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ASSESSING CONTRAST SENSITIVITY BEHIND CLOUDY MEDIA

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Summary—It is often important to assess visual neural function despite the presence of cloudy optic media. A psychophysical method is described which may prove useful for assessing visual function behind cloudy optics. Contrast thresholds for grating patches were measured in the presence and absence of one dimensional dynamic noise as a function of spatial frequency and noise level. These measurements provide a means for analyzing a visual deficit into optical-like and non-optical-like components. Given certain assumptions, a purely optical loss will depress the contrast sensitivity function, relative to a normal control, in the absence, but not in the presence of high levels of added visual noise. This predicted effect was borne out by simulation and found in one patient. Because optical-like neural deficits may mimic this behavior, it is necessary to look for and study neural deficits which are not optical-like. Out of 11 patients with normal optics, but with neural loss, only one showed an optical-like deficit.

Key words—Contrast sensitivity; optical deficits; neural deficits; visual noise.

INTRODUCTION

How can one assess neural visual function when optic media deficits obscure the view of both the subject and examiner? In older patients, cataracts are often accompanied by macular degeneration or other disorders. Techniques which accurately predict visual function following cataract surgery or corneal implantation would benefit both clinician and patient. Our approach to this problem is to measure the contrast sensitivity function in the presence and absence of visual noise. These measurements provide additional data which may bear on the causes of loss of sensitivity.

Several psychophysical methods have been investigated to assess visual neural function behind cloudy media. Green used a laser interferometer to form sine-wave gratings at the retina (Green, 1970; Green and Cohen, 1971). This technique can be used to bypass the optical modulation transfer function (MTF) if two clear points can be found in the entrance pupil to image two point sources. This has the potential advantage of assessing acuity up to the sampling limit imposed by the receptor spacing (Campbell and Green, 1965). As Green points out, in practice it can be difficult to find any clear points in the pupil. Each higher spatial frequency tested requires finding a more widely separated pair of clear spots. Further, scattering by particles along the optical path leads to laser speckle which can completely obscure the target grating. Another technique, which also requires a small clear region in the pupil, is to present an acuity target in Maxwellian view (Cavonius and Hilz, 1973). Although, it has practical advantages over interferometry (Minkowski *et al.*, 1983), like interferometry, it works best with clear regions of the optic media. The technique we report may complement these acuity measures by providing information about contrast sensitivity at low spatial frequencies, whether or not there are clear regions in the optics.

Another technique for discounting the peripheral causes of loss in contrast sensitivity was developed by Hess and Bradley (1980; Hess *et al.*, 1983). Contrast matching functions show little effect of the high or low spatial-frequency fall-off in sensitivity generally attributed to peripheral mechanisms, both optical and retinal (Georgeson and Sullivan, 1975). Using a contrast matching paradigm, Hess and Bradley measured contrast matching for amblyopes and showed that the contrast loss at threshold is not carried over to suprathreshold levels, whereas in optic neuritis, the threshold and suprathreshold losses are comparable (Hess, 1983).

It has been known for some time that certain vernier-acuity stimuli can be substantially blurred with little drop in performance (Stigmar, 1971; Westheimer, 1979). This means that sufficient information for this task is carried in the low spatial frequencies which in turn suggests a primarily neural rather than optical limit to hyper-acuity. Recently, Enoch and Williams (1983) developed a clinical version of this technique which may be useful in predicting visual function following cataract surgery. The technique holds promise by virtue of the relative speed and simplicity of the measurement.

One advantage of extending contrast sensitivity function (CSF) measurements to higher contrasts using a noise mask, is that the CSF has become advocated as a clinical aid to diagnosis (Bodis-Wollner, 1972; Regan *et al.*, 1977; Hess and Woo, 1978; see Legge and Rubin, 1986 for a critique). Our knowledge of the way in which different neural deficits affect the contrast sensitivity function is growing rapidly. Designing a differential diagnostic test around this approach allows one to capitalize on this information. Furthermore, CSFs have a rich theoretical and empirical research base, partly due to the power of sine-wave amplitude- and phase-transfer functions to completely characterize linear shift-invariant systems.

Another advantage is that the interpretation of contrast thresholds in visual noise has received theoretical attention recently. Comparing thresholds in noise with those in the absence of noise has led to an account of two different limits to visual sensitivity (Barlow, 1977; Pelli, 1981; Burgess *et al.*, 1981; Kersten, 1983). Squared-contrast thresholds as a function of noise level are well fit by a linear function. The slope and intercept of this line are related to two types of limits to visual contrast sensitivity (see Appendix; Legge *et al.*, 1987). One limit, *equivalent noise*, goes up if there is either increased additive internal noise or if contrast gain prior to internal additive noise goes down. Equivalent noise often goes up due to bad optics, but may also be due to neural factors (Legge *et al.*, 1987). Below, we use equivalent noise to define a measure of optical-like efficiency. This is motivated by the observation that an optical deficit causes a theoretically predictable increase in equivalent noise (see Appendix). A second component, *sampling efficiency* (SE), goes down if a non-optimal decision strategy is used. Even though the contrast of the signal may be transmitted normally up to some point, the signal may not be processed efficiently. The term "decision strategy" does not necessarily imply cognition, but lumps together a number of factors other than a change in contrast gain which affect sensitivity. For example, the signal may not be processed efficiently simply because of a loss of image samples due to a complete scotoma. The Appendix develops a simple model of detection in white Gaussian noise which illustrates these ideas. In this research we investigate whether optical-like and sampling efficiency have the potential to tease apart optical from certain types of neural visual loss.

We do this by comparing the contrast sensitivity for sine-wave gratings in the presence and absence of added visual noise. It is tentatively assumed that signal and noise are equally attenuated by optical loss over a narrow frequency band, and it is this signal-to-noise ratio which determines detectability (Stromeyer and Julesz, 1972). If the loss is optical or optical-like, then both the test signal and visual noise contrast are attenuated equally in a narrow spatial frequency band near the signal, leaving the signal-to-noise ratio unchanged. Thus, the observer's threshold is the same as a normal's. In order to usefully quantify these descriptions, we define the observer's *relative sampling efficiency*, (relative SE) as the ratio of the patient's to normal's SE. However, relative SE is most usefully thought of as approximately the squared ratio of the normal to patient contrast threshold in high noise. *Relative optical-like efficiency* (relative OLE) is the ratio of the patient's equivalent noise to that of the normal*. If a visual deficit is only due to a loss of contrast gain, relative sampling efficiency should be near 1, but optical-like efficiency less than 1. If a visual deficit is due to a non-optimal decision strategy, relative sampling efficiency should be less than 1. A third measure which is useful is *relative detection efficiency* (relative DE), which is the product of relative SE and relative OLE. Relative DE is useful because it is inversely proportional to the square of contrast threshold in the absence of noise and is thus directly proportional to the square of the standard contrast sensitivity measurement (Pelli, 1981).

*Relative optical-like efficiency is the ratio of the patient to normal *transduction efficiency*—a term coined by Pelli (1981).

Our question can be reduced to asking whether there is some range of spatial frequencies over which the patient has normal contrast sensitivity in the presence of visual noise. In cases of cloudy media, if the patient's contrast sensitivity function in high levels of noise is the same as a normal observer's, this may be evidence in favor of the patient having normal neural function over the spatial frequency range measured. If this does not occur, and the contrast deficit remains unchanged or increases relative to a normal observer, then the patient may suffer from more than loss of optical-like efficiency over this range. However, if this technique is to be useful, we need to find out whether optical and neural deficits manifest themselves as differential changes in optical-like and/or sampling efficiencies.

We conducted two experiments. In the first, we estimate sampling efficiency and equivalent noise by measuring contrast thresholds in increasing levels of noise, for a fixed spatial frequency. In the second experiment, we measure contrast sensitivity functions in the presence and absence of visual noise. These data permit us to derive average measures of the relative efficiencies over the entire spatial frequency range of an observer's CSF. We were also interested in an efficient measure of psychometric function slope which might be useful in a clinical setting. This was motivated in part by the suggestion of Patterson *et al.* (1980) that abnormally shallow psychometric functions in optic neuritis might be a consequence of internal variability or noise. Thus, we also concurrently estimated psychometric function slope during a testing session.

