AAPM REPORT NO. 15

PERFORMANCE EVALUATION AND QUALITY ASSURANCE IN DIGITAL SUBTRACTION ANGIOGRAPHY

Published by the American Institute of Physics
for the American Association of Physicists in Medicine

AAPM REPORT NO. 15

PERFORMANCE EVALUATION AND QUALITY ASSURANCE IN DIGITAL SUBTRACTION ANGIOGRAPHY

A REPORT OF THE DIGITAL
RADIOGRAPHY/FLUOROGRAPHY TASK GROUP
DIAGNOSTIC X-RAY IMAGING COMMITTEE
AMERICAN ASSOCIATION OF
PHYSICISTS IN MEDICINE

Maynard High, Chairman
Gerald Cohen
Pei-Jan Paul Lin
Arthur Olson
Lawrence Rothenberg
Lois Rutz
Gerald Shapiro
Chorng-Gang Shaw
Louis Wagner

May 1985

Published for the
American Association of Physicists in Medicine
by the American Institute of Physics

Further copies of this report may be obtained from

Executive Secretary
American Association of Physicists in Medicine
335 E. 45 Street
New York, NY 10017

Library of Congress Catalog Card Number: 85-72829
International Standard Book Number: 0-88318-482-6
International Standard Serial Number: 0271-7344

Published by the American Institute of Physics, Inc.,
335 East 45 Street, New York, New York 10017

Printed in the United States of America

TABLE OF CONTENTS

I. INTRODUCTION

Radiographic systems capable of digital subtraction angiography (DSA) are now to be found in most medium-sized and large hospitals, as well as in private offices and outpatient clinics. At present there is no standard method of evaluating these units, nor is there much guidance from the manufacturers regarding quality assurance. The Digital Radiography/Fluorography Task Group has prepared this report with the following goals; (A) define a set of performance parameters that relate to DSA image quality, (B) describe the components of a digital subtraction system and the way in which they contribute to image quality, and (C) recommend a method of measuring the defined performance parameters so they can be evaluated, optimized and monitored in a quality assurance program.

The Task Group feels that the simplest and clinically most meaningful way of measuring and monitoring the performance parameters is by us of a properly designed non-invasive test phantom. Although it is possible to evaluate the individual components of the system (such as focal spot, image intensifier, TV chain, etc.), and indeed it is advised that one should do this in a proper acceptance test, it is actually the final image from a chain of interrelated components that is used in the clinical diagnosis. A phantom that simulates the clinical situation allows meaningful evaluation of the DSA system without the time-consuming disassembly needed for evaluation of the individual components in the imaging chain. This report does not recommend one specific phantom design, but rather a rational method of measuring performance parameters. A certain phantom design will be described for the purpose of illustrating the concepts of this report; however, any phantom which can adequately measure the performance parameters described in this report is equally acceptable. The user must always be aware that, for meaningful comparisons of DSA systems, for comparisons with manufacturers' claims, or for verification of compliance with bid specifications, the same phantom must be used, and in the same way. The Task Group feels strongly that the manufacturers of DSA systems must include a properly designed phantom with each unit. This should be specified in the purchase document.

II. DESCRIPTION OF SYSTEM AND PERFORMANCE PARAMETERS

A. Digital Subtraction Angiography Systems

Commercial DSA systems come in many forms with varying features, but they all share a common approach: subtraction of a mask image (background) from an image obtained at a slightly different time that contains a contrast signal due to vascular injection of iodine contrast material. The visual contrast of the subtracted

image is enhanced in DSA because the background is subtracted away and the remaining digitized signal can be enhanced through filtering, averaging, and windowing. Many modes of image acquisition are available, the most common being pulse mode, in which a high tube current (100-1000 mA) tine-type x-ray pulsed image is collected at the rate of few frames/second during injection of contrast material and subtracted from a stored mask obtained just before injection. Also common is continuous x-ray mode, in which a lower tube current (5-50 mA) remains on continuously during the study, high frame-rate (3Of/s) image collection is used, and a number of frames are averaged before subtraction to decrease quantum noise. Also available is a fluoroscopy mode, in which contrast images can be 'obtained at the rate of 30 frames/second by subtraction of subsequent frames from previous frames, sometimes with averaging or weighted averaging. Some DSA systems can be operated in more than one, or all, of the above modes.

In this report we will concentrate on simple mask-mode time subtraction DSA methods and will not address the other developing methods such as multiple energy subtraction, "hybrid" subtraction, matched filtering, and recursive filtering. It is likely that these systems can also be evaluated by the methods described in this report, although some additional modules would have to be developed for the phantom to allow for the dynamic change in contrast required for these systems to operate.

Figures 1 & 2 are block diagrams of typical DSA systems. Most systems employ an image intensifier (II) as the x-ray detector, although there are systems with solid-state detection assemblies. The x-ray and fluoroscopy system is quite standard; however, a very-low-noise TV camera is often employed, sometimes with a progressive-scan or slow-scan readout. A remotely controlled aperture is often placed before the TV camera since, in the pulsed mode, the luminance at the II output phosphor may be 1000 times higher than that in the fluoroscopy mode, and the dynamic range of the camera is not large enough to handle this range of light intensities.

The video signal from the TV camera is digitized in the analog-to-digital converter (ADC) and sent to one of two image memories. If the image is a pre-injection mask, it is placed in one memory, where it remains for the duration of the run. All subsequent images are placed in the other memory, where they will each replace the previous image residing there. The mask image is then subtracted from each image in the other memory, as these images are acquired, to give a series of difference images. The difference images are then "windowed and leveled" to enhance the contrast visibility, changed back to an analog signal by the digital-to-analog converter (DAC), and finally sent to the viewing monitor.

Figure 1

Typical Block Diagram Of DSA System Using Digital Storage

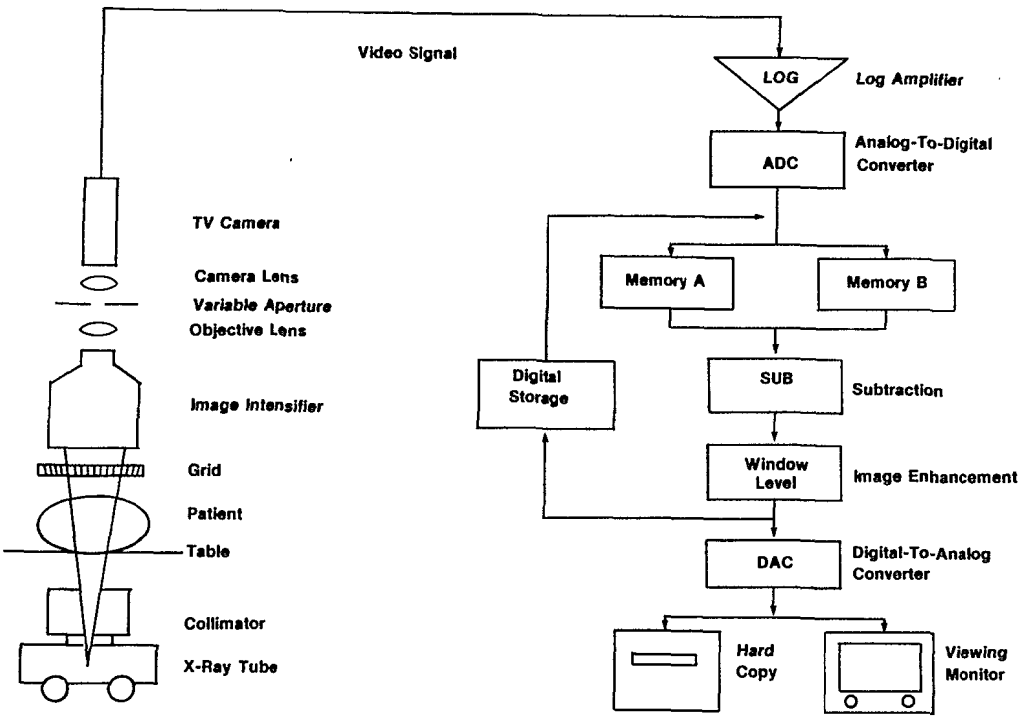
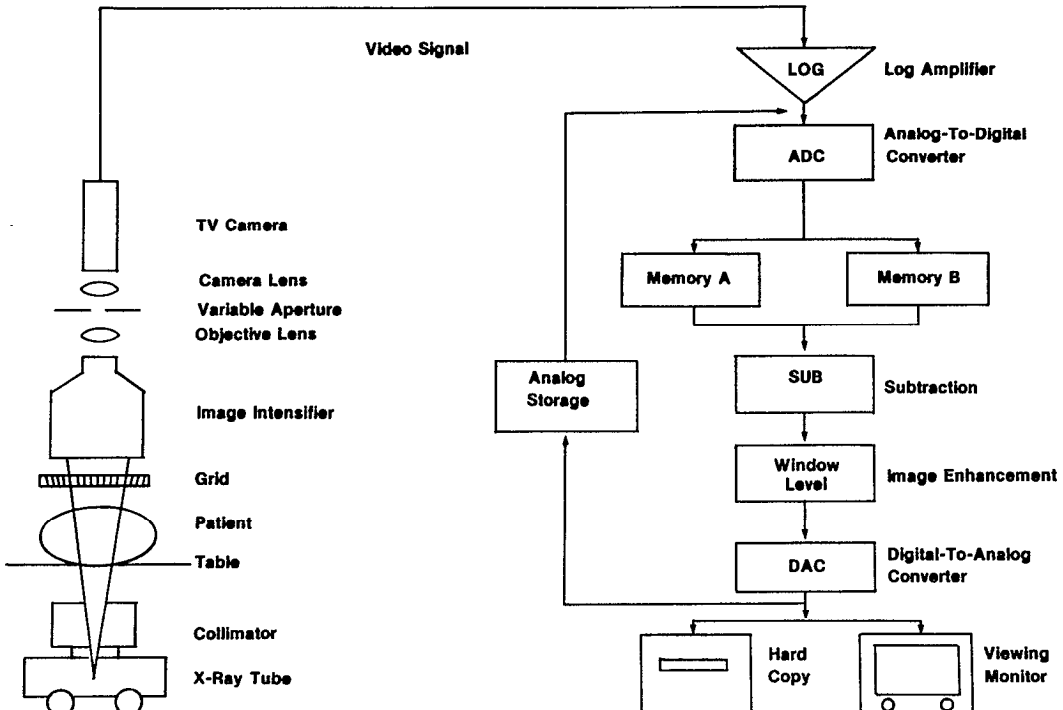


