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Generalized Performance Evaluation of X-ray Image Intensifier compared with a Microangiographic System

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Abstract

Standard objective parameters such as MTF, NPS, NEQ and DQE do not reflect complete system performance, because they do not account for geometric unsharpness due to finite focal spot size and scatter due to the patient. The inclusion of these factors led to the generalization of the objective quantities, termed GMTF, GNNPS, GNEQ and GDQE defined at the object plane. In this study, a commercial x-ray image intensifier (II) is evaluated under this generalized approach and compared with a high-resolution, ROI microangiographic system previously developed and evaluated by our group. The study was performed using clinically relevant spectra and simulated conditions for neurovascular angiography specific for each system. A head-equivalent phantom was used, and images were acquired from 60 to 100 kVp. A source to image distance of 100 cm (75 cm for the microangiographic system) and a focal spot of 0.6 mm were used. Effects of varying the irradiation field-size, the air-gaps, and the magnifications (1.1 to 1.3) were compared. A detailed comparison of all of the generalized parameters is presented for the two systems. The detector MTF for the microangiographic system is in general better than that for the II system. For the total x-ray imaging system, the GMTF and GDQE for the II are better at low spatial frequencies, whereas the microangiographic system performs substantially better at higher spatial frequencies. This generalized approach can be used to more realistically evaluate and compare total system performance leading to improved system designs tailored to the imaging task.

Keywords

MTF; NPS; GMTF; GNNPS; GDQE; Image Intensifier; generalized; microangiography; performance

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1. INTRODUCTION

X-ray image intensifiers are the most commonly used detector in clinical studies. Their detector performance has been evaluated in terms of the standard parameters, Modulation Transfer Function (MTF), Noise Power Spectrum (NPS), Noise Equivalent Quanta (NEQ), and Detective Quantum Efficiency (DQE). These standard parameters are defined for the bare detector performance irrespective of the effect of clinical environment during patient studies where scatter due to the patient and geometric unsharpness due to the finite size of the focal spot play a significant role in determining the image quality. The generalized methodology of detector evaluation was developed by including the clinically relevant factors affecting the image quality during patient studies. The properties of the detector, x-ray tube, and the patient were combined together to define a total x-ray imaging system for the generalization of detector performance evaluation parameters. The generalized imaging system evaluation parameters, termed Generalized Modulation Transfer Function (GMTF), Generalized Normalized Noise Power Spectrum (GNNPS), Generalized Noise Equivalent Quanta (GNEQ) and Generalized Detective Quantum Efficiency (GDQE) include the effect of scatter due to the patient as well as the spatial distribution of scatter and the focal spot blurring referenced to the object plane. The effect of focal spot blurring, magnification and scatter on the MTF has already been investigated by many authors.^{1,2,3,4} Kyprianou et. al.^{5,6} developed a theoretical formulation of the generalized performance evaluation parameters, and experimentally verified this for a specific high-resolution, region-of-interest, microangiographic prototype detector developed by our group.^{7,8,9}

Considering the clinical importance of the standard x-ray image intensifiers in image guided procedures, it is important to evaluate their performance under realistic clinical conditions for such procedures. The generalized system evaluation approach incorporates the properties of the patient as well as the geometry of the system affecting the final imaging quality. In this study, we present a detailed generalized performance evaluation of a commercial x-ray image intensifier system and compare it with the high-resolution, region-of-interest, microangiographic detector under simulated conditions for neurovascular angiography specific to each detector. A comprehensive comparison of the total system GMTF, GNNPS, GNEQ, and GDQE as a function of varying air-gap, radiation field size, magnification, and per-frame-exposure to the detector is presented for simulated neurovascular studies. The effect of anti-scatter grid with the II system is also evaluated in terms of GDQE.