METHOD

Apparatus

Stimuli were presented on the face of a Joyce Electronics CRT display by Z-axis modulation. The display had a white P4 phosphor and an unmodulated mean luminance of 380 cd/m^2. The screen was 30 cm wide and 16 cm high. Luminance waveforms were synthesized digitally by a PDP 11/40 computer. A 12-bit multiplying DAC generated a sinusoidal waveform modulated by the temporal envelope. The output of the DAC in turn modulated a horizontal envelope produced by a second multiplying DAC. The output of the second DAC was low-pass filtered to prevent aliasing and was passed to a 9-bit programmable attenuator. The output of the attenuator was passed to the Z-axis input of the display. This signal was multiplied along the raster lines by an 8-bit fast-buffer store via the high-bandwidth multiplying Z-axis imput of the display. The contrast was controlled in quarter dB steps by the attenuator.

Stimuli

The signal was a sine-wave grating with Gaussian enveloped windows in the horizontal, vertical and temporal dimensions (Plate 1). The spatial frequency at the screen was kept at 0.25 cycles/cm, except for the 0.0625 c/deg conditions, where it was 0.125 c/cm. For all but this latter condition, the windows extended vertically and horizontally to 6 cm measured between 1/e points of the envelope. To keep a constant number of cycles under the window, the windows were twice this size for the 0.0625 c/deg grating. The duration between 1/e points was 500 msec.

The noise was dynamic and varied only in the horizontal direction*. It had a flat temporal spectrum out to about 50 Hz (half the frame rate) and spatial spectrum out to 6.2 c/cm. Binary noise was synthesized at 8 MHz via a 31-bit shift register (Horowitz and Hill, 1980) and then passed through a pair of Krohn–Hite 6-pole low-pass filters with cut-offs of 40 Khz. As a consequence of the filtering and the Central Limit Theorem, the output voltages were approximately Gaussian distributed. The filters also served to amplify the noise. The noise from the filter was added to the signal grating by an analog summing junction in the display. The r.m.s. contrast of the noise at the screen was constant at 56%. This limited reliable contrast sensitivity measurements to above 6 dB (i.e. below 50%).

Procedure

Contrast thresholds were measured with a temporal two-alternative forced-choice procedure. In order to estimate psychometric function slope, the Quest algorithm (Watson and Pelli, 1983) was adapted to estimate the 70 and 90% points of the psychometric function. A

*If the noise had been two-dimensional, the noise power would have been distributed over a broader total band, thus reducing the noise spectral density (and perceptual "noisiness" of the noise). For some of the low-vision observers the equivalent noise was too high to permit two-dimensional noise—no noise level was high enough to change thresholds from the no noise condition.

block consisted of 100 trials—on the even and odd trials the algorithm tracked the 70 and 90% points respectively. The slopes were calculated by taking the difference of the logarithms of d' at 90 and 70% correct divided by the difference of the corresponding log contrasts.

Experiment 1. In order to get reliable estimates of sampling efficiency and equivalent noise, contrast thresholds were measured for 6 observers as a function of noise level for a fixed viewing distance and spatial frequency. For AM1, the viewing distance was 10 m, and the spatial frequency equaled 5 c/deg; for AM2, MD2, ON1, and ON4 the viewing distance was 75 cm and the spatial frequency 0.25 c/deg; for N1 and CM1 (N1 with diffuser), the viewing distance was 114 cm and the spatial frequency 0.5 c/deg. For AM1, AM2 and MD2, thresholds were also measured on their good eyes.

Experiment 2. Contrast thresholds were measured as a function of spatial frequency for two noise levels at each spatial frequency. For all but the 0.06 c/deg condition, spatial frequency with respect to the observer was varied by changing viewing distance appropriately: 460, 228, 114, 57 and 28 cm, resulted in spatial frequencies of 2, 1, 0.5, 0.25 and 0.125 c/deg respectively. Both the 0.12 and 0.06 c/deg signals were produced at the 28 cm viewing distance. Noise spectral density is defined as the r.m.s. contrast squared divided by the two-sided bandwidth of its spectrum

$$\frac{c^2_{\text{r.m.s.}}}{B_x B_t}$$

where B_x and B_t are the two-sided spatial and temporal frequency bandwidths. This in our case gives $(0.56)^2/(12.4 \times 100)$ equals 2.5×10^{-4} sec-cm. Thus the noise spectal densities at the viewing distances 28, 57, 114 and 228 cm were 500, 250, 125 and 63 micro-sec-deg respectively.

The data for one subject (CM3) and a normal control (N3) were collected in another laboratory under slightly different conditions. The viewing distance and noise level were kept fixed at 14 cm and 10^{-3} sec-deg, respectively, for all spatial frequencies tested. In addition, the grating was vignetted by a Gaussian envelope of width 64° between $1/e$ points. The duration was 160 msec between $1/e$ points. The display was a Joyce but with P31 phosphor and the mean luminance 340 cd/m^2. Data at 0.03 c/deg were collected in addition to the other spatial frequencies. Thresholds were collected at the 75% correct point only. Other details were as above. A minimum of one block was run for each noise level and/or spatial frequency.

Subjects

Table 1 summarizes the clinical details for the patient population. In all, there were three normal observers: N1, N2 and N3; three with cloudy media: one simulated cataract, CM1; one with cataract, CM2; one with corneal scarring CM3; four with optic neuritis: ON1, ON2, ON3, ON4; four with macular degeneration: MD1, MD2, MD3, MD4; and three amblyopes: AM1, AM2 and AM3.

RESULTS

The data are presented in two sections. In section 1, Figs 1–4 represent contrast sensitivity as a function of noise level for Experiment 1. Figure 1 illustrates contrast energy and the calculation of sampling efficiency and equivalent noise. Figures 2–4 show contrast sensitivity as a function of noise level. In section 2, contrast sensitivity functions, from Experiment 2, are shown. Figures 4–13 show contrast sensitivity

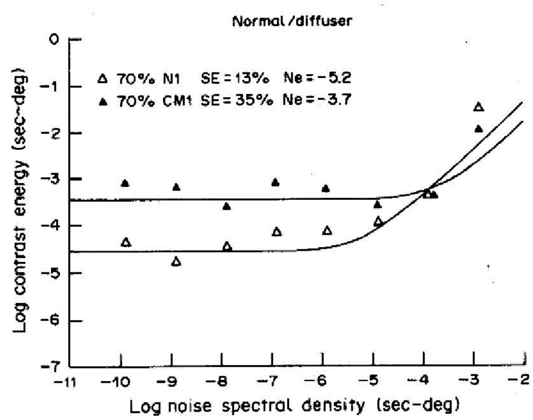


Fig. 1. Log contrast energy thresholds are plotted as function of log noise spectral density for observer N1 with (solid triangles) and without (open triangles) the diffuser for just the 70% correct points. (The data averaged over 70 and 90% correct points are plotted in terms of contrast thresholds in dB in Fig. 2.) The lines plotted represent the linear regression fits in *linear coordinates* to equation (5). Because contrast thresholds generally have standard errors which are constant in log coordinates, regression was performed assuming the variance was proportional to the mean value of the contrast energy. The sampling efficiencies (SE) are 13% and 35% for the normal and normal with diffuser respectively. *Ne* represents the log equivalent noise. The difference in levels at the left hand side of the graph is approximately the logarithm of relative DE. The difference in the threshold levels at the right hand side of the graph is approximately the logarithm of relative SE.

