Oak Ridge

Presented at the "Small Vessel
Angiography" meeting New York,
April 1972. Will be published
in a book
6/12/72

COMPUTER SIMULATION OF SMALL BLOOD VESSEL IMAGING IN MAGNIFICATION RADIOGRAPHY*

ACRH-1000-356
CONF-720457-2

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*Supported in part by USPHS Grant GM 15806-01A1

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INTRODUCTION

Radiographic magnification technique has been applied to the visualization of small blood vessels in angiography, and the improvement of diagnostic accuracy resulting from better detail visibility of angiographic images has been reported by several clinical investigators.^{1,2,3,4} In addition, the modulation transfer function (MTF) and the resolution of the magnification system have been measured and applied to explain the technical advantage of magnification radiography in comparison with conventional radiography.^{5,6,7,8} Technical effects, which can be attributed to the magnification and also may contribute to the improvement of diagnostic accuracy, are: (1) The change of the image distribution which is related to the MTF, i.e., to resolution and sharpness; (2) the enlargement of the image which might help the visual recognition; and (3) the increase of the signal-to-noise ratio caused by an increase of the x-ray exposure per unit area of an object. However, it does not seem to be clear which technical effect has the most important impact on the clinical usefulness of the magnification technique. This may be due partly to the fact that the magnification process involves many technical factors, such as the magnification ratio, the focal spot size of the x-ray tube, the amount of scattered radiation, the screen-film system, the x-ray tube kilovoltage and the exposure time. It is, therefore, difficult to investigate the individual or combined effect of all these factors on angiographic images accurately by experiments alone: It is necessary to make an appropriate choice of these factors so that detail visibility in magnified images is better than in conventional images.

In this investigation the process of imaging small blood vessels in magnification angiography is simulated by a digital computer, and the effects of several technical factors on magnified images are studied. A merit of the computer simulation is that parameters can easily be changed over a wide range and the effect of the complicated combinations of technical factors can be

investigated within a small computation error.

COMPUTER SIMULATION

The process of imaging one-dimensional angiographic objects is simulated in this investigation. A cylinder, which corresponds to a blood vessel filled with a contrast medium, is placed in the space between the focal spot of an x-ray tube and a screen-film system. The position of the cylinder is restricted to the center of the radiation field and is specified by only the magnification, m . The entire imaging process, which includes the focal spot of x-ray tubes, the screen-film system, and a cylindrical radiological object, is assumed to be linear and isoplanatic. Some of these fundamental prerequisites have been verified indirectly by several investigations.^{9,10,11,12}

A diagram illustrating the computer simulation for magnification angiography is shown in Fig. 1. An object profile or x-ray pattern which is an x-ray intensity distribution transmitted through a blood vessel filled with a contrast medium, is calculated first by assuming monochromatic radiation. An object profile, which is experimentally obtained, can also be used for the computer simulation. This was not done in the present computation, because the normalized distributions of two object profiles, calculated and measured, were found to be almost the same and the object profile generated by the computer has more flexibility for studying magnification angiography than have the experimental data. An image profile, which is an illuminance distribution of the blurred image corresponding to the object profile, is calculated by two successive convolutions of the object profile with the line spread function (LSF) of the focal spot and with the LSF of the screen-film system. The convolutions are performed on the image plane, but image profiles are displayed on the image and object planes. The magnification is, therefore, used for computing the object

profile and the LSF of the focal spot in the image plane, and the image profile in the object plane. An image density distribution is calculated from the image profile by means of a characteristic curve.

In the spatial frequency domain an object Fourier spectrum, which is a Fourier transformation of the object profile, is calculated by a Bessel function series expansion of the object Fourier spectrum.¹³ An image Fourier spectrum is calculated by multiplying the object Fourier spectrum with the OTF's of the focal spot and the screen-film system.

