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**DEVELOPMENT AND INVESTIGATION OF SINGLE-SCAN TV RADIOGRAPHY
FOR THE ACQUISITION OF DYNAMIC PHYSIOLOGIC DATA**

**Semi-Annual Report
November 1, 1973 through April 30, 1974**

**National Aeronautics and Space Administration
Research Grant No. NGR 05-009-257***

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INTRODUCTION

In the six month period covered by this report we have conducted research and development activities in six distinct areas all involving the use of a low dose electronic radiography system developed in this laboratory.

The first area concerned the localization and tracking of tantalum screws implanted on the inner wall of the right ventricle of the heart. A computer program was developed to take the position coordinates from an electronically generated cross-hair, frame-by-frame, and then to plot position both absolute and relative of the screws, compute the enclosed area, and plot wall motion (position). Velocity and acceleration can also be computed. Correlation of inner and outer wall points can also be presented. We are presently working on a program which will allow the use of a loosely placed electronically generated window in place of the cross-hair which requires exact operator positioning. The system can be used either with film or on-line with a video disc recorded image.

The second area investigated was the use of the cross-hairs to outline inner or outer heart wall contours. For use with the inner wall the presence of a contrast agent or implanted screws is required. A complete system using the cross-hairs is now operational. Programming for the use of the electronic window is under way. We are also planning to take the next logical step, that is, the use of a simple pattern recognition technique. The system will then be completely automated and capable of processing investigative data on-line so that the experimenter can have the advantage of modifying experimental techniques during the course of each phase of his experiment.

A third area of interest is the use of the system as a quantitative measure of anatomical components which are either stationary in size or changing size dynamically. Data has been accumulated on precision, accuracy, and general capability.

A fourth important area is the utilization of such systems for obtaining dynamic quantitative data from roentgenologic or fluoroscopic procedures. An extensive investigation into such capabilities has been completed.

A major difficulty in commercially available fluoroscopic x-ray systems is the limitation due to size and curvature of the input screen. We have developed a new type of large area flat screen-low light level TV fluoroscopic imaging system.

Finally, we have supplied working drawings and schematics to NASA Ames personnel for implementation of the first phase of their involvement. This first step will allow the processing of data accumulated in past experiments and which has been recorded on 35 mm cine film to be processed expeditiously and accurately.

RESEARCH SUMMARY

1. Localization and Tracking of Implanted Screws

The localization and tracking of screws implanted on the inner wall of the right ventricle of the heart is an important tool for the study of cardiac function, damage, stress, etc. Roentgenologic images can be recorded either on cine film, video tape, or on video disc. For on-line processing of such data the best and most efficacious method is recording on video disc with direct transfer to the computer. However, much data has been recorded using cine roentgenographic techniques. To expedite processing of such data we have devised a system for such processing. The basic system consists of a TV camera operated in the single scan mode,¹ scan converter, cross-hair-window generator, and computer equipped with D-A converters.² Each form (either cine or video disc) is displayed on a TV monitor and the cross-hair positioned at the center of the image of the screw. The coordinates of the cross-hair are then read out by the computer. Successive screws are handled in the same manner. With reasonable care, a positioning accuracy of one part in 200 is easily attained.

-
1. Capabilities of a Single Scan TV-Radiographic System for Digital Data Acquisition. N.A. Baily and R.L. Crepeau, *Invest. Radiol.* 6, 273-279 (1971).
 2. Applications of a Digitalized Radiographic-Fluorographic Processing System to Physiologic Data Extraction. R.L. Crepeau, N.A. Baily, and N.R. Silverman, *Proc. of the San Diego Biomed. Symp.*, 11, 317-322 (1972).
-

The computer program is capable of handling this data in innumerable ways. Some examples are given below.

A. Screw position as a function of time is illustrated in Fig. 1. Each point represents the position of the same screw on a series of frames taken from a cine roentgenographic record.

B. Distance between two screws. This is illustrated by Fig. 2, where the distance between two adjacent screws is plotted vs. time. Each point shown represents the distance between the screws as recorded every $1/30$ of a second.

C. Position of screws from a fixed marker placed either in the roentgenographic or optical field. This is illustrated by Fig. 3. Here position of a series of screws relative to a fixed field marker is given for a series of four frames.

D. To eliminate motion due to respiration, etc., we can plot position relative to one screw. This is illustrated in Fig. 4 where screw position for four frames is plotted with reference to one screw located at the apex of the right ventricle. A, B, C, and D are successive in time. A lateral view computed and measured from the same ciné sequence is shown in Fig. 5.

E. The enclosed area is also easily calculated and plotted. This is illustrated in Fig. 6.

The second step in this part of the program is to allow the operator to substitute a small electronically generated window for the cross-hair. Each picture element included within this window is digitized in accordance with its intensity. Our programs utilize 32 gray levels. At present we can utilize this technique with a number of different algorithms for the localization of the screws. The simplest algorithm for this procedure is variations of what might be loosely termed 'thresholding'. The simplest approach would be to use an averaging procedure over groups of elements (of just sufficient area to cover a screw) and the program then eliminates all such areas below a previously selected threshold. This is entirely satisfactory where the screws are isolated within the window and sufficient contrast is present. In cases of lower contrast we have added a refinement to this algorithm which functions by plotting a histogram of the digitized gray levels. There are two options which can then be selected. The peak area can either be selected manually or the plot can be bypassed and the computer instructed to compute the center of mass of the distribution. As a refinement to this procedure we have added the capability of removing either constant or constant slope, backgrounds so that the levels depicting the screws will be considerably different from other areas in the window and thereby making the selection of the screw location more precise.

