessment of minimal luminal diameter ($p < 0.05$), which was more pronounced for the larger stenosis diameters.

Cinefilm Assessment of Minimal Luminal Diameter with Catheter Calibration

Using catheter calibration the measurements of minimal luminal diameter by CAAS gave an accuracy of 0.09 mm and a precision of 0.23 mm. Again, there was good correlation between the values of minimal luminal diameter and phantom diameter ($r = 0.89$; $y = 0.19 + 0.74x$, $SEE = 0.19$), although as with calibration at the isocenter a non-significant trend towards overestimation was observed for smaller phantom sizes (Fig. 5C). The measurement precision using this approach to calibration was similar to calibration at the isocenter.

Digital Assessment of Minimal Luminal Diameter with Catheter Calibration

The digital measurements of minimal luminal diameter using the DCI system with the calibration performed

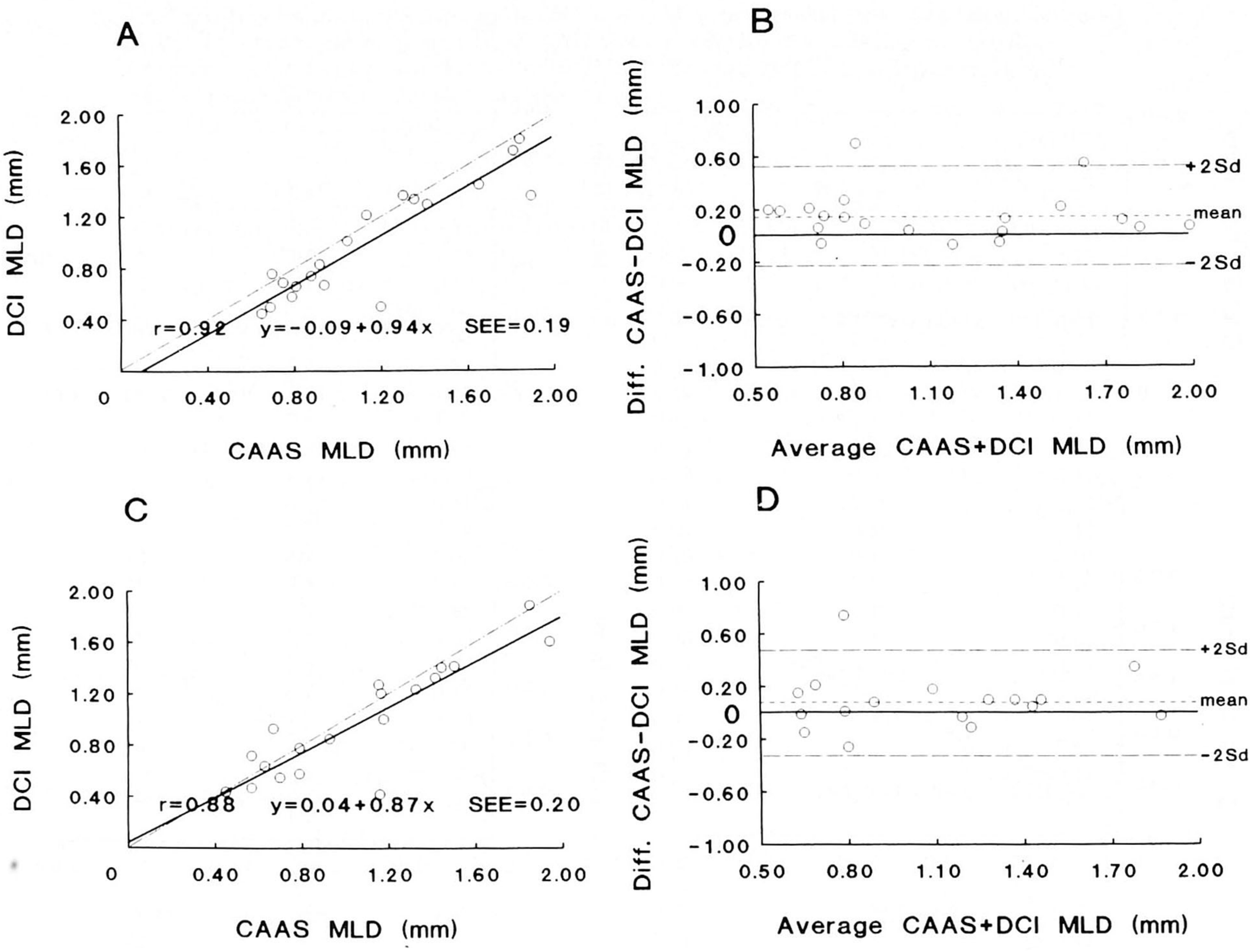

Fig. 6. Comparison between digital and cinefilm measurements of minimal luminal diameter (MLD) using calibration at the isocenter (A, B) and catheter calibration (C, D). Left: plots of digital (DCI) against cinefilm (CAAS) measurements with the linear regression analyses and lines of identity. Right: plots of differences between the MLD measurements acquired by the two systems vs. means of the measurements, with the mean difference and 2-fold standard deviation displayed.