2. MATERIALS AND METHODS

2.1. X-ray system and experimental set up

A total x-ray imaging system consists of an x-ray tube with a given focal spot size, the patient (phantom), a patient table, and the detector. In this study, two different detectors were employed: a standard commercial x-ray image intensifier unit mounted on a C-arm (Infinix, Model RTP12303J-G9E, Toshiba Medical Systems Corporation), and a high-resolution, region-of-interest microangiographic detector.^{7,8,9} The microangiographic detector is a full frame 12 bit, $1k \times 1k$ CCD camera optically coupled to a CsI (TI) scintillator via a minifying fiber optic taper, with a $4.5 \text{ cm} \times 4.5 \text{ cm}$ field of view having an effective pixel size of 0.043 mm, and capable of acquiring the images at 5 fps. The x-ray

were averaged to calculate the edge-spread function (ESF) at 60, 70, 80, 90 and 100 kVp in the scatter-free geometry. The MTF was calculated by the Fourier Transformation of the line-spread function (LSF) obtained by taking the first derivative of the ESF. An exponential tail fit of the LSF below the 2 percent of maximum level was used.¹³ Figure 2 shows the typical detector MTF of the image intensifier and the microangiographic system measured at 80 kVp.

2.2.2 Focal Spot MTF

The spatial distribution of the focal spot blurring was obtained by imaging the same edge test device when kept exactly in between the source and image plane.^{14,15} A 20 mm thickness of Al was used as a beam hardener instead of the phantom in this case. An average of 60 images was used to calculate the focal spot MTF. The Focal Spot Optical Transfer Function (OTF) goes to zero at around 1.65cycles/mm in the plane of the focal spot and then oscillates near zero reflecting a spatial phase shift. Assuming that the negative part of the OTF is anomalous resolution and unusable, a Gaussian tail extrapolation was used from the 1.6 cycles/mm to the Nyquist frequency of the detector in calculating the Focal Spot MTF. This anomalous behavior of the MTF is only a factor at high object magnification. Figure 3 shows the typical low frequency behavior of the Focal Spot MTF (MTF_F) for the 0.6 mm focal spot measured at 80 kVp. Focal Spot MTF is a property of the x-ray tube and is independent of detector properties.

2.2.3 Scatter Fraction and Veiling Glare

For the image intensifier system we estimated the veiling glare as well as the scatter, and then the veiling glare fraction was separated from the scatter. The standard lead beam-stop method¹⁶ was used to calculate the scatter fractions in the with-scatter geometry where the phantom was close to the detector. These scatter-to-primary ratio values also include the effect of veiling glare. In order to separate the veiling glare, the same lead beam-stop technique was used to estimate the veiling glare fraction without any scattering materials in the beam. The scatter fraction was calculated by subtracting the veiling glare fraction from the scatter-to-primary ratio measured previously with the scattering material. In order to associate the magnification studies and effect of radiation field sizes, sets of images were acquired as a function of air-gaps: 2.5 cm, 5.0 cm, 7.5 cm, 10 cm; and radiation field sizes: 6cm × 6cm, 8 cm × 8 cm, 9 cm × 9 cm and the full field of view of the image intensifier in the 5 inch mode. The resulting scatter fractions were fit to a surface following the empirical relation⁶ between the scatter fraction (ρ), air-gap (AG), and the radiation field area (α)

$$\rho(\alpha, AG) = \sum_{i=1}^4 \sum_{j=0}^2 a_{i,j} \alpha^{i/2} (AG)^j \quad (1)$$

where $a_{i,j}$ are constants to be determined from the fit. This surface fit allows us to evaluate the scatter fractions at any intermediate air-gaps and radiation field sizes. Figure 4 shows the surface plot of scatter fraction with respect to the air-gap and radiation field size. There is no veiling glare in the microangiographic system because of the absence of light intensification and a lens coupling stage in the hardware of the detector as compared to the veiling glare for the II, which is apparent in the low frequency drop of the Image intensifier MTF in Figure 2.

2.2.4 Scatter MTF

A difference edge-spread function technique was employed to calculate the spatial distribution of scatter for image intensifier and microangiographic system.¹⁷ The spatial distribution of scatter is appreciable only at very low spatial frequencies. Because we need the scatter MTF all the way up to the Nyquist frequency of each detector for further calculations in generalized system parameters, we used an exponential tail extrapolation up to the Nyquist frequency of each detector. Figure 3 shows the low spatial frequency behavior of the scatter MTF (MTF_S). Because the spatial distribution of scatter is detector independent for a given patient or phantom, we used the same scatter MTF for both the detectors. This scatter spatial distribution was measured with the microangiographic detector. Any potential grid induced changes in the scatter distribution were not considered in this work.