Figure 2

Typical Block Diagram Of DSA System Using Analog Storage



If the DSA system uses a digital storage device such as a digital disk, the digitized images are normally stored as shown in Figure 1. If the system uses an analog storage device, such as a video disk or a video tape, normally only the analog images are stored, and redigitization is required for any post-processing of the data. Long term storage may be accomplished on video hard-copy film, floppy disk, magnetic tape, or video tape.

Logarithmic processing is often used because it achieves density uniformity of the vessels in the subtracted image as they pass under bone and varying thicknesses of overlying tissue. This may be done by using a logarithmic amplifier on the incoming analog signal, or by modifying the digitized signal using a look-up table.

The operator's console allows initiation of the exposures and choice of the imaging mode and frame rate; it controls the injector and allows post-processing of the data. Post-processing may include window and level setting, remasking and reregistration to remove motion artifacts,' and quantitative analysis of the iodine concentration.

B. PERFORMANCE PARAMETERS AND THE FACTORS AFFECTING THEM

Performance parameters for DSA systems which are related to image quality and which can be evaluated with a non-invasive phantom are the following:

1) Spatial resolution
2) Low-contrast performance
3) Contrast and spatial uniformity
4) Contrast linearity
5) Radiation exposure
6) Subtraction artifacts

1) Spatial Resolution

Spatial resolution is a measure of the capacity of an imaging system to resolve adjacent high contrast objects (vessels). Spatial resolution can be expressed quantitatively in terms of the modulation transfer function (MTF). Measurement of the MTF is complex; therefore, in this report the spatial resolution is expressed as the visual cut-off frequency of a standard bar phantom and is specified in line pairs per millimeter (lp/mm). The cut-off frequency of the subtracted image will depend on the product of the MTF's of the individual components in the imaging system. The factors which have a major influence on spatial resolution are:

a) Image intensifier
b) Geometric magnification
c) Focal spot size
d) Display matrix size
e) TV chain

Items a, b, c, and e affect the amount of image blurring, whereas d and e affect the sampling rate.

a) Image Intensifier: Current cesium-iodide image intensifiers have cut-off frequencies of 4 to 5 lp/mm. This generally exceeds the spatial resolution limits of present DSA systems. Therefore, the II is usually not a limiting factor. The active input phosphor size is inversely related to the spatial resolution and is one parameter which may be selectable. It is important to realize that the active input-phosphor size is almost always less than the manufacturer's stated nominal size, and it should be measured. It is common for II's to have electronic magnification, which changes the active input phosphor size (i.e., dual-mode 9"/6"). For a fixed display matrix, the pixel size will increase with increasing active input phosphor size, thus resulting in reduced spatial resolution.

b) Geometric Magnification: In all DSA systems, considerable geometric magnification is present. The size of the object presented to the input-phosphor is enlarged by the magnification factor. Hence the effective resolution of the imaging system increases by the same magnification factor, at the expense of a decreased field of view and increased patient exposure. After a certain point, increasing the magnification will decrease the effective spatial resolution due to increased geometric unsharpness resulting from the finite focal spot size.

c) Focal Spot Size: The spatial resolution limit is inversely related to the focal spot size. The choice of a focal spot size in pulsed-mode DSA involves a trade-off between the increased geometric unsharpness resulting from a larger focal spot, and the decreased patient motion blurring resulting from the higher allowed mA's and correspondingly shorter exposure times associated with larger focal spots. The measured size of the focal spot may be 30-50% greater than the nominal size, and it may be even larger at high mA's. Although the intensity distribution is never uniform, one can calculate the resolution limit from a uniform square focal spot of known dimension:

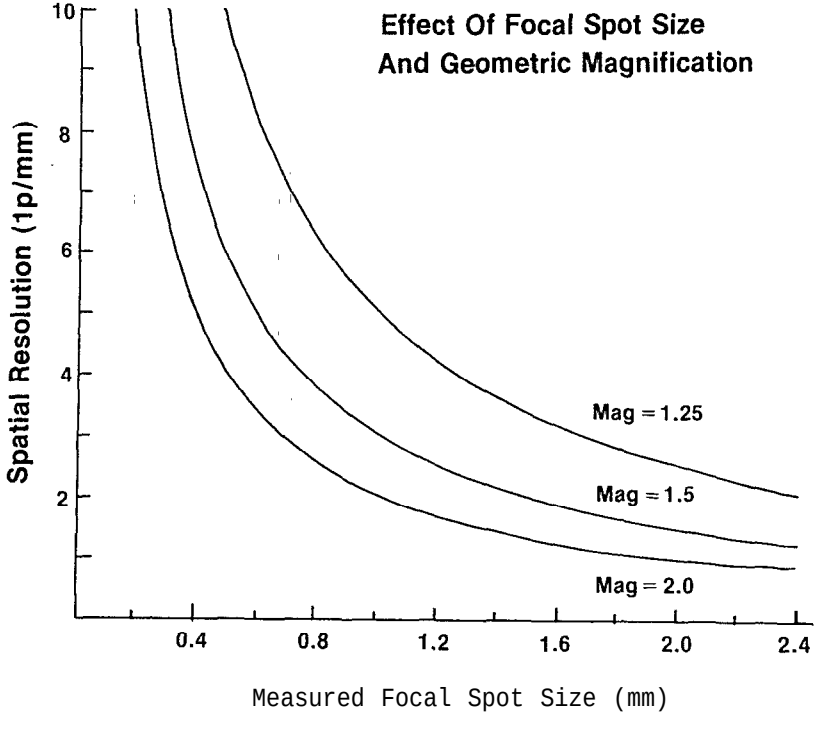
$$R = \frac{m}{F(m-1)}, \qquad (1)$$

where R = spatial resolution limit (lp/mm)
F = measured focal spot size in mm

m = geometric magnification

Figure 3 shows the effect of focal spot size and magnification on the spatial resolution. A magnification of at least 1.25 is almost always present due to the geometry of the equipment. Larger geometric magnifications may be used in practice.

Fig. 3



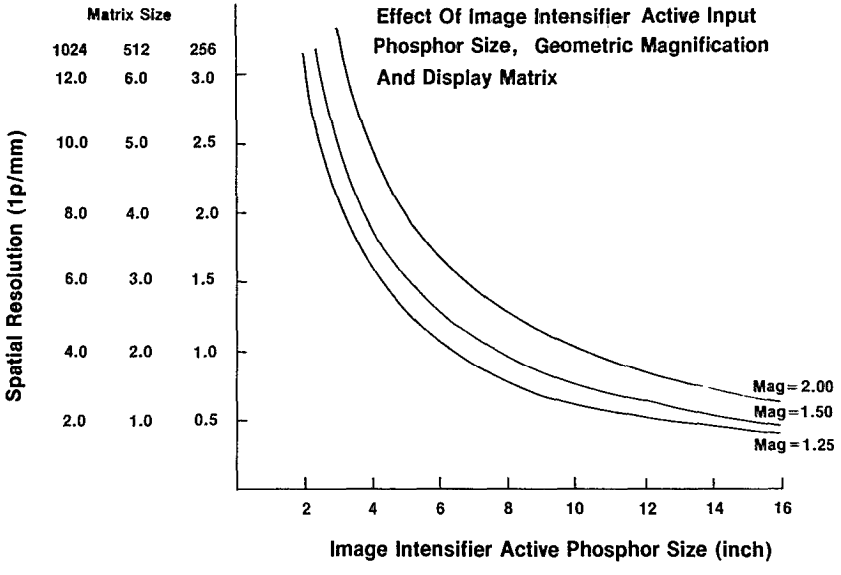
d) Display Matrix Size: The spatial resolution can never exceed the limit imposed by the pixel size. The pixel size depends on the display matrix (number of pixels) and on the diameter of the image in the patient plane (geometric magnification and active input-phosphor size). Typical display matrices used in DSA are 256 x 256, 512 x 512 and 1024 x 1024. The theoretical limit for spatial resolution due only to the display matrix is given by

$$R = \frac{m}{F(m-1)}, \qquad (2)$$

where R = resolution limit (lp/mm)
M = matrix size (ie, 256, 512, 1024)
m = geometric magnification
D = II active input-phosphor size or diameter (mm)
2 converts pixels/mm to lp/mm

Figure 4 shows the effect of active input-phosphor size, geometric magnification, and display matrix size on spatial resolution.