on the catheter yielded an accuracy of 0.18 mm and a precision of 0.21 mm. Although there was good correlation ($r = 0.92$, $y = 0.09 + 0.76x$, $SEE = 0.17$) between minimal luminal diameter measurements and phantom diameter values (Fig. 5D), the t-test for paired values again showed a significant underestimation of true stenosis phantom diameters ($p < 0.001$) as was the case with calibration at the isocenter. The differences in precision between both calibration methods were not significant (Pitman's test).

Comparison Between Digital and Cinefilm Measurements

A direct comparison between DCI and CAAS measurements is shown in Figure 6. As demonstrated, there was good correlation between both measurements using calibration at the isocenter ($r = 0.92$, $y = -0.09 + 0.94x$, $SEE = 0.19$) and catheter calibration ($r = 0.88$, $y = 0.04 + 0.87x$, $SEE = 0.20$), depicted in A and C, respectively, of Figure 6. The plot of differences between CAAS-MLD and DCI-MLD values versus the mean values from both shows satisfactory agreement between digital and cinefilm measurements over the whole range of phantom sizes. This holds for calibration at the isocenter (Fig. 6B) as well as for catheter calibration (Fig. 6D).

DISCUSSION

Quantitative coronary arteriography, originally designed as an off-line cinefilm analysis technique on the Cardiovascular Angiography Analysis System (CAAS), has recently been adapted for on-line use with the Digital Cardiac Imaging System (DCI). The latter approach is

expected to make an important contribution to interventional cardiology, because it enables the operator to assess the size of interventional devices as well as to objectively define the result of interventions during the catheterization procedure [6]. The variable shape of human coronary artery stenoses [18] has prompted the use of noncircular stenosis phantoms for the validation of quantitative coronary angiographic analysis systems [9]. This approach seems to be particularly relevant for the measurement of minimal cross sectional area by densitometry [19]. Cylindric phantoms, however, fulfill the requirements for the application of two-dimensional geometric measurements and therefore are eminently satisfactory as surrogate of coronary obstructions.

In the present study two calibration methods have been investigated. Calibration at the isocenter [20] is normally used for in vitro phantom trials, so our results may be directly compared with these. Catheter calibration, in contrast, represents the calibration method conventionally used in clinical studies [21].

The use of angiographic catheters for the calibration of quantitative coronary analysis systems may influence the outcome of minimal luminal diameter measurements. Varying catheter composition may result in varying X-ray attenuation [22] and therefore in differences in the automated detection of the contour points. In our study only one type of catheter was used for calibration and therefore the influence of different materials on calibration was excluded. A further geometric error is introduced if the planes of calibration and measurement are not identical [20]. This error can be circumvented by out of plane correction as proposed by Wollschläger [23], or by calibration at the isocenter of the X-ray system.

The results of our study show that, in general, the values of both digital and cinefilm measurement with catheter calibration are smaller than with calibration at the isocenter. Theoretically, a greater distance between image intensifier and catheter tip than between image intensifier and isocenter would result in out-of-plane magnification producing smaller calibration factors. This could explain the smaller values of measurements when catheter calibration was applied.

Validation in vitro of minimal luminal diameter assessments has already been performed with CAAS and the DCI system. Reiber et al. found an overall accuracy of -0.03 mm and a precision of 0.09 mm for the measurement of minimal luminal diameter from plexiglass phantoms using CAAS [15]. The variability of measurements from clinical cineangiograms was 0.10 mm, whereas the medium-term variability in an angiographic follow-up was 0.22 mm [7]. In vitro phantom studies assessing the DCI system yielded an accuracy of -0.02 mm and a precision of 0.09 mm [24]. From digital coronary arteriograms, a medium-term measurement variability of 0.17 mm has been reported [25].