Table 1

Case	Age	Visual acuity	Visual field	Clinical details	Ishihara test No. of errors (max 12)	Optic disc
Optical loss						
CM2	65	1/60	Full	Nuclear cataract	—	—
CM3	35	1/60	Full	Secondary corneal vascularization due to Steven Johnsons Syndrome	—	—
Optic neuropathy						
ON1	34	6/60	Dense paracentral scotoma	Optic atrophy due to mulitple sclerosis	12	Optic atrophy
ON2	39	1/36	Dense central scotoma	Optic atrophy due to multiple sclerosis	12	Optic atrophy
ON3	69	3/60	Generalised depression	Bilateral idiopathic optic atrophy	12	Optic atrophy
ON4	26	6/36	Relative paracentral scotoma	Optic atrophy due to multiple sclerosis	12	Optic atrophy
Amblyopia						
AM1	28	6/24	Full	Anisometropic amblyope central fixation	—	Normal
AM2	42	1/60	Full	Strabismic amblyope, 3° esotropia, central fixation	—	Normal
AM3	58	2/60	Full	Strabismic amblyope (esotropic) central fixation	—	Normal
Age related Maculopathy						
MD1	68	1/60	Dense central scotoma	Pseudodisciform degeneration	—	Normal
MD2	65	4/60	Central scotoma	RPE degeneration subretinal neovascularization	—	Normal
MD3	71	3/36	Central scotoma	Disciform degeneration	—	Normal
MD4	60	6/18	Full	Disciform degeneration	—	Normal

functions in the presence and absence of noise for the cloudy media (Figures 5–7), optic neuritis (Figs 8 and 9), macular degeneration(Figs 10 and 11) and amblyopia conditions (Figs 12 and 13). Figures 14 and 15 summarize the results of Experiments 1 and 2, respectively.

Contrast sensitivity as a function of noise level

In order to extract summary measures from the data in Figs 2–4, we have fit curves to contrast energy thresholds as a function of noise spectral density using linear regression on equation (5) (see Appendix). Figure 1 illustrates two such fits for the normal and normal plus diffuser condition. Note that, although the data are actually plotted on logarithmic coordinates, the fits are to a linear equation. The logarithms of the equivalent noise, Ne, is an estimate of the position of knee of the curve. It can be thought of as that additional noise at the screen which raises the contrast energy threshold by a factor of 2. The sampling efficiency, SE, is estimated from the slope of the linear regression fit. It is the asymptotic value of the ratio of the ideal's

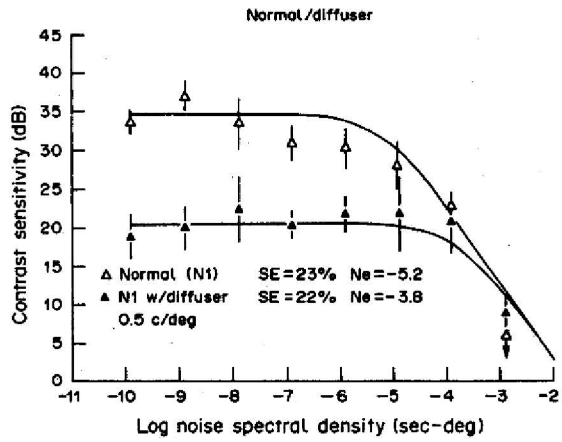


Fig. 2. Average contrast sensitivity as a function of noise level for the diffuser condition. The fact that sampling efficiencies are close (23 and 22%) is reflected in the contrast thresholds for the diffuser data approaching the normal thresholds in high noise, at about -4 log units. The optical deficit raises the equivalent noise by about 1.4 log units, from -5.2 to -3.8 log sec-deg. (These figures differ from those in Fig. 1, because they represent averages over the 70 and 90% correct points.)

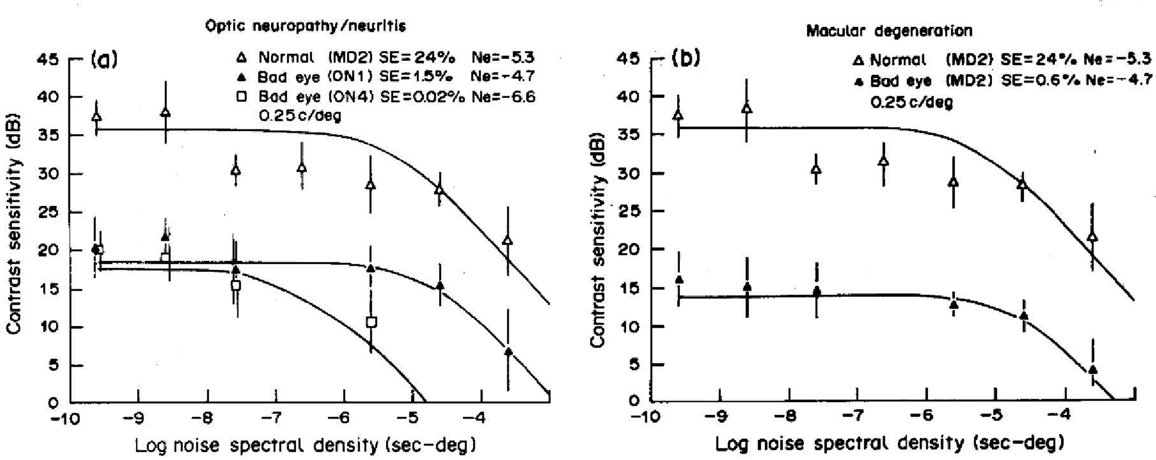


Fig. 3.(a) Data for the optic neuritis condition for subjects ON1 and ON4 collected at a spatial frequency of 0.25 c/deg. Data from MD2's normal eye are also shown for comparison. Note that even in the highest noise levels contrast thresholds do not approach those for the normal. Partly in consequence of this, the sampling efficiencies are low for both observers' bad eyes—1.5 and 0.02% for ON1 and ON4 respectively. This should be contrasted with the high sampling efficiency for MD2's and N1's good eyes—24 and 23% respectively. Further note that at low noise levels, both optic neuritis subjects had thresholds comparable to those of N1 with the diffuser (about 20 dB), but they drop more rapidly with increased noise with respect to the normal data. Although the equivalent noise for ON1 is larger than normal, ON4's is lower. (b) Contrast thresholds for macular degeneration case MD2 at 0.25 c/deg. Again contrast thresholds do not approach those of the normal eye in high levels of noise. The sampling efficiencies for the normal and abnormal eye are: 24 and 0.6% respectively. The equivalent noise is about 0.6 log units higher for MD2's bad eye.

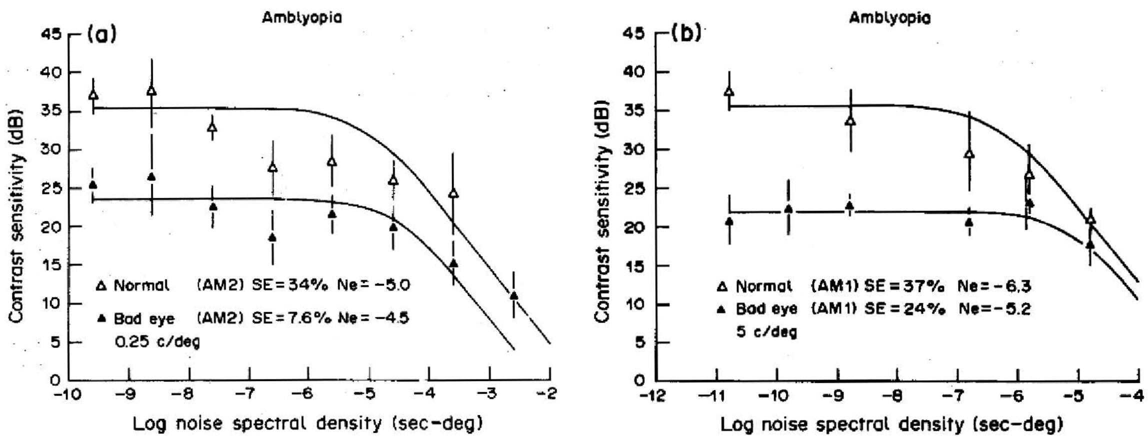


Fig. 4.(a) Data for the good and bad eyes of a strabismic amblyope AM2. As with the optic neuritis and macular degeneration cases, thresholds for the bad eye never reach those of the good eye in high noise levels—the loss is not optical-like. This is despite the fact that the thresholds in the absence of noise are only 12 dB apart— less than for the diffuser case in Fig. 2. The sampling efficiencies for the good and bad eyes were 34 and 7.6% respectively. Equivalent noise for the bad eye goes up by half a log unit. (b) Contrast thresholds for an anisometropic amblyope AM1 for normal and abnormal eyes. Unlike the previous amblyope [AM2, Fig. 4(a)], AM1's thresholds for his bad eye do approach those of the normal eye in high noise levels. This amblyope behaves as if there was an optical-like deficit. The sampling efficiencies for the good and bad eyes were 37 and 24%, respectively. The deficit is reflected primarily by a 1.1 log unit increase in equivalent noise.