A linear absorption coefficient of the blood vessel filled with contrast medium is estimated as 0.618 mm^{-1} corresponding to an iodine content of 282 mg/ml of Conray, Mallinckrodt Pharmaceuticals and a mass attenuation coefficient of iodine, $21.9 \text{ cm}^2/\text{g}$, at 40 keV, which is estimated from the measured x-ray emission spectrum of the diagnostic x-ray tube¹⁴ as a first approximation. A uniform distribution is assumed for the LSF of the focal spot, and an analytical form of the optical transfer function (OTF), the Fourier transformation of the uniform distribution,¹⁵ is employed. Experimental data are used for the LSF and the MTF of the screen-film system.¹⁶

RESULTS

The computer program can be applied to various problems in magnification radiography. In this investigation the following problem was studied.

When motion unsharpness is unavoidable unless a very short exposure time can be used, or when the x-ray exposure per frame is limited by the capacity of the x-ray equipment, the use of a fast screen-film system is imperative and results in poor detail visibility of angiographic images. The question then arises whether the magnification technique can be employed to improve detail visibility. What

should the focal spot size and the magnification be? What is the appearance of the magnified blood vessel image? In the following some of the answers obtained in this study are discussed:

Imaging performance of a radiographic system is frequently evaluated by the OTF or the MTF which describes the signal transmitting capability of the imaging system in the spatial frequency domain. The filtering effect is often utilized to explain the resolution and sharpness of images by using the OTF or the MTF.

The OTF's of the focal spot and the screen-film system on the object plane are shown in Fig. 2 when the magnification varies from unity to twenty. A focal spot size of 0.3 mm and Radelin TF-2 X-ray Intensifying Screens were selected as examples. At a magnification of unity, the MTF of the screen-film system is the only factor degrading the image because the OTF of the focal spot is unity at all spatial frequencies, which corresponds to an ideal system performance. As the magnification increases, the MTF of the screen-film system improves slowly toward an ideal MTF at a magnification of infinity, but the OTF of the focal spot changes in the opposite direction until, finally, it is the only factor responsible for the system performance. The total OTF of the magnification process is a product of these two OTF's. Optimum magnification is obtained when the total OTF reaches a maximum. This is the basic concept underlying the optimum magnification theory based on OTF's. The evaluation of the OTF itself, however, has little meaning unless the object Fourier spectrum is taken into account. This is because the OTF is a filter function describing transmission of the object Fourier spectrum through the system. The best OTF should, therefore, be the most effective filter function for the transmission of the object Fourier spectrum. The total OTF's, and the object and image Fourier spectra are shown in Fig. 3. The object is a blood vessel of 0.3 mm diameter.

The total OTF in the low spatial frequency region improves slowly with increasing magnification, but in the high spatial frequency region the negative and oscillating components appear at high magnification. The image Fourier spectra reflect these OTF variations. As the magnification increases, the image Fourier spectrum approaches the object Fourier spectrum at low spatial frequencies, but an oscillation is observed at high spatial frequencies. These changes of the OTF and the image Fourier spectrum appear to indicate that, with the TF-2 screen and an 0.3 mm focal spot, a blood vessel of 0.3 mm diameter is better imaged at high magnification than at low magnification. The object and image profiles of this blood vessel at various magnifications are shown in Fig. 4. The higher the magnification becomes, the more the image profile approaches the object profile. The contrast of the blood vessel image, which is the relative illuminance decrease at the center of the image profile, increases by 67% when the magnification increases from unity to twenty.