Fig. 4



e) TV Chain: The spatial resolution of the TV chain is affected by the electronic focusing of and the size of the electron beam in the TV camera, the optical focusing of the TV camera lens to the II output plane, and by factors which are different for the horizontal and vertical directions. The vertical resolution is affected primarily by the number of active scanning lines and by the Kell factor. The Kell factor is an empirical factor which expresses the fraction of the active scanning lines that are effective in preserving detail in the vertical direction. It has been determined experimentally to be about 0.7 for most television systems. The nominal number of scanning lines used in the U.S. television standard is 525, but systems with more lines (eg., 875, 1023) are common in medical imaging. Usually only about 480 (or 91%) of these lines are actually used in the formation of the image; the remainder are used for display control. Thus, the vertical resolution is proportional to (nominal number of TV lines) x (91%) x (Kell factor), up to the limit imposed by the TV camera beam size and focusing.

There are two methods of reading out the target or faceplate of the TV camera: the interlaced mode is the conventional method, while progressive scan readout is sometimes used in DSA systems. In the interlaced mode, one 525-line TV frame is read out in two passes over the target. The first pass scans only the odd-numbered lines and the second pass scans only the even-numbered lines. Each pass is called a field and consists of 262-l/2 lines. A progressive scan system reads out the entire target with one pass comprised of 525 lines. As each system uses the same number of scanning lines, it might appear that the spatial resolution would be equal. However, in a pulsed-mode DSA system, the image is stored on the target of the TV camera. If the pulse rate is 30 per second or less, and the image is read out in the interlaced mode, most of the signal (70-80%) is read out during the first field. This would effectively reduce the 525-line system to a 262-line system, hence decreasing the resolution. For systems operating in the fluoroscopic or frame-averaging mode, the image on the camera target is continually being refreshed between field readouts by new information from the image intensifier, and there is no loss of spatial resolution in the interlaced mode (for a stationary image).

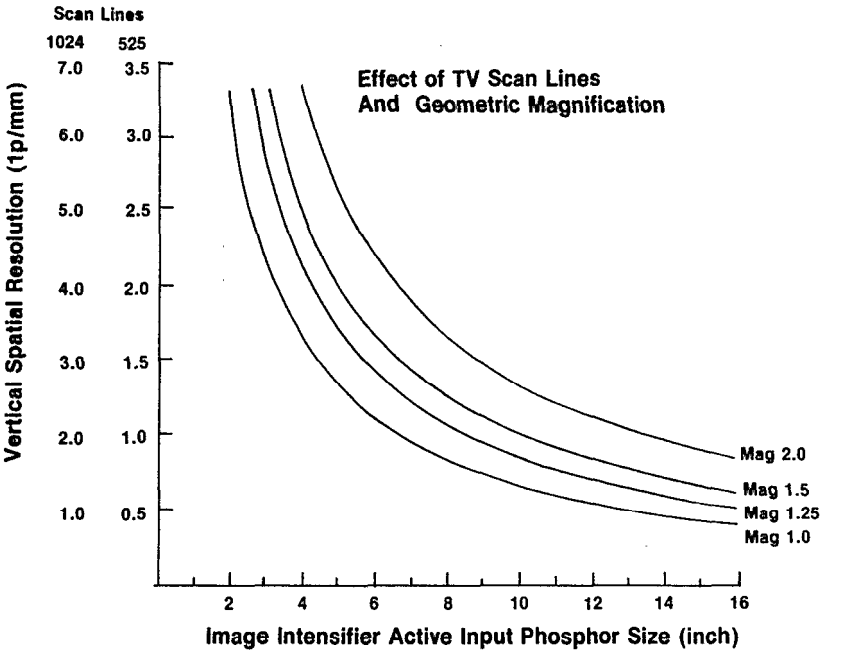
The TV spatial resolution limit in the vertical direction is determined by the effective number of scan lines per mm in the patient plane, which depends on the geometric magnification and the active input-phosphor size:

$$R\ (\text{vertical}) = \frac{SaKm}{2D}\ , \qquad (3)$$

where R (vertical) = resolution in vertical direction (lp/mm)
S = nominal number of TV scan lines (i.e., 525, 1023)
a = fraction of TV scan lines active in image formation (~.91)
K = Kell factor
m = geometric magnification
D = active input-phosphor size (mm)
2 converts lines/mm to lp/mm

Figure 5 demonstrates the effects of number of scan lines, geometric magnification and input phosphor size on spatial resolution.

Fig. 5



The spatial resolution limit in the horizontal direction is determined primarily by the number of scan lines per video frame and by the bandwidth of the video amplifier (i.e., how many times per scan line the video signal amplitude can be changed), up to the limit imposed by the beam size and focusing of the TV camera. In conventional TV systems, the bandwidth is usually set just high enough to provide equal horizontal and vertical resolution. However, in DSA systems, the bandwidth is often reduced by electronic filtering so that sampling artifacts are avoided and for reduction of electronic noise, which increases with bandwidth. In general, the sampling rate of the ADC should be about twice the bandwidth of the TV chain. Commonly used ADC's have sampling rates of 10-15 MHz, which limits the TV bandwidth to about 5 MHz. This may be sufficient to give approximately equal vertical and horizontal spatial resolution for a 525-line TV system, but not for a 1023-line system. The TV target in a 1023-line system would have to be read out with a slow horizontal sweep (slow scan TV) in order to have a horizontal spatial resolution that is even equivalent to a standard 525-line TV because of the bandwidth limitation imposed by the ADC sampling rate. Thus, it is important that the vertical and the horizontal spatial resolution be evaluated separately in DSA systems, since they may be quite different.

2) Low-Contrast Performance

Although the spatial resolution in digital subtraction angiography is generally inferior to that of conventional film radiography, DSA is useful because its low-contrast sensitivity is superior due to its capacity to enhance contrast through better subtraction, display windowing and centering. DSA is thus capable of imaging with smaller concentrations of iodine in the vessels under study. It is this low-contrast performance that has made possible angiography via intravenous injection of contrast material and the use of lower, less hazardous amounts of contrast material in intra-arterial angiography. Ideally, the evaluation of the low contrast performance should be closely related to the clinical imaging task of observing patency, stenosis, ulceration, aneurysms and calcification of blood vessels. These tasks are difficult to simulate and quantify with phantoms; however, all involve the ability to image an iodine-contrast-filled vessel well enough for detection of abnormalities in the vessel walls. Thus, a reasonable measure of low-contrast performance is the smallest cylindrical vessel that can be imaged with good visualization of the vessel edges. The test vessel should have an iodine concentration similar to that normally encountered in a patient's vessels during angiography.

The low-contrast resolution depends on the contrast signal presented to the image intensifier (II) and the capacity of the system to detect, preserve and display this signal. The contrast

signal from the patient depends on the iodine concentration, vessel size, and beam quality. The signal presented to the II is degraded by the scatter reaching the II, which depends on magnification, the contrast improvement factor of the grid, beam quality, field size, and patient thickness. The contrast detected by the II is generally degraded by the inability of the components of the imaging, digitizing, subtraction, storage and display chain to preserve this contrast. In addition, the presence of noise in the signal will degrade the perceptibility of low contrast. The low-contrast performance of the DSA system image is ultimately determined by the signal-to-noise ratio (SNR) of the system.

Two kinds of noise are present in the system, fixed pattern noise and random process noise. Fixed pattern noise is usually due to manufacturing variations in individual system components such as II input-screen non-uniformities, granularity of the II output-phosphor, TV camera target imperfections, and ADC step size nonuniformities. In the subtracted difference image, these are normally subtracted and do not degrade the image. However, when reregistration of the mask is carried out to correct for patient motion, the fixed pattern noise will appear in the difference image. Noise resulting from random processes fluctuates from frame to frame and from pixel to pixel, and thus limits the low-contrast detectability in the difference image. These random processes are due to statistical fluctuations in the detected x-ray fluence (quantum noise) and to the electronic noise in the imaging chain. The major sources of electronic noise are TV camera electronic noise, TV camera time jitter, ADC quantization noise, and electronic noise and time jitter in analog storage systems.

The difference image SNR is defined as the ratio of the contrast signal for an opacified object in the subtraction image to the RMS signal fluctuation in the neighboring background region. The contrast signal to be used in the calculation of the SNR is the signal from an opacified vessel less the neighboring background signal in the subtraction image. In DSA units with a region-of-interest (ROI) function, the contrast signal can be found if one places the ROI over an opacified vessel, obtains the average pixel value, and subtracts from this the average pixel value when the ROI is placed over the neighboring background. The noise to be used in the calculation may be the pixel standard deviation in the background region. This difference image SNR, along with the vessel or object size, determines the detectability of the vessel or object, and hence the low-contrast performance.

In any evaluation of the low-contrast performance of a DSA system, it is important that one understand the role of the operator-adjustable parameters, and that these parameters are set to typical clinical values after they have been properly

optimized. (See Section III) The operator-adjustable parameters which have the greatest influence on image quality are geometric magnification, pixel size, beam quality, and radiation exposure at the II input. It is most important when comparing the low-contrast performance of two DSA systems to use exactly the same operator-adjustable parameters in both systems.

a) Beam Quality: In general, the difference image SNR is optimized for a highly filtered beam in the range of 50-60 kVp for thin body parts (IO cm), 60-70 kVp for medium body parts (20 cm), and 65-75 kVp for thick body parts. These guidelines should be used for evaluation with phantoms with the understanding that the kVp may have to be raised due to mA limitations on radiographic equipment.

b) Geometric Magnification: Within the limitation imposed by focal spot blurring, geometric magnification will improve the detection of low contrast objects, so long as the II input exposure is held constant. In any evaluation of low-contrast performance, one should use a typical magnification encountered in clinical studies, such as 1.25.

c) Pixel Size: Pixel size depends on the matrix size, II active input phosphor size, and geometric magnification. Decreasing the pixel size decreases the resultant SNR when the radiation exposure to the II is low and the camera video output signal is relatively high.

d) Entrance Exposure to the II: The two major contributions to the noise in DSA are quantum noise and TV camera noise. Above a certain level of II input exposure, the quantum noise is low and low-contrast resolution is dominated by TV noise. In this region, the performance is relatively independent of patient or II exposure, and increased exposure will not result in increased image quality. This is the region of needlessly high patient dose.