Figs 2–4. The five panels show the average of contrast thresholds at the 70 and 90% points (in dB relative to 100% contrast) as a function of log noise spectral density. Open symbols and solid symbols represent data from a normal eye and bad eye respectively. As in Fig. 1, sampling efficiency and equivalent noise are indicated by SE and *Ne*. The ends of the vertical bars indicate 70% and 90% threshold estimates.

contrast energy threshold to the subject's as the noise level increases. SE and Ne were used to calculate the relative efficiencies. Because contrast is a more familiar measure than contrast energy, for the rest of the conditions, data are plotted as the average (over the 70 and 90% correct points) contrast thresholds in dB vs. log noise spectral density. (One dB is equivalent to about a 12% drop in contrast.) The smooth curves show fits to equation (5), in which d' is averaged over the 70 and 90% points.

Figure 14 summarizes relative DE, SE and OLE for each condition in Figs 2–4. Note that, as might be expected, all the relative detection efficiencies [Fig. 14(A)] are at least a factor of 10 below 1. The relative sampling efficiency [Fig.

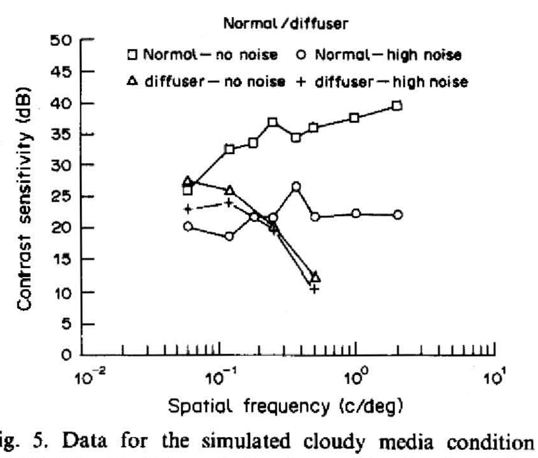


Fig. 5. Data for the simulated cloudy media condition (CM1), N1 with diffuser. In the absence of noise, thresholds are close only for the lowest spatial frequency (0.06 c/deg), whereas thresholds in the presence of additive noise are comparable out to 0.25 c/deg after which they begin to drop. This shows us the spatial frequency region over which the convergence of sensitivities seen in Fig. 2 hold. The noise level is the highest used in the experiments of the previous section. Note that normal contrast sensitivity in noise does not show the low-spatial frequency drop off typical of low contrast threshold sensitivity functions.

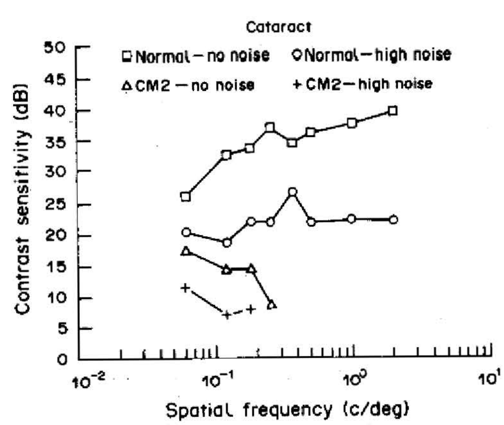


Fig. 6. Data for cataract patient CM2. Over the frequencies tested, neither thresholds in noise nor in the absence of noise approached normal thresholds.

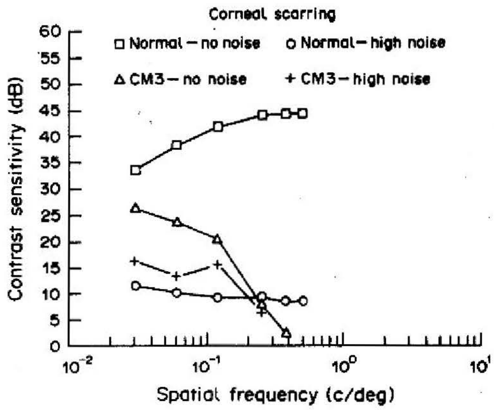


Fig. 7. Data for observer CM3, an observer with a cloudy cornea. Similar to CM1, CM3 has severely abnormal contrast sensitivity in the absence of noise, but actually did better than the normal control over frequencies from 0.03 to 0.125 c/deg when thresholds were measured in noise.

14(B)] is above 1 for the diffuser (CM1) and anisometropic amblyope (AM1). The SEs of all the rest are below 1. As we would predict, the optical-like efficiency [Fig. 14(C)] is very low for CM1. However, only ON4 has an optical-like efficiency above 1 and in fact, is much larger than 1 (the actual value, off the graph, is 14.6). Because there are no known optical deficits in patients other than CM1, it is clear that relative optical-like efficiency is not necessarily near 1 for cases of clear optics. This is consistent with a neural basis for the loss in optical-like efficiency.

Contrast sensitivity as a function of spatial frequency

Figures 5–13 show contrast sensitivity functions in the presence and absence of noise for the optic media deficit, optic neuritis, macular degeneration and amblyopia conditions. Except for Fig. 7, thresholds in the absence (open triangles) and presence (solid triangles) of noise for the average of normal eyes N1 and N2 are replotted on each graph. Because only two points are required to fit equation (5), sampling efficiency and equivalent noise can be estimated for each pair of thresholds at a given frequency. However, SE and Ne are very susceptible to error with only two points. We have sum-

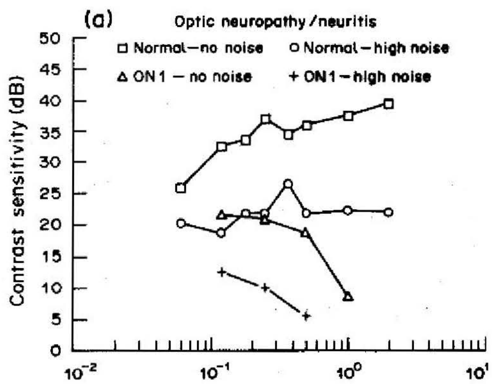


Fig. 8(a)

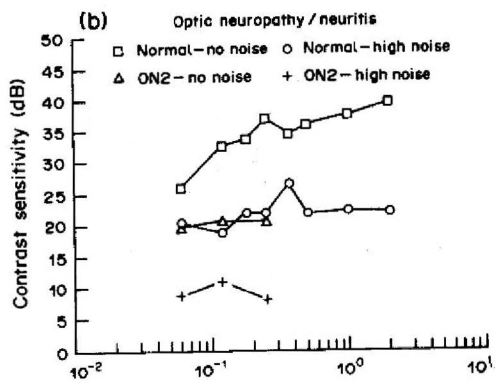


Fig. 8(b)

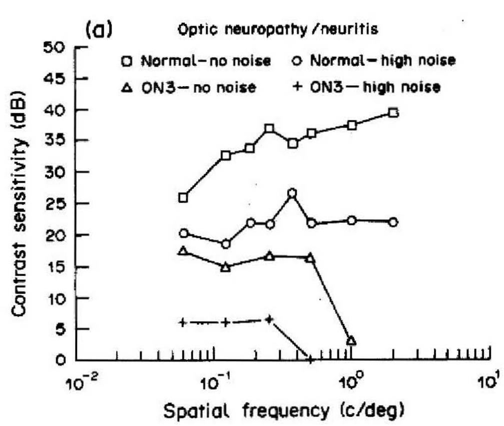


Fig. 9(a)

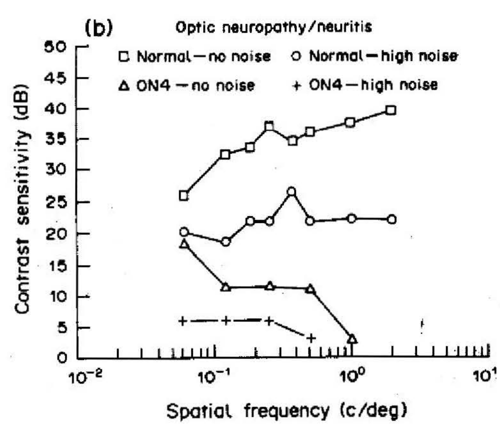


Fig. 9(b)

Figs 8–9. The four panels show contrast sensitivity functions for four optic neuritis patients.

