

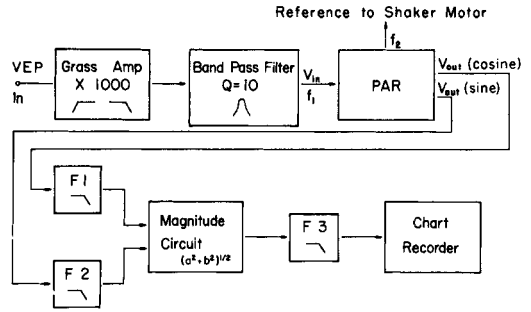
## A reconsideration of visual evoked potentials for fast automated ophthalmic refractions.

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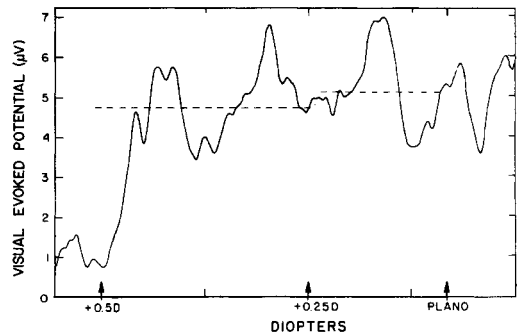
*Visual evoked potentials (VEP's) recorded from the scalp are sensitive to retinal image sharpness and thus to changes in the refractive state of the eye. Initially the responses to checkerboard flash or reversal stimuli were computer-averaged in order to raise the signal above the noise, primarily the electroencephalogram (EEG). Recently analogue Fourier signal analysis has been proposed for using the VEP for rapid clinical refraction. We have confirmed that this method can measure the spherical refractive state to within  $\pm 0.50D$ . However, because of large slow-wave fluctuations, measurements of changes within  $\pm 1.00D$  are not always clear. Despite its promise the method does not appear to be clinically useful at present.*

Visual stimulation of the normal eye results in the activation of neural elements in the retina. These neurally processed signals are transmitted to the brain via the optic nerve. Most of these visual input fibers synapse in the lateral geniculate body and then project to the visual cortex. The volume potential signals generated in the brain by this incoming volley of neural activity are called visual evoked potentials (VEP's) and can be recorded noninvasively from the occipital scalp at or near the inion, which is the protuberance at the base of the cranium on the midline of the posterior skull. Since the VEP's apparently have not been processed intermodally by the higher centers of the brain, they should, in principle, provide an excellent objective method of refraction through its sensitivity to retinal image sharpness.

Spehlmann<sup>1</sup> established the sensitivity of the retina to different patterns of flashed stimuli by comparing the "late wave" (200 msec) components. Rietveld et al<sup>2</sup> studied the VEP response to form. Harter and White<sup>3</sup> demonstrated that the transient VEP to flashed patterns is quite sensitive to focus. Ludlam and Meyers,<sup>4</sup> using a flashed pat-

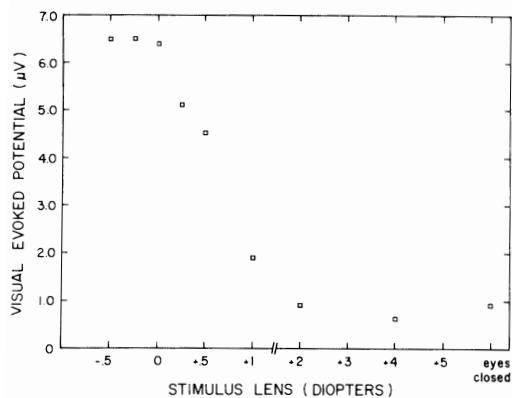


**Fig. 1.** Block diagram of system used in analogue Fourier analysis of VEP's. Filters F1 and F2 are internal to the PAR and had a time constant of  $T_{F1} = T_{F2} = 10$  sec.  $V_{in}$  contains the frequency of interest,  $f_1$ , and other EEG noise. The PAR had an internal 3 Hz oscillator which generated  $f_2$  (the reference frequency) and drove the shaker motor. The magnitude circuit gives the square root of the sum of the squares of the magnitude components of the quadrature sine and cosine wave forms. Filter F3 smooths the magnitude before it is recorded on the chart recorder.