In the low II input-exposure range, the x-ray photon quantum noise dominates the TV camera noise, and the low-contrast performance can be described approximately by the following expression:

$$[\mu / \rho]_I \, t_I \cdot d \propto \frac{1}{\sqrt{R}} \qquad (4)$$

where $[\mu / \rho]_I$ = effective iodine mass attenuation coefficient (cm^2/mg)

t_I = projected iodine thickness (mg/cm^2)

d = vessel diameter (cm)

R = radiation exposure at input of the II

This expression simply indicates that, for a given iodine contrast level, the smallest detectable diameter of a vessel varies inversely as the square root of the exposure. Conversely, it indicates that to see the same vessel at half the iodine concentration requires 4 times the exposure. This relationship can be plotted in a series of contrast vs. vessel diameter curves called contrast-detail-dose (CDD) curves. There will be a separate curve for each exposure up to the point where TV camera noise becomes dominant. These CDD curves characterize the low-contrast performance of a DSA system and can guide the selection of x-ray techniques.

3) Contrast and Spatial Uniformity:

Generally, DSA imaging systems use logarithmic processing of the video signal to insure that a vessel of uniform diameter and uniform iodine concentration appears in the subtracted image with uniform diameter and contrast, regardless of overlying structures such as bone, air and varying tissue thickness. The logarithmic processing increases the dynamic range of the unsubtracted video signals which the system can detect, and hence is usually necessary when a large range of x-ray attenuation is present in the image. The processing must be adjusted properly to insure a uniform contrast signal in a vessel crossing areas of drastically differing attenuation, such as regions containing bony structures. The contrast uniformity is degraded in a less noticeable way by a nonuniform distribution of radiation scatter and veiling glare in the video image.

Good spatial uniformity means a uniform magnification factor over the entire field of the II. This is seldom achieved due to the curved input surface of the II and nonlinearities in the electron optics in the II and TV chain, and in the optical lenses in the imaging chain. It most often appears as "pin-cushion" distortion which is normally "edited out" by the viewing radiologist, but may become important if quantitative measurements are to be made on the image.

4) Contrast Linearity:

A second benefit of logarithmic processing is that the difference contrast signal is directly proportional to iodine thickness, and independent of transmitted x-ray fluence. This allows DSA images to be used for the quantitative measurement of physiological

quantities related to a patient's cardiovascular system.

To see this, consider the video signal levels in a subtraction of a mask from an opacified image using logarithmic processing:

$$M = \log (a N), \quad (5)$$

where M = mask image signal

N = number of detected photons per pixel in the mask

a = a conversion factor mapping detected x-ray fluence to video-signal level

Also, $$C = \log [\alpha N \exp(l-[\mu / \rho]_I t_I)] \quad (6)$$

or $$C = \log [\alpha N] - [\mu / \rho]_I t_I$$

where C = signal of image containing iodine contrast

$[\mu / \rho]_I$ = effective iodine attenuation coefficient (cm^2/mg)

t_I = projected iodine thickness (mg/cm^2).

Thus, the difference signal is

$$D = M-C$$

or $$D = [\mu / r]_I t_I , \quad (7)$$

which is <u>proportional</u> to iodine thickness and <u>independent</u> of the transmitted x-ray fluence N.

The contrast linearity is dependent on the proper adjustment of the logarithmic processor, the luminance linearity of the image intensifier and TV camera, and the linearity of the ADC. It is thus important that the II input exposure not be so high that its luminence becomes nonlinear with input. The effective f-stop of the TV camera must also be carefully chosen to maintain video signal linearity. When these conditions are satisfied, the contrast linearity should be good for a localized small iodine signal. For large opacified structures, such as the cardiac chambers or the aortic arch, nonlinearity often occurs as the result of varying radiation scatter, veiling glare, and large variations in iodine thickness. In the presence of such nonlinearities, the difference signal can be approximated by:

$$D \cong \frac{[\mu/\rho]_I t_I}{(1+s)(1+g)} , \quad (8)$$

where s = scatter radiation ratio

g = veiling glare ratio.

In order to compare DSA systems and to verify specifications, one must understand what is meant by "iodine contrast." This is variously referred to as percent contrast, projected iodine thickness (mg/cm^2), or iodine concentration (mg/cm^3). Each of these has a different meaning and use:

Iodine Concentration (mg/cm^3) This is the concentration of iodine in a vessel and is the quantity of interest in quantitative flow studies. It is also the easiest way to specify the amount of iodine to be placed in a vessel phantom since this quantity is independent of beam quality, phantom thickness, and vessel shape and size. Iodine concentration forms the link between simulated vessels and patient studies, since the iodine concentrations in patient vessels from intravenous injections are at the same levels as the iodine concentrations in the phantom vessel insert. The iodine concentration depends on the type and dose of the contrast agent, the injection technique, and the dilution curve for the patient and the vessel site under study.

Projected Iodine Thickness (mg/cm^2): This is the total thickness of iodine seen by the x-ray beam in any given pixel. It is dependent not only on the concentration of iodine in a vessel, but also on the thickness and shape of the vessel. It is independent of x-ray beam quality and phantom thickness. For example, a round vessel with a given iodine concentration will have a smaller mg/cm^2 value at the edges of the vessel than at its center. This is the quantity proportional to the numerical pixel values in a subtracted image when logarithmic processing is used.

Percent Object Contrast: This is a traditional term from radiography which refers to the percentage attenuation of primary x-rays by the iodine present. The percent object contrast is dependent on the iodine concentration, the vessel thickness, and the x-ray beam quality within the pixel of interest. Percent object contrast is a useful term for comparison of the low contrast performance of DSA systems to other radiographic systems, but projected iodine thickness is more useful quantity for the specification of low-contrast performance because it is

directly related to the clinical iodine contrast injection techniques and is independent of beam quality.

Percent Image Contrast: This is the iodine contrast signal in the subtracted image and is defined as the mask image signal for a certain pixel minus the corresponding signal from the image containing iodine. In a perfectly linear imaging system with no scatter, the percent image contrast would be equal to the percent object contrast. In reality however, image contrast is reduced by the presence of scatter and veiling glare. Like object contrast, image contrast is dependent on iodine concentration, vessel thickness, and x-ray beam quality.

F) Radiation Exposure:

The radiation exposure produced in a DSA examination will affect the quality of the images produced, as well as the magnitude of the potential radiation risk to the patient and to the personnel conducting the examination. Several different exposure parameters are associated with the DSA procedure:

a) Patient Entrance Exposure: A measurement of the exposure produced in the plane at which the beam enters the patient will be necessary for comparison with other procedures, or for determination of the dose to specific organs. Since the exposure is normally different for different settings of the II field size, it is important to make these measurements at all the II field sizes used.

b) Exposure at Image Intensifier Input Phosphor: A major factor determining the quality of the final image is the x-ray exposure (or exposure rate) reaching the input surface of the II. If the exposure (or exposure rate) is low, the system noise will be dominated by the x-ray quantum noise rather than by the system electronic noise. The exposure at the II input phosphor will therefore play an important role in determining the low-contrast performance and hence the CDD curve for the system. In order to compare the image performance of one system to that of another, the CDD curves must be labelled with the exposure (or exposure rate) at the II input-phosphor.

i) Continuous Mode: When the system is run continuously, as in fluoroscopy for localization or in fluoroscopy-mode subtraction, a measurement of the exposure rate at the II input surface will provide the fluoroscopic input exposure rate sensitivity (FIERS). The FIERS depends on the f-stop of the TV camera lens, the TV camera target voltage, and the automatic brightness control settings for the x-ray generator. Typical values are 30-600 uR/s for the 6" mode

and 15-300 uR/s for the 9" mode.

ii) Pulsed or Cine Mode: The image intensifier input exposure sensitivity (IIIES), typically expressed in units of uR/frame, is the parameter which determines system information content when the system is run in a pulsed x-ray mode. Typical IIIES values vary from 100 to 1000 uR/frame and normally are inversely proportional to the active input-phosphor area for II's of different size.

c) Personnel Exposure: Exposure to personnel will normally come from radiation scattered by the patient to the side of the x-ray table, and from the leakage radiation fron the x-ray tube housing. The magnitude of this hazard will depend upon the technique factors, beam quality, field size, intensifier size and position, location of drapes and shields, and the position of personnel in the room. The location of the x-ray tube strongly influences personnel exposure. Exposure to the operator is generally much less when the x-ray tube is under the table, rather than over the table. Iso-exposure curves for various conditions should be measured. Typical examples are given in AAPM Report No. 12, "Evaluation of Radiation Exposure Levels".

6) Subtraction Artifacts

If the pixel location of a high contrast background object such as bone, air, bowel gas, or metal should move during mask-mode subtraction, the result of this misregistration between the mask and the opacified image will be either the introduction of an artifact or incomplete removal of overlying structures. This, in turn, may obscure the visibility of an opacified vessel.