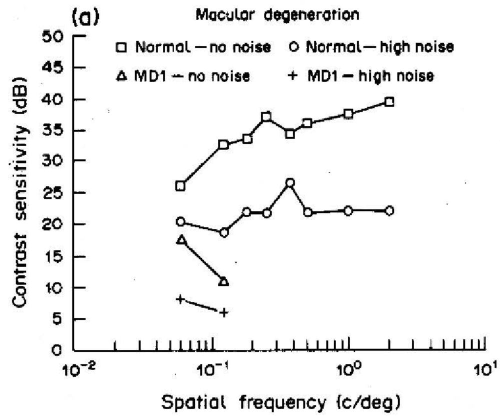


Fig. 10(a)

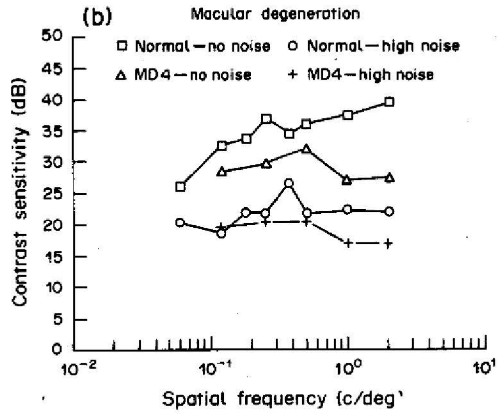


Fig. 10(b)

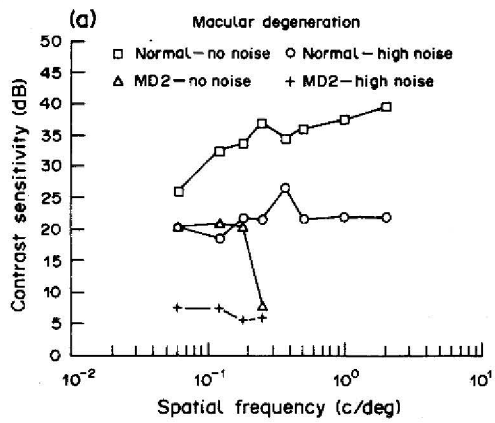


Fig. 11(a)

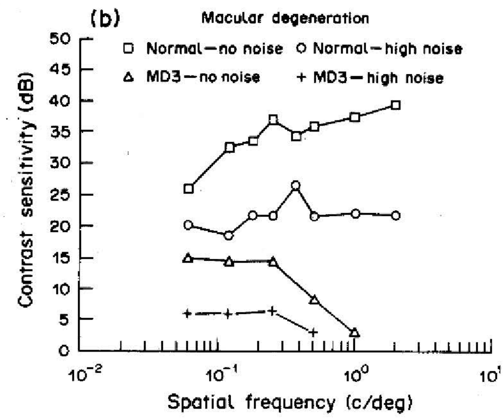


Fig. 11(b)

Figs 10–11. The four panels show contrast sensitivity functions for the macular degeneration patients. For none of the seven subjects represented by Figs 8(a, b), 9(a, b), 10(a) and 11(a, b), do thresholds in the presence or absence of noise approach normal over the range of spatial frequencies tested. In other words, for no spatial frequencies are their sensitivity losses optical-like. However, for MD4 [Fig. 10(b)], thresholds in noise are fairly close to normal from 0.125 to at least 0.5 c/deg.

marized these data by averaging relative detection, sampling and optical-like efficiencies across spatial frequency. Figure 15 shows the average relative efficiencies for Experiment 2. As before, relative detection efficiencies [Fig. 15(A)] are all substantially less than 1. However, relative sampling efficiency [Fig. 15(B)] is 1 or higher for both CM1 and CM3. Only MD4 and AM1 approach the performance of the two optical deficit conditions, CM1 and CM3. However, CM2's relative SE is considerably below normal. Whether or not this is indicative of a neural deficit would require post-operative testing. In contrast to Experiment 1, Fig. 14(C), the relative OLEs of many of the patients with neural loss are well within the norm of 1. However, none of the patients in the optical-loss category have normal OLEs.

We did find some differences in psychometric function slope between individuals. However, these differences were small and we may not be able to reliably distinguish small psychometric function slope differences with this technique*.

*For Experiment 1, log–log slopes were compared using a two-tailed pair-wise t-test across noise levels between each subject's good and bad eyes where possible (AM1, AM2, MD1, CM1). A pair-wise two-tailed t-test was used to compare slopes across spatial frequency from bad eyes with the normal average (for N1 and N2) in Experiment 2. The slopes of optic neuritis subjects were compared with the slopes of MD1. In only two of the 19 comparisons was $P < 0.05$. This was only for the comparisons of slopes as a function of spatial frequency ($P < 0.05$ for MD4 and $P < 0.01$ for ON4, both in the absence of noise). For MD4, the average slope across spatial frequency was 1.31 (SE = 0.11, N = 5) slightly higher than 0.96 (SE = 0.06, N = 7)—the average of N1 and N2 (over spatial frequency). ON4's average slope, 0.82 (SE = 0.02, N = 4) was slightly lower than the average normal.

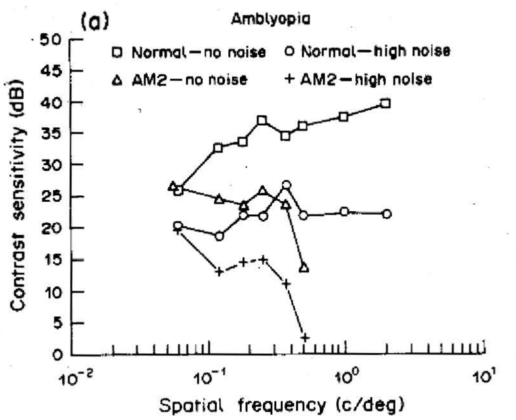


Fig. 12(a)

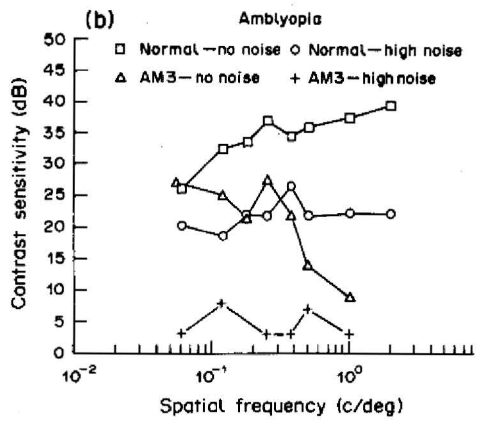


Fig. 12(b)

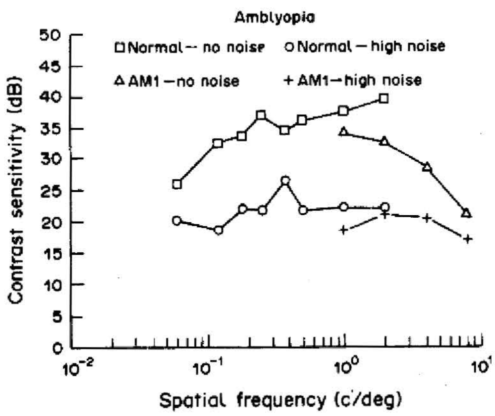


Fig. 13

Figs 12–13. The three panels show contrast sensitivity functions for the three amblyopes. Figure 13 shows data for anisometropic amblyope AM1. Thresholds in noise are normal at least to 2 c/deg and probably beyond [see Fig. 4(b) for data collected at 5 c/deg on AM1]. As before, AM1 has an optical-like component to the deficit. In contrast to AM1, Fig. 12(a) and (b) show evidence for a loss of sampling efficiency at spatial frequencies above 0.06 c/deg. At 0.06 c/deg, contrast thresholds in the absence of noise for both AM2 and AM3 are normal. This is what one might expect of an optical deficit, where at low enough spatial frequencies, the optical MTF is near one. However, despite this, thresholds in noise at higher frequencies are not normal and their sensitivity loss cannot be described as an optical-like loss. AM2 and AM3 are strabismic amblyopes.