The results of this study also show high accuracy and precision of geometric measurements obtained by CAAS with an accuracy of -0.07 mm and a precision of 0.21 mm using calibration at the isocenter. The corresponding values for catheter calibration differed only slightly (accuracy = 0.09 mm; precision = 0.23 mm). A tendency toward overestimation of small diameters was observed and represents a phenomenon that has already been described for other automated coronary measurement systems, in which no correction was applied for the limited resolution of the entire X-ray chain [10].

In comparison to the cinefilm determination of MLD, the digital analysis underestimated the true stenosis phantom diameter. This underestimation was shown to be significant for the calibration at the isocenter (0.08 mm; $p < 0.05$) as well as for the catheter calibration (0.18 mm; $p < 0.001$).

From Figure 5, it is also apparent that, particularly for the smaller stenosis dimensions, the digital measurements using the Automated Coronary Analysis Package (ACA) are very close to the true phantom dimensions, whereas CAAS clearly overestimates these dimensions. This is probably due to the ACA-package correcting for the limited resolution of the entire X-ray imaging chain. If such a correction procedure is not carried out, as on the CAAS, overestimations occur which are particularly apparent for the sizes below about 1.0 mm.

The data from this study clearly show the great advantage of the newer approach, which represents a novel contribution to the field of quantitative coronary angiography where obstruction dimensions in the range of 0.5–1.5 mm are important. The reason why the larger lumen dimensions of phantoms are underestimated with the digital system may be an overcorrection for the limited resolution of the X-ray imaging chain. In addition, the ACA-package does not correct for pincushion distortion, which is especially relevant to catheter calibration technique, where the catheter image may inadvertantly be slightly magnified due to the distortion at the periphery of the image field. Since the catheter is used as a calibration device, it is clear that structures imaged at locations where there is less distortion (such as at the phantom positions) will be measured as being smaller than they really are.

The linear regression analysis of digital measurements where calibration at the isocenter had been performed yielded the highest correlation with true stenosis phantom diameters as well as the smallest standard error of the estimate, implying that the ACA package provides highly reliable geometric measurements.

Comparing digital and cinefilm assessments in terms of the different calibration methods, it should be pointed

out that the mean difference of the cinefilm measurements changes from −0.07 (calibration at the isocenter) to +0.09 (catheter calibration), whereas the mean difference of the digital measurements changes from 0.08 (calibration at the isocenter) to 0.18 (catheter calibration). Taking these differences into account, a minor influence of catheter calibration on the accuracy of digital measurements can be assumed. In contrast, the actual digital and cinefilm measurements demonstrate that conventional catheter calibration introduces additional variability, which is most pronounced for the digital measurements, although the difference in variabilities between the calibration methods was not shown to be significant (Pitman's test). It appears that a more radiopaque structure (the drill bit) gives rise to less variation in calibration factors, and thus in stenosis measurements.

The somewhat lower accuracy and precision values of our in-vivo results in comparison to the findings of in vitro phantom studies can be explained by the influence of radiographic inhomogeneity of surrounding tissue (beam scattering) as well as by motion blurr. This latter disturbing factor was reduced to a minimum, as we selected end diastolic frames and interrupted ventilation during contrast injection. It is possible that microthrombi may have formed within the phantoms making an additional contribution to the measurement variability.

In principle, the use of minimal luminal diameter as the parameter of choice for comparison with true phantom stenosis diameter can be criticized. The size of the stenosis channel theoretically could be underestimated if the automatic edge detection algorithm is influenced by the presence of cellular debris collected in the phantom lumen during insertion or by the development of microthrombosis. These occurrences may also explain the frequency of underestimation of the true phantom lumen by all techniques. In our study, the minimal luminal diameter has been 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.

With respect to the calibration technique as used in clinical practice, it must be taken into account that on-line assessment of coronary dimensions is not compatible with the measurement of catheter tips using a micrometer prior to the angiographic procedure unless such a measurement could be carried out under sterile conditions. On-line calibration using the catheter sizes indicated by the manufacturer would interfere with the accuracy of digital coronary measurements because of the well known variability of true catheter diameters from that indicated on the package. This is more pronounced with nylon than woven dacron catheters [26].

In conclusion, the automated measurement of obstruction diameters in coronary vessels can be performed with a high degree of accuracy both on-line from digitally acquired images and off-line from cineangiograms. Superior results are obtained when systems are calibrated using a well defined structure at the radiographic isocenter. Conventional catheter calibration results in a slightly lower level of precision. The new software technology for the digital assessment of geometric coronary dimensions provides highly reliable measurements.

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