Misregistration can be caused by imaging chain deficiencies or by motion of the patient, x-ray tube, or II. Imaging-chain misregistration artifacts can be due to TV camera sweep instabilities, II power supply fluctuation, and time base jitter. Artifacts from these sources reflect improper functioning of the equipment and should not be tolerated.

Patient motion is the primary cause of misregistration artifacts and of unsatisfactory studies. Motion artifacts can often be decreased by remasking, a process in which one of the frames taken after the motion occurred is used as the new mask and subtracted from subsequent images. A second technique is reregistration, in which the two images to be subtracted are shifted with respect to each other and sometimes rotated. It appears that shifting by a fraction of a pixel is necessary, and that the whole image cannot be perfectly subtracted, but that one can optimize the subtraction in a vessel area of interest.

Incomplete subtraction can occur due to instability of the x-ray tube output, which might fluctuate by several percent during the imaging sequence mode. However, because of the logarithmic processing of the video image before subtraction, this fluctuation is turned into a uniform brightness change from one image to another. Although this brightness change obviously needs correction for quantitative use of the image data, such as in blood flow analysis, it does not affect the visualization of the opacified vessels in subtraction images. On the other hand, excessive brightness change in an image sequence indicates possible malfunction of the synchronization of x-ray pulsing to image acquisition, the television camera, or the x-ray tube itself.

III. PERFORMANCE EVALUATION WITH A PHANTOM

The performance evaluation method to be outlined in this section is non-invasive and involves the use of a patient-simulating phantom. Also needed will be standard fluoroscopic resolution test devices and a dosimeter, whose characteristics are described in the exposure measurement section.

A. Typical Phantom

Figure 6 illustrates a possible phantom design which would be acceptable for performance evaluation. The basic attenuation phantom could be made of lucite or tissue-equivalent plastic and should allow assembly or stacking in such a way that the thickness of any part of the body could be simulated. There should be a step-wedge section which has an x-ray attenuation dynamic range greater than 15:1 to simulate the large range of signals encountered in DSA examinations. There should be a bone section which further simulates extremes in dynamic range, the high contrast overlying structures in actual patients, and beam hardening effects. The bone could be actual bone or bone equivalent plastic, but it should be very similar to bone in attenuation and scatter characteristics for use in systems that are capable of dual energy subtraction.

The phantom should have a slot somewhere near mid-plane where the resolution inserts can be placed with a magnification simulating clinical conditions. Figure 7 shows the blank insert which is to be used for mask images (pre-contrast). It is important that this blank insert be made of the same material and be the same thickness as the resolution inserts that will be replacing it in this slot.

Figure 8 illustrates a possible vessel insert with simulated

vessels of 0.5, 1, 2, and 4 mm diameter. These vessels should be filled with iodinated epoxy, with a different concentration for each set of vessels. Concentrations of 10 mg/cm^3, 5 mg/cm^3, and 2.5 mg/cm^3, of iodine simulate typical clinical conditions for intravenous injection. If intraarterial injection techniques are used, an insert of higher concentration, such as 60 mg/cm^3, would be useful to assure proper operation of the unit under such conditions. The vessel insert would be used for the evaluation of low-contrast performance and uniformity. It is desirable that the simulated vessels be circular (or semi-circular) in cross section to provide the same difficulty of clearly imaging vessel edges that occurs with actual blood vessels.

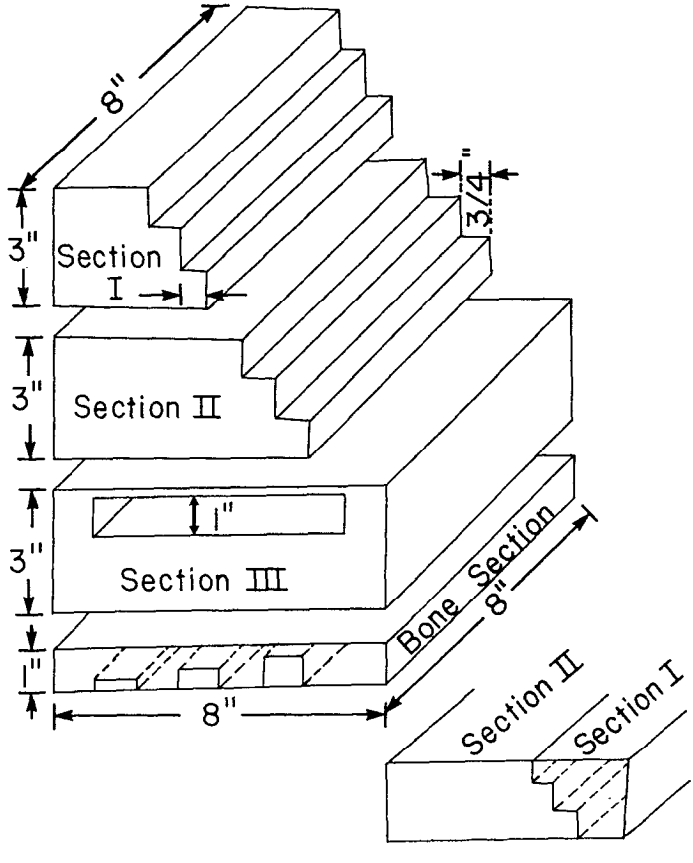
Although the vessel insert described is necessary to simulate the clinical situation, particularly for quantitative analysis and uniformity, a line-pair type of insert as seen in Figure 9 may be more useful for optimizing the system. A line-pair target test is less observer-dependent than determining the smallest perceptible single vessel with indistinct edges. The line-pair insert has channels of rectangular cross section to give high contrast edges. All channels (1, 0.7, 0.5, 0.35, 0.25, 0.175, 0.125 lp/mm) have the same 1 mm depth; therefore, the contrast (mg/cm^2) is constant for all sizes, as opposed to the vessel insert, whose contrast decreases as the vessel size decreases. Filling these channels with iodinated epoxy at iodine concentration of 5 mg/cm^3 results in an iodine thickness of 0.5 mg/cm^2. Similarly, a set with 10 mg/cm^3 iodine concentration would give an iodine thickness of 1.0 mg/cm^2.

Figure 10 shows a misregistration insert. It is a standard l/16" (1.6 mm) thick perforated aluminum plate with l/8" (3.2 mm) diameter holes which provide many high contrast edges.

Figure 11 illustrates a linearity insert. It contains six regions of different iodine thickness; 0.5, 1, 2, 4, 10, and 20 mg/cm^2. It should be made of the same material as the blank insert.

The manufacture of iodinated plastic phantoms requires considerable skill and practice. The typical mixture of iodobenzene and casting resin requires very careful measurement of small amounts of iodine to insure accuracy, careful mixing to insure uniformity, and vacuum curing to eliminate air bubbles. There is also a question about the possible diffusion of iodine out of the channels over time.

Figure 6. Attenuation Phantom. Stepwedge Sections I and II Can Be Combined To Form A Uniform Thickness Of 3 Inches.



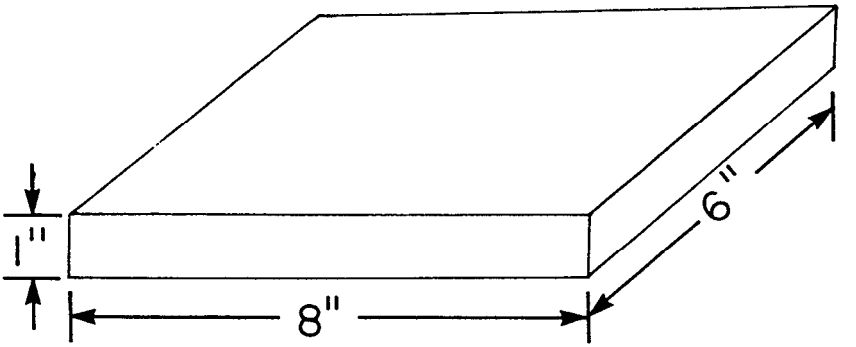


Figure 7. Blank Insert.

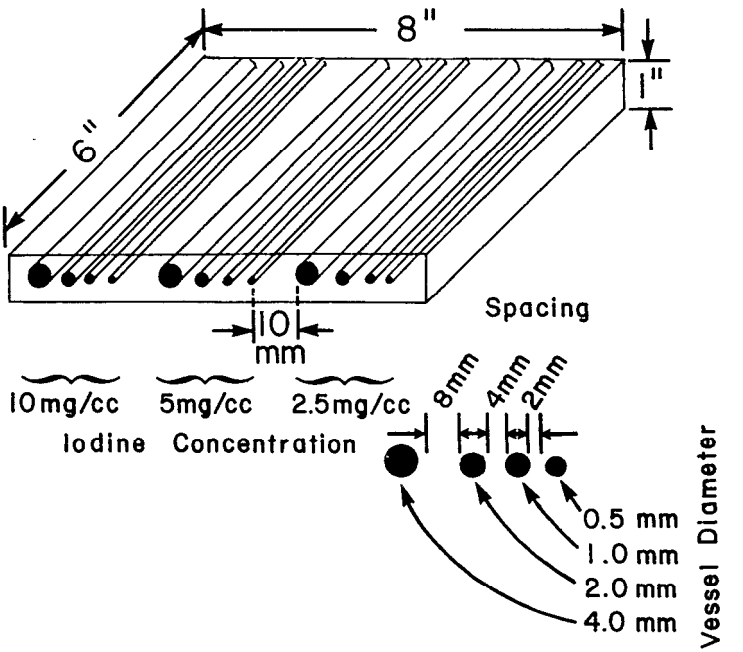
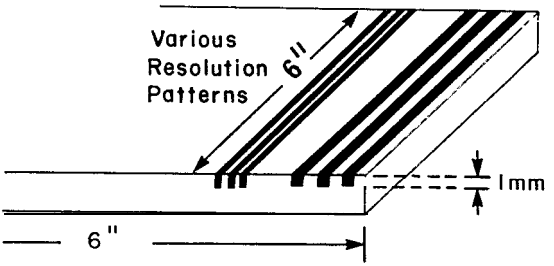


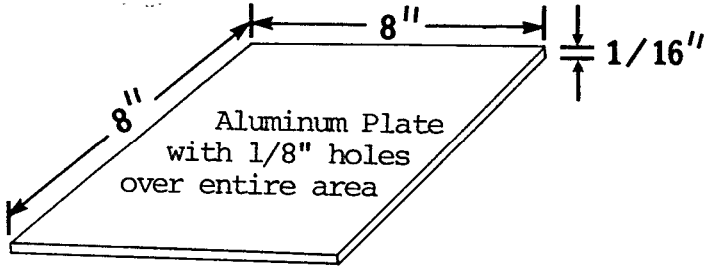
Figure a. Vessel Insert.