Figs 5–13. These panels show contrast sensitivity (in dB relative to 100%) as a function of log spatial frequency (in cycles per degree) for the cloudy media condition (squares). Open squares and triangles symbols represent thresholds in the absence of noise. Open circles and pluses represent thresholds in the presence of noise. The averaged thresholds in the presence and absence of noise (open circles and triangles, respectively) for N1 and N2 are plotted for comparison.

DISCUSSION

We have shown how measurement of contrast thresholds in noise enables one to measure the relative sampling and optical-like components of contrast sensitivity loss. Relative sampling efficiency compares the contrast sensitivity of a patient to normal in high levels of visual noise. Relative optical-like efficiency compares the level of noise required to raise contrast thresholds of a normal observer to that of a patient. In all three optical cases, optical-like efficiency was below normal. However, the presence of clear optics did not always imply a normal OLE. This suggests that a low optical-like efficiency may be a necessary, but not sufficient indication of optical loss.

For those patients with manifest neural deficit, sampling efficiency was consistently below normal. It would be useful if this was generally the case, so that low sampling efficiency would be a necessary and sufficient indication of neural loss. However, two observations suggest this conclusion would be premature. One is the low sampling efficiency of the cataract patient, CM2, which may or may not

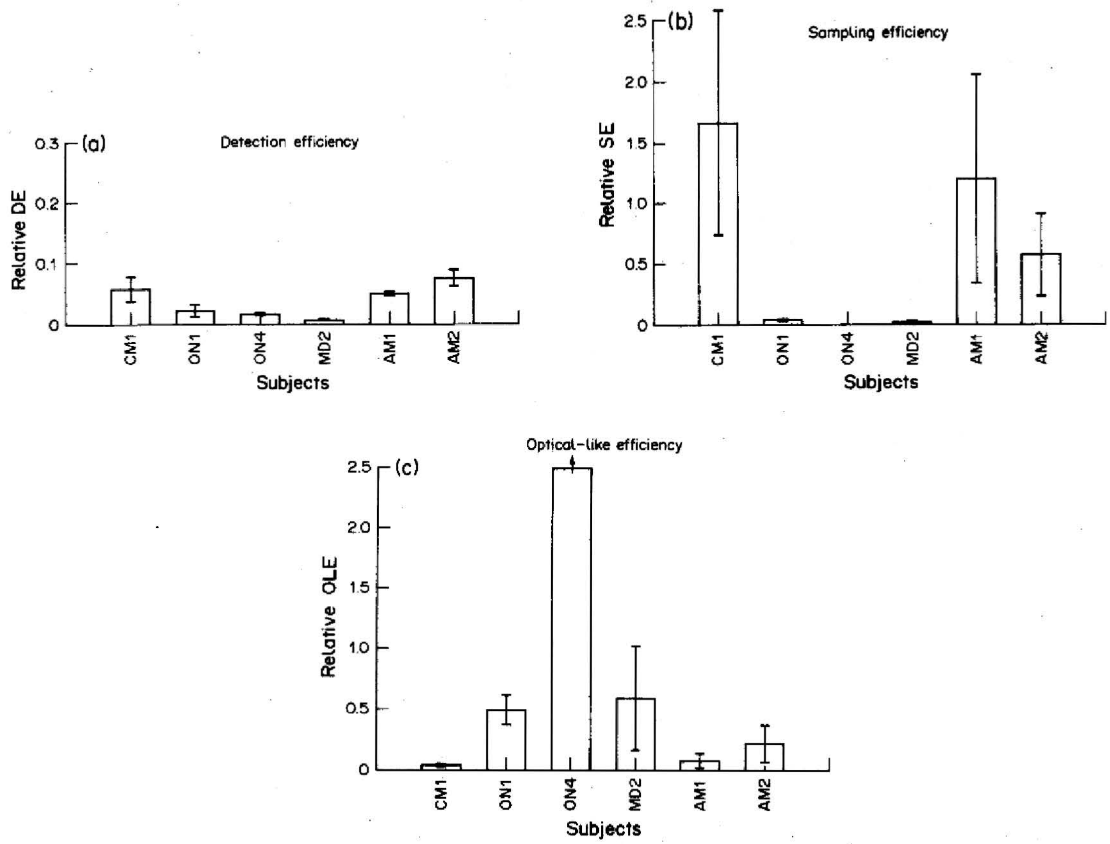
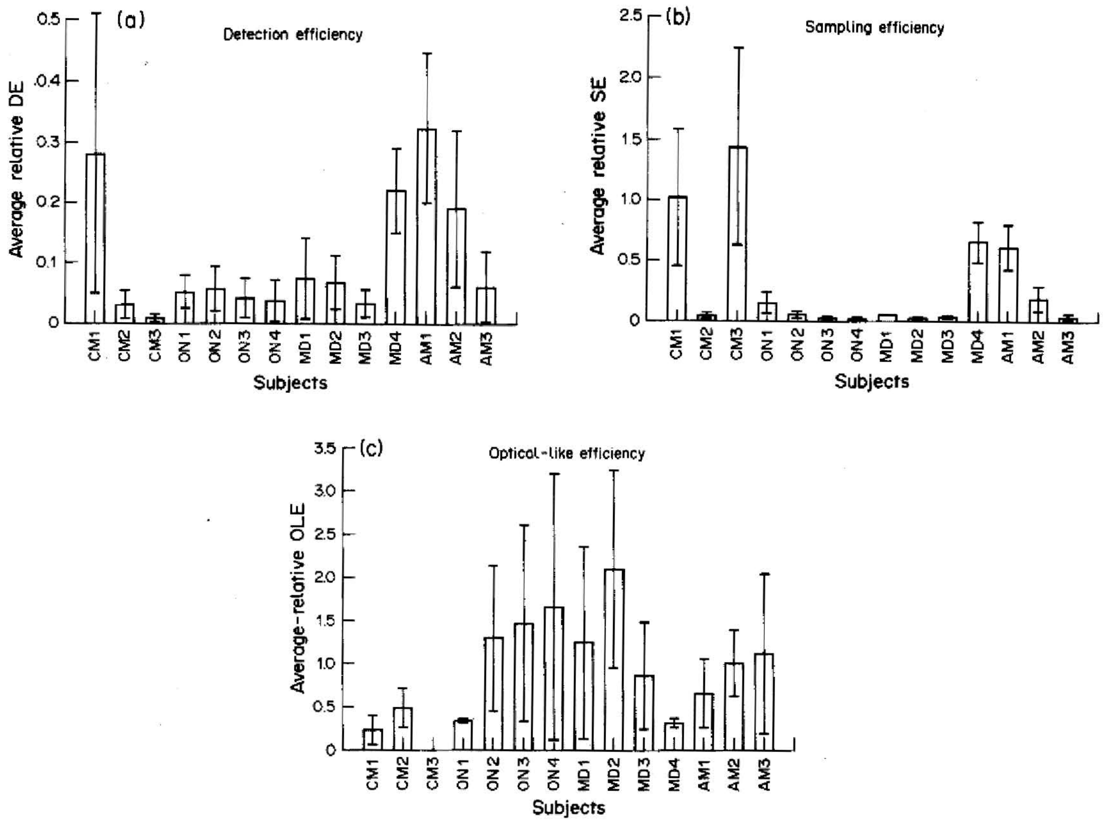


Fig. 14(a, b, c). The relative detection, sampling and optical-like efficiencies for Experiment 1. Here the efficiencies were computed separately for the 70 and 90% correct points by linear regression fits to equation (5), rather than on the averaged contrast energy thresholds as for Figs 2–4. The efficiencies were then averaged and the error bars on the graph indicate the range of the 70 and 90% estimates of these data. The actual relative OLE for ON4 was 14.6 ± 1.9. Different normative data were used for comparison and thus, the large variability between subjects is due, in part, to the variability in both the numerator and denominator for the 70 and 90% correct points. Averaging efficiencies after the linear regression fit can lead to different results, as can be seen when comparing the relative SE for AM1 under the two procedures [Fig. 4(b) and 14(b)]. When an outlier for the 70% correct point was removed (at the highest noise level), the relative SE dropped to 0.66, closer to the 0.65 figure obtained from Fig. 4(b).

indicate neural loss. Further, the sampling efficiencies of several of the neural loss cases were near or just below normal (AM1, AM2, MD4). Nevertheless, our preliminary studies suggest that these measures should be explored further. It is also clear that sampling and optical-like components are not entirely independent. High sampling efficiencies tend to go along with low optical-like efficiencies. It may be useful in future work to combine these two measures into one which would yield a more reliable indicator of optic vs neural loss than either of them separately.