2.2.5 Noise Power Spectrum

The detector noise power spectrum is defined as the squared modulus of the Fourier Transformation of the fluctuations about the mean of a flat-field image, multiplied by the pixel area with units of mm^2 . For this purpose the uniformly illuminated full field of view of the image intensifier in the 5 inch mode was used. Sixty images at 3 fps were acquired for the fixed tube voltage of 80 kVp and for 6 different entrance exposure values before the grid ranging between 0.1 to 0.4 mR per frame. The phantom was placed at a distance of 2.5 cm from the detector allowing the scatter to enter. The two dimensional normalized noise power spectra were obtained by the standard Fourier transform method.¹⁸ To be consistent with the calculation of NNPS of the microangiographic system, a radial averaging was performed to obtain the one-dimensional noise power spectrum from the corresponding two-dimensional NNPS. To obtain the zero frequency NNPS, we used a fourth-order-polynomial fit over the NNPS data excluding the first four data points because of their unusual behavior due to some low frequency artifacts. The 1-D NNPS was saved into 128 bins up to the Nyquist frequency 4.3cycles/mm. Figure 5(a) shows a 2-D NNPS at entrance exposure 0.4 mR before the grid. Figure 5(b) shows a comparison of the smoothed polynomial and the measured NNPS data calculated from the radial averaging of the 2-D NNPS shown in Figure 5(a). The same procedure was applied to fit the microangiographic NNPS for further calculation of generalized parameters.

Assuming the linearity of the detector along with flat-field and offset corrected images, the $NNPS_D$ can be expressed as^{5,19,20}

$$NNPS_D(f, X) = \frac{a(f)}{X} + \frac{b(f)}{X^2} \quad (2)$$

where $a(f)$ and $b(f)$ are constants dependent on frequency, and are obtained by fitting the measured $NNPS_D$ at different detector entrance exposures X . Figure 6 shows a comparison of fit to equation (2) to the measured data points for spatial frequencies of 1, 2, 3 and 4cycles/mm. This interpolated function allows us to evaluate the NNPS at any intermediate detector entrance exposures.

2.3 Generalized System Parameters

The generalized system parameters are defined by considering the focal spot blurring and the image degradation due to scatter as parallel processes¹⁹ at the detector entrance followed by the detector degradation as the cascade in the imaging chain. The detector MTF (MTF_D), the Focal Spot MTF (MTF_F) and the Scatter MTF (MTF_S) along with the scatter fraction (ρ) are combined together to define the Generalized Modulation Transfer Function^{5,6,11}:

$$GMTF(f, \rho, m) = \left[(1 - \rho) MTF_F \left(\frac{m-1}{m} f \right) + \rho MTF_S \left(\frac{f}{m} \right) \right] MTF_D \left(\frac{f}{m} \right) \quad (3)$$

where m is magnification factor and f the spatial frequency in mm^{-1} .

The scatter fraction is defined as $\rho = S/(S+P)$, where S and P being the scatter and primary components detected by the image receptor, respectively. It is to be noted that in this equation the scatter fraction is independent of magnification m and the object is assumed to remain fixed with respect to the detector. The term magnification as used here is defined for the objects within the patient or phantom, e.g., blood vessels or stents within the skull, keeping the patient or phantom at fixed distance from the detector. When using this approach to magnification radiography, the scatter fraction as a function of air-gap is required. The factor $(1-\rho)$ in the first term ensures that only the primary component contributed to focal spot blurring. The factor $(m-1)/m$ in the focal spot term rescales the spatial frequency into the object plane from the physical focal spot in the x-ray tube. The second term has only the scatter contribution and spatial frequency is rescaled to the object plane from the image plane using a factor $1/m$ with the spatial frequency.⁴ The detector MTF is used as a multiplicative term assuring the cascade process, and again the spatial frequency is rescaled to the object plane from the detector plane. Thus, the total system Generalized Modulation Transfer Function (GMTF) is defined at the object plane in contrast to the bare detector MTF which is defined at the detector plane.

The Generalized Normalized Noise Power Spectrum (GNNPS) in the object plane is defined as

$$GNNPS(f, X, m) = \frac{NPS_D \left(\frac{f}{m}, X \right)}{m^2 d^2(X)} = \frac{NNPS_D \left(\frac{f}{m}, X \right)}{m^2} \quad (4)$$

where $NPS_D(f, X)$ is the detector Noise Power Spectrum and $d(X)$ is the detector output signal measured as the average pixel value of the flat-field image, provided that the detector is linear or linearized. There is no distinction between primary and scattered photons in terms of the normalized-noise contribution; therefore there is no scatter term in this equation. A division by a factor of m^2 is used to project the effective pixel area from the image plane to the object plane.