Resolution Pattern	0.125	0.175	0.25	0.35	0.5	0.7	l.O(lp/mm)
Groove Width	4.0	2.8	2.0	1.4	1.0	0.7	0.5(mm)
SpaceBetween Groups	8.0	5.6	4.0	2.8	2.0	1.4(mm)	



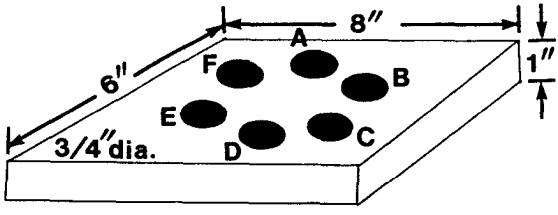
One insert is filled with 10 mg/cm^3 of Iodine.
A second insert is filled with 5 mg/cm^3 of Iodine.

Fig. 9. Low Contrast Line Pair Insert



(Use of l/8" perforated aluminum plate is acceptable)

Figure 10. Misregistration Insert.



Iodine Concentration of the Discs:

(A) 0.5 mg/cm^2

(B) 1.0 mg/cm^2

(C) 2.0 mg/cm^2

(D) 4.0 mg/cm^2

(E) 10.0 mg/cm^2

(F) 20.0 mg/cm^2

Figure 11. Linearity Insert.

B. Spatial Resolution

1) Method: The spatial resolution should be determined at the location where the radiologist will view the image. Typically, this is at the display console. This provides a measurement of the overall performance of the system rather than of the operation of individual components. It may also be necessary to check the performance of these components to diagnose problems or to optimize the system performance.

The spatial resolution may be different in the horizontal and vertical directions due to the bandwith and the number of TV lines, therefore it is recommended that both horizontal and vertical spatial resolution be determined, as well as the resolution at 45°. This can be done by placement of a standard lead bar line-pair pattern parallel, perpendicular, and at 45° to the TV raster lines. The pattern should go up to 5 lp/mm, and the lead thickness should be no greater than 0.1 mm. If this cause3 dynamic range problems, a thinner lead pattern (0.01 mm) or a wire mesh pattern might be used. The three patterns should be placed in a 15 cm (6 inch) thick phantom to simulate a patient. It is recommended that a magnification of 1.25 be used if this is consistent with the system geometry. The spatial resolution should be determined by viewing of both unsubtracted and subtracted images. It will be necessary to remove the pattern3 for the mask image.

2) Evaluation: The spatial resolution cut-off frequency is a complex function of the II active input-phosphor size, geometric magnification, focal spot size, number of TV scan lines and display matrix size. Table I summarizes the theoretical cut-off frequencies for a number of combinations of the variables. It should be understood that the cut-off frequency listed in the table is that of the component in the imaging chain which ha3 the poorest cut-off frequency ("weakest link") and ignore3 the degrading influence of the other components. Thus, in practice, the measured cut-off frequency may be slightly below these values. Interpolation for combinations not shown can be made with the formula3 given in Section II-B1.

Gross deviation3 from the value3 in the table, or degradation over time, should be investigated by isolation of the influence of the various component3 and variables. Some are easily isolated, but others require additional test equipment. Focal spot size and geometric magnification influence3 can be eliminated if the test pattern3 are placed against the II input surface, in which case the magnification would be very close to 1.0. To eliminate the loss of contrast due to scatter and to increase the contrast of the test pattern, one can remove the phantom and use a very low kVp. To isolate the II, one can view

TABLE I

THEORETICAL LIMIT OF SYSTEM SPATIAL RESOLUTION

4	512	1.25	0.90	525	2.30 TV Limit
6	"	"	"	"	1.53 "
9	"	"	"	"	1.02 "
14	"	"	"	"	0.65 "
4	512	1.50	0.90	525	2.77 TV Limit
6	"	"	"	"	1.84 "
9	"	"	"	"	1.22 "
14	"	"	"	"	0.78 "
4	512	2.0	0.90	525	2.30 F.S. Limit
6	"	"	"	"	2.30 "
9	"	"	"	"	1.62 TV Limit
14	"	"	"	"	1.05 "
4	512	1.25	1.4	525	2.30 TV Limit
6	"	"	"	"	1.53 "
9	"	"	"	"	1.02 "
14	"	"	"	"	0.65 "
4	1024	1.25	0.90	1024	4.46 TV Limit
6	"	"	"	"	2.98 "
9	"	"	"	"	2.00 "
14	"	"	"	"	1.27 "
4	512	1.25	0.90	1024	3.46 Matrix Limit
6	"	"	"	"	2.30 "
9	"	"	"	"	1.54 "
14	"	"	"	"	1.00 "
4	1024	1.5	0.90	1024	3.30 F. S. Limit
6	"	"	"	"	3.30 "
9	"	"	"	"	2.38 TV Limit
14	"	"	"	"	1.52 TV Limit
4	1024	2.0	0.45	1024	3.90 F. S. Limit
6	"	"	"	"	3.90 "
9	"	"	"	"	3.57 "
14	"	"	"	"	2.30 "

II = Nominal image intensifier size (inches); calculation assumes active II size = 0.90 nominal size
M = Display matrix size
m = Geometric magnification
F.S. = Actual focal spot size (mm)
S = Nominal TV scan lines; calculation assumes a=0.91, K=0.7
R = Theoretical spatial resolution cut-off frequency

the output phosphor of the II directly by using a telescope focused-at infinity. It may be necessary to remove the TV camera for this. The adjustment of the II electronic focusing is best made in this manner. AAPM Report #4 provides more details on the evaluation of an II. For optimization of the TV electronic and optical focusing, test pattern3 should be placed against the II input surface with no phantom in place, a low kVp used, and adjustment3 should be made during fluoroscopy. AAPM Report #4 provide3 additional details.

The subtracted image should be recorded on the video hard copy imager and the spatial resolution compared with that obtained on the display monitor.

C. Low Contrast Performance:

1) Method: The attenuation phantom should be assembled with the blank insert to give a uniform (no step wedge) thickness of 15 cm (6 inches) and should be set up in a geometry similar to that used for patient studies. The x-ray tube potential is to be set at 70 kVp. Scout pulse3 are taken and the mAs and/or camera aperture adjusted until they are within the manufacturer's recommended operating range. The mask image is obtained. Next the vessel insert (or low-contrast iodine line pair insert) is put into the phantom in place of the blank, and subtracted images are obtained without adjustment of any parameters. The kVp, mAs, exposure time, focal spot size, focal spot-to-object distance, focal spot-to-image receptor distance, II active input-phosphor size, aperture setting, phantom thickness, and all other pertinent variable3 are recorded. The test may be repeated for other phantom thicknesses for simulation of the type of studies being done at a particular facility. It is useful to measure the patient and/or II input-exposure at the same time (See section III.F, Radiation Exposure).

NOTE: Units that operate only in the fluoroscopic mode cannot be evaluated with the above method; the iodine thickness must be changing to be seen. Motorized vessel insert3 would be needed for accurate quantitative evaluation of the low-contrast performance of these DSA systems. However, it may be possible to image vessels by manually moving the vessel insert at an appropriate speed perpendicular to the direction of the vessels.

2) Evaluation: The smallest-diameter vessel (or finest line-pair) seen for each iodine concentration is determined by optimizing of the display window and level controls. This information can then be compared with manufacturer or purchase document specifications, can be used for comparison of two DSA systems, used for optimization of component3 and operating variables, or recorded periodically for quality assurance

purposes. CDD curve3 may be drawn which correlate the data.

The optimization of the imaging variables, such a3 kVp, mAs, beam filtration, aperture setting, frame averaging, and magnification, is best accomplished with the use of a phantom. Especially important is the optimization of the input exposure to the image intensifier. In general, raising the radiation exposure will allow smaller vessel3 with lower concentrations of iodine to be seen, because the quantum noise will be reduced (the TV camera aperture may have to be reduced for increased exposure). The point at which increasing the exposure no longer reduces the noise (and no longer increase3 vessel visibility) is the point at which the TV camera noise dominates. Radiation exposures above this point should never be used.