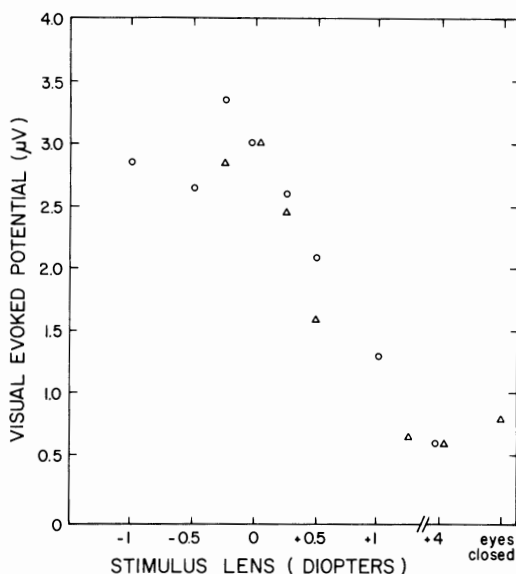


**Fig. 2.** Raw data showing large amplitude variations in the VEP. Record typical of our results with all subjects. The means (dashed lines) were approximated by counting squares underneath the curve. The mean value for +0.5D is  $4.76 \mu V$  and for +0.25D is  $5.11 \mu V$ . The peak-to-peak signal variation was  $2.84 \mu V$  for +0.5D and  $2.70 \mu V$  for +0.25D. The signal-to-noise ratios are 1.68 and 1.90 for +0.5D and +0.25D, respectively. Arrows denote a change in lens power.

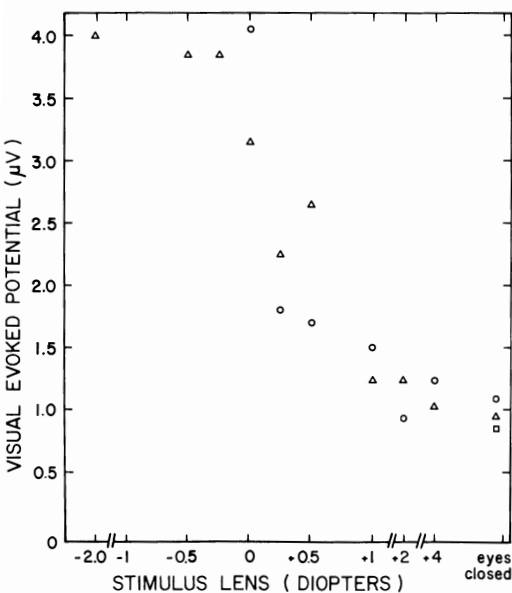
tern stimulus and computer averaging, obtained 0.25D sensitivity. Likewise it has been found that the amplitude of the VEP induced by continuous pattern reversal of a fine checkerboard pattern is sensitive to the sharpness of focus of the pattern on the retina.<sup>5</sup> Millodot and Riggs<sup>6</sup> used this type of



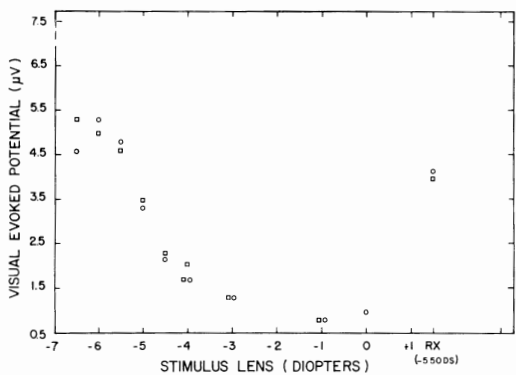
**Fig. 3A.** Subject L. K., one trial.  $T = 10$  sec;  $T_{F3} = 1$  sec; Rx: +0.5D. Left eye.



**Fig. 3C.** Subject B. P., two trials.  $T = 10$  sec;  $T_{F3} = 1$  sec, w/contacts (recent Rx).



**Fig. 3B.** Subject J. H., two trials.  $T = 10$  sec;  $T_{F3} = 1$  sec; no Rx. Left eye.



**Fig. 3D.** Subject C. F. L., two trials.  $T = 10$  sec;  $T_{F3} = 1$  sec; Rx: -5.50D. Right eye.

visual stimuli and analogue signal averaging to obtain a 0.25D sensitivity.