There are a number of shortcomings of our general method. One is that because the equivalent noise grows as the contrast sensitivity function drops, it is difficult to estimate sampling efficiency at frequencies approaching the spatial frequency cut-off. This is where laser interferometry or Maxwellian view acuity measurements would have an advantage. Another problem is that our interpretation depends on our model of the optical MTF—there should be no large regions of zero transfer at frequencies lower than the cut-off. Future research into the nature of optical deficits is needed. In order to distinguish optical loss from optical-like neural loss requires more detailed understanding of how neural loss can affect equivalent noise while leaving sampling efficiency unaffected. This could result, for instance, from a deficit whose only effect was to depress rather than reduce to zero the signal-to-noise ratio across a range of synapses. Another criticism is that we had to assume that only noise frequencies near that of the signal affect thresholds. Although this is a reasonable assumption over much of the frequency range of normal observers, it is not



Figs 15(a, b, c). Relative detection, sampling and optical-like efficiencies from Figs 5 to 13 (Experiment 2) averaged over spatial frequency.

known to what extent it holds true for cases of visual deficit. One way our method could be made less dependent on this assumption is to use narrow band masking noise. This would, in effect, make the task a contrast discrimination experiment. This would have the added advantage of permitting a greater range of noise spectral density. Also, clearly more subjects and a wider range of neural deficits need to be considered to evaluate the utility of this technique, both as a potential clinical tool and as a means to understand the nature of a particular visual loss.

To the extent that the contrast sensitivity function becomes a useful clinical tool, we hope this straightforward extension of adding visual noise to existing contrast detection procedures may prove a useful adjunct. It allows any depressed sensitivity to be dissected into optical-like and non-optical-like components. In the majority of cases, the non-optical-like responses corresponded to a neural loss to one or other sites in the retinal-cortical pathway. If this becomes a general finding, then it may be a useful way to use a simple psychophysical test to see behind cloudy optics.

Consider the illustration in Plate 2. The plate illustrated on the top was originally produced by Campbell and Robson in the sixties as a demonstration of the contrast sensitivity function of the normal human eye. Both contrast and spatial frequency are represented logarithmically and at a distance of arms length, the observer can see the well-known inverted U-shape function that characterizes our visual contrast sensitivity. In disease, this function is perturbed and can be reduced in a wide variety of different ways depending on the pathology. Unfortunately this is also true for optical anomalies. When optical and neural anomalies occur together as they often do in the elderly, it is difficult to assess their relative contributions. Our results suggest that if noise is added to the stimulus, if the loss is optical it will become normal (in noise), whereas if it is neural, it will be further depressed. If our results can be generalized, then patients with optic neuropathy, strabismic amblyopia and some forms of macular degeneration who see their sensitivity functions reduced below normal when they look at the plot on the top (without noise) will see it reduced further in noise (bottom photograph of

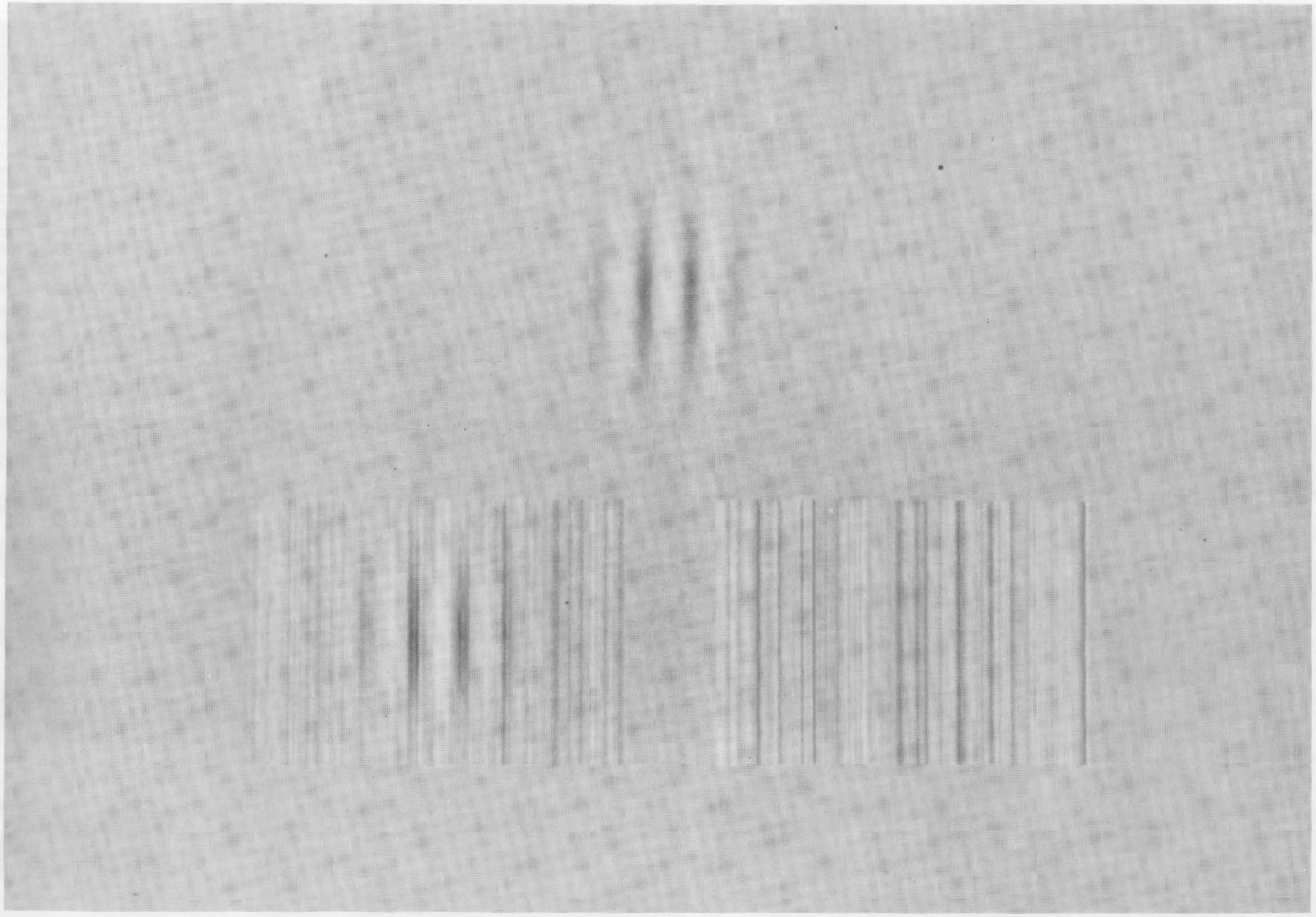


Plate 1. Examples of the stimuli used. The stimulus at the top shows the grating patch in the absence of noise. On the lower left is the grating patch in additive one-dimensional noise. The lower right shows noise alone. In the actual experiment, the noise vs signal plus noise pair (or uniform field vs signal pair) stimuli were presented using temporal two-alternative forced-choice.