The effect of the magnification technique on image contrast for 0.1 mm and 0.3 mm diameter blood vessels is shown in Fig. 5 when various focal spot sizes are used. For an 0.3 mm blood vessel, the contrast obtained with a focal spot smaller than 0.3 mm increases gradually but considerably as the magnification increases. On the other hand, the contrast obtained with a focal spot larger than 0.6 mm increases slightly at first and then decreases as the magnification increases. These contrast changes due to magnification are dependent on the diameter of the blood vessel. For 0.1 mm vessels, magnification with an 0.1 mm focal spot greatly improves the contrast, but with a focal spot larger than 0.3 mm the contrast improvement is very small. From these results, it can be concluded that magnification with an 0.1 mm focal spot would result in better detail visibility of 0.1 mm and 0.3 mm blood vessels, and might improve the diagnostic accuracy because of the contrast increase and the increased diameter

of the blood vessel images. Magnification with an 0.3 mm focal spot, however, might also improve detail visibility of an 0.1 mm blood vessel, resulting in increased diagnostic accuracy only if the increased diameter of the blood vessel is the essential technical effect on the usefulness of the magnification.

The effect of increased x-ray absorption of the contrast medium on the image contrast of the magnified blood vessel is shown in Fig. 6. The dotted curves represent the image contrasts of an 0.1 mm blood vessel filled with a contrast medium, the x-ray absorption of which is twice that represented by the solid curves. The relative changes of image contrast by magnification on two blood vessels with different x-ray absorption are almost the same, but on an absolute scale the contrast increase of the blood vessel with the high x-ray absorption is greater than that with the low x-ray absorption. This might be a meaningful result in a situation in which the detection of a blood vessel image was critically affected by low image contrast, because it would indicate that the use of magnification with a highly absorbing contrast medium is an effective way to improve the image contrast.

The optimum magnification, which is here defined as the magnification yielding maximum contrast of the blood vessel image, is shown in Table 1. The optimum magnifications are amazingly high for the small, uniform focal spot, and depend on the diameter of the blood vessel. The larger the diameter of the blood vessel, the higher the optimum magnification becomes. It is realized that the optimum magnification depends on the intensity distribution of the focal spot of the x-ray tube. The magnitude of this dependence is not yet known, and will be determined in the future.

An increase in image contrast similar to the increase due to magnification can also be obtained by increasing only the concentration of the contrast medium

without using magnification. The contrast change due to magnification can, therefore, be evaluated in terms of the corresponding relative increase of concentration, which will be called a "relative equivalent concentration of the contrast medium." Fig. 7 shows the effect of magnification on the diameter of the magnified blood vessel and the relative equivalent concentration for blood vessels of 0.1 mm and 1.0 mm diameter. For the larger vessel, the increase of the relative equivalent concentration obtained by the magnification is very small. Even for the maximum contrast of a 1.0 mm vessel obtained with an 0.1 mm focal spot and at twenty times magnification, an increase of only 30% in concentration of the contrast medium results in better image contrast than that obtained with the magnification technique. For a small blood vessel magnified with an 0.1 mm focal spot, however, the relative equivalent concentration becomes extremely high. At a magnification of twenty, for example, the relative equivalent concentration exceeds four. This means that to obtain the same contrast in the non-magnified image as that in the magnified image the concentration of the contrast medium must be increased more than four times. This is one way of indicating that the magnification technique is especially useful for small blood vessels.

For an 0.1 mm diameter vessel, the choice of focal spot size and magnification is very important. High magnification with an 0.1 mm focal spot improves the blood vessel image considerably, but the improvement with the 0.3 mm focal spot is small. A comparison of image profiles resulting from magnification with 0.1 mm and 0.3 mm focal spots is shown in Fig. 8. At low magnification, the image profiles obtained with the two different focal spot sizes are almost the same. The higher the magnification, the more noticeable the difference in the image profiles becomes. The image profile with the 0.1 mm focal spot is very close to the object profile at a magnification of twenty, although it might not be possible to use this high magnification in practice

because of the resulting increase in radiation to the patient.

DISCUSSION

The optimum condition of technical factors in magnification angiography depends on the diameter of the blood vessel as shown in this investigation. It is, therefore, rather difficult to specify the focal spot size and other technical factors that will result in optimum magnification, unless the diameter of the diagnostically significant blood vessel is known. This indicates that the study of not only the imaging performance but also of the object is necessary for designing magnification angiography effectively.