Although such techniques showed promise as a laboratory refraction method, they were unattractive clinically because of the long periods of time required by the averaging techniques. A major advance was reported by Regan,<sup>7</sup> who used continuous pattern reversal stimulation and analogue Fourier analysis of the resulting VEP's. Although computer averaging may require anywhere from 20 to 400 stimulus presentations, requiring long periods of time, Regan reported that using his method he could find the best lens power (to

within 0.50D) in 10 to 20 sec. This rapid speed of refraction is clinically attractive where time is important. It also reduces errors due to possible subject fatigue and habituation and hence should improve the reliability of the technique.

The purpose of the present experiments was to reproduce these results on a wider sample of clinical patients using commercially available Fourier analysis equipment and filters. We found, to our surprise, more variability in the results of the technique than have been generally reported even though the equipment employed was technically more advanced than that previously used. The

amount of variability was large enough to render the technique not clinically useful at the present time.

**Material and methods.** A black-and-white checkerboard pattern alternating at 3 Hz was used as the stimulus. A checkerboard gives a larger VEP signal than a grating pattern.<sup>2</sup> The checkerboard pattern was created by a transparency in a slide projector. A lens was placed in front of the projector, and the lens was modulated transversely by a shaker motor at a rate of 3 Hz to produce shifting antiphase of the checks. Since each cycle results in two phase changes (light-dark and dark-light) the pattern has a contrast (phase change) frequency of 6 Hz. The 6 Hz VEP's are generated by phase changes, not by pattern oscillation changes, which occur at 3 Hz.

The checkerboard pattern has a constant average luminance which avoids the luminous on-off response which would result from a flashed stimulus. However, pattern oscillation changes can cause local flicker VEP's which resemble pattern reversal VEP's and the second harmonic would be at 6 Hz. There is no way to avoid this contamination, but its contribution to the signal can be made small by making the check size small. The contrast between the black (65 c/m<sup>2</sup>) and white (86 c/m<sup>2</sup>) squares was approximately 28%. This is low enough to avoid the saturation effects of large contrasts.<sup>3</sup> The check size was about 14 min of arc and the total subtense was 9.5 by 14 degrees of arc. The check size was chosen to be well below the saturation limit for this parameter, found by Regan and Richards<sup>8</sup> to be about 25 to 35 min of arc. The screen was placed 20 feet from the subject to inhibit accommodation and minimize any possible proximal accommodation and convergence effects. The subject was asked to fixate on a small red dot in the center of the stimulus field. Natural pupils were used.

The VEP was recorded differentially from two electrodes placed on the skin over the mastoid bone and at another point 3 to 5 cm above the inion (after Halliday and Michael<sup>9</sup>). The other mastoid bone area was used as a ground for the Grass P-15 biological amplifier, which has an input impedance of 200 megohms and gain of 1,000. Scalp impedances were maintained between 3K and 10K ohms through appropriate cleaning of the skin and electrode paste application. The Grass amplifier has a low-pass, one-pole filter which was set at 100 Hz and a high-pass, one-pole filter which was set at 0.3 Hz (½ amplitude frequencies). A bandpass filter with a Q = 10 and gain of 10 (henceforth abbreviated as BP(Q = 10)) fol-

lowed the Grass amplifier. The purpose of this filter is to reduce alpha wave and muscle artifact interference and thus improve the dynamic reserve and reduce the dynamic range. The alpha waves are between 10 and 100 times larger than the signal in amplitude and vary in phase, frequency, and amplitude with time. They complicate the signal extraction problem for any scheme of electronic signal processing. The bandpass filter also removes any higher harmonics.

The method used in this investigation to extract continuously just that component of the brain waves produced by the alternating visual pattern was an analogue Fourier analysis based on a Princeton Applied Research Corporation Model 129A Two Phase/Vector Lock-In Amplifier (PAR). The PAR has high-gain, high-Q, adjustable-filter time constants, and is a low-noise device, which makes it ideal for this application. This signal is filtered with a BP(Q = 10) and the result is multiplied by the reference square wave. The following equations describe the multiplier output:

$$V_{out} = \sum_{n=0}^{\infty} \frac{2 V_{in}}{(2n+1)\pi} \cos\{2\pi(f_1 - (2n+1)f_2)t + (\varphi_1 - (2n+1)\varphi_2)\} - \sum_{n=0}^{\infty} \frac{2 V_{in}}{(2n+1)\pi} \cos\{2\pi(f_1 + (2n+1)f_2)t + (\varphi_1 + (2n+1)\varphi_2)\}$$