The Generalized Noise Equivalent Quanta (GNEQ) from the total imaging system point of view is defined by using the GMTF and GNNPS as follows

$$GNEQ(f, \rho, X, m) = \frac{GMTF^2(f, \rho, m)}{GNNPS(f, \rho, X, m)} \quad (5)$$

The Generalized Detective Quantum Efficiency for the total system is defined by normalizing the absolute image quality with the effective photon fluence at the object plane

$$GDQE(f, \rho, X, m) = \frac{GNEQ(f, \rho, X, m)}{m^2 \Phi_{in}(X, m)} \quad (6)$$

where $\Phi_{in}(X, m)$ represents the number of quanta per unit area at the detector entrance. The denominator $m^2 \Phi_{in}(X, m)$ is the number of incident quanta per unit area referenced to the object plane using the inverse square increase of the number of quanta measured at the detector entrance. The number of quanta per unit area at the detector entrance is determined by using the normalized x-ray spectrum as a function of energy^{21,22,23} and the measured detector entrance exposure.

The scatter fraction and magnification can be expressed in terms of geometry dependent explicit parameters like air-gap and radiation field size for a given SID and depth of the object within the phantom.

3. RESULTS

The generalized system performance evaluation parameters were calculated following the equations defined in section 2.3. For the performance comparison of the two systems, we present the comparison of the generalized parameters measured at 80 kVp. Figure 7 shows a plot of GMTF as a function of spatial frequency at a fixed air-gap (AG) of 2.5cm and varying the radiation field sizes measured at the detector plane. Both the II and the microangiographic system show similar trends with respect to the variation in radiation field size. A more realistic comparison of the two systems according to their commonly used configuration is to compare the full field of view of the image intensifier with grid and 16 cm² radiation field size of the microangiographic system. Generally, increase in radiation field size degrades the GMTF of both the systems compared to their bare detector MTF. Figure 8 shows the behavior of the GMTF with respect to the change in air-gap, while radiation field size was kept constant at 36 cm² for both the detectors. The air-gap 2.5 cm is more realistic for clinical studies in neurovascular angiography where the detector should be close to the patient. The larger air-gaps provide results for magnification radiography.

Figure 9 shows the effect of magnification on the GMTF of the image intensifier and microangiographic system. The magnifications 1.1 and 1.2 were defined by keeping the head phantom at a fixed air-gap of 2.5 cm and changing the depth of the object within the phantom, whereas the magnification 1.3 was defined for the air-gap of 8.0 cm because of the constraint on the depth of the object within the limited height of the phantom. Increasing magnification causes a substantial degradation in the high frequency GMTF of the microangiographic system, but at the same time it improves the high frequency GMTF of the image intensifier system.

Figure 10 shows a representative plot of the GNNPS of the II and microangiographic system at three different detector entrance exposures. It infers that the microangiographic detector has less system noise than the II detector for a given detector entrance exposure. Figure 11 shows the relative comparison of GNEQ curves of the two systems, which physically represent the absolute measure of image quality from the total system point of view. The microangiographic system generally leads over the II system in terms of the image quality.

Figure 12(a) represents a relative performance comparison of GDQE curves of the two systems at three different exposures to the detector. The GDQE of the two systems has no appreciable dependence on exposure as expected. Since the GNNPS does not depend on the scatter fraction or radiation field size for a constant detector exposure, the GDQE will be governed by the changes in GMTF through the variation of these parameters, except a rescaling with the GNNPS term. Therefore, the qualitative behavior of the GDQE will be the same as that of the GMTF with respect to the variation in radiation field size and air-gap.

In general, the GDQE of the microangiographic system is better than that of the II system with the grid. In order to make an alternate performance comparison of the two systems, a plot of the GDQE of microangiographic system and II without the grid is shown in Figure 12(b). In this case the zero frequency GDQE of the II system is higher than that of the microangiographic system. However, the later performs substantially better at the high spatial frequencies which become important when imaging small objects of neuro-interventional significance.