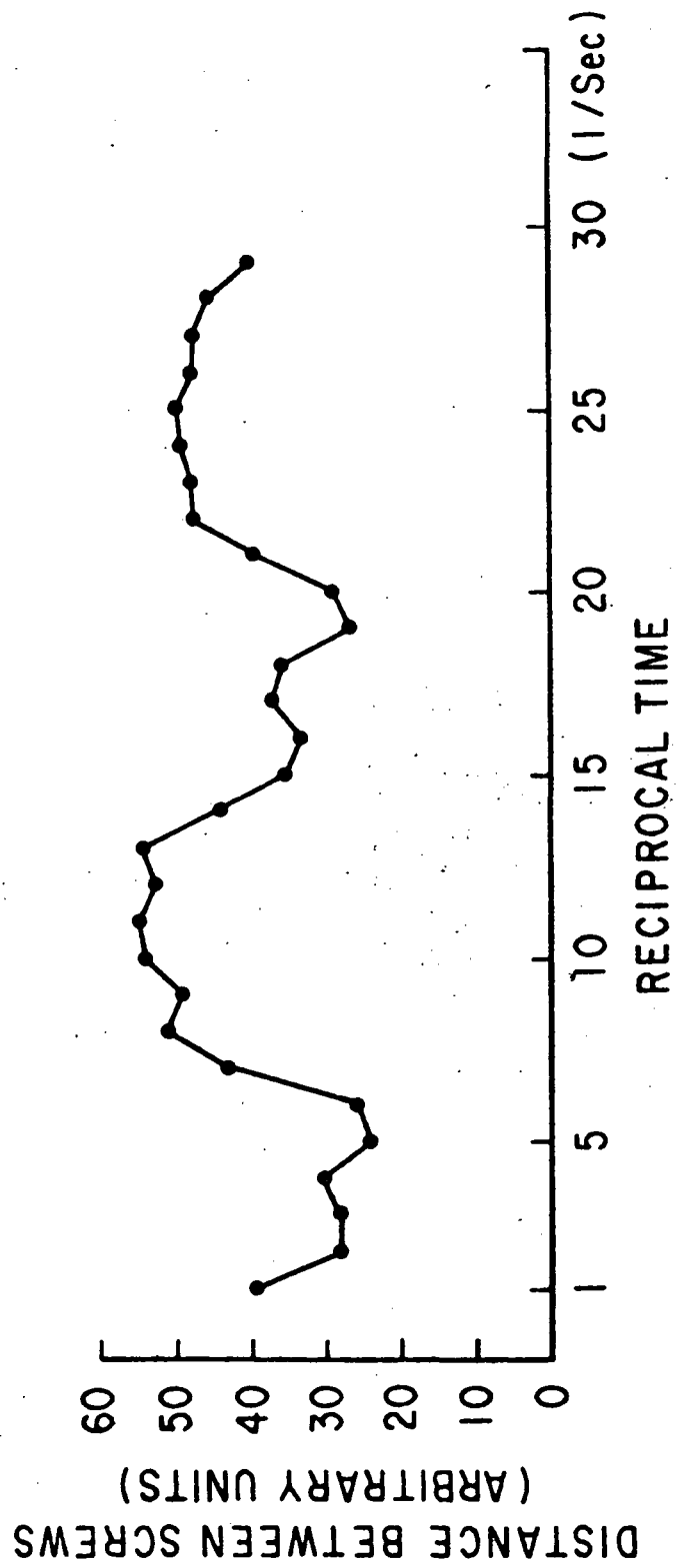


Fig. 2

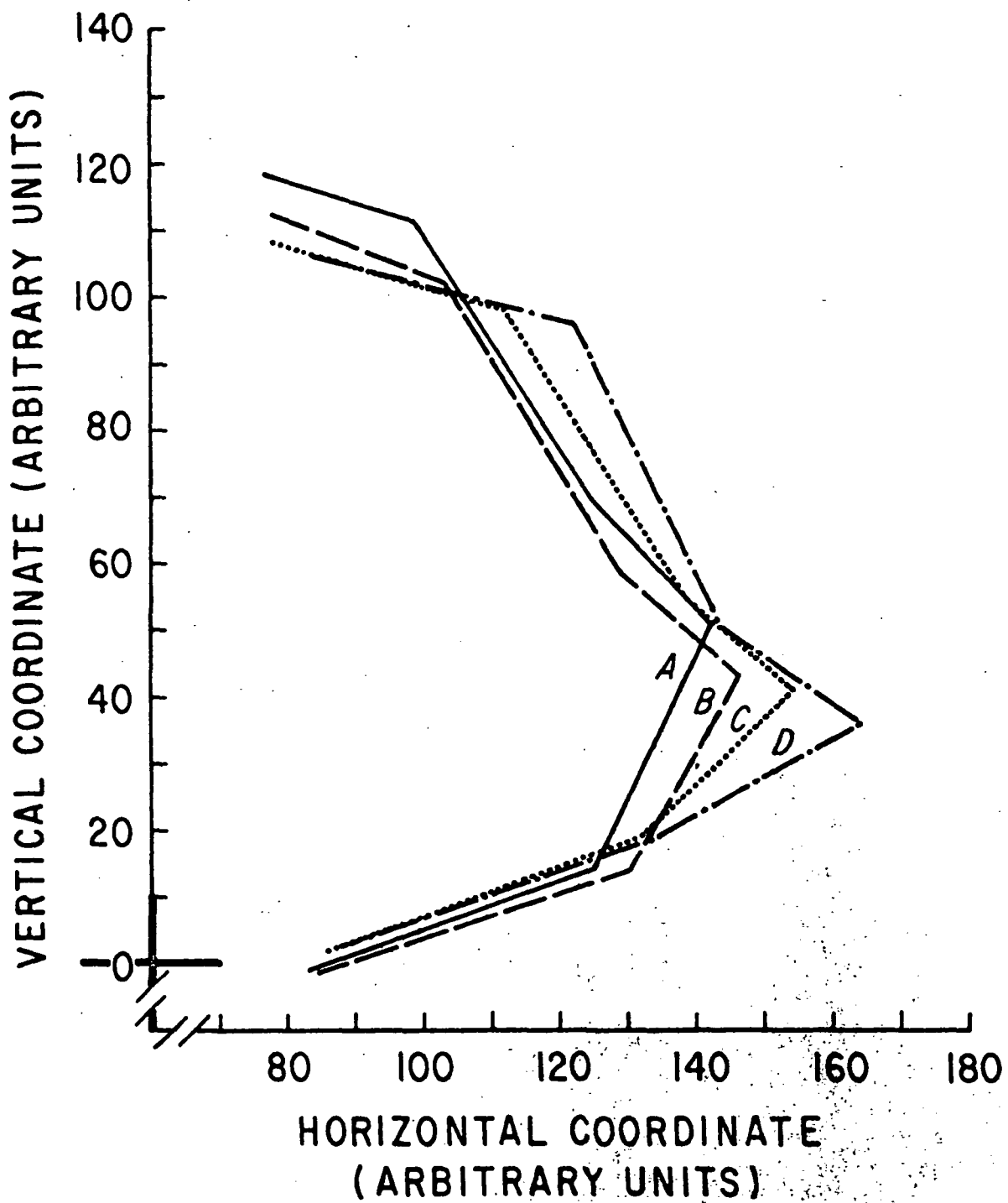


Fig. 3

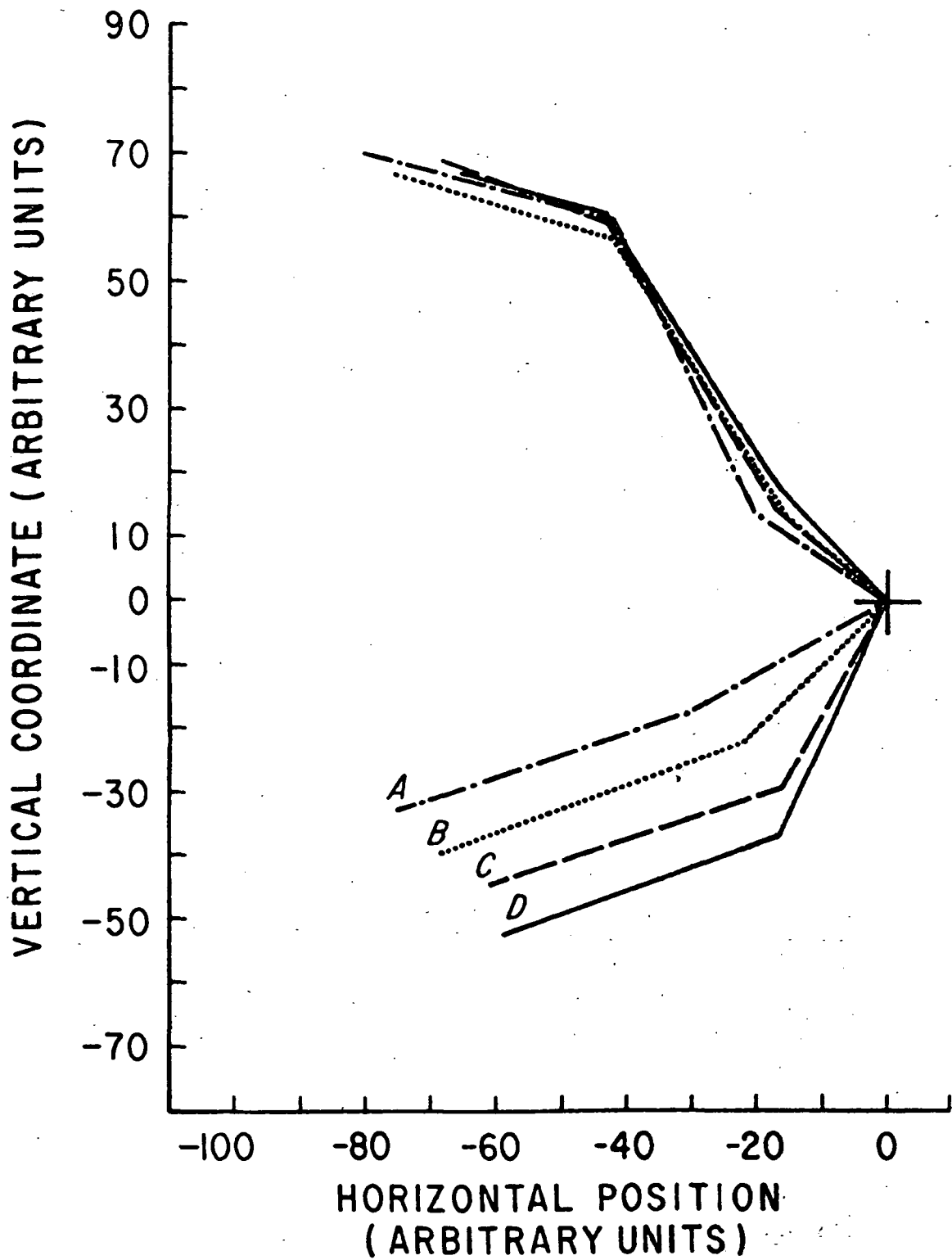


Fig. 4

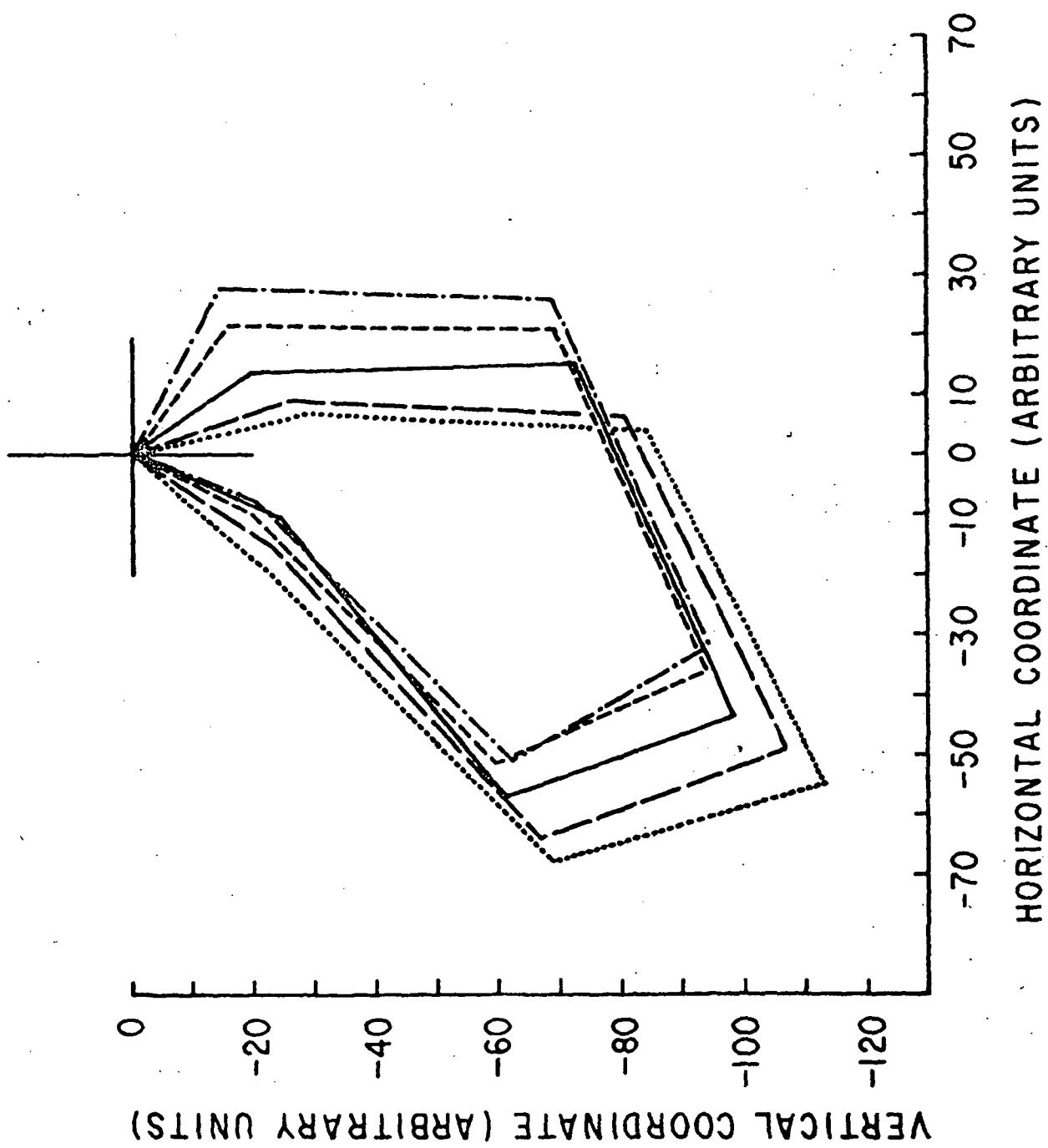


Fig. 5

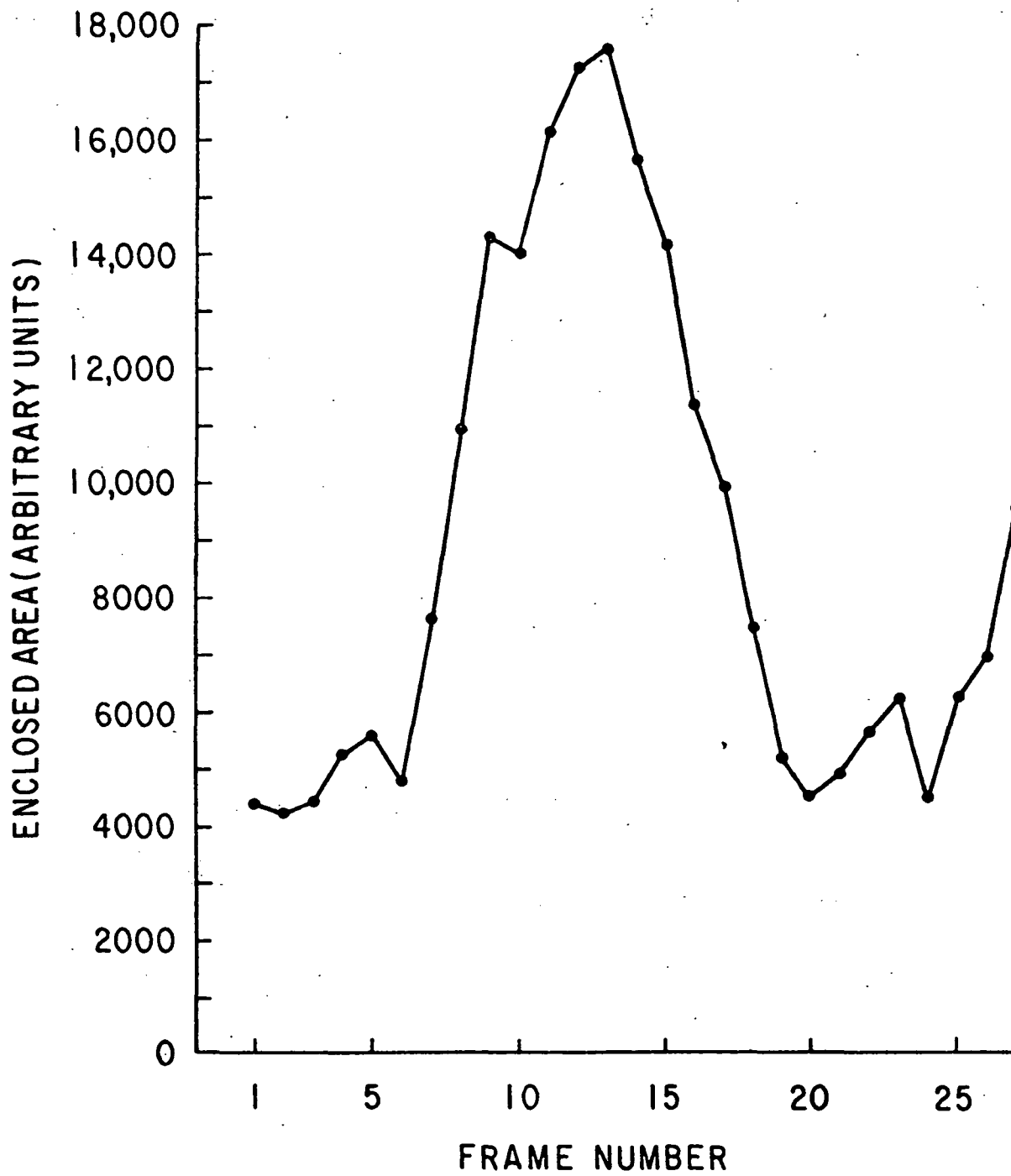


Fig. 6

In the next period we shall investigate, quantitatively, the ability of these programs and algorithms to extract the desired information on screw position. The variables which must be investigated are: size of object, relative contrast between surrounding and object, noise level, and interference by anatomical structures such as ribs, blood vessels, etc. Quantitative evaluation will be done first using a fixed set of digital patterns having noise (random, and statistically generated) superimposed upon the pattern ranging from very low to very high levels, i.e., large to small signal-to-noise levels. A second set will contain, in addition to the noise levels, simulated ribs or arterial structures.

A second phase of this quantitative evaluation will be accomplished through the use of optical patterns identical to those used for the computer simulation. This should provide data identical to that recorded roentgenographically and by electronic radiography.

2. Contour Identification

If screws are implanted, the problem is trivial and the methodology would be identical to the above with the addition of a simple program for connecting the individual points. This technique, however, is more of interest where either the outer wall of the heart, lung outline, or contrast filled chambers or vessels, etc. are desired to be outlined. As a simple approach readily implemented, we have used the cross-hair technique. Figs. 7A and 7B show a contrast filled left ventricle followed over one expansion cycle. The program also computes the enclosed area.

During the next period algorithms employing directional derivatives will be developed. These will make the use of electronically generated cross-hairs or windows unnecessary, thereby simplifying the transition to on-line data processing which will be accomplished directly from the video disc recorded images.

3. Metrology

An important application for any physiologic experiment is the ability of the system to make accurate and precise measurements. These might be cardiac volumes, vessel diameters, pulmonary function, bronchi size, gallstone size and stone volume, etc. A series of experiments designed to measure precision and accuracy are now under way.

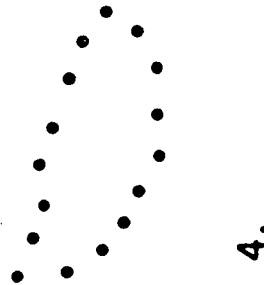
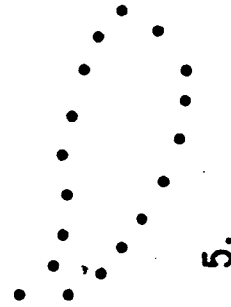
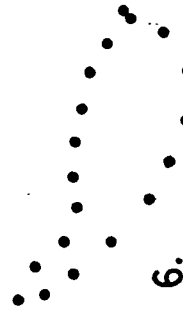
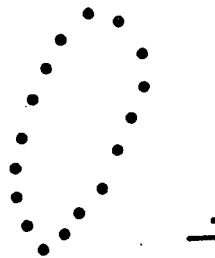
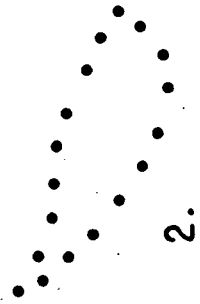
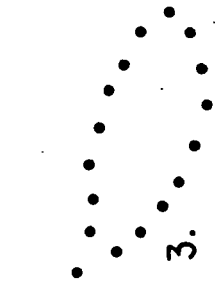
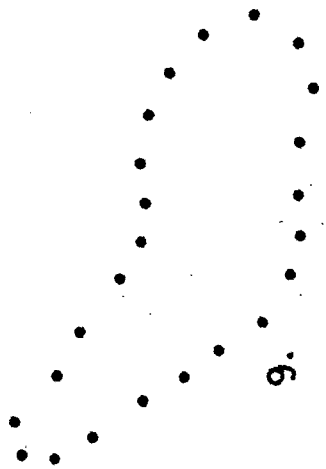


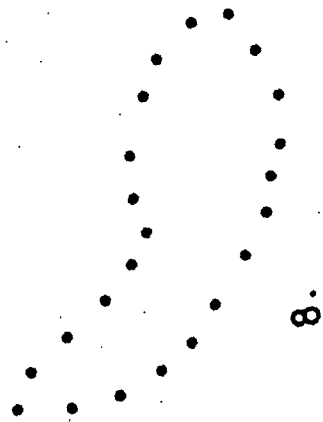
Fig. 7A



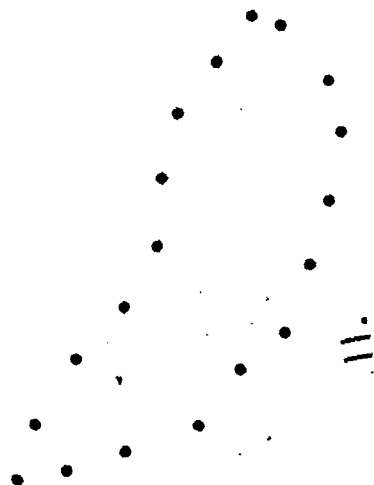
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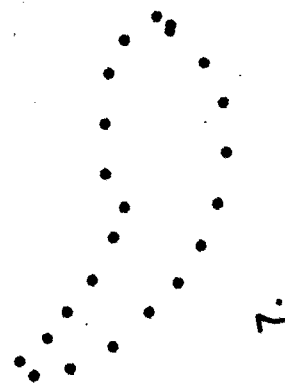
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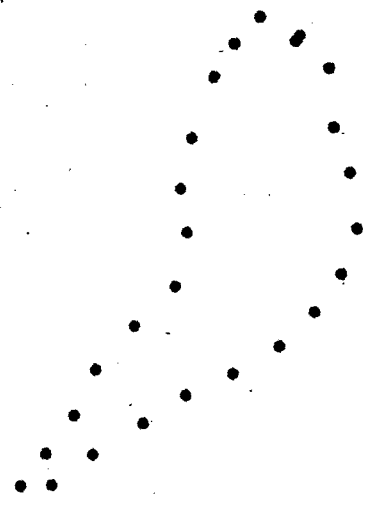
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Fig. 7B

The first studies undertaken were designed to determine the spread in gray levels recorded in the computer memory for a constant amplitude input signal. This will be a function of the system noise and the size of the discrete computer segments chosen. In the case of 32 gray level partition of the 0-1 volt conversion, our system never showed an error greater than one segment. This uncertainty can be reduced to a negligible value by a statistical averaging of the digitized data. This will be illustrated in Appendix A. In brief, however, the reproducibility can be improved to such a degree that very small differences in x-ray absorption can be reliably determined.

The second factor required for accurate measurement is system linearity using constant amplitude signals. Here two distinct portions of the system are involved. First, we are concerned with the relationship of the image size to the pulse height supplied to the A-D converter of the computer. Fig. 8 shows this relationship using the full monitor screen. A slight non-linearity is detectable. Fig. 9 shows this response for the low end of the scale (small images). Here again a non-linearity is evident but in this region it occurs at the low end of the response curve. These non-linearities although slight will be corrected by improvements in the electronics of the window generator.

The portion following this output is the A-D conversion and transfer to the storage bins of the computer. Computer output as a function of size displayed on the monitor is shown in Fig. 10. The non-linearity displayed here is identical to that found previously, indicating that the A-D conversion, storage, and print-out are not significantly contributing to any further non-linearity. The initial portion response (small images) is shown in Fig. 11. Here, although the non-linearity at the low end is similar to that of the window generator, it is perhaps greater. This could be due to noise which is electrically generated and is being investigated further.

During the next period we shall attempt to correct the causes of these irregularities. Although the non-linearity is slight and goes in a regular manner and could therefore be corrected by a very simple computer program we feel that correction of the electronic circuitry is preferable.