For the case where  $n = 0$  and  $f_1 = f_2$  ( $f_1$  is the input frequency and  $f_2$  is the reference frequency) and  $\varphi_1 = \varphi_2$  ( $\varphi_1$  is the input phase and  $\varphi_2$  is the phase of the reference signal) the cosine of the differences becomes a DC term. If  $f_1 \approx f_2$ , a low-frequency sine wave output results. Since the DC term is all that is desired, the low-pass, DC-coupled filters are necessary to filter out any low-frequency components. The largest practical time constant was 10 sec. This gave a calculated value of  $Q = 240$ . The cosine of the sum of the frequencies term will have a frequency well beyond that of the low-pass filter at the output, and it can be disregarded. Likewise, a similar equation for sine waves is available. However, when  $f_1 = f_2$ ,  $\varphi_1 = \varphi_2$  and  $n = 0$ , the sine argument becomes a multiple of  $2\pi$  and the result is equal to zero.

A magnitude circuit of the form  $(a^2 + b^2)^{1/2}$ , where  $a$  and  $b$  are the cosine and sine magnitude components, followed the PAR. There was another filter after the magnitude circuit. The PAR internal oscillator also drove the stimulus shaker motor (see Fig. 1 for the system used).

**Results.** Results were obtained from four sub-

jects studied in five sessions. Fig. 2 shows a typical example of the type of raw record obtained on the chart recorder as different trial lenses were inserted before the subject's eye at the times marked by the arrows. When the value of the trial lens approached within about 1.00D of the subject's correct prescription, there was a consistent, rapid, and unmistakable increase in the value of the processed VEP. At further steps of 0.25D the changes in signal amplitude showed a random slow-wave fluctuation about this elevated level. We estimated the mean value of signal during the placement of each lens by counting the number of squares on the chart recorder paper contained by the signal fluctuations. Two such estimates of the mean are shown on Fig. 2. Peak-to-peak variabilities of the signal about this estimated mean were typically  $\pm 10\%$  to  $\pm 30\%$  of the mean value. Thus it is seen that estimation of the best lens directly from the output of the chart recorder would be highly unreliable over a range of less than  $\pm 1.00D$  about the correct prescription.

Averaged results on the four subjects are shown in Figs. 3A to 3D.

**Discussion.** This experiment was designed to test the effectiveness of using VEP's in a clinical situation to find the best prescription as rapidly as possible. The VEP is buried in noise that is many times larger in amplitude. To extract the induced wave form an electronic system was used that functioned as a very narrow band filter. Although a definite increase in amplitude was observed as the correct prescription was approached, a large slow-wave fluctuation was superimposed on the VEP signal. This noise could not be filtered out with the present system.

The source of the noise was not determined. Nearby frequencies, such as alpha waves (8 to 14 Hz), would have a negligible effect on the output signal. Frequencies very near the center frequency would affect the signal, as would a slight time-varying frequency shift. A physiologically induced amplitude or phase variation of the signal could also cause the slow-wave fluctuations noted in the raw records.

The possible effect of fluctuations in accommodation causing variation in retinal image sharpness was considered. A control experiment was devised which used a 1 mm diameter pinhole aperture placed before the eye. The pinhole is known to eliminate blurring resulting from errors or fluctuations in accommodation. The resulting output showed the same kind and magnitude of variations as those trials without the pinhole. It does not appear that the noise can be attributed to accom-

modation. These results indicate that the noise would likewise be unaffected by cycloplegia.

Results from four subjects show that the best corrective refraction can be determined reliably only to within about 1.00D, which is not adequate when compared to objective clinical procedures now used. If the raw record is further averaged (as in Figs. 3A to 3D) results can be improved to 0.50D and 0.25D on some subjects. Due to the large amount of signal overlap at these tolerances, the best corrective state is somewhat ambiguous.

In principle, VEP's should provide a useful objective measure of the sharpness of focus of an image on the retina, closer to that of subjective methods than objective clinical procedures now used. In our opinion, a more stable signal will be required before this method can become clinically useful.

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**Key words:** objective refraction, visual evoked potential, continuous pattern reversal, analogue Fourier analysis, fast automated ophthalmic refraction

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