4. DISCUSSION

The GMTF estimates the resolution properties of the total imaging system including the effect of patient and focal spot blurring. From Figure 7, the effect of increasing radiation field size is to degrade the system resolution because of the increased amount of scatter at the entrance to the detector. From the relative comparison of figure 7(a) and 7(b), the GMTF of the image intensifier system is relatively less degraded compared to the microangiographic system and in general, the GMTF of the II system at low spatial frequencies is better compared to the microangiographic system, but the microangiographic system performs substantially better at higher spatial frequencies, i.e., in the regime of high resolution neuro-interventional procedure requirements. This difference is largely due to the effect of the anti-scatter grid with the II system compared to the absence of the grid with the microangiographic system. Assuming the most usual operating field of views of the two detectors, i.e., the full field of view of the II and 16 cm² of the microangiographic system, the GMTF of the two systems are very close to each other even at low spatial frequencies. Thus, considering the region-of-interest and high-resolution requirement in the neuro-interventional procedures the overall performance of microangiographic system is in general better compared to the II system.

From figure 8 it is noticeable that up to a certain spatial frequency the GMTF of both systems improves with increasing air-gap because of the scatter reduction, and then reaches a cross-over point. Beyond this cross-over point the GMTF of the microangiographic system starts degrading whereas that of the II system continues to improve even well than the

detector MTF. This shows that the image intensifier performs better in magnification radiography, at least for low magnifications. The effect of increasing air-gap has in turn two factors that contribute to the GMTF simultaneously: the scatter fraction – dominant at low spatial frequencies, and the focal spot blurring – dominant at higher spatial frequencies. The improvement in the GMTF at low spatial frequencies is because of the decreased scatter to the detectors with increasing air-gap. The higher frequency behavior of GMTF is attributed to the difference in the effective pixel size of the two detectors, in combination to the focal spot size. For the same focal spot (0.6mm nominal), the focal spot blurring will have a greater impact on image captured with the microangiographic detector, which has a pixel size of 0.043mm, than on image captured with the II system, which has a pixel size of 0.116mm. The relative difference in the low frequency drop of the GMTF of the two systems for the same air-gap is again because of the effect of the anti-scatter grid with the II system. In the typical set-up for high resolution neuro-interventional procedures (patient should be close to the detector), we can compare the case of the 2.5 cm air-gap where the microangiographic system has better GMTF at higher frequencies compared to the II system having a better GMTF at low frequencies. When the air-gap was fixed and the depth of the object inside the phantom was varied ($m = 1.1$ and 1.2), the impact of increasing magnification on the GMTF (Figure 9) is basically focal spot blurring, which is dominant at higher frequencies. For the $m = 1.3$ case, where the air-gap was different than the previous two, the scatter as well as focal spot blurring affected the GMTF, as in the case of magnification radiography.

The GNNPS of the two systems represents the noise characteristics of the total x-ray imaging system relative to the object plane (Figure 10). The microangiographic system is less noisy compared to the II system for the same exposures to the detector entrance. Increasing the exposure to the detector decreases the relative noise because of the larger number of photons entering the detector. The system NNPS, as defined by equation (4), is independent of the scatter fraction because these detectors do not distinguish between scatter and primary when a photon is detected. An anti-scatter grid with the II system leads to considerable reduction in the number of photons reaching the detector, and hence resulting in higher relative system noise compared to the microangiographic system for the same exposure exiting the patient.

The GNEQ is a measure of absolute image quality which includes all factors influencing image formation under realistic clinical conditions. The GNEQ of the microangiographic system is better compared to the II system for the same exposure exiting the patient and the usual operating geometry of the two detectors (Figure 11). This is a consequence of equation (5), where the GNEQ improves according to the square of the GMTF and degrades with increasing GNNPS. The higher GMTF and smaller GNNPS of the microangiographic system at usual operating conditions lead to improved image quality compared to the image intensifier system.

The GDQE is a measure of the absolute system efficiency in detecting primary photons. The GDQE of the two systems did not show any appreciable dependence on the detector entrance exposure, as expected. This is a result of the normalization by the number of photons relative to the object plane. The GDQE of the microangiographic system is

substantially better compared to that of the II system with the grid (Figure 12(a)). This lower GDQE of the II system compared to the microangiographic system is because of the anti-scatter grid that decreases the effective number of photons reaching the II and causes higher quantum. The low frequency abrupt drop in the GDQE of the two systems is due to the large scatter at the smaller air-gap (2.5 cm). Compared to the detector DQE (no scatter, no blurring) the low frequency drop in the GDQE of the microangiographic system is relatively greater than that in the II system. This is because the grid with the II system reduces the scatter fraction. The zero frequency GDQE of the II with the grid (0.20) is fairly low as compared to the microangiographic system (0.47). An alternate GDQE comparison of the two systems can be made by removing the grid. In this case the GDQE of the II system is better at very low spatial frequencies compared to the microangiographic system (Figure 12 (b)); however, overall the microangiographic system performs substantially better compared to the image intensifier system. The effect of focal spot blurring on the GDQE was qualitatively similar to that on the GMTF.