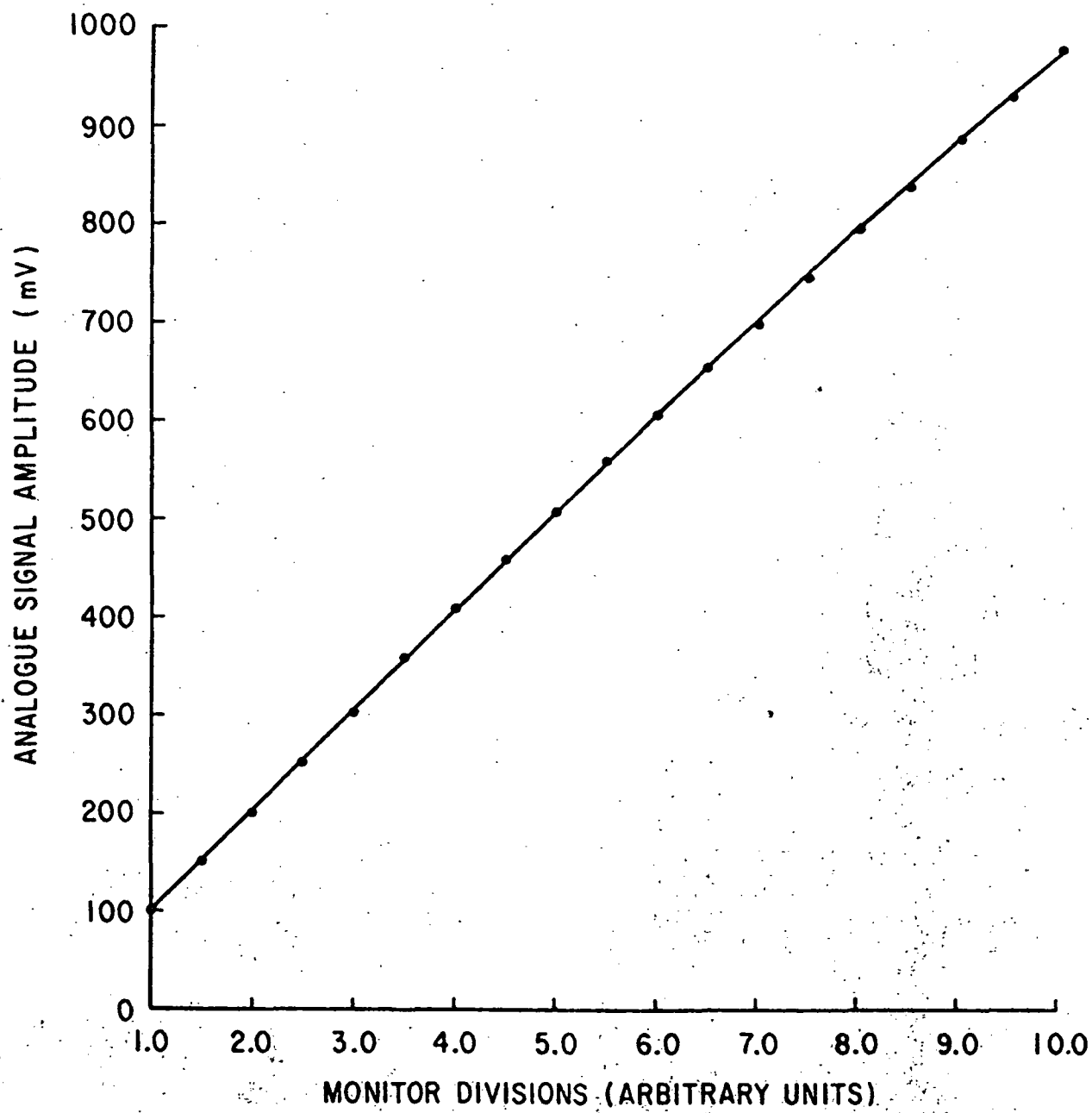


Fig. 8

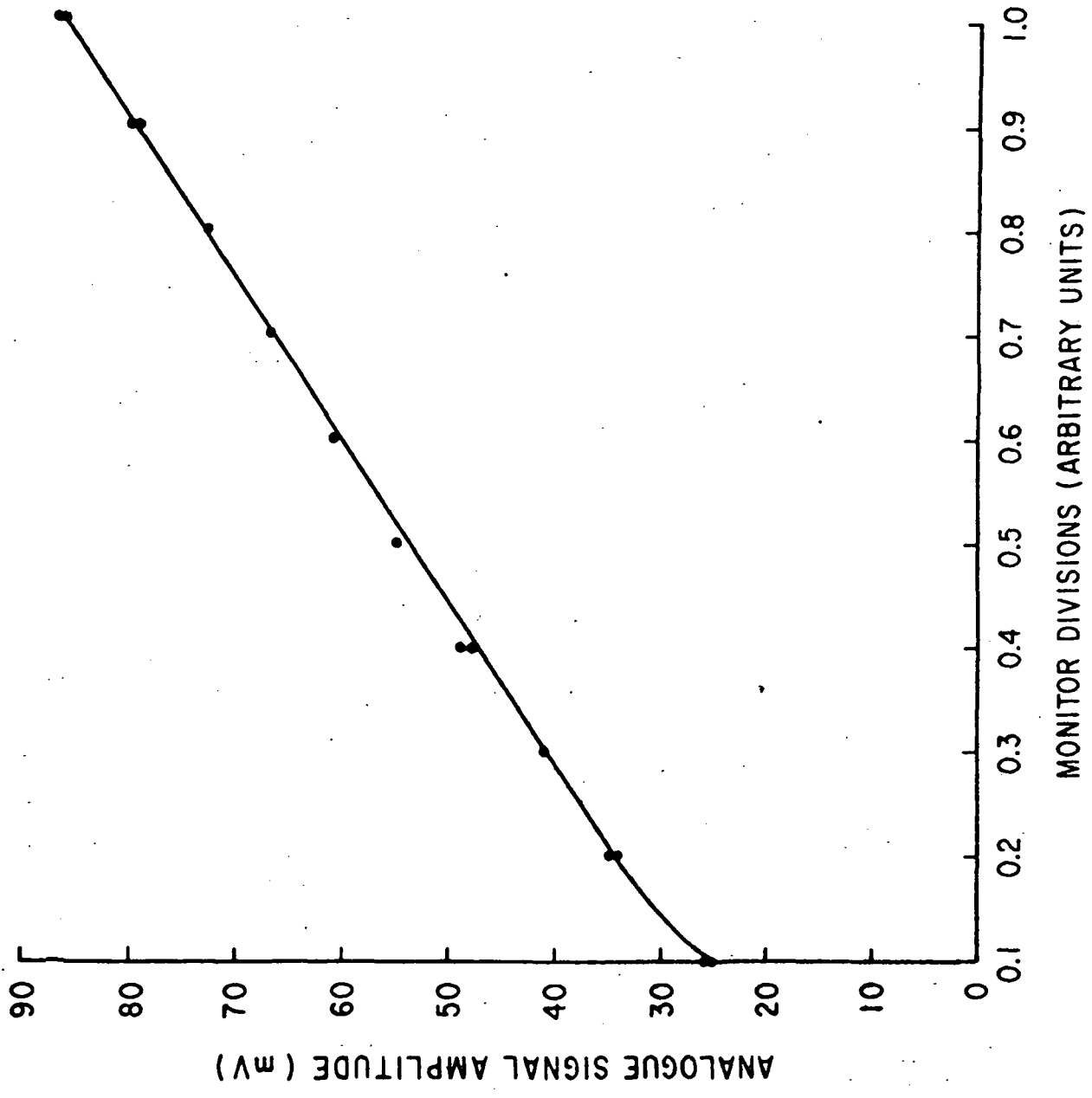


Fig. 9

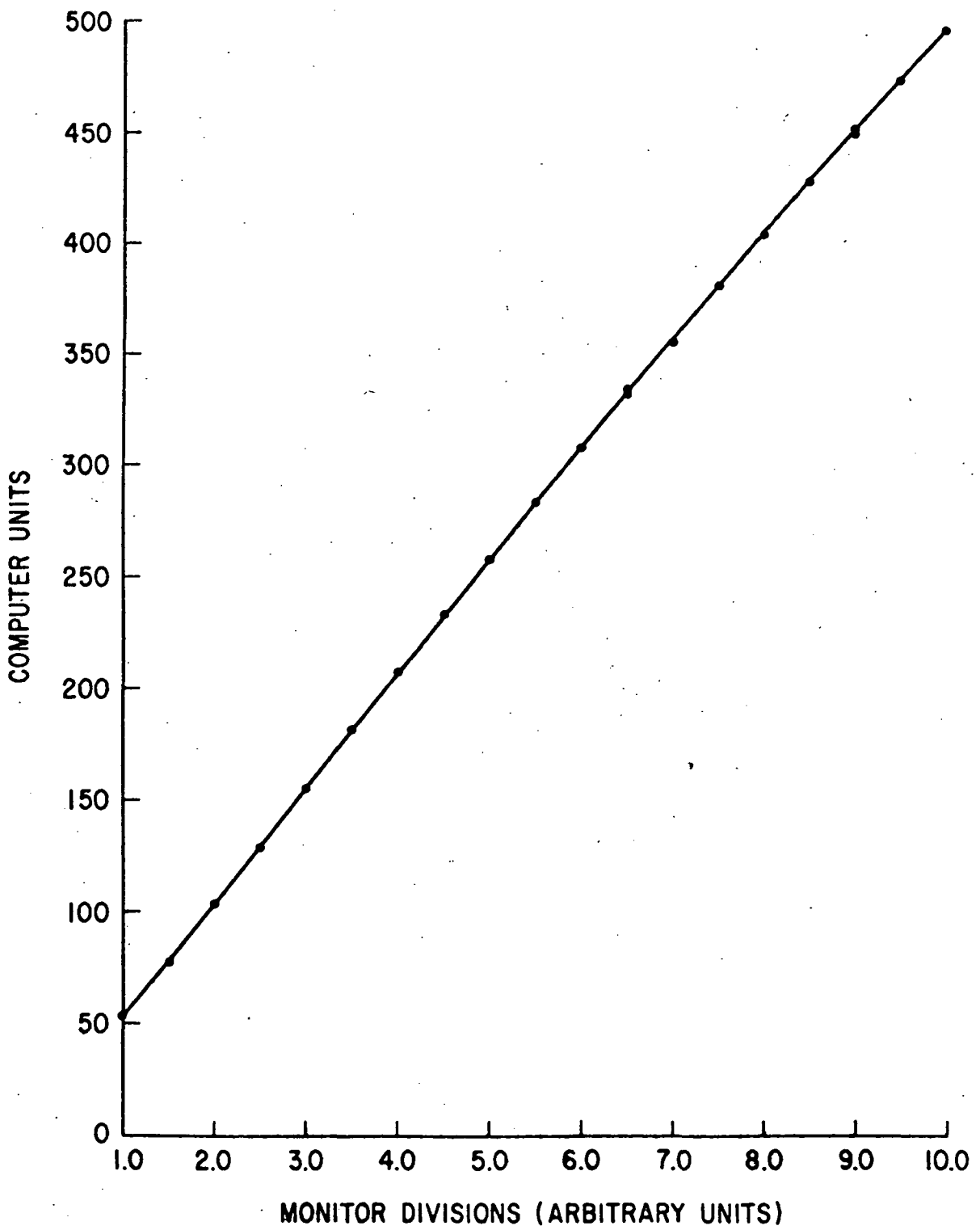


Fig. 10

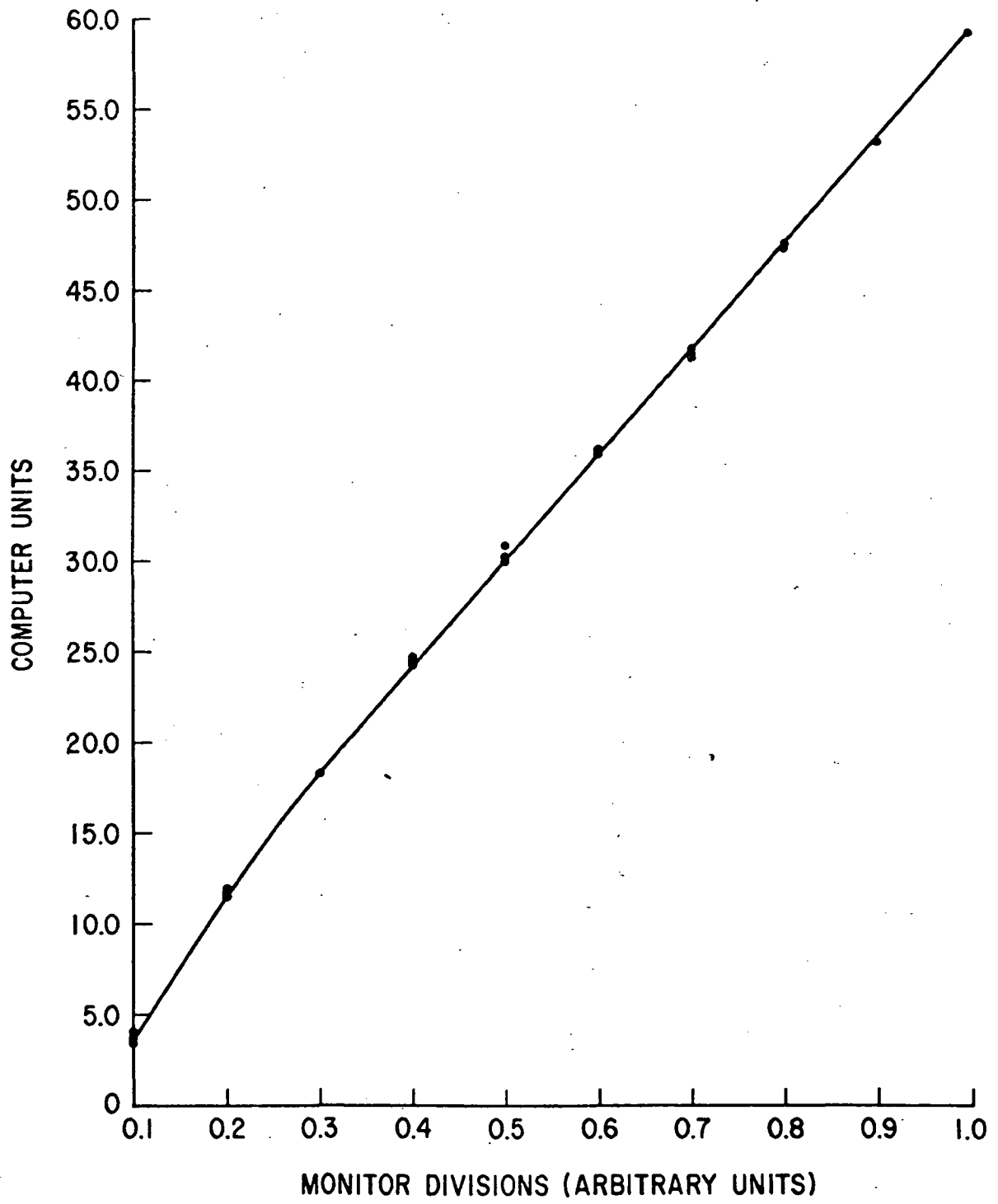


Fig. 11

4. Differential X-Ray Transmission Sensitivity

The degree to which small changes in x-ray transmission between two areas in the image (fluoroscopic) can be detected and quantitated clearly determines the type and amount of physiological data than can be extracted from the image. A paper on this topic has been written and recently submitted for publication to RADIOLOGY. The submitted paper is included in this report as Appendix A.

5. Large Area-Flat Screen-Low Dose-Fluoroscopic Imaging Systems

Two systems of this type have been investigated. The first was a system developed by a commercial laboratory and has been reported upon at a Society of Photo-Optical Instrumentation Engineers (SPIE) meeting. A paper which is attached as Appendix B will be published in the proceedings of this meeting. A second system having a considerably larger input screen (14" x 17") has been built in this laboratory. Also, a three-stage light amplifier was used giving the system considerably higher sensitivity. The three main advantages of these systems are: (1) large input screen allowing a greater anatomical area to be presented; (2) the elimination of screen curvature which would distort quantitative measurements; (3) all components needed can be obtained in ruggedized form making possible the use of this system in centrifuge and space applications.

To date we have only investigated the resolution of the improved system at the center of its input screen and in the axis perpendicular to the x-ray tube cathode to anode direction. The horizontal modulation transfer function was measured for three line rates; 525, 875, and 1225. Each line rate was measured at bandwidth of; 8, 16, and 32 MHz. The results for 525 lines are shown in Fig. 12. The noise level increases with bandwidth and therefore the signal-to-noise ratio is decreased with increased bandwidth. Although the ultimate resolution is not affected, increasing the bandwidth from 8 to 16 MHz does depress the response in the low spatial frequency range. However, increasing the bandwidth to 32 MHz will essentially produce a better transfer function for all spatial frequencies. Curves A, B, and C are the transfer functions for bandpasses of 8, 16, and 32 MHz, respectively. Using an 875 line rate the results shown in Fig. 13 were obtained. Here there is essentially no difference shown for the 8 and 16 MHz data, but a definite improvement in resolution was obtained at 32 MHz bandwidth.

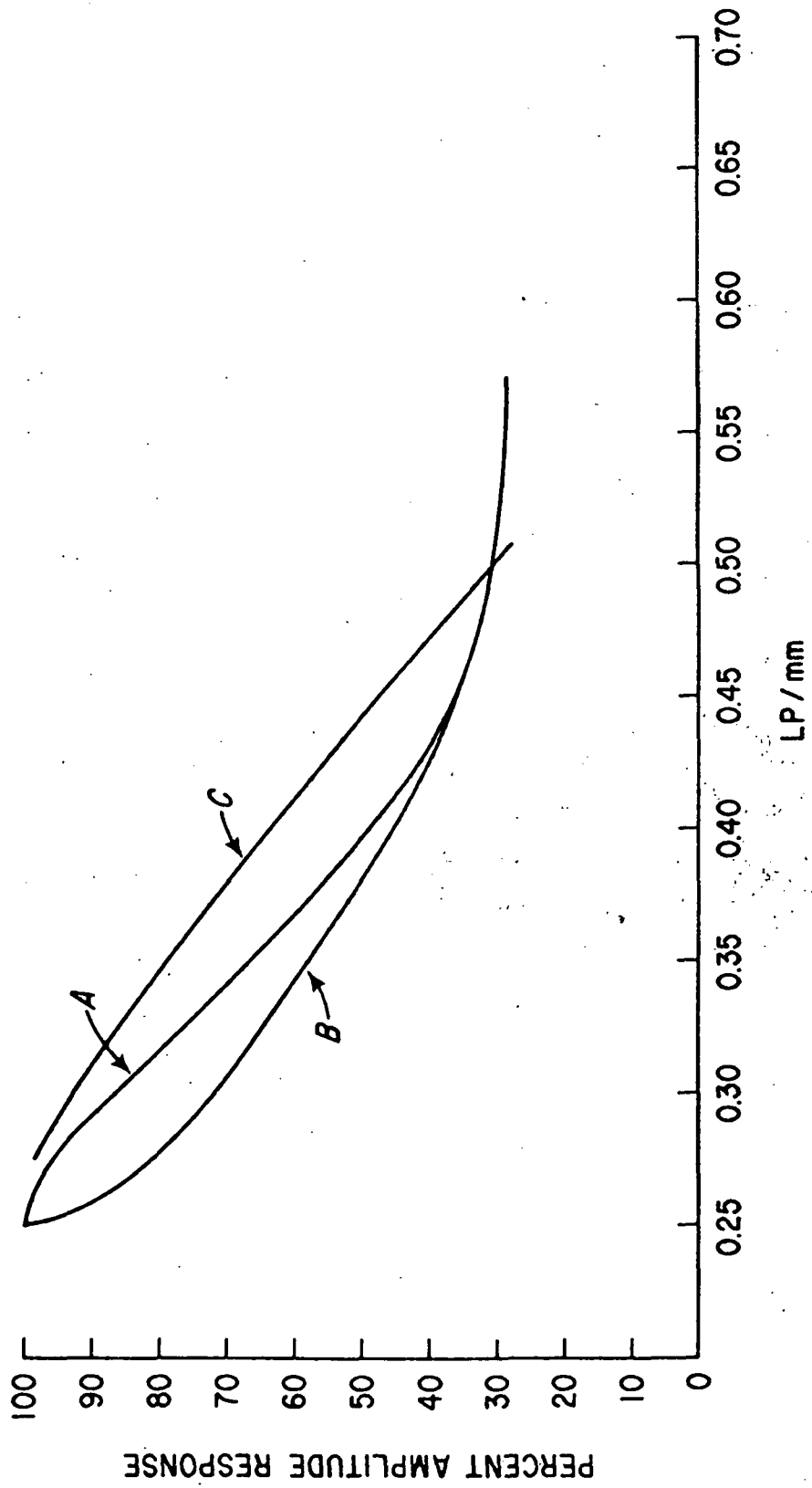


Fig. 12

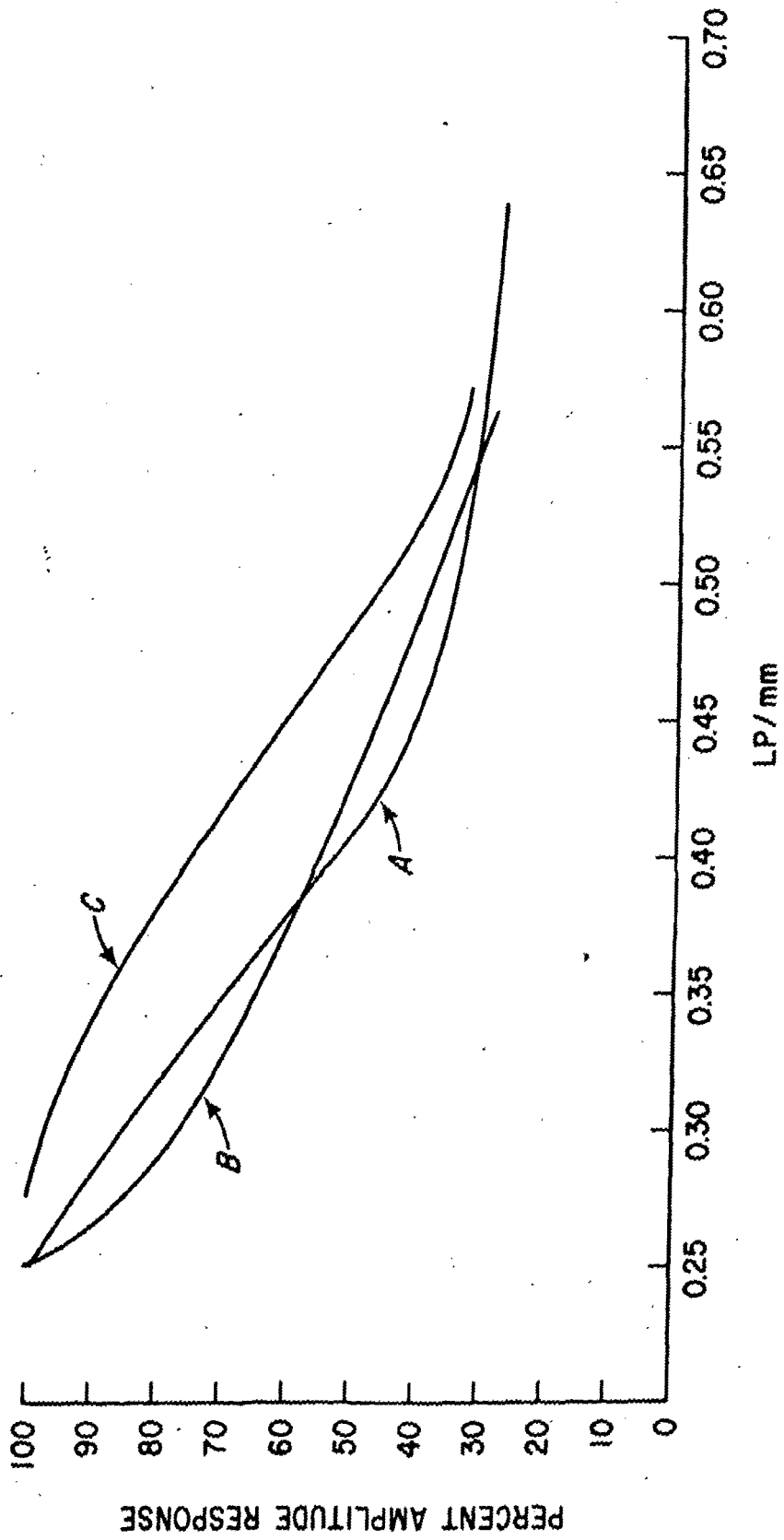


Fig. 13

Curves A, B, and C again represent the data for 8, 16, and 32 MHz bandwidths. The lack of sufficient bandwidth at 16 MHz results in the depressed response at high spatial frequencies in Curve B. This was further demonstrated when the line rate was raised to 1225. Here again, Curves A, B, and C of Fig. 14 represent the transfer functions obtained at 8, 16, and 32 MHz. The vertical resolution is of course governed completely by the line rate except where amplifier bandwidth is limiting.