The optimum exposure level may not be the same for different clinical procedures which involve different vessel sizes and different iodine concentrations. Each clinical situation should be simulated with the phantom (proper phantom thickness) and the optimum input exposure (IIIES) determined for the desired perceptibility. Evaluation should also be made with and without a grid to determine if a grid is necessary for all procedures. Without a grid both patient and personnel exposures are greatly reduced.

D.1 Contrast Uniformity:

1) Method: The phantom is used with the step wedge in place and a subtracted image of the vessel insert is obtained following the procedure outlined in Section III-C. The logarithmic amplifier must be on. The direction of the vessel3 should be perpendicular to the steps. The test should also be done with the bone section added to the step wedge.

2) Evaluation: The vessels should appear uniform in density and width a3 they pass under the steps and bones. There will be increased noise for those portions of the vessel3 that pa33 under the thickest portion3 of the phantom, due to added photon attenuation. Some steps may disappear due to the limited dynamic range.

If the uniformity of density and width is not adequate, particularly for the highest contrast vessels, some adjustment of the logarthmic processor may be needed. The service engineer should be consulted.

This test also measure3 the dynamic range of the system. The number of steps which can be seen (by the presence of noise changes) is related directly to the dynamic range. A system which saturates and cannot see all six steps will most likely require bolus material around the patient, or a shaped filter at

the collimator.

D.2. Spatial Uniformity:

1) Method: A subtracted image of the vessel insert is obtained with a thin uniform phantom (no step wedge).

2) Evaluation: The difference between any two vessels is measured at the center and edges of the image with the electronic calipers, or with a ruler on the hard copy. This is a measure of the amount of distortion, which should be taken into account when size measurements are performed.

E. Contrast Linearity:

1) Method: A subtracted image is obtained with the linearity insert and the uniform phantom, following the methods of Section III-C. The average pixel value is obtained for each of the linearity sections. For systems with quantitative analysis packages the ROI function is used. For systems not having this, a narrow window can be set and the level setting scrolled until the mean pixel value is found for each section.

2) Evaluation: The pixel values are plotted as a function of the iodine thickness (mg/cm^2) in each section. They should fall on a reasonably straight line. If not, the logarithmic processor may not be properly adjusted. The user should be aware that the logarithmic processor is often adjusted to compensate for the effects of scatter and veiling glare for a certain phantom thickness and kVp, and it may not be possible to achieve precise linearity for all clinical situations. The service engineer should be consulted. This method also calibrates the contrast scale of the unit; this is useful for quantitative flow studies and for comparison of different DSA systems.

F. Radiation Exposure:

1) Method:

a) Instrumentation: Three different dosimeters may be required for these measurements.

i) Entrance exposure - The ionization chamber and associated electrometer should be capable of measuring exposure rates up to 200 R/min or an accumulated exposure of 50 R per run. The energy response should be calibrated over a range of beam qualities from 1.5 mm Al to 7.0 mm Al HVL.

ii) IIIES or FIERS - The ion chamber system should be capable of measuring exposure rates as low as 0.1 mR/min or an accumulated exposure as low as 0.1 mR per run. The chamber should be small enough to be contained within the field of view of the intensifier and relatively thin so that it can be located near the input surface of the II.

iii) Personnel exposure - An ionization chamber survey meter capable of reading scatter exposure rates up to 1 R/hr should normally be used.

b) Phantom Thickness: At least two phantom thicknesses should be employed for these measurements; a 10 cm (4 inch) thick section to simulate the neck or cervical spine and an 18 cm (7 inch) thick section to simulate the abdomen. This thicker section will probably require technique factors which are also appropriate for cerebral examinations.

c) Experimental Arrangement: The phantom is placed on the table top and centered in the x-ray beam. The II input surface is positioned 30 cm above the table top. For C-arm units, the target-intensifier distance is adjusted to 76 cm (30 inches), if possible. If the phantom does not take up the full field of view of the II, the x-ray field is collimated to within the area of the phantom. The field size is recorded.

i) FIERS or IIIES measurements are made with the phantom in place. The ion chamber is to be supported at the input surface of the II between the anti-scatter grid and the II input surface. For positioning the ion chamber, the anti-scatter grid will have to be removed temporarily.

C A U T I O N

For image intensifier systems in which the anti-scatter grid can not be easily removed, removal of the grid MUST be performed by a qualified person. This is especially important when the grid is a built-in type design or is an integral part of the II housing. The II tube is both an implosion hazard and very expensive to replace if it is damaged. The removal of this type of grid housing may cause room light to affect the II and the operation of an automatic exposure system.

ii) For patient entrance exposure measurements, the ion chamber is placed on the table top (or at 30 cm from the II input-surface for C-arm units) centered in the beam. The anti-scatter grid should be in place on the II, if the unit is used this way clinically.

The unit is run at typical clinical technique factors for these measurements. If typical factors are not available, the unit is set for 70 kVp and the mAs is adjusted according to the DSA manufacturer's instructions for the phantom thickness used; or the automatic exposure technique factor control circuitry is used for determination of the technique factors.

It is generally desirable to perform measurements with the dosimetry system operated in the rate mode. Exposure times should be long enough to permit automatic brightness control (if any) of the x-ray system and the ionization chamber measurement system to reach stable values. The exposure/frame is obtained by division of the measured exposure rate (R/min) by the known frame rate ((f/s) X 60). The exposure measurement mode may be appropriate for those systems which produce clearly separated individual exposures at rates less than 1 frame/sec. The exposure per frame is then just the total exposure measured, divided by the total number of frames. The particular model of dosimeter used should be carefully evaluated as to whether it gives correct readings for pulsed beams in the readout mode being used.

If the grid cannot be removed, the exposure rate into the grid must be divided by the grid transmission factor. distance corrections may also be necessary if the ion chamber is not located close to the II input surface.

2) Evaluation:

a) The patient entrance exposure is determined either by multiplication of the measured entrance exposure rate in R/min by the total time of the exposure expressed in minutes, or by multiplication of the measured exposure in mR/f by the total number of frames used. To obtain the total entrance exposure for the examination, one must remember to add the exposures required to obtain scout images, subtraction mask(s), and/or fluoroscopy localization exposures to those required for the actual subtraction images.

b) The IIIES will normally be expressed as the II input exposure in uR/f. If several frames are averaged to yield an image, the total exposure for the several frames should be stated. The FIERS should be given as an exposure rate in uR/s or divided by the television frame rate of 30 f/s to provide an effective value of uR/f. Typical values of the IIIES and FIERS are given in Section II-B.5.

c) Scatter measurements within the examination room under typical clinical conditions will provide information necessary for estimation of risk to personnel performing the examination, or for determination of the need for additional shields or drapes within the room.

G. Artifacts

1) Method: The misregistration insert is placed in the uniform phantom. This image is then subtracted from itself, i.e., the same phantom is used for the mask and the live image. This method can be used in exactly the same manner for subtraction fluoroscopy. This test should be carried out over a time equal to or larger than a typical clinical run, as instabilities usually increase with time.

2) Evaluation: The auto-subtraction should result in a featureless image with only differences in noise. Any evidence of hole edges, whether over the entire image or in bands, usually indicates problems such as TV camera sweep instability, time base jitter, or II power supply instabilities. The amount of misregistration can often be quantized by pixel shifting until the image disappears. The number of pixels shifted quantifies the extent of the problem.

IV. QUALITY ASSURANCE

A complete program of quality assurance involves (a) specifying performance in the purchase document, (b) acceptance testing of the performance at installation, and (c) regular monitoring of performance after installation.

Items To Be Monitored and Frequency

1) X-ray and Fluoroscopy System: This should be routinely monitored and calibrated exactly like other conventional radiographic equipment. Results must be recorded in a log book. (See AAPM Report #4. Ref. 53).

2) Digital Subtraction System: The performance parameters outlined in Section II should be monitored with a phantom such as in Section III. All parameters should be evaluated on at least a six month schedule. Each day a vessel insert subtraction with the step wedge phantom should be run fixed operating variables and examined for consistency. Results are recorded in a log book.

3) Video Image Hard Copier and Film Processor: The hard copy film should reproduce as closely as possible the image on the viewing monitor. The film processor must be monitored daily as outlined in AAPM Report #4. In addition, it is advisable to photograph and process a digitally stored gray wedge such as the SMPTE test pattern (Ref. 54) each day and compare it visually to a master in order to detect deviations in the hard copier. The density and contrast of the processed gray wedge should be plotted at least monthly and entered in the log book.

REFERENCES

General Review:

1. R.A. Kruger and S.J. Riederer, Basic Concepts of Digital Subtraction Angiography, (Hall, Boston, 1984).
2. C.A. Mistretta, A.B. Crummy, C.M. Strother, "Digital angiography: a perspective", Radiology 139,273-276 (1981).
3. C.A. Mistretta, "The use of a general description of the radiological transmission image for categorizing image enchancement procedures," Optical Eng. 13,134-138 (1974).

Development:

4. R. Brennecke, T.K. Brown, J. Bursch, et al., "Computerized video-image processing with application to cardioangiographic roentgen image series," edited by H.H. Nagel. Digital image processing," (New York, Springer-Verlay, 1977) p.244.
5. T.W. Ovitt, M.P. Capp, H.D. Fischer, et al., "The development of a digital video subtraction system for intravenous angiography," Proc SPIE. 167,61-66 (1978).
6. R.A. Kruger, C.A. Mistretta, T.L. Houk, et al., "Computerized fluoroscopy in real time for noninvasive visualization of the cardiovascular system", Radiology 130,49-57 (1979).