5. CONCLUSION

A generalized performance comparison of a commercial x-ray image intensifier and a high resolution, region-of-interest, microangiographic system under simulated neurovascular angiographic conditions was performed. A detailed performance analysis in terms of the GMTF, GNNPS, GNEQ and GDQE was presented. These generalized performance comparison parameters included the effect of scatter due to the patient and focal spot blurring combined with the detector performance parameters, which led to a total system-based analysis for the clinical situation. The effects of varying radiation field size, air-gap and magnification on the generalized performance parameters of the two systems were presented. We conclude that the microangiographic imaging system will be a better detector system for the image-guided neurovascular interventional procedures, because it has substantially better performance at higher spatial frequencies which are important in such interventional procedures. If a small focal spot (0.3 mm) were used, there would be less degradation in the generalized parameters at higher spatial frequencies, and that would be beneficial for the microangiographic system performance in such procedures. This generalized, total system-based analysis can provide realistic evaluation of such systems leading to improved system designs tailored to the imaging task.

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Figure 1.

Picture showing the microangiographic detector in place with a human head phantom. The II is moved all the way up to insert the microangiographic detector during image acquisition; the microangiographic detector can be moved out of the field by moving the arm, and II can be positioned as needed.

Figure 2.

Typical detector MTF of the image intensifier and the microangiographic detector measured at 80 kVp: the plot terminating at 4.3cycles/mm corresponds to the II and that terminating at 11.6cycles/mm corresponds to the microangiographic system.

Figure 3. Low spatial frequency behavior of the Focal Spot MTF (MTF_F) and the Scatter MTF (MTF_S) measured at 80 kVp. The Scatter MTF is significant only at very low spatial frequencies.

Figure 4. Plot showing the surface fit of scatter fraction (ρ) as a function of radiation field size (α) and air-gap (AG). Points show the corresponding measured data points.

Figure 5(a).

Two-dimensional NNPS of the II at the detector entrance exposure of 0.4 mR, measured before the grid at tube voltage of 80 kVp.

Figure 5(b).

Comparison of the polynomial fit and the measured one-dimensional Normalized Noise Power Spectrum corresponding to Figure 5(a).

Figure 6. Comparison of the measured NNPS and the fit with equation (2) for a range of spatial frequencies 1, 2, 3 and 4 cycles/mm (from top to bottom respectively).

Figure 7.

Plots showing the effect of varying radiation field size on the GMTF at 80 kVp with fixed air-gap of 2.5 cm: **7(a)** Image intensifier system; **7(b)** microangiographic system shown to the same frequency scale as in the II; **7(c)** comparison of the two systems on the same plot till the Nyquist frequency of each system.

Figure 8.

Plots showing the effect of varying air-gap on the GMTF at 80 kVp with fixed radiation field size of 36 cm^2 : 8(a) Image intensifier system; 8(b) microangiographic system shown to the same frequency scale as in the II; 8(c) comparison of the two systems on the same plot till the Nyquist frequency of each system.

Figure 9.

Comparison of GMTF of the II and the microangiographic system at magnifications 1.1, 1.2 and 1.3: Plots terminating at 4.3 cycles/mm correspond to the II system and the others terminating at 11.6 cycle/mm correspond to the microangiographic system

Figure 10. Comparison plots of GNNPS of the II with grid and the microangiographic system: plots terminating at 4.3 cycles/mm correspond to the II system

Figure 11. Comparison plots of GNEQ of the II with grid and the microangiographic system: plots terminating at 4.3 cycles/mm correspond to the II system

Figure 12(a).
Comparison plots of the GDQE of the *II* with grid and the microangiographic system.

Figure 12(b).
Comparison plots of the GDQE of the *II* without grid and the microangiographic system.