During the next period several avenues for system improvement will be investigated. For greater resolution and a better transfer characteristic, improved light amplifiers will be investigated. At present, resolution and edge distortion are limited by the performance of this element.

The system noise is primarily being generated in the camera pre-amplifier. It would appear that a factor of between 2 and 10 could be attained here. This would allow images of contrast modulation in the range of 1 to 10 percent to be detected, instead of the 30 percent limitation which we have at present.

Special filters designed to remove high frequency noise are required for use of analog signal processing. These will be investigated and some breadboard designs evaluated.

Contrast and edge enhancement will produce a significant improvement in picture quality. Fuchs, et al,³ have shown that this can be accomplished effectively by a delay line-subtractive technique known as aperture equalization. This approach will be evaluated with both visual and quantitative applications in mind.

3. Electronic Detail Enhancement in Roentgen Television. W.A. Fuchs, V. Messerschmid, V. Herren, and W. Steck, Invest. Radiol. 7, 140-146 (1972).

6. Cooperative Equipment Development

During this period we have supplied drawings and schematics to NASA Ames personnel for the construction and assembly of a system to process x-ray ciné roentgenographic studies previously recorded.

During the next period we shall assist in its assembly and testing. In addition, we shall supply the necessary software required for its computerization. Following this, we will supply software and schematic drawings for on-line operation of a fluoroscopic adaptation of this system so that data will be processed and presented to the investigator during the course of an experiment.

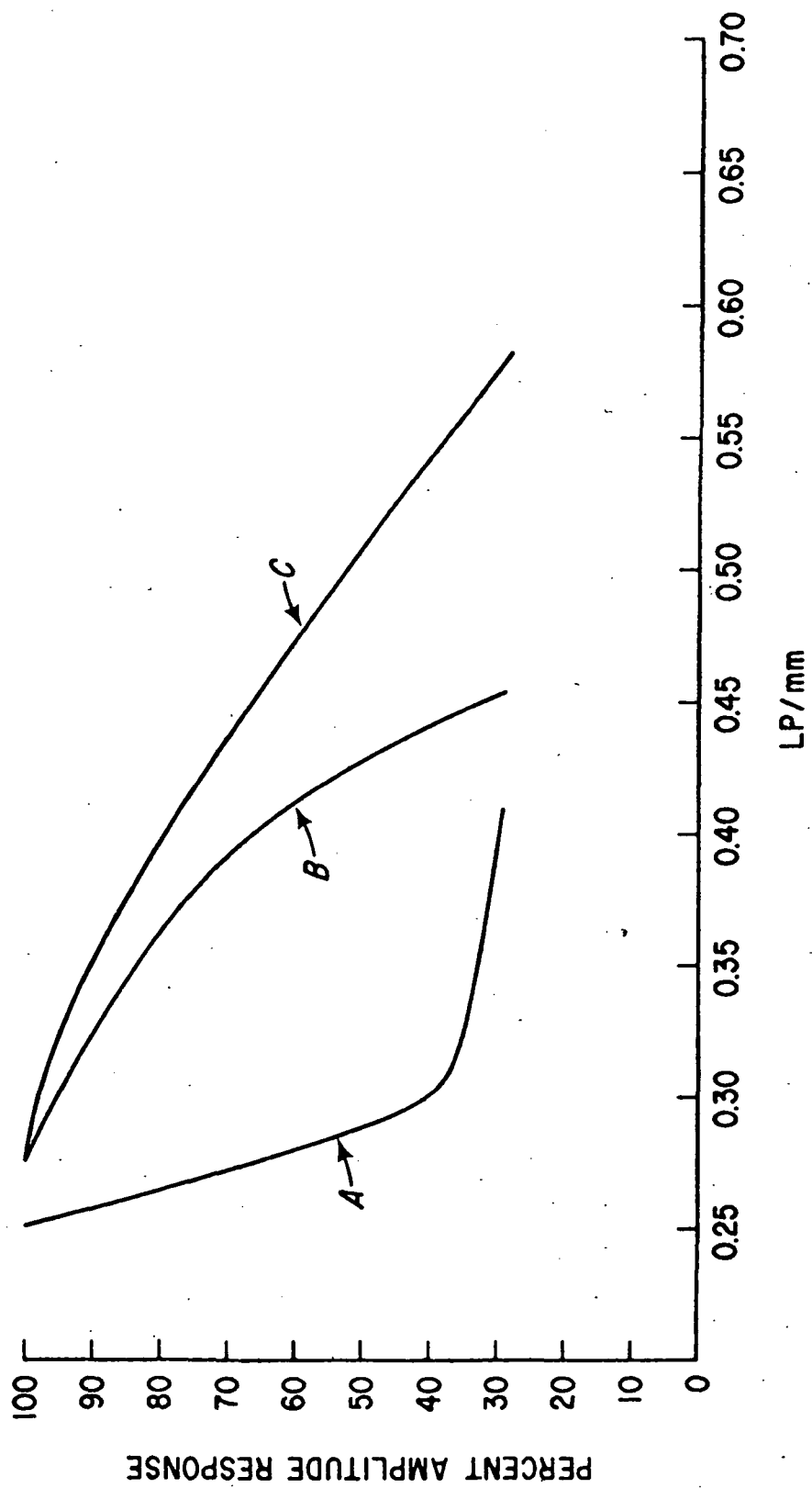


Fig. 14

Following the laboratory trials, a ruggedized system will be developed for use on the man-rated centrifuge to be initially used with animals and, hopefully, followed by human studies. Such studies are made possible because of the system's very high sensitivity combined with computer processing of the image features. This allows many studies to be accomplished at very low dose levels and without the use of invasive techniques.

PAPERS PRESENTED AT MEETINGS

1. Baily, N.A. and Crepeau, R.L., Capabilities of Fluoroscopic Systems To Determine Differential Roentgen Ray Absorption. Radiological Soc. of No. America, Annual Meeting (1973).
2. Baily, N.A. and Crepeau, R.L., Performance of a Large Screen Fluoroscopic Imaging System. S.P.I.E. Seminar, Application of Optical Instrumentation In Medicine II, December 1973.

PAPERS IN PRESS

1. Performance of A Large Screen Fluoroscopic Imaging System. N.A. Baily and R.L. Crepeau, Proc. of the S.P.I.E.
2. Fluoroscopic Tomography. N.A. Baily, R.L. Crepeau, and E.C. Lasser, Invest. Radiol. Mar-Apr (1974).

PAPER SUBMITTED FOR PUBLICATION

N.A. Baily and R.L. Crepeau, Capabilities of Fluoroscopic Systems to Determine Differential Roentgen Ray Absorption, Radiology.

PERSONNEL PARTICIPATING IN PROGRAM

1. Norman A. Baily, Professor of Radiology
2. John E. Steigerwalt, Assistant Research Radiation Physicist
3. Earl M. Raeburn, Lab Technician
4. Elliott C. Lasser, Professor of Radiology
5. Robert A. Keller, Postdoctoral Scholar

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APPENDIX A

CAPABILITIES OF FLUOROSCOPIC SYSTEMS TO DETERMINE DIFFERENTIAL ROENTGEN RAY ABSORPTION *

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ABSTRACT

The capabilities of a clinical fluoroscopic x-ray unit, used in conjunction with a television image digitalization system, for determining differential absorption between two areas in the same field has been investigated. Fractional contrasts and minimum detectability for air, a number of Renografin-60 concentrations, and aluminum have been studied when varying layer thicknesses have been introduced into a water phantom. A number of x-ray potentials and phantom thicknesses have been used.

INDEX TERMS: Fluoroscopy, video-densitometry, digitalization, differential x-ray absorption.

* Work partially supported by NASA Grant NGL 05-009-103.

INTRODUCTION

Roentgenographic and fluoroscopic images contain a good deal of quantitative information which in general is not used by the radiologist or his colleagues in other branches of medicine for either clinical diagnosis or in research programs involving physiological function. In the recent past a number of articles have appeared detailing methods and systems for obtaining quantitative data from roentgenographic or fluoroscopic images. Application of these to cardiovascular and pulmonary studies¹⁻⁵ are typical. However, very little data is available on the capabilities of systems which have been used to quantify radiographic and fluoroscopic images. We have previously published data on resolution, gray scale, and linearity for a simple system designed for digitizing fluoroscopic images.⁶ Some preliminary data on detection capability has been published by the present authors.⁷ Although we have used digital techniques in both our past work and that to be presented in this paper the results should be valid and applicable to comparable analog fluoroscopic-videodensitometric systems.

The process of measuring video signal levels is fundamental to quantitative physiologic or anatomic data extraction from fluoroscopic images. These video signal levels may be measured on either a relative or absolute basis and the measured values used to calculate spatial relationships, and/or dynamic (temporal) changes. The work presented in this paper describes; the system response to changes in roentgen absorptivity, and the minimum detectability of differential x-ray absorption in the presence of a large amount of scatter such as would be present in normal diagnostic radiological applications.

METHOD

A. EQUIPMENT

The basic fluoroscopic system used in all of our experiments was an under-table installation of a 9"-6", G.E. image intensifier with an overall target-to-screen distance of 46 inches. Tabletop-to-screen distance was 6 inches and a 6:1 stationary grid was used. The single scan TV system previously described by Baily and Crepeau⁶ was used to record the fluoroscopic images on a video disc recorder. These were then transferred to the computer by the method described in this same paper. Control of the portions of the video field selected for digitalization was provided by manually variable electronic windows whose positions and sizes are sensed by the computer. With this system, one picture point or resolution cell corresponds to a tabletop area of approximately 0.8 mm (vertical) by 0.9 mm (horizontal) for the 525 line TV system and the geometry described above. For the digitalization process the video range was divided into 32 gray levels.

B. DATA PROCESSING AND METHODOLOGY

The precision and the values of the relative contrast or the change in gray level associated with a change in x-ray beam attenuation are a function of the scatter, noise, and the dc level associated with the recorded video signal.

The process of obtaining quantitative information from a fluoroscopic or roentgenographic image involves the measurement of electronic signal levels. Quantitation consists of measurements of changes in a videometric level associated with a change in x-ray attenuation within the object under examination. As with any measurement, the measurement of a videometric level is subject to statistically random fluctuations. These fluctuations when unrelated to true absorbtivity changes constitute the noise of the system.

Such measurements may be applied to at least two basic types of mass density change bringing with it a change in x-ray absorbtivity. In the first instance we might have a variation with respect only to position, such as a contrast filled vessel occupying one particular region of the entire image. In the second case, we might have a similar image changing in time, such as the volumetric changes associated with heart or lung function.

In the first case, the precision with which the signal can be measured will influence how well one can locate edges/boundaries or the precision with which one can compute diameters, volumes, etc. In the second case the ability to measure the signal accurately will directly affect our ability to accurately track videometric level changes associated with physiological functions. In some cases, such as the tracking of the motion of the heart wall, both types are involved.

The precision of the measured value of the relative contrast or the change in videometric level associated with a change in x-ray beam attenuation is a function of the system noise. The system noise originates from numerous sources. These include fluctuations in x-ray tube output, scattered radiation, image amplifier/TV noise, recorder noise and digitalization noise. These noise sources place a real limitation on the precision to which either digital or analog measurement of video signal levels can be made.

The most straightforward method for measuring a change in videometric signal level associated with a change in x-ray attenuation is to observe or compare the levels associated with two 'single' points, i.e., the smallest resolvable elements (resolution cells) of the system. Single-point analysis, however, has the major disadvantage of providing the least precise measurement. The advantage of such single-point analysis is that maximum spatial resolution is obtained.

An increase in accuracy can usually be obtained by averaging more than one resolution cell. One averaging method that can be employed involves comparing the videometric levels of two resolution cells, each of which represents the average of that resolution cell's signal level when sampled at different times. This technique reduces the statistically random system noise, preserves the maximum system resolution, but decreases the temporal resolution at the points involved. No true signal change is required to have taken place during the period of the averaging.

A second method useful for image contrast measurements involves averaging numerous resolution cells in the two areas of interest. This averaging technique preserves temporal resolution, but results in a loss of spatial resolution.

Noise or videometric fluctuations mainly originate in one or all of the three principal steps involved in the production and analysis of the fluoroscopic image. These steps are: Fluoroscopy; the image recording; and, the image playback and digitalization procedure. The different modes of averaging have varying degrees of efficiency with respect to the types or sources of noise. The noise generated at each of the phases is amenable to reduction by the appropriate type of averaging. For instance, averaging the mean videometric levels of a region over several digitalizations of the same recording can improve the estimate of the mean videometric level recorded, but can do nothing to reduce the effects of the recorded noise. Increasing the area over which the mean videometric level is computed will reduce some of the image formation and recording fluctuations, and the image playback and digitalization noise effects, but cannot improve the situation with respect to fluctuations caused by the x-ray generating equipment.

One artifactual nuance present in typical fluoroscopic systems is the x-ray field inhomogeneities, resulting in a non-uniform videometric level versus position within the field. The field inhomogeneities are due to inverse square effects taking place in the x-ray beam, non-flat image intensifier input screens, image intensifier distortion, non-uniform scatter, vignetting, and heel effects. These effects are illustrated in Fig. 1A. This is a digitized recorded fluoroscopic image of a 10 cm x 10 cm field at the surface of an 11 cm thick polyethylene slab whose dimensions were 30 cm x 30 cm. This particular cross section represents the recorded field approximately at its vertical center. Each point is an average of twelve consecutive picture points plotted at the geometric midpoint of the twelve.

To overcome this non-uniformity and to reduce the effects of recorded noise, we have used a subtraction method to improve the ability of making precise and sensitive measurements. That is we have recorded a field such as shown in Fig. 1A and subtracted it from that recorded, for example, with an absorber in the field. A digitized recording made with a 0.05" aluminum absorber in half of the field is shown in Fig. 1B. Subtracting these two on a point-by-point basis yielded the results shown in Fig. 2. Although Fig. 2 shows a flat field (after the subtraction) in both halves, this will not always be the result. This would not be the case, for example, if amplifier gains, x-ray beam intensity, etc. were to shift during the interval between recordings.

In order to overcome the effects of this non-uniformity when measuring video levels, a normalization process was employed. This normalization process involved recording the fluoroscopic image of interest, then recording an image representing a uniform absorbing material. This latter recording was made under identical circumstances as the former recording. Then, on a point-by-point basis, using digital processing, the phantom of interest was normalized to the uniform phantom. By this technique any aberrations resulting in a non-uniform image are eliminated and comparisons of the videometric gray level in various portions of the field can be made with a reasonable degree of accuracy.

RESULTS

A preliminary experiment was conducted to verify the efficacy of averaging to improve the precision of videometric gray level measurements. A region at approximately the-center of a recorded x-ray field was selected.

To determine the fractional standard deviation of the measured mean video levels due to the noise of the image playback and digitalization portions of the system, a region approximately in the center of a recorded x-ray field was selected. This region was varied in size and for each size digitized 16 times. Each time a mean value, the standard deviation about the mean value of the 16 means, and the fractional standard deviation (standard deviation divided by the mean of the 16 means) were calculated. These were then plotted vs. the size of the region. The results are shown in Fig. 3. The solid curve is the least squares fit to a theoretical curve whose value decreases inversely with the square root of the number of picture points averaged for each area. The experimental data was processed in two different ways. First they were averaged over successively larger areas. Second, in lieu of averaging a large area, a smaller area was averaged a given number of times so that the total number of picture elements were equal to the larger area. We found that the resultant fractional standard deviations were identical in both instances. These values then place a limit on the precision with which signals generated by the fluoroscopic image can be measured from recorded images using the type of system discussed in this paper and described previously.⁶

In reporting systems response and minimum detectability, averaging was used over 320 picture elements (1.5 cm^2 tabletop area) on six individual recordings. In this manner, recorded, playback and digitization random noise fluctuations were minimized.