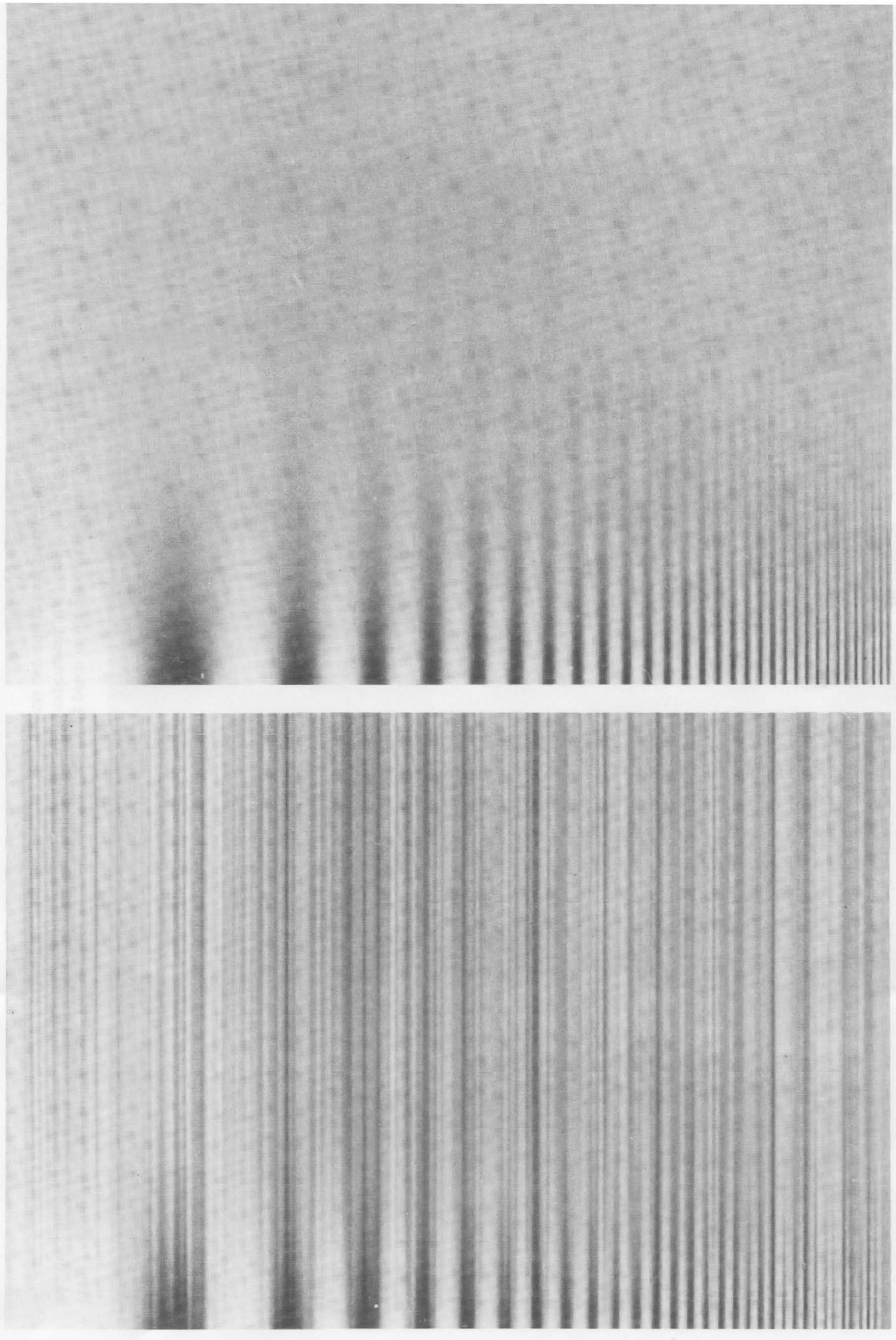


Plate 2. The top panel is a photograph of a sine-wave grating whose contrast drops exponentially from the bottom to the top of the picture and whose spatial frequency increases exponentially from left to bottom. This demonstration was initially devised by F. W. Campbell and J. G. Robson. The panel on the bottom represents the contrast sum of the grating from the top panel and spectrally flat Gaussian noise. At arms length, the reader can view his/her own contrast sensitivity function in the presence and absence of noise in log coordinates.

Plate 2). Patients with optical anomalies will see no further reduction to their sensitivity functions in noise. This is the basis of the test.

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APPENDIX

Sampling Efficiency and Equivalent Noise

Let $c(x, y, t)$ be the contrast function of the pattern

$$c(x, y, t) = \frac{L(x, y, t) - L_0}{L_0} \tag{1}$$

where $L(x, y, t)$ and L_0 are the luminance at (x, y, t) and mean luminance over some interval, respectively. The contrast energy, E, is the integral over x, y, and t of the squared contrast function

$$E = \int\int c(x, y, t)^2 \, dx \, dy \, dt \tag{2}$$

If N is noise spectral density (the r.m.s. contrast squared per squared c/deg per Hz) of the noise at the screen, then ideal performance d' is given by

$$d' = \sqrt{\frac{E}{N}} \tag{3}$$

where d' is $\sqrt{2}$ times the z-score of the proportion correct in a two-alternative forced-choice measurement (Green and Swets, 1970; Kersten, 1984). We now introduce two constraints on otherwise ideal performance. First the signal is modified by a linear shift-invariant spatial filter with (possible complex) transfer function $H(f_x,f_y)$ and MTF $|H(f_x,f_y)|$. This may describe the effect of the eye's optics, for example. For completeness, we also assume a linear temporal filter $H(f_t)$. The spectrum of the signal is now $C(f_x,f_y,f_t)H(f_x,f_y)H(f_t)$, where $C(f_x,f_y,f_t)$ is the Fourier transform of $c(x,y,t)$. Second, we presume that the observer behaves as if there is more noise than just N, by adding an internal noise, N_i following the filter. Adding an equivalent noise prior to the filter. Ideal performance given these constraints is represented by

$$d' = \sqrt{\int\int\frac{|C(f_x,f_y,f_t)H(f_x,f_y)H(f_t)|^2}{N|H(f_x,f_y)H(f_t)|^2+N_i}\,\mathrm{d}f_x\,\mathrm{d}f_y\,\mathrm{d}f_t} \quad (4)$$

(Burgess, 1984). Note that if $N_i \ll N|H(f_x,f_y)H(f_t)|^2$, (e.g. because of high display noise), so that we can neglect N_i, then the MFT drops out and the equation simplifies to equation (3). That is, performance for this sub-optimal detector approaches the ideal's in high noise levels. (This follows by noting that, by Parseval's theorem, contrast is equal to the squared spectrum integrated over spatial and temporal frequency.) Although in actual practice, human performance does not become ideal in high levels of noise, in many cases it comes close. Burgess *et al.* (1981) report efficiencies as high as 70% for grating patch detection in static noise and Kersten (1984) reported efficiencies near 30% in one-dimensional dynamic noise.

From an empirical point of view, introducing factors *sampling efficiency* S (or SE elsewhere in this paper), and *equivalent noise* N_{eq} allow good fits to most data so far collected

$$d' = \sqrt{\frac{ES}{N+N_{eq}}} \quad (5)$$

(Burgess *et al.*, 1981; Pelli, 1981; Legge *et al.*, 1986). Equivalent noise can be interpreted as that extra noise which would have to be added to the input to account for the measured thresholds. Sampling efficiency is the asymptotic absolute statistical efficiency as noise level grows (absolute efficiency is Nd'^2/E). In order to understand the significance of sampling efficiency, note that contrast energy can be expressed as r.m.s. contrast squared times the extent and duration of the signal. Any decrease in the extent or duration (e.g. due to the image falling on a scotoma) causes a drop in d', which is captured by a drop in S. The term sampling efficiency is meant to reflect loss of statistical samples, perhaps due to loss of measurement area or duration, which decreases sensitivity.

Depression of the MTF, with no drop in cut-off frequency, raises the equivalent noise. This can be easily seen in the context of the equation (4) by noting that N_i can be written as $N_{eq}H(f_x,f_y)H(f_t)^2$. Sampling efficiency remains unaffected. However, if there are holes in the sampling array (or complete bands missing from the spatial spectral sensitivity range), sampling efficiency drops. Sampling efficiency will also drop if there is multiplicative noise. One assumption of the research presented here is that there are few low-frequency regions where the MTF of a patient remains near zero. It is well known that the amplitude of the optical transfer function can pass through zero (spurious resolution), however these points are scarce. Thus generally, sampling efficiency should remain unaffected by a bad optical MTF, well below the cut-off frequency.