This computer simulation of the radiographic system has just begun, and has at present several limitations. Some of these limitations, however, can be removed in the near future. For example, experimentally obtained LSF's and MTF's of non-uniform focal spots could be used in the computation, and the effect of scattered radiation on images could be taken into account if data for the various angiographic procedures were available. Motion unsharpness could also be included in the computer simulation if object motion was described in terms of time and position axes. Moreover, it might be possible to include the x-ray emission spectrum of the x-ray tube precisely when exact data for the energy response of the screen-film system become available.

The present computer program cannot answer directly which of the three technical effects of magnification radiography described in the introduction has the greatest effect on the clinical result. From the computer simulation, however, it is possible to know accurately how the blood vessel is imaged physically in magnification angiography, which means that one of three technical effects can be investigated in detail.

SUMMARY

A computer program to simulate small blood vessel imaging in magnification angiography has been written, and magnification in conjunction with a uniform focal spot and a fast screen-film system has been simulated. The optimum magnification, determined from the contrast of blood vessel images, depends on the diameter of the blood vessel. For a blood vessel of 0.1 mm diameter, the image contrast increases almost four times when a magnification of twenty is used with an 0.1 mm focal spot.

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FIGURE CAPTIONS

- Fig. 1 - Block diagram of computer simulation for blood vessel imaging in magnification angiography.
- Fig. 2 - OTF's of focal spot, $G(u)$, and screen-film system, $S(u)$, on object plane at magnification of 1, 1.6, 3 and 20. Focal spot size: 0.3 mm. Screen: TF-2
- Fig. 3 - Total OTF, $T(u)$, object Fourier spectrum, $O(u)$, and image Fourier spectrum, $I(u)$, on object plane at various magnifications. Object diameter: 0.3 mm
- Fig. 4 - Object, $O(x)$, and image, $i(x)$, profiles of 0.3 mm blood vessel with 0.3 mm focal spot at various magnifications. Screen: TF-2
- Fig. 5 - Contrasts of 0.1 mm and 0.3 mm blood vessel images magnified with 0.1 mm, 0.3 mm, 0.6 mm and 1.2 mm focal spot. Screen: TF-2
- Fig. 6 - Contrasts of 0.1 mm blood vessel images with two different x-ray absorptions magnified with 0.1 mm, 0.3 mm and 0.6 mm focal spot
- Fig. 7 - Relative equivalent concentrations of contrast medium of 0.1 mm and 1.0 mm blood vessels magnified with 0.1 mm, 0.3 mm, 0.6 mm and 1.2 mm focal spot
- Fig. 8 - Object, $O(x)$, and image profiles of 0.1 mm blood vessel magnified with 0.1 mm (----) and 0.3 mm (—) focal spot
- Table 1 - Optimum magnification for maximum contrast of blood vessel image by TF-2 screen

Fig. 1

Fig. 2

Fig. 3

Fig. 4

Fig. 5

Fig. 6

Fig. 7

Fig. 8

BLOOD VESSEL DIAMETER FOCAL SPOT SIZE	BLOOD VESSEL DIAMETER			
	0.1 mm	0.3 mm	0.5 mm	1.0 mm
0.1 mm	> 20	> 20	> 20	> 20
0.3 mm	4	> 20	> 20	> 20
0.6 mm	1.2	1.5	3	4
1.2 mm	1.2	1.2	1.2	1.7

Table 1

CONTRAST OF BLOOD VESSEL IMAGE

RELATIVE EQUIVALENT CONCENTRATION
OF CONTRAST MEDIUM

CONTRAST OF BLOOD VESSEL IMAGE

BLOOD VESSEL DIAMETER FOCAL SPOT SIZE	0.1 mm	0.3 mm	0.5 mm	1.0 mm
0.1 mm	> 20	> 20	> 20	> 20
0.3 mm	4	> 20	> 20	> 20
0.6 mm	1.2	1.5	3	4
1.2 mm	1.2	1.2	1.2	1.7