Experiments were then conducted to determine the minimum detectability and response of a clinical fluoroscopic system with a phantom which divided the image field into two different contrast levels, and provided a realistic clinical condition. The experimental set-up is shown in Fig. 4. The target-to-tabletop distance was 40 inches. The dimensions of the water phantom were 50 cm x 50 cm x 50 cm. In all experiments the field size was 10 cm x 10 cm at the tabletop. Various thicknesses of absorbers or air cavities were placed so as to cover half of the x-ray field and centered at the midpoint of the water scatterer. The water depth ranged between 10 cm and 20 cm, thereby introducing scattering comparable to that which would be present in a clinical situation. The contrasts calculated are all referred to the unperturbed half of the field after normalization as described above. Experiments were conducted using pure aluminum (as an approximation for bone), Renografin-60, and air.

The results are shown in Figs. 5 through 8. The values of the ordinate were computed by selecting two adjacent areas occupying a tabletop area of approximately 1.5 cm^2 (320 picture elements) each area having its vertical center line at the center of the x-ray field and interior edges 1 cm from the absorber or cavity edge. The mean gray level was calculated for each area and these means used to calculate the contrast between the areas.

Fig. 5 shows the response curves obtained for aluminum absorbers. A was obtained using 120 kVp and a water depth of 20 cm. B represents the data for 100 kVp and 15 cm of water. C was obtained using 90 kVp and 10 cm of water.

Fig. 6 shows the results obtained for various concentrations and thicknesses of Renografin-60. A is for a 10 percent solution, B for 25 percent, and C for 50 percent. The thickness of the water phantom was 20 cm and the tube potential 120 kVp. The Renografin-60 solutions were contained in plexiglass containers having a thickness of 0.25". These were centrally supported in the water scatterer on a platform of 0.25" polystyrene. Equal thicknesses of these materials were placed in the other half of the field. The water level was kept fixed for all recordings.

Figs. 7 and 8 show the results obtained with various thicknesses of air cavities introduced into the water phantom. The geometry was identical to that used to obtain the Renografin-60 data. Fig. 7 shows the complete response while Fig. 8 shows the data obtained for thin cavities on an expanded scale. The data in Fig. 7 was obtained using 120 kVp and 20 cm water. Curve A, in Fig. 8 is the same data plotted on an expanded scale, while B is that obtained using a potential of 100 kVp and 20 cm water.

Using the slope of the initial linear portions of these curves (or straight lines) and the standard deviation of the experimental points, one can obtain a value for the minimum degree of contrast that can be reliably differentiated in each case. A value of plus or minus two standard deviations was assumed necessary to reliably determine a change in image composition. The results are shown in Table I. In all cases less than 1 mm of material can be detected even in the presence of a large scattering volume. In previous work it was found that the addition of a 12:1 moving grid was found to improve the contrast of the image substantially. However, in clinical use this would, of course, increase the patient dose.

DISCUSSION

In all of our experiments the videometric response, when treated as contrast, shows a linear response with absorber thickness up to considerable thicknesses. This is not a case of small absorber thickness giving a truly exponential response and being approximated by a linear function as the data does not plot linearly when treated semi-logarithmically. If we ignore scattering and other perturbing factors such as curvature of the input screen and inverse square effects, the contrast or relative video signal levels are given by:

$$C = 1 - e^{-\mu_2 t} e^{-\mu_1 t} \quad (1)$$

where; μ_2 = attenuation coefficient of water
 μ_1 = attenuation coefficient of absorber
 t = absorber thickness.

For small t , this reduces to:

$$C = \mu_1 t (1 + \mu_2 t) - \mu_2 t \quad (2)$$

At larger absorber thicknesses the contrast relationship does not remain linear but decreases in the case of absorbers which are more dense or have a greater linear absorption coefficient than water and increases more rapidly with cavity size in the case of air. In the manner in which our data was computed Equation (1) becomes for air cavities,

$$C = 1 - e^{-\mu_2 t} e^{\mu_1 t} \quad (1')$$

and for small cavities,

$$C = \mu_2 t (1 - \mu_1 t) - \mu_1 t \quad (2')$$

where; μ_1 = attenuation coefficient of air

t = cavity thickness

The minimum detectable contrast seems to be mainly limited by scatter and electronic noise. Both of these are amenable to improvements in the experimental set-up. However, decreasing scatter by the addition of a grid with a higher ratio would involve an increase in patient dose. Signal-to-noise ratios can be increased by any number of electronic improvements in circuitry, system configuration, and digitization equipment.

The linearity of the system's response to changes in contrast over a rather large range of clinically interesting values will tend to make quantitative determinations simple, accurate, and allow extrapolation of experimental data for other calculations. The values given in this paper would certainly be different using another system or with a change in

components. However, we feel the values given, the form of the system's response, and minimum detectable contrasts are typical of what can be expected from this type of system.

CONCLUSIONS

1. The response of digitized fluoroscopic imaging systems is linear with contrast over a rather wide range of absorber and cavity thicknesses.
2. Contrast changes associated with the addition of aluminum, iodine containing contrast agents and air of thicknesses 1 mm or less can be detected with a 95% confidence level.
3. The standard deviation associated with such determinations using clinically available x-ray generators and video disc recording should be less than 1 percent.

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TABLE I-A

MINIMUM DETECTABLE THICKNESS - t_{min}

Material	Water Depth (cm)	Tube Potential (kVp)	Standard Deviation (percent)	t_{min} (mm)
Al	10	90	0.849	0.6
Al	15	100	0.465	0.5
Al	20	120	0.723	0.6
10%, Renografin-60	20	120	1.10	0.8
25%, Renografin-60	20	120	0.578	0.2
50%, Renografin-60	20	120	0.828	0.1
Air	20	100	0.797	0.4
Air	20	120	0.802	0.9

FIGURE CAPTIONS

Fig. 1-A: Digitized video signal level as recorded after penetration of 10 cm x 10 cm x-ray beam through 11 cm of polyethylene scatterer. A was recorded with only the scatterer in the field while B was recorded with the addition of 0.05" of aluminum in one-half of the field.

Fig. 2-A: Video signal levels after subtraction of Fig. 1A from Fig. 1B.

Fig. 3-A: Fractional standard deviation versus area averaged.

Fig. 4-A: Water phantom used to determine system response to variations in contrast levels.

Fig. 5-A: System response to varying contrast levels produced by aluminum absorbers. A: 120 kVp, 20 cm water scatterer. B: 100 kVp, 15 cm water scatterer. C: 90 kVp, 10 cm water scatterer.

Fig. 6-A: System response to varying contrast levels produced by several concentrations of Renografin-60. The water depth was 20 cm and the tube potential, 120 kVp. A, 10 percent concentration; B, 25 percent, and C, 50 percent.

Fig. 7-A: System response to varying contrast levels produced by air cavities when introduced into 20 cm of water using 120 kVp.

Fig. 8-A: System response to varying contrast levels produced by air cavities when introduced into 20 cm of water. Curve A, is for 120 kVp (same data as used in Fig. 7) and Curve B, is for 100 kVp.

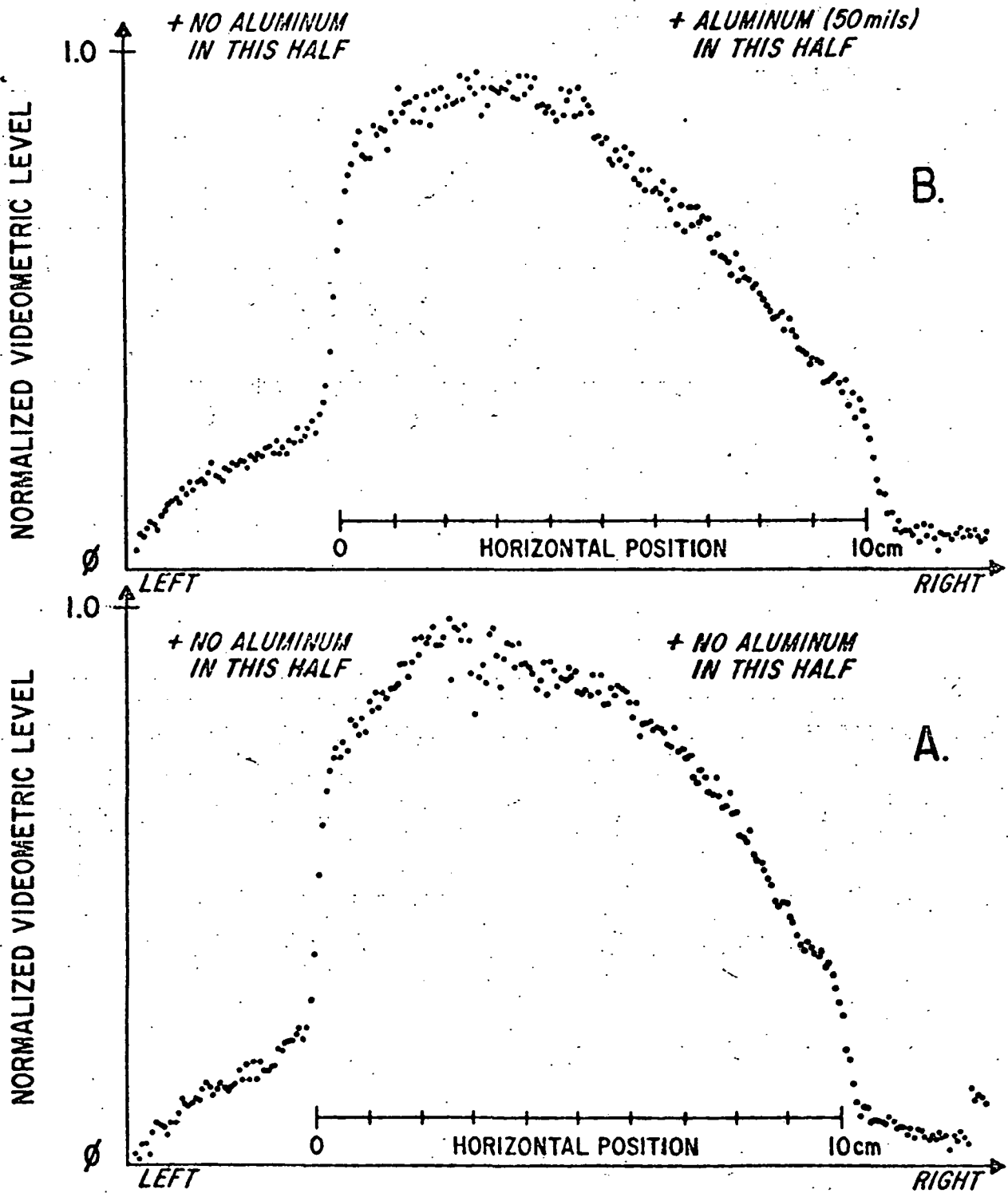


Figure 1-A

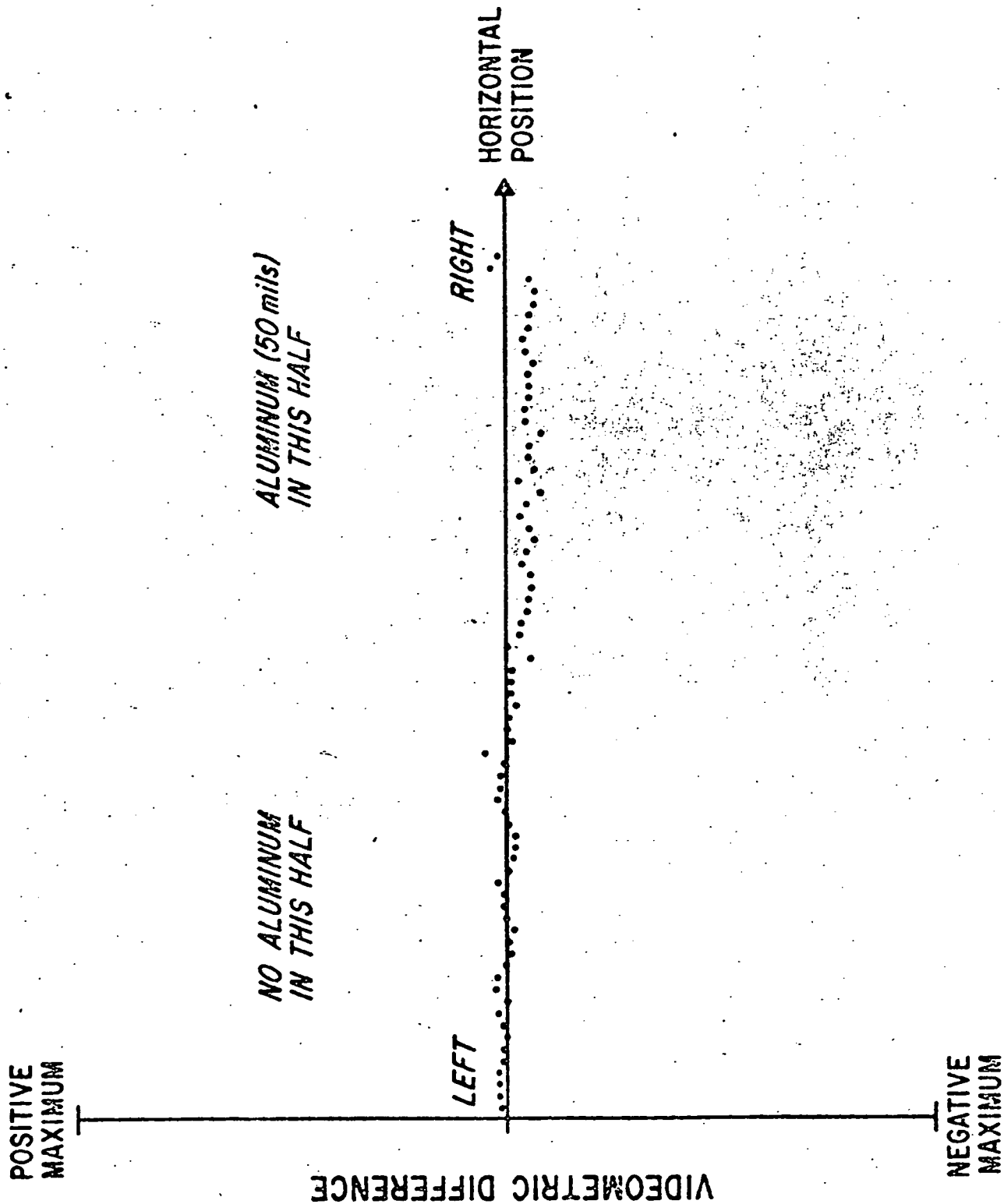


Figure 2-A

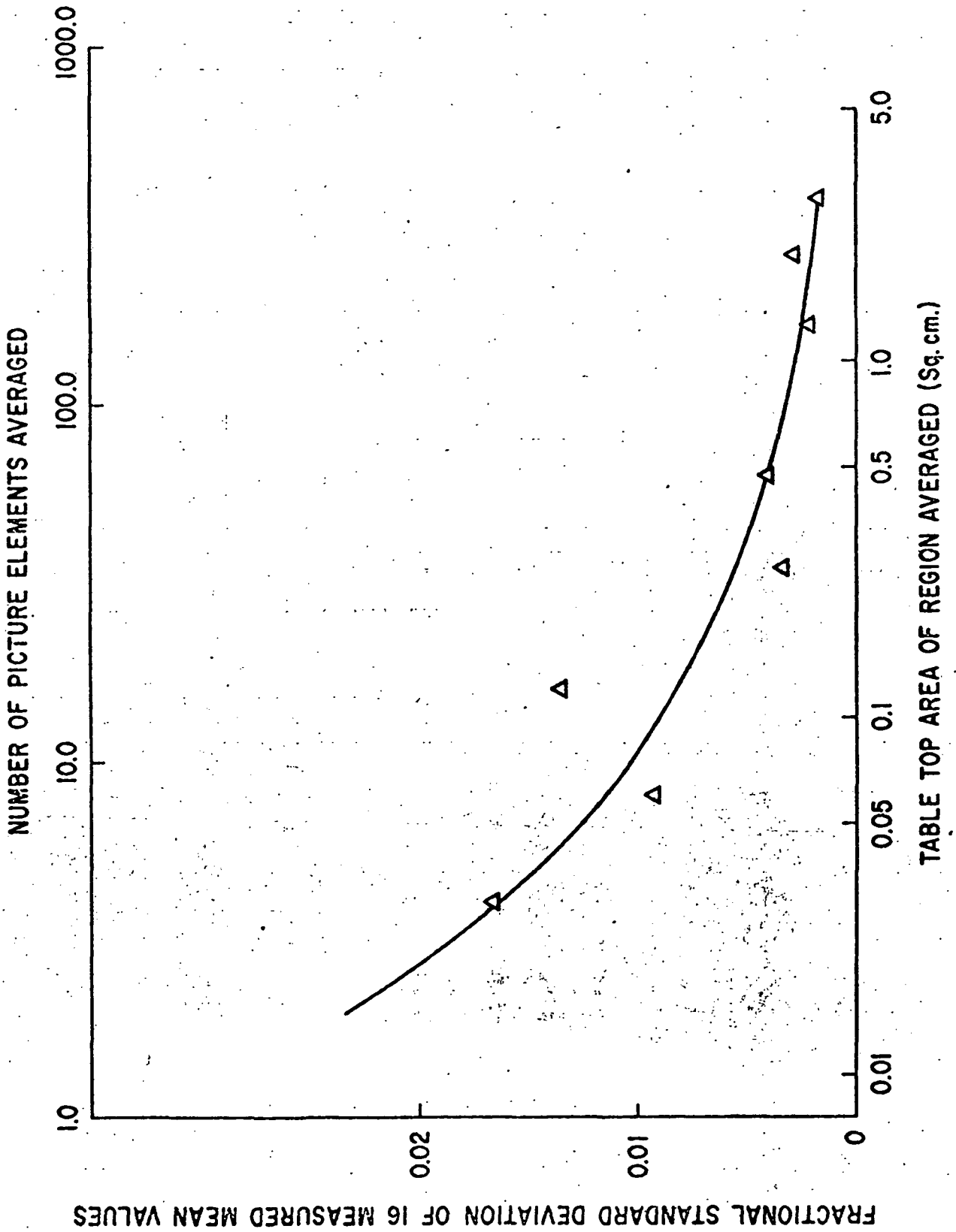


Figure 3-A

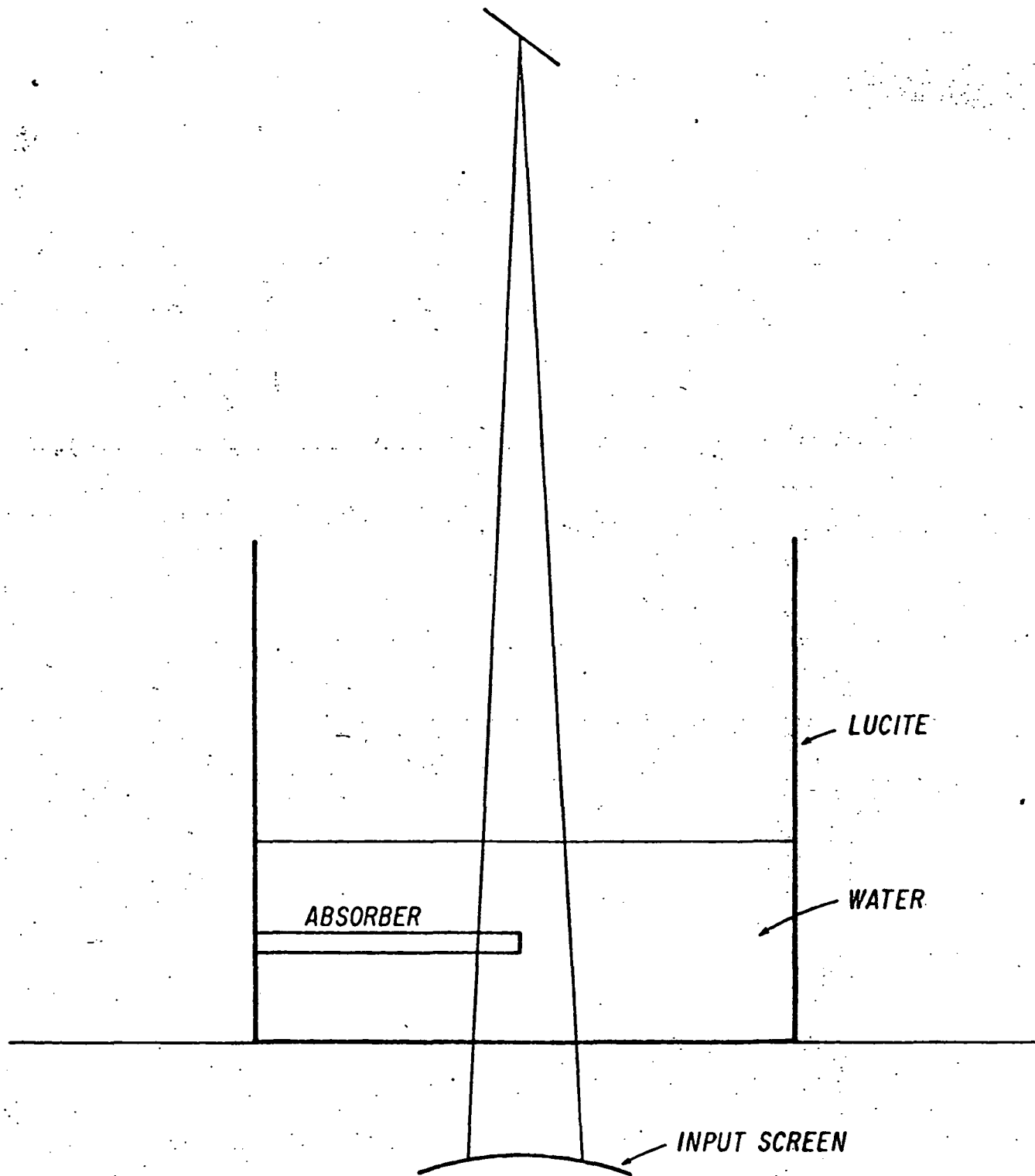


Figure 4-A

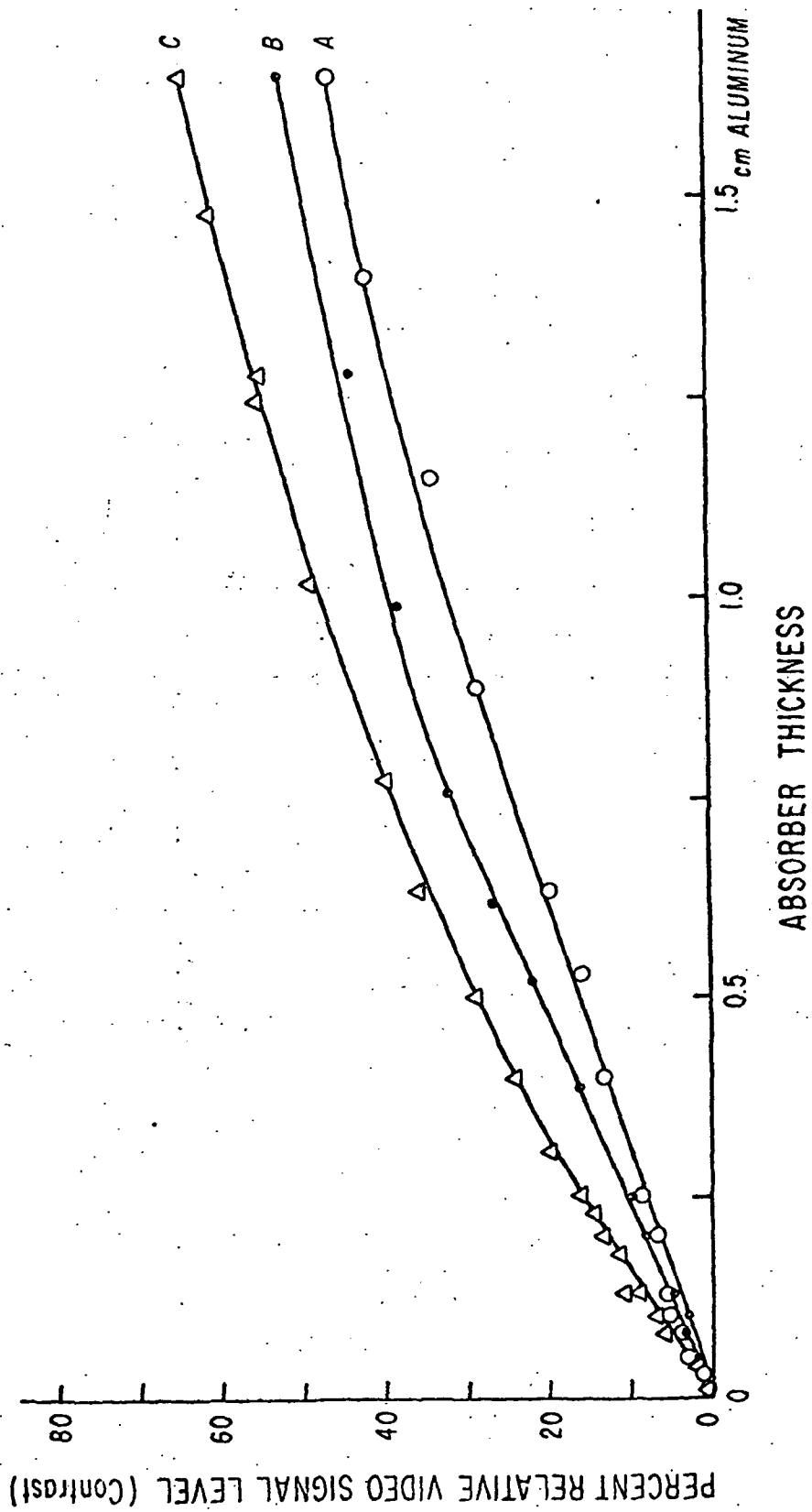


Figure 5A

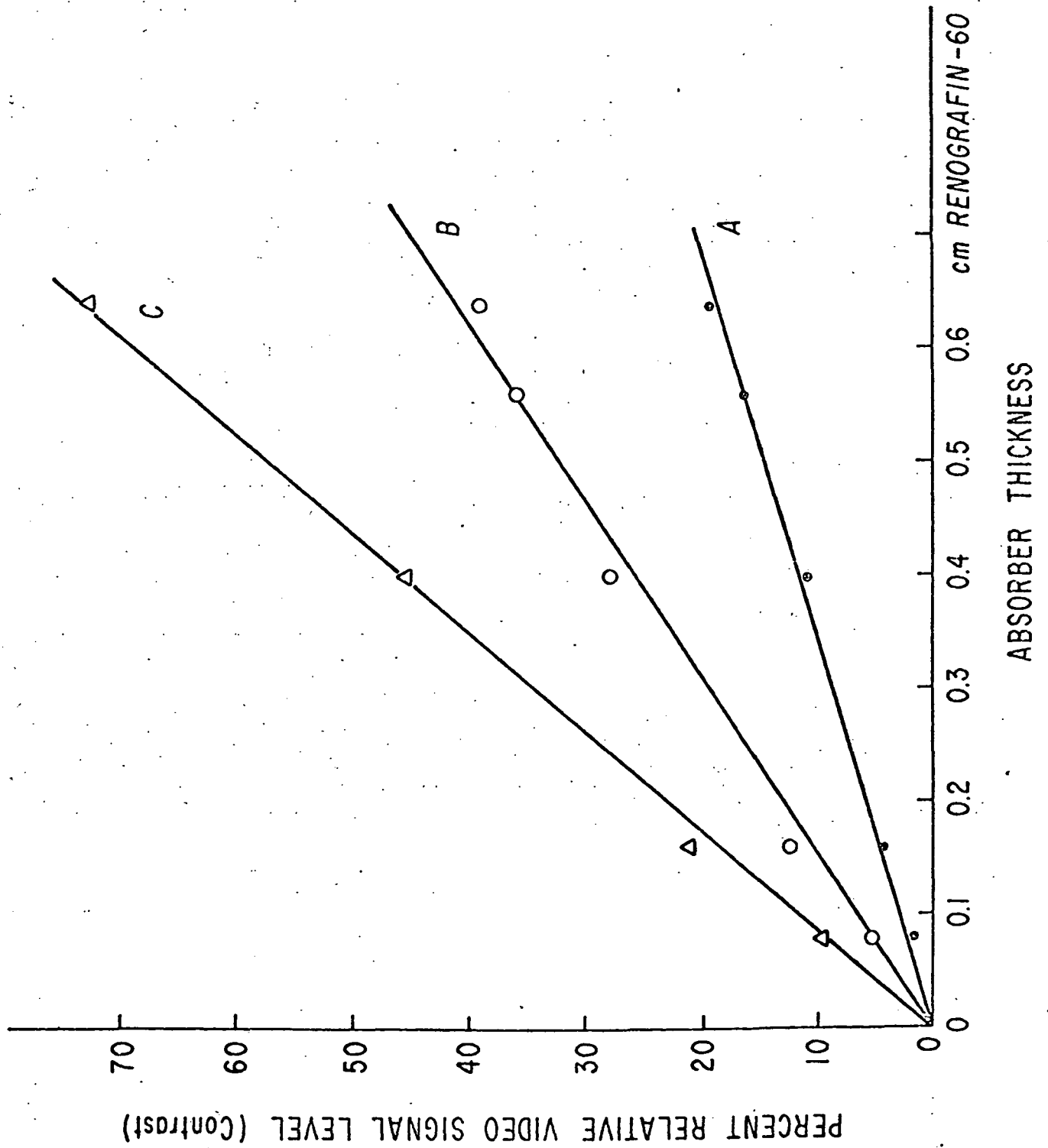


Figure 6-A

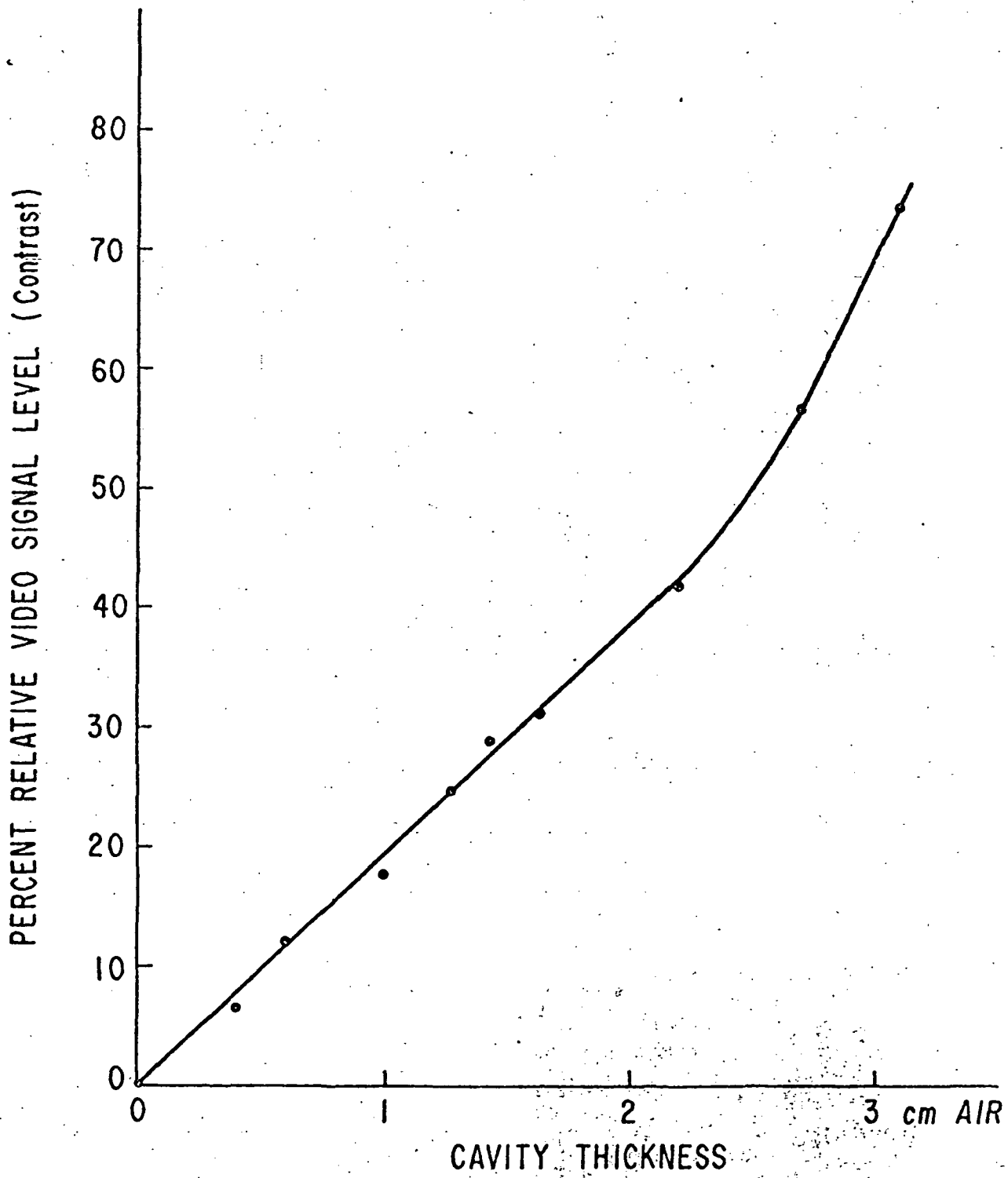


Figure 7-A

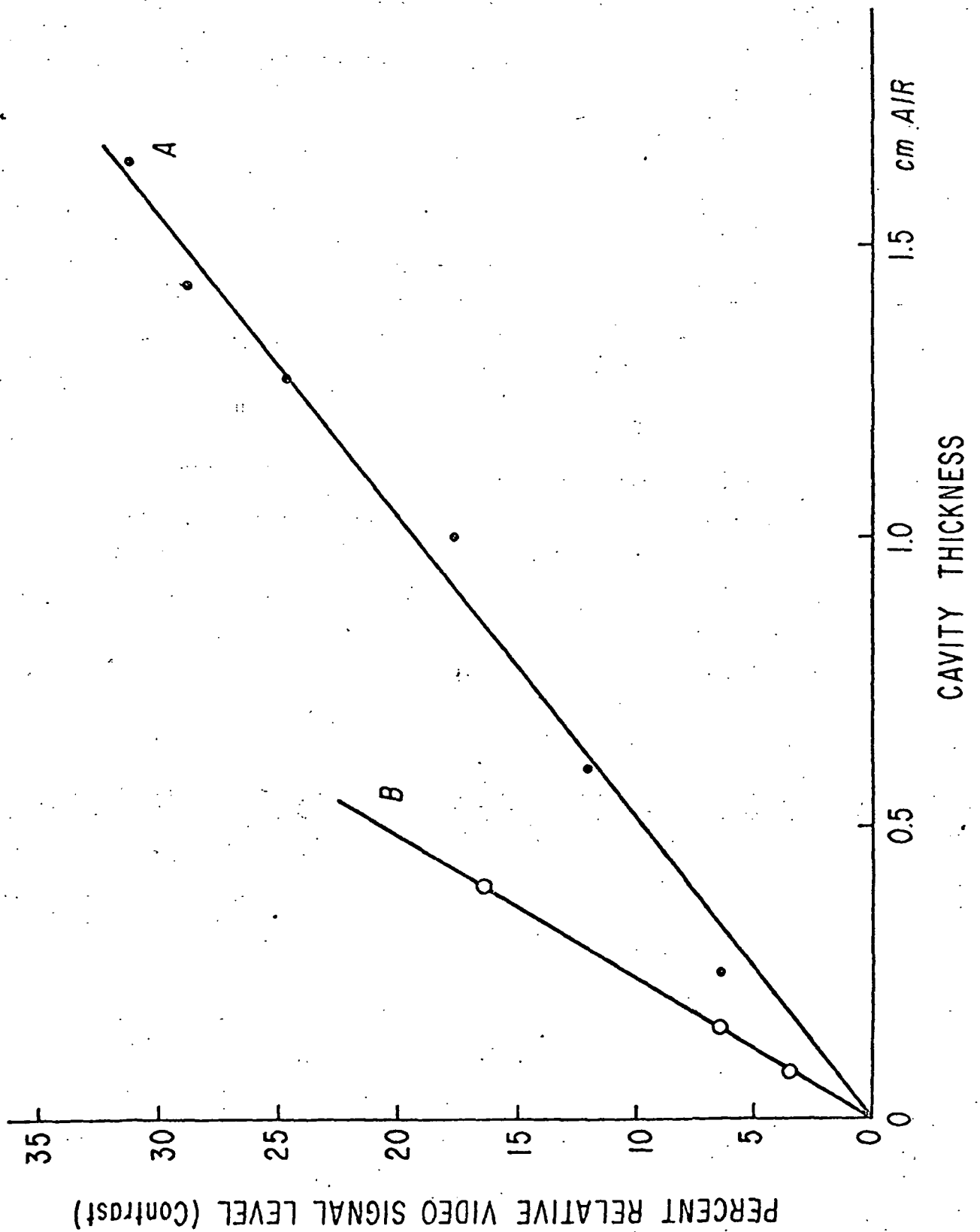


Figure 8-A

APPENDIX B

PERFORMANCE OF A LARGE SCREEN FLUOROSCOPIC IMAGING SYSTEM[†]

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Introduction

In recent years the need for large flat screen fluoroscopic imaging systems has been pointed up by the development and subsequent use of new technologies and new procedures in diagnostic radiology. Videodensitometry, if it is to be extended to clinical applications involving studies of the thorax, kidneys, etc., requires an input screen having an area greater than that of the usual x-ray image intensifier (Ref. 1). A similar requirement exists for optimal use to be made of fluoroscopic tomography (Refs. 2, 3, 4). Further, if these very promising developments are to be made truly quantitative rather than relative, the distortion of the resultant optical intensity due to screen curvature needs to be eliminated. The use of digitized systems with on-line processing and with their capabilities for making precision measurements at the levels of minimum detectability, makes this development even more desirable (Refs. 5, 6). Examples of this need can also be found in the qualitative or imaging aspects of diagnostic radiology.

The concept of using a large flat input screen viewed by a low light level TV camera was described by Baker, et al (Ref. 7). This type of system became feasible because of the development of high resolution light amplifiers. These amplifiers are available as single-, two-, or three-stage assemblies, and with diameters of 18 mm, 25mm, or 40 mm. These range in luminous gain from 100 to 30,000 and have center resolution capabilities of 28 to 64 line pairs/mm. Even greater capabilities may be expected when electromagnetically focused light amplifiers become more generally available.

System Configuration

A schematic of the basic large screen fluoroscopic system is shown in Figure 1. Both the light amplifier and the vidicon used have fiber optic faceplates. These are optically coupled with fiber optics and the interfaces optically matched with cedar oil. The system we have employed* used a stationary grid and a 12" x 12" type CB-2 fluorescent screen. The conventional lens used had a 67 mm focal length and an aperture of f 1.4. The light amplifier was a 3-stage, 18 mm diameter device and was coupled to a 1-inch type 8507 vidicon. The TV electronics incorporated shading circuits to compensate for the fall-off of light at the edges of and non-uniformity within the field.

Results

The examples, comparisons, and MTF measurements that are presented in this paper are preliminary and represent results obtained with a less than optimum large screen fluoroscopic system. System optimization was necessarily sacrificed due to limited resources for making system changes. In addition, the fact that the large screen fluoroscopic system was not the property of the authors precluded certain other changes. Considerable improvement in image quality is obtainable by optimization of this system. At this time, though, the object of the testing was to make a qualitative comparison of the images obtained with this system as compared to those obtained with a standard clinical x-ray image intensifier. The conventional unit used for this comparison was a G.E. 6"-9" fluoricon in its 9" format.

Figure 2 shows a comparison of images between the large screen and conventional units using a diagnostic type pelvic phantom. Photograph A is the image obtained using the conventional intensifier and B that obtained using the 12" x 12" flat screen device.

[†] Work partially supported by USPHS Grant No. 1 R17 HL 24160

* Manufactured by Sierra Scientific Corp.,
Menlo Park, California

both were obtained in the standard fluoroscopic geometry with equal magnification. A comparison of the upper portion of B with the image of A illustrates the increased coverage obtained. Both exposures were made at 110 kVp. Tube current was 3 mA for the conventional amplifier and 5.25 mA for the flat screen device. These images were recorded on a video disc using both TV fields and then photographed from the monitor using the disc playback.

In order to get a direct image comparison of the same phantom area, the focal spot-to-phantom distance for the flat screen amplifier was decreased to obtain a reduction in phantom area coverage. All other factors were kept constant. The image obtained is shown in Figure 3 and stands in comparison to Figure 2A.

In previous work we have found that operation of the TV system in the single scan mode results in both a higher resolution capability and decreased patient dose (Refs. 2, 4). Figure 4 shows recordings similar to those in Figure 2, but obtained with the TVs operated in the single scan mode, i.e., with only a single field recorded on the video disc. Using a 250 msec delay in the start of the vidicon sweep, the tube current used with the conventional amplifier was dropped to 0.5 mA. Due to a difference in dark current between the large screen and conventional system vidicons the longest delay usable with the large screen system was 150 msec and a reduced vidicon target voltage was required. This resulted in a requirement for an x-ray tube current of 5 mA. This is one point where the system could be optimized. This would be done through the use of a vidicon with a target material of higher resistivity.

Operation in the single scan mode also allows the use of very short x-ray exposures since there is no requirement for any type of timing. This allows recording of single images using multiple very short exposures with high x-ray tube currents, integrating the series of short exposures to obtain an increase in the image signal-to-noise ratio. This mode is also useful for stopping motion since high mA and short times are possible. An example of the integration technique is shown in Figure 5. The images shown here are the sum of seven exposures. Photograph A was obtained using a conventional intensifier and B, the resultant image recorded by the flat screen-low light level TV system.

Figure 6 shows an image recorded of a human chest during clinical examination. This patient has a unilateral tubercular condition of the left lung. Photograph A shows a fluoroscopic image recorded on the flat screen intensifier. In this instance the unit was substituted for a conventional amplifier and

the mechanical constraints required for mounting of the unit. Photograph B shows the same region recorded on film.

In addition to these comparisons we have made measurements of the MTF of the entire x-ray system used with this experimental large screen x-ray intensifier. The x-ray tube focal spot size was 1 mm. To obtain these transfer functions, a lead test pattern (Nuclear Associates #07-553) was placed on the surface of the input fluorescent screen. The field size of the x-ray beam was adjusted so as to just cover the test pattern (6.4 cm x 4.5 cm). Both beam and pattern were moved to different positions of the input screen. A single horizontal TV line through the pattern was analyzed to obtain the system MTF. Figure 7 shows the MTF curves obtained in four different areas. The circles depict the response measured at the center of the screen; the squares, that obtained midway between center and edge; the triangles, at the midpoint of the vertical edge of the field. The crosses represent the modulation transfer function obtained at the corners of the field.

Discussion

It is stressed that because of the difficulties associated with the photography of an image from a television monitor, the examples shown in this paper (Figures 2 through 6) are severely degraded. The images presented to the viewer on the TV monitor are very much superior to these photographs. As such, applications of electronic radiography calling for permanent and clinically acceptable radiographic image storage will have to await improvements in techniques of permanently recording video images.

Presently available video recording equipment (for electronic playback) has limited bandwidth. At present 4 MHz to 6 MHz is what usually can be obtained. Recent advances have made possible video disc recorders, with restricted storage capacity, having bandwidths up to 10 MHz. The availability of such equipment will substantially improve the modulation transfer function and the image quality of radiographic disc recording systems.

Increasing system resolution capability can also be obtained through the use of a 1.5" vidicon. This would increase the usable vidicon target area from 0.1875 in² to 0.48 in². Such an increase would then allow the use of a larger light amplifier. An increase in light amplifier size has a number of benefits. For example, increasing the light amplifier diameter from 16 mm to 40 mm automatically drops by a factor of two the resolution requirements of the light amplifier needed to give the same resultant

the target of a 1.5" vidicon only has a diameter of 25 mm, one can afford to lose, in the case of a 40 mm diameter light amplifier, the extreme outer portion of the useful amplifier screen area. This has three beneficial effects since in the typical light amplifier, as one moves from center to edge, there is; a decrease of about 10 percent in resolution, an increase in distortion approaching 25 percent, and image magnification of about the same 25% magnitude.

At present we are carrying out experiments in our laboratory using a 1.5" vidicon and a 40 mm three-stage light amplifier coupled with tapered fiber optics. This system has been designed to use a 14" x 17" input screen. We intend to study the effect of increasing the grid ratio on resolution, contrast, and patient dose. Preliminary experiments using a 12:1 moving grid indicate that improvements are possible without an excessive increase in patient dose. The experimental unit is now equipped with a 15" x 18", 10:1, 103 lines per inch grid. The system is also set up so that the input fluorescent screen is easily changed. In addition to the CB-2 screen we will study the response with both E-2 and DuPont Lightning Plus materials. If a higher efficiency screen can be used, then a reduction in light amplifier gain will be possible. Use of a two-stage amplifier (30,000 to 3,000 in gain) would result in an increase of light amplifier resolution of about 40 percent. Using the three-stage amplifier and an E-2 screen, images of test patterns require 0.1 mA at 65 kVp.

We are also intending to introduce electronic image processing into the system. We are examining the effects of using both linear and non-linear filtering for improvement in the image signal-to-noise level and the use of an all-directional aperture equalizer as described by Fuchs, et al (Ref. 8).

Conclusion

Preliminary investigations and limited trials leading to the results presented, indicate that both clinical and research applications of this type of x-ray image intensifier are entirely feasible. Further, it is felt that future developments and improvements will lead to applications which will make important contributions in both diagnostic radiology and radiation therapy. In addition, applications to physiological data acquisition using radiological methodology will be facilitated by this development.

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Figure Captions

Figure 1: Schematic diagram of large flat screen x-ray image intensifier.

Figure 2: A. Monitor photograph of recorded fluoroscopic image using a conventional x-ray image intensifier.
B. Monitor photograph of recorded fluoroscopic image using a 12" x 12" flat input screen-low light level TV intensifier.

Figure 3: Magnified fluoroscopic image of the same area in the phantom as shown in Figure 2A, but obtained using the experimental intensifier.

Figure 4: Single scan images recorded as single fields.
A. Conventional image intensifier.
B. Flat screen intensifier.

Figure 5: Summation of seven exposures, single-scan recorded after integration on the vidicon target.
A. Conventional x-ray image intensifier.
B. Flat screen-low light level TV intensifier.

Figure 6: A. Fluoroscopic recording of a tubercular chest made with the flat screen device.
B. The same region recorded on film.

Figure 7: Modulation transfer functions obtained for the 12" x 12" flat screen-low light level TV image intensifier at various areas of the input screen. The circles are for the center; squares, midway between center and edge; triangles, the response at an edge of the vertical midpoint; and the crosses at a corner.

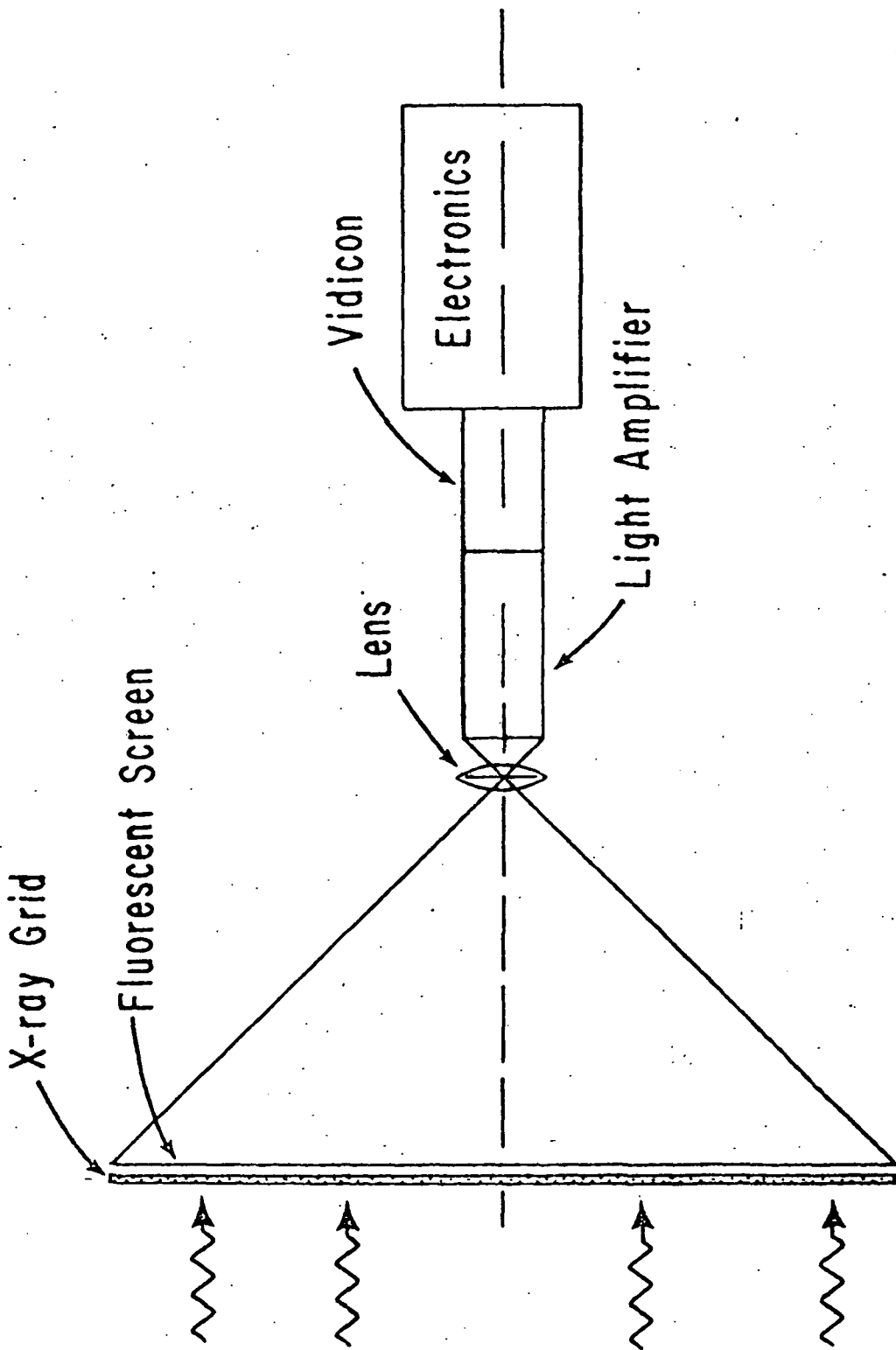
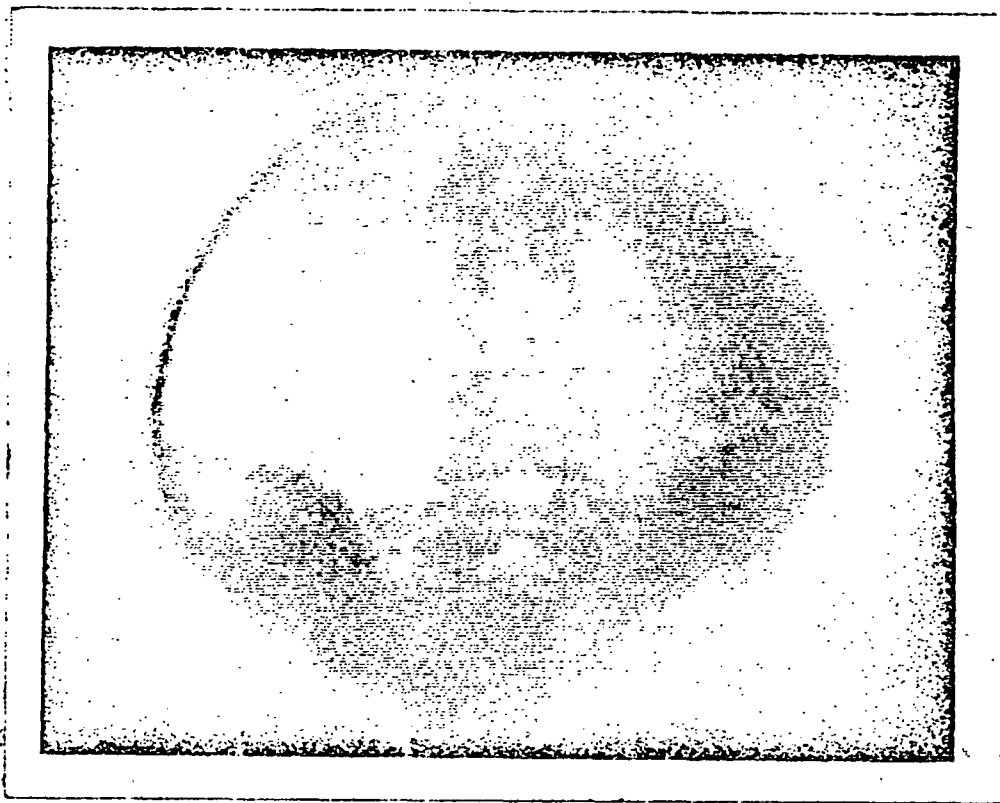


Figure 1-B



A



B

Fig. 2-B

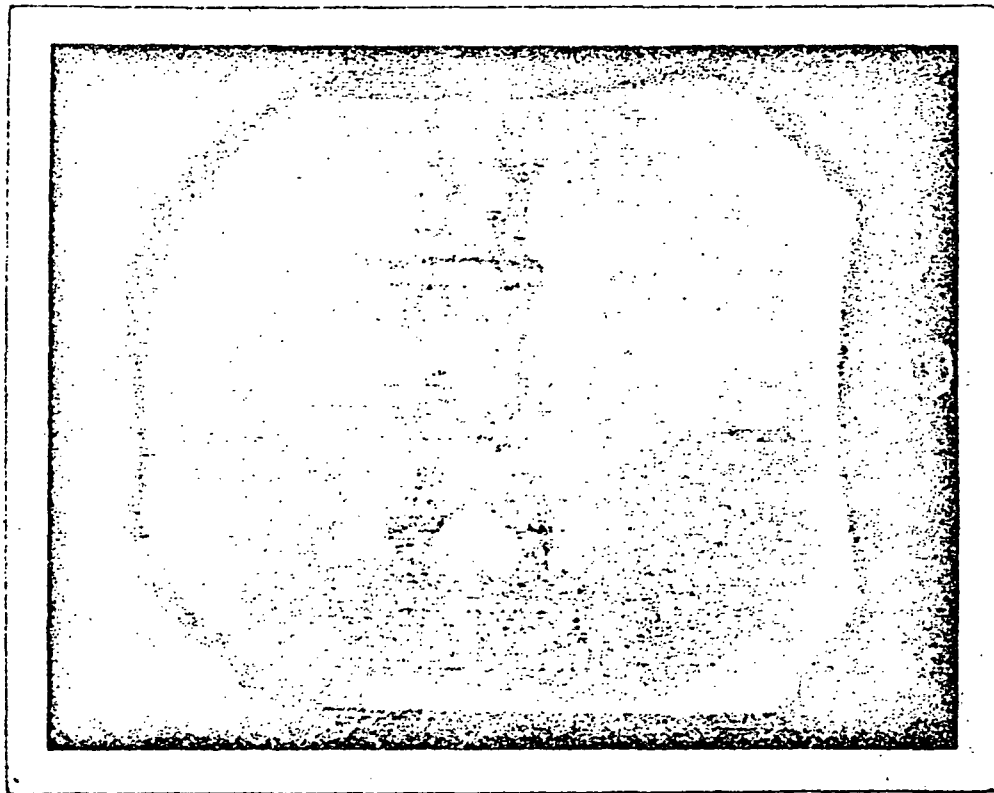
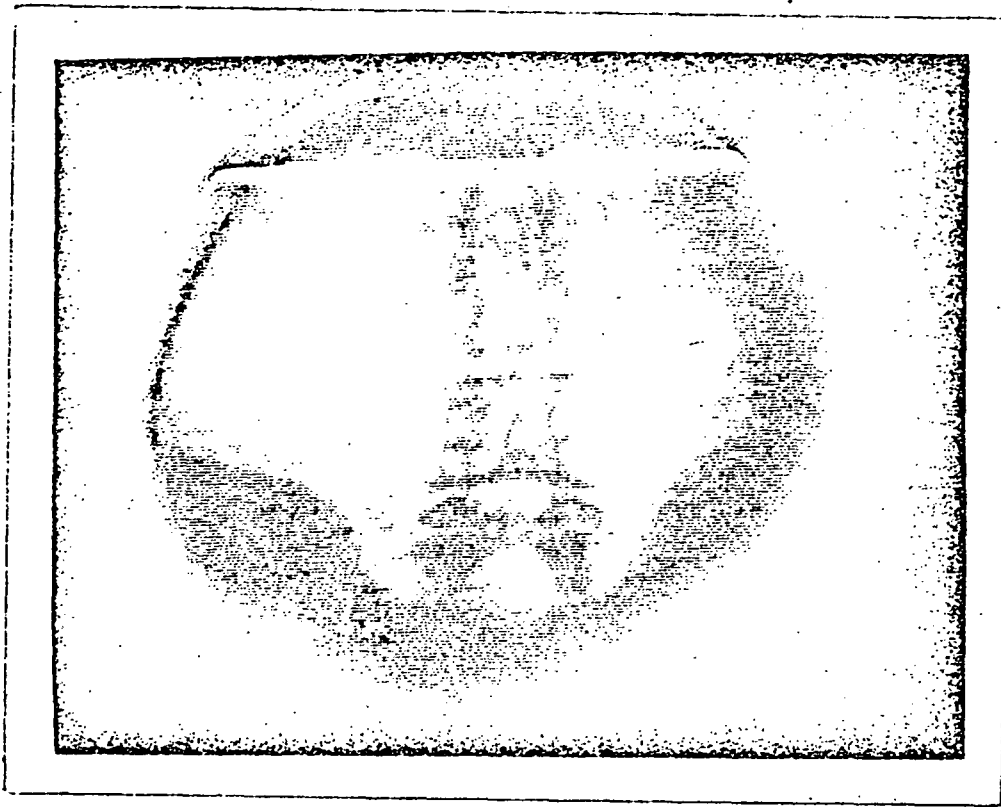
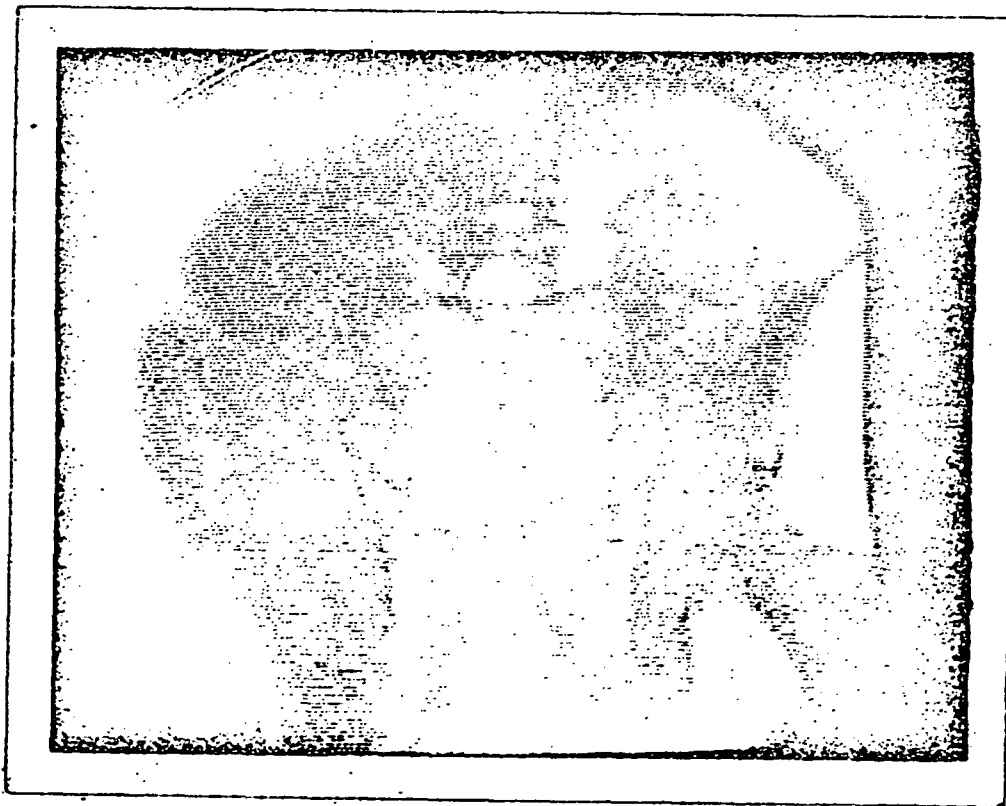


Fig. 3-B

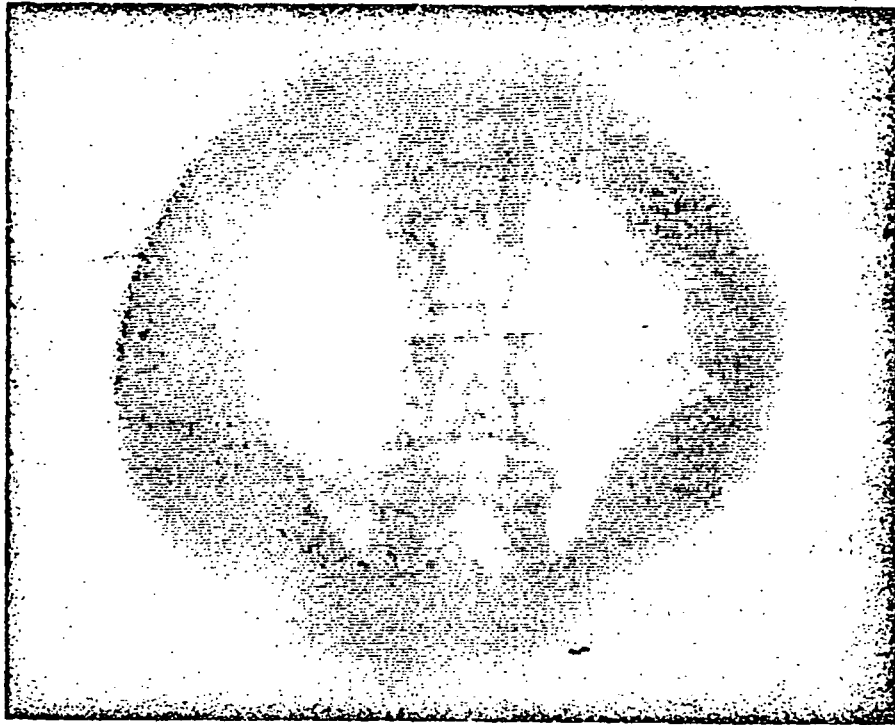


A

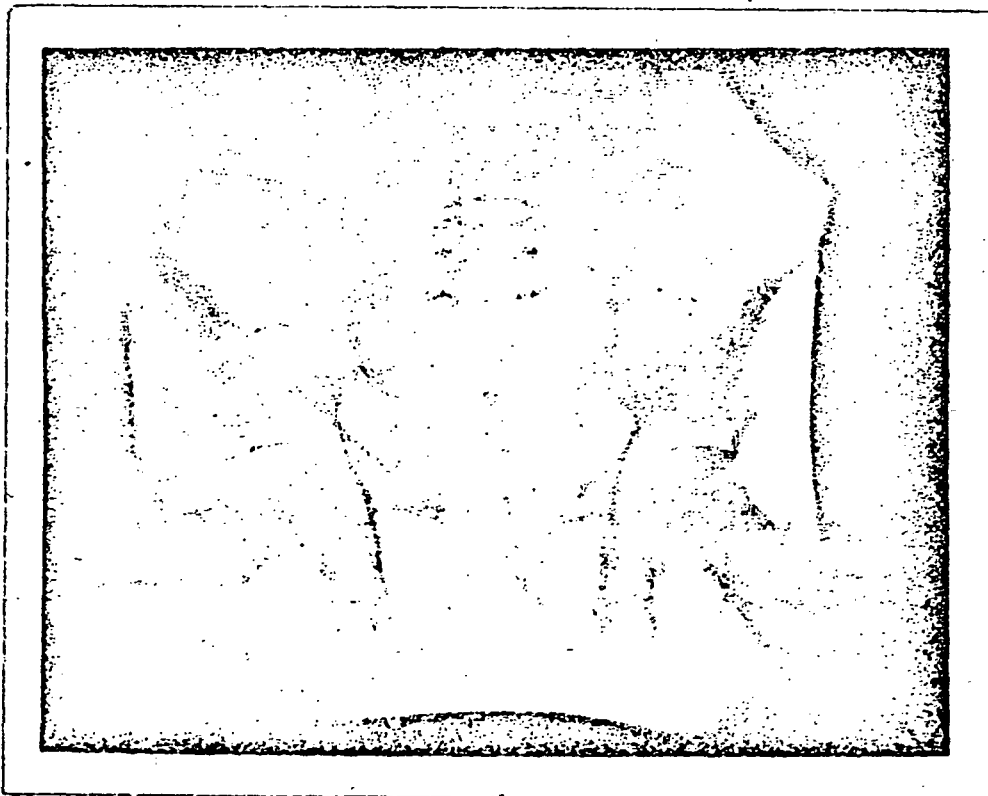


B

Fig. 4-B

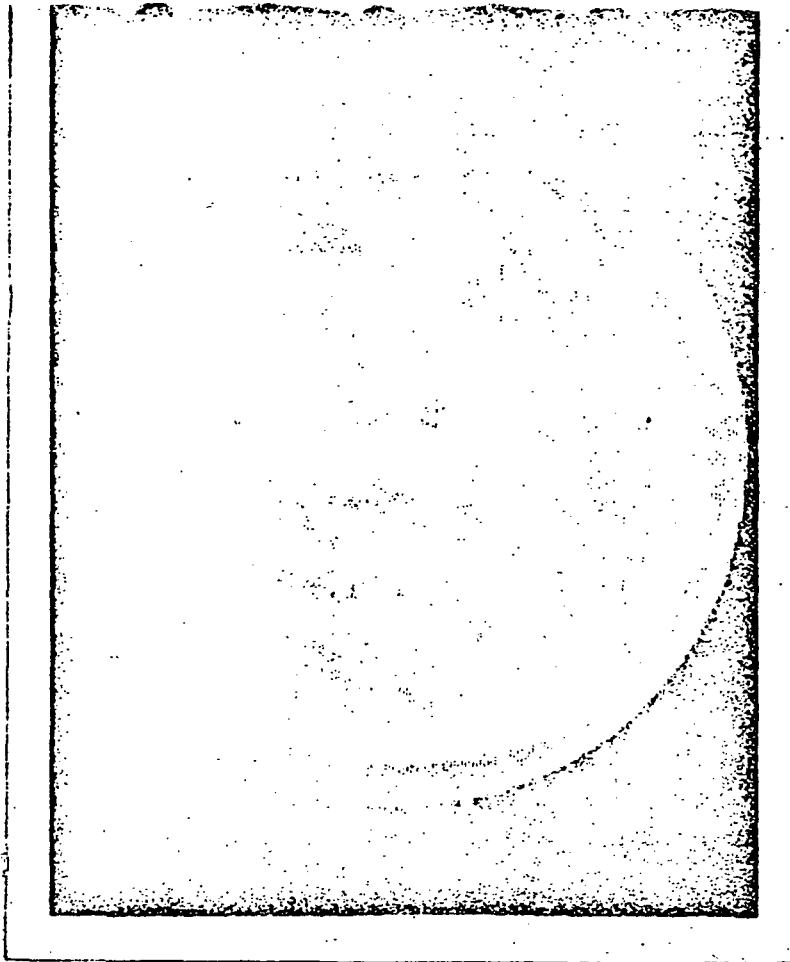


A



B

Fig. 5-B



A



B

Fig. 6-B

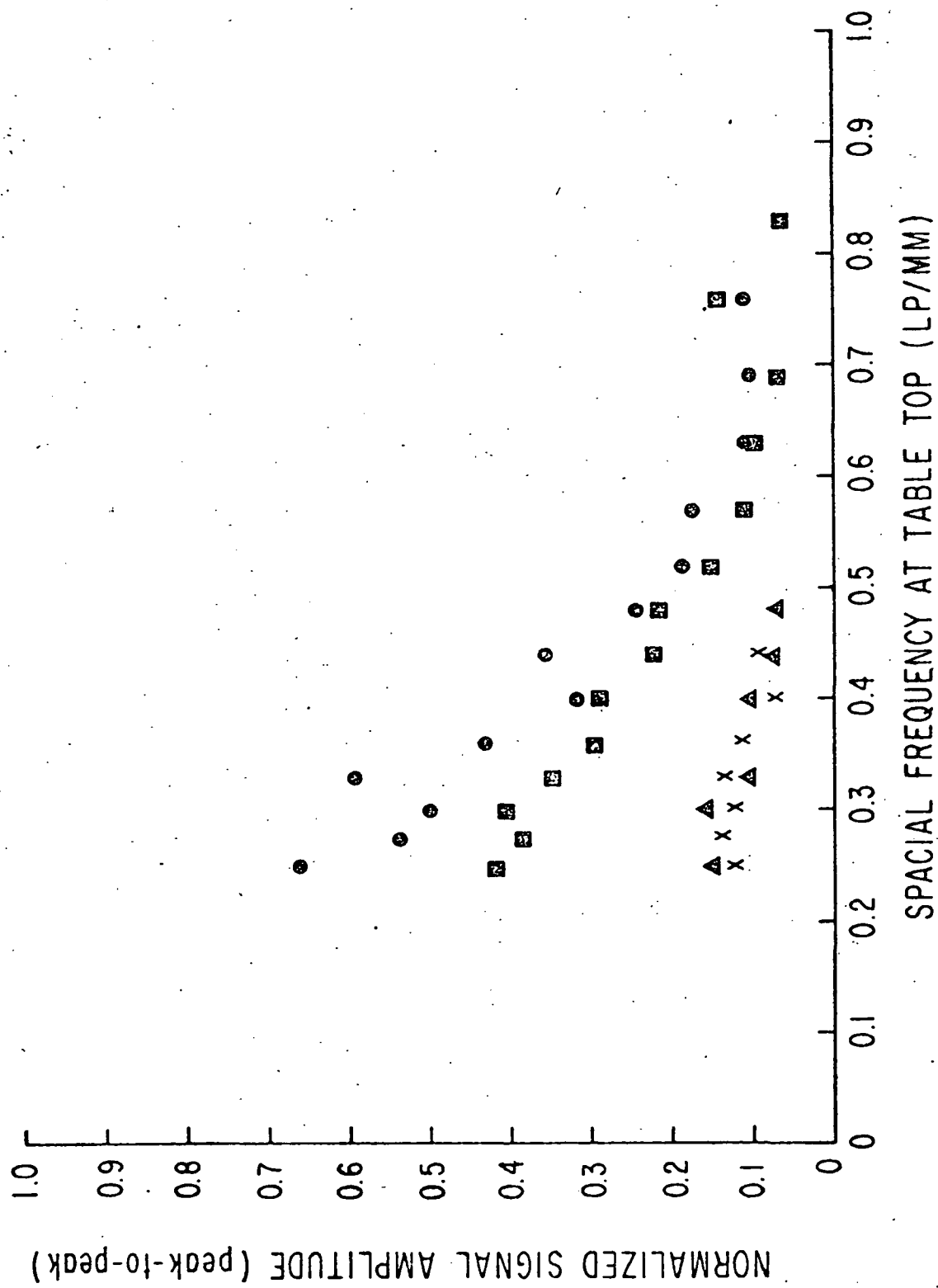


Figure 7-8