Basic Physics:

7. R.A. Kruger, C.A. Mistretta, S.J. Riederer, "Physical and technical considerations of computerized fluoroscopy difference imaging," IEEE Transactions on Nuclear Science, NS-28,206-212.
8. S.J. Rieder, A.L. Hall, G.S. Keyes, N.J. Pelc, "Contrast sensitivity of digital fluorographic systems," Proc IEEE 1982, Int'l. Workshop on Physics and Engineering in Medical Imaging, IEEE Computer Society, 66-73.

Intravenous Applications:

9. A.B. Crummy, C.M. Strother, J.F. Sackett, et al., "Computerized fluoroscopy: digital subtraction for intravenous angiocardiography and arteriography, "AJR 135,1131-1140 (1980).
10. R.C. Christenson, T.W. Ovitt, H.D. Fisher, et al., "Intravenous angiography using digital video subtraction: intravenous cervicocerebrobascular angiography," AJR 135,1145-1152 (1980).
11. T.F. Meaney, M.A. Weinstein, E. Buonocore, et al., "Digital subtraction angiography of the human cardiovascular system," AJR 135,1153-1160 (1980).
12. C.A. Strother, J.F. Sackett, A.B. Crummy, et al., "Clinical applications of computerized fluoroscopy: the extracranial carotid arteries, " Radiology 136,781-783 (1981).

13. W.A. Chilcote, M.T. Modic, W.A. Pavlicek, et al., "Digital subtraction angiography of the carotid arteries: a comparative study of 100 patients," Radiology 139,287-295 (1981).
14. E. Buonocore, T.F. Meaney, G.P. Borkowski, et al., "Digital subtraction angiography of the abdominal aorta and renal arteries: comparison with conventional aortography," Radiology 139,281-286 (1981).
15. N.J. Hillman, T.W. Ovitt, S. Nudelman, et al., "Digital video subtraction angiography of renal vascular abnormalities," Radiology 139,277-280 (1981).
16. G.D. Pond, R.W. Osborne, M.P. Capp, et al., "Digital subtraction angiography of peripheral vascular bypass procedures," AJR 138,279-281 (1982).
17. D.A. Enzmann, W.R. Brody, S.J. Riederer, et al., "Intracranial intravenous digital subtraction angiography," Neuroradiology 23,241-251 (1982).
18. F.H. Burbank, W.R. Brody, A. Hall, G. Keyes. "A quantitative in vivo comparison of six contrast agents by digital subtraction angiography," Invest Radio1 17,610-616 (1982).
19. G.B.J. Mancini, D.R. Ostrander, R.A. Slutsky, et al., "Intravenous vs. left ventricular injection of ionic contrast material: hemodynamic implications for digital subtraction angiography," AJR 140,425-430 (1983).
20. M.T. Modic, M.A. Weinstein, W. Pavlicek, et al., "Intravenous digital subtraction angiography: peripheral versus central injection of contrast material," Radiology 147,711-715 (1983).
21. W.D. Foley, E.T. Stewart, J.R. Milbrath, et al., "Digital subtraction angiography of the portal venous system," AJR 140,497-499 (1983).
22. F.H. Burbank, "Determinants of contrast enhancement for intravenous digital subtraction angiography (IV-DSA)," presented at the meeting of the Association of University Radiologists, Mobile, AL, March 22-24, 1983.
23. M.L. Wagner, E.F. Singleton, M.E. Egan. "Digital subtraction angiography in children," AJR 140,127-133 (1983).
24. M.T. Modic, M.A. Weinsten, D.L. Starnes, et al., "Intravenous digital subtraction angiography of the intracranial veins and dural sinuses," Radiology 146,383-389 (1983).
25. E. Buonocore, W. Pavlicek, M.T. Modic, et al., "Anatomic and functional imaging of congenital heart disease. with digital subtraction angiography," Radiology 147,647-654 (1983).
26. J.W. Ludwig, L.A.J. Verhoeven, J.J. Kersbergen, T.T.C. Overtoom. "Digital subtraction angiography of the pulmonary arteries for the diagnosis of pulmonary embolism," Radiology 147,639-645 (1983).
27. B.M. Brown, D.R. Enzmann, M.L. Hopp, R.A. Castellino. "Digital subtraction laryngography," Radiology 147,655-657 (1983).

IA Applications:

28. C.M. Strother, M.F. Stieghorst, P.A. Tursky, et al., "Intraarterial digital subtraction angiography," Proc SPIE 3l4,235-238 (1981).
29. A.B. Crummy, M.F. Stieghorst, P.A. Tursky, et al., "Digital subtraction angiography: current status and use of intraarterial injection," Radiology 145,303-307 (1982).
30. F.J. Miller, D.E. Mineau, P.R. Koehler, et al., "Clinical intraarterial imaging using recursive digital filtration techniques injecting only small amounts of contrast material or carbon dioxide," Submitted to Radiology.
31. M. Brant-Zawadski, R. Gould, D. Norman, et al., "Digital subtraction cerebral angiography by intraarterial injection: comparisonwith conventional angiography," AJNR 3,593-599 (1982).
32. M.A. Weinstein, et al., "Intraarterial digital subtraction angiography," Radiology 147,717-724 (1983).

Energy Subtraction:

33. R.A. Kruger, C.A. Mistretta, A.B. Crummy, et al., "Digital K-edge subtraction radiology, " Radiology 125,243-245 (1977).
34. T.L. Houk, R.A. Kruger, C.A. Mistretta, et al., "Real time digital K-edge subtraction fluoroscopy," Invest Radio1 14,270-278 (1979).
35. S.J. Riederer, R.A. Kruger, C.A. Mistretta, "Limitations to iodine isolation using a dual-beam non-K-edge approach," Med Phys 8,54-61 (1981).
36. W.R. Brody, D.M. Cassel, F.G. Sommer, et al., "Dual energy projection radiography: initial clinical experience," AJR 137,201-205 (1981).
37. S.J. Rieder, R.A. Kruger, C.A. Mistretta, "Three beam K-edge imaging of iodine using difference between video fluoroscopic images: theoretical principles," Submitted to Med Phys.
38. S.J. Riederer, R.A. Kruger, C.A. Mistretta, D.L. Ergun, C.G. Shaw, "Three beam K-edge imaging of iodine using differences between video fluoroscopic images: experimental results," Submitted to Med Phys.

Hybrid Subtraction:

39. W.R. Brody, "Hybrid subtraction for improved arteriography," Radiology 141,828-831 (1981).
40. G.S. Keyes, S.J. Riederer, B.F Belanger, W.R. Brody, "Hybrid subtraction in digital fluorography, " Proc SPIE 347,34-41 (1982).
41. A.B. Crummy, M.F. Stieghorst, P.A. Tursky, et al., "Digital subtraction arteriography (DSA) with time-energy (hybrid) subtraction: clinical trials," presented at the Sixty-eighth Scientific Assembly and Annual Meeting of the Radiological Society of North America, Chicago, Nov 27-Dec 3, 1982.

42. M.S. Van Lysel, J.T. Dobbins, W.W. Peppler, et al., "Hybrid temporal-energy subtraction in digital fluoroscopy: preliminary results, " Radiology 147,869-874 (1983).

Temporal Filtering Techniques:

43. R.A. Kruger, "A method for time domain filtering using computerized fluoroscopy, " Med Phys 8,466-469 (1981).
44. R.G Gould, M.J. Lipton, P. Mengers, R. Dahlberg, "Investigations of a video frame averaging digital subtraction fluoroscopic system, " Proc SPIE 314,184-191 (1981).
45. R.A. Kruger, P. Liu, W. Bateman, J. Nelson, "Time domain filtering using computerized fluoroscopy: intravenous angiography applications," Proc SPIE 314,319-324 (1981).
46. J.A. Nelson, F.J. Miller, R.A. Kruger, P-Y Liu, W. Bateman, "Digital subtraction angiography using a temporal bandpass filter: initial clinical results, " Radiology 145,309-313 (1982).
47. R.A. Kruger, P-Y Liu, "Digital angiography using a matched filter." IEEE Trans Med Imag 1,16-21 (1982).
48. R.A. Kruger, F.J. Miller, J.A. Nelson, et al., "Digital subtraction angiography using a temporal bandpass filter; associated patient motion properties," Radiology 145,315-320 (1982).
49. D.R. Enzmann, W.T. Djang, S.J. Riederer, et al., "Low-dose high'-frame rate versus regular-dose low-frame-rate digital subtraction angiography," Radiology 146,669-676 (1983).
50. S.J. Riederer, D.R. Enzmann, W.R. Brody, A.L. Hall, "The application of matched filtering to contrast dose reduction in digital subtraction angiography, " Radiology 147,853-858 (1983).
51. S.J. Riederer, W.R. Brody, D.R. Enzmann, et al., "The application of temporal filtering techniques to hybrid subtraction in DSA, " Radiology 147,859-862 (9183).
52. S.J. Riederer, D.R. Enzmann, A.L. Hall, et al., "The application of matched filtering to x-ray exposure reduction in digital subtraction angiography: clinical results," Radiology 146,349-354 (1983).

Quality Assurance:

53. M. Siedband, et al., AAPM REPORT No. 4: Basic quality control in diagnostic radiology, (American Associates of Physicist in Medicine, New York, 1978).

54. J.E. Gray, K.G. Lisk, D.H. Haddick, et al., "Test pattern for video displays and hard-copy cameras," Radiology 154, 519-